The influence of heel lift on the loading of the Achilles tendon in running

THE INFLUENCE OF HEEL LIFT DEVICES ON THE LOADING OF THE ACHILLES TENDON IN RUNNING

by
Sharon J. Dixon

A Doctoral Thesis

Submitted in partial fulfilment of the requirements for the award of Doctor of Philosophy of the Loughborough University

November 1996

In Homer's Iliad, the Greek chieftain Achilles was mortally wounded
by an arrow which pierced his Achilles tendon, his only unprotected area.
An immersion in the River Styx had made the remainder of his body invulnerable.
Since then, the Achilles tendon has reflected the weak point of the human body.

ABSTRACT

The incidence of Achilles tendon injury amongst distance runners is high, resulting in lost training and deterioration in performance. The occurrence of this injury has been associated with the repeated loading of the tendon in running. The influence on tendon loading of heel lift intervention, a common treatment of Achilles tendon injury, is not known. The aim of this research was to investigate the influence of heel lift devices on Achilles tendon loading during running. Estimation methods were developed using synchronised ground reaction force and lower extremity kinematic data. Evaluation of methods was achieved using a sensitivity analysis and magnetic resonance imaging techniques.

Due to a variable response to footwear interventions across individuals, a single subject approach was employed throughout the study. The influence of heel lift manipulation on maximum Achilles tendon force was initially investigated for a rearfoot striking subject. Subsequently, the influence on maximum Achilles tendon forces of firm tapered heel lift devices was investigated for three barefoot subjects with distinctly different running styles using a 'multiple single subject' design. The same experimental design was then employed to investigate the influence of heel lift manipulation on maximum Achilles tendon force, Achilles tendon loading rate, sagittal plane joint angles, and frontal plane rearfoot angle for eight rearfoot striking subjects.

Over the course of these studies, maximum Achilles tendon forces ranging from 6 BW to 18 BW were obtained. In the initial study of a single rearfoot striking subject, it was found that the attachment of shock absorbing heel lifts to the plantar surface of the barefoot reduced the maximum Achilles tendon force ($p<0.01$). In the study of three subjects with different running styles, it was found that a rearfoot striker and a midfoot striker demonstrated an increase in maximum Achilles tendon force with the use of firm heel lift devices, whereas no changes in this force were observed for a forefoot striker ($p<0.05$).

In the study of eight rearfoot strikers, four subjects demonstrated a significant decrease in maximum Achilles tendon force with increased heel lift, whilst one subject demonstrated a significant increase ($p<0.05$). For these eight subjects, a trend for a later occurrence of maximum Achilles tendon force with increased heel lift resulted in a decrease in average Achilles tendon loading rate. With increased heel lift, seven of the eight subjects demonstrated a reduction in ankle dorsi-flexion at impact of between three and eight degrees ($p<0.05$), and all eight subjects demonstrated a similar reduction in maximum ankle dorsi-flexion ($p<0.05$). The resulting ranges of ankle angle from ground impact to maximum dorsi-flexion were consistent across all conditions. There was no clear pattern in rearfoot angle response to heel lift manipulation.

This research has highlighted that, despite similar sagittal plane kinematic responses to heel lift, individual assessment of Achilles tendon forces is required. Even when runners demonstrating the same ground contacting style were studied, the response of individuals to common heel lift interventions was varied.

ACKNOWLEDGEMENTS

I would like to thank my supervisor, David Kerwin, for his guidance, encouragement and support throughout my studies. I thank also Dr Fred Yeadon for his advice and helpful comments.

I am grateful to Darren Atkin and Lesley Griffin for their help with data collection, and to my subjects, who performed repeatedly without complaint.

Special thanks to Jennifer Hogbin for her continued support and encouragement.

Finally, I would like to thank the Department of Physical Education, Sports Science and Recreation Management, Loughborough University for their financial support of this work.

PUBLICATIONS AND PRESENTATIONS

This research has been presented in the following publications and conference presentations.

Publications

Dixon, S.J. and Kerwin D.G. (1993). Estimating Achilles tendon force using quasi-static moments. *Journal of Sports Sciences* **11**, p.170.

Dixon, S.J. and Kerwin D.G. (1996). The influence of heel lift on Achilles tendon forces during running. *Journal of Applied Biomechanics* (under review).

Conference presentations

Dixon, S.J. and Kerwin, D.G. (1993). Estimating Achilles tendon force using quasi-static moments. Partners in Performance, BASS, Manchester.

Dixon, S.J. and Kerwin, D.G. (1994a). The influence of heel raises on Achilles tendon force in barefooted running. Second World Congress of Biomechanics, Amsterdam.

Dixon, S.J. and Kerwin, D.G. (1994b). The influence of heel raises on Achilles tendon force in running. Biomechanics Section of BASES, London.

Dixon, S.J. and Kerwin, D.G. (1995a). The influence of firm heel lifts on maximum Achilles tendon force during barefooted running. XVth Congress of the International Society of Biomechanics, Jyvaskyla, Finland.

Dixon, S.J. and Kerwin, D.G. (1995b). Precision in estimating maximum Achilles tendon forces in running. Proceedings of the Biomechanics Section of BASES, Leeds.

IN MEMORY OF

Sylvia Parker
Team Manager and friend
Parkside (Harrow) A.C.

CONTENTS

		Page
	Abstract	iii
	Acknowledgements	iv
	Publications and Presentations	v
	Table of Contents	vii
	List of Tables	xii
	List of Figures	xv
CHAPTER 1	**INTRODUCTION**	1
CHAPTER 2	**REVIEW OF LITERATURE**	7
2.1	Introduction	7
2.2	The Achilles Tendon	7
	(i) Frequency of Achilles Tendon Injury Occurrence	7
	(ii) Achilles Tendon Structure and Function	8
	(iii) Isolated Tendon Properties	10
	(iv) Identification of Achilles Tendon Injury	13
	(v) Etiology of Achilles Tendon Injury	13
	Repeated loading	14
	Running shoes	15
	Lower extremity kinematics	15
	Other factors	15
	(vi) Treatment of Achilles Tendon Injuries	16
	Conservative methods	16
	Orthoses and shoe inserts	16
	Other methods	17
2.3	Achilles Tendon Loading	18
	(i) Direct Measurement of Achilles Tendon Force	18
	(ii) Indirect Measurement of Achilles Tendon Force	18
	Ankle moment	19
	Distribution problem	20
	Moment arm estimation	22
	(iii) Estimation of Achilles Tendon Strain	24
	(iv) Achilles Tendon Loading Results	26
2.4	Lower Extremity Kinematics	27
	(i) Obtaining Kinematic Data	27
	Data collection	27

	Direct linear transformation	28
	Smoothing and differentiation	29
	(ii) The Ankle and Subtalar Joints	32
	(iii) Frontal Plane Kinematics	36
	Rearfoot motion	36
	Factors influencing rearfoot motion	38
	(iv) Sagittal Plane Kinematics	39
	Quantification of sagittal plane kinematics	39
	Factors influencing sagittal plane kinematics	42
2.5	Ground Reaction Force	43
	(i) Measurement of Ground Reaction Force	43
	(ii) Interpretation of Ground Reaction Force	44
	(iii) Factors Influencing Ground Reaction Force	46
	Running speed and style	46
	Running shoes and shoe inserts	46
2.6	Experimental Methods and Analysis of Results	48
	(i) Experiments	48
	(ii) Methods for Detection of Differences	48
	(iii) Single Subject Studies	49
	(iv) Analysis of Results from Single Subject Studies	51
	Graphical analyses	51
	Times series research	51
	Statistical analyses	52
2.7	Summary and Conclusions	53
CHAPTER 3	**DEVELOPMENT OF METHODS FOR ESTIMATION OF ACHILLES TENDON FORCE IN RUNNING**	
3.1	Introduction	54
3.2	Methods	56
	(i) Questionnaire	56
	(ii) Estimation of Achilles Tendon Forces	56
	Data collection	56
	Digitising	58
	Scaling methods	59
	Synchronisation	60
	Muscle moment calculations	61
	Achilles tendon force calculations	61
	Influence of ankle marker location	65
	(iii) The Influence of Shoes and Heel Lifts	65
3.3	Results	67

(i) Questionnaire 67
(ii) Estimation of Achilles Tendon Forces 67
Scaling methods 67
Ground reaction force 67
Ankle moments 68
Achilles tendon force 69
Influence of ankle marker location 71
(iii) The Influence of Shoes and Heel Lifts 71
Ankle joint moments 71
Achilles tendon force 72
Ground reaction force 73
Moment arms 75
Use of different methods 76
Digitisation 76
Tendon marker reliability 76
Power analysis 77
3.4 Discussion 78
(i) Questionnaire 78
(ii) Estimation of Achilles Tendon Forces 80
Simplifications and assumptions 80
Calibration 81
Sampling frequency 81
Marker locations 82
Achilles tendon forces 83
Number of trials 83
(iii) The Influence of Shoes and Heel Lifts 83
Ground reaction force 83
Ankle moment 85
Achilles tendon force 85
(iv) Response to Questions 89

CHAPTER 4 THE INFLUENCE OF HEEL LIFT ON PEAK ACHILLES TENDON FORCE AND ANKLE DORSI-FLEXION
4.1 Introduction 91
4.2 Methods 93
4.3 Results 98
(i) General Observations 98
(ii) Ankle Moment 101
(iii) Maximum Achilles Tendon Force 102
(iv) Achilles Tendon Angle 103

	(v) Achilles Tendon Moment Arm	106
	(vi) Evaluation of Methods	106
4.4	Discussion	111
	(i) Ankle Moment	111
	(ii) Achilles Tendon Force	112
	(iii) Achilles Tendon Moment Arm	115
	(iv) Evaluation	117
	(v) Response to Questions	119
CHAPTER 5	**EVALUATION USING MAGNETIC RESONANCE IMAGING**	
5.1	Introduction	121
5.2	Methods	123
5.3	Results	127
5.4	Discussion	132
CHAPTER 6	**THE INFLUENCE OF HEEL LIFT ON ACHILLES TENDON LOADING IN REARFOOT STRIKERS**	
6.1	Introduction	135
6.2	Methods	136
	(i) General Procedures	141
	(ii) GRF Calculations	141
	(iii) Joint Moment Calculations	141
	(iv) Achilles Tendon Loading Calculations	142
	(v) Measurement Precision and Accuracy	144
	(vi) Data Analysis	145
6.3	Results	146
	(i) Shock Absorption Conditions	146
	(ii) Ground Reaction Force	147
	(iii) Assessment of Methods	153
	(iv) Measurement Precision and Accuracy	157
	Ankle plantar-flexion moments	157
	Ankle dorsi-flexion moments	157
	Maximum Achilles tendon force	157
	Achilles tendon loading rate	157
	(v) Data Smoothing, Velocity and Acceleration Calculations	158
	(vi) Ankle Moment	159
	(vii) Maximum Achilles Tendon Force	170
	(viii) Achilles Tendon Loading Rate	176
6.4	Discussion	182
	(i) Shock Absorption Conditions	182
	(ii) GRF	182

	(iii) Ankle Moments	183
	(iv) Achilles Tendon Force	185
	(v) Achilles Tendon Loading Rate	187
	(vi) General Findings	189
CHAPTER 7	**THE INFLUENCE OF HEEL LIFT ON LOWER EXTREMITY KINEMATICS IN REARFOOT STRIKERS**	
7.1	Introduction	191
7.2	Methods	192
	(i) Sagittal Plane Calculations	192
	(ii) Frontal Plane Calculations	194
7.3	Results	196
	(i) Sagittal Plane Joint Angles	196
	Standing calibration	196
	Characterisation of angle plots	202
	Evaluation of methods	205
	Impact angles	205
	Sagittal plane joint angles during stance	211
	Joint angles at maximum Achilles tendon force	219
	(ii) Frontal Plane Movement	221
7.4	Discussion	225
	(i) Evaluation of Methods	225
	(ii) Standing Calibration	225
	(iii) Impact Angles	226
	(iv) Angles during Stance	227
	(v) Realtionship between Achilles Tendon Moment Arm and Ankle Angle	229
	(vi) Relationship between Kinematics and Achilles Tendon Loading	230
	(vii) Rearfoot Movement	234
CHAPTER 8	**SUMMARY AND GENERAL DISCUSSION**	
8.1	Summary of Studies	236
8.2	General Discussion and Future Research	237
8.3	Practical Implications	241
8.4	Summary of Results	244
REFERENCES		245
APPENDICES		260

A	Error if 10 degree variation in tendon line of action	260
B	Sample questionnaire	261
C	Correction for centre of pressure	262
D	Basic program for quasi-static calculations	264
E	ANOVA from Chapter 3	275
F	Inverse dynamics calculations	276
G	Template spreadsheet	277
H	Statistics spreadsheet	328
I	Chapter 7 results tables	

LIST OF TABLES

Table 2.1	Typical ultimate stress and strain values for tendon	12
Table 2.2	Typical stress and strain values measured in-vivo	13
Table 2.3	Stiffness and modulus of elasticity values	13
Table 2.4	Maximum hip / thigh extension and flexion angles reported in the literature	42
Table 2.5	Knee extension and flexion angles reported in the literature	43
Table 2.6	Ankle plantar and dorsi flexion angles reported in the literature	43
Table 3.1	Measured distances on the model and on the subject	63
Table 3.2	Mean age, and mileage for athletes for different AT injury incidence	67
Table 3.3	RMSD between DLT and linear scaling methods	68
Table 3.4	Achilles tendon force	70
Table 3.5	Maximum ankle moment	72
Table 3.6	Ankle moment at maximum Achilles tendon force occurrence	72
Table 3.7	Maximum Achilles tendon force	72
Table 3.8	Ground reaction force corresponding to maximum ankle moment	73
Table 3.9	Maximum loading rate for conditions B, S and S2	75
Table 3.10	Length of Achilles tendon moment arm about ankle joint centre	75
Table 3.11	Length of moment arm of GRF about ankle joint centre	76
Table 3.12	Standard deviation for conditions B2 and S using each of the methods	76
Table 3.13	Maximum Achilles tendon force using different marker combinations	77
Table 3.14	Variation in power for a significance level of 0.01 (u=5, f=0.7)	78
Table 3.15	Variation in power for a significance level of 0.05 (u=5, f=0.7)	78
Table 4.1	Maximum ankle moment and Achilles tendon force (SD) for each subject and condition.	100
Table 4.2	Peak Achilles tendon angle, Achilles tendon force and ankle moment, for 50 Hz and interpolated data (Subject A)	100
Table 4.3	Factors influencing maximum Achilles tendon force (i) Subject A, (ii) Subject B	105
Table 4.4	Peak Achilles tendon angle (degrees) and time of occurrence	105
Table 4.5	The influence of different cursors on precision of locating points	109
Table 4.6	RMSD for one, four and ten digitisations of known points	109
Table 4.7	Influence of ankle joint and tendon marker variation of 0.4 mm	110
Table 5.1	Achilles tendon moment arm lengths obtained using a moving centre of rotation, fixed centre of rotation, and skin marker methods	128
Table 6.1	Mean age, height, mass, foot length and tendon measurement for each subject	146
Table 6.2	Peak ankle moments and Achilles tendon force for Subject 7	147
Table 6.3	Average and instantaneous GRF loading rates and time of instantaneous loading rate for each subject/condition combination	149
Table 6.4	Mean vertical GRF variables for each subject/condition combination	150

Table 6.5	RMSD from criterion locations for seven calibration frame markers, and four small markers	154
Table 6.6	RMSD from the mean for seven calibration frame markers, and four small markers	155
Table 6.7	Mean peak plantar-flexion ankle moment magnitude for each subject/condition combination	168
Table 6.8	Mean peak ankle dorsi-flexion moment magnitude for each subject/condition combination	169
Table 6.9	Mean time of occurrence of maximum ankle plantar-flexion moment for each subject/condition combination	169
Table 6.10	Mean maximum Achilles tendon force for each condition and subject, and maximum stress for barefoot condition	171
Table 6.11	Time of occurrence of maximum Achilles tendon force for each condition and subject	172
Table 6.12	Factors contributing to maximum Achilles tendon force value	173
Table 6.13	Calculation of Achilles tendon loading rate: sample data	177
Table 6.14	Mean maximum Achilles tendon loading rates for each subject and condition	178
Table 6.15	Mean time of occurrence of maximum Achilles tendon loading rate for each subject and condition	179
Table 6.16	Average Achilles tendon loading rate for each subject and condition	179
Table 7.1	Mean standing angles for each subject/condition combination	198
Table 7.2	Standard deviation of markers (mm) during calibration	199
Table 7.3	Changes in ankle angle (degrees) with heel lift increase	201
Table 7.4	Foot angle at impact for each subject/condition combination	207
Table 7.5	Ankle angle at impact for each subject/condition combination	208
Table 7.6	Impact angles relative to standing calibration angles	210
Table 7.7	'Flat foot' angle	213
Table 7.8	Maximum ankle angle	214
Table 7.9	Times of occurrence of maximum ankle angle	215
Table 7.10	Ankle angle (degrees) at the time of maximum Achilles tendon force	220
Table 7.11	Knee angle (degrees) at the time of maximum Achilles tendon force	220
Table 7.12	Peak rearfoot angle (degrees) for each condition	220

LIST OF FIGURES

Figure 2.1	The Achilles tendon	9
Figure 2.2	Cross-section of the Achilles tendon	9
Figure 2.3	Muscle-tendon complex for a unipennate muscle	10
Figure 2.4	A typical stress-strain curve for isolated tendon (Abrahams, 1967)	11
Figure 2.5	Proposed relationship between tendon loading, cellular response and overuse injury (Archambault et al., 1995)	14
Figure 2.6	Sagittal plane view of the ankle joint and the subtalar joint	34
Figure 2.7	Pronation	34
Figure 2.8	Supination	34
Figure 2.9	Mean orientation of the subtalar axis in the sagittal and frontal planes	35
Figure 2.10	Mitred hinge representation of subtalar joint motion	35
Figure 2.11	Conventions for measuring sagittal plane joint angles of the lower extremity	41
Figure 2.12	A typical vertical GRF time history	45
Figure 3.1	Arrangement of data collection apparatus	57
Figure 3.2	Body landmarks and wobble points for digitisation	58
Figure 3.3	Calibration plane with marker locations	60
Figure 3.4	Calculation of quasi-static moments about the ankle joint centre	61
Figure 3.5	Forces and moments acting about the ankle joint centre	62
Figure 3.6	Relationship between digitised points and Achilles tendon insertion	64
Figure 3.7	Estimated location of Achilles tendon insertion	64
Figure 3.8	A typical trace of vertical GRF interpolated to approximately 500 Hz	68
Figure 3.9	A typical trace of ankle moment for a single trial (50 Hz)	69
Figure 3.10	Achilles tendon force trace for a single running trial	70
Figure 3.11	Mean Achilles tendon force over 10 trials using each of the methods	70
Figure 3.12	Vertical GRF traces for conditions B, S and S2	74
Figure 3.13	Vertical GRF traces for conditions B, B1 and B2	74
Figure 3.14	Illustration of Achilles tendon markers	77
Figure 4.1	Illustration of angles used for calculation of Achilles tendon angle	94
Figure 4.2	The two cursors used for comparison of cursors	95
Figure 4.3	A hierarchical model illustrating the factors influencing Achilles tendon force	97
Figure 4.4	Typical geometry of the system at the time of maximum Achilles tendon force	97
Figure 4.5	Ankle moment, Achilles tendon force, Achilles tendon and foot angles (Subject A)	99
Figure 4.6	Barefoot ankle moment time histories for each subject	101
Figure 4.7	Barefoot Achilles tendon force time histories for each subject (interpolated data)	102
Figure 4.8	Barefoot Achilles angle time histories for each subject	104

Figure 4.9	Typical plots of Achilles tendon moment arm against Achilles tendon angle	107
Figure 4.10	Achilles tendon moment arm against Achilles tendon angle using data from Burdett (1982); Bobbert et al. (1986); and the present study	107
Figure 4.11	Geometry relating Achilles tendon insertion and ankle joint centre	115
Figure 5.1	Procedure for calculation of centre of rotation using Reuleaux methods	125
Figure 5.2	Procedure for calculation of rotation angle	125
Figure 5.3	Achilles tendon line of action using marker images	126
Figure 5.4	Sagittal plane magnetic resonance images for each ankle angle	129
Figure 5.5	Achilles tendon moment arm against ankle orientation using each method	130
Figure 5.6	Location of fixed and moving centres of rotation relative to the talus	130
Figure 5.7	Typical transverse images showing the cross-section of the Achilles tendon	131
Figure 5.8	Illustration of recommended procedures for calculation of moment arms	134
Figure 6.1	Calibration frame for MacReflex system, showing axes orientation	139
Figure 6.2	Force plate showing axes orientation and location, and marker placement	139
Figure 6.3	Marker locations and attachment of heel lifts	143
Figure 6.5	Performance index against marker size	145
Figure 6.6	Example of the smoothing influence of using mean vertical GRF data	148
Figure 6.7	Example plots of instantaneous GRF loading rate	151
Figure 6.8	Typical GRF time histories	152
Figure 6.9	Theoretical influence of projection of three-dimensional data onto the yz plane	156
Figure 6.10	Smoothing using a Reinsch (1967) spline with different parameter values	161
Figure 6.11	Angular velocity of foot segment obtained using a two point procedure	162
Figure 6.12	The influence of using different algorithms for acceleration calculation	163
Figure 6.13	Ankle moments obtained using inverse dynamics and quasi-static methods	164
Figure 6.14	Achilles tendon force using inverse dynamics and quasi-static methods	165
Figure 6.15	Typical ankle moment time histories	166
Figure 6.16	Peak ankle plantar-flexion moment for each subject/condition combination	167
Figure 6.17	Typical Achilles tendon force time histories	174
Figure 6.18	Maximum Achilles tendon force for each subject/condition combination	175
Figure 6.19	Sample traces of Achilles tendon moment arm and vertical GRF	180
Figure 6.20	Sample traces of GRF moment arm and centre of pressure	180
Figure 7.1	Identification of the medial-lateral foot orientation	194
Figure 7.2	Marker placement and definition of lower leg and calcaneus angles	195
Figure 7.3	Transverse and sagittal plane dimensions with foot in standardised position with 5 degrees medial rotation	197

Figure 7.4	Foot angle for each condition for each subject	200
Figure 7.5	Typical angle time histories for the foot, ankle, knee and thigh	203
Figure 7.6	Characteristic foot traces at the beginning of stance	204
Figure 7.7	Impact ankle angles for each subject/condition combination	209
Figure 7.8	Illustration of foot calibration angle (θ_0) and impact angle (θ_1)	210
Figure 7.9	Maximum ankle angle for each subject/condition combination	216
Figure 7.10	Time of occurrence of peak ankle angle for each subject/condition combination	217
Figure 7.11	Peak knee angle for each subject/condition combination	218
Figure 7.12	Typical rearfoot angle time histories	222
Figure 7.13	Changes in maximum rearfoot angle with heel lift	224
Figure 7.14	Isometric force length relationship	232

GLOSSARY OF TERMS

GRF	ground reaction force
F_x, F_y, F_z	orthogonal ground reaction force components
a_x, a_y	centre of pressure coordinates
R	resultant ground reaction force
F	Achilles tendon force
d_1	Achilles tendon moment arm about the ankle joint centre
d_2	ground reaction force moment arm about the ankle joint centre
σ	common within population standard deviation
u	degrees of freedom
a	significance level
α	lower leg sagittal plane angle
β	foot sagittal plane angle
θ	Achilles tendon sagittal plane angle
RMS	root mean square
ψ	medial - lateral foot angle
ϕ	lower leg frontal plane angle
γ	calcaneus frontal plane angle
η	rearfoot angle

CHAPTER 1 INTRODUCTION

Approximately 32,000 runners, aged between 18 and 85, participated in the Great North Run on 15 September 1996. Both elite competitors and fun-runners were represented in this World Record breaking half marathon. Running is clearly a popular activity, appealing to all ages and abilities. Despite the growth of more fashionable new activities such as aerobics and gym work, running still appeals to around 7% of the population of England and Wales as a sport or recreational activity (Nicholl et al., 1991). Running is a cheap activity, that does not require extensive organisation, facilities or equipment.

The main item of equipment required for running is a suitable pair of running shoes. The vast number of running shoes available on the market is highlighted by the numerous running shoe advertisements in sports magazines (eg. Runners' World, Today's Runner). Several design characteristics, claimed to provide improved performance and protection from injury, are spotlighted in these advertisements. The commercial nature of the running shoe has led to the larger companies funding research into shoe design characteristics and their influence on biomechanics (eg. Nike Sport Research Laboratory, Oregon, USA). Despite its relatively small weight and volume, it has been extensively demonstrated that the shoe has an influence on the way an individual runs. Measurement of external ground reaction forces, and measurement of lower extremity kinematics, have demonstrated that running shoes can influence running biomechanics (Nigg, 1986; Clarke et al., 1983). Geometrical factors such as shoe heel height and width, and material properties, such as the amount of shock absorption provided, have been found to be influential (Stacoff and Kaelin, 1982; Light et al., 1980).

Cavanagh and Lafortune (1980) identified three distinct methods of ground contact employed during running. The response of runners to variations in running shoes and materials placed in the shoe has been found to vary depending on the method of ground contact adopted by an individual (Therrien et al.,1982; Lees and McCullagh, 1984). If differing behaviour exists across subjects in response to heel lift intervention, the chance of observing this behaviour can be increased by the use of subjects with different running styles.

Runners are prone to injuries of the lower extremity (Clement et al., 1984). Van Mechelen (1992) stated that between 37% and 56% of all runners are injured over a period of one year. The annual incidence of running injuries in England and Wales has been estimated as at least two million (Nicholl et al., 1991). Many individuals who have used running as a means of gaining cardiovascular fitness, retire from participation completely when injured, resulting in a decline in fitness and general health. For the elite competitor, lost training due to injury results in declined performance levels. The prevalence of injury incidence at the elite level is highlighted by consideration of distance runners of the British team at the 1996 Atlanta Olympics. For the women's team, Kelly Holmes struggled on through competition over 800 m and 1500 m, despite injury. Of the three 5000 m runners, two were unable to progress to the final due predominantly to the influence of injury. With established competitors absent due to injury, no British runners were present in the 10,000 m. The situation was mirrored by the British men.

With increased pressure to succeed, and increased financial incentive, the training load required to be successful has increased. Knowledge regarding the incidence of injury to elite competitors is limited. Increased research is required in all areas of running shoe design with investigation into possible relationships with injury occurrence. In the present research, elite performers are studied.

Selected running shoe design characteristics have been associated with a reduction in injury. Increased heel height (Subotnick, 1979), and increased shock absorption in the rear of the shoe (MacLellan and Vyvyan, 1981), have both been shown to reduce the frequency of lower extremity injuries. Evidence directly relating this injury reduction with changes in biomechanical variables during running has not been presented (James and Jones, 1990).

Foot orthoses have been designed to ensure that 'normal' foot movement is achieved during the ground contact phase of running. The prescription of these devices has become increasingly common for treatment of injuries to the lower extremity (Kilmartin and Wallace, 1994), and has been described as an art rather than a science (Costain, 1993; personal communication). Essentially the podiatrist uses experience to detect an abnormality and select an appropriate intervention. Epidemiological evidence is available to demonstrate the successful treatment of particular injuries by the use of orthotic devices (eg. Subotnick, 1979). Scientific evidence has been published demonstrating the biomechanical influence of foot orthoses (Nigg, 1986; Smith et al., 1986). These effects have not been correlated with the reduced incidence of injuries. To increase understanding of the mechanisms by which selected interventions are successful in the treatment of injury, knowledge of the influence of interventions on the loading of internal structures of the foot is required.

Achilles tendon injury is common amongst runners (James et al., 1978; Clement et al., 1984). Several research papers have discussed the possible contributors to the occurrence of this injury, with overuse being highlighted (Kvist, 1993; Archambault et al., 1995). Load on body structures during sport is a required stimulus, acting to increase or maintain the strength of biological materials. However, if the load is excessive, damage may result. Nigg (1985) described how the accumulation of cyclic forces that are below the critical limit of a biological structure can lead to damage. Insufficient recovery time prior to the next exercise session results in accumulated damage to the structure and subsequent injury occurrence (Archambault et al., 1995). Following this suggested mechanism for the occurrence of overuse injuries, the study of Achilles tendon injury requires methods to quantify the forces experienced by the Achilles tendon during running.

Methods have been described in the literature for the estimation of forces acting internally on the joints, muscles and tendons of the lower extremity during running. Achilles tendon force values have been obtained directly using in-vivo force transducers (eg. Komi, 1990). Invasive methods interfere with the normal environment of the tendon, and are not practical or ethical for use with human subjects. Indirect estimation methods are therefore favoured, and have been commonly used (eg. Burdett, 1982; Scott and Winter, 1990). Estimation methods require monitoring of the ankle joint moment and the moment arm of the

Achilles tendon about the ankle joint centre during the ground contact phase of running (Groh and Bauman, 1976; Burdett, 1982). It has been found that estimations of Achilles tendon moment arm are subject specific (Burdett, 1982), and vary according to ankle angle (Spoor et al., 1990).

Running shoe design characteristics and the use of orthotic devices have been suggested to influence the incidence of Achilles tendon injury. The use of shock absorbing heel lifts placed in the rear of the shoe has been demonstrated to reduce Achilles tendon pain (MacLellan and Vyvyan, 1981; Fauno et al., 1993). The wearing of shoes with an increased heel lift has also been found to be successful in the treatment of Achilles tendon injury (Subotnick, 1979). The mechanism by which these interventions are successful has not been documented.

To perform research which is aimed at influencing the frequency of occurrence of a particular injury, Nigg and Bobbert (1990) described recommended approaches. The first, termed the 'cause-effect approach', involved the estimation of maximum stress experienced by an internal body structure, and the subsequent comparison of this value with the ultimate stress. The errors inherent in the methods used to estimate internal stress magnitudes, and to obtain ultimate stresses were described. Nigg and Bobbert (1990) therefore concluded that the 'cause-effect approach' is unlikely to be successful in contributing to a reduction in injury occurrence.

The alternative approach of investigating the factors influencing stress magnitude in a defined structure, termed the 'comparison technique', was preferred. Nigg and Bobbert (1990) described this approach as the detailed analysis of a single subject to investigate whether the forces or stresses in a specific structure are smaller for a defined condition, than under an alternative condition. This technique is based on the assumption that all errors which are inherent in the estimation of internal forces are systematic, and thus do not influence comparisons between conditions. The 'comparison technique' was considered to be a suitable approach for the studies performed in this research, allowing determination of factors influencing the loading of the Achilles tendon in running.

The detailed analysis of a single subject for the study of injury occurrence, as suggested by Nigg and Bobbert (1990), has been supported by other authors. Reboussin and Morgan (1996) described how research is typically conducted in stages, commencing with an investigation of the effect of an intervention, to provide an indication as to whether future research in the area is warranted. This is followed by a well-controlled single subject study, on one or more subjects. Finally, providing sufficient evidence has been gained from the initial studies, the use of a well-controlled, randomised group design has been suggested, with the aim of providing an indication of the influence of the intervention on a population. Reboussin and Morgan (1996) stated that the study of multiple subjects does not preclude the repeated observation of the response of individuals, and described the increased popularity of 'multiple single subject' designs.

The interpretation of results from the study of the influence of running shoes and shock absorbing heel lifts on Achilles tendon forces is limited by the lack of control possible in such studies. Running shoes have many varied geometrical and material design characteristics. Shock absorbing heel inserts provide both cushioning and heel lift. It is therefore not possible to deduce which design characteristics have influenced an observed variation in Achilles tendon force under these conditions. Lafortune, speaking at the Second Symposium on the Biomechanics of Functional Footwear in Cologne (1995) highlighted the lack of well-controlled studies in the investigation of factors influencing running biomechanics. This investigator suggested that more detailed study of the response to single interventions is required.

It has been demonstrated that maximum Achilles tendon force occurs during the middle of the stance phase, when the foot is flat on the ground (Burdett, 1982; Komi, 1990; Scott and Winter, 1990). When heel lifts are employed, they are likely to influence the geometry of the lower extremity at this stage of stance. In contrast, shock absorbing properties provided in the rear of the shoe are most likely to influence the impact phase occurring at the start of ground contact. Thus, evidence in the literature points to the raising of the heel as the property of heel lifts that is most likely to influence maximum Achilles tendon force.

When utilising estimation methods for the determination of internal forces, various assumptions must be made. For example, for the estimation of Achilles tendon forces, the proportion of the triceps surae muscle group contributing to net muscle moment about the ankle joint is required. Errors associated with the assumption that skin markers can be used to represent the movement of underlying body structures should be investigated. To support the use of the 'comparison technique', evaluation methods are required to determine whether errors in absolute force values are systematic (Nigg and Bobbert, 1990). For confident use of the method developed in this research, the evaluation of the technique used for measurement of the Achilles tendon moment arm is required. Quantification of the error in measurement of this length requires anatomical knowledge of the locations of the ankle joint centre and the Achilles tendon line of action. Rugg et al. (1990) have demonstrated the use of magnetic resonance imaging techniques to determine the Achilles tendon moment arm length for a range of ankle and knee joint angles.

Factors other than maximum force have been associated with Achilles tendon injury. Tendon is a viscoelastic material, and is therefore sensitive to loading rate (Hawkins, 1993). Tendon has been shown to be subjected to higher stresses when the rate of loading is increased (Frankel and Nordin, 1980). The loading rate of the Achilles tendon may therefore be important when considering the etiology of Achilles tendon injury.

Strain experienced by the Achilles tendon has also been associated with injury occurrence (Clement et al., 1984). The overall combined length of the Achilles tendon and associated muscle, and thus the strain experienced by the tendon, is influenced by the ankle and knee joint angles (Grieve et al., 1978). Conventions for the monitoring of lower extremity sagittal plane joint angles have been described in the literature (Milliron and Cavanagh, 1990). Variations in material properties of running shoes have been found to affect these angles, suggesting that the strain experienced by the Achilles tendon has been influenced. Evidence is not available concerning the influence of shoe geometry on sagittal plane joint angles.

A further factor that has been associated with Achilles tendon injury is movement of the calcaneus relative to the lower leg, commonly referred to as rearfoot motion (Pagliano et al., 1987; Winter and Bishop, 1992). Contrasting results have been presented in the literature regarding the influence of heel lift manipulation on rearfoot motion (Stacoff and Kaelin, 1982; Clarke et al., 1983). This contrasting evidence may have been the result of these studies using running shoes to vary heel height, providing a complex intervention under which variables additional to heel lift are likely to have been influenced.

The aim of the current research was to investigate the influence of heel lift devices on Achilles tendon loading during running. Details of the associated literature are reported in Chapter 2. The development of a method for the estimation of Achilles tendon loading during running is described in Chapter 3. These methods differ from those previously presented in the literature by monitoring the Achilles tendon moment arm throughout the stance phase of running. A series of specific questions are addressed, ranging from whether shoe and heel lift interventions can influence maximum Achilles tendon force to what influence heel lift manipulation has on Achilles tendon loading. Separate studies are described which address these questions, beginning with the one reported in Chapter 3 which involves the use of a single subject to investigate the influence of heel lift manipulation on maximum Achilles tendon force. The main question addressed in this chapter is ' Can maximum Achilles tendon force be influenced by shoe and heel lift interventions ? '.

When using different running shoe and heel lift combinations to manipulate heel lift it is not clear whether any observed differences in Achilles tendon loading are the result of changes in heel lift or in shock absorption properties. Thus, in the study described in Chapter 4, the isolated intervention of raising the heel is applied using firm heel lift devices. The influence of heel lift manipulation is studied using three subjects with distinctly different running styles, increasing the likelihood of detecting different responses to heel lift variation between subjects. The main questions addressed are ' Can raising the heel with firm lifts reduce maximum Achilles tendon force ? ', and ' Do runners with distinctly different running styles differ in their response to heel lift ? '.

When estimating the loads on internal structures an evaluation of methods is required. In Chapter 5, errors in Achilles tendon moment arm lengths are assessed using magnetic resonance imaging techniques. These techniques are also used to obtain a value for the cross-sectional area of the tendon, allowing the estimation of maximum Achilles tendon stress magnitudes. The main question addressed in Chapter 5 is ' Are accurate anatomical data obtained using skin markers ? '.

The use of one subject with each style can provide strong conflicting evidence, but an increased number of subjects are required to provide supportive evidence. Thus, in the studies described in Chapter 6 and Chapter 7, eight rearfoot striking subjects are employed. Variables additional to maximum Achilles tendon force are analysed. In Chapter 6, the development of methods for the estimation of Achilles tendon loading rate are described. The results of a study investigating the influence of heel lift on maximum magnitude and rate of change of Achilles tendon force in rearfoot strikers are presented. Maximum Achilles tendon force values are combined with the tendon cross-sectional area obtained in Chapter 5, to determine the maximum stresses experienced in running. The main question addressed in this chapter is ' How does heel lift influence the Achilles tendon loading of rearfoot strikers ? '.

In Chapter 7, sagittal plane joint angles, monitored simultaneously to the Achilles stress data collection of Chapter 6, are presented. These angles are used to provide an indication of the influence of heel lift on Achilles tendon strain. The relationship between maximum Achilles tendon stress and strain is investigated. Additionally, results are presented on the influence of isolated heel lift intervention on rearfoot motion. The questions addressed in Chapter 7 are ' Are sagittal plane joint angles influenced by heel lift manipulation ? ', ' Can joint angles be employed to adequately represent Achilles tendon strain in running ? ', and ' Is maximum rearfoot eversion reduced by the use of heel lifts ? '. A final general discussion and summary are presented in Chapter 8.

Research has been described involving a progression of experiments, consistent with the suggestions of Reboussin and Morgan (1996). A single subject design is initially employed for the detailed study of one subject. Subsequently, a 'multiple single subject' design is employed to investigate selected behaviour across subjects. The use of an estimation method for the determination of Achilles tendon loading is described, and the use of the 'comparison technique', as described by Nigg and Bobbert (1990), allows the confident detection of differences in Achilles tendon loading across heel lift conditions.

2.1 Introduction

This chapter contains a review of literature, beginning with a consideration of aspects of Achilles tendon injury in runners. Suggested etiology and treatments of this injury are presented, with the successful use of heel lifts in treating and preventing Achilles tendon injury discussed. The lack of present understanding of the mechanism by which heel lift intervention is successful is highlighted. For investigating the loading of the Achilles tendon in running, knowledge of the properties of tendon is required. An overview of the Achilles tendon structure, function and properties is therefore also included in Section 2.2.

To investigate the mechanism by which heel lifts are successful in the treatment of Achilles tendon injury, methods for estimating the loading experienced by this tendon during running are required. Methods that have been presented in the literature for the estimation of internal forces are therefore described in Section 2.3. In particular, estimation methods which employ inverse dynamics techniques are highlighted.

Inverse dynamics methods involve the use of synchronised kinetic and kinematic data. Additionally, selected aspects of ground reaction force and lower extremity kinematics have been associated with Achilles tendon injury occurrence. Thus, the methodology presented in the literature for the collection of ground reaction force and kinematic data are considered in Section 2.4 and Section 2.5, respectively.

For the investigation of the influence of heel lifts on Achilles tendon loading, methods are required for the comparison of selected variables across different heel lift conditions. Section 2.6 therefore contains relevant experimental techniques and statistical methods available to the investigator.

Finally, Section 2.7 summarises the literature most relevant to the present study, and draws together the most important observations.

2.2 The Achilles Tendon

(i) Frequency of Achilles Tendon Injury Occurrence

Achilles tendon injury has been found to be the third most common injury sustained by runners, following knee injuries and 'shin splints' (James et al., 1978). Surveys have provided figures for the percentage of runners suffering from this injury. James et al. (1978), in a survey of 180 runners attending an injury clinic, found 11% of injuries were to the Achilles tendon. Clement et al. (1984) found 6.5% of the running injuries presented at their clinic over a two year period were to the Achilles tendon. Subotnick and Sisney (1986) quoted a personal communication with Pagliano and Jackson (1984) stating that the frequency of Achilles tendon injury occurrence in runners was 5%. Published data by Krissoff and Ferris

(1979) quoted 18% incidence of this injury. Johansson (1986) found in a one year study that 7% of elite orienteers suffered from an Achilles tendon injury. Lysholm and Wiklander (1987) quoted a 9% annual incidence of Achilles tendon injury in runners.

Thus figures ranging from 5% to 18% have been provided for the incidence of Achilles tendon injury in runners. Differences between studies, particularly in the time period of data collection and the method of subject selection, have limited the comparison of findings. However, it is evident that the incidence of Achilles tendon injury in runners warrants the investigation of factors influencing this injury.

(ii) Achilles Tendon Structure and Function

For the investigation of the factors associated with Achilles tendon injury occurrence, knowledge is required of the structure and function of this tendon. The Achilles tendon is the common insertion of the gastrocnemius and soleus muscles onto the posterior superior aspect of the calcaneus of the foot (Figure 2.1). The gastrocnemius and soleus muscles, termed the triceps surae muscle group, are the dominant generators of ankle plantar-flexion. Two synovial bursae protect the insertion of the Achilles tendon on the calcaneus. The subcutaneous bursa is located between the skin and the tendon, and the retrocalcaneal bursa is found between the tendon and the calcaneus.

The cross-sectional structure of the Achilles tendon is illustrated in Figure 2.2. The entire tendon is surrounded by the epitenon, a fine tissue sheath. The epitenon is surrounded by paratenon, a thin layer of tissue. The epitenon and paratenon together are known as the peritendon, and function as an elastic sleeve to allow free movement of the tendon against surrounding tissues.

The relationship between the part of the tendon into which the muscles insert, termed the aponeurosis, the external tendon and the muscle fibres is illustrated in Figure 2.3. The muscle-tendon complex presented is for a unipennate muscle, consistent with the structure of the soleus and the gastrocnemius muscles. The pennation angle, θ, represents the angle between the muscle fibres and the tendon. Ker et al. (1988) demonstrated how θ is small in humans, and described negligible error in estimated tendon forces if this angle is assumed to be zero. Using this evidence, it can be assumed that the maximum force in the tendon is equal to the maximum force generated by the muscle.

Tendon is a passive structure which is brought under tension by the contraction of the associated muscle. The function of tendon is to attach muscle to bone or fascia, and transmit tensile loads from the muscle. Tendon consists almost entirely of collagen fibres arranged in close to totally parallel alignment, making the tendon suited to sustaining high tensile loads.

Tendon is vascularised minimally, reflecting its low metabolic requirements. The result is a slow healing process if damage occurs. For the Achilles tendon, further reduced blood supply has been found between 2 cm and 6 cm above the insertion (Leach et al., 1981). Achilles tendon rupture can occur in this area, leading to the suggestion that reduced blood supply plays an etiological role in the rupture of the Achilles tendon (Smart et al., 1980).

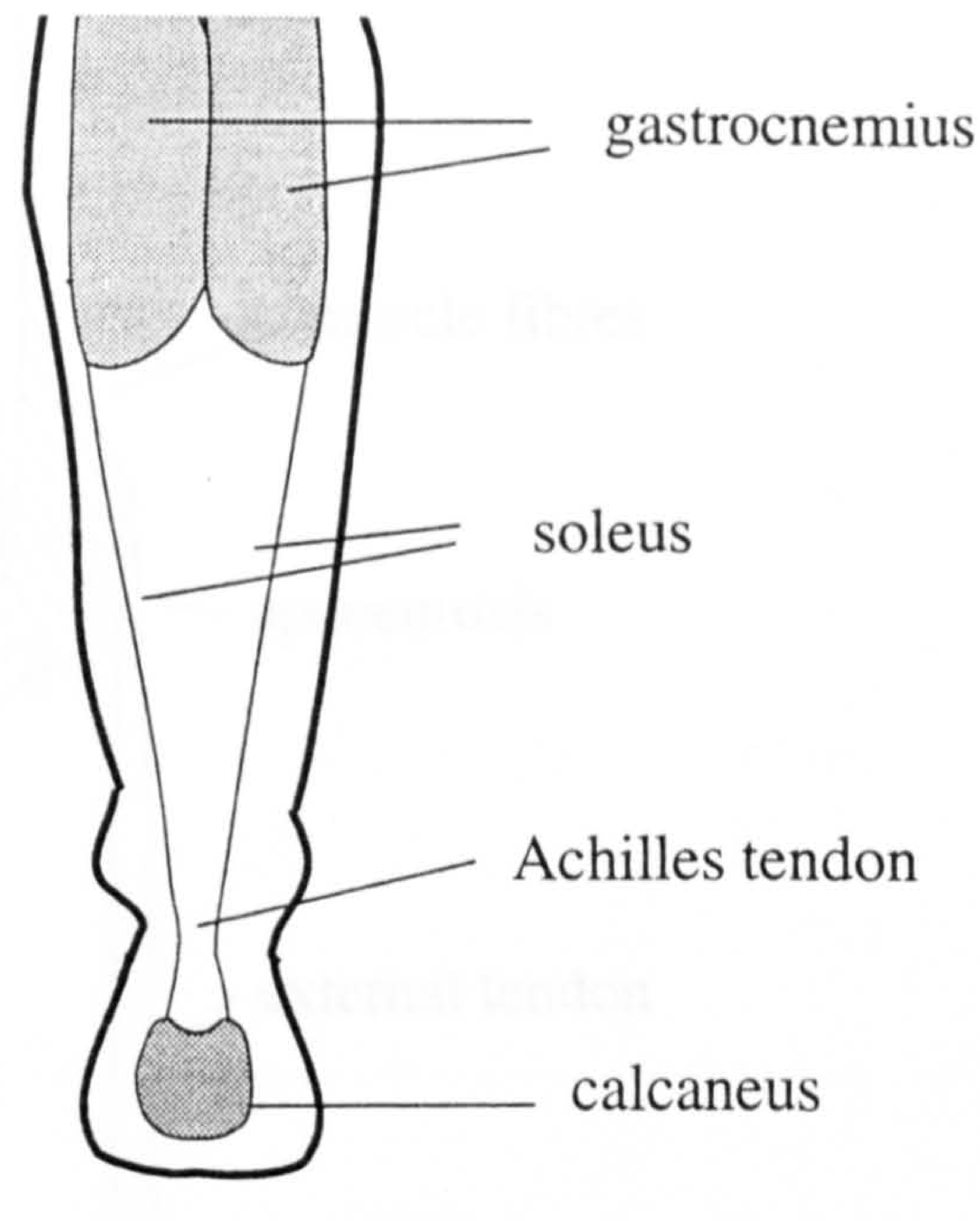


Figure 2.1 The Achilles tendon (rear view)

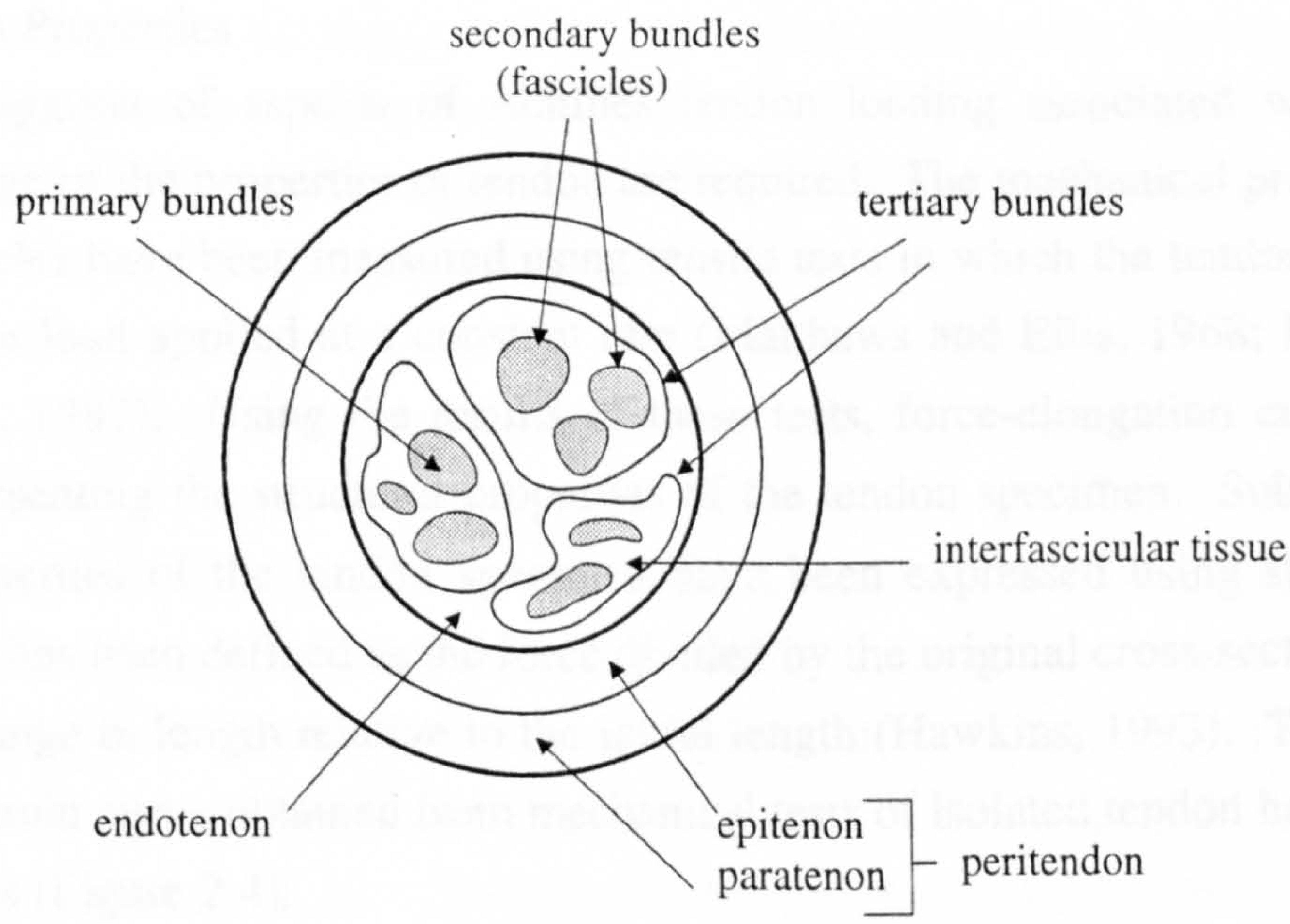


Figure 2.2 Cross-section of the Achilles tendon

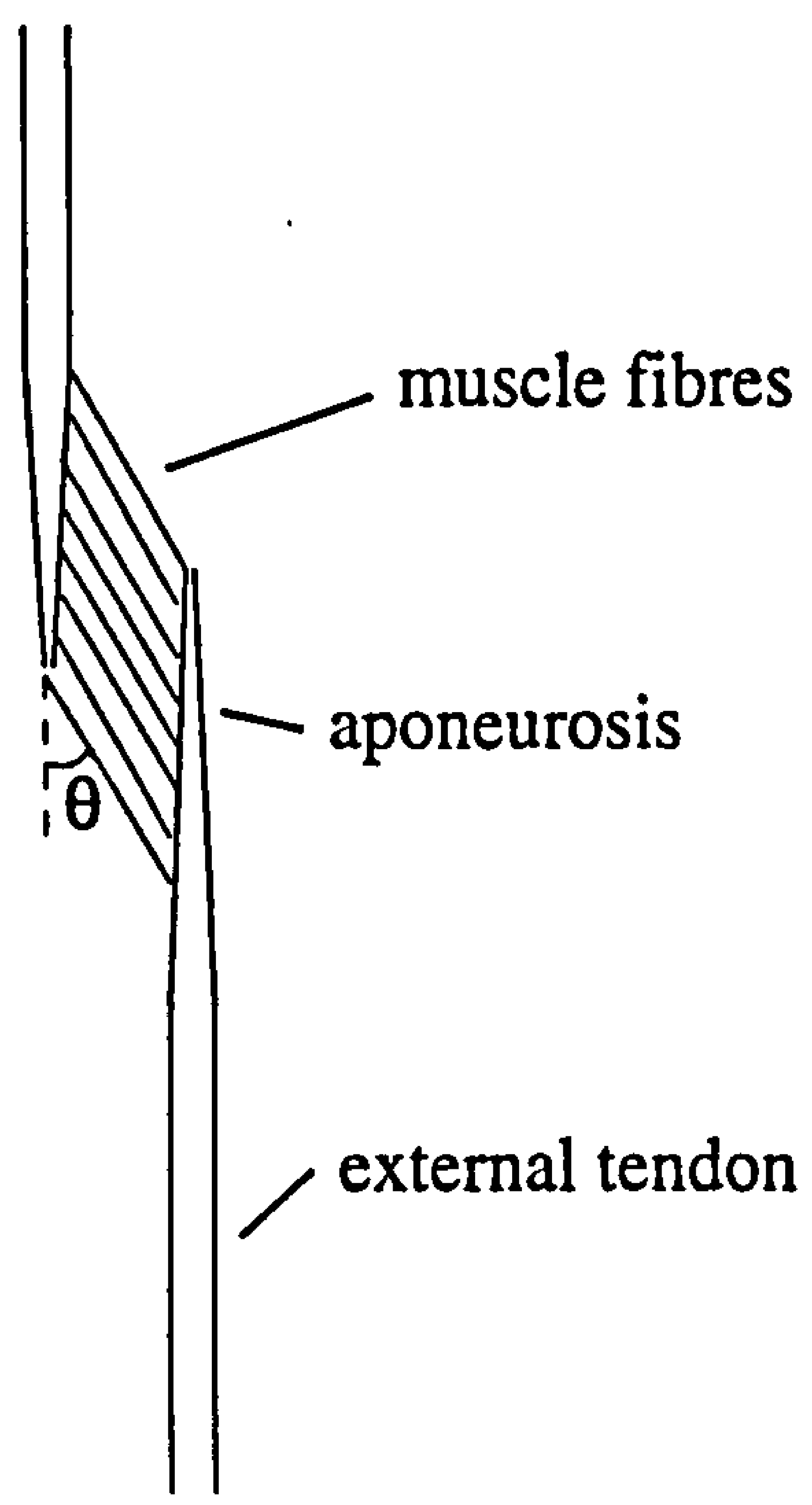


Figure 2.3 Muscle-tendon complex for a unipennate muscle

(iii) Isolated Tendon Properties

For the investigation of aspects of Achilles tendon loading associated with injury occurrence, knowledge of the properties of tendon are required. The mechanical properties of isolated tendon samples have been measured using tensile tests in which the tendon has been subjected to a tensile load applied at a constant rate (Matthews and Ellis, 1968; Ker, 1981; Proske and Morgan, 1987). Using the results of these tests, force-elongation curves have been obtained, representing the structural properties of the tendon specimen. Subsequently, the mechanical properties of the tendon specimen have been expressed using stress-strain curves, where stress has been defined as the force divided by the original cross-sectional area, and strain as the change in length relative to the initial length (Hawkins, 1993). The general shape of the stress-strain curve obtained from mechanical tests of isolated tendon has been the same between studies (Figure 2.4).

Three distinct regions are seen on the stress-strain curve. The first region, termed the 'toe' region, demonstrates strain deformation with little corresponding stress, and occurs from approximately 0 - 3% strain (Abrahams, 1967). The low initial stiffness of the tendon in this region has been suggested to correspond to the straightening of the wavy collagen fibres when loads are first applied (Rigby et al., 1959).

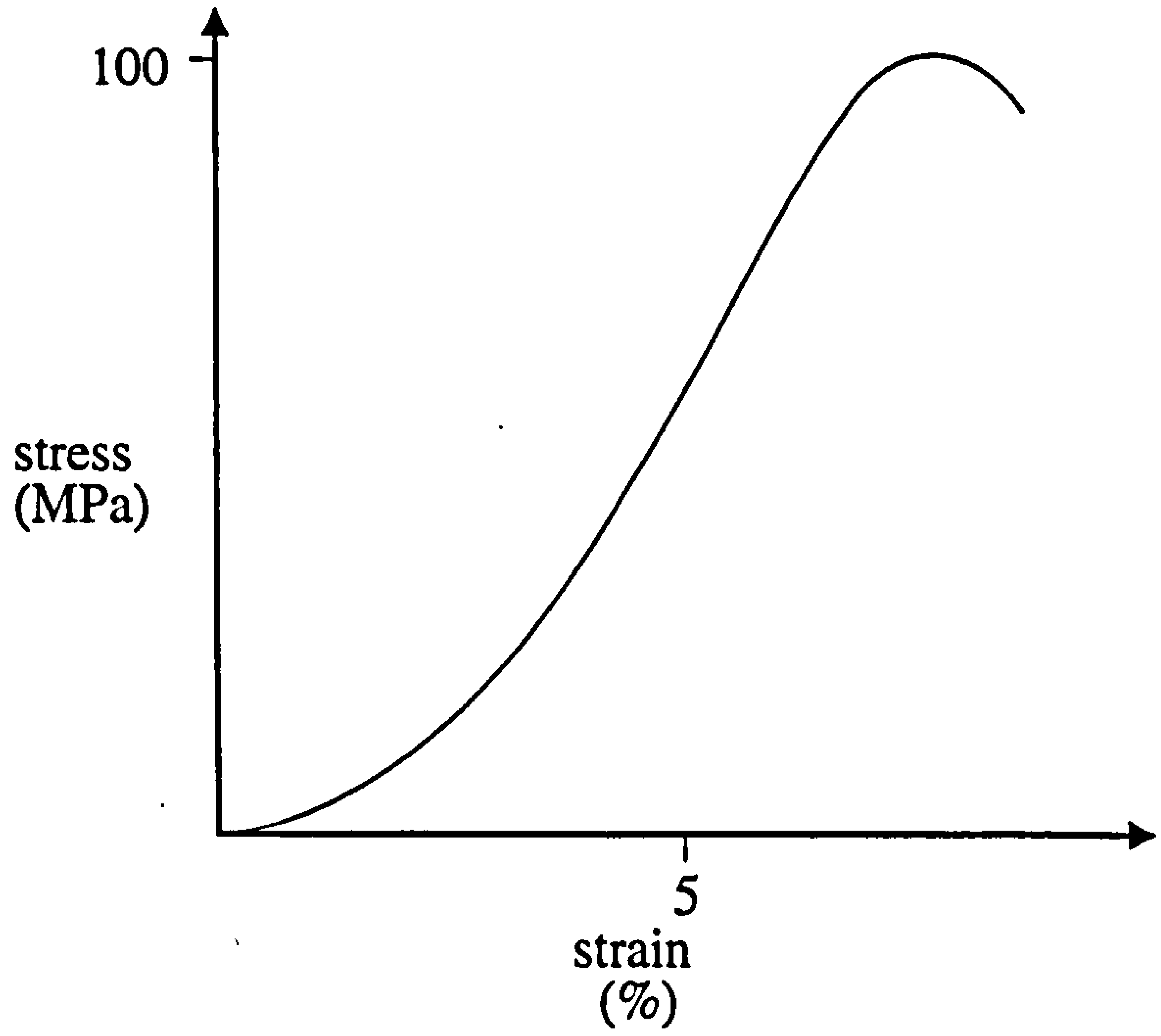


Figure 2.4 A typical stress-strain curve for isolated tendon (Abrahams, 1967)

Increased loading stiffens the fibres, resulting in a linear relationship between stress and strain from approximately 3% to 5% strain (Abrahams, 1967). Non-elastic deformation then occurs until failure. The slope of the linear portion of the stress-strain curve is commonly used to define the modulus of elasticity (Equation 2.1)

$$\text{modulus of elasticity} = \text{stress} \,/\, \text{strain} \qquad (2.1)$$

Tendon is a viscoelastic material, demonstrating time-dependent behaviour including stress relaxation under maintained stretch, creep under constant load, and mechanical hysteresis under cyclic loading and unloading (Schwerdt et al., 1980). As a viscoelastic material, tendon is sensitive to strain rate (Hawkins, 1993). Abrahams (1967) found that tendon mechanical properties were influenced by the rate of strain. Other authors have found that the rate of strain has not influenced tendon mechanical properties (Schwerdt et al., 1980; Ker et al., 1987). Differing ranges of strain rate or different testing procedures may have contributed to the discrepancies in behaviour across studies. Ker et al. (1987) suggested that the Achilles tendon is not sensitive to strain rate within the range of loading rates expected during human running. This suggestion was supported by Gregor (1993).

Tendon ultimate stress and strain, modulus of elasticity, and stiffness have been obtained from isolated tendon testing (Benedict et al., 1968; Schwerdt et al., 1980; Ker, 1981). Large variations in these values have been demonstrated between studies. Possible reasons for these variations have been identified as the method of testing, the use of tendon from different species (Table 2.2), the level of activity of the mammal from which the tendon was obtained (Butler et al., 1984), the age of the donor from which the tendon was obtained (Yamada, 1970), and the method of storage (Matthews and Ellis, 1968).

The percentage strain at which failure of the tendon begins to occur has been found to vary from 5% to 10% (Table 2.1). Typical stress and strain values for tendon measured using in-vivo procedures are presented in Table 2.2, with a range of results demonstrated across studies. Differing magnitudes for modulus of elasticity of tendon have been presented (Table 2.3). Ker et al. (1988) suggested the region of the stress-strain curve corresponding to the stresses typically experienced by the tendon during activity should be used to obtain modulus of elasticity values. For example, these authors quoted an estimated stress of 53×10^6 N.m^2 in the human Achilles tendon during running (Table 2.2). A stress of this magnitude falls in the linear region of the stress-strain curve, indicating that the gradient of this region should be used as the modulus of elasticity. This finding suggests there is a linear relationship between stress and strain for the Achilles tendon during running.

Varying stiffness values for tendon have been presented in the literature (Table 2.3). Stiffness of tendon has been shown to be dependent on force, with an initial region of low stiffness, corresponding to the 'toe' region of the stress-strain curve, followed by an approximately constant stiffness value (Proske and Morgan, 1987). The values presented in Table 2.3 refer to constant stiffness values calculated for tendon.

In order to obtain stress values corresponding to a particular tendon force, it is necessary to measure the cross-sectional area of the tendon. For isolated tendon samples, methods used have included measurement of wet and dry weight per unit length, wet and dry displacement per unit volume, use of an area micrometer, shadow amplitude contour reconstruction, use of vernier callipers, and measurements of histological sections. Ker et al. (1987) calculated a mean cross-sectional area for the human Achilles tendon of 89 mm^2. Abrahams (1967) described how it can be assumed that the cross-sectional area of the tendon remains unchanged during movement, as only small increases in length of tendon occur when transmitting large forces. Thus measurements of tendon cross-sectional area from a static subject may be combined with Achilles tendon force values for the determination of tendon stress.

Table 2.1 Typical ultimate stress and strain values for tendon

Source	Breaking Stress (N.m^{-2})	Breaking Strain (%)	Tendon
Abrahams (1967)	34.5×10^6	5 - 6	typical human
Bennett et al. (1986)	100×10^6	8	typical human
Yamada (1970)	56×10^6	10	human Achilles
Ker et al. (1988)	67×10^6	5.2	human Achilles

Table 2.2 Typical stress and strain values measured in-vivo

	Max. Stress ($N.m^{-2}$)	Max. Strain (%)	Tendon	Activity
Ker et al. (1987)	53×10^6		human Achilles	running
Komi (1990)	111×10^6		human Achilles	running
Alexander et al. (1982)	18×10^6		camel running	'pacing'
Herrick et al. (1978)		10	horse	galloping
Dimery et al. (1986)	$28\text{-}75 \times 10^6$		deer	galloping

Table 2.3 Stiffness and modulus of elasticity values

	Stiffness ($N.m^{-1}$)	Mod. of Elasticity ($N.m^{-2}$)	Tendon
Rigby et al. (1959)		1000×10^6	rat tail
Benedict et al. (1968)	250×10^3		various human
Ker (1981)	785×10^3	1650×10^6	sheep
Bennett et al. (1986)		1500×10^6	various

(iv) Identification of Achilles Tendon Injury

Inflammation of the peritendon, termed peritendinitis, has been associated with a swelling of the paratenon (Puddu et al., 1976). Subotnick and Sisney (1986) described how adhesions form between the paratenon and the tendon, causing pain and stiffness. Pathology within the tendon itself, termed tendinosis, has symptoms including tenderness and swelling, and has been associated with limited passive ankle dorsi-flexion (Puddu et al., 1976). Achilles tendon rupture may also occur, with complete failure of the tendon. This occurrence has been associated with the sustaining of an exceptionally large force.

Subotnick and Sisney (1986) highlighted the importance of recognising the type of Achilles tendon injury before treatment. Ultrasonic and magnetic resonance imaging have been used as diagnostic tools for differentiating between types of Achilles tendon injury (Williams, 1993).

(v) Etiology of Achilles Tendon Injury

Several different factors have been implicated as contributors to or causes of injury to the Achilles tendon in runners, although the exact cause of this injury remains unclear. A combination of factors are likely to be responsible. Factors which have been associated with Achilles tendon injury occurrence are described in the following sections.

Repeated loading

The repeated loading of the Achilles tendon during running has been cited by many authors as a likely cause of this injury (Leach et al, 1981; Subotnick and Sisney, 1986). The suggestion that lower extremity injuries are caused by repeated loading has been supported by evidence that repeated forces of a relatively low magnitude have been shown to have a degenerative effect on the cartilage of sheep (Radin, 1982). Impact forces in particular have been implicated, as highlighted by Radin et al. (1991) who found that subjects experiencing anterior knee pain demonstrated a higher loading rate of vertical ground reaction force than a group of control subjects.

The suggestion that impact forces are related to Achilles tendon injury has been supported by the successful treatment of this injury through the use of shock absorbing heel lifts (MacLellan and Vyvyan, 1981). However, it has been shown that external loading conditions do not necessarily reflect the internal loading. For example, it has been demonstrated that the peak Achilles tendon force in running occurs in the middle of the stance phase rather than at initial ground contact (Komi, 1987; Winter and Bishop, 1992). Consequently, it appears unlikely that changes in the initial impact force directly influence the maximum force experienced by this tendon.

Archambault et al. (1995) described a suggested mechanism for the development of tendon overuse injuries (Figure 2.1). Tendon has been found to adapt to exercise by breaking down structurally, before remodelling with an increased strength. During this process, a period of temporary weakness is suggested to exist, indicating that if the tendon is loaded again without sufficient rest time, the structure may be susceptible to tissue injury. Archambault et al. (1995) further described how, once injured, the tendon may not return to its normal structure and strength.

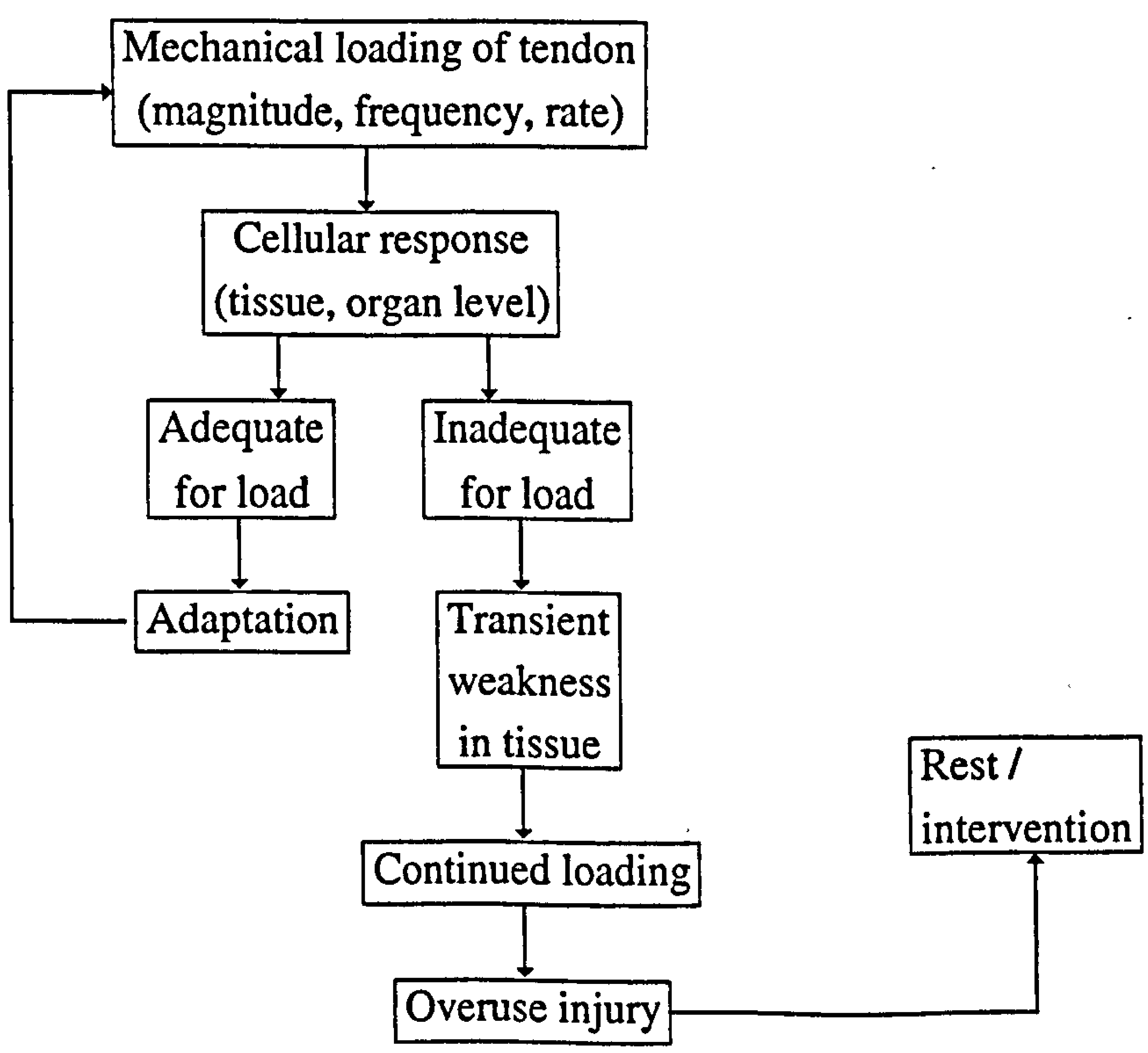


Figure 2.1 Proposed relationship between tendon loading, cellular response and overuse injury (Archambault et al., 1995)

Running shoes

The possibility of ill-fitting shoes rubbing and causing inflammation of the paratenon, or aggravation of an existing Achilles tendon injury has been described (Subotnick and Sisney, 1986; Grisogono, 1989). It has also been suggested that a stiff shoe sole increases the length of the moment arm of the ankle joint about the point of resultant ground contact, causing an increase in the net muscle moment about this joint (Winter and Bishop, 1992). This has been suggested to lead to an increased loading of the Achilles tendon (Clement et al., 1984).

Inadequate shoe heel height has been suggested to cause injury by increasing the Achilles tendon force and stretch (Clement et al, 1984). Running shoes with insufficient medial-lateral support in the heel area have also been implicated, with Smart et al. (1980) describing how soft, loose heel counters and narrow heel bases do not provide adequate stability for the subtalar joint (see Section 2.4).

Lower extremity kinematics

Aspects of an individual's lower extremity biomechanics, particularly the kinematics of the ankle and subtalar joints, have been implicated. A description of these joints is provided in Section 2.4. Increased range and rate of pronation have been identified as causes of injury to the Achilles tendon (Pagliano, 1987a).

Winter and Bishop (1992) described how, in an individual who demonstrates 'excessive' pronation, the external rotation generated by knee extension may conflict with the internal rotation associated with pronation. Clement et al. (1984) speculated that these conflicting rotatory forces on the tendon result in 'wringing out' of the vessels in the tendon and peritendon, causing vascular impairment and subsequent injury. Nigg (1995, personal communication) has suggested that the uneven distribution of stress across the tendon resulting from the conflicting rotations at its proximal and distal ends may cause high local stresses and subsequent injury. No literature has yet been published to support these suggestions.

Excessive supination during the push-off phase of running may increase stress on the structures controlling this motion, including the Achilles tendon.

Other factors

Clement et al. (1984) and Leach et al. (1981) have described how inadequate flexibility of the triceps surae muscles increases the strain on the Achilles tendon during running, as the tendon is required to contribute an increased proportion of the overall stretch of the muscle-tendon complex. Clement et al. (1984) described the possible influence of fatigue of the triceps surae muscle group on Achilles tendon injury occurrence. It was speculated that glycogen depletion of these muscles decreases their ability to function adequately during the eccentric contractions of the shock absorption phase of support, causing increased stretch of the Achilles tendon.

Smart et al. (1980) suggested that a high volume of training or sudden increase in the training intensity, may result in injury. Smart et al. (1980) also described how a sudden return to running after a period of relative inactivity may cause Achilles tendon injury. These authors described how a decrease in activity is likely to cause a decrease in the vascularisation of the tendon. When returning to training, an insufficient blood supply may fail to meet the nutritional requirements of the tendon, resulting in degeneration and possibly tendon rupture.

Several rheumatic conditions, such as gout, have been implicated in the development of Achilles tendon injuries (Smart et al., 1980). A large bodyweight and height have been associated with overuse injuries, although the relevance of these factors in Achilles tendon injury is not known (Kvist, 1994). Increased temperature in the tendon has been associated with Achilles tendon injury occurrence (Kvist, 1994). However, Ker (1981) demonstrated that theoretically this was not possible. Increased Achilles tendon injury occurrence has been demonstrated with age (Marti et al., 1988). Kvist (1994) described how leg length discrepancy may be associated with this injury, although no significant relationships have been demonstrated. An increased prominence of the calcaneal superior tuberosity has been related to inflammation of the distal part of the Achilles tendon. The Achilles tendon is thought to rub on the tuberosity, especially when shoes with a low heel height are worn (Kvist, 1994, Reinschmidt and Nigg, 1995).

(vi) Treatment of Achilles Tendon Injuries

Conservative measures

The initial use of conservative measures including ice, massage, oral anti-inflammatories, contrast baths, regular stretching and reduced activity has been recommended, particularly in the treatment of peritendinitis (Leach et al., 1981; Clement et al., 1984; Subotnick and Sisney, 1986; Davidson and Taunton, 1987; Williams, 1993).

Orthoses and shoe inserts

Orthoses are shoe inserts designed to control foot function during stance to provide support around the ideal neutral position. The aim has been to eliminate or reduce abnormal motions and stresses in the foot and lower extremities. The measurement of lower extremity movement is described in Section 2.4. The successful treatment of Achilles tendon injury using orthotic devices has been documented (Clement et al, 1984; Pagliano, 1987b). However, no relationships between this injury and particular interventions have been established (Kvist, 1994; Kilmartin and Wallace, 1994). To increase knowledge of the influence of orthoses, studies investigating the influence of these devices on forces in internal structures are required.

In an attempt to reduce the stress and strain on the Achilles tendon, the use of shoe inserts providing increased heel lift or increased shock absorption has been recommended (Davidson and Taunton, 1987; Clement et al., 1984; Leach et al., 1981, MacLellan, 1984). Shoe inserts are generally not considered to be a form of orthoses, as they are not designed to encourage a neutral foot during the middle of the stance phase. Smart et al. (1980)

recommended a heel height of 12 mm to 15 mm in all shoes. Leach et al. (1981) described the use of a small felt pad of thickness 6 mm to 10 mm placed in the rear of the shoe in addition to the heel lift of the shoe. Leach et al. (1981) suggested that increasing the height of the heel in relation to the forefoot during midstance reduces the maximum ankle dorsi-flexion during ground contact, thus reducing the strain on the Achilles tendon. Clement et al (1984) suggested the maximum force experienced by the Achilles tendon was reduced by raising the heel. It has also been suggested that the decreased dorsi-flexion will result in a decrease in the range of pronation, since dorsi-flexion is a component of this movement (Clement et al., 1984).

MacLellan and Vyvyan (1981) found the incidence of Achilles tendon pain in sports participants was reduced by the use of shock absorbing heel inserts placed in the rear of the shoe. Fauno et al. (1993) demonstrated the frequency of Achilles tendon injury was less in soccer referees using shock absorbing heel inserts, than those wearing only their normal footwear during games. These authors noted the mechanism by which Achilles tendon injury was reduced was not clear, since the lifts provided both increased shock absorption and heel lift. Lowden et al. (1984) found no evidence to support the use of shock absorbing heel lifts in the treatment of Achilles tendon injury. The simultaneous provision of increased shock absorption and increased heel lift may contribute to the discrepancy demonstrated in results across studies.

Shoe inserts have been found to influence selected aspects of lower extremity biomechanics such as tibial acceleration (Light et al., 1980; Bojsen-Moller, 1983), ground reaction force peak loading rate (Lees and McCullagh, 1984), and peak rearfoot movement (Stacoff and Kaelin, 1983). However, scientific evidence correlating the reduced incidence of Achilles tendon injury with biomechanical variables has not been presented. The clinical studies described in this section have been performed by experienced practitioners whose knowledge in treating Achilles tendon injury cannot be ignored. Various shoe inserts have clearly contributed to the successful treatment of Achilles tendon injury, but the mechanism by which these interventions have been successful is not clear.

Other methods

Steroid injections have been employed following unsuccessful use of conservative measures and shoe inserts (Leach et al., 1981; Williams, 1993). Smart et al. (1980) described how steroids can relieve the symptoms of a partial rupture to allow an athlete to continue activity, but this increased activity may result in rupture of the tendon. Leach et al. (1981) described how steroid injections in an inflamed tendon may slow the natural healing process and weaken the tendon, increasing susceptibility to tendon rupture.

When all other treatments have failed, surgery has been used (Leach et al., 1981; Subotnick and Sisney, 1986; Williams, 1993). Leach et al. (1981) described different surgical procedures including stripping of inflamed paratenon and removal of calcium deposits or an inflamed retrocalcaneal bursa from the tendon.

2.3 Achilles Tendon Loading

Methods are required for the determination of the Achilles tendon stress and strain during running, for comparison with ultimate stress and strain values obtained from the testing of isolated tendon specimens, and for the investigation of the behaviour of tendon in-vivo.

(i) Direct Measurement of Achilles Tendon Force

Direct methods have been used to measure the force experienced by the Achilles tendon in running. Komi (1990) implanted a buckle transducer around the tendon, and obtained maximum Achilles tendon force values of 12.5 BW for a subject running at 6 $m.s^{-1}$. Fukashiro et al. (1993) used a buckle transducer in analysis of continuous two-legged hopping, a movement involving similar muscle action to running, and obtained values of 5 BW for each Achilles tendon.

The use of this invasive technique for the measurement of internal forces has not been widespread in studies on human subjects. It is limited to use with tendons such as the Achilles, where the tendon is sufficiently distant from bone and other structures. The actual presence of the transducer around the tendon may affect the action of the tendon and thus the measured force. Difficulties are encountered during calibration of these transducers. In animal studies, at the end of the experimental procedure it is possible to isolate the muscle/tendon complex under investigation and calibrate the transducer by applying known forces to the complex. Clearly this is not practical when using human subjects. The calibration methods of Komi (1990) have involved the application of known moments about the ankle joint in the sagittal plane by the use of a pulley system. This results in the system being calibrated to record the magnitude of the total force acting about the joint, rather that the force solely due to muscle action of the triceps surae group. Komi (1990) justified this by reference to the dominant action of this group about the ankle joint.

Despite the limitations involved with the use of direct measurement of Achilles tendon force using transducers, these methods have a potential in the evaluation of other techniques. For example, Fukashiro et al. (1993) compared in-vivo measurements of Achilles tendon force with tendon force estimates obtained using inverse dynamics calculations. These authors found close agreement between the forces obtained using the two methods.

(ii) Indirect Measurement of Achilles Tendon Force

Due to the practical and ethical problems encountered with direct measurement techniques, indirect methods for estimation of tendon forces have been developed. In particular electromyography (EMG) and inverse dynamics methods have been utilised. EMG provides a measure of the electrical activity of a muscle. This EMG measurement indicates the input to muscle action, but the resulting output also depends on the mechanical properties of the muscle. Investigators have developed relationships between integrated EMG (iEMG) and muscle force for the prediction of force from iEMG measurement (Hof et al., 1983). Under static conditions, iEMG has been shown to be linearly proportional to the force of contraction of the muscle (Lippold, 1952). Under dynamic conditions the relationship between EMG and muscle force output is more complex, since the relationship between these variables will vary throughout a particular movement.

The difficulties described in using EMG measurements to estimate tendon forces has limited the widespread use of these techniques (Gregor, 1993). More common has been the use of inverse dynamics procedures. For estimation of Achilles tendon forces, these techniques involve the calculation of net muscle moment about the ankle joint, determination of the proportion of this moment that can be attributed to the triceps surae muscle group, and measurement of Achilles tendon moment arm about the ankle joint centre (Burdett, 1982; Scott and Winter, 1990).

Ankle moment

Determination of net moment about the ankle joint has been achieved by use of inverse dynamics procedures (Winter, 1983; Scott and Winter, 1990) and quasi-static methods (Burdett, 1982; Morlock and Nigg, 1988). Calculation of ankle joint moment using inverse dynamics requires the quantification of the foot mass and centre of mass, and the moment of inertia of the foot segment about the principal axes. Different methods have been described for the quantification of these inertia characteristics.

Historically, the first attempts to determine these values was by measurement of cadavers. Braune and Fischer (1889), Dempster (1955), Clauser et al. (1969), and Chandler et al. (1975) used cadaver studies to determine mass, volume, density, and centre of mass values for selected body segments, and for the total body mass. Alternative experimental methods presented in the literature include volume contour mapping, computerised axial tomography, the quick release method, the relaxed oscillation method, the gamma-scanner method, and magnetic resonance imaging (Nigg, 1994). Hatze (1980) suggested that, although these experimental techniques appear appropriate for determination of segment mass, centre of mass and volume, theoretical methods are more suitable for the calculation of principal moments of inertia and orientations of the principal axes since measurements are taken from the individual subjects. Using theoretical methods, the body has been represented as a mathematical model, allowing the inertia properties to be determined mathematically (Hanavan, 1964; Jensen, 1978; Hatze, 1980; Yeadon, 1989). These techniques generally involve taking measurements from subjects for input into the model.

Inertia data that have been obtained using experimental methods are easy to utilise in inverse dynamics calculations, but do not provide data that are specific to the subject under study. Theoretical methods provide personalised inertia data, but require the time-consuming process of taking measurements from individual subjects. For example, 95 anthropometric measurements are required for the model of Yeadon (1989). An approach which provides relatively easy determination of inertia characteristics for an individual is the use of regression equations, developed from measurements taken from a group of subjects. This technique was applied by Zatziorsky and Seluyanov (1985). Using anthropometric data obtained from 100 subjects using the gamma ray scanner technique, a series of regression equations for mass, centre of mass, and moments of inertia about the principal axes were developed. Subject mass and height are required as input into the regression model for determination of inertia values for an individual.

The choice of an appropriate method for determination of inertia characteristics depends on the requirements of the study. Nigg (1994) described how the obtaining of accurate inertia data for a subject is important if absolute joint, muscle or tendon forces are to be calculated. However, accuracy of these measures is less important if the aim of the study is to make comparisons of results obtained for one subject under different conditions. The choice of method will also depend on the subject characteristics. For example, if the subject is an athlete, inertia data measured from young sports people is likely to be more suitable than that obtained using cadaver data from relatively old, untrained individuals.

The determination of accurate inertia data has been shown to be of less importance for ankle joint moments than for moments about joints higher up the body. Alexander and Vernon (1975) found that inertia effects had a negligible influence of less than 2 N.m (1%) on peak ankle moments during running. Burdett (1982) justified neglecting inertia effects in a study of running, arguing that the foot has a small mass and moment of inertia, and that throughout the stance phase of running the foot experiences small accelerations. Morlock and Nigg (1988) demonstrated that omitting inertia effects had a negligible influence on the net moment about the ankle joint in running, even during impact with the ground, when accelerations are high. These authors termed the resulting moments 'quasi-static'. Alexander and Vernon (1975) described how inertia effects are likely to be large immediately following ground impact, indicating that quasi-static methods may not be suitable for analysis of the initial impact phase. The suitability of quasi-static methods is likely to depend on the subject and conditions, and should therefore be assessed for individual studies.

A joint moment tending to cause flexion is generally given a positive sign, and a moment tending to cause extension of the joint is given a negative sign. Thus, for human locomotion in two dimensions moving in a direction from left to right, an anti-clockwise ankle moment causing ankle dorsi-flexion is taken to be positive, and a clockwise moment causing ankle plantar-flexion negative. It has been demonstrated that beyond the first 10% of stance there is a negative muscle moment about the ankle joint in running (Winter, 1983; Scott and Winter, 1990). Theoretically, this negative moment can be balanced mathematically by the use of a single muscle group acting to cause plantar flexion. In practice, the main plantar flexion muscle group of the ankle joint is the triceps surae group. Other muscles with the ability to contribute to plantar flexion are the plantaris, flexor hallucis longus, flexor digititorum longus, tibialis posterior, proneus longus, and proneus brevis.

Distribution problem

In order to estimate the force in the Achilles tendon, it is necessary to determine the relative contribution of the different plantar-flexion muscles to the resulting moment. The various methods developed to quantify this relative contribution of agonist muscles acting across a particular joint include EMG, systematic elimination, optimisation and simplification.

EMG measurements have been used to provide an indication of the relative contribution of each muscle to a particular movement, and provide evidence for the elimination of inactive muscles to reduce the number of unknown muscle forces. Scott and Winter (1990) demonstrated using EMG methods that the activity levels of each of the plantar-flexion muscles follow similar patterns of activity during running, and the antagonist muscle, the tibialis anterior, demonstrates a relatively low level of activity. These authors used this evidence to support the assumption that negligible antagonist muscle action occurs about the ankle joint during ankle plantar-flexion in running.

Systematic elimination involves the selection of random combinations of muscles to solve the equations of motion. Multiple solutions are obtained, but generally many of these can be eliminated. For example, Burdett (1982) calculated the ankle joint force in running using the assumption that only two tendons were exerting force at any time, with every possible combination of two tendons being considered.

Optimisation methods involve the use of the assumption that the body's central nervous system minimises certain physiological quantities when generating movement. The function to be minimised is known as the objective or cost function. Seireg and Arvikar (1973) calculated muscle forces in the lower extremity during walking using muscle and joint forces and moments. An objective function which minimised a function of the muscle forces and resultant moments was used. The functional and neuromuscular control responses of joints depend on many factors, such as psychological motivation, training status, fatigue, morphometric properties, and pathological conditions. Therefore, optimisation of a single factor is unlikely to satisfy all requirements. It is also unlikely that an objective function developed for one particular activity will be suitable for another activity.

Various simplifications can be used to ensure that a series of determinate equations are obtained when estimating internal forces. The simplest of such methods involves the assumption that the action of specific muscle groups is negligible, thus reducing the number of unknowns. For example, Alexander and Vernon (1975) assumed that the plantar-flexion moment during running is generated entirely by the triceps surae muscle group. An alternative simplification is to combine muscles into functional groups according to their proximity to each other and their action about the joint in question. In this manner, the number of unknowns can be reduced. The resulting force results will refer to the force in the equivalent muscle, and thus a method is required to determine the relative contribution of the individual muscles. The most straight forward such approach is to assume that the contribution of each individual muscle is a constant percentage of the total force. Relative muscle mass (Inman, 1947), anatomical cross-sectional area (Procter and Paul, 1982) and physiological cross sectional area (Burdett, 1982; Scott and Winter, 1990) have been used to determine percentage contributions. Physiological cross-sectional area is the volume of a muscle divided by the average length of its muscle fibres. This method of providing a measure of the number of fibres contributing to the force is suggested to be more appropriate than anatomical cross-sectional area, since most muscle fibres do not extend the full length of the muscle (Burdett, 1982). By assuming that there is equal stress in each of the relevant muscles, the magnitude of force exerted by each muscle group can be estimated using physiological cross-sectional area.

Different figures have been provided across studies regarding the percentage contribution of the triceps surae group to the total physiological cross-sectional area of the ankle plantar-flexors. Alexander and Vernon (1975) provided a percentage of 65%, compared with 80% by Murray et al. (1976) and 85% by Scott and Winter (1990). Scott and Winter (1990) demonstrated that the effect of a reduction of the contribution from 85% to 65% required an impossible increase in the stress of the remaining plantar-flexors. These authors also stated that if energy expenditure is considered, then the muscles with the largest moment arm about the ankle joint centre would provide greater force outputs than those provided using other estimates, since a muscle with a longer moment arm has greater potential to contribute to the total moment acting about the joint centre. Since the triceps surae group has the largest moment arm, a percentage of greater than 85% was therefore indicated. For the determination of absolute Achilles tendon stress values, the use of a percentage contribution of at least 85% appears to be appropriate. For the comparison of Achilles tendon forces across different conditions for a single subject, the chosen percentage contribution of the triceps surae group to ankle plantar-flexion will not influence results. Thus, as it is not possible to determine exactly what the relative contribution of this group is, the assumption of a 100% contribution is acceptable and convenient for comparison across conditions.

Moment arm estimation

In order to calculate muscle and tendon forces from joint moments it is necessary to measure the length of the moment arm of the muscle or tendon about the joint centre. For estimation of Achilles tendon force, a method for calculation of Achilles tendon moment arm about the ankle joint centre is therefore required. This length is defined as the perpendicular distance between the ankle joint centre of rotation and the line of action of the Achilles tendon, and thus depends on the location of the ankle joint centre and the line of action of the Achilles tendon. It has been demonstrated that the centre of rotation of the ankle joint is not at a fixed point (Engsberg, 1987; Siegler et al., 1988). Rugg et al. (1990) and Gregor et al. (1991) have found that there is no significant difference in moment arm lengths obtained using a fixed and moving ankle joint centre of rotation. The determination of a fixed ankle joint centre of rotation therefore appears to be adequate for measurement of Achilles tendon moment arm length.

In general, a fixed centre of rotation represented by a marker on the most prominent point of the lateral malleolus has been used in sagittal plane studies to represent the centre of rotation of the ankle joint (Scott and Winter, 1990). The variation in location of the ankle joint centre obtained across studies is described in Section 2.4.

Several authors have assumed a constant moment arm length throughout the range of ankle joint movement. Morlock and Nigg (1991) assumed a constant moment arm length of 0.06m in running. Bauman (1981) assumed a constant moment arm of 0.05 m during sprinting. For the stance phase of running, variation in Achilles tendon moment arm length obtained using anatomical data has been demonstrated with ankle angle (Burdett, 1982; Scott and Winter, 1990).

Bruggemann (1985) placed a single marker on the tendon and the moment arm was determined by measurement of the distance from a marker representing the centre of rotation of the ankle joint and the Achilles tendon marker. It was assumed that the line joining the two markers was normal to the tendon line of action. This would not have been the case throughout the movement, resulting in errors in moment arm estimation.

Magnetic resonance imaging techniques have been used to calculate Achilles tendon moment arm lengths. Rugg et al. (1990) used magnetic resonance imaging (MRI) techniques to investigate the degree of error in moment arm estimations caused by the use of a constant moment arm length. The foot was positioned over a range of ankle angles and sagittal plane images of the bones, muscles, tendons and ligaments were obtained, allowing direct measurement of the moment arm lengths from the image. Rugg et al. (1990) found that there was a 20% change in moment arm length when moving from a position of maximum dorsi- to maximum plantar-flexion. A 20% error in moment arm measurement will result in a 20% error in Achilles tendon force calculation, assuming no change in net ankle moments. Thus the use of a constant moment arm length will clearly result in significant errors in subsequent force calculations. The accurate determination of moment arm length is clearly important over the range of joint angles experienced in running.

Cadaver data have been used to obtain information on the point of insertion of the Achilles tendon and the line of pull for different orientations of the foot in relation to the lower leg (Brand et al., 1982; Burdett, 1982; Procter and Paul, 1982). Burdett (1982) found that the angle between the line of pull of the Achilles tendon and a line representing the lower leg was less than 10 degrees throughout the range of ankle joint movement. Burdett (1982) used these findings to justify the approximation of the line of action of the Achilles tendon as being parallel to the lower leg. Taking a typical moment arm length of 0.028m, and ankle to tendon insertion distance of 0.03m, a 10 degree deviation in tendon line of action results in a 9% difference in estimated Achilles tendon force (Appendix A). Thus, due to the relatively small length of the Achilles tendon moment arm about the ankle joint, approximating the line of action as being parallel to the lower leg can clearly have a marked influence on estimated Achilles tendon forces.

Groh and Bauman (1975) combined knowledge of the anatomical locations of the origin and insertion points obtained from cadaver studies with x-rays of the lower extremity for the subject studied, to calculate lines of action of muscles and thus their respective moment arm lengths.

Brand et al. (1982) demonstrated large differences in origin and insertion points when comparing small and large cadavers. These authors found that the use of average or individual data resulted in large differences in moment arm measurements. Burdett (1982) also obtained significantly different moment arm lengths when comparing individual cadaver origin and insertion points to those obtained using average data (n=5). It therefore appears important that data on muscle points of origin and insertion are specific to the subject under consideration. Minimisation of error has been achieved either by use of techniques such as x-ray or MRI, by appropriate scaling of measurements, or by use of cadaver data from subjects similar to those under investigation.

An alternative method for calculation of the moment arm of the Achilles tendon line of action has been described by Bobbert et al. (1986). Using the methods of Grieve et al. (1978), the gastrocnemius muscle length was predicted from angular data for the knee and ankle joints. The methods of Grieve et al. (1978) allowed prediction of the change in gastrocnemius muscle length relative to the leg length at knee and ankle angles of 90 degrees. Bobbert et al. (1986) established a relationship between the rate of change of the length of the gastrocnemius muscle, the rate of change of the angle between the calcaneus and the lower leg, and the length of the moment arm of the gastrocnemius about the ankle joint. The moment arm of the gastrocnemius was then calculated as a percentage of the lower leg segment length. Using this method, the gastrocnemius muscle line of action was assumed to be a straight line from the point of origin to the point of insertion of the muscle. Since the Achilles tendon is a continuation of the gastrocnemius, the moment arm length of this muscle about the ankle joint was assumed to be equal to the Achilles tendon moment arm length. Thus, an alternative method for calculation of the Achilles tendon moment arm length, using leg length and angular data on the knee and ankle joints, has been described.

The studies of Burdett (1982) and Bobbert et al. (1986) have provided methods of generating subject-specific Achilles tendon moment arm data using selected anthropometric measurements. Burdett (1982) determined the point of insertion of the tendon from the average cadaver data, with the coordinates of this point being scaled by use of foot length, medial-lateral malleoli distance and vertical distance from the plantar surface of the foot to the origin of the coordinate systems used. In the study carried out by Bobbert et al. (1990) leg length and joint angles were used to estimate Achilles tendon moment arm lengths. These methods can be employed to obtain subject specific lengths for use in the estimation of Achilles tendon forces.

(iii) Estimation of Achilles Tendon Strain

The properties of isolated tendon samples have been described in earlier sections. Muscle and tendon are in series and function together. Komi (1990) described how increased knowledge of the relative contribution of muscle and tendon to overall length changes in the muscle-tendon complex during running is required to increase understanding of the relationship between tendon force and elongation, and thus stress and strain. Overall length changes of the triceps surae muscle-tendon complex have been estimated using ankle and knee joint angles, with ankle angle being found to be most influential (Grieve et al., 1978). An increased dorsi-flexion of the ankle joint results in an increase in the length of the overall muscle-tendon complex. For determination of Achilles tendon strain, the contribution of the Achilles tendon to this change in length is required. Hoffer et al. (1989) found using an ultrasound technique that the muscle fibres and associated tendon of the cat gastrocnemius did not shorten or stretch in phase. Rack and Westbury (1984) described how the distribution of change in length of the muscle-tendon complex will depend on the relative stiffness values of the muscle and tendon.

Under passive conditions, muscle fibres are more compliant than tendon, but when activated the muscle fibres have an increased stiffness. Although impact with the ground in running has been described as passive (Nigg, 1986), it has been suggested that activation of muscles occurs before ground contact in anticipation of impact (Bobbert et al., 1992). The activation level chosen will determine the stiffness of the muscles during the impact phase. No change in activation occurs until the termination of a finite time period of approximately 100 ms (Bobbert et al., 1992), corresponding to the middle of the stance phase in running. Muscle fibres exhibit an initial high short range stiffness when stretched, beyond which the stiffness is reduced (Proske and Morgan, 1984). Evidence in the literature indicates that the short range stiffness of muscle is exceeded during locomotion (Walmsley and Proske, 1981). The relative stiffness of muscle and tendon therefore appears to vary during the stance phase of running, demonstrating the complex nature of the determination of the relative contribution of tendon to the overall length change of the muscle-tendon complex.

The muscle fibres of the triceps surae group have been found to be between 60 mm and 70 mm in length, compared with an overall muscle-tendon length of between 400 mm and 500 mm (Hof et al., 1983). The maximum stretch of muscle fibres has been measured directly as between 1% and 2% (van Ingen Schenau, 1984), compared with a maximum stretch of tendon of at least 5% (Ker et al., 1988). Using a tendon length of 260 mm, and a maximum tendon stretch of 5%, van Ingen Schenau (1984) concluded that the stretch of the triceps surae muscle tendon complex observed during running is contributed to mainly by the Achilles tendon, with muscle fibres playing a negligible role. A tendon stretch of 13 mm was obtained for an ankle angle of 16 degrees dorsi-flexion. The negligible contribution of muscle fibres to changes in the length of the triceps surae muscle-tendon complex in running has been supported in the literature (Bobbert et al., 1986; Caldwell, 1995). Using the assumption that changes in ankle angle indicate changes in length of the triceps surae muscle-tendon complex, and the assumption that observed changes in length are contributed predominantly by the Achilles tendon, ankle angle changes may therefore be used to indicate changes in Achilles tendon strain.

An alternative approach to the estimation of Achilles tendon strain in running has been described by Ker et al. (1988). These authors investigated theoretically the maximum possible stress and strain of the Achilles tendon using relative mass and cross-sectional areas. A maximum Achilles tendon stress of 67×10^6 $N.m^{-2}$ was obtained. The corresponding strain was obtained from a tendon stress-strain plot, and was found to be approximately 5.2%. Since these authors have indicated that there is a linear relationship between Achilles tendon stress and strain during running, strain values can subsequently be calculated for running using estimated Achilles tendon stress values.

(iv) Achilles Tendon Loading Results

Several investigators have estimated the forces transmitted by the Achilles tendon during walking. Procter and Paul (1982) estimated the maximum force to be approximately 2.5 bodyweights (BW). Brewster et al. (1974, cited in Procter and Paul, 1982) obtained higher values of 3.5 BW. For the same activity, Groh and Baumann (1975) obtained values of between 2.2 BW and 3.1 BW.

Bruggemann (1985) calculated Achilles tendon force during the take-off of a flic-flac, obtaining maximum values of 16 BW. Smith (1975) calculated a maximum Achilles tendon force of 6.1 BW for a subject performing a drop jump from a height of 42 inches. Baumann (1981) quoted an approximate force of 6000 N acting on the Achilles tendon during sprinting. For a typical mass of 70 kg this force corresponds to 8.5 BW.

For running at 4.5 $m.s^{-1}$ (6 $minutes.mile^{-1}$), Burdett (1982) obtained maximum Achilles tendon force values of between 5.3 BW and 10 BW, using different sets of anatomical data. Scott and Winter (1990) obtained maximum Achilles tendon force values of between 6100 N and 6300 N, corresponding to an average of 9.2 BW, for three subjects performing running trials at an unspecified speed. Harrison et al. (1987) estimated the mean maximum force in the gastrocnemius for four subjects running at 4.5 $m.s^{-1}$ as 7.9 BW. Komi (1990) measured Achilles tendon force directly by use of an in-vivo force transducer. He quoted a maximum Achilles tendon force of 12.5 BW for a single subject performing barefooted running at 6 $m.s^{-1}$.

Fukashiro et al. (1993) compared directly measured Achilles tendon forces with estimated values, for an individual subject performing jumping activities. The jumps used were maximal vertical jumps with and without counter-movement, and submaximal two-legged hopping at preferred frequency. The maximum force values were obtained during hopping, with values of approximately 5 BW for both the directly measured and estimated forces. The value of 5 BW was for two-legged support and thus the Achilles tendons of each foot were sharing the total load. For a single-legged support, this value equated to 10 BW (2 x 5 BW).

The results of a number of studies estimating the force experienced by the Achilles tendon during running have been provided. Each of these studies have presented similar results for maximum Achilles tendon force, in the range of 8 BW to 12 BW.

Achilles tendon strain during running has been less frequently documented. Using the linear relationship between Achilles tendon stress and strain in running described by Ker et al. (1988), the stress during running quoted by these authors of 53 x 10^6 $N.m^{-2}$ corresponded to a strain of 4.1%.

2.4 Lower Extremity Kinematics

Kinematic data are required for input into the inverse dynamics equations for determination of ankle joint moments. These data are also required for comparison across conditions of kinematic variables that have been associated with Achilles tendon injury.

(i) Obtaining Kinematic Data

Data Collection

Cavanagh (1987) described how historically high speed cinematography has been employed to record body movement during running, allowing frame by frame assessment of the action. Coordinate data have subsequently been obtained by the digitisation of cine frames. Recently video has also been employed for collection of kinematic data. The advantages over cine-film include low cost of video tapes, the ease of use and the immediate availability of recordings. Although it has been found by some authors that there is a reduced accuracy obtained by use of video analysis due to the limited resolution and reduced image quality (Angulo and Dapena, 1992), steps have recently been taken to improve the measurement resolution and image quality provided by video systems (Kerwin and Challis, 1994). Kerwin and Challis (1994) demonstrated that comparable accuracy can be obtained using video and 16 mm cine-film, supporting the continued use of video.

A disadvantage of kinematic data collection using video is the slow sampling frequency provided by conventional recorders, with 50 Hz and 60 Hz being the commonly provided rates. Equipment for collection of high speed video recordings is becoming more readily available, although obtaining the data at increased sampling frequencies requires the laborious digitisation of a large number of data fields. The problem of digitisation can be overcome by the use of opto-electronic techniques (Selspot, MacReflex) and high speed video systems (Motion Analysis, Vicon), which automatically track body markers in space. These systems can provide increased data sampling rates, whilst avoiding the increased data analysis required when cine-film or manual video digitisation systems are used.

The sampling frequency for any data collection is based on the highest frequency in the signal. Williams et al. (1991) and Hamill et al. (1992) have demonstrated that the highest frequency in the signal of frontal plane rearfoot motion data is between 15 Hz and 16 Hz. The Nyquist theorem suggests that the sampling frequency should be at least two times the highest frequency (Winter, 1991). However, signal aliasing has been found to be a problem when sampling at exactly two times the highest frequency (Hamill et al., 1994). Oppenheim et al. (1983) have suggested that the sampling frequency for rearfoot data collection should be at least five times the highest frequency in the signal. Using the data presented in the literature, this indicates that a sampling frequency of at least 80 Hz be used for collection of rearfoot data. This suggestion has been supported by Hamill et al. (1994) who demonstrated identical rearfoot angle time histories when data were collected at 200 Hz and 100 Hz, whereas clear differences were detected when 50 Hz data were used. Similar considerations must be made for the choice of sampling rate for data collection in the sagittal plane.

Williams (1993) highlighted the different ground contact velocities calculated during running using 67 Hz data compared with 200 Hz data. These differences resulted from discrepancies in the detected time of foot contact due to the use of different sampling rates. Other variables, such as peak joint angles, were found not to be markedly influenced by this variation in sampling rate.

The choice of sampling frequency clearly depends on the requirements of the study in question, and the available equipment. The Achilles tendon acts predominantly in the sagittal plane, and the maximum Achilles tendon force occurs during the middle of the stance phase when sudden changes in force are not evident (Burdett, 1982; Komi, 1990). A sampling frequency of 50 Hz, obtainable using a conventional video recorder, may therefore be appropriate for analysis of maximum Achilles tendon force. However, if frontal plane rearfoot data or Achilles tendon loading close to impact are of interest, an increased rate is required.

Several authors have demonstrated differences in lower extremity kinematics with running speed (Nigg, 1986; Milliron and Cavanagh, 1990). It is therefore important to specify and/or control the speed of running. If a treadmill is employed, then running speed is easily controlled. For running trials over ground, the time taken to cover a specified distance can be monitored and the average running speed over this distance subsequently calculated (Cavanagh and Lafortune, 1980; Cole et al., 1995). An envelope of acceptable running speeds is normally specified. For example, Cole et al. (1995) employed a running speed of 4.5 ± 0.2 m.s^{-1}, with any trials outside of this envelope being rejected. The running speed most commonly used in running studies is 3.83 m.s^{-1} (7 min.mile^{-1}) (Milliron and Cavanagh, 1990).

Direct Linear Transformation (DLT)

A video or cine-camera provides a two-dimensional image of the recorded action. For two-dimensional analyses the plane being considered must be clearly defined. Appropriate scaling methods are required to obtain the locations of the points of interest. The earliest method of scaling was by using a 'multiplier', a scaling factor obtained by the placement of an object of known length in the movement plane (Cureton, 1939). If the movement is not planar, or the camera is not positioned as required, a systematic error, known as projection or perspective error, will occur.

If three-dimensional analysis of a movement is considered to be necessary, data from several two-dimensional images can be combined. During digitising, coordinates of the points relative to the origin of the digitising tablet or monitor screen are obtained. The object space coordinates have therefore been transformed first to image coordinates and then to digitiser coordinates. To obtain the object space co-ordinates of a selected point, it is necessary to transform digitised points to image co-ordinates and then to object space co-ordinates. A method known as Direct Linear Transformation (DLT) has been developed to do these transformations in one step (Abdel-Aziz and Karara, 1971). This method involves the estimation of a series of parameters to correct and scale the digitised data. Six external camera parameters and five internal camera parameters are required. These eleven parameters are estimated by the use of a calibration volume with at least six points marked at measured locations, positioned in the volume of interest.

The DLT procedure can also be used to obtain two-dimensional data. A calibration plane, rather than a volume, is employed. The coordinates provided by the two-dimensional DLT are the locations of the object space coordinates projected onto this plane. The axis normal to the plane of interest is set to zero in the DLT, and calibration coefficients are calculated as for the three-dimensional procedure. If a two-dimensional analysis is performed in the yz plane, x is set to zero. This reduces the equations provided by Abdel-Aziz and Karara (1971) to two equations with eight unknowns (Equation 2.2, Equation 2.3).

$$q = (P1Y + P2Z + P3) / (P7Y + P8Z + 1) \tag{2.2}$$

$$r = (P4Y + P5Z + P6) / (P7Y + P8Z + 1) \tag{2.3}$$

(q,r) are two-dimensional digitiser coordinates
(Y,Z) are object space coordinates
Pi (i=1 to 8) are the calibration coefficients

At least four points are required in the calibration plane. Following digitisation of these points, at least four pairs of (q,r) and (Y,Z) will be available to solve Equation 2.2 and Equation 2.3 for the eight unknowns. Where more than four points are available, a least squares approach can be used to solve the equations. Once the eight calibration coefficients have been calculated by use of the known two-dimensional coordinates of the calibration plane, Equation 2.2 and Equation 2.3 can be rearranged to allow reconstruction of unknown points from digitised coordinates. Unlike the two-dimensional techniques described earlier, the image plane is not required to be parallel to the plane of movement, since the object space coordinates are projected onto the two-dimensional plane that has been calibrated. This technique is therefore preferred for obtaining two-dimensional kinematic data.

Smoothing and differentiation

All methods employed for the collection of kinematic data introduce an amount of noise. This noise may not be evident in the displacement data, but it will be amplified if the data are differentiated to obtain velocity and acceleration values. In addition to the true signal, the sampled signal will consist of both systematic and random noise. The influence of systematic noise can be reduced by identification of the source, and subsequent removal or reduction of this noise. Sources of systematic error include incorrect marker placement, skin and marker movement and errors in calibration. Random noise remains in the signal, and thus some form of data processing is required to reduce this noise.

It has been demonstrated in studies of human movement that noise tends to be of a higher frequency than the frequency components of the true signal (Winter et al., 1974; Lesh et al., 1979). Methods for removal of the high frequency components have therefore been developed. The importance of selecting procedures most appropriate for the data being analysed has been highlighted (Lees, 1980). Filtering can be described as any process which selectively removes high frequency components from a signal. The Butterworth filter has been commonly used in analysis of human locomotion (Winter et al., 1974; Pezzack et al., 1977; Bobbert et al., 1986).

Smoothing techniques have also been developed for the reduction of noise. The smoothing of a signal is considered to be equivalent to the use of a low-pass filter. In the biomechanics literature the terms filtering and smoothing tend to be used interchangeably. The use of least squares polynomials to determine the best fit curve through data points, and the use of Fourier series, have been described, (eg. Wood, 1982). A disadvantage of polynomial techniques has been their inability to adequately fit regions of varying complexity. For example, Zernicke et al. (1976) demonstrated that a polynomial curve over-smoothed rapid changes in the data. The use of spline functions for data smoothing has provided a method of fitting data with varying curvature in different time regions (Zernicke et al., 1976; McLaughlin et al., 1977). A spline is a series of polynomials joined together to represent a signal. The polynomials are of the same order and join at positions called knots. Challis (1991) described how the Reinsch (1967) cubic spline has been used regularly in biomechanics. This routine has knots at all data points and the degree of smoothing is adjusted using a smoothing parameter which controls the closeness of fit to the raw data. The data are represented as (x_i,y_i), where x_i are the time steps and y_i are the corresponding data points. A smoothing function with continuous second derivative $g''(x)$ is fitted through the data points such that the constructed function $g(x)$ minimises the integral of the approximate curvature (Equation 2.4), subject to the constraint defined by Equation 2.5. δy_i corresponds to the expected error in y_i.

$$\int_{x_1}^{x_n} (g''(x))^2 \, dx \tag{2.4}$$

$$\sum_{i=1}^{n} (g(x_i) - y_i)^2 \,/\, \delta y_i^2 \leqslant S \tag{2.5}$$

Comparisons have been made between splines of different orders regarding their ability to provide displacement, velocity and acceleration data. In particular cubic and quintic splines have been compared. Several authors have demonstrated that cubic splines are adequate for smoothing of displacement data, but quintic splines have been favoured for obtaining second derivatives (Wood and Jennings, 1979; Challis, 1991). The end points of the second derivative of a cubic spline are zero, causing a tapering of data when approaching these end points. The problems with the second derivative being set to zero for cubic splines can be lessened by collecting additional data either side of the time period of interest. The number of additional points required is likely to vary depending on the data and the sampling frequency, as indicated by the differing suggestions in the literature. For example, at least 20 additional points at each end of the sample has been recommended by McLaughlin et al., (1977), whilst Zernicke et al., (1976) suggested that three points are sufficient. The number of points required is likely to be influenced by the movement being analysed and the sampling frequency of data collection. Quintic splines have been found to sometimes produce artifacts when interpolating (van den Bogert and Glossop, 1996). To avoid erratic behaviour, the use of a cubic spline appears most suitable for smoothing and interpolation of displacement data, particularly if accelerations are not required.

When using filtering, a cut-off frequency is specified above which all data are discarded. Data published in the literature may be employed for specification of a required cut-off frequency. For example, Winter et al. (1974) have quantified the frequency of noise during human walking, and recommended the use of a specified cut-off in all similar studies. Ideally the frequency of noise in the data of the study being undertaken should be identified, rather than obtaining typical values from the literature.

For the selection of the required amount of smoothing, some investigators have used visual inspection of curves before and after smoothing has been applied (Wood, 1975; Vaughan, 1982). More commonly, the degree of smoothing is specified by varying the value of a smoothing parameter according to the amount of expected error in the signal (δy_i, Equation 2.5). Setting this value to zero provides interpolated data. However, it has been recommended that a value close to zero is more appropriate if interpolation is required, since erratic behaviour may result from forcing the spline through every data point (Yeadon, 1981). Automated techniques have been developed for selection of an appropriate smoothing parameter for a specified set of data (Woltring, 1986). Challis (1991) argued that automatic procedures based on mathematical methods are preferred, primarily owing to the repeatability of these techniques. However, by definition, automatic techniques will have limited flexibility.

McLaughlin et al. (1977) stressed the importance of quantifying the amount of random noise in data, for selection of the required amount of smoothing. Repeated digitisation of data points was described as a possible method for identification of the required amount of smoothing, using the mean standard deviation over typical data points. This procedure provides the average variation in data and can therefore be used to represent the level of precision. The estimated error value can be employed in the Reinsch spline to control the closeness of fit (δy_i, Equation 2.5). The approach of performing repeated digitisations is time consuming, but can also be used as an alternative method of filtering data (Challis, 1991). If sufficient repeated digitisations are performed, the random error in the data will be reduced to zero, providing data containing negligible random noise.

An alternative approach to the use of a single method to obtain displacement, velocity and acceleration values has been to smooth the displacement data and subsequently differentiate this data using finite difference techniques to provide velocity and acceleration values (Smith, 1975; Lees, 1980). Miller and Nelson (1973) described the use of a two-point finite difference algorithm (first central difference), or a four point algorithm (second central difference) for calculation of velocity. The use of prior smoothing of the displacement data was recommended. For calculation of acceleration data, several different finite difference methods have been described. Lanczos (1957) suggested five-, seven- and nine-point procedures that may be employed. Lees (1980) compared the use of these different procedures, describing how prior smoothing was not always necessary when using the seven- and nine-point methods.

Wold (1974) described the importance of identifying exactly what information is required from the collected data prior to making a choice of the most appropriate tool for extracting this information. Cubic splines appear to be appropriate for the determination of smoothed displacement data and interpolation. Finite difference techniques following prior smoothing, or quintic splines, appear more suited to the obtaining of accelerations.

(ii) The Ankle and Subtalar Joints

The triceps surae muscle-tendon complex has origins at the knee and lower leg, and inserts on the rear of the calcaneus of the foot. Thus, the loads experienced by the Achilles tendon will vary with changes in the orientation of the leg relative to the foot. Movement of the foot in relation to the lower leg is a result of motion at the ankle and subtalar joints (Figure 2.6). The ankle joint provides the connection between the tibia bone of the lower leg and the talus bone of the foot. The subtalar joint provides the connection between the talus and calcaneus bones. It has an oblique axis, allowing movement in the three cardinal planes of the body.

Combined motion about the ankle and subtalar joints is described using the terms pronation and supination. During running, the foot generally makes initial ground contact with the lateral edge and rear part of the sole. Following initial ground contact, the foot rolls inwards onto the medial side of the foot (eversion). The toes move outwards away from the midline of the body (abduction), and the distance between the toes and the lower leg is decreased (dorsi-flexion). Pronation is the word used to describe this simultaneous eversion, abduction and dorsi-flexion of the foot relative to the lower leg (Figure 2.7).

During the push-off phase, the foot rolls outwards so that ground contact moves from the medial to the lateral surface of the sole (inversion). The toes move inwards towards the midline of the body (adduction), and the toes move downwards away from the lower leg (plantar-flexion). The combined movement of inversion, adduction and plantar-flexion is known as supination and is the opposite motion to pronation (Figure 2.8).

Pronation and supination are required components of running gait. Pronation upon ground contact allows adaptation to varying surfaces and also contributes to shock absorption. Supination allows the foot to act as a rigid lever towards the end of the stance phase to propel the runner forward and upward. Injury occurrence through pronation or supination is only likely when the rate or range of movement is excessive, increasing the stress placed on the structures of the leg and foot.

The ankle joint has been approximated as a hinge joint with an axis passing from side to side in the frontal plane (Gray, 1858). Detailed cadaver study of the ankle axis orientation has provided differing descriptions. Scott and Winter (1990) described the axis as being normal to the sagittal plane and passing 2.2 cm below the lateral malleolus. Inman (cited by Burdett, 1982) described the ankle joint axis as being on a line passing less than 1 cm distal to the tips of the most prominent parts of the medial and lateral malleolus, and less than 1 cm anterior to the tip of the lateral malleolus. Hicks (1953) found that the ankle joint axis had a differing orientation depending on whether plantar-flexion or dorsi-flexion was occurring. For dorsi-flexion, a straight line passing from a point 1.5 cm anterior to the tip of the medial malleolus to a point 0.5 cm inferior to the tip of the lateral malleolus was described. For plantar-flexion, a line passing from a point 1.5 cm anterior and 1 cm below the tip of the medial malleolus to a point 0.5 cm above the lateral malleolus was described. The most common method for approximating the ankle joint axis is as a straight line normal to the sagittal plane, passing through the most prominent point of the lateral malleolus (Scott and Winter, 1990).

Large variations have been found between subjects in the orientation of the subtalar joint axis (Inman, 1976; Engsberg, 1987). Inman (1976, cited by Cavanagh, 1990) provided mean data from 46 cadaver feet on the orientation of the subtalar joint. These data indicated an axis inclined 23 degrees medially from the midline (standard deviation 11 degrees) and tilted 42 degrees upward (standard deviation 9 degrees). These orientations are illustrated in Figure 2.9.

Inman and Mann (1973, cited in Cavanagh, 1990) represented motion about the subtalar joint as a mitred hinge inclined at an oblique angle, indicating that internal/external rotation of the leg is related to subtalar joint movement (Figure 2.10). This representation has been questioned recently, with Engsberg (1987) and Siegler et al. (1988) both describing translational as well as rotational movements at the subtalar joint.

Siegler et al. (1988) examined the relative contribution of the ankle and subtalar joints to plantar/dorsi flexion, abduction/adduction and inversion/eversion. The ankle joint was found to contribute primarily to plantar/dorsi flexion, whereas the subtalar joint was shown to contribute to movement in all three cardinal planes, with a relatively large contribution to inversion/eversion.

The oblique orientation of the subtalar joint means that pronation/supination cause internal/external rotation of the leg. Pronation is accompanied by internal rotation of the tibia, whilst supination is accompanied by external rotation of the tibia. Knee flexion is accompanied by internal rotation of the tibia, and knee extension is accompanied by external rotation of the tibia. Generally, pronation and knee flexion occur simultaneously, and supination and knee extension occur simultaneously. This relationship has been described by Siegler et al. (1988) as 'kinematic coupling'. Differences in the timing of pronation/ supination and knee flexion/extension result in contradictory rotations of the tibia.

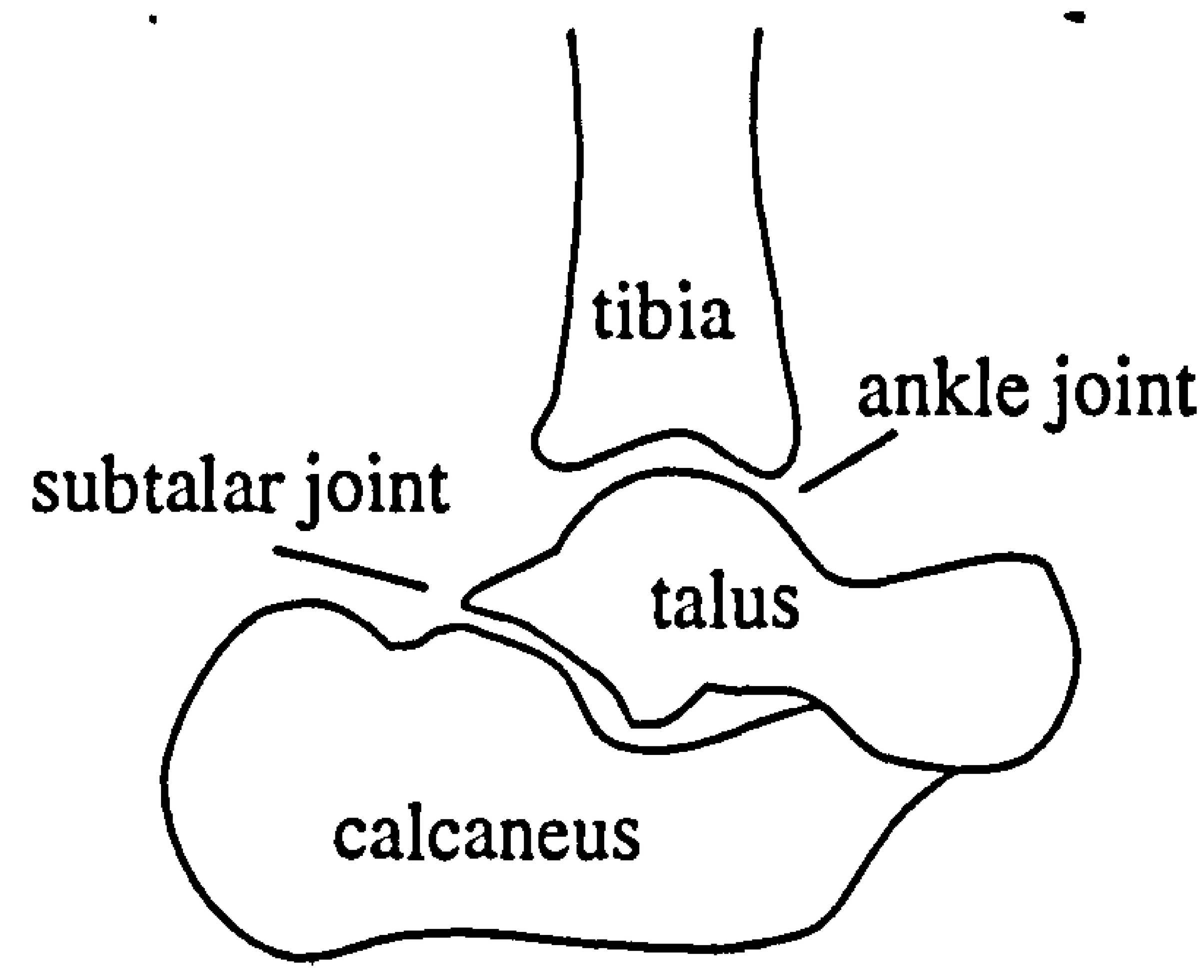


Figure 2.6 Sagittal plane view of the ankle joint and the subtalar joint

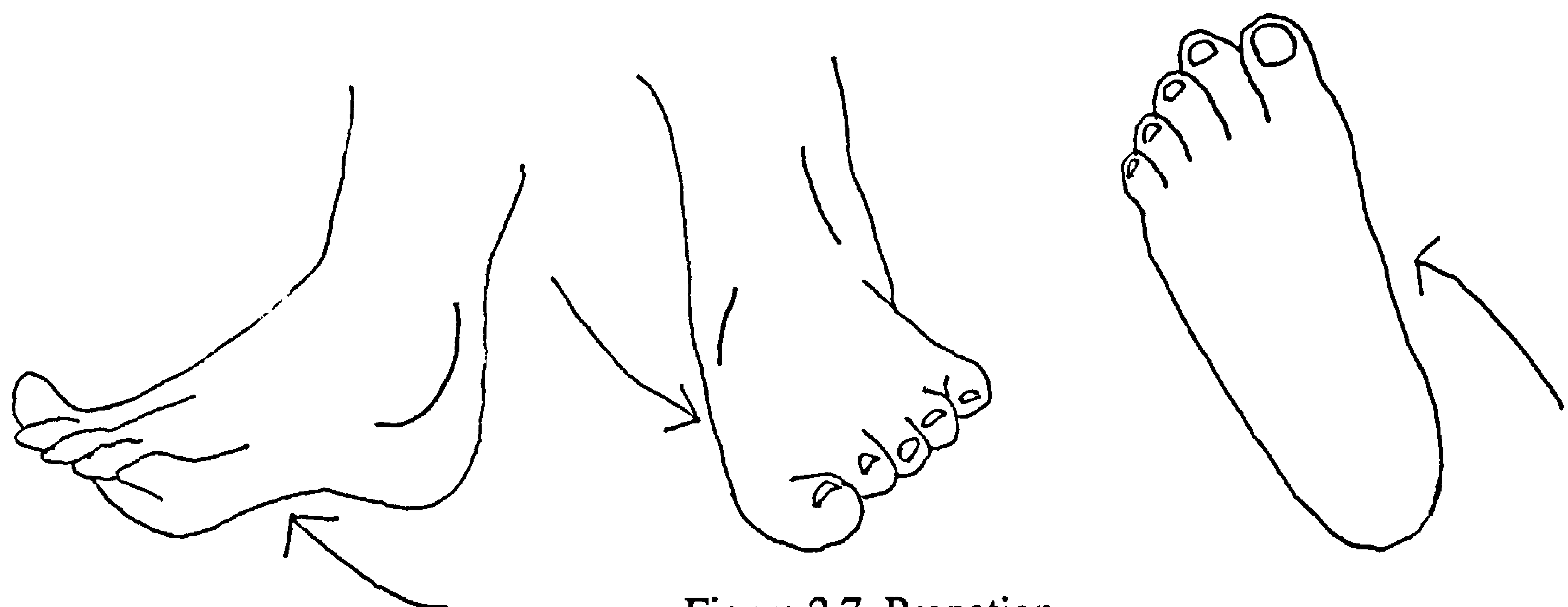


Figure 2.7 Pronation

Figure 2.8 Supination

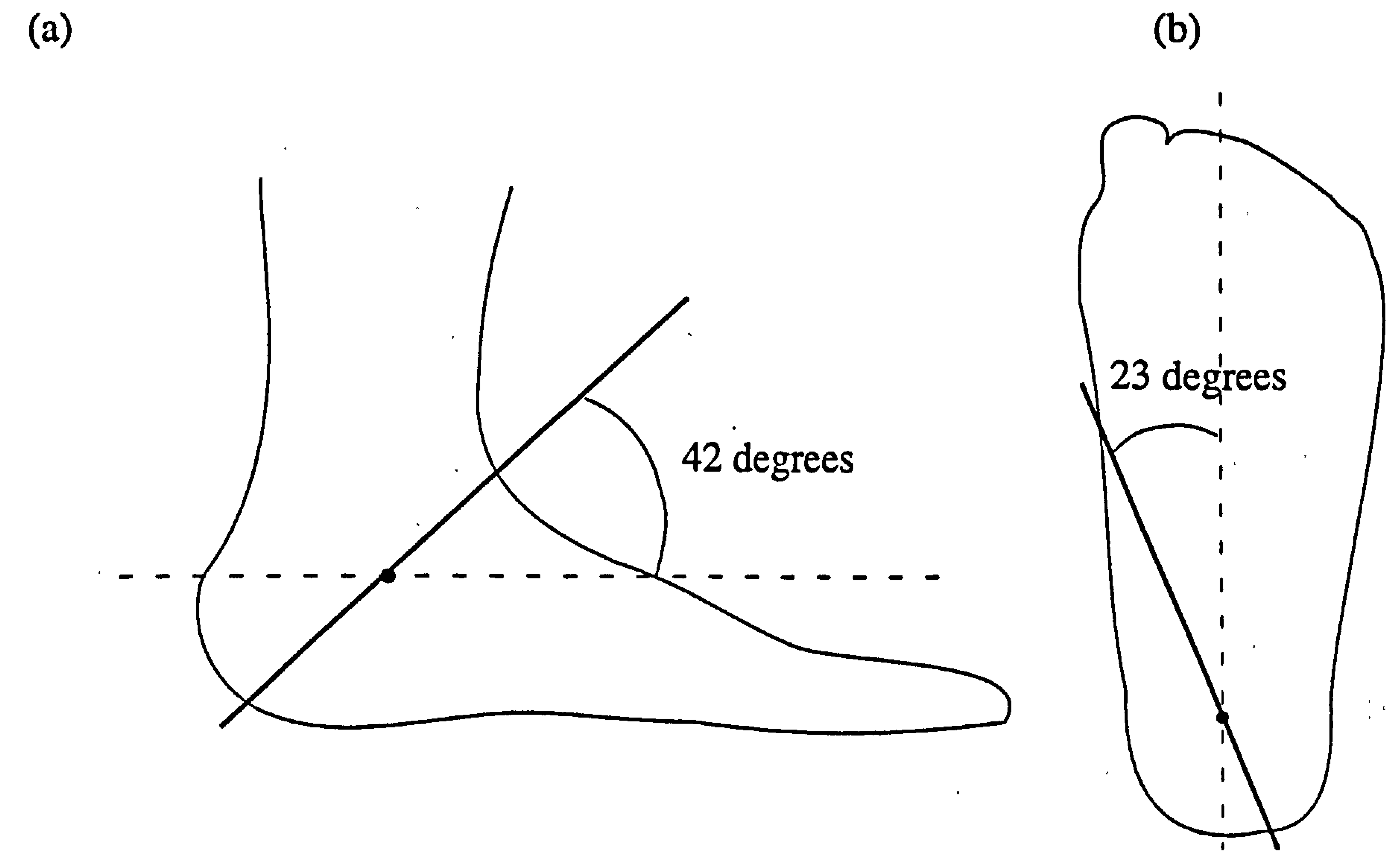


Figure 2.9 Mean orientation of the subtalar joint axis in the sagittal plane (a) and the transverse plane (b)

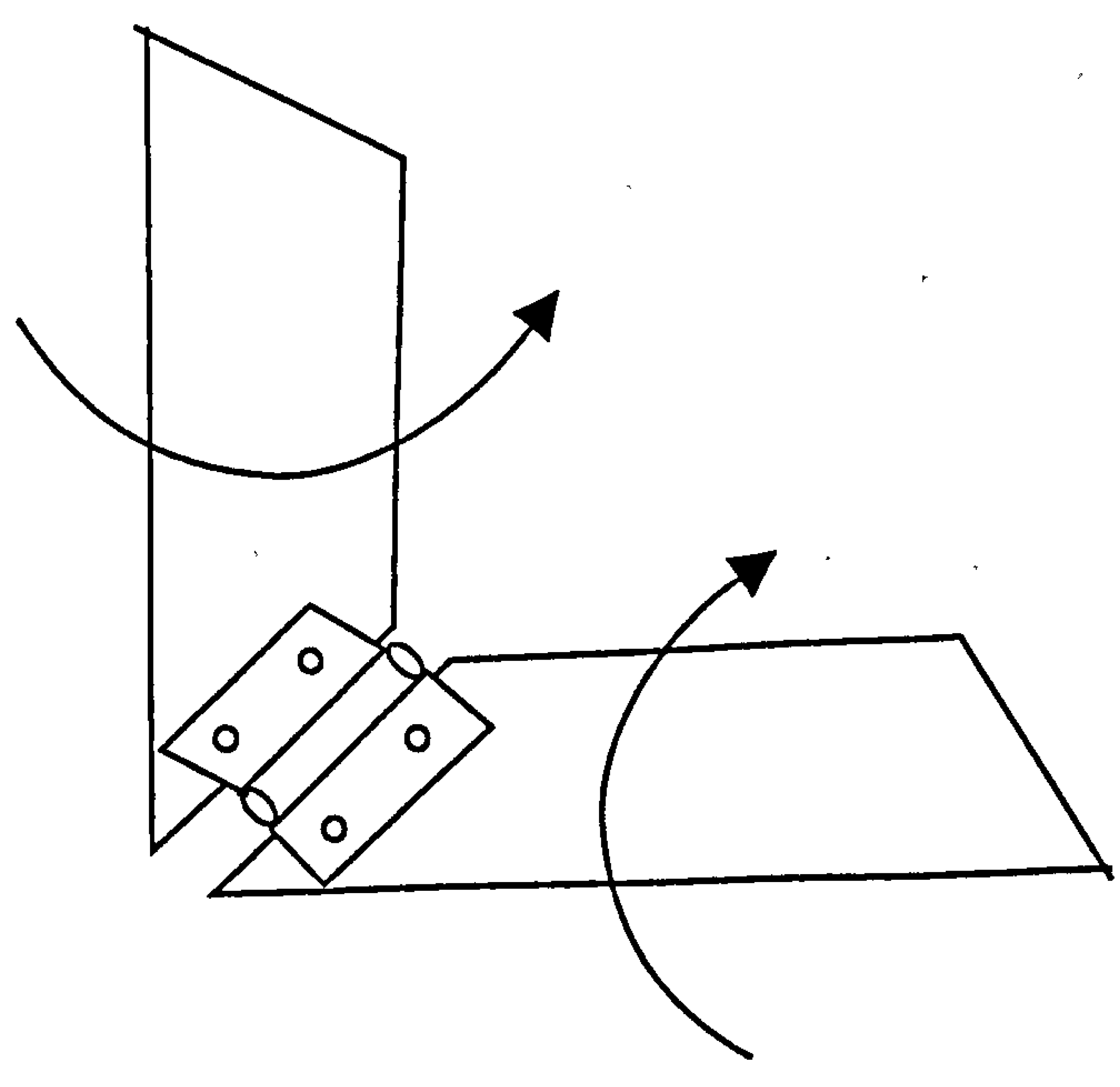


Figure 2.10 Mitred hinge representation of subtalar joint motion

(iv) Frontal Plane Kinematics

Rearfoot Motion

Movement about the subtalar joint is difficult to quantify due primarily to the oblique joint axis. It is apparent that three-dimensional analysis is required for monitoring of the full movement of this joint. However, a generally accepted procedure has been established which involves the quantification of pronation/supination of the subtalar joint by approximation of the motion to the two-dimensional frontal plane measurement of inversion/eversion of the calcaneus relative to the lower leg. During foot-ground contact in running, eversion is always accompanied by a dependent amount of dorsi-flexion and abduction, and thus eversion has been assumed to be an accurate estimator of pronation (Clarke et al., 1983a). Eversion/ inversion is the component of subtalar joint motion which is the most independent of motion at other joints and is the simplest to measure (Edington et al., 1990). Studies of frontal plane movement usually use the expression 'rearfoot motion', rather than 'subtalar joint motion'.

The degree of error resulting from the two-dimensional approximation of subtalar joint motion has been investigated (Soutas-Little et al., 1987; Arebald et al., 1990). It has been demonstrated that results from the two-dimensional technique are influenced by projection errors, dependent on the alignment of the body segments with the plane of filming. Arebald et al. (1990) found that alignment errors were smallest during midstance, indicating that the two-dimensional approach is most suited to analysis of this phase of stance.

There is no universally accepted method for measurement and representation of rearfoot motion parameters. Several subtly different methods have been adopted by different investigators, the differences being in marker placement, angles specified in results and standardisation between subjects and between tests. In 1992, the ASTM sub-committee F08.54 established a working committee to recommend standards for the testing of stability of athletic footwear. Hamill et al. (1994) detailed the standards pertaining to the two-dimensional measurement of rearfoot motion for comparison of stability of selected running shoes. For marker placement, the methods described by Clarke et al. (1984) were recommended. Hamill et al. (1994) also discussed the angle conventions in the literature, and made recommendations on the standards that should be used to allow comparison of results between subjects and between studies. The most commonly used convention in the literature was found to be that of Clarke et al. (1983b). It was recommended that variables concerned with the maximum pronation angle should be reported, as indicated by Arebald et al. (1990), since other times during stance the lower leg is not in the frontal plane, and so measured angles will not represent rearfoot motion. The suggested variables for representing rearfoot motion included: maximum pronation angle; maximum leg angle; maximum heel angle; and time to maximum pronation.

The methods described by Clarke et al. (1983b, 1984) involved the use of four markers to monitor rearfoot motion. Two markers were placed on the lower leg during standing, with the distal marker on the midline of the Achilles tendon between the medial and lateral malleolus, and the other 20 cm above on a line bisecting the leg, the markers being joined to represent the lower leg orientation. Identification of the bisection of the leg was kept consistent by use of an adjustable clamp. The clamp was placed around the knee from behind allowing determination of the geometric centre of the joint in the frontal plane. A string was then used to join this centre to the marker on the centre of the Achilles tendon. The proximal leg marker was placed on this line 20 cm above the Achilles marker.

Two markers were placed on the shoe on a line approximating the bisection of the posterior of the calcaneus, and these markers were joined by a straight line to represent the orientation of the calcaneus. Clarke et al. (1983b) standardised the position of the heel markers by the use of a repeatable stance position during marker placement. Wooden blocks were used as guides to ensure the feet were placed externally rotated by 7 degrees and with 5 cm between the heels. Markers were then placed vertically on the shoe heel and joined to form a line representing the orientation of the calcaneus. The angles that the leg and calcaneus lines made with the vertical were calculated, and the difference between these two angles was termed the rearfoot angle. Thus, a measure of the orientation of the rearfoot in relation to the lower leg was provided.

Different marker conventions, calibration positions and terminology have been employed in the literature (Edington et al., 1990; Nigg, 1986a), but the general principles have been consistent with those of Clarke et al. (1983b). It is important to be aware of the methods used in marker placement and in terminology before comparing results across studies. If relative movement between the lower leg and the calcaneus is presented, then differences in marker placement are not critical. Varying marker placement should have a systematic influence on measured angles and thus will not influence the comparison between relative angles measured using different marker placement methods. However, it is often the case that measurements of eversion/inversion of the calcaneus, are used to represent absolute amounts of pronation/supination. For example angles of 'excessive' pronation are often stated (Pagliano, 1987a). For quoting of absolute angles in this manner, consistency must exist in definition of angles measured and in marker placement.

Cavanagh (1990) demonstrated the significant influence on rearfoot angle measurements of small variations in shoe marker placement. A lateral/medial movement of 3 mm of one of the shoe markers was found to change the rearfoot angle by more than 4 degrees. For shoe markers placed closer together, this variation in rearfoot angle was increased to at least 8 degrees. The repeatability of marker placement is clearly essential for comparison across studies, and accuracy and reliability in obtaining marker coordinates is paramount.

Movement of the calcaneus within the shoe has generally been assumed to be negligible, and thus the measured angles have been used to represent the relative movement of the lower leg and the calcaneus. The degree of movement of the calcaneus within the shoe during running has been investigated using holes cut in the rear of the shoe (Clarke et al., 1980; Nigg et al., 1986a; Stacoff et al., 1992). Nigg et al. (1986a), Clarke et al. (1980), and Stacoff et al. (1992) have found that the movement of the shoe and rearfoot vary systematically, indicating that movement of the rear of the shoe and the posterior aspect of the lower leg can be used to represent the relative movement of the calcaneus and the lower leg.

Factors influencing rearfoot motion

Running shoes have been found to influence rearfoot motion variables, with several investigators detecting differences between running barefoot and with shoes. Lower touchdown angles, increased time to peak rearfoot angle, and decreased total rearfoot movement have been found for barefoot running compared with running in shoes (Bates et al., 1978; Smith et al., 1986; Nigg et al., 1986b).

Differences in relative rearfoot angles have been demonstrated between different footwear conditions, but no cause-effect relationship with injury occurrence has been established (Kilmartin and Wallace, 1994). Variations in running shoe materials and the geometry of the shoe have been found to be influential.

Standardised material tests have been developed to quantify the hardness of materials, allowing quantification of the materials of different running shoes. Shore values have commonly been obtained by measuring the resistance of a material to penetration by a defined object under a defined pressure. Denoth (1986) described the Shore A and Shore D units of hardness measurement, where the difference between these two methods of hardness measurement was in the dimensions of the object used to penetrate the material. At present, SATRA Footwear Technology Centre (Shoe and Allied Trades Research Association, UK) use International Rubber Hardness Degrees (IRHD) to quantify hardness of shoe materials. This test is intended to provide a rapid measurement of rubber stiffness, unlike the previously described hardness tests which measure resistance to permanent deformation. The hardness is measured from the depth of indentation into the rubber sample of a spherical indentor under a specified force. A hardness scale of between zero and 100 is used, with zero representing the hardness of a material with a Young's modulus of zero, and 100 representing a material of infinite Young's modulus. In the British Standards describing this test it is stated that, for highly elastic materials, the scales of IRHD and Shore A durometer are comparable.

Drop tests have been described for the evaluation of the shock absorption characteristics of a shoe material. For example, SATRA employ a mass of 8.5 kg dropped from a height of 5 cm onto the shoe, aiming to provide a controlled simulation of the impact occurring in running. An accelerometer is used to measure the peak deceleration, providing a 'g' value. The lower the 'g' value, the more shock absorption provided. Typical 'g' values of between 9 g and 15 g have been measured for running shoes, compared with up to 42 g for town shoes (The SATRA Bulletin, 1991).

Clarke et al. (1983b) demonstrated that the wearing of soft shoes of less than 35 Shore A durometer resulted in a significant increase in peak rearfoot angle and total rearfoot movement compared with wearing harder shoes. In contrast, Nigg et al. (1986) found that total rearfoot movement was lower in shoes of 25 Shore A than in harder shoes of 35 Shore A durometer. Edington et al. (1990) suggested that this discrepancy may be due to differences in geometry of the shoes used in the two studies. Quantification of the hardness of shoe materials facilitates the comparison of results across studies. However, it is apparent from the contrasting results in the literature that specification of shoe geometry is also required.

Increased heel height has been demonstrated in two separate studies to influence the amount of rearfoot pronation by decreasing both the peak rearfoot angle and the period of rearfoot motion (Bates et al., 1978; Stacoff and Kaelin, 1983). In contrast, Clarke et al. (1983b) found no relationship between increased heel height and rearfoot motion variables. The conflicting results from these studies may be due to differing combinations of shoe design factors. For example, variations in heel height may also influence the shock absorbency of the shoe due to increased thickness of material. There appears to be no data available concerning the influence on rearfoot motion of manipulating heel height using heel lifts.

Lateral heel flare has also been shown to influence the amount of rearfoot movement, with a greater lateral heel flare being associated with an increase in initial rearfoot angle and initial rearfoot velocity (Nigg and Morlock, 1987; Nigg and Bahlsen, 1988). The increased pronation was associated with an increase in the length of the moment arm of the resultant ground reaction force about the subtalar joint, resulting in an increase in moment about this joint, and thus an increase in the velocity of movement about this joint.

The influence of the introduction of orthotic devices on rearfoot motion variables has also been investigated. A significant reduction in peak rearfoot angle has been found by several investigators by the use of orthoses to build up the medial side of the shoe (Nigg, 1986; Smith et al., 1986). Other studies have not been able to demonstrate a significant influence on rearfoot motion (Bates et al., 1979). The orthoses used in these investigations have not been standardised between studies, and thus discrepancies in results are not surprising.

Even though it must be stressed that the entire movement of the ankle and subtalar joints cannot be represented solely by the motion of the rearfoot, this method clearly allows the detection of changes in foot function in response to variations in shoe design. Since rearfoot motion has been found to be influenced by variations in shoe geometry and material properties, the investigation of one particular design characteristic clearly requires the careful control of the remaining shoe characteristics.

(v) Sagittal Plane Kinematics

Quantification of sagittal plane kinematics.

Due to the predominantly two-dimensional nature of running, many investigators have limited their analyses of running to the sagittal plane (Dillman, 1975; Vaughan, 1984). Engsberg (1987) demonstrated that the ankle joint acts predominantly in the sagittal plane. The findings of Soutas-Little et al. (1987) and Arebald et al. (1990) that sagittal plane joint angles obtained using two-dimensional data are comparable with joint angles measured in three-dimensions supports the use of sagittal plane measurements to quantify ankle joint motion. The knee and hip joints also act predominantly in the sagittal plane during running.

Cavanagh (1987) described the typical timing of phases of running gait for a subject running at 3.83 $m.s^{-1}$ using a rearfoot ground strike. Following footstrike, taken to be zero time, movement of the foot to a position flat on the ground was found to take 50 ms. The following phase, from 50 ms to 180 ms was termed the mid-support phase, also referred to as midstance (Hlavac, 1977). During this phase the foot was flat on the ground. The phase from midstance to toe-off was found to take 50 ms, providing a total ground contact time of 230 ms. The descriptions by Cavanagh (1987) were similar to those of Hlavac (1977) in which the first 25% of stance was described as the contact phase, the period from 25% to 75% stance was referred to as midstance, and the final 25% of ground contact was called the propulsive phase.

For the measurement of sagittal plane kinematics, marker placement conventions have been adopted. The use of markers on the skin to signify body landmarks has commonly been used, with ankle, knee and hip markers in particular being used for measurement of joint angles (Figure 2.11). Skin markers are only approximations of the bone movement that they are being used to represent. Both the knee and hip joints are difficult to mark, due in particular to their varying locations during movement. Milliron and Cavanagh (1990) described how markers placed on the foot and ankle are less problematic since the required bony landmarks are generally better defined, and the amount of skin movement is relatively small. Maslen and Ackland (1994) found, using radiographic techniques to study inversion/ eversion, that minimal displacement of skin markers occurred relative to the lateral malleolus during standing, providing markers were placed under a weight-bearing condition.

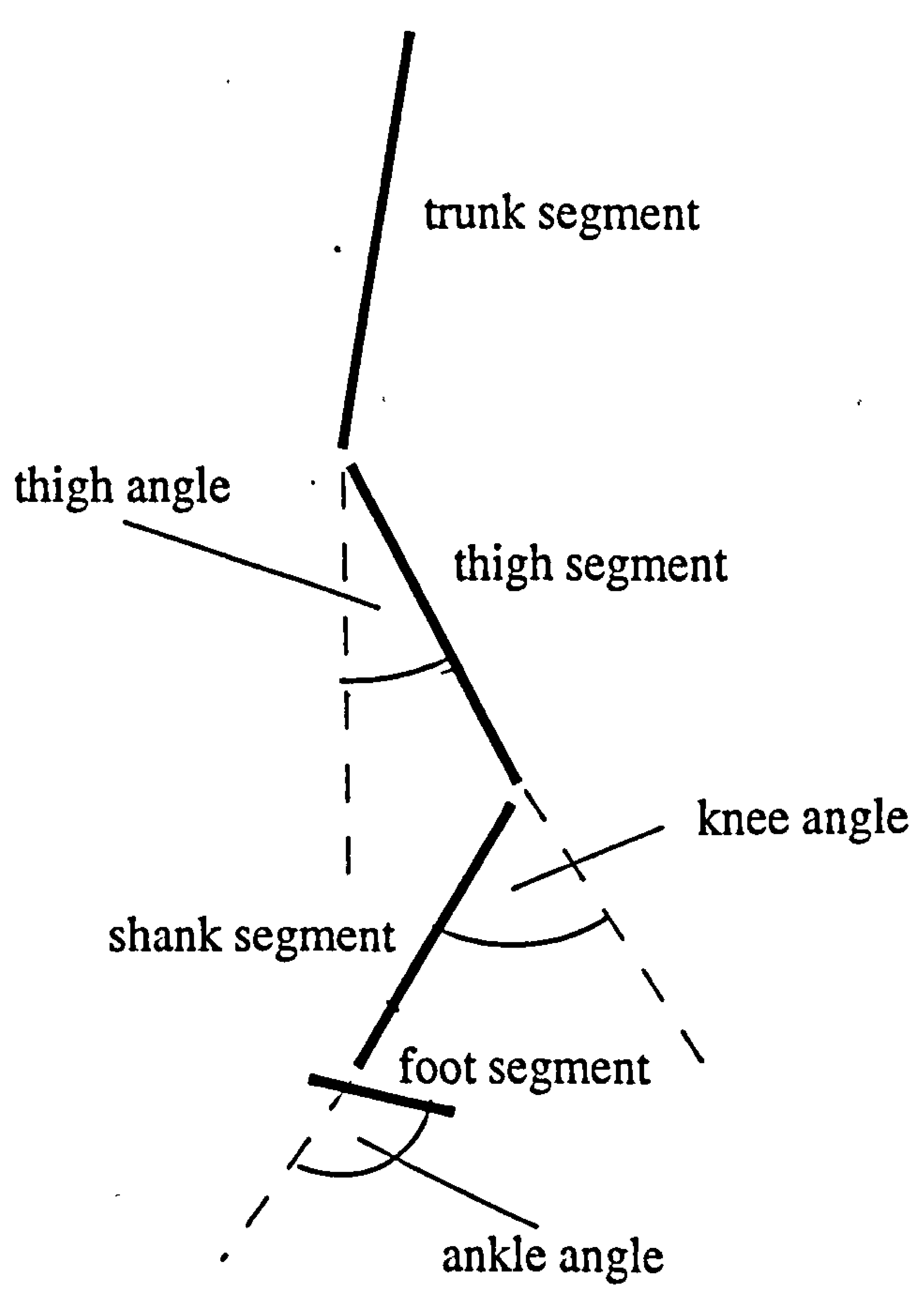


Figure 2.11 Conventions for measuring sagittal plane joint angles of the lower extremity

The hip joint angle is particularly difficult to measure and to standardise, due mainly to difficulties in defining the proximal end of the trunk segment. Many investigators have therefore chosen to present thigh angles relative to the vertical, rather than hip angles (Williams, 1980; Milliron and Cavanagh, 1990). By convention, the hip marker is generally placed on the superior border of the greater trochanter. Milliron and Cavanagh (1990) described the movement of the thigh during a running stride. After initial ground contact in hip flexion, flexion was found to continue until approximately 50% of the stance phase, when simultaneous hip and knee extension occurred at similar rates. Toe-off occurred around the time of maximum extension, coinciding with maximum knee extension. In Table 2.4 thigh and hip angles are presented as reported in the literature.

The general description of knee angle changes during running is consistent across the literature, but absolute joint angles at different stages of the gait cycle have been found to vary (Table 2.5). Milliron and Cavanagh (1990) provided a general description of knee movement during running. These authors described how the instantaneous centre of rotation of the knee joint moves depending on the joint angle, limiting the ability of a single marker to represent this joint centre. The knee was found to be between 10 and 20 degrees of flexion at ground contact. Immediately following ground contact, knee flexion occurred up to approximately 50% of the stance phase, providing cushioning of the impact. The knee then extended up to toe-off, with the knee angle at toe-off being approximately equal to the angle at ground contact.

Milliron and Cavanagh (1990) described how ankle angle can be defined as the angle between the shank and the foot segments, where the shank segment is represented by a straight line joining the knee and ankle joint centres, and the foot segment by a straight line joining a marker on the heel to one on the fifth MTP joint centre. Typical recorded ankle angles are presented in Table 2.6. There are no bony landmarks on the heel that are suitable for standardising the placement of this marker, and thus the convention of placing the heel marker in a position to create an ankle angle of 90 degrees when the subject is in a standing position has generally been adopted. Using this procedure, Milliron and Cavanagh (1990) measured ankle angles of approximately 90 degrees at footstrike. After initial ground contact, a small amount of ankle joint plantar-flexion of about 5 degrees was found to occur, influenced by simultaneous ankle plantar-flexion and knee flexion. The continued knee flexion once the foot is flat on the ground was shown to cause ankle dorsi-flexion up to an ankle angle of 110 degrees, occurring at approximately 50 % of the stance phase, coincident with the time of maximum knee flexion. A maximum plantar-flexion angle of 20 degrees, corresponding to an ankle angle of 70 degrees, was found to occur slightly after toe-off. During the swing phase, a gradual dorsi-flexion of the ankle joint was demonstrated, to an approximate ankle angle of 90 degrees in preparation for ground contact. Milliron and Cavanagh (1990) found that sometimes more dorsi-flexion than required occurred and a small amount of plantar flexion was then found to occur to attain the approximate 90 degree ankle angle for ground contact.

A large variability in running kinematics has been found to exist between subjects

(Bates et al., 1983a). The varied results presented in Tables 2.4, 2.5 and 2.6 support this suggestion. The results presented in these tables refer to studies where differing numbers of subjects were used. Milliron and Cavanagh (1990) presented mean results for four subjects. Williams (1985) presented mean results for 31 subjects. Miller (1978) presented sagittal plane kinematic data for an individual female distance runner of national running standard. Due to the large variability in running kinematics between subjects, it may be appropriate to present data for individual subjects, eliminating the possibility of obscuring the behaviour of individuals.

Factors influencing sagittal plane kinematics

Sagittal plane kinematic data have been found to vary for barefoot versus shod running (Clarke et al., 1983a; Frederick, 1986). Differences have been demonstrated in sagittal plane kinematics in response to varying amounts of running shoe midsole hardness. Frederick (1986) described how maximum knee flexion velocity was increased with an increased hardness, whilst Clarke et al. (1983a) demonstrated increased ankle dorsi-flexion immediately prior to ground contact with shoes of increased midsole hardness. The influence of increased heel lift on sagittal plane kinematics does not appear to be known. It has been suggested that an increased heel lift reduces the maximum ankle dorsi-flexion, contributing to a decreased strain experienced by the Achilles tendon. Investigation is required of the influence of heel lift on sagittal plane kinematics.

Table 2.4 Maximum hip / thigh extension and flexion angles reported in the literature

Authors	Thigh Angles ext.	Thigh Angles flex.	Hip Angles ext.	Hip Angles flex.
Miller (1978)			-26.1	35.2
Milliron and Cavanagh (1990)			-23.0	39.3
Nilsson et al., (1985)	-11.0	23.8		
Sinning and Forsyth (1970)	-8.0	26.0		
Teeple (1968)				31.0
Williams (1985)			-25.8	33.5

Table 2.5 Knee extension and flexion angles reported in the literature

Authors	Stance Angles footstrike	Stance Angles max. flex.	Swing Angles max. flex.
Miller (1978)	19.2	42.0	104.0
Milliron and Cavanagh (1990)	9.9	38.6	109.3
Nilsson et al., (1985)	14.0	37.0	106.0
Sinning and Forsyth (1970)	12.0	27.0	82.0
Williams (1980)	9.4	42.2	101.0

Table 2.6 Ankle plantar- and dorsi-flexion angles reported in the literature

Authors	Stance Angles plantar	Stance Angles dorsi	Swing Angles plantar	Swing Angles dorsi
Milliron and Cavanagh (1990)		108.0	54.9	
Nilsson et al., (1985)	84.3	107.5	57.0	94.0
Sinning and Forsyth (1970)		104.0	64.0	
Williams (1980)	74.0	115.0	68.0	98.0

2.5 Ground Reaction Force

(i) Measurement of Ground Reaction Force

The ground reaction force (GRF) is the force that reacts to the contact of the foot on the ground. It is equal in magnitude and opposite in direction to the push applied by the foot. The force is distributed over the entire contacting surface, but is represented as a resultant vector with magnitude, direction and point of application. Aspects of GRF have been associated with Achilles tendon injury occurrence (eg. Light et al., 1980). These data are also required for synchronisation with kinematic data for the estimation of ankle joint moments using inverse dynamics.

GRF data are obtained by means of force plates. The data provided are the resultant force components in three orthogonal directions (F_x,F_y,F_z), the point of resultant force application (a_x,a_y), and the free moment (M'_z). The convention of the Kistler force plate system is such that the medial-lateral force component is termed F_x, and the anterior-posterior (breaking-propulsion) force is termed F_y. The natural frequency of a force plate must be verified to ensure it is sufficiently higher than the highest frequency to be measured. The data should be appropriately filtered to guard against aliasing.

(ii) Interpretation of GRF Data

The most commonly used GRF variables for the analysis of running include duration of ground contact, maximum vertical impact force, time from initial ground contact to impact peak, maximum impact force loading rate, maximum active (propulsive) vertical force, anterior-posterior force peaks (positive and negative), and medial-lateral force peaks (positive and negative).

For running, the vertical component of ground reaction force has by far the greatest magnitude of the GRF components. Typical vertical GRF time-histories in running have been presented extensively in the literature (Figure 2.12). Nigg (1983) described three different time histories of vertical GRF, with that illustrated in Figure 2.12 being the most commonly observed. This trace has a single impact peak and an active peak. Double, and sometimes triple, peaks are seen at impact, whilst the impact peak is sometimes absent. Total time of ground contact, or stance time, is generally calculated using this vertical GRF component.

The impact peak is the maximum vertical force occurring during the first 50ms of ground contact. This peak is often referred to as the passive impact peak, since the short time over which it occurs is not sufficient for any muscular response (Nigg, 1986). Magnitudes of this parameter of between 2 bodyweights (BW) and 3 BW for running speeds of between 3 $m.s^{-1}$ and 6 $m.s^{-1}$ have been recorded (Cavanagh and Lafortune, 1980). For some subjects, impact force peaks are not visible on force traces (Cavanagh and Lafortune; 1980). Maximal vertical force loading rate, which exists in every force-time curve, is often used to describe the load during impact. Different force ranges have been adopted in the literature over which vertical GRF loading rate has been calculated. Miller (1990) described the use of the time taken for a 1 BW rise in force from a 50 N ground contact criterion. A limitation of using a 50 N vertical force to signify ground contact is the possibility that the loading rate immediately following impact is overlooked. A criterion selected following the careful observation of the data to identify the start of ground contact may be more appropriate. Lees and Haynes (1995) presented vertical GRF loading rate data calculated instantaneously using 1000 Hz data. Other authors have calculated the average loading rate using the impact peak and time to attain this peak (Bourassa and Therrien, 1981). The approximately linear rise in vertical force from initial ground contact to peak impact force, results in this procedure providing an approximation of the peak loading rate during impact. Calculation of the average loading rate allows comparison with the literature where impact force and time of occurrence have been provided.

The maximum vertical active force peak is the second peak seen on the vertical force-time history (Figure 2.12), and represents the force applied during the propulsive phase of contact. This peak has been shown to occur between 35% and 50% of total stance time, and to have a magnitude of between 2.5 BW and 2.9 BW when running at speeds of between 3 $m.s^{-1}$ and 6 $m.s^{-1}$ (Cavanagh and Lafortune, 1980; Munro et al, 1987).

The point of application of the resultant force is generally termed the centre of pressure. The location of the centre of pressure relative to the ankle joint centre is of interest in the present study since this variable, together with the GRF resultant force magnitude and direction, determine the ankle joint moment.

Variability of GRF data within subjects has been demonstrated. Bates et al. (1983b) assessed subject variability and suggested that a minimum of eight trials are required in order to obtain stable subject-condition values for GRF's, and 10 trials for a 95% confidence level. The condition for stable data required that all successive mean data values were within one-quarter standard deviation of the 10-trial mean for that variable. Cavanagh (1987) suggested that the mean result from at least six trials is required to attain reliable GRF data. Kinoshita et al. (1985) collected GRF data for three subjects for 15 successive trials to investigate the number of trials required to obtain stable mean data, and obtained results in agreement with those of Bates et al. (1983a). It is apparent that at least 10 trials are required to obtain stable GRF data. The exact number will depend on factors such as the subject, the conditions and the GRF variables of interest. Thus, the requirements of a particular study must be considered when determining the appropriate number of trials.

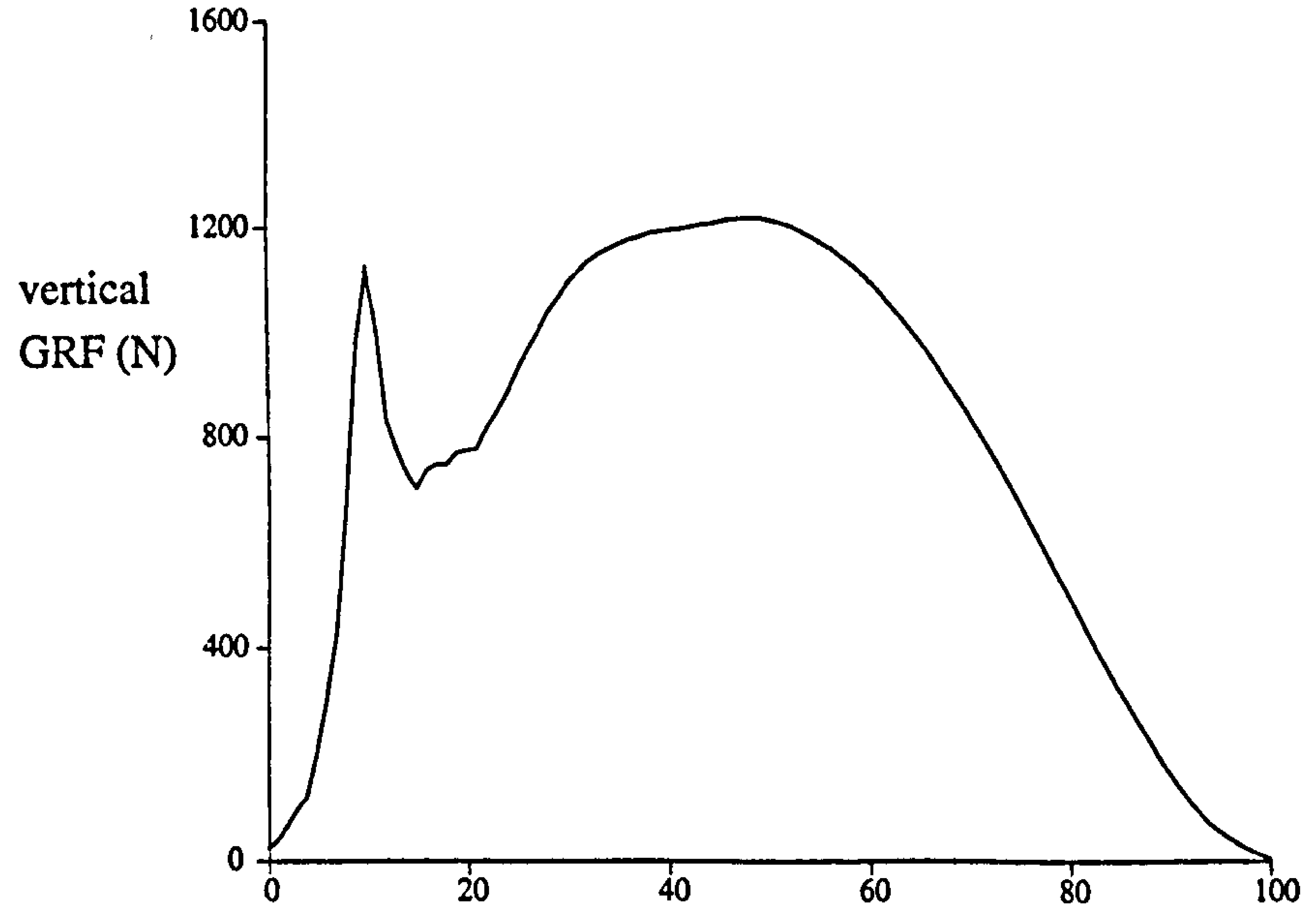


Figure 2.11 A typical vertical GRF time history for a rearfoot striker running at 3.83 m.s-1

(iii) Factors Influencing GRF

Running speed and style

Nigg et al. (1987) demonstrated that GRF variables are influenced by running velocity. These authors found that with increased running velocity the maximum impact force magnitude increased linearly, whilst the magnitude of the GRF loading rate increased non-linearly. The time of occurrence of the peak impact force and the maximal loading rate were found to decrease linearly with increased speed. As with kinematic data collection, it is clearly important to specify the running speed at which GRF data are collected.

Differences in running style have been shown to influence GRF variables. Cavanagh and Lafortune (1980) developed a footstrike index to quantify the location of the centre of pressure at initial ground contact. Runners were described as a rearfoot strikers if the point of initial contact was in the rear third of the foot, as midfoot strikers if it was in the middle third of the foot, and as a forefoot strikers if it was in the front third of the foot. Wide variations were demonstrated across subjects. Characteristic GRF traces were described for the different methods of footstrike. In particular, rearfoot strikers were shown to demonstrate a distinct impact peak, whilst midfoot strikers demonstrated a reduced, smooth impact peak much later and less distinct than that observed for the rearfoot strikers. No forefoot strikers were detected in the Cavanagh and Lafortune (1980) study. Forefoot strikers have been shown to have smoothed, sometimes undetectable, impact peaks occurring relatively late in stance (Lees and McCullagh, 1984). The time history of the centre of pressure in relation to foot position on the plate (Cavanagh and Lafortune, 1980), and time history of the anterior-posterior GRF component (Cavanagh, 1987) have been shown to differ for subjects with different running styles. The characteristic GRF force data for particular running styles facilitates the use of GRF data alone to describe the running style adopted during a specified running trial.

Running shoes and shoe inserts

Studies of the influence of shoes of different hardness on GRF variables have provided contrasting results. Peak forces measured using impact tests have provided increased values with increased hardness (Clarke et al., 1983a; Nigg et al., 1986a). The results of running studies have not been consistent with these findings. Clarke et al. (1983a) and Nigg et al. (1987) found no difference in peak impact force values for shoes of varying hardness. Kaelin et al. (1985) and Nigg and Bahlsen (1988) found that shoes with softer midsoles were measured with higher impact peaks than those with hard midsoles. Several explanations have been provided for the apparent inconsistencies between shoe midsole hardness measured using material tests and GRF results. An increased rate of rearfoot movement for a harder shoe, due to an increase in moment arm length of the GRF about the subtalar joint, has been suggested (Nigg et al., 1987; and Kaelin et al., 1985). Kinematic adaptation in the form of increased knee flexion, suggested to compensate for inadequate shock absorption, has been demonstrated (Clarke et al., 1983a). 'Bottoming out' may occur when softer midsoles reach their limit of compression during impact (Bates et al., 1983b). The fact that the vertical GRF represents the acceleration of the total body, and not the accelerations of components of the lower extremity, has been highlighted (Miller, 1990). The time taken to reach the impact peak has been shown to be increased with reduced hardness, indicating that, for unchanged magnitudes of impact force, the average loading rate has been reduced (Clarke et al., 1983a, Therrien et al., 1982; Snel et al., 1985). In contrast with this suggestion, Nigg and Bahslen (1988) found that increased hardness caused a reduction in vertical impact force peak loading rate.

Shoe geometry has also been found to influence GRF results. In particular, Nigg and Bahlsen (1988) demonstrated that the vertical impact force peak decreased significantly with reduced heel flare. This was attributed to a decrease in moment arm length about the subtalar joint associated with the decreased flare, resulting in a reduction in pronation velocity and subsequent decrease in GRF. For softer shoes, the lateral side of the heel was compressed easily at initial contact, shortening the length of the moment arm when compared to a shoe of the same geometric specifications but harder material.

The contrasting results obtained across studies are likely to have been due to different combinations of midsole hardness and shoe geometry. There appears to be no direct relationship between shoe design characteristics and GRF variables, indicating that care should be observed when interpreting force plate results. Manipulation of heel lift may influence impact forces, since the moment arm of the GRF about the ankle joint centre is likely to be influenced by the amount of heel lift. Investigation of this suggestion is required.

The influence of shock absorbing shoe inserts on GRF variables has been investigated. Lees and McCullagh (1984) demonstrated general trends in reduced impact force peak and loading rate with the use of viscoelastic insoles in running shoes, with the changes in loading rate being most marked. In contrast, Nigg et al. (1988) demonstrated that there were no significant differences in GRF variables when viscoelastic insoles were placed in running shoes. Differences in the shoes and the insoles employed may account for the conflicting findings of these two studies, although these differences could not be quantified. The behaviour of individual subjects may have been overlooked by Nigg et al. (1988), since these authors grouped the results for all subjects, whereas Lees and McCullagh (1984) reported individual subject results.

Consideration of the results of studies of running shoes and shoe inserts has indicated that the loading rate of impact force appears to be more sensitive to changes in shock absorption than the impact force magnitude, indicating that this variable is more suitable for detection of shock absorption differences.

2.6 Experimental Methods and Analysis of Results

(i) Experiments

In order to confidently detect differences in dependent variables across heel lift conditions, a controlled experiment is required. Cox (1968, 1990) outlined the requirements for such an experiment as being the avoidance of systematic error, the maximisation of precision, the applicability of results, simplicity, and calculation of the level of uncertainty in results. With regard to systematic error, it is important to ensure that experimental trials under one condition differ in no systematic way from those under another condition. This can be achieved by randomising the order of the conditions (Cox, 1968). Precision is maximised by minimising the amount of random variation. By minimising systematic error and maximising precision, the internal validity of an experiment is maximised. The level of applicability of results of an experiment to general applications is known as the external validity. Yeadon and Challis (1994) described how internal and external validity are in conflict, since the more experimental control imposed, the greater the likelihood of reducing external validity. The determination of the relative importance of these two forms of validity is therefore required when planning an experiment. Cox (1968) described how an experiment should be as simple as possible, especially if the investigation is in a new area of work, and how an experiment should address a clearly defined question. Consideration of each of these factors is required when investigating the influence of heel lift on Achilles tendon loading.

The conduction of research in stages, as suggested by Reboussin and Morgan (1996), has been described in Chapter 1 of the present research. An interesting observation made by Reboussin and Morgan (1996) was the importance of continued re-assessment of methods at each stage of the experimental investigation. The possible interaction between scientific question, experimental design, and statistical analysis at each stage, was highlighted. A flexible approach was suggested, allowing for the possibility of design changes influencing the question, and methods of analysis helping to refine the design of the experiment.

(ii) Methods for Detection of Differences

When comparing the response of subjects to two or more conditions, traditional inferential statistical methods have generally been used (Cohen and Holliday, 1982). A null hypothesis is tested at a defined level of significance, stating the probability that random variation causes an apparent difference.

The probability of detecting a relationship that does not exist (Type I Error) is under the direct control of the investigator, since it is determined by the level of significance. The probability of missing a relationship that does exist (Type II Error) depends on both the level of significance and whether the hypothesis is actually true or false. This error cannot therefore be controlled directly. It is, however, influenced by the design of the experiment. The relationship between error types and level of significance is such that as the probability of a Type I Error is decreased, the probability of a Type II Error is increased.

The power of a statistical test is the ability of the test to detect a relationship that exists.

It is therefore the opposite of a Type II Error (Power = 1 - (Probability of Type II Error)). The magnitude of the power of a test is influenced by sample size, variability, control of extraneous variables, precision of equipment, operators and methods, the type of statistical analysis, and the level of significance used. In order to maximise the power, the amount of variability and the influence of extraneous variables should be minimised, and precision should be maximised. The most common method used to obtain adequate power values is by increasing the sample size (Bates et al., 1992).

Cohen (1969) described the comparison of mean values across conditions, termed the effect size. An effect size may be tested for clinical or practical significance. Normalised effect size can be obtained by dividing the effect size by the common population standard deviation, allowing comparison of results between studies of different variables. Cohen (1969) identified categories of effect size as small (<0.2), medium (>0.2 and <0.8), and large (>0.8).

With knowledge of the effect size and significance level, standard tables can be used to obtain the sample size necessary to obtain a particular statistical power. Bates et al. (1992) investigated the influence of sample size on power in group and single subject studies. The number of subjects and the number of trials per subject were varied to illustrate the effect on statistical power. These authors found, to attain adequate power, single subject studies required a greater number of trials per subject than group studies. Although these authors provided figures for the number of trials required to obtain defined effect size, each experimental design should be considered individually for determination of an appropriate number of trials.

An increase in the size of the sample is sometimes not possible or practical. A less stringent level of significance can then be used to increase power. Each experiment must be considered individually to determine the relative cost of Type I and Type II errors. Franks and Huck (1986) described how Type II errors are costly in exploratory studies because they may result in no further study being carried out in a particular area. In their final recommendations, Franks and Huck (1986) suggested that researchers should decide the significance level based on the available sample size, variability, desired effect size, and relative importance of Type I and Type II errors.

(iii) Single Subject Studies

The averaging of results over a group has been found to obscure individual behaviour (Herson and Barlow, 1976; Bates et al., 1983). General results are obtained which are unlikely to reflect the response of individuals in the group. The variation in response between subjects has led to the alternative approach of single subject study (Bates, 1996; Reboussin and Morgan, 1996).

Bates (1996) described how a single subject approach is required if variations in movement between subjects are the result of individual subjects using different strategies to perform the same task. This author defined a strategy as a selected neuromusculoskeletal solution for the performance of a motor task, and described how patterns of movement are constrained to react to mechanical, morphological and environmental variations. Bates (1996) described how response patterns vary along a continuum from purely mechanical, where the outcome of the perturbation can be predicted from the principles of Newtonian mechanics, to totally neuromuscular, where the system adjusts to maintain the value of the dependent variable by modifying other variables.

Hlavac (1977) and Subotnick (1979) have described the differing foot structure and function between subjects and the resulting varying strategies used for running. Variations in running kinetics and kinematics have been demonstrated across subjects. Cavanagh and Lafortune (1980) defined three distinctly different methods of foot strike in running, as previously described. Bates (1989) highlighted the fact that each individual has many possible response patterns in the control of the human body on impact with the ground in running. Winter (1980) demonstrated the different combinations of lower extremity joint moments adopted by different individuals for production of similar total support moment values. Miller (1990) described variations across all subjects in the anterior-posterior and the medial-lateral force-time histories.

The response to changes in external conditions, such as shoes or shoe inserts, has also been found to vary between subjects. For example, Bates et al. (1983) demonstrated varied ground reaction force responses to changes in running shoes across subjects, and Lees and McCullagh (1984) demonstrated differing responses to the introduction of insoles in a running shoe.

This brief summary of the behaviour variations between subjects highlights the need for study of individual subjects when investigating the influence of footwear or shoe insert interventions on running mechanics. Group studies, in which efforts are made to generalise results to the remainder of the population, only appear appropriate if the mean results resemble the response of individuals in the group. Bates et al. (1992) described how practitioners are usually concerned with the responses of an individual to interventions, rather than with the behaviour of a theoretical average person. For the design of personalised running shoes for elite athletes, and for the treatment or prevention of running injuries by the prescription of shoe inserts, an understanding of individual response patterns is clearly important. For the design of running shoes for the general population, however, it is common to obtain group responses. The danger in this approach is that running shoes are designed for the 'average' runner, who in reality will rarely exist. Bates (1996) suggested that a combination of group and single subject designs may be appropriate. It is concluded that, for the testing of the response of individuals to shoe design characteristics, the use of single subject studies appears most suitable, with subsequent generalising of results only where appropriate. This approach has been supported by Reboussin and Morgan (1996) who described the use of 'multiple single subject' designs, in which a series of single subject studies are performed. An indication of the consistency of any intervention effect can therefore be obtained.

Advantages to using single subject studies include the possibility of intensive investigation of individual subjects (Bryan, 1987; Smith, 1988), and the provision of strong contrasting evidence (Smith, 1988). The main drawback of the single subject approach is the limited knowledge gained on the behaviour of the remainder of the population. It should be clearly stated that the results refer only to the subject studied.

(iv) Analysis of Results from Single Subject Studies

Contrasting opinions exist on the most suitable methods for analysis of results from single subject studies.

Graphical analysis

Bryan (1987) described how researchers in the field of human behaviour have often used visual analysis to detect the effect of interventions, since these investigators are often more concerned with the practical or clinical significance of results, rather than the statistical significance. Graphical methods of analysis may be suitable for initial visual detection of differences in footwear conditions in running, and for illustration of differences. The practical or clinical significance of an effect size may be assessed if the appropriate knowledge is available. In particular, with regard to Achilles tendon loading, it could be argued that a decrease in Achilles tendon loading is significant only if the change results in a reduction in the likely incidence of Achilles tendon injury. The present knowledge on tendon loading likely to cause injury is not sufficient to allow the use of such a criterion. In the future, such an approach may be appropriate.

Time series research

Single subject studies have generally involved the use of some form of time series experiment (Kratochwill, 1978). In such studies, a dependent variable is monitored in the absence of an independent variable. This is known as the baseline condition (condition *A)*. An experimental change is then introduced by varying the independent variable, known as the treatment or experimental condition (condition *B)*. The result is an *AB* sequence of trials. The results of the intervention in time series experiments are demonstrated by any discontinuity in the recorded dependent variable. Differing designs have been implemented, with variations on the *AB* sequence described.

A variation on the commonly used methods of time series analysis is the randomised block design. The order of conditions applied should be selected randomly, to eliminate the possibility of maturation effects, such as familiarisation, boredom and fatigue, influencing results. Each block within the randomised block design should contain one trial for each of the conditions, and the order of conditions within each block should be randomly assigned. This approach allows for as many trials as required to be performed under each condition, whilst protecting against maturation effects. An example containing four trials under each of three conditions might be: *BAC, ABC, CAB, CBA* (*A* = baseline, *B*=condition 1, *C*=condition 2).

Statistical analysis

It has been suggested that traditional statistical methods are not appropriate for analysis of data obtained from single subject studies (Levin et al., 1978; Bryan, 1987). Cohen and Holliday (1982) defined statistics as the drawing of inferences about large groups on the basis of observations made on smaller ones. In statistical studies, the observed sample is assumed to be drawn randomly from the studied population. Since this is generally not the case in single subject studies, it may be suggested that traditional methods of analysis are not appropriate. However, in group studies, the sample is rarely obtained randomly from the population under consideration. It is therefore not possible to generalise the results to the entire population on statistical grounds. It is, however, acceptable to use traditional methods to investigate whether a relationship exists for the subjects studied, and then to use logical grounds to generalise the results for the remainder of the population. Edgington (1967) argued that a similar approach, using a combination of statistical methods and logical arguments, can be applied in single subject studies.

Traditional statistical procedures, which require assumptions about the population, are known as parametric methods. The most common methods employed in group studies are analysis of variance (ANOVA) procedures. For using ANOVA methods, it is assumed that the sample is taken from a normal population, that the subjects have been randomly sampled from this population, that treatment conditions are applied randomly to the subjects, and that independent errors occur. Many of these assumptions are not met in single subject studies (Levin et al., 1978). ANOVA methods have been found to be robust to violations of all assumptions, except for the independence of errors.

Caster et al. (1994) described how criticism of the use of ANOVA methods for single subject studies has centred on the possible violation of the assumptions of a normal distribution and independence of errors. These authors carried out within-subject tests for normality and independence, for selected biomechanical measures of 35 subjects. The biomechanical measures included ground reaction force variables, joint moments and joint angles. Across all biomechanical measures obtained, it was found that 33% of the data sets were significantly non-normal. Since most statistical tests are robust to deviations from normality, this finding was not considered to limit the use of traditional procedures in single subject studies. Caster et al. (1994) found that the biomechanical measures obtained in their study were statistically independent, and individual subjects were described as random trial generators. It was suggested that this is not surprising given the complex combination of neural, muscular and skeletal systems involved in activities such as running. Caster et al. (1994) supported the use of traditional statistical methods in single subject studies of running.

Non-parametric methods of statistical analysis allow detection of differences between conditions, without making assumptions about the distributions of the sampled populations. Non-parametric tests tend to involve the use of simple formulae which are quick and easy to use, often producing a rank ordering, rather than numerical results.

Since non-parametric tests do not require the use of any assumptions, it could be argued that they should be preferred, and routinely used. There are, however, limitations involved with the use of these tests. A disadvantage of using non-parametric tests is that they generally have relatively low statistical power, compared to parametric tests. Additionally, as described by Campbell and Machin (1990), non-parametric tests are not flexible. For example, they do not allow for analyses such as multiple regression and analysis of covariance.

2.7 Summary and Conclusions

The investigation of the loading of the Achilles tendon in running has been justified by reference to the frequency of occurrence of this injury. Several suggested mechanisms by which this injury occurs have been described, but all evidence to support these suggestions has been found to be circumstantial. Possibilities highlighted as injury causes have included the maximum stress and strain experienced by the tendon, the repeated nature of this loading in running, and localised high stresses resulting from rearfoot movement. The requirement for the development of methods for the analysis of these aspects of Achilles tendon loading in running has been highlighted.

A review of Achilles tendon injury treatments has demonstrated the lack of knowledge of the mechanisms by which treatments have been successful. In particular, the use of heel lifts has been highlighted as an intervention that is commonly used successfully, but has limited scientific support. Inverse dynamics procedures have been highlighted as suitable for the comparison of forces across conditions. The requirement of a method for reliable measurement of subject-specific Achilles tendon moment arm lengths has been stressed.

Peak ankle dorsi-flexion angle and peak rearfoot angle have been identified as variables associated with Achilles tendon injury. It has been demonstrated that these aspects of lower extremity kinematics may be influenced by changes in running shoe geometry and material properties. The investigation of the influence of heel lift manipulation on Achilles tendon loading should therefore ideally include measurement of lower extremity kinematics. The inconsistent responses in kinematics and GRF variables across subjects and studies with variations in running shoe properties has been demonstrated. A controlled experiment for examining the influence of heel lift on Achilles tendon loading has been shown to be necessary.

Practitioners are concerned with the treatment of individuals, not with average behaviour. Thus, for the investigation of the influence of heel lift on Achilles tendon loading, a single subject analysis appears to be most appropriate. The suitability of traditional ANOVA procedures in the analysis of results from single subject studies of running mechanics has been demonstrated.

CHAPTER 3 A METHOD FOR ESTIMATING ACHILLES TENDON FORCES

3.1 Introduction

The common occurrence of Achilles tendon injury in runners has been highlighted in the literature review (Chapter 2). The published incidence of Achilles tendon injury in the running population has ranged from 5% to 18% (Clement et al., 1984; Krissoff and Ferris, 1979). In the study described in the present chapter, the frequency of Achilles tendon injury occurrence in elite female distance runners is investigated using a questionnaire distributed to members of the British National Squad.

Overuse has been highlighted as a possible cause of Achilles tendon injury (Archambault et al., 1995). To investigate this suggestion, knowledge of the loading of the tendon during running is required. Indirect methods have been presented in the literature for estimation of internal forces (Burdett, 1982, Scott and Winter, 1990). For Achilles tendon force estimation, it is necessary to calculate the length of the moment arm of the tendon about the ankle joint centre throughout ground contact. The development of a method for measurement of Achilles tendon moment arm during running, for use in determination of maximum Achilles tendon force, is presented in the present chapter. The location of the ankle joint centre and the line of action of the Achilles tendon force are obtained. The influence on estimated Achilles tendon moment arm lengths of using different methods of approximation of tendon line of action is studied, and the most suitable method identified for use in future work.

A factor suggested to influence Achilles tendon injury is heel lift, which can be influenced by shoe heel height (Subotnick, 1979) or the use of heel lifts in the shoe (Smart et al., 1980). A reduction in the incidence of Achilles tendon injury by the use of heel lifts made from Sorbothane, a viscoelastic polymer, has been reported in the literature for a variety of sports participants (MacLellan and Vyvyan, 1981; Fauno et al., 1993). The placement of heel lifts or insoles in a shoe is likely to influence the geometry and the cushioning properties of the shoe. Additionally, the presence of any material in the shoe will affect the positioning of the foot relative to the shoe upper, and thus is likely to influence the amount of stability provided. In the current study, heel lift and cushioning are varied by a combination of shoe and heel lift conditions. The influence of placing commercially available Sorbothane heel lifts in the running shoe is investigated, as this is a commonly recommended action for treatment of Achilles tendon injury. Additionally, conditions are included under which heel lifts are attached to the plantar surface of the foot, thus providing heel lift and increased cushioning, whilst avoiding some of the difficulties involved with controlling the intervention when running shoes are worn.

Reinschmidt and Nigg (1995) used ankle plantar-flexion moment to investigate the influence of heel height on Achilles tendon loading. These authors made the assumption that ankle moment changes directly represent changes of force in the tendon. This assumption may be misleading, since changes in the moment arm of the Achilles tendon will also influence the magnitude of Achilles tendon force. The estimation of both ankle plantar-flexion moment and Achilles tendon forces in the present chapter, allows investigation of the relationship between these two variables.

Statistical power has been defined as the ability of a statistical test to detect a relationship, when a true relationship exists. The data obtained in the present study are used in a post-hoc power analysis to investigate the influence of variations in significance level, sample size and effect size on power values. This procedure allows the identification of an appropriate experimental design for future similar studies.

The appropriate level of significance for detecting differences in the present study is only identifiable after the data have been analysed. The indiscriminate selection of a 0.05 significance level without consideration of power has been questioned (Franks and Huck, 1986). Thus, for the identification of significant differences between conditions in the present study, a range of significance levels are used.

The method developed for estimation of maximum Achilles tendon force is used to investigate whether this force can be influenced by selected shoe design variations, with heel lift in particular being manipulated. As suggested by Reboussin and Morgan (1996), this approach is used to fuel speculation concerning the factors influencing Achilles tendon loading. Where differences are detected, the mechanism by which these occurs is investigated.

In summary, the aim of the present study is to address the question:

- Can maximum Achilles tendon force be influenced by shoe and heel lift interventions ?

Additional questions are:

- Can wearing running shoes decrease the maximum Achilles tendon force ?
- Can the additional use of heel lifts result in a further decrease in maximum force ?
- Does attachment of lifts to the plantar surface of the foot influence maximum force ?
- Do changes in peak plantar-flexion moment indicate changes in Achilles force ?

It is noted that this is an initial study into the influence of heel lift on maximum Achilles tendon force. Therefore, where considered appropriate, variables additional to Achilles tendon force are presented to provide indications of possible influences of heel lift for stimulation of future work. For example, visual observations are made of variables such as GRF impact peak.

3.2 Methods

(i) Questionnaire

A questionnaire surveying the frequency of Achilles tendon injury occurrence was posted to all female distance runners included on the British Athletic Federation (B.A.F.) list of distance athletes. All of these athletes had competed for Great Britain in distance events, ranging from 800 m to marathon. A sample questionnaire is provided in Appendix B.

To facilitate the description of the sample, factors which may influence Achilles tendon injury, such as age (Yamada, 1970) and training load (Archambault et al., 1995) were included. The average total mileage covered by each athlete was obtained as a quantification of the training load. Since the majority of runners vary their average mileage according to the time of year, mileage was obtained for summer and winter separately. Age was also obtained for each athlete questioned.

With regard to the frequency of Achilles tendon injury occurrence, the athletes were asked whether they had ever experienced an Achilles tendon injury which had caused them to miss more than one week of their usual training. If they answered yes, an additional question of whether this injury occurred within the last three years was asked.

(ii) Estimation of Achilles Tendon Forces

Data collection

A single subject performed barefoot running trials with a rearfoot ground strike. The subject was an elite female distance runner of 26 years, with height 1.64 m and mass 54 kg. For each running trial, force data were collected at 1000 Hz using a force plate (Kistler 9281B12), for a right foot ground contact. Video data in the sagittal plane were collected simultaneously at a sampling rate of 50 Hz (Panasonic F15 camera and AG7350 sVHS recorder). Time code was recorded on the video tape during data collection. A video field of view of 1.8 m was used, containing the entire lower extremity during contact with the force plate. Apparatus were set up as illustrated in Figure 3.1. The global axes used throughout this research were defined such that the x-axis was orientated normal to the direction of progression, the y-axis in the direction of progression, and the z-axis vertically, with the origin being located at the centre of the force plate.

Data were collected for ten successful running trials at 3.83 $m.s^{-1}$ (seven minute $mile^{-1}$ pace). A trial was considered to be successful if the speed of running was within 5% of that required and right-footed contact was made with the force plate without any obvious variation in stride during the approach of approximately 10 metres. The average speed of running over a three metre distance was recorded with photocells positioned three metres apart, 1.5 metres either side of the force plate centre line. Breaking of the photocell beam as the subject passed through the first sensor triggered a light-emitting diode (L.E.D.) in the field of view of the sagittal plane camera, started a digital clock recording time taken to travel between the photocells, and triggered the start of force data collection. This facilitated the synchronisation of video and force plate data.

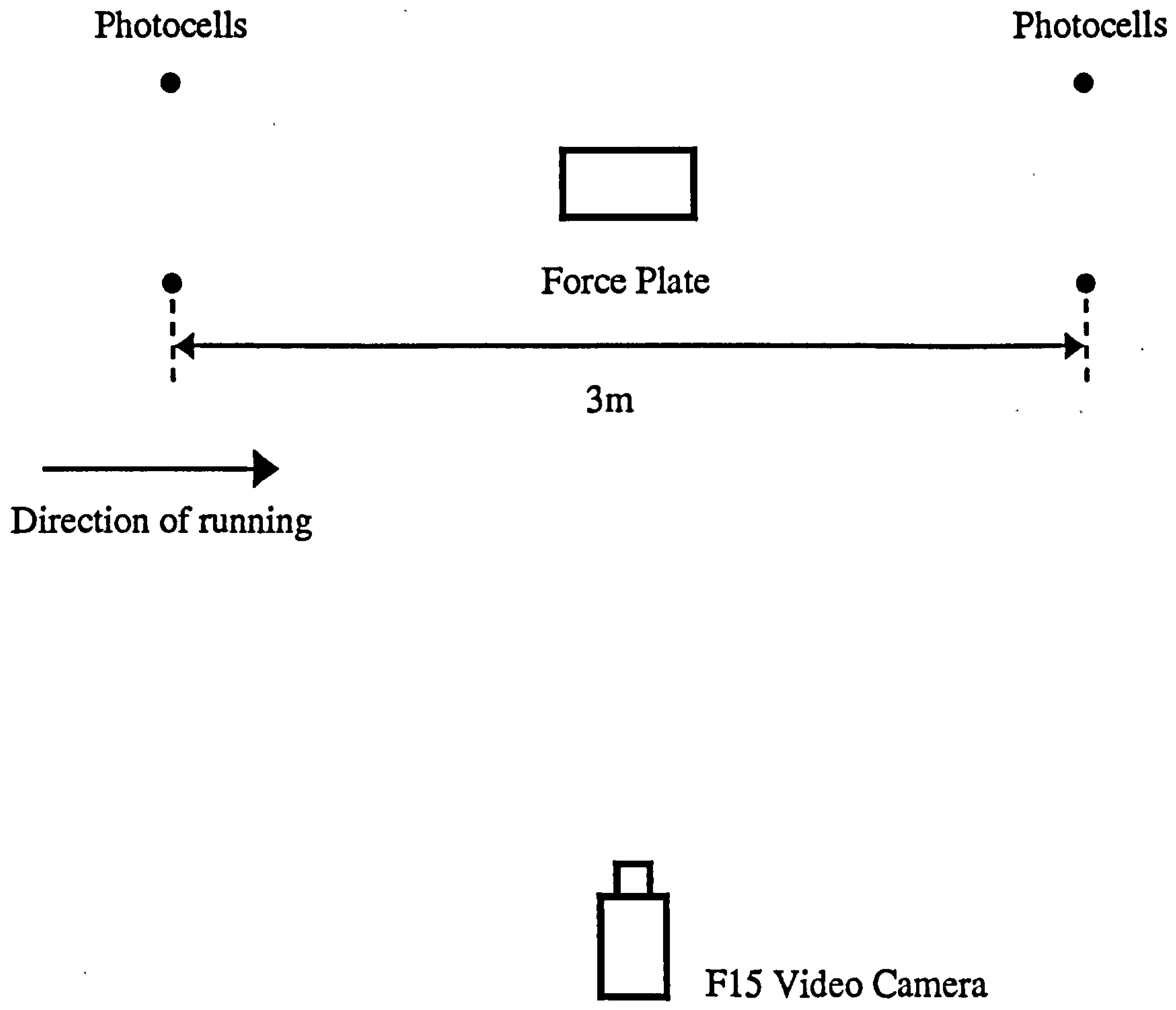


Figure 3.1 Arrangement of data collection apparatus

Digitising

A fine felt pen was used to draw markers of approximately 1cm diameter on the lateral side of the subject for identification of body landmarks during digitisation. Points representing the centres of rotation in the sagittal plane of the knee and metatarsalphalangeal (MTP) joints were located by visual assessment of joint movement in this plane prior to data collection. The ankle marker was placed on the most prominent point of the lateral malleolus. A marker was drawn to represent the centre of the distal phalange of the fifth metatarsal, and three markers were drawn on the skin covering the visually located centre line of the Achilles tendon. A heel point was defined as the point on the rear of the heel such that, for a flat foot condition, this point corresponded to the rear most point of the foot. This location was considered to be the most reproducible point on the heel. To aid in identification of this marker, a horizontal line was marked through its centre around the rear of the calcaneus.

The body landmarks were digitised for each field of the ground contact phase (Figure 3.2), with two additional stationary markers in camera view being digitised as control points (wobble points) to check for any camera movement during data collection, and to allow correction for image movement between fields when digitising. The digitising system used was the Millipede Prisma with a measurement resolution of 768 x 576.

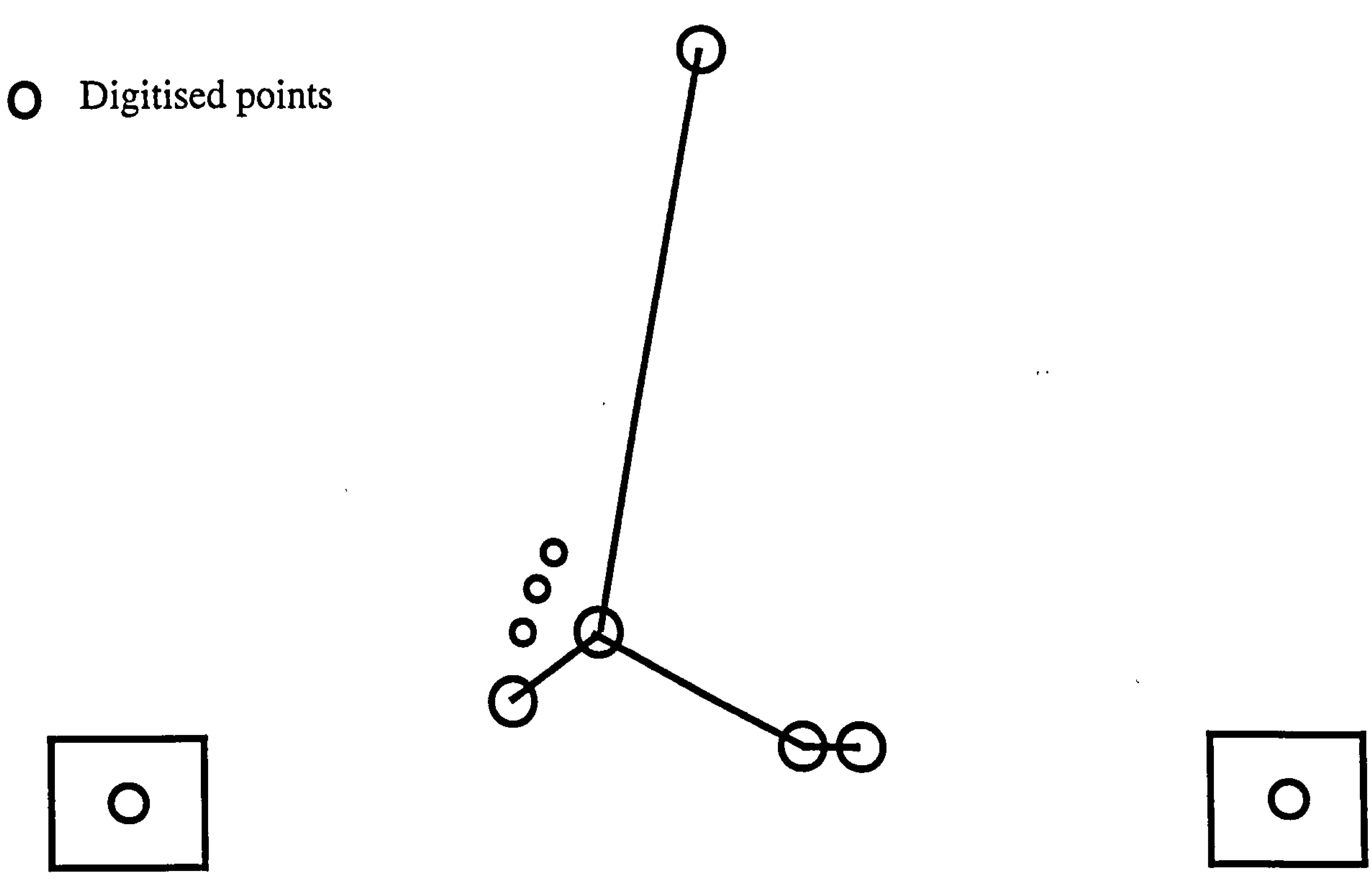


Figure 3.2 Body landmarks and wobble points for digitisation

Scaling methods

A calibration plane was constructed by placement of eight markers on a flat surface to enclose a square area of 0.6 m x 0.6 m. The exact positions of the markers were measured relative to a point on the edge of the plate which was to be placed vertically above the centre of the force plate and vertically below the bottom centre marker, when the plate was positioned in the specified vertical plane (Figure 3.3). These positions were measured to the nearest 0.5 mm using a metre stick, with each marker location being obtained on three separate occasions, and mean values used. The surface of the calibration plane was then positioned vertically and orientated square to the force plate such that the bottom centre marker was directly above the force plate centre, and the calibration plane was in line with the centre line of the force plate in the sagittal plane of motion (yz plane). The measured vertical and horizontal distances in the resulting scaling plane therefore represented locations relative to the force plate centre in the two-dimensional plane used in the study. Before the first running trial, the calibration plane image was recorded in this position.

Horizontal and vertical scaling factors were calculated independently. The five distances y_i (i=1,5) were averaged to obtain the horizontal scaling factor. Similarly, the five distances z_i (i=1,5) were averaged to obtain the vertical scaling factor (Figure 3.3).

Horizontal Distances		**Vertical Distances**	
y_1	1 to 2	z_1	1 to 8
y_2	1 to 3	z_2	1 to 7
y_3	1 to 4	z_3	1 to 6
y_4	1 to 5	z_4	1 to 5
y_5	1 to 6	z_5	1 to 4

The effectiveness of this method of linear scaling in two-dimensions was assessed by comparison of scaled results for two running trials with the results obtained using a two-dimensional Direct Linear Transformation method (2D DLT; Challis, 1993). For all digitised points for the ground contact periods of two separate running trials, the root mean square differences (RMSD) between the data points obtained using the two different methods of scaling were calculated in the horizontal and vertical directions, to provide a measure of the difference in reconstructed data between the two methods.

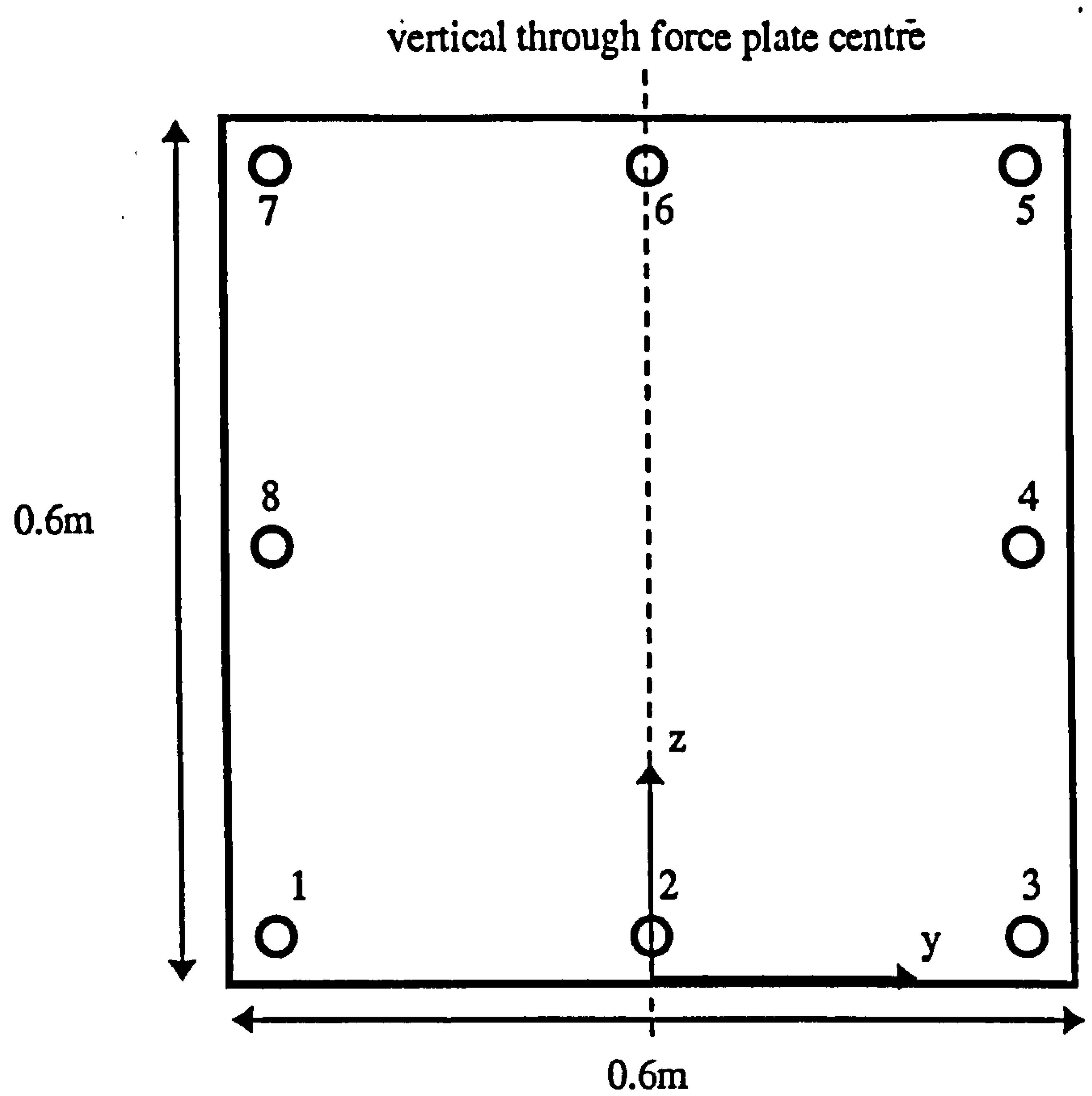


Figure 3.3 Calibration plane with marker locations

Synchronisation

The 1000 Hz force data were exported at 50 Hz to provide force data for every 0.02 s, matching up with the sampling rate of the video data. Code was written in BBC BASIC to synchronise the force and video data by use of the time of the field showing the L.E.D. Ground contact was defined as the time at which the vertical component of GRF exceeded 50 N, as commonly used in the literature (Cavanagh and Lafortune, 1980; Burdett,1982).

The following points were digitised for each video field:

point no.	**anatomical location**
1	wobble point 1
2	wobble point 2
3	knee joint centre
4	ankle joint centre
5	heel landmark
6	metatarsalphalangeal joint centre
7	toe landmark
8	Achilles point 1
9	Achilles point 2
10	Achilles point 3

These points were linked together to form a rigid link system representing the lower leg and foot, as illustrated in Figure 3.2.

Muscle moment calculations

Quasi-static moment methods as described by Morlock and Nigg (1988) were used for calculation of joint moments about the MTP, ankle and knee joint centres. The procedures used for calculation of ankle joint moment are provided in Figure 3.4 (Equation 3.1). Since it was not possible to ensure that the sagittal plane of the subject coincided with the calibrated two-dimensional plane, the centre of pressure x-coordinate (ax) corresponding with each video field of ground contact was used to correct for projection error (Appendix C).

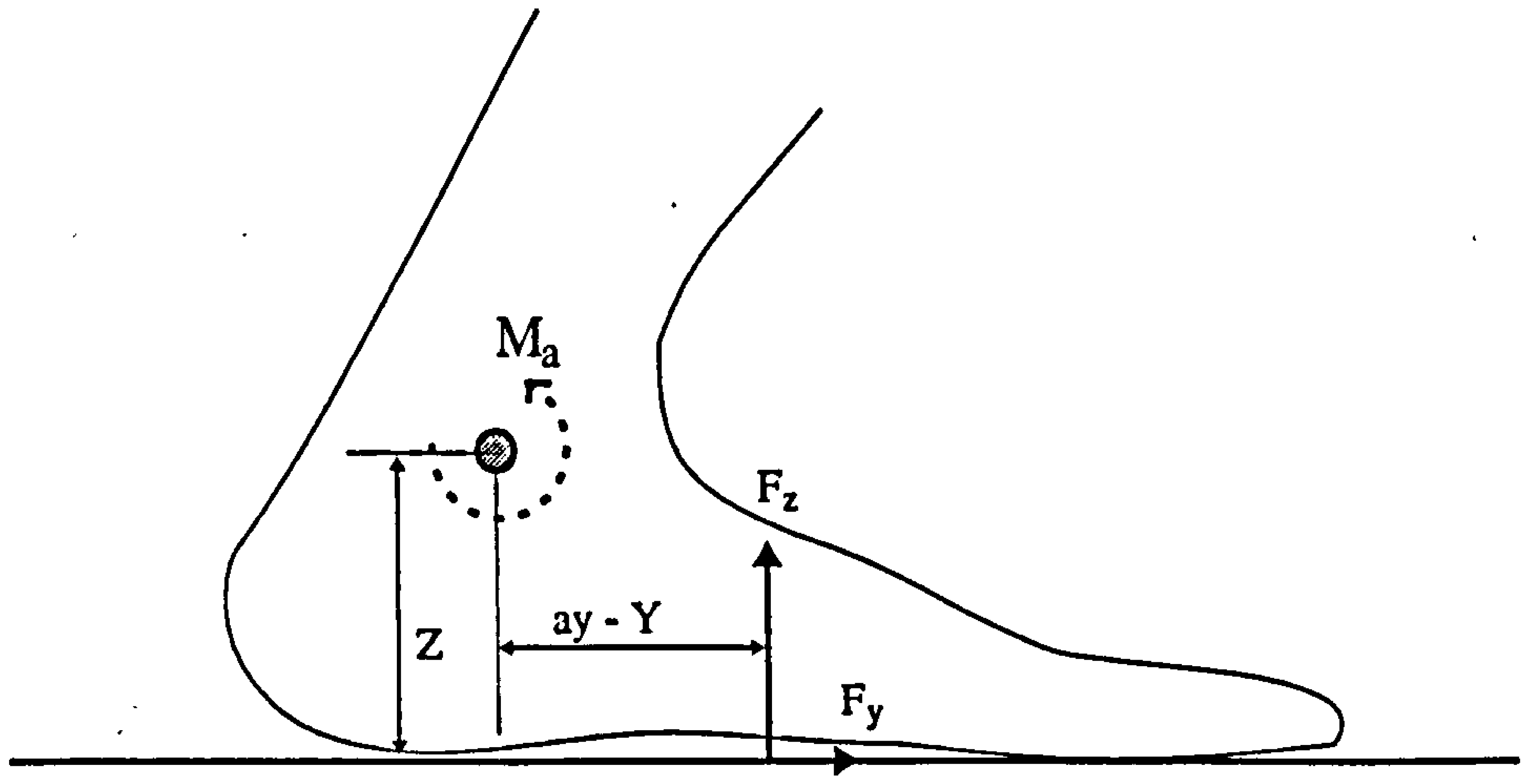


Figure 3.4 Calculation of quasi-static moments about the ankle joint centre

M_a = resultant moment of GRF about the joint
Y, Z = horizontal and vertical distances respectively
ay = y coordinate of the centre of pressure

$$M_a = (ay - Y).Fz + Z.Fy \tag{3.1}$$

Achilles tendon force calculations

It was assumed in the present study that the force in the Achilles tendon acts in the sagittal plane only. Additionally, it was assumed that the entire ankle plantar-flexion moment was developed by the triceps surae muscle group and was thus transmitted by the Achilles tendon. It was assumed that the direction of this force was in a straight line acting from the insertion to the origin of the muscle group, and that there was no antagonist muscle action. The calculation of Achilles tendon force using ankle moment data is illustrated in Figure 3.5.

The resultant muscle moment about a joint is defined as the product of the resultant force acting about the joint due to muscle action and the moment arm length of this resultant force (Equation 3.2). Thus, if the resultant moment and the moment arm length are known, it is possible to calculate the resultant force acting about the joint centre (Equation 3.3, Figure 3.5), in this case the Achilles tendon force.

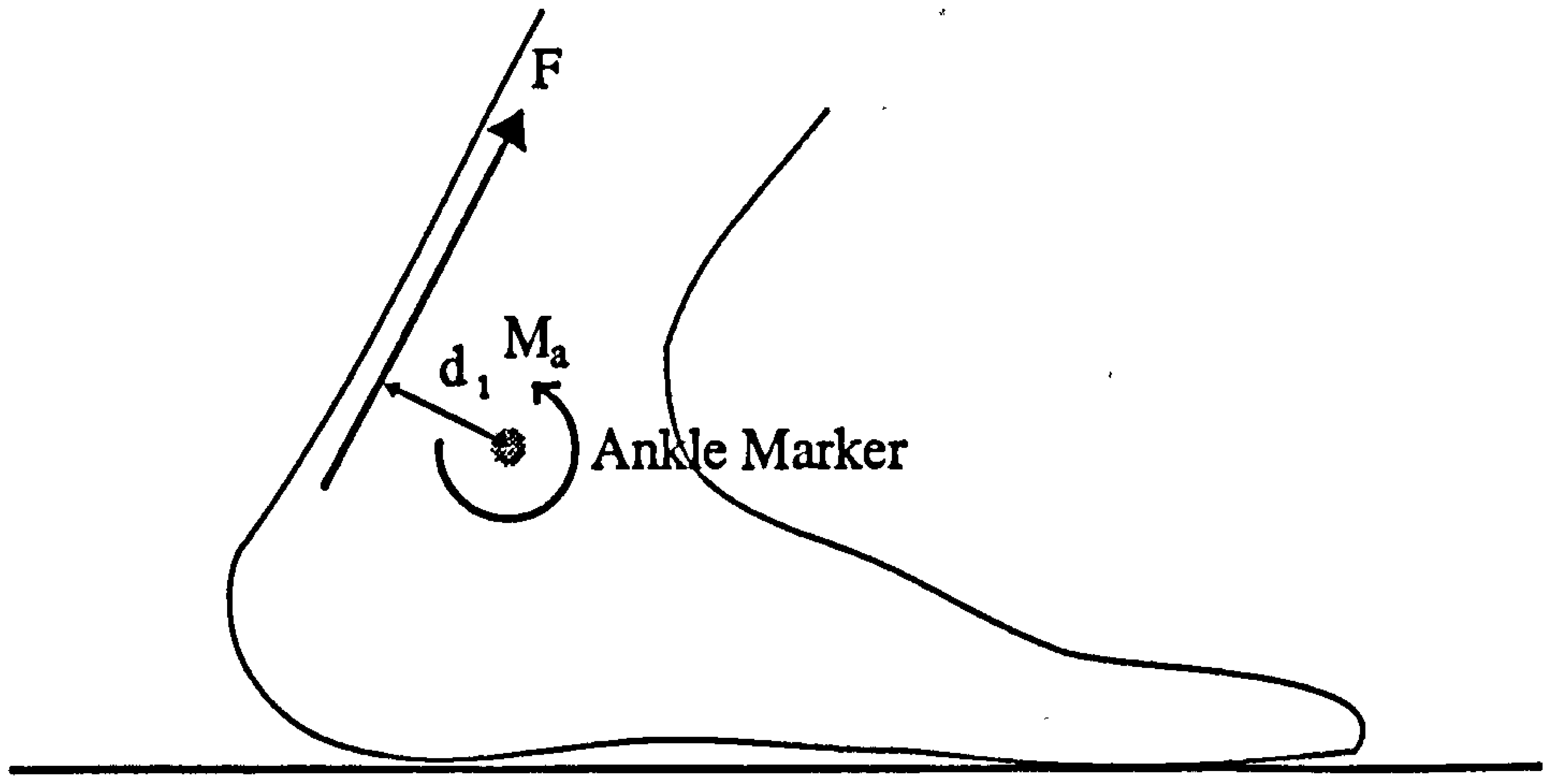


Figure 3.5 Forces and moments acting about the ankle joint centre

F = Achilles tendon force
d_1 = length of moment arm of resultant force about the ankle joint centre

$$M_a = d_1.F \qquad \text{(muscle moment)} \qquad (3.2)$$

$$F = M_a / d_1 \qquad (3.3)$$

For the initial barefoot study, four different methods were used for the calculation of the Achilles tendon line of action for use in estimation of the moment arm of the tendon (d_1, Figure 3.5). The Achilles tendon force values obtained when using the different methods were compared to investigate the suggestion that the method of Achilles tendon approximation influenced the estimated tendon force.

Method 1 involved the approximation of the point of insertion as being at the digitised heel point, and the line of action as being parallel to the line representing the lower leg orientation. The use of a line parallel to the lower leg was consistent with the method of Burdett (1982). However, in the Burdett study the insertion of the tendon was estimated by the use of cadaver data.

For Method 2, the Achilles tendon line of action was approximated as acting on a line from the digitised heel point to the knee marker.

The line of action of the Achilles tendon is likely to be somewhere between the Method 1 and Method 2 approximations, since these methods use extreme points of origin of the gastrocnemius muscle. A method using knowledge on the actual points of origin and insertion is likely to be more appropriate.

Method 3 involved the use of anatomical models of the foot and lower leg (Adam

Rouilly Anatomical Models). The point of insertion of the Achilles tendon was determined by the use of a foot model. The line of action of the tendon was assumed to run directly from the point of origin of the gastrocnemius to the insertion point of the tendon. The point of gastrocnemius origin was determined using a second anatomical model which included the lower leg. For determination of the point of tendon insertion, horizontal and vertical distances on the model from the ankle joint centre to the point of tendon insertion were measured in the sagittal plane. These measurements were scaled using the length of the foot for horizontal scaling, and the distance from the heel base to the ankle joint centre for vertical scaling for both the models and the subject (Table 3.1).

The calculated distance from the point of Achilles tendon insertion to the ankle joint for the subject was $= (39.2^2 + (74-29.6)^2)^{½} = 59.2$mm (Table 3.1).

Table 3.1 Measured distances on the model and on the subject
(figures in *italics* have been calculated using the appropriate scaling factor)

Distance	Model (mm)	Subject (mm)	Scaling Factor
Foot Length	253	236	0.933
Ground - Ankle	75	74	0.987
Horizontal Insertion		42	*39.2*
Vertical Insertion		30	*29.6*
Ankle - Heel (hori)	56	*52*	
Ground - Heel (vert)	14	*13.8*	

Due to the structure of the calcaneus and talus bones and their articulation, it was assumed that the distance from the ankle joint to the point of Achilles tendon insertion remained constant throughout the movement, and the distance from the ankle joint to the heel point on the calcaneus remained constant. The acute angle between a straight line from the ankle joint marker to the point of tendon insertion, and a straight line from the ankle joint marker to the heel marker was termed alpha (Figure 3.6). Since the rearfoot was assumed to be a rigid structure, the angle alpha remained constant, allowing calculation of the position of the Achilles tendon insertion point from the digitised data points. The angle alpha was found to be close to zero (0.4 degrees). Since this angle was determined using several assumptions, and was so small, the assumption was made that this value was zero throughout the movement. Thus, the point of Achilles tendon insertion was assumed to lie on a straight line from the ankle joint to the heel point (Figure 3.7), at a constant location on this line.

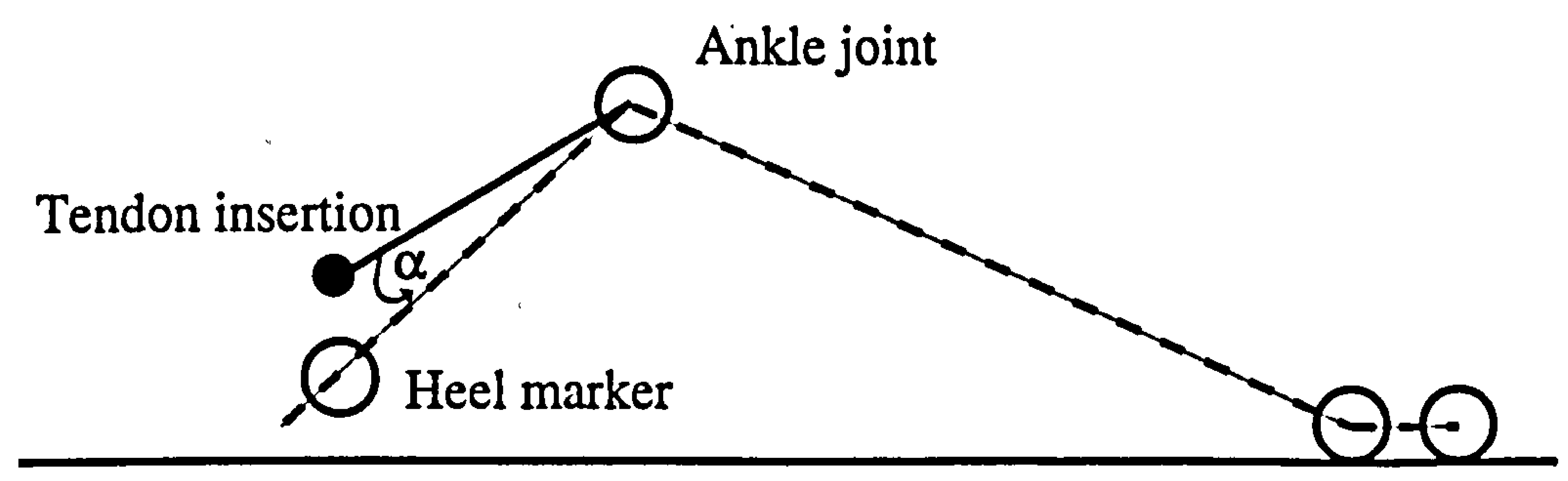


Figure 3.6 Relationship between digitised points and Achilles tendon insertion

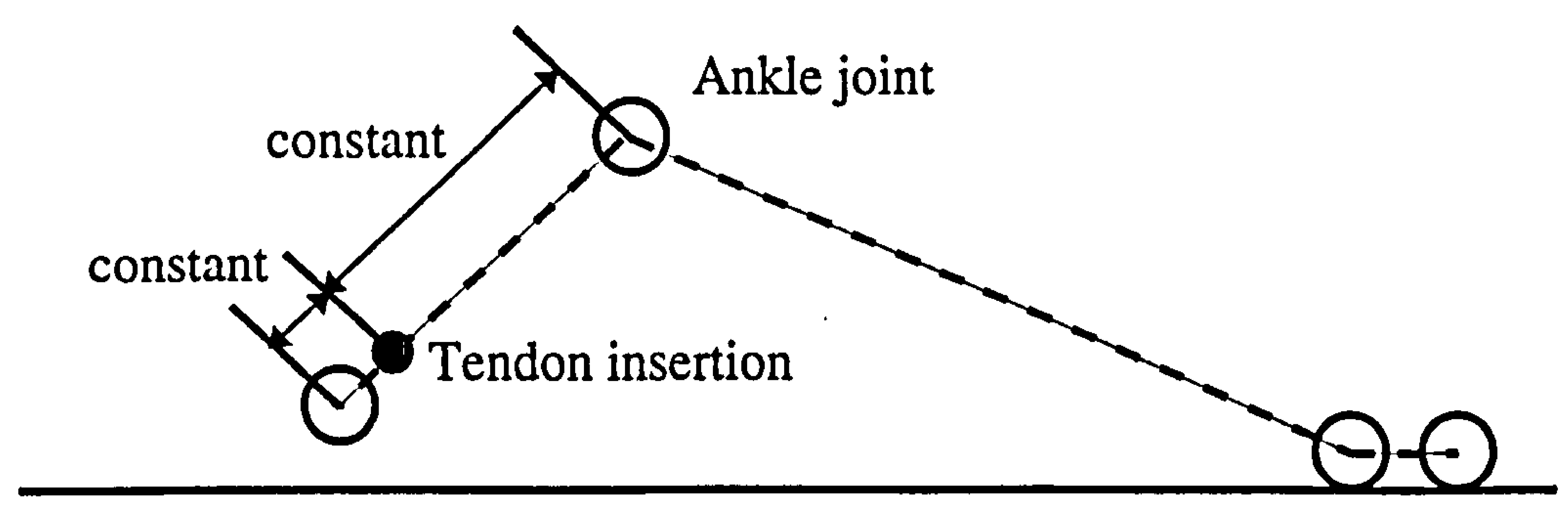


Figure 3.7 Estimated location of Achilles tendon insertion

It was assumed that the point of origin of the gastrocnemius was on a horizontal line through the knee joint centre. The location of the tendon origin was located on the model and the distance from the knee joint centre to the origin measured. A horizontal scaling factor was calculated by measurement of foot length on the model and the subject. This factor was used to estimate the location of the gastrocnemius origin on the subject. The horizontal location of the origin relative to knee joint centre was calculated as -46 mm. The estimated origin and insertion points were joined by a straight line to represent the line of action of the Achilles tendon in each field of motion. The associated moment arm length of the Achilles tendon about the ankle joint centre was calculated, allowing determination of the Achilles tendon force.

For Method 4, points marked on the skin covering the Achilles tendon were digitised. Three markers were drawn on the skin to represent the centre line of the tendon in the sagittal plane. The two outer points were used to represent the line of action of the tendon for the barefoot trials.

Coding for all calculations of ankle moments and Achilles tendon forces was performed in BBC BASIC (Appendix D).

Influence of ankle marker location

The precision with which the most prominent point of the lateral malleolus could be identified using a marker was quantified subjectively to be within an area of 1 cm^2. The influence of a variation of ±0.5 cm in location of the ankle marker on GRF moment arm and Achilles tendon moment arm was therefore quantified. Typical GRF magnitude and moment arm length and Achilles tendon moment arm values of 1450 N, 0.115 m and 0.04 m respectively, were used to assess the subsequent influence on ankle moment and Achilles tendon force magnitudes.

(iii) The Influence of Shoes and Heel Lifts

10 running trials at 3.83 $m.s^{-1}$ were performed by the same subject as employed in Section 3.2 (iii) under each of the following conditions:

Condition B : barefoot
Condition B1 : barefoot with 6 mm heel lift (Sorbothane heel lift)
Condition B2 : barefoot with 12 mm heel lift
Condition S : running shoes (Reebok Ladies Pyro)
Condition S1 : running shoes with 6 mm heel lift (Sorbothane) in the shoe
Condition S2 : running shoes with 12 mm heel lift (Sorbothane) in the shoe

Commercially available heel lifts made from Sorbothane, a viscoelastic polymer, were used for the heel lift conditions. For conditions B1 and B2, these lifts were attached to the plantar surface of the foot below the calcaneus by the use of surgical tape. For the running shoe conditions, a 6 mm heel lift was inserted in each shoe for condition S1, and a 12 mm heel lift in each shoe for condition S2. The use of a 12 mm heel lift in method S2 was designed to allow investigation into the influence of raising the heel to a greater extent than normally achieved by the use of a standard 6 mm heel lift. When the subject was wearing shoes, the bottom of the three Achilles tendon markers was obscured by the shoe. Thus, for the running shoe trials, the top two Achilles markers were used to approximate the line of action of the tendon.

Force and video data were collected as described in Section 3.2 (ii). Mean values for peak ankle plantar-flexion moments were calculated over the 10 trials for each condition. Maximum Achilles tendon force values were estimated using Method 4, and mean values over the 10 trials for each condition were calculated. A one way ANOVA was used to test for significant differences between conditions. A post-hoc Tukey test was used to detect differences between particular conditions.

Where differences in ankle moment and Achilles tendon force were detected between conditions, the mechanism by which these changes occurred was investigated. Using the quasi-static methods employed in the present study, peak ankle plantar-flexion moment was dependent on the resultant GRF and the moment arm of this force about the ankle joint centre. Thus, the magnitude of these variables at the time of maximum ankle plantar-flexion moment were obtained. The maximum Achilles tendon force magnitude depended on the ankle moment and moment arm length of the Achilles tendon. The contribution of these variables to changes in maximum Achilles tendon force was investigated by obtaining their values at the time of maximum Achilles tendon force occurrence.

The magnitude and time of occurrence of the vertical GRF impact peak, and maximum loading rate of vertical GRF were obtained for each condition.

Accuracy in locating markers was quantified by calculation of the RMSD between the known calibration plane coordinates and the locations obtained over 10 repeated digitisations. The influence of operator random error on measured marker locations was assessed by repeated digitisation of a randomly selected video field 10 times. The mean RMSD over the digitised markers was calculated using deviation from the mean marker location over the 10 repeated digitisations. To investigate the reliability of the calculated Achilles tendon line of action obtained using the procedures of Method 4, the differences in maximum Achilles tendon force values obtained using different combinations of the three tendon markers were quantified. The influence of the number of wobble points employed was investigated by comparison of RMSD values obtained using zero, two and four wobble points.

A power analysis was performed using the power tables for F-tests provided by Cohen (1982; pp. 282-347). Individual power tables are illustrated for different significance levels (a) and degrees of freedom of the numerator of the F ratio (u). Six conditions were employed in the present study, and thus a u value of 5 (number of conditions -1) was set. The influence of using significance levels of 0.01 and 0.05 were compared, using the respective power tables (a=0.01, u=5; a=0.05, u=5). From each of these tables it was possible to read off the power corresponding to a defined effect size index (f) and number of trials (n). The effect size index (f) was calculated using the formulae provided by Cohen (1982). The necessary formulae are described by Equation 3.4 and Equation 3.5, with descriptions of character abbreviations provided below the equations. Statistical powers corresponding to the calculated f value were obtained from the tables for significance levels of 0.01 and 0.05 (u=5, n=10).

$$f = \sigma_m / \sigma \qquad (3.4)$$

$$\sigma_m = \sqrt{\frac{\sum_{i=1}^{k} (m_i - m)^2}{k}} \qquad (3.5)$$

σ is the common within population standard deviation
k is the number of means being compared
m_i are the individual sample means
m is the mean of the combined populations

Cohen (1990) defined the common within population standard deviation (σ, Equation 3.5) as the square root of the error variance of the F-test which is to be performed. The error variance corresponds to the denominator for the F-test in question. The common within population standard deviation across the six conditions employed in the present study was therefore calculated using the Var_{within} value obtained from the ANOVA results table (Appendix E).

3.3 Results

(i) Questionnaire

Mean and standard deviation values for the age, summer weekly mileage, and winter weekly mileage are presented in Table 3.2. Data are provided separately for all subjects, subjects who have ever experienced an Achilles tendon (AT) injury, and subjects who have experienced an Achilles tendon injury in the past three years. Of the 50 athletes who responded to the questionnaire, 22 (44%) reported an occurrence of Achilles tendon injury at some stage in their running career. 16 athletes (32%) reported an Achilles tendon occurrence within the past three years.

Table 3.2 Mean age, and mileage for athletes for different AT injury incidence (SD)

	n	age	summer mileage	winter mileage
all athletes	50	28.4 (6.0)	51.3 (15.9)	61.6 (14.7)
athletes reporting AT injury	22	28.4 (5.6)	50.5 (15.5)	61.3 (16.0)
athletes reporting no AT injury	28	28.4 (6.4)	52.0 (16.4)	61.8 (14.0)
AT injury within past 3 years	16	27.1 (3.9)	51.6 (16.5)	61.9 (15.8)
no AT injury within past 3 years	34	31.0 (7.4)	49.8 (14.9)	60.8 (16.0)

(ii) Estimation of Achilles Tendon Forces

The following results refer to 10 successful, barefoot running trials.

Scaling Methods

It was found that the kinematic data obtained using the two different methods of scaling were similar, with a maximum RMSD of 1.6 mm (Table 3.3). RMSD was found to be 7-8 times greater for locating horizontal coordinates, than for locating vertical coordinates.

Table 3.3 Root mean square differences between DLT and linear scaling methods

	Total number of points considered	RMSD in y (m)	RMSD in z (m)
Trial 1	72	0.0016	0.0002
Trial 2	80	0.0014	0.0002

Ground reaction force

A typical trace of vertical ground reaction force is presented over fraction of stance time in Figure 3.8. To provide data relative to total stance time, 1000 Hz data were interpolated to 100 points over total stance. Since total stance time was approximately 0.2 s, the resulting data were presented at around 500 Hz. The mean total stance time over the ten barefooted trials was 0.194 s (±0.021 s). Over the ten trials, the mean peak vertical impact force was 1390 N or 2.6 multiples of bodyweight (BW). This force occurred at 4% total stance time, which corresponded to approximately 0.007 s (7 ms) after initial ground contact.

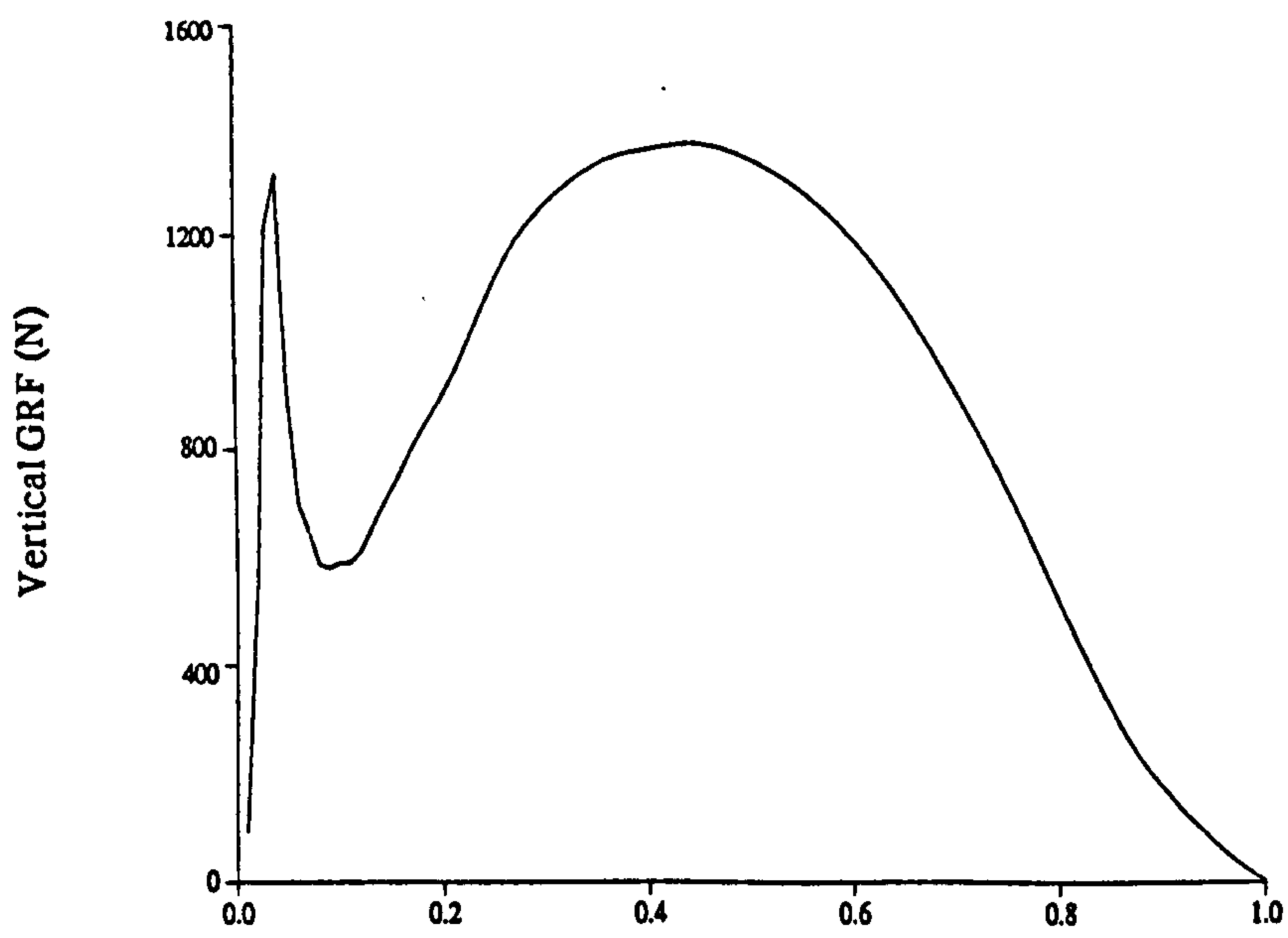


Figure 3.8 A typical trace of vertical GRF against normalised stance time

Ankle moments

A typical plot of 50 Hz ankle moment against time is provided in Figure 3.9. It is evident that 50 Hz data results in a fluctuating graph, as opposed to a smooth trace. In order to provide mean plots over the 10 trials which had varying ground contact times and were not in phase, percentage of total stance time was used. For each set of ankle moment data, interpolation was used to provide 100 data points over the entire stance period. The peak ankle plantar-flexion moment occurred at approximately 50% of total stance time.

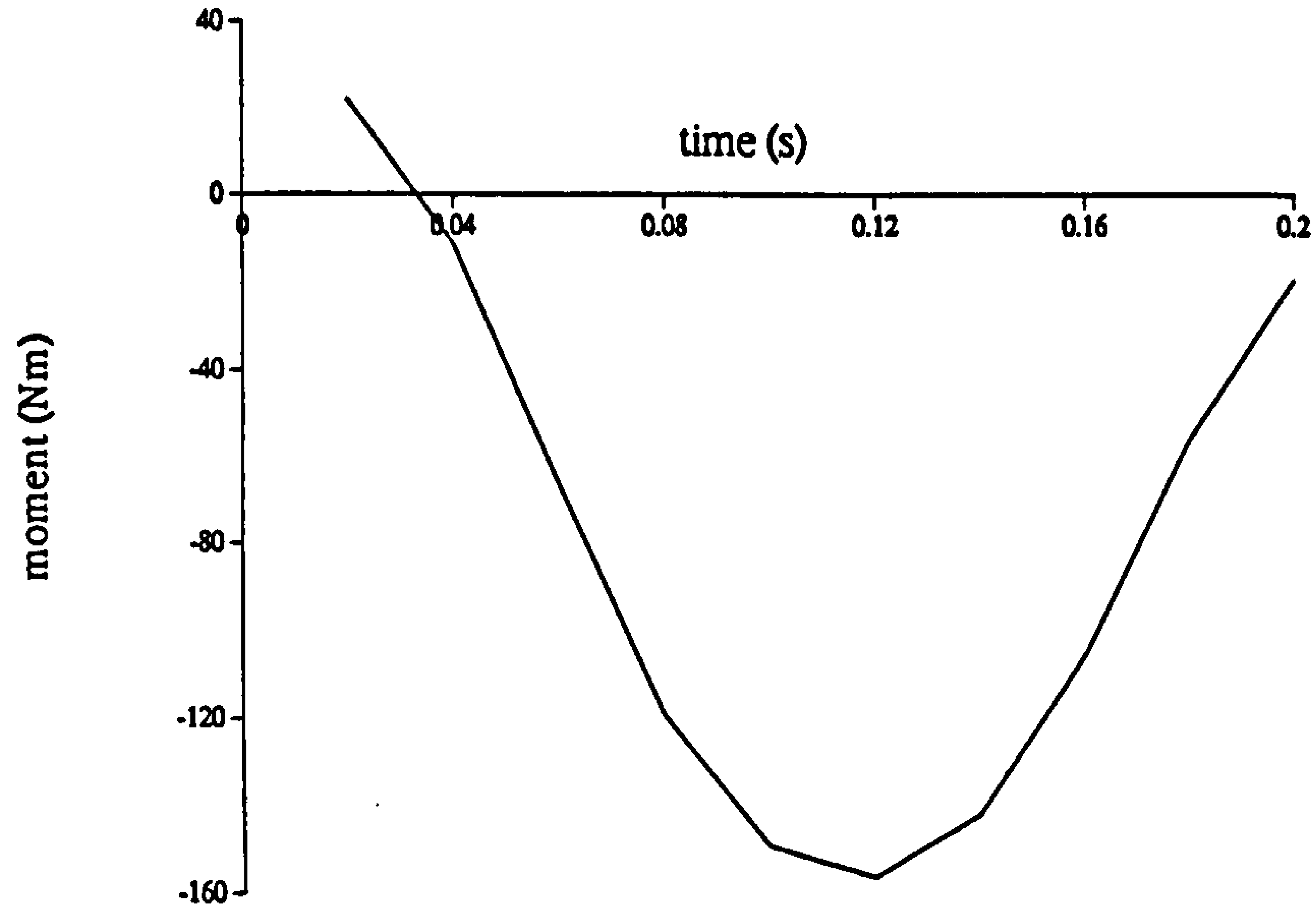


Figure 3.9 A typical trace of ankle GRF moment for a single trial (50 Hz)

Achilles tendon forces

Table 3.4 provides Achilles tendon force data calculated using each of the four methods. Mean data over the ten trials are presented for the maximum Achilles tendon force and the mean of the Achilles tendon forces throughout ground contact. The force data are presented as multiples of body weight of the subject (BW). A 50 Hz Achilles tendon force trace obtained for a single trial using Method 4 is presented in Figure 3.10. These data are presented in real time to illustrate the timing of the generation of Achilles tendon force and the influence of using 50 Hz data.

Figure 3.11 illustrates the variation in Achilles tendon force calculated using each of the four methods of Achilles tendon representation. Mean results are presented for the same 10 successful trials for each of the methods, with interpolation to 100 points being used to synchronise the trials. The general pattern of Achilles tendon force trace is similar across methods, but clear differences are demonstrated in the peak Achilles tendon force values attained using each of the Achilles tendon line of action approximations.

The peak value occurs at approximately the same time using each of the methods of tendon approximation. This peak is at approximately 50% of the total contact time for each of the methods.

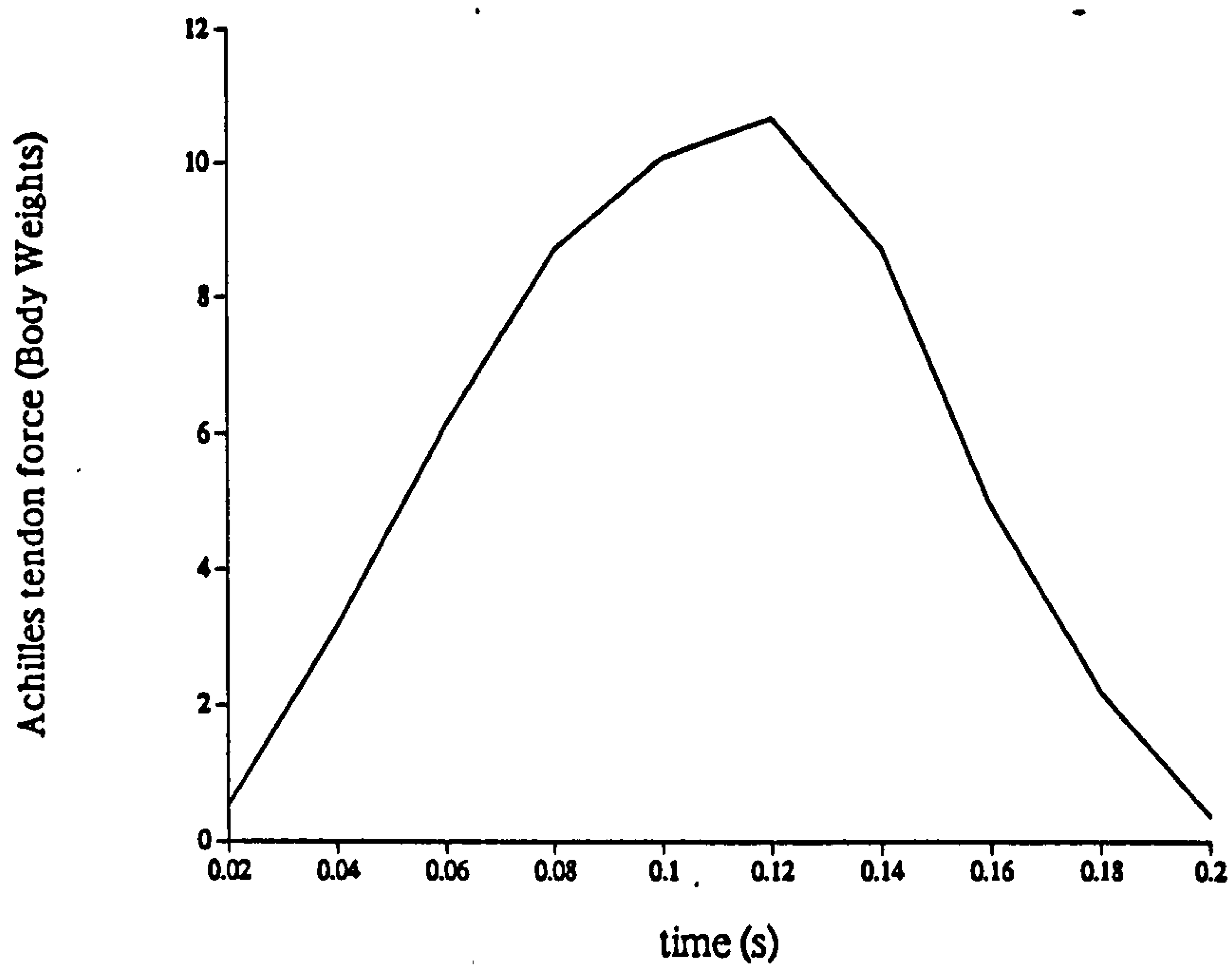


Figure 3.10 Achilles tendon force trace for a single running trial

Table 3.4 Achilles tendon force (BW±SD)

	Method 1	Method 2	Method 3	Method 4
Maximum	13.7±1.22	15.7±1.40	16.3±1.18	10.8±0.42
Mean	6.6±0.42	7.6±0.47	8.1±0.51	5.9±0.45

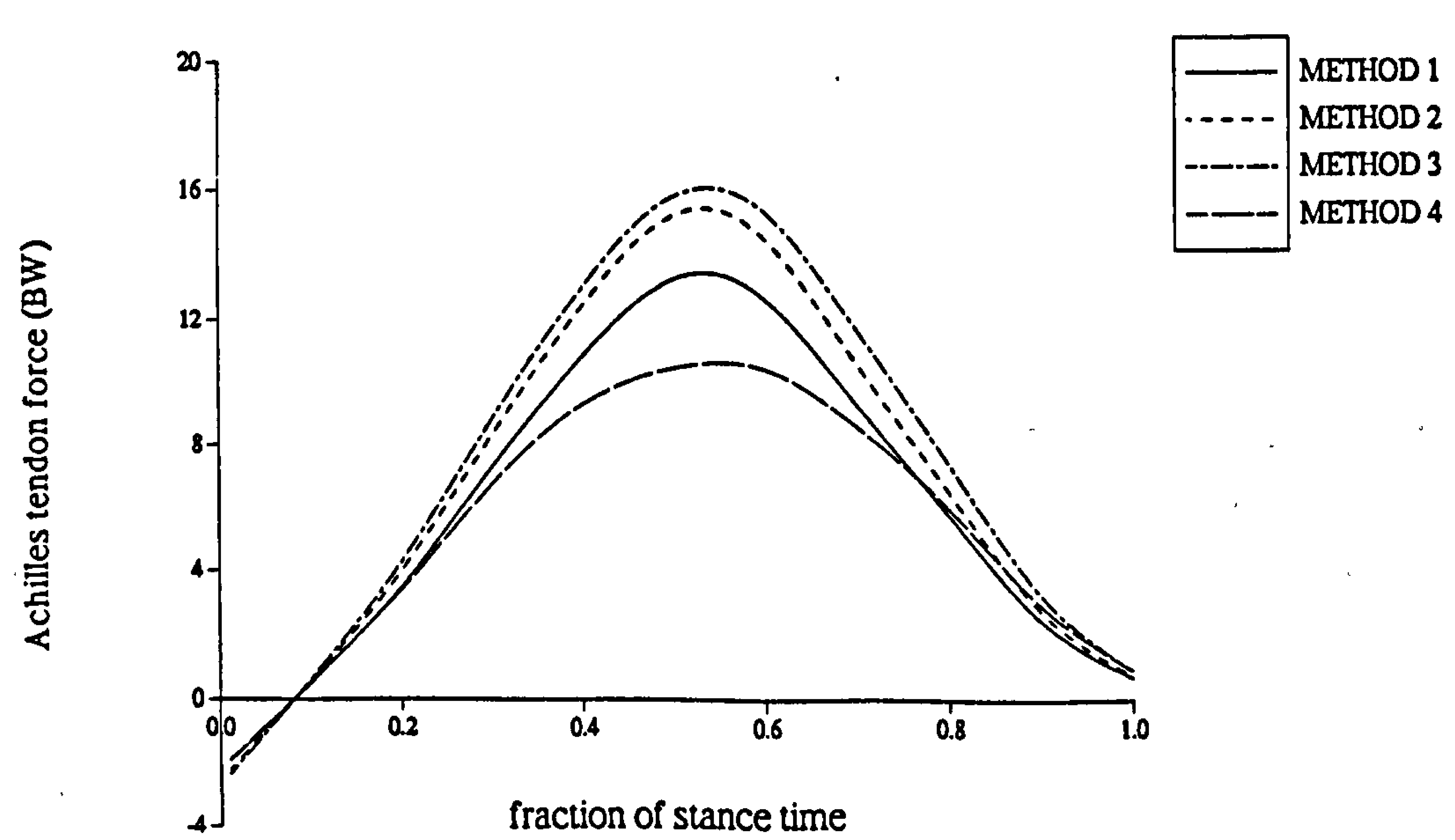


Figure 3.11 Mean Achilles tendon force over 10 trials using each of the four methods

Influence of ankle marker location

The resultant GRF was close to the vertical at the time of maximum Achilles tendon force occurrence. Thus a variation of 0.005 m in ankle joint marker location was found to result in a maximum variation in moment arm length of approximately 0.005 m. A resulting maximum possible variation in ankle moment of ±7.25 N.m (4.4%) was found. For typical values at the time of maximum Achilles tendon force of resultant GRF and GRF moment arm length of 1450 N and 0.115 m respectively, an ankle moment value of between 159.5 N.m and 174 N.m was obtained, depending on whether the moment arm length was increased or decreased by the error.

Variation in ankle marker location will also influence the length of the moment arm of the Achilles tendon. Visual observation of the video recordings demonstrated that the line of action of the Achilles tendon was orientated at approximately 45 degrees to the horizontal in the middle of the stance phase when maximum Achilles tendon force was found to occur. A displacement of 0.005 m in the horizontal and vertical directions simultaneously would therefore result in a maximum variation of approximately 0.007 m. Thus, for maximum error, the moment arm of the GRF would vary by 0.005 m and the moment arm of the Achilles tendon would vary by 0.007 m. For a typical Achilles tendon moment arm length of 0.04 m, a maximum Achilles tendon force magnitude of between 3702 N and 4833 N was obtained (6.9 BW and 9.0 BW). This is a variation of 1131 N, or 2.1 BW.

It was therefore concluded that variation in ankle marker location has a relatively small influence on estimated ankle joint moments. However, this variation in moment value together with the variation in the moment arm length of the Achilles tendon, causes a variation in estimated Achilles tendon force of up to approximately 17%.

(iii) The Influence of Shoes and Heel Lifts

Achilles tendon forces were estimated using Method 4 under different heel lift conditions. For each variable, mean values over the 10 trials for each condition are presented, with standard deviations in parenthesis. Statistically significant differences between barefoot and heel lift intervention conditions are noted below results tables.

Ankle joint moments

Table 3.5 provides data on peak ankle plantar-flexion moments under each of the conditions. Attaching heel lifts to the plantar surface of the foot was found to reduce the peak ankle moment compared with barefoot running. Wearing running shoes caused a further reduction in the peak moment. Compared with barefoot running, the reductions in ankle moment were significant for all conditions ($p<0.05$). No significant differences in peak ankle moment were detected when comparing across the three running shoe conditions.

Ankle moment values at the time of maximum Achilles tendon force are provided in Table 3.6. Since maximum Achilles tendon force and peak ankle moment occurred at the same video field for most trials, the values of ankle moment at maximum Achilles tendon force were very close to the values of maximum ankle moment.

Table 3.5 Maximum ankle moment (N.m±SD)

CONDITION	B	B1	B2	S	S1	S2
Maximum Ankle Moment	161.92 ± 6.32	156.27 ± 5.22	154.76 ± 6.46	149.03 ± 6.91	148.36 ± 9.26	148.33 ± 4.60

$p<0.05$: B versus B1; B2; S; S1; S2

Table 3.6 Ankle moment at maximum Achilles tendon force occurrence (N.m±SD)

CONDITION	B	B1	B2	S	S1	S2
Ankle moment	162 ±6	154 ±7	155 ±6	146 ±6	147 ±7	145 ±9

$p<0.05$: B versus B1; B2; S; S1; S2

Achilles tendon forces

Maximum Achilles tendon force values attained during ground contact using Method 4 for each of the conditions are presented in Table 3.7. Maximum Achilles tendon force was reduced significantly when heel lifts were attached to the plantar surface of the foot, with the use of the 12 mm heel lift having a greater influence than the use of the 6 mm lift. However, wearing running shoes did not reduce the maximum Achilles tendon force compared with the barefoot condition.

Table 3.7 Maximum Achilles tendon force (BW±SD)

CONDITION	B	B1	B2	S	S1	S2
Maximum Achilles Force	10.77 ±0.42	10.06 ±0.55	9.06 ±0.53	10.88 ±0.90	10.85 ±1.66	11.20 ±1.06

$p<0.001$: B versus B2; B1 versus B2

$p<0.001$: B2 versus S; S1; S2

$p<0.01$: B versus B1

Examination of the results for maximum Achilles tendon force (Table 3.7) revealed that the standard deviation values were of a similar magnitude for each of the barefoot conditions (B, B1, B2). This observation was supported by the use of a homogeneity of variance test for these three conditions ($p<0.05$). The standard deviations for the running shoe conditions are approximately twice the magnitude of the barefoot condition values (Table 3.6). A homogeneity of variance test revealed that the running shoe conditions demonstrated equal variance at a significance level of 0.01, but not at a level of 0.05. A homogeneity test for all six conditions revealed that the sample variances were not equal. The use of an ANOVA test for analysis of results was justified since these tests are robust with regard to the assumption of equal sample variance.

Ground reaction force

Normalised vertical GRF traces over 10 trials for each condition are provided in Figure 3.12 for the barefoot, running shoe and running shoe with 12 mm heel lift conditions, and in Figure 3.13. for the barefoot, barefoot with 6 mm heel lift and barefoot with 12 mm heel lift conditions. The wearing of running shoes resulted in a reduction in the peak impact force and a later occurrence of this force, with the use of a 12 mm heel lift within the shoe resulting in a further reduction in this variable. Similarly, compared to barefoot running, the attachment of heel lifts to the plantar surface of the foot resulted in a reduction in the peak impact force and an increase in the time to peak force. Differences from the barefoot condition in the peak impact force were more marked for the running shoe conditions than for the barefoot heel lift conditions. The vertical GRF traces were similar across all conditions after the initial impact phase.

Table 3.8 provides the mean value over 10 trials of the vertical GRF at the time of maximum ankle moment for each condition. No significant differences were identified across conditions in the magnitude of this GRF variable.

Table 3.8 Ground reaction force corresponding to maximum ankle moment (N±SD)

CONDITION	B	B1	B2	S	S1	S2
Maximum GRF	1374.5 ±41.6	1361.8 ±43.3	1372.4 ±49.4	1368.4 ±56.6	1331.2 ±49.8	1355.0 ±48.86

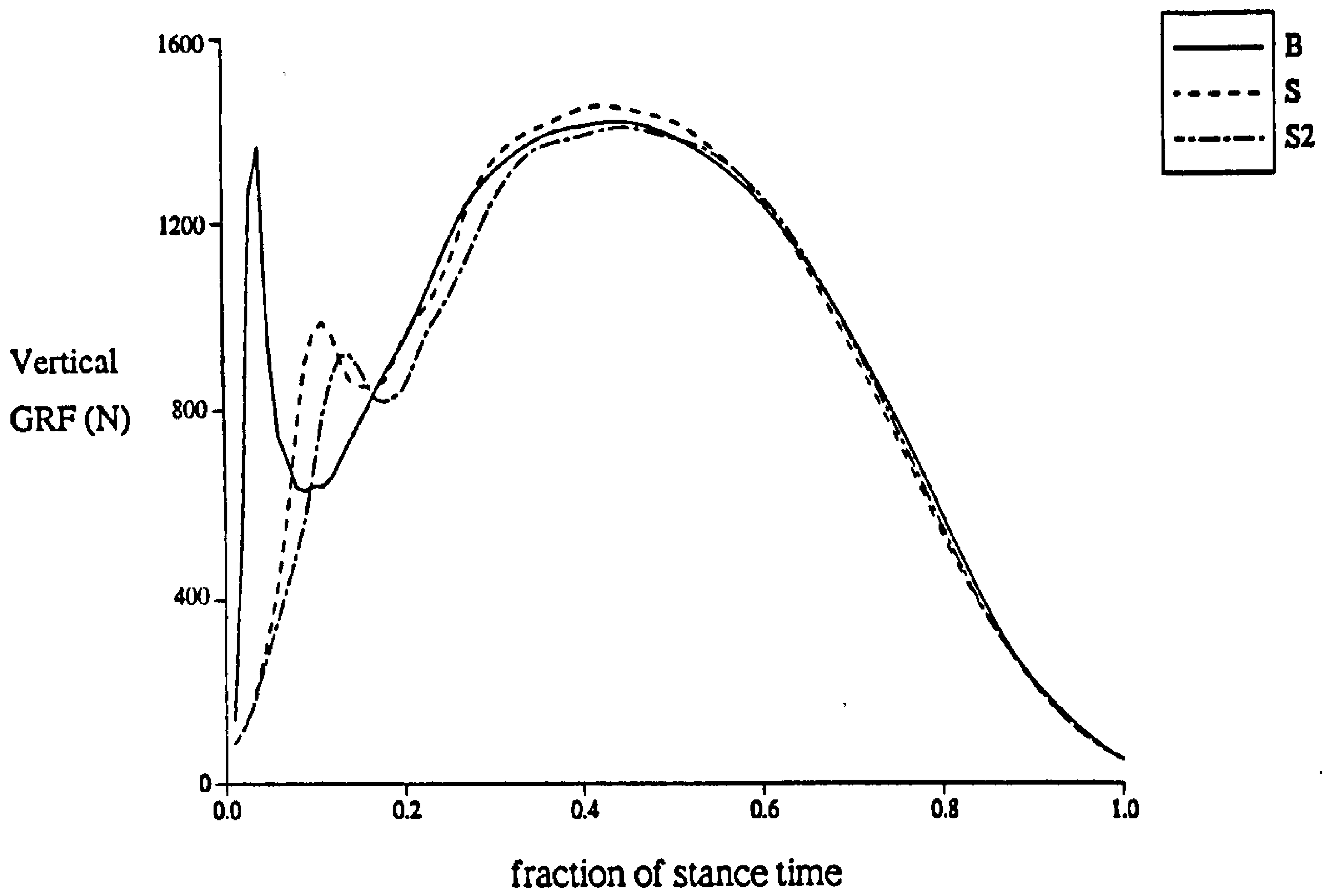


Figure 3.12 Vertical GRF traces for barefoot (B), shoes (S) and shoes with lifts (S2)

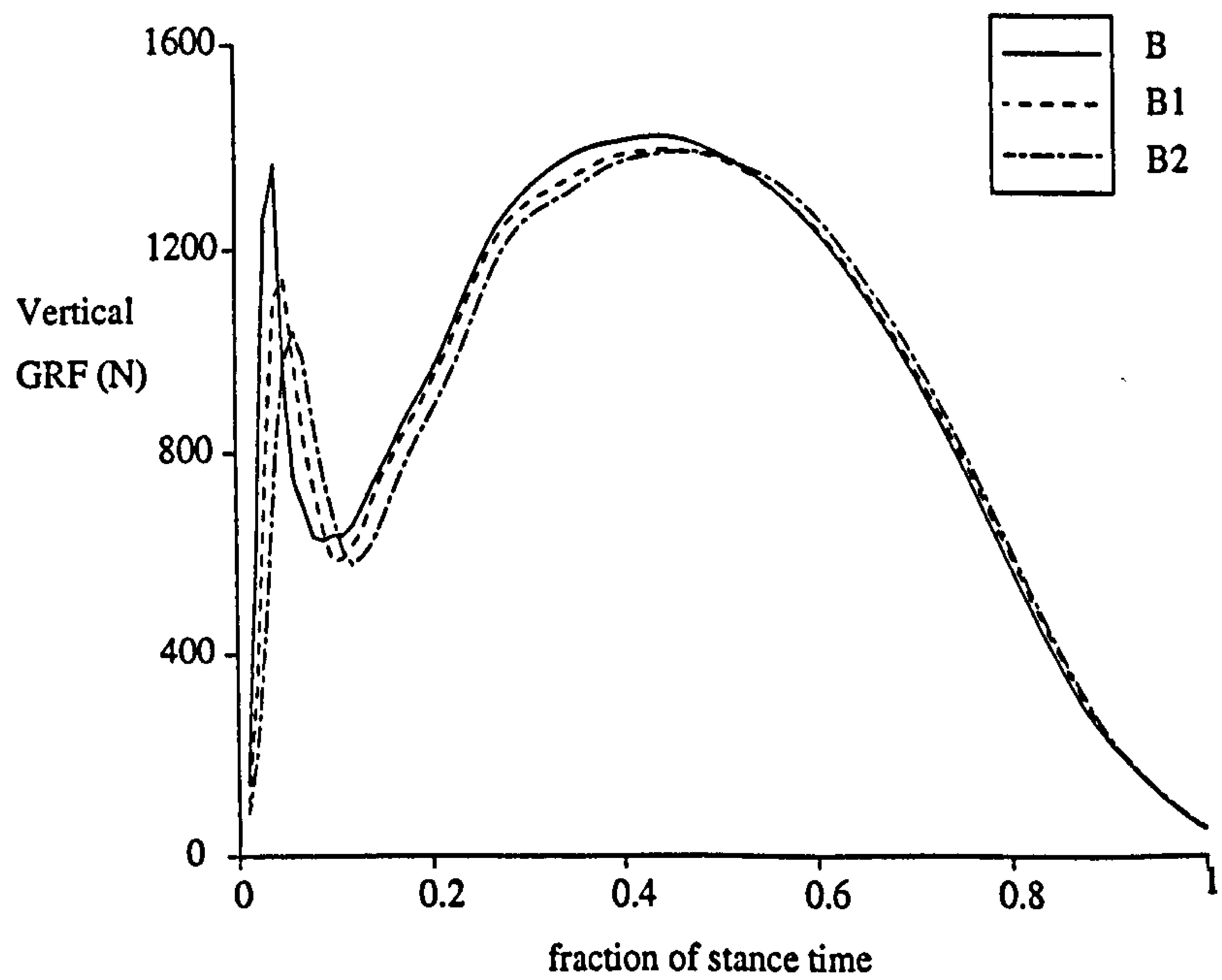


Figure 3.13 Vertical GRF traces for barefoot (B), 6 mm lifts (B1) and 12 mm lifts (B2)

The maximum rate of increase of vertical GRF, or loading rate, for the conditions barefoot, running shoe, and running shoe with 12 mm heel lifts are presented in Table 3.9. GRF loading rate was reduced by the wearing of shoes, and to a greater extent when heel lifts were used in addition.

Table 3.9 Maximum loading rate for conditions B, S and S2

CONDITION	B	S	S2
Maximum Loading Rate (BW. s^{-1})	604	196	151

Moment arms

Achilles tendon moment arm length values corresponding to the time of maximum Achilles tendon force occurrence are presented in Table 3.10. The moment arm length was significantly increased by the raising of the heel in relation to the forefoot using the 12 mm heel lifts. In contrast, wearing running shoes caused the moment arm length to be reduced, with this length being reduced further by the introduction of heel lifts in the shoe.

GRF moment arm lengths for each condition are presented in Table 3.11. This moment arm length was reduced when the heel was raised in relation to the forefoot, either by the use of heel lifts or the use of shoes. The placement of heel lifts in the running shoes did not further reduce the length of this moment arm.

Table 3.10 Length of Achilles tendon moment arm about ankle joint centre (m±SD)

CONDITION	B	B1	B2	S	S1	S2
Achilles tendon moment arm	0.0280 ±0.001	0.0284 ±0.002	0.0321 ±0.002	0.0258 ±0.001	0.0250 ±0.001	0.0247 ±0.002

$p<0.01$: B versus B2

Table 3.11 Length of moment arm of GRF about ankle joint centre (m±SD)

CONDITION	B	B1	B2	S	S1	S2
Moment arm of GRF	0.1178 ±0.004	0.1129 ±0.004	0.1128 ±0.002	0.1090 ±0.004	0.1114 ±0.002	0.1095 ±0.004

$p<0.01$: B versus B2; S; S1; S2

$p<0.05$: B versus B1

Use of different methods

To assess the reliability of Method 4 under conditions other than barefoot, the standard deviations obtained using each of the original four methods were calculated for the 12 mm heel lift (B2) and the running shoe (S) conditions. These values are presented in Table 3.12. The smallest standard deviation values were demonstrated for Achilles tendon force values obtained using Method 4.

Table 3.12 Standard deviation for conditions B2 and S using each of the four methods (BW)

CONDITION	B2	S
Method 1	1.25	0.98
Method 2	1.40	1.10
Method 3	1.36	1.11
Method 4	0.53	0.90

Digitisation

The level of accuracy in reconstruction of point locations, represented using RMSD from the measured coordinates on the calibration frame, was found to be 0.0004 m in the horizontal (y) direction, and 0.0006 m in the vertical (z) direction.

Over the 10 repeated digitisations, RMSD values of 0.0006 m in the y direction and 0.0004 m in the z direction were obtained. For the analysed video field, the subsequent RMSD in Achilles tendon moment arm length and Achilles tendon force value were 0.0012 m and 0.56 BW respectively. The use of different numbers of wobble points was found to have no influence on the accuracy or reliability of digitised points.

Tendon marker reliability

The maximum Achilles tendon force values obtained using different combinations of tendon points are presented in Table 3.13. The Achilles tendon marker placement is illustrated in Figure 3.14. No significant differences in estimated Achilles tendon forces were demonstrated for each of the combinations of Achilles tendon markers ($p<0.001$).

Table 3.13 Maximum Achilles tendon force using different marker combinations (BW±SD)

Combination	A1 / A2	A1 / A3	A2 / A3
Maximum Tendon Force	10.97 ± 0.63	10.96 ± 0.41	11.18 ± 0.91

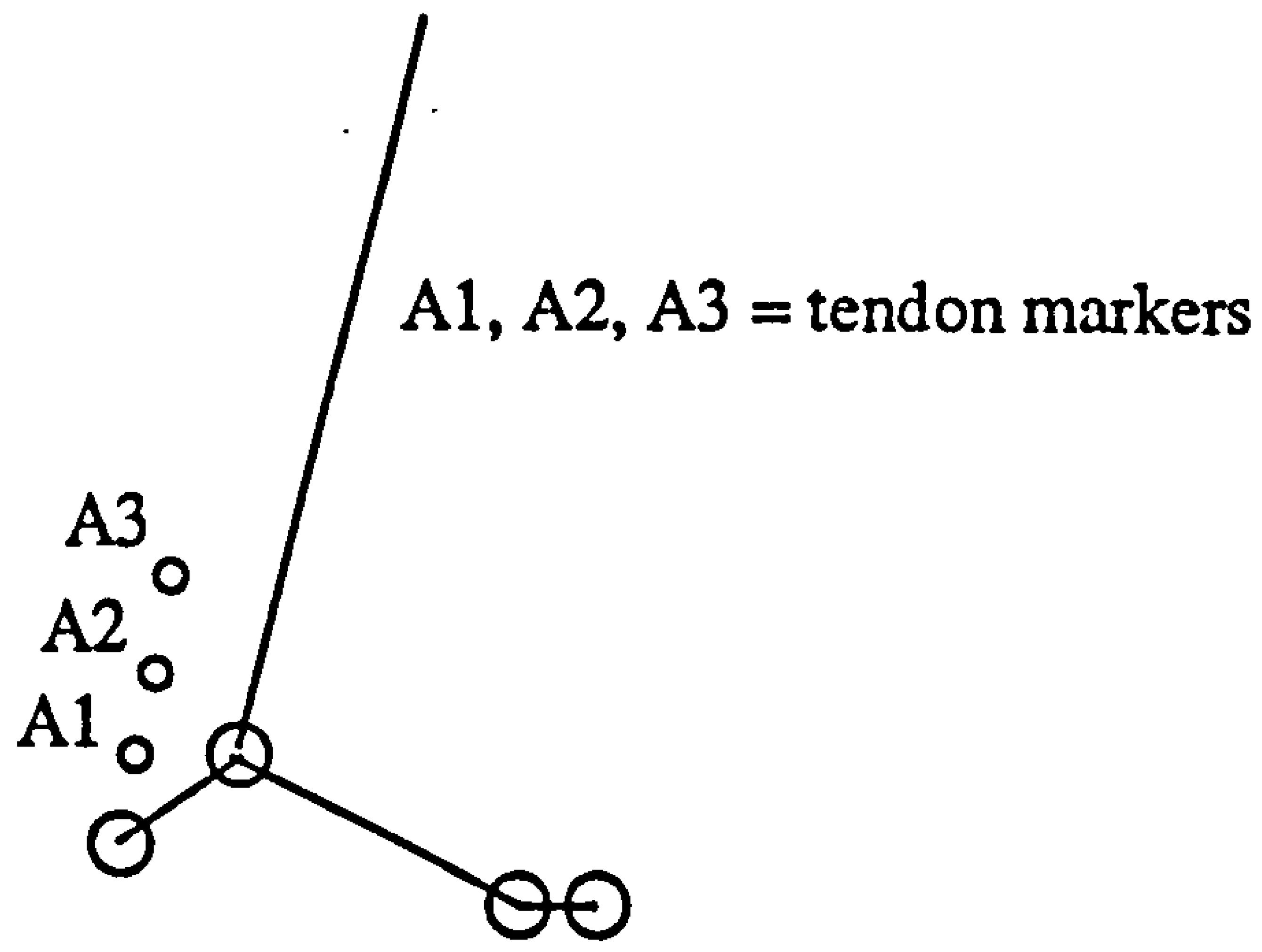


Figure 3.14 Illustration of Achilles tendon markers

Power analysis

It was assumed that the range of maximum Achilles tendon force values and associated standard deviations obtained in the present study was typical. A power analysis was therefore performed using the results for all six conditions. From Appendix E, a Var_{within} value of 0.88 BW was obtained, providing a common within population standard deviation of 0.94 BW. Substitution of the maximum Achilles tendon force results into Equation 3.5 provided a σ_m value of 0.7 BW. Thus, an effect size index (f) of 0.74 (0.7/0.94) was obtained. This value was used to represent the expected effect size in studies of the influence of footwear interventions on maximum Achilles tendon force. The power of the F-test used in the present study was calculated using n=10 and u=5, for different levels of significance. Power values of 0.95 ($p<0.01$) and 0.99 ($p<0.05$) were obtained. Since high powers were obtained for these levels of significance, a significance level of 0.1 was not considered. The increase in power when moving from a level of 0.05 to 0.1 would be negligible compared with the 5% increase in the probability of a Type I Error. The influence of varying sample size for significance levels of 0.01 and 0.05 is illustrated in Table 3.14, and Table 3.15.

Table 3.14 Variation in power for a significance level of 0.01 (u=5, f=0.7)

sample size (n)	power
2	0.07
4	0.32
6	0.64
8	0.85
10	0.95
12	0.98
*13	0.99

* for n>13, power > 0.995

Table 3.15 Variation in power for a significance level of 0.05 (u=5, f=0.7)

sample size (n)	power
2	0.21
4	0.61
6	0.86
8	0.96
*10	0.99

* for n>10, power > 0.995

3.4 Discussion

(i) Questionnaire

Two methods of collecting data on injury occurrence have frequently been described in the literature. Data have been collected by recording the number of occasions specified injuries have been presented at injury clinics (Clement et al., 1984), or competitors in selected races have been questioned. It was felt by the author of the present research, an elite female distance runner, that the true frequency of Achilles tendon injury occurrence is greater than implied by the available literature. The questionnaire designed in the present study was aimed at quantifying the percentage of female distance runners competing at an elite level who have been forced to interrupt their training due to an Achilles tendon injury.

In order to quantify the occurrence of Achilles tendon injury, it was necessary to define the term 'injury'. A survey of sports injuries in the UK (Nicholl et al., 1991) defined injury as pain preventing an individual from maintaining usual activity levels for more than 48 hours.

As it would not concern the athlete, it was felt that any injury that did not interfere with usual training was not important. Thus, injury was defined in the present study as damage to the Achilles tendon which resulted in disruption of usual training over a defined time period. A period of one week was chosen, as this would cause a noticeable disruption to training, and likely deterioration of fitness levels. Achilles tendon injury was therefore defined, for the purposes of the questionnaire of the present study, as pain in the Achilles tendon which resulted in usual training being missed for one week or more. It was felt that the word 'usual' was required since many athletes will continue with a reduced amount of training when feeling pain.

The figure of 44% obtained in the present study for frequency of Achilles tendon injury occurrence, demonstrates that a large proportion of elite runners have their training interrupted by this injury. No time specification was set with regard to when this injury occurred. The athletes will have been competing for differing numbers of years, and thus some would possibly be more or less likely to have injury of any type than others. For this reason, the second Achilles tendon injury occurrence question was set, asking whether the athlete had experienced the Achilles tendon injury within the last three years. The period of three years was chosen as all the athletes were of senior age (over 20), and thus were likely to have been competing at a high level for at least this number of years. A greater number of years would have increased the likelihood of differences occurring between athletes in the number of years competing and training at a high level. A smaller number of years may have resulted in some athletes who had experienced an Achilles tendon injury whilst competing and training at this level having their injury incidence overlooked in the study.

The frequency of 32% of athletes experiencing an Achilles tendon injury during the last three years is greater than any of the figures provided in the literature, but comparisons are limited by the different procedures used to obtain values. James et al., (1978) and Clement et al., (1984) surveyed athletes attending injury clinics, and thus provided data on the percentage of all the injuries presented that were to the Achilles tendon. The figures presented in these studies clearly cannot be compared directly to those obtained in the present study. An alternative method employed in the literature has been to survey runners competing in selected races. The obvious appeal of this approach is the opportunity to question large numbers of athletes in a single session. However, a limitation of this approach is the fact that those runners not competing due to injury will be omitted from the study. In a report published by the Sports Council (Nicholl et al., 1991) sports injury sites were identified by use of a postal survey in which subjects' addresses were obtained through their General Practitioners. Unfortunately, sites of injury as being the foot or ankle were used as categories, providing no information on the frequency of Achilles tendon injury. The aim of the Sports Council report was to obtain information on sports injury occurrence that was representative of the whole population, whereas a small unique group was of interest in the present study. By surveying athletes in the British squad, the aim in this study was to quantify the percentage of female, elite distance athletes who have experienced disruption in training, and subsequent deterioration in fitness levels.

The data on age and mileage were collected for descriptive purposes, and thus no attempt was made to test statistically for relationships between these variables and Achilles tendon injury occurrence. However, the descriptive statistical measures obtained do not highlight any obvious differences between those subjects who have experienced Achilles tendon injury and those who have not. By using athletes in the present study who were all elite (international standard) and female, a homogeneous group has been ensured as far as possible, limiting the possibility of extraneous factors such as age, ability, training load and type of training influencing results. This is supported by the similar values attained for age, summer mileage and winter mileage for athletes reporting different responses to the Achilles tendon injury incidence questions.

(ii) Estimation of Achilles Tendon Forces

Simplifications and assumptions

The use of quasi-static moment methods to obtain ankle moment values has been investigated and justified in the literature (Morlock and Nigg, 1988). In particular, these methods are considered to be suitable for analysis of the midstance phase (Alexander and Vernon, 1975). Since peak ankle plantar-flexion moment and maximum Achilles tendon force were found to occur during the midstance phase, quasi-static moment methods were considered to be suitable for the determination of ankle moments in the present study.

It is apparent from the literature that the triceps surae muscle group is the main contributor to ankle plantar flexion (Alexander and Vernon, 1975; Scott and Winter, 1990). Figures provided in the literature for the relative contribution of this muscle group to net ankle plantar-flexion moment are variable, ranging from 65% to 100%. If a constant percentage contribution was assumed throughout ground contact, the use of differing values would have a systematic influence on maximum Achilles tendon force results. Since the methods in the present study were developed for the comparison of maximum Achilles tendon force magnitudes across conditions, and not for providing absolute Achilles tendon force values, it was felt that the assumption of the triceps surae muscle group being the sole contributor to ankle plantar-flexion was more suitable than using a percentage contribution obtained from the literature (Scott and Winter, 1990; Gregor, 1990). A consequence of this assumption will be a possible over-estimation of Achilles tendon force. The assumption in the present study that there is no antagonistic muscle action, has been justified using the EMG results of Scott and Winter (1990), and Komi (1990). This will result in the ankle moment value obtained using quasi-static methods being the minimum values possible for the equilibrium condition to be satisfied. It is noted that when comparing maximum Achilles tendon forces obtained in the present study with those presented in the literature, consideration must be made of the different assumptions used.

The assumption in the present study that ankle plantar-flexion is a two-dimensional action and thus the force in the Achilles tendon acts in the sagittal plane, is consistent with common sense and knowledge of the anatomy of the human ankle joint. In support of this assumption, Burdett (1982) found that estimated forces in the Achilles tendon during running obtained using using two-dimensional methods were similar to those obtained using three-dimensional methods. Reinschmidt and Nigg (1995) found good agreement between maximum ankle plantar-flexion moment values obtained in their three-dimensional study and two-dimensional values presented by other authors, concluding that a two-dimensional analysis is adequate for determination of ankle flexion moments during running. Additionally, these authors demonstrated that there were no significant changes in the magnitude of ankle moments in the transverse and frontal planes when comparing different heel height conditions. Fukashiro et al. (1993) found that Achilles tendon forces obtained using sagittal plane estimation methods were similar during the push-off phase of continuous jumping to those obtained by direct measurement. This finding provides further evidence that Achilles tendon force components out of the sagittal plane are negligible. It is therefore concluded that the two-dimensional assumptions adopted in the present study are justified.

Calibration

The RMSD between coordinates obtained using the linear scaling methods and the two-dimensional DLT methods was found to be greater in the horizontal (Y) direction than in the vertical (Z) direction. It is suggested that this may be the result of the camera not being positioned such that the optical axis was perpendicular to the calibration plane. The linear scaling method used the assumption that this was the case. If the camera was panned horizontally off the required orientation, then this would result in errors in the horizontal (Y) data. The two-dimensional DLT method does not use the assumption that the video image is parallel to the calibration plane, and thus results using this method will not be influenced by movement of the optical axis of the camera. To avoid the requirement of the optical axis of the camera being perpendicular to the plane of movement, all subsequent work in this study was conducted using the two-dimensional DLT technique. The DLT technique is commonly used for the reconstruction of movement data in biomechanical studies (eg. Challis, 1991).

Sampling frequency

The most appropriate sampling frequency will vary depending on the requirements of the study (Williams, 1993). In the present study, maximum ankle plantar-flexion moment and maximum Achilles tendon force were the variables of interest. Ankle moment traces obtained during ground contact in the present study were generally similar in shape to those presented in the literature. It has been demonstrated in the literature that there is normally an initial small dorsi-flexion moment at the start of the ground contact phase for rearfoot strikers, with a plantar-flexion moment occurring beyond 10% to 20% of total stance (Winter, 1983; Reinschmidt and Nigg, 1995). This initial dorsi-flexion moment was not present for some of the trials in the present study. The use of 50 Hz data, and thus one data point for every 0.02 s, will have resulted in the possibility of the dorsi-flexion moment not being detected. The variables of concern in the present study were peak ankle plantar-flexion moment and maximum Achilles tendon force, both of which have been monitored in the literature at sampling rates greater than 50 Hz, and have been found to occur during midstance, when variations over time are small (Reinschmidt and Nigg, 1995; Komi, 1990). For the determination of peak ankle plantar-flexion moment and maximum Achilles tendon force values, 50 Hz data were therefore considered to be adequate. Caution is advised when analysing 50 Hz data around the initial impact phase.

Marker locations

The most prominent area of the lateral malleolus was used to estimate the centre of rotation of the ankle joint in the sagittal plane. Investigation into the anatomy of the ankle joint has concluded that the centre of rotation is close to this point, but not actually at this location (Hicks, 1953; Isman and Inman, 1969; quoted in Scott and Winter, 1990). However, the lateral malleolus point is generally used in sagittal plane studies to approximate the centre of rotation of this joint (Morlock and Nigg, 1988; Scott and Winter, 1990). The peak ankle moment values obtained in the present study are consistent with those quoted in the literature (Winter, 1983; Scott and Winter, 1990).

Errors of up to 4.4% in peak ankle moment and 17% in maximum Achilles tendon force were found to result from a variation in ankle marker location of 0.005 m. This result supports the findings of Burdett (1982) that a 1 cm variation in the ankle joint centre location will have a significant influence on estimated force values. Differing locations of the ankle marker across studies will have a systematic influence on ankle moment and Achilles tendon force results, and thus will not influence the comparison of values for a subject within a testing session.

Achilles tendon forces

For each of the methods of Achilles tendon moment arm determination, the traces over time of Achilles tendon force were similar in shape to those presented in the literature (Burdett, 1982; Komi, 1990). However, differing maximum Achilles tendon force values were obtained for the different methods. The selection of the most appropriate method for representing Achilles tendon line of action is clearly important. When estimating Achilles tendon line of action, the importance of using Achilles tendon location data relating directly to the subject under consideration has been clearly demonstrated by Burdett (1982) and Brand et al. (1982). Burdett (1982) found, for example, that the use of data from five individual cadavers resulted in maximum ankle joint forces varying by up to 40%. In the present study, both Method 1 and Method 2 used information which was obtained directly from the individual subject, but no direct information was obtained on the line of action of the tendon for the subject. Method 3 used scaled data from an anatomical model, and thus used the assumption that scaling of anatomical data in this manner is acceptable. Method 4 involved the use of points on the Achilles tendon to approximate the Achilles tendon line of action, and thus used data directly related to the individual subject. The use of appropriate camera placement and lighting resulted in the Achilles tendon of the subject being clearly defined on the video image, highlighting the apparent suitability of the marked points for representation of the line of action of the tendon.

Methods using scaled cadaver measurements have been provided in the literature for the determination of Achilles tendon moment arm lengths (Grieve et al., 1978; Burdett, 1982). The use of these methods have not been considered in the present study, since it was felt that the development of subject specific methods for a running subject was more appropriate. The Achilles tendon moment arm lengths obtained using the methods of the present study are later compared with those obtained using the methods of other authors in Chapter 4.

Of the methods used, Method 4 was found to provide Achilles tendon force results with the lowest standard deviation between trials. Over the 10 barefoot trials, maximum Achilles tendon force results using Method 4 had a standard deviation of 0.42 BW (4% of maximum force), compared to 1.22 BW (9%), 1.40 BW (9%) and 1.18 BW (7%) for Methods 1, 2 and 3, respectively. This demonstrated that the data obtained using Method 4 were more reliable than those obtained using the other methods. Since Method 4 used Achilles tendon data on the individual subject and provided results with a relatively small variance, this method was used in subsequent work.

Number of trials

The number of trials used for each condition will influence the likelihood of Type I and Type II errors occurring, and the power of a test. Bates et al. (1983a) concluded that a minimum of eight running trials are required to obtain stable GRF data, and 10 trials are necessary to obtain results at the 95% confidence level. 10 trials were practically possible in the present study, and variation over 10 trials in the present study was found to be relatively small. Thus, the recommendations of Bates et al., (1983a) were followed, and 10 trials were used for each condition in subsequent studies.

(iii) The Influence of Shoes and Heel Lifts

Ground reaction force

The focus of the present study was on the estimation of maximum Achilles tendon force, which has been found to occur in the middle of the stance phase. Analysis has therefore been limited predominantly to this phase of stance, with the impact phase not being examined in detail. However, graphical presentation of vertical GRF data has clearly illustrated differences in the magnitude and time of occurrence of the impact peak for the heel lift and running shoe conditions compared with the barefoot condition. This finding supports those published in the literature (Dickinson et al., 1985; Snel et al., 1985). The GRF data were collected at 1000 Hz and normalised to produce 100 points over a ground contact of approximately 0.2 s, resulting in data close to 500 Hz. The clear illustration of shoe and heel lift conditions influencing the impact phase indicates that, additional to the analysis of the middle of the stance phase, monitoring of the loading of the Achilles tendon during the impact phase may be necessary in future work. Examination of the 50 Hz GRF data employed for synchronisation with the kinematic data has revealed that the GRF impact peak is not detectable, demonstrating that an increased sampling frequency is required for detailed analysis of this phase.

The reduction in vertical GRF impact peak and peak loading rate with shock absorbing heel lifts for a rearfoot striker is consistent with the findings of Lees and McCullagh (1984), and supports the suggestion that the reduction in Achilles tendon injury demonstrated when using viscoelastic heel lifts is due to a reduction in the peak acceleration at heel strike resulting from increased shock absorption (MacLellan and Vyvyan, 1981). The observed reduction in GRF variables is in contrast with the findings of Nigg et al. (1988), who found no change in GRF variables when shock absorbing insoles were placed in running shoes. Additionally, several studies have found that running shoes constructed from materials providing increased shock absorbency have not reduced the magnitude of the GRF impact peak (Nigg et al., 1987; Kaelin et al., 1985). Possible kinematic adaptations to inadequate shock absorption have been suggested to account for the absence of differences in GRF measurements (Clarke et al., 1983a). Differences in response across subjects, or the use of different conditions, may account for contrasting GRF results across studies (Lees and McCullagh, 1984). Kinematic analysis in future work is suggested, to allow investigation into the possible kinematic adaptation of an individual to interventions. It must be noted that analysis of the impact phase in the present study has been limited to visual assessment, and thus confidence in observations has not been quantified. More detailed analyses of GRF during the impact phase are required in future work.

The investigations in which a decrease in lower extremity accelerations have been demonstrated using heel inserts have generally been walking studies where heel lifts have been introduced in shoes made of a relatively hard material (Light et al., 1980; Bojsen-Moller, 1983). In a running study with shoes of midsole hardness spanning the range of available sports shoes, Clarke et al. (1983) found that there were no significant differences in tibial acceleration across conditions. It is suggested that the optimum amount of shock absorption for a running shoe is within the range available on the market, with differences between shoes of different material properties being too small to be detected. In contrast, the conventional shoes available on the market for everyday use do not provide adequate shock absorption for walking, and thus the additional shock absorption provided by the introduction of heel inserts can be detected. Using the same argument with reference to the results of the present study, the finding that maximum Achilles tendon force was not reduced when the heel lifts were placed in a running shoe, but were reduced when the same lifts were attached to the barefoot, may have been due to the shoes providing adequate shock absorption.

In the present study, any significant differences in maximum Achilles tendon force between conditions were found to be a result of changes in the geometry of the system due to raising of the heel, rather than changes in GRF magnitude. This result could be interpreted to indicate that differences detected in Achilles tendon force across conditions were the result of the geometric, rather than shock absorbing, properties of the heel lifts. However, the differences found in geometry during midstance may occur in response to the impact force variations occurring during the impact phase. To investigate further the relative influence of the material properties and the geometric properties of the heel lifts on Achilles tendon loading, a study is required in which only one of these design characteristics is varied.

Ankle moment

The ankle moment magnitudes and traces obtained in the present study were similar to those presented in the literature (Winter, 1983; Scott and Winter, 1990). The significant reduction in ankle moment for the running shoe condition compared to barefoot running is consistent with the results of Reinschmidt and Nigg (1995). The finding that the raising of the heel in the shoe did not reduce maximum ankle plantar-flexion moment is in agreement with the results Reinschmidt and Nigg (1995), who found no significant difference in ankle moment in the sagittal plane when comparing shoes of different heel heights. Heel heights varying from 2.1cm to 3.3cm were used in the study by Reinschmidt and Nigg (1995). In the present study, a shoe with heel height approximately 2.5 cm was used. The heel lifts were each approximately 0.6 cm thick. Thus, heel heights of approximately 2.6 cm to 3.7 cm were used in the present study, heights similar to those used by Reinschmidt and Nigg (1995). Despite no differences being observed in ankle moment, the subject reported an obvious difference in feeling for the running shoe with heel lift conditions, compared with the running shoe only condition.

The variation of 0.5 cm used to assess the influence of ankle marker placement on results was chosen subjectively, and thus the possibility of a variation in ankle marker location of a greater amount cannot be neglected. Analysis of the level of repeatability in ankle marker placement on a subject is limited by difficulties in guaranteeing a stationary position of the subject throughout recording of repeated marker location. An alternative approach is suggested for future work, where rather than testing the level of variation of marker location, the subsequent influence of marker variation on ankle moments and Achilles tendon force values is quantified.

Achilles tendon force

To optimise the ability of the methods used in the present study to confidently detect differences between conditions, the reliability or level of precision should be maximised. In this initial study it has been found that random errors in digitisation have an influence of approximately 0.6 BW on maximum Achilles tendon force values. This figure is close to the effect size observed between some of the conditions. Errors resulting from confidence in GRF data have not been considered. For the comparison of the barefoot and running shoe conditions in the present study, a significant difference in ankle moment was found, but no difference was detected in the value of the maximum Achilles tendon force. This discrepancy was the result of a difference of 2.2 mm in the Achilles tendon moment arm length between the conditions. With the level of precision in each Achilles tendon moment arm measurement being 1.2 mm, a difference of up to 2.4 mm may occur between conditions due to random variation alone. Increased reliability in the identification of body landmarks is required to improve the level of precision attained in Achilles tendon moment arm length. In future work, it is suggested that methods for improving reliability in digitisation of markers are investigated. Additionally, it is suggested that the level of precision of GRF data is included in the calculation of overall precision in estimated Achilles tendon forces.

Difficulties in the evaluation of estimated internal force measurements due to the necessary assumptions have been highlighted by Nigg and Bobbert (1990). Fukashiro et al., (1993) evaluated methods similar to those of the present study by direct measurement of Achilles tendon forces. These procedures were considered not to be practical or ethical for use in the present study. Additionally, the calibration methods used by Fukashiro et al., (1993) were limited by the inability to directly calibrate the transducer, as can be done in animal studies (Gregor et al., 1988). It is of interest that the magnitudes of maximum Achilles tendon force obtained in the present study are within the range presented in the literature (Burdett, 1982; Scott and Winter, 1990; Komi, 1990). However, evaluation of methods by direct comparison of results with those published in the literature is limited due to differences in assumptions and methodology employed. In the present study, the assumption that the triceps surae group contributes solely to ankle plantar-flexion has been justified. For comparison of results with those obtained in other studies it is important to adjust results accordingly. Differences in the definition of the ankle joint centre and in determination of the Achilles tendon line of action will also influence comparisons between studies. Marker variation will have a systematic influence on results, and thus will not influence comparisons between conditions for a single subject. Of paramount concern is the influence of random digitisation error on estimated Achilles tendon force values, since random errors influence the confidence with which comparisons can be made across conditions. In future work it is therefore suggested that a detailed sensitivity analysis is employed to quantify the possible error in maximum Achilles tendon force values resulting from predicted variations in marker locations.

The use of Method 4 to calculate Achilles tendon forces across all conditions was supported by the finding that this method provided the most reliable results under all conditions of the present study. This supports the future use of this method for the estimation of Achilles tendon forces in running.

Differences in maximum Achilles tendon forces have been highlighted across conditions in the present study. Where differences in maximum Achilles tendon force were not detected, interventions were found to have a significant influence on other variables such as maximum ankle moment or GRF variables. Heel lift intervention has been found to influence maximum Achilles tendon force, maximum ankle moment, GRF moment arm, and Achilles tendon force moment arm values. It is not clear from the findings of the present study which specific joint angles and GRF variables contribute to the observed differences in moment arm values across conditions. To understand more clearly the influence of heel lift intervention on running kinematics, it is suggested that in future work monitoring of joint angles across conditions is included.

The suggestion in the literature that the reduced Achilles tendon injury following heel lift intervention is the result of a reduction in the maximum Achilles tendon force was not supported by the findings of the present study. No differences in maximum Achilles tendon force were observed between the three running shoe conditions with differing amounts of heel lift. These results were for a single subject, and thus cannot be used to infer behaviour for the remainder of the running population. However, it clearly cannot be assumed that the placement of commercially available heel lifts in a running shoe will result in a decrease in the maximum Achilles tendon force.

The shock absorbing heel lifts employed in the present study provided increased shock absorption in the rear of the shoe, and raised the heel in relation to the forefoot. Both of these interventions have been associated with a reduction in the incidence of Achilles tendon injury. Although it was observed in the present study that the magnitude and time of occurrence of maximum impact force were influenced by the heel lift interventions, it was found that during this period of stance the Achilles tendon was not loaded. Vertical impact force variations therefore appear unlikely to indicate changes in Achilles tendon loading.

The raising of the rear of the foot in relation to the forefoot has been suggested to reduce the maximum amount of ankle dorsi-flexion during the midstance phase, and thus the maximum amount of stretch of the Achilles tendon (Clement et al., 1984). The findings in this initial study highlight the significance of changes in geometry during midstance, with these apparently contributing to changes in maximum Achilles tendon force values. From the results obtained in the present study, it therefore appears that the effectiveness of shock absorbing heel lifts in the treatment of Achilles tenon injury is the result of the raising of the heel in relation to the forefoot.

The possibility that changes during impact resulting from increased shock absorption properties of the heel lifts cause subsequent differences in geometry during midstance cannot be eliminated. To investigate further the relative contributions of shock absorbency and geometric effects of heel lifts a study is required in which only one of these interventions is applied. Since the results of the present study indicate that geometric effects are influential, it is suggested that in future work the amount of heel lift provided is manipulated, with the shock absorption provided being maintained at a constant level. An increased understanding of the mechanism by which heel lifts are effective is likely to be of interest to clinicians, such as chiropodists and podiatrists, who often prescribe shock absorbing heel lifts for patients or athletes with Achilles tendon pain. Knowledge as to whether the emphasis should be on increased shock absorption, increased heel lift, or a combination of these factors should help in providing the most effective treatment.

It has been assumed in the present study that the magnitude of the maximum Achilles tendon force is the factor influencing injury occurrence. This is consistent with suggestions presented in the literature. Clement et al., (1984) recommended the use of heel lifts to relieve the tension in the Achilles tendon. These authors also described the increased flexibility and force in the Achilles tendon when there is inadequate heel wedging in the shoe. Clement et al., (1984) appear to have used the terms 'strain' and 'tension / force' interchangeably, and thus it is not clear what exactly their rationale in prescribing heel lifts was. It is not unreasonable to use these terms interchangeably if a change in tension (force) is proportional to a change in strain. If tendon obeys Hooke's Law then this is true. Ker et al., (1988) demonstrated that the region of the stress-strain curve corresponding to typical strains experienced by the Achilles tendon during running was linear, indicating that a change in stress indicates a proportional change in strain, and visa-versa. Even if the relationship is non linear, stress-strain curves indicate that an increase in strain results in a corresponding increase in stress. The rationale for prescribing heel lifts may therefore be to decrease both stress and strain. It is generally accepted that there is little change in the cross-sectional area of the Achilles tendon during running (Abrahams, 1967), and thus a decrease in stress corresponds to a decrease in tension or force. The minimising of ankle dorsi-flexion, and thus the stretching of the muscle-tendon complex, and the decreasing of maximum Achilles tendon force, therefore amount to the same thing. The inclusion of joint angle measurements in future work, as previously suggested, will allow further investigation into the relationship between Achilles tendon stress and Achilles tendon strain.

A single subject demonstrating a rearfoot strike running style has been used in the present study. It cannot be assumed that the results obtained for this subject are typical for all runners. The use of a single subject approach has been supported by the findings of several authors that individual responses are often overlooked when results are grouped for analysis (Bates et al., 1992; Reinschmidt and Nigg, 1995). The use of a single subject approach when performing initial investigations has also been suggested (Reboussin and Morgan, 1996). Running style has been found to influence the response of an individual to footwear interventions (Therrien et al., 1982; Lees and M^{c}Cullagh, 1984; Komi, 1990; Reinschmidt and Nigg, 1995). Since running style is likely to influence the response of an individual to heel lift intervention, it is suggested that in future work subjects with differing running styles are employed. It is proposed that subjects demonstrating rearfoot strike, midfoot strike, and forefoot strike, as described by Cavanagh and Lafortune (1980) are chosen.

For the comparison of magnitudes of maximum Achilles tendon force in the present study, a power analysis has demonstrated powers of 0.99 and 0.95 for significance levels of 0.01 and 0.05, respectively. Using the assumption that the effect size (difference in sample means) and the standard deviation values obtained in the present study were typical, the results of this power analysis were used to determine the most appropriate sample size and significance level for future studies. The probability of a Type I Error occurring is controlled directly by the definition of the significance level. The choice of significance level is made based on the available sample size, variability, desired effect size, and the relative importance of Type I and Type II Errors. As the probability of a Type II Error occurring is reduced, the probability of a Type I Error occurring will be increased. Summarising the results of the present power analysis, a probability of a Type II Error occurring of 5% existed for a Type I probability of 1% ($p<0.01$), and a probability of a Type II Error occurring of 1% occurred for a Type I probability of 5% ($p<0.05$). Franks and Huck (1986) suggested that for exploratory studies, as this initial study can be described, Type II Errors should be minimised, and therefore power maximised. Thus, if all other factors remain unchanged, a significance level of 0.05 is recommended for future similar studies.

In the preceding paragraph, an argument has been presented for the use of a significance level of 0.05, based on the assumption that only significance level has been changed. It is possible that other factors influencing power can be varied. In the present study, the most easily influenced of these factors was the number of trials. Using the classifications of Cohen (1990), large power values were obtained for both of the significance levels considered in the present study. However, it was apparent that a reduction in sample size (the number of trials per condition) had a more marked influence when the smaller significance level of 0.01 was employed. For this significance level, a reduction of trials from 10 to 8 was shown to result in a reduction in power from 0.95 to 0.85. The same reduction in trials at a significance level of 0.05 reduced power from 0.99 to 0.96, thus maintaining similar probabilities for the occurrence of a Type II Error and a Type I Error. Two conclusions are made from the investigation of the influence of number of trials on power value. Firstly, the use of 10 trials per condition within the experimental design of the present study is appropriate for attaining acceptable power at both of the significance levels investigated. Secondly, the reduction in power if any trials are discarded or lost is less marked for the 0.05 significance level.

The use of 10 trials per condition has also been suggested by Bates et al., (1983a) based on the number of running trials required to obtain stable GRF data. This figure was increased in a future study to 25 trials per condition (DeVita and Bates, 1988). Yeadon and Challis (1994) have, however, outlined inconsistencies in this more recent study. Bates (1989) described the importance of considering each study individually for optimum experimental design. For future similar studies, the use of 10 trials per condition and the use of a significance level of 0.05 are recommended.

(iv) Response to Questions

In response to the original question, it is apparent from the results of the present study that the magnitude of maximum Achilles tendon force can be influenced by the wearing of running shoes, or by the attachment of lifts to the rear of the foot. This finding is for a single subject, and thus no generalisations can be made to the remainder of the running population. However, by demonstrating that maximum Achilles tendon force magnitude can be influenced by footwear manipulations for the subject in the present study, further investigation into the influence of heel lift intervention has been stimulated. If the maximum Achilles tendon force can be influenced by heel lift intervention, then more detailed study of the mechanism of the response, and the study of the response of other runners is justified.

The attachment of 12 mm heel lifts to the plantar surface of each foot was found to significantly reduce the maximum force transmitted by the Achilles tendon ($p<0.001$). Knowledge of the mechanism by which changes in maximum Achilles tendon force occur will further the understanding of how external factors such as shoes and heel lifts may help to reduce the occurrence of injury. Achilles tendon force has been defined in the present study as the ankle moment divided by the moment arm of the Achilles tendon about the ankle joint centre. Thus, a change in either the ankle moment or the length of the moment arm, or both, could result in a change in the estimated tendon force. The introduction of heel lifts resulted in a reduction in the maximum ankle moment, occurring at approximately the same percentage of stance time as the maximum Achilles tendon force. Using quasi-static methods, ankle moment was calculated as the product of the magnitude of the resultant GRF and the length of the moment arm of this force about the ankle joint centre. Thus, any difference in ankle moment was due to changes in either or both of these variables. It was found that the moment arm of the GRF was decreased significantly by the attachment of heel lifts. The magnitude of the GRF when maximum ankle moment occurred was found to be the same under the two conditions. Thus, the significant difference in maximum ankle moment between the two conditions was demonstrated to be entirely due to the change in geometry of the system caused by the heel lifts. Maximum Achilles tendon force was estimated by the division of ankle moment by the moment arm of the Achilles tendon. Compared to the barefoot condition, the Achilles tendon moment arm at maximum Achilles tendon force was increased significantly by the introduction of heel lifts. Thus, the decreased Achilles tendon force for the heel lift conditions compared with the barefoot condition was demonstrated to be due to the combined effect of decreased GRF moment arm and increased Achilles tendon moment arm.

The placement of heel lifts in a running shoe was found to have no significant influence on maximum Achilles tendon force or maximum ankle moment. These findings are consistent with those of Reinschmidt and Nigg (1995) who found that maximum ankle plantar-flexion moment was not influenced by an increase in heel height. Reinschmidt and Nigg (1995) suggested several possibilities for the inability of lifting the heel to produce a detectable change in maximum Achilles tendon force. The suggested possibility of ankle plantar-flexion moment not necessarily representing changes in Achilles tendon force has been tested in the present study The limitations of using ankle moment alone to indicate changes in Achilles tendon force has been highlighted. The possibility of heel lifts being effective in the treatment of Achilles tendon force by mechanisms other than reducing maximum Achilles tendon force was also noted by Reinschmidt and Nigg (1995). Differences in the amount of calcaneal friction, or in the amount of rearfoot movement with subsequent changes in shear or bending forces were suggested.

Contrary to the suggestions of Reinschmidt and Nigg (1995), it has been demonstrated in the present study that changes in ankle moment may not directly reflect changes in Achilles tendon force. A significant reduction in ankle moment was found when wearing running shoes compared with the barefoot condition, but no subsequent reduction in maximum Achilles tendon force was detected. The reduction in ankle moment was found to be due to a reduction in the GRF moment arm, with no significant difference demonstrated in the magnitude of the GRF at the time when maximum ankle moment occurred. At the time of maximum Achilles tendon force occurrence, a decrease was demonstrated in the moment arm of the Achilles tendon. The combined effect of the decrease in ankle moment and in Achilles tendon moment arm was to cause no significant difference in maximum Achilles tendon force between conditions. Caution is therefore advised in using differences in peak ankle plantar-flexion moment to infer changes in maximum Achilles tendon force.

CHAPTER 4 THE INFLUENCE OF HEEL LIFT ON PEAK ACHILLES TENDON FORCE AND ANKLE DORSI-FLEXION

4.1 Introduction

In Chapter 3 it was found that maximum Achilles tendon forces were not influenced significantly by the placement of shock absorbing heel lifts in a conventional running shoe. However, the regular prescription of heel lifts by practitioners (Subotnick, 1979), together with the epidemiological evidence in the literature (MacLellan and Vyvyan, 1981; Fauno et al., 1993), supports the use of heel lifts in the treatment of Achilles tendon injury. It was demonstrated in the Chapter 3 study that maximum Achilles tendon force in running can be reduced by the attachment of shock absorbing heel lifts to the plantar surface of the foot. It was not possible to conclude whether the reduction in force was the result of increased shock absorption, increased heel lift, or a combination of these factors.

Both the traditional approach of employing running shoes with differing design characteristics, and the approach of creating a controlled setting under which systematic manipulation of one variable is facilitated, were utilised in the study of Chapter 3. Due predominantly to the difficulties encountered when attempting to vary only one design characteristic of a running shoe, the alternative approach of attaching material to the plantar surface of the foot has been adopted in the study of the present chapter. It has been noted that a limitation of the use of shock absorbing heel lifts of Chapter 3 was the fact that both the amount of shock absorption, and the degree of heel lift were changed. In an attempt to manipulate heel height alone, lifts constructed from a firm material are used in the study presented in the present chapter. An additional factor influencing the decision to use barefoot conditions for the remainder of the study, rather than running shoe conditions, was the finding in Chapter 3 that maximum Achilles tendon force was not influenced by the placement of heel lifts in the rear of the shoe.

The use of a single subject approach allows the detailed study of the behaviour of distinct subjects. The likelihood of the method of ground contact adopted by a runner influencing the response to footwear interventions has been discussed in Chapter 3. Three ground contact possibilities have been identified as rearfoot, midfoot and forefoot strike. To build on the results of the Chapter 3 study, three subjects demonstrating distinctly different ground contact styles are employed in the present chapter. The possibility of these subjects responding differently to heel lift interventions is investigated.

Nigg and Bobbert (1990) highlighted the fact that to justify the use of the 'comparison technique', sources of error influencing estimated forces must be demonstrated to be systematic. To investigate the suitability of this technique in the present study, a detailed analysis of random errors is performed. Following the identification of error sources and magnitudes, it is necessary to quantify their possible influence on Achilles tendon results. A sensitivity analysis is employed in the study of the present chapter to investigate the reliability of estimated Achilles tendon force values.

For the calculation of ankle moments, quasi-static moment methods have been used in Chapter 3. Although the error in maximum ankle plantar-flexion moment when using quasi-static methods has been found to be small (Morlock and Nigg, 1988), and is likely to be systematic across conditions, effort should be made to minimise all errors. Thus, both quasi-static and inverse dynamic procedures are utilised for the calculation of ankle moments in the present chapter, allowing quantification of the magnitude of the difference between the two methods.

It has been demonstrated in the literature that sagittal plane kinematic variables may be influenced by the wearing of different footwear (Frederick, 1986). A sagittal plane Achilles tendon angle is employed in the present chapter to represent changes in ankle angle. Changes in this angle with heel lift manipulation are monitored. The data of Burdett (1982) and Bobbert et al. (1986) are used to obtain Achilles tendon moment arm lengths for the joint angles measured in this chapter. The resulting relationships between Achilles tendon moment arm and ankle angle are compared to those obtained using the methods of the present study.

A problem of precision was identified in Chapter 3. It was found that the random error occurring in the digitisation of marker points resulted in the level of precision in estimated maximum Achilles tendon force being inadequate for the confident detection of differences between conditions when individual trials were compared. Effort is therefore concentrated on maximising the measurement precision of the Achilles tendon moment arm length in the present chapter, to improve the level of precision attained in Achilles tendon forces.

It is possible that time dependent factors such as familiarisation, boredom and fatigue influenced the findings of Chapter 3, since for each condition the 10 trials used for analysis were performed consecutively. A randomised block design is used in the present chapter to eliminate the possibility of maturation effects influencing results.

The main questions addressed in the present Chapter are:

- Can raising the heel with firm lifts reduce maximum Achilles tendon force ?
- Do runners with distinctly different styles differ in their response to heel lift ?

Additional questions are:

- Is the maximum amount of ankle dorsi-flexion reduced by the use of heel lifts ?
- Is the level of measurement precision attained in this study sufficient for confident detection of differences in maximum Achilles tendon forces across conditions ?

4.2 Methods

Three elite female middle distance runners of similar mass (54.7±0.5 kg), performed barefoot running trials at seven minute mile pace (3.8 $m.s^{-1}$± 5%). The subjects were selected due to the three distinctly different shod running styles they exhibited. Preliminary running trials without shoes verified that the runners adopted the same ground contact styles when running barefooted as when wearing shoes. Description of running styles was initially based on field by field visual assessment of the sagittal plane video recordings of ground contact, and was supported by characteristic vertical GRF traces as described by Cavanagh and Lafortune (1980). Subject A demonstrated a rearfoot ground contact, Subject B contacted the ground with a flat foot, and was thus referred to as a midfoot striker, and Subject C demonstrated a forefoot ground contact, with the rearfoot not contacting the ground at any stage of stance. Each subject performed 10 trials under each of three conditions: barefoot (B); barefoot with heel lifts of 7.5 mm (B1); barefoot with heel lifts of 15 mm (B2). A randomised block design was utilised. 10 blocks of the three conditions were used, with random assignment of the conditions within each block. Heel lifts were made from a high density Ethyl Vinyl Acetate (EVA) material, tapered along a length of 88 mm (Frank Lord Foot Appliance Materials). The apparent hardness, measured by SATRA (Footwear Technology Centre, UK), was 65 IRHD (International Rubber Hardness Degrees; BS 903 / ISO 48). The peak deceleration when the material was tested using a standard drop test (SATRA) was 23.8 g. The heel lifts were attached to the plantar surface of the foot using light, micropore surgical tape. The tape was attached to the skin covering the calcaneus, and did not cross the ankle joint, thus minimising the possibility of tape attachment influencing running mechanics.

The collection of kinematic and force data was performed using the same equipment and procedures as described in Chapter 3. The field of view was reduced from 1.8 m to 0.88 m, providing an increased foot image size. Synchronisation, calibration and DLT reconstruction procedures were as described in Chapter 3. For each video field during ground contact, four reference points, a point on the most prominent point of the lateral malleolus representing the ankle joint centre, a point representing the 5th metatarsalphalangeal (MTP) joint centre, and two points on the skin surface over the Achilles tendon were digitised. As previously, body locations were identified by circular markers of approximately 1 cm diameter drawn on the skin using a black felt pen. Five fields encompassing the midstance phase were each digitised four times, and the mean Achilles tendon force calculated over the four repeated digitisations. Detailed analyses were limited to this phase because maximum Achilles tendon force and maximum dorsi-flexion of the ankle joint occurred during this time. A 24 bit colour video frame store (Millipede Apex Imager) with 768 x 576 pixel image resolution, and software generated sub-pixel cursor movement (Target), provided a measuring resolution of 12288 x 9216. The precision of digitised locations attained in the present study, and subsequent influence on maximum Achilles tendon force reliability, were compared with those attained in Chapter 3, in which a measurement resolution of 768 x 576

(Prisma system) and a reduced foot image size were used. To establish the relative contribution of measurement resolution and foot image size on any observed differences, selected trials from the present study (increased foot image size) were also digitised using the Prisma system, and resulting levels of precision were compared.

Inverse dynamics and quasi-static moment calculations were used to provide two estimates of sagittal plane ankle joint moment (M_a). Procedures for inverse dynamics estimation of ankle moment are provided in Appendix F. For these inverse dynamics calculations, foot weight and centre of mass were estimated using the data presented by Clauser et al. (1969), and moment of inertia from the data of Whitsett (1963). The peak ankle plantar-flexion moment values obtained using the two methods were compared. The method developed in Chapter 3 was used for the estimation of maximum Achilles tendon forces.

The foot segment was represented by a straight line joining the ankle and MTP markers. The angle which the Achilles tendon line of action made with the vertical (α), and the angle which the foot made with the horizontal (β) were measured throughout stance. The angle between the Achilles tendon line of action and the foot was defined as the Achilles tendon angle (θ), and was measured throughout the stance phase (Figure 4.1).

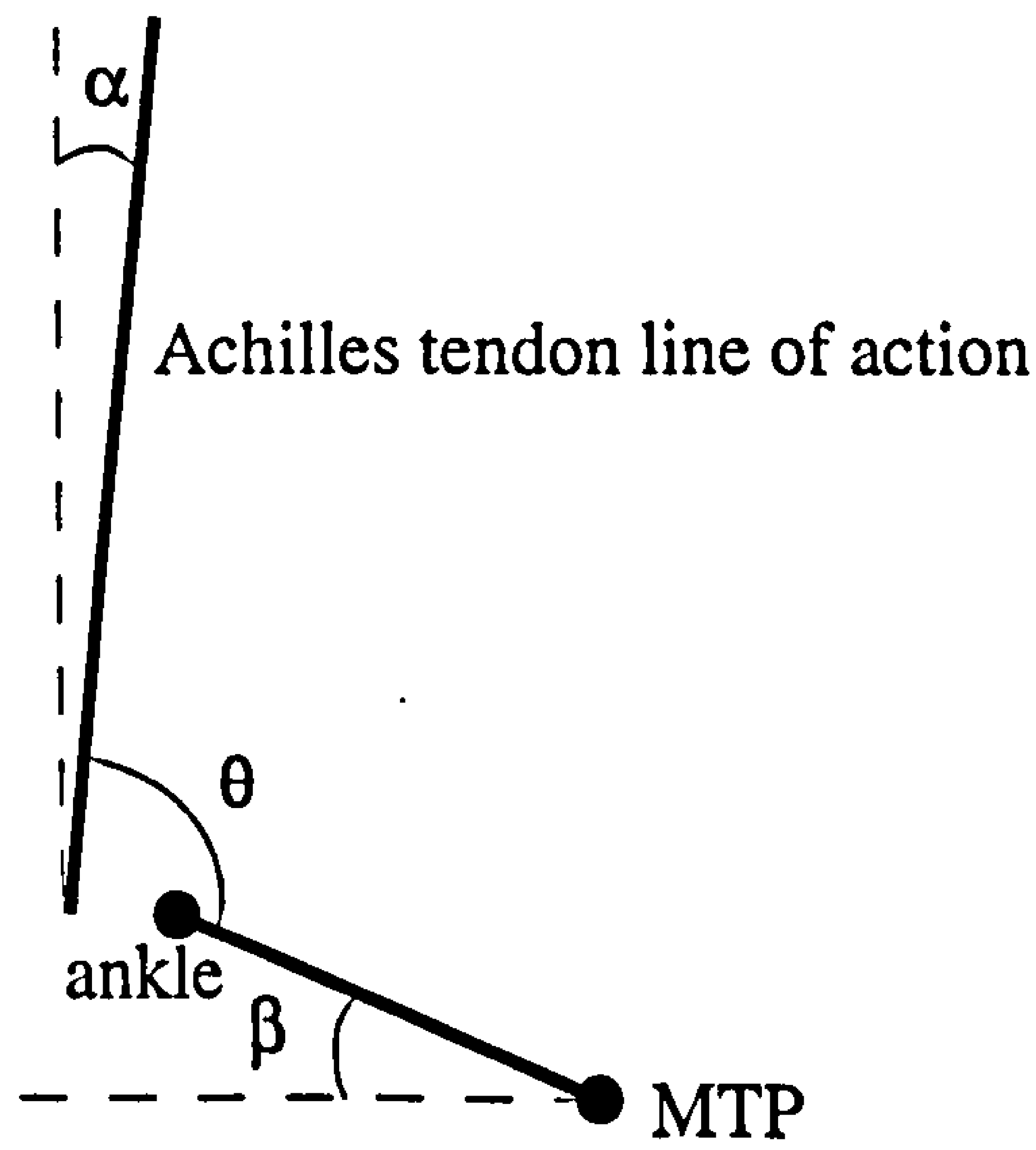


Figure 4.1 Illustration of angles used for calculation of Achilles tendon angle (θ)

The influence on estimated moment arms of using ankle angle-Achilles tendon moment arm relationships available in the literature was investigated for Subject A. There would be a difference between ankle angles obtained in the present study and those obtained using the literature convention of representing the foot segment as a straight line from the heel to the MTP joint (Milliron and Cavanagh, 1990). To account for this for input of ankle angles into the relationships presented by Burdett (1982) and Bobbert et al. (1986), the systematic difference between ankle angles was quantified by taking anatomical measurements for the standing subject. The line of action of the Achilles tendon, estimated from digitisation, was used to represent the lower leg, justified by reference to Burdett (1982). The cadaver data provided by Burdett (1982) were scaled using foot length, distance between medial and lateral malleolus, and lateral malleolus height for Subject A. The methods of Bobbert et al. (1986), using the equations on gastrocnemius length supplied by Grieve et al. (1978), were used for the estimation of Achilles tendon moment arm as a function of ankle angle, using leg length of Subject A.

The degree of random digitising error, or noise, influencing the maximum Achilles tendon forces was assessed using 10 randomly selected trials for Subject A. For each trial, the root mean square (RMS) variation in marker coordinates was calculated over ten repeated digitisations of the video field corresponding with maximum Achilles tendon force. The subsequent influences of this variation on estimated ankle moment, Achilles tendon moment arm and maximum Achilles tendon force were calculated. The influence on precision of utilising different numbers of reference points as wobble points was investigated. Precision attained using zero, two and four wobble points was compared.

Accuracy of the digitisation process was assessed as described in Chapter 3. The Target software employed in this study provided the additional, previously unavailable feature of selection from a choice of cursors for digitisation. The influence on accuracy and precision of using different cursor designs was investigated by quantification of RMSD between known points and their digitised coordinates, and comparison of values attained for two different cursor designs. The chosen cursors are illustrated in Figure 4.2.

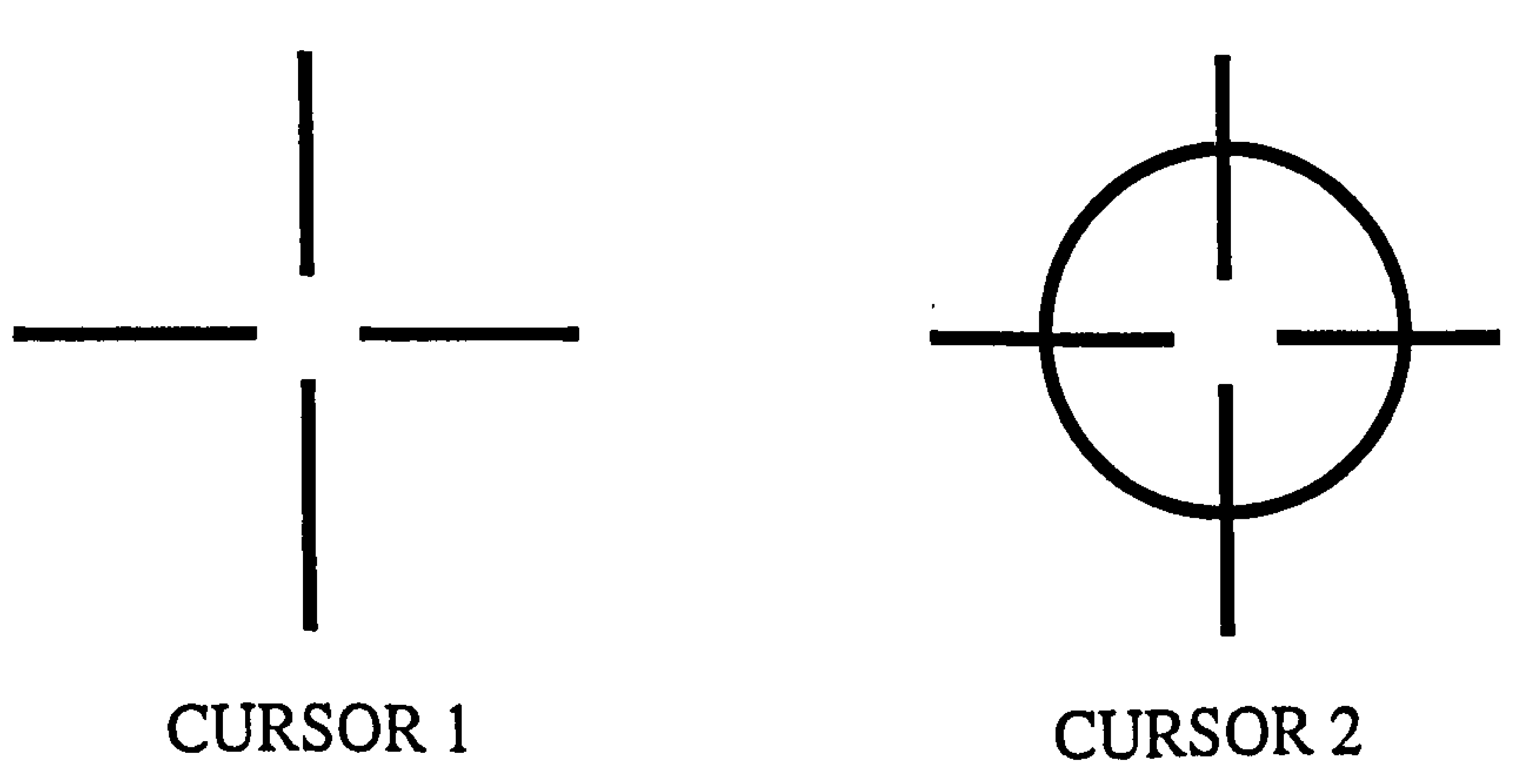


Figure 4.2 The two cursors used for comparison of cursors

The variation in maximum Achilles tendon force due to different marker placement was assessed by comparing values obtained over three separate testing sessions for a single subject. For each testing session, Subject A performed barefoot running trials and new Achilles tendon and ankle markers were used.

For the evaluation of methods, a sensitivity analysis was used to assess the influence on estimated maximum Achilles tendon forces of measurement errors. A hierarchical model (Figure 4.3) was used to highlight the factors influencing the estimated maximum Achilles tendon forces. Ultimately, the digitised marker locations, the horizontal and vertical force plate components and the centre of pressure location determined the maximum Achilles tendon force values obtained. The influence of marker digitisation error was assessed by varying marker locations by magnitudes which could reasonably be expected during digitisation, as represented by the calculated RMS variation values when repeated digitisations were performed. It was necessary to simultaneously vary the ankle marker coordinates, and the two Achilles tendon marker coordinates to quantify the worst possible propagated error in maximum Achilles tendon force due to random variation in the digitising process. Figure 4.4 shows the geometry of the system for a typical trial at the time of maximum Achilles tendon force. The effect of variations in each marker location in the positive and negative horizontal and vertical directions were first investigated to identify the variations which had the most influence on estimated maximum Achilles tendon forces. These variations were combined in the sensitivity analysis to calculate the maximum variation in ankle moment, Achilles tendon moment arm length and maximum Achilles tendon force. Variations in force plate data were chosen using the manufacturer's specifications. The influence on estimated Achilles tendon forces of a variation of 2 mm in centre of pressure location, and 1% in force magnitude, were investigated.

To assess the influence of possible synchronisation error when using 50 Hz video data and 1000 Hz force plate data, the influence of ±0.01 s difference in the selected force plate data was investigated. For a typical running trial, force plate data from the time 0.01 s before and 0.01 s after the identified occurrence of maximum Achilles tendon force were combined with a consistent set of kinematic data, providing figures on the maximum variation in maximum Achilles tendon force due to synchronisation error.

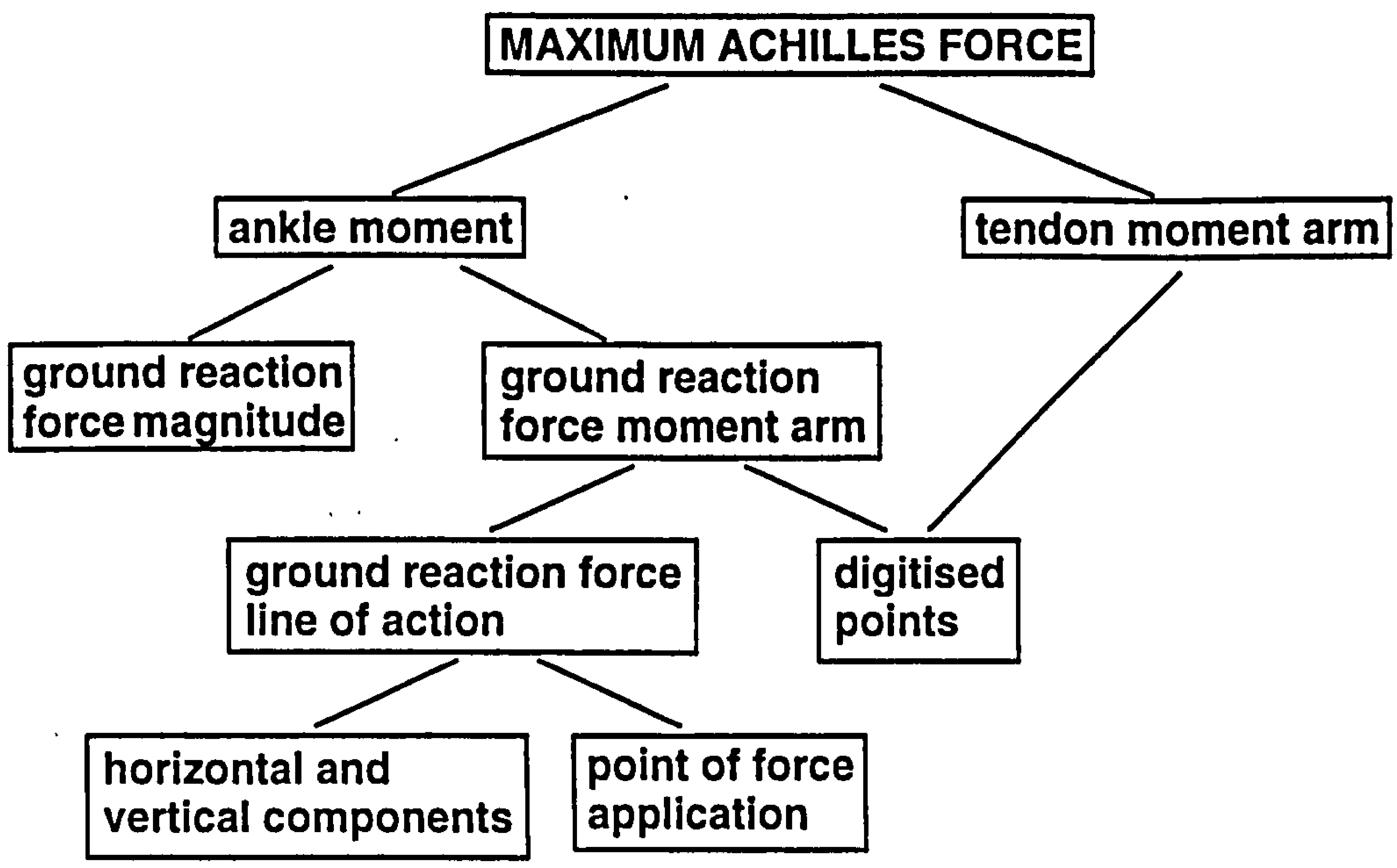


Figure 4.3 A hierarchical model illustrating the factors influencing Achilles tendon force

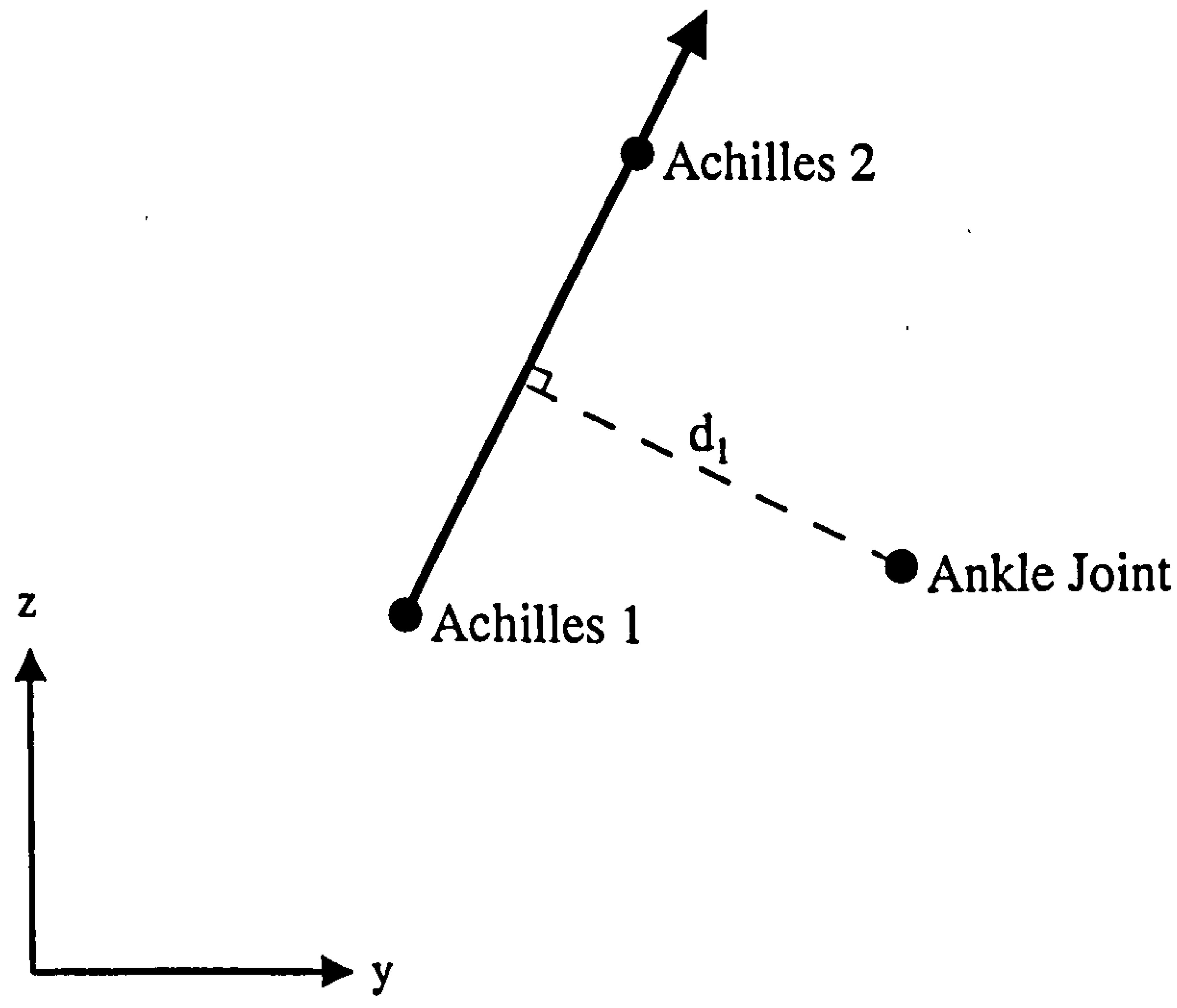


Figure 4.4 Typical geometry of the system at the time of maximum Achilles tendon force

Mean values for maximum ankle moment (M_a), maximum Achilles tendon force (F) and minimum Achilles tendon angle (θ) were obtained over the 10 trials for each condition and each subject, using the 50 Hz data. Data for each subject obtained under the three different conditions were compared using analysis of variance with repeated measures ($p<0.05$). The significance level of 0.05 was chosen using the findings Chapter 3. A Tukey test was used to identify differences between mean values for the three conditions. Maximum values obtained using interpolated data to 100 points over stance time were also calculated, and were compared to those obtained using 50 Hz data mean values.

Where significant differences in maximum Achilles tendon forces were detected between conditions, the variables which had contributed to these changes were obtained for the time of maximum Achilles tendon force occurrence, by reference to the hierarchical model (Figure 4.3). This allowed investigation into the factors most influential in the determination of the maximum Achilles tendon force magnitudes. For comparisons of maximum Achilles tendon force across conditions, statistical power values attained in the present study were calculated for each subject.

4.3 Results

(i) General Observations

For all subjects, maximum ankle plantar-flexion moment, maximum Achilles tendon force and minimum Achilles tendon angle were found to be coincident. This is illustrated for Subject A, the rearfoot striker, using mean interpolated plots over 10 trials for each condition of ankle moment, Achilles tendon force, Achilles tendon angle, and foot angle (Figure 4.5). The shapes of the ankle plantar-flexion moment and Achilles tendon force curves were similar for all subjects. Mean values for peak ankle plantar-flexion moment and maximum Achilles tendon force for each subject and each of the three conditions are presented in Table 4.1.

In Table 4.2, the peak values and calculated times of occurrence attained using interpolated data are compared to values calculated at 50 Hz, for Subject A. Similar values were obtained for both methods for each of the variables.

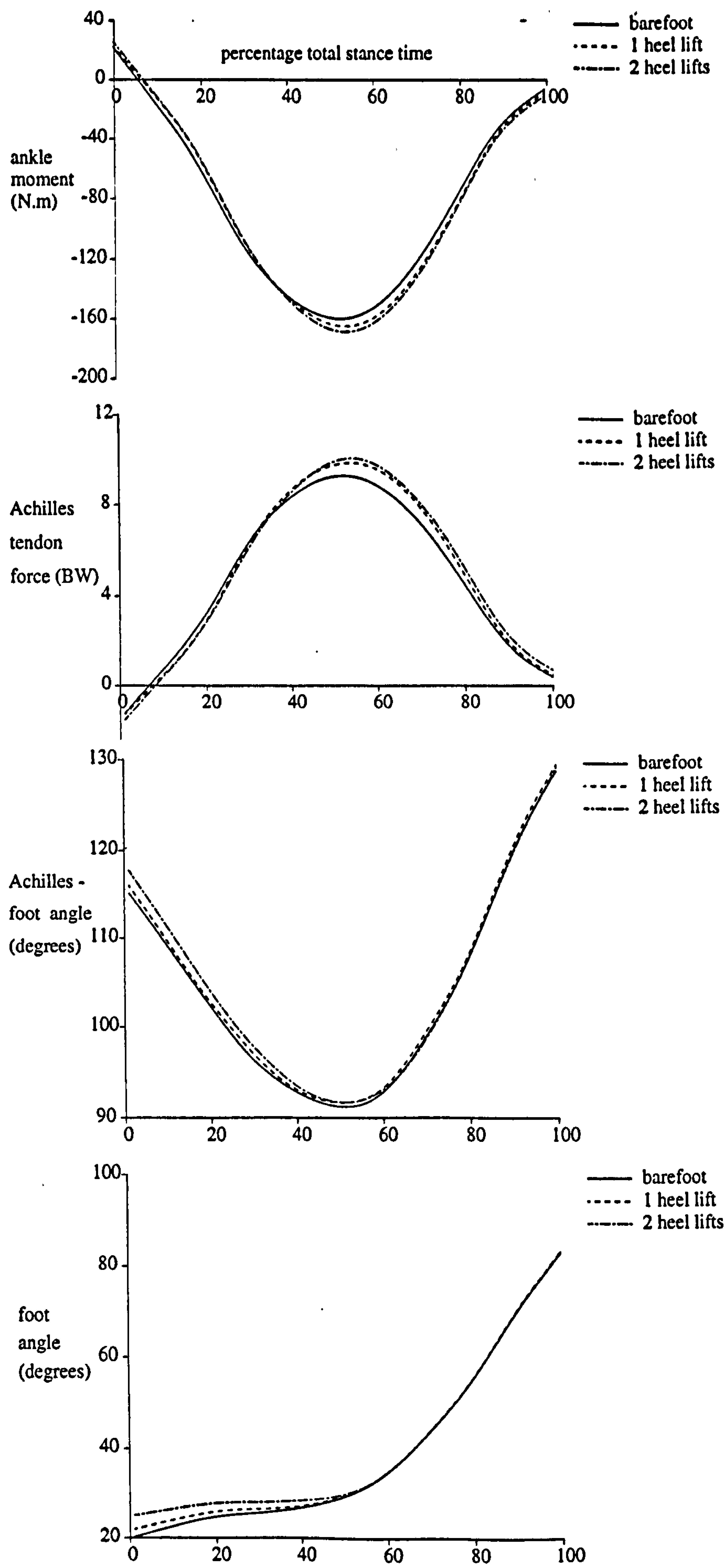


Figure 4.5 Ankle moment, Achilles tendon force, Achilles - foot and foot angles (Subject A)

Table 4.1 Maximum ankle moment (M_a) and Achilles tendon force (F) for each subject and condition (SD in parenthesis). Condition B = barefoot, Condition B1 = 7.5 mm heel lift, Condition B2 = 15 mm.

Condition	B		B1		B2	
	M_a (N.m)	F (BW)	M_a (N.m)	F (BW)	M_a (N.m)	F (BW)
Subject A	159 (8)	9.3 (0.5)	164 (6)	9.9 (0.5)*	169 (5)*	10.2 (0.5)*
Subject B	163 (6)	8.0 (0.3)	171 (6)*	8.5 (0.3)*	170 (5)*	8.5 (0.3)*
Subject C	231 (4)	17.8 (0.5)	234 (6)	18.0 (1.0)	230 (6)	17.9 (0.7)

* significant difference from barefoot condition ($p<0.05$)

Table 4.2 Peak Achilles tendon angle (θ), Achilles tendon force (F) and ankle moment (M_a), with corresponding times of occurrence, for 50 Hz and interpolated data (Subject A)

	θ (degrees)		F (BW)		M_a (N.m)	
	peak	%	peak	%	peak	%
B (50 Hz)	90.7	50	9.3	52	159	53
B (interp)	91.3	50	9.3	52	160	53
B1 (50 Hz)	91.2	53	9.9	54	164	54
B1 (interp)	91.7	50	9.9	54	165	54
B2 (50 Hz)	90.6	55	10.2	54	169	54
B2 (interp)	91.8	51	10.1	54	168	54

(ii) Ankle Moment

The peak ankle plantar-flexion moment values obtained using inverse dynamics and quasi-static methods were compared to the nearest 0.1 N.m for 10 trials, and found to be identical. Mean interpolated data for ankle joint moments are presented in Figure 4.6, for each of the subjects for the barefoot condition.

Significant differences in maximum ankle plantar-flexion moment were identified between conditions (Table 4.1). For Subject A, the rearfoot striker, a significant increase in maximum ankle moment was found for the 15 mm heel lift condition compared with the barefoot condition ($p<0.05$). For Subject B, the midfoot striker, significant increases in maximum ankle plantar-flexion moment were detected for both heel lift conditions compared with the barefoot condition ($p<0.05$). For Subject C, the forefoot striker, no significant differences in maximum ankle plantar-flexion moment were identified between conditions. No significant differences were identified in peak ankle moment across trials within each condition for any of the subjects.

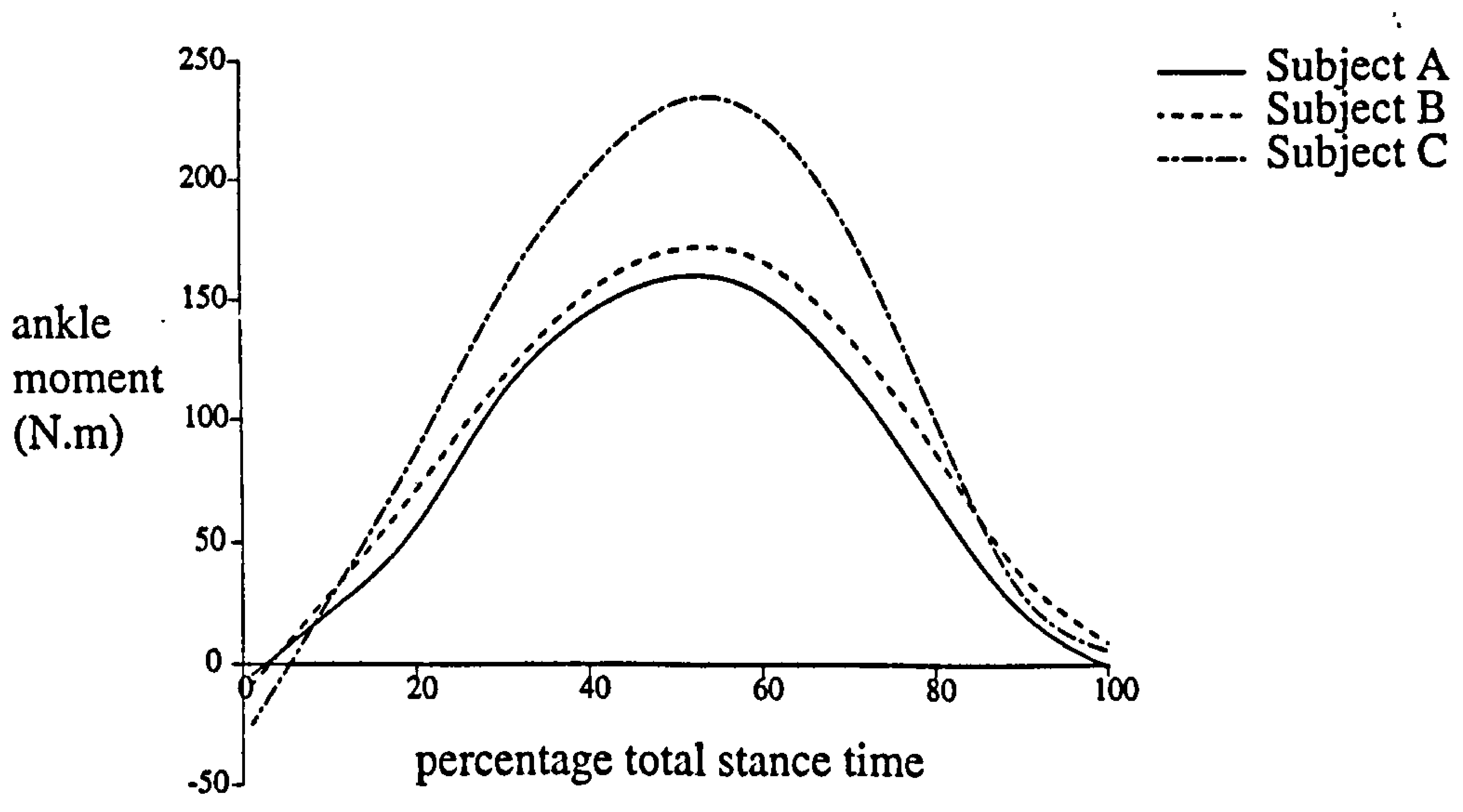


Figure 4.6 Barefoot ankle moment time histories for each subject (interpolated data)

(iii) Maximum Achilles Tendon Force

Mean interpolated data for Achilles tendon forces are presented in Figure 4.7, for each of the subjects for the barefoot condition. For Subject A and Subject B, the rearfoot striker and midfoot striker respectively, standardised effect sizes in maximum Achilles tendon force of 0.8 were observed. For a significance level of 0.05, the resulting statistical power was 0.97 for both subjects.

Significant differences in maximum Achilles tendon force were detected between conditions (Table 4.1). For Subject A and Subject B, significant increases in maximum Achilles tendon force were found for both heel lift conditions compared with the barefoot condition ($p<0.05$). For Subject C, the forefoot striker, a standardised effect size of 0.1 was observed. No significant differences in maximum Achilles tendon force were identified between conditions for this subject. No significant differences were identified across trials within each condition for each subject.

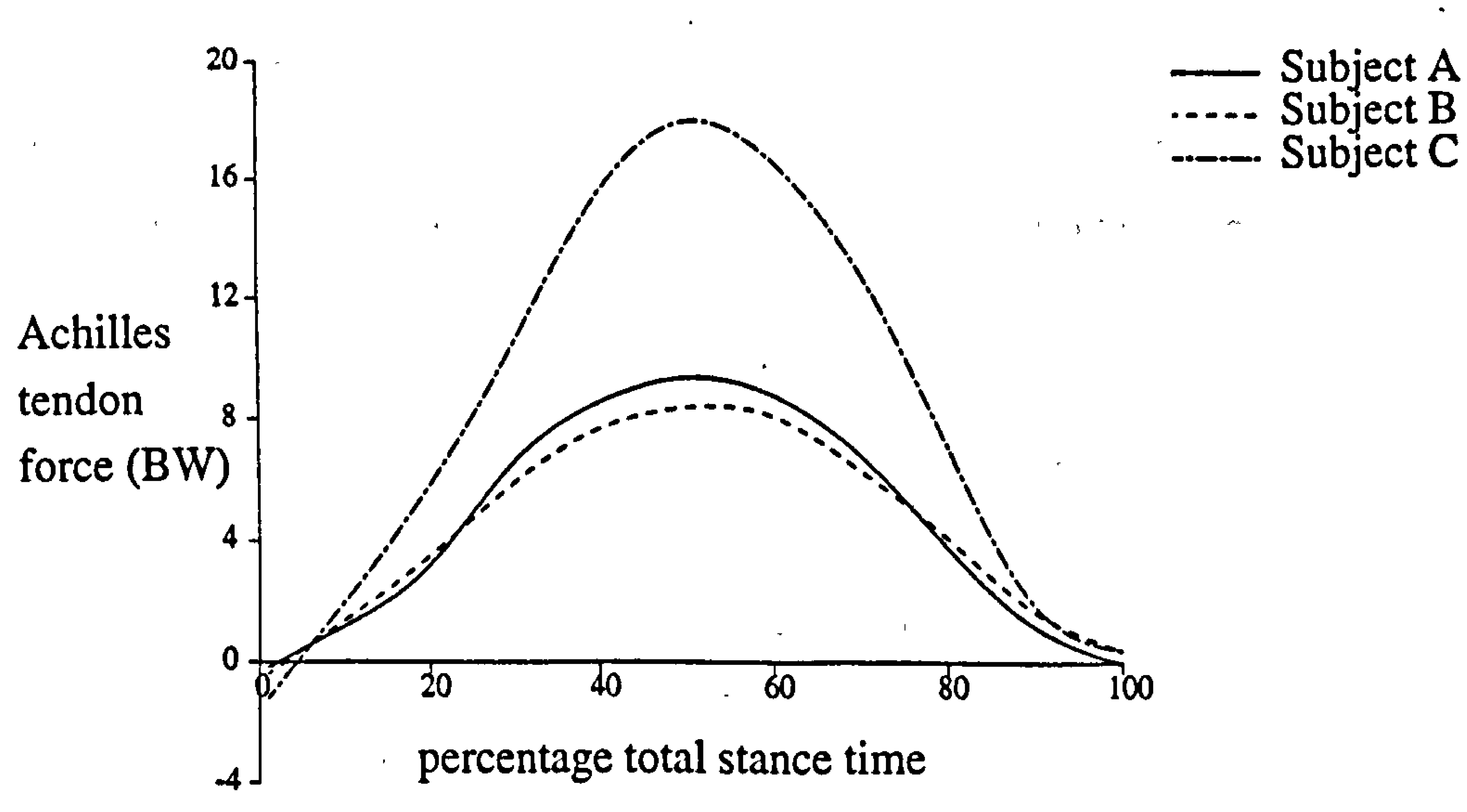


Figure 4.7 Barefoot Achilles tendon force time histories for each subject (interpolated data)

Table 4.3 provides values for the variables contributing to maximum Achilles tendon force variations between conditions, for Subject A and Subject B. Ankle moment, Achilles tendon moment arm, resultant ground reaction force (GRF) and moment arm of GRF at the time of maximum Achilles tendon force occurrence (d_2) are presented. An increase in ankle moment was found with heel lift for both subjects. With increased heel lift, a small decrease in Achilles tendon moment arm was detected. The increase in maximum Achilles tendon force with increased heel lift was therefore a result of the combined influence of an increase in ankle moment, and a decrease in the length of the Achilles tendon moment arm about the ankle joint centre.

The increase observed in ankle moment with heel lift was a result of increases in both GRF magnitude and moment arm about the ankle joint centre. It was found that the resultant GRF was close to vertical at the time of maximum Achilles tendon force occurrence, with the horizontal component being small and having both positive and negative values across trials within conditions. Increases in the length of the GRF moment arm with heel lift were therefore found to be predominantly the result of a more anterior point of application of resultant GRF relative to the ankle joint centre for increased heel lift conditions.

(iv) Achilles Tendon Angle

For all three subjects, similar shaped Achilles tendon angle plots were obtained during stance. Typical plots for each subject are provided in Figure 4.8 using mean interpolated barefoot data. The forefoot striker demonstrated a larger range of angles and larger magnitude of angles for the same periods of stance, than the other two subjects.

For the subjects which demonstrated significant differences in maximum ankle plantar-flexion moment and maximum Achilles tendon force across heel lift conditions, the rearfoot striker and the midfoot striker, changes in Achilles tendon angle were studied in detail. Angles at the time corresponding to maximum Achilles tendon force occurrence are presented in Table 4.3 for each condition. Similar Achilles tendon angles existed across conditions at the time of maximum Achilles tendon force occurrence for each of the subjects. Analysis of the contributing angles of foot angle and angle of Achilles tendon line of action to the vertical, revealed negligible change across conditions in these variables for the time corresponding to maximum Achilles tendon force occurrence.

Peak (minimum) Achilles tendon angle and time of occurrence are provided in Table 4.4 for each subject. The times of occurrence for peak angle were found to be similar to the times of maximum Achilles tendon occurrence. No changes in Achilles tendon angle magnitude or time of occurrence were found across conditions for Subject A, the rearfoot striker, or Subject C, the forefoot striker. For Subject B, the midfoot striker, the 15 mm heel lift condition resulted in a reduction in the maximum amount of ankle dorsi-flexion observed, indicating reduced dorsi-flexion of the ankle joint. For this subject, the time of occurrence of maximum ankle dorsi-flexion was increased for both heel lift conditions compared with the barefoot condition. The rearfoot striker and the midfoot striker demonstrated similar magnitudes for the peak Achilles tendon angle. The forefoot striker was found to have larger peak Achilles tendon angles, indicating a smaller amount of maximum ankle dorsi-flexion for this subject. Peak Achilles tendon angle also occurred later in the stance phase for this subject than for the other two subjects.

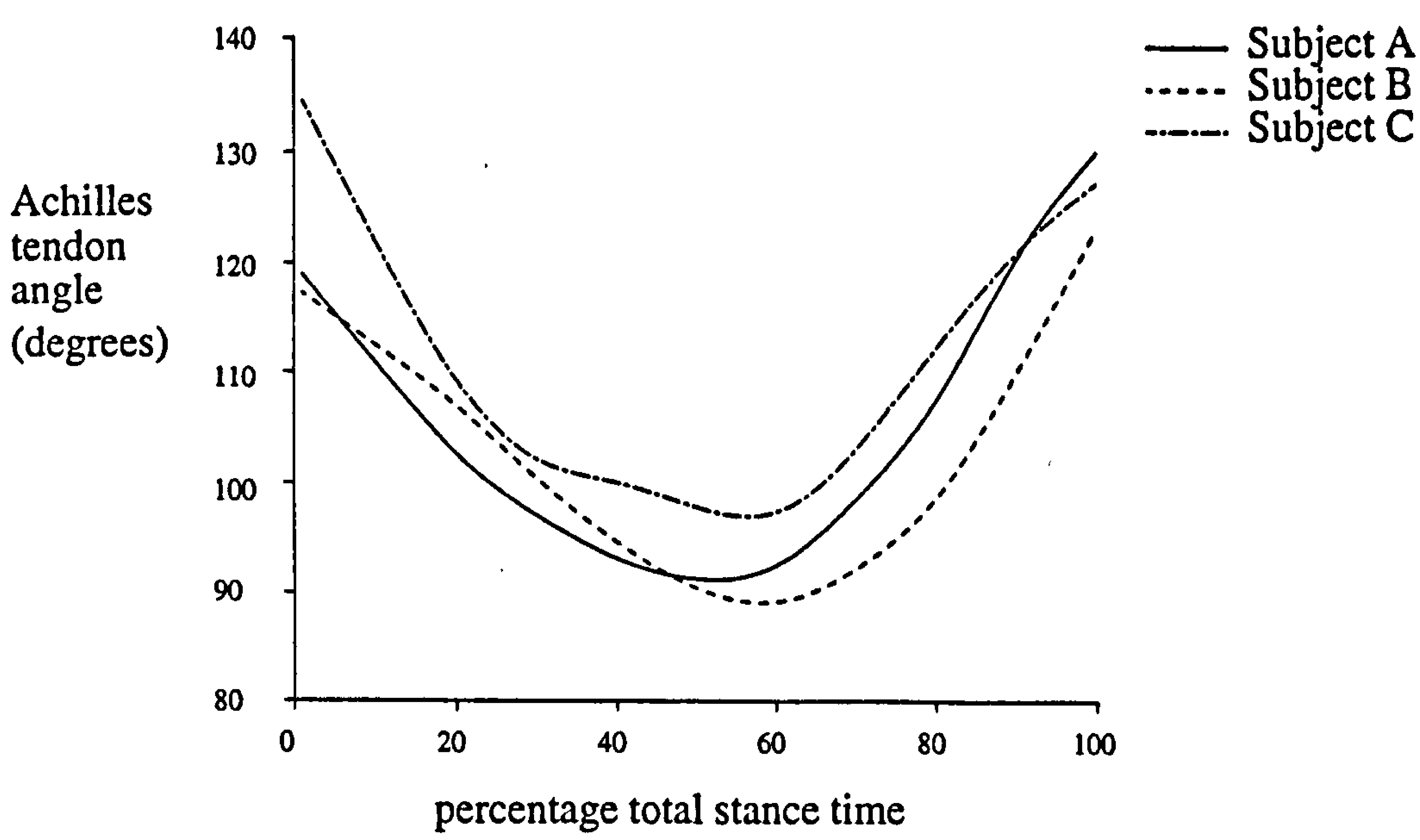


Figure 4.8 Barefoot Achilles angle time histories for each subject (interpolated data)

Table 4.3 Maximum Achilles tendon force and contributing factors ankle moment (M_a), Achilles tendon moment arm (d_1), GRF, GRF moment arm (d_2) and Achilles tendon angle (q) for (i) Subject A, and (ii) Subject B

(i) condition	max. F (BW)	max. M_a (N.m)	d_1 (mm)	resultant GRF (N)	d_2 (m)	θ (degrees)
barefoot	9.3 (0.5)	159 (8)	31.9 (0.6)	1372 (50)	0.1161	91.3
7.5 mm lift	9.9 (0.5)	164 (6)	30.7 (0.7)	1377 (39)	0.1192	91.9
15 mm lift	10.2 (0.5)	169 (5)	30.7 (0.7)	1420 (27)	0.1188	91.9

(ii) condition	max. F (BW)	max. M_a (N.m)	d_1 (mm)	resultant GRF (N)	d_2 (m)	θ (degrees)
barefoot	8.0 (0.3)	163 (6)	38.7 (0.7)	1395 (38)	0.1169	88.4
7.5 mm lift	8.5 (0.3)	171 (6)	38.0 (1.4)	1440 (70)	0.1190	88.5
15 mm lift	8.5 (0.4)	170 (5)	38.0 (1.0)	1430 (31)	0.1192	88.6

Table 4.4 Peak Achilles tendon angle (q, degrees) and time of occurrence (% total stance)

	B		B1		B2	
	peak	%	peak	%	peak	%
Subject A	91.3	50	91.7	50	91.8	51
Subject B	89.6	49	88.8	55	91.3	53
Subject C	96	63	96	62	97	62

(v) Achilles Tendon Moment Arm

Typical plots of Achilles tendon moment arm with varying Achilles tendon angle are provided for each subject in Figure 4.9. For the rearfoot striker and the midfoot striker, increased Achilles tendon moment arm occurred with increased ankle angle. For the forefoot striker, Achilles tendon moment arm initially increased with the decreasing ankle angle following ground impact. With a continued reduction in angle, the Achilles tendon moment arm was reduced. The Achilles tendon moment arm then showed a steady increase with increasing ankle angle, corresponding to the push-off phase.

Using anatomical measurements, a systematic difference of 28 degrees between foot angle measured in the present study and that obtained using conventional methods was calculated for Subject A. Subtraction of 28 degrees from measured Achilles tendon angles therefore provided data suitable for comparison with ankle angles in the literature. Figure 4.10 illustrates plots of moment arm lengths against ankle angle obtained using the methods of the present study, and those obtained using the relationships of Burdett (1982) and Bobbert et al. (1986), for a barefoot running trial for Subject A. Differences of up to 5 mm in length were found between moment arms estimated using the methods of the present study and those calculated using relationships from the literature. Differences in shape of the angle-moment arm plots were also observed.

(vi) Evaluation of Methods

The RMSD in locating known points are compared for the chosen cursors in Table 4.5. The use of different cursor designs was found to have no influence on the accuracy attained in locating marker coordinates.

The mean values for maximum Achilles tendon force for the barefoot condition for Subject A over three separate testing sessions, with new marker placement for each session, were 10.8 BW, 10.2 BW and 9.3 BW. A maximum variation of 1.5 BW was therefore demonstrated between testing sessions. Within each testing session with the use of a single set of markers, a maximum variation in maximum Achilles tendon force of 1.5 BW, and standard deviation of 0.5 BW were obtained over 10 trials. Thus for a 95% confidence interval, a maximum variation of ±0.3 BW is expected due to random variation. The variation in marker placement between testing sessions will have a systematic influence on maximum Achilles tendon force results, and thus should not influence the results of comparisons between conditions for a single subject during a testing session.

With a single set of markers used within a testing session, the naturally occurring amount of random error in the digitising process was found to result in a RMS variation of 0.4 mm for each of the points. The subsequent variations in moment arm length and maximum Achilles tendon force were 0.4 mm and 0.14 BW, respectively. The digitisation of the increased foot image size using the Prisma digitising system, as in Chapter 3, provided a RMS variation of 0.14 BW, a value identical to that attained using the new Apex system.

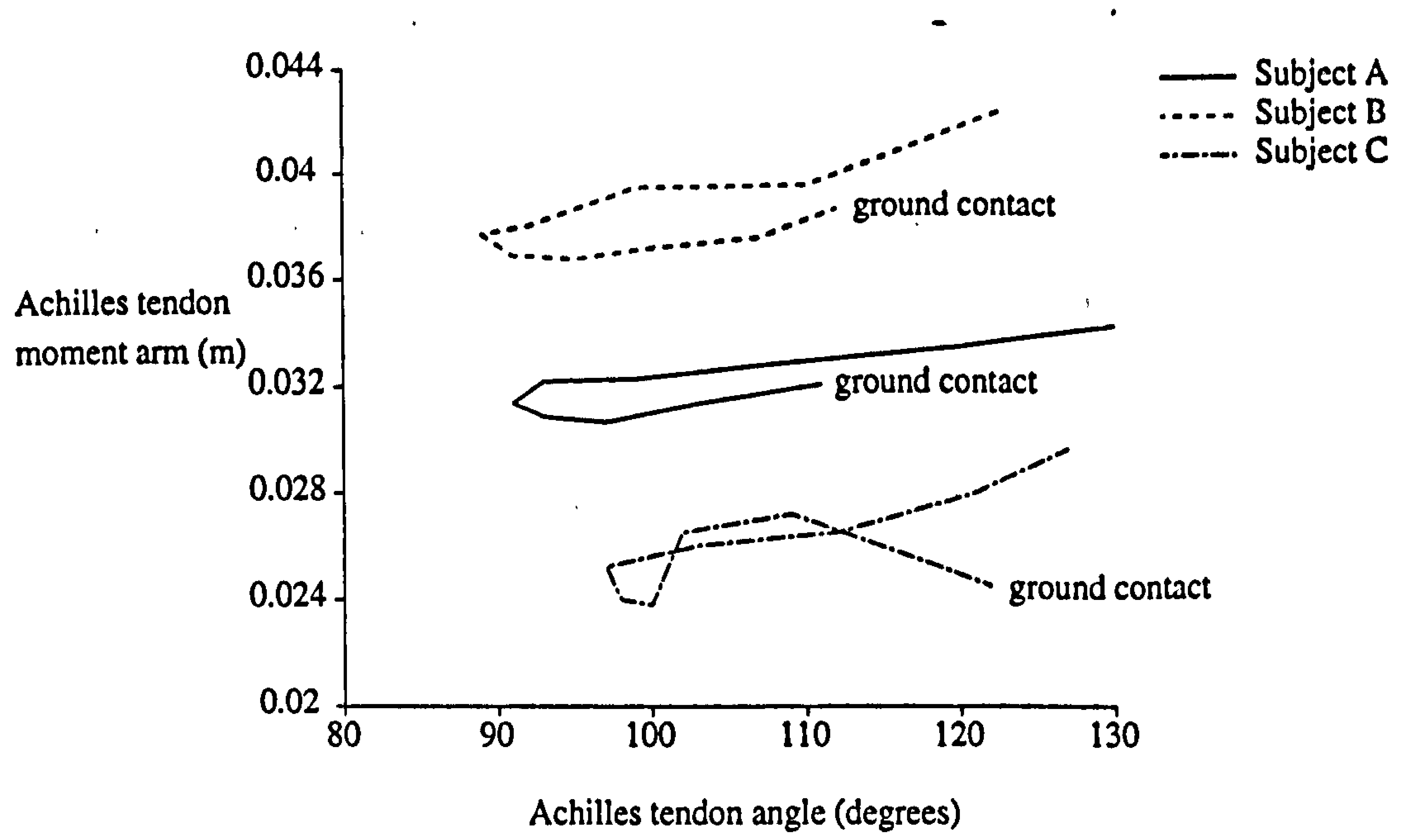


Figure 4.9 Typical plots of Achilles tendon moment arm against Achilles tendon angle

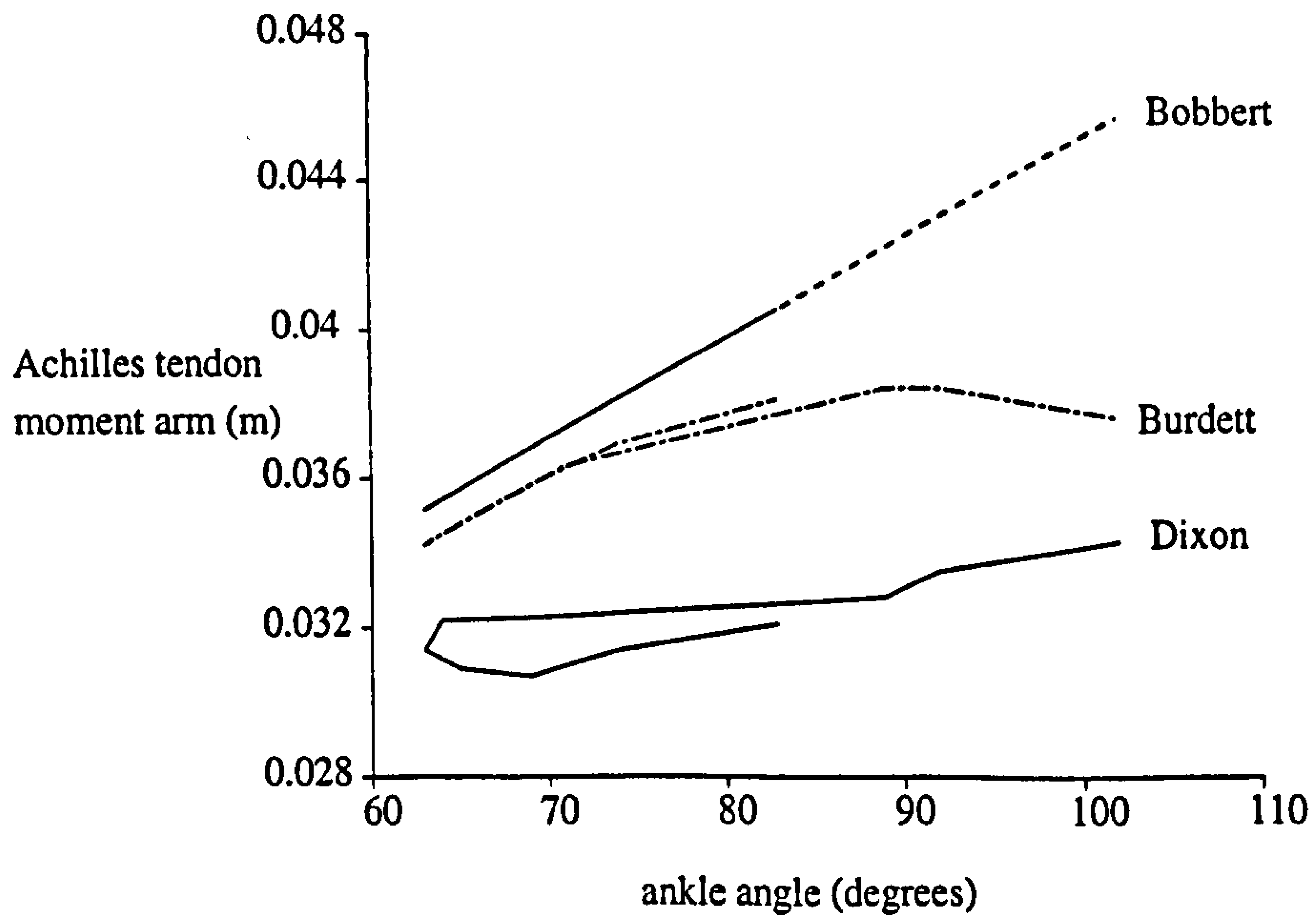


Figure 4.10 Achilles tendon moment arm against Achilles tendon angle using data from Burdett (1982); Bobbert et al. (1986); and the present study

Table 4.6 provides RMSD values when locating known points, with differences presented between the actual coordinates and their estimated locations for assessment of accuracy. The influence of repeated digitisation on estimated locations was assessed by relocating the five known points using one digitisation, four repeated digitisations, and ten repeated digitisations. This procedure also provided values to illustrate the degree of random variation occurring during digitisation. Over the five points, a mean RMSD of 0.4 mm was found in the horizontal (y) direction, and 0.7 mm in the vertical (z) direction when each of the points were digitised once. Using mean locations over four repeated digitisations, mean RMSD values were reduced to 0.2 mm and 0.5 mm, in the y and z directions respectively. The use of ten repeated digitisations did not reduce the RMSD values further. The RMSD in measured locations using zero, two and four wobble points were identical.

For the evaluation of the methods used in this study, a sensitivity analysis was employed. For a typical trial, it was found that a variation in the centre of pressure location by 2 mm resulted in a 3 N.m (2%) difference in ankle joint moment, and a 0.2 BW (2%) difference in maximum Achilles tendon force. A 1% variation in the magnitude of the resultant ground reaction force will result in a 1% variation in ankle moment, and thus a 1% variation in the Achilles tendon force value. Thus the combined influence of errors in forceplate data was 0.3 BW (3%). The possible influence of 0.4 mm digitising error on estimated ankle moment, Achilles tendon moment arm and maximum Achilles tendon force are presented in Table 4.7. The ankle moment was not influenced by the Achilles tendon marker locations, and the variation in ankle marker location of 0.4 mm had only a small influence on ankle moment (±0.4%). A maximum possible variation in the Achilles tendon moment arm of 1.1 mm (4%) was found. The combined influence of these changes in ankle moment and Achilles tendon moment arm resulted in a maximum variation of 0.5 BW (5%) in the estimated maximum Achilles tendon force.

Table 4.5 The influence of different cursors on precision of locating points

Point	Cursor 1 RMSDY	Cursor 1 RMSDZ	Cursor 2 RMSDY	Cursor 2 RMSDZ
Ankle	0.0009	0.0009	0.0008	0.0009
Ach.1	0.0008	0.0011	0.0006	0.0007
Ach.2	0.0008	0.0008	0.0010	0.0007

Table 4.6 RMSD for one, four and ten digitisations of known points (m)

	1 Digitisation		4 Digitisations		10 Digitisations	
Point	RMSDY	RMSDZ	RMSDY	RMSDZ	RMSDY	RMSDZ
1	0.0001	0.0010	0.0005	0.0007	0.0005	0.0007
2	0.0007	0.0007	0.0002	0.0005	0.0003	0.0005
3	0.0002	0.0008	0.0000	0.0006	0.0000	0.0006
4	0.0002	0.0000	0.0000	0.0001	0.0001	0.0005
5	0.0005	0.0005	0.0001	0.0003	0.0001	0.0003
Mean	0.0004	0.0007	0.0002	0.0005	0.0003	0.0005

Table 4.7 Influence of ankle joint and Achilles tendon marker variation of 0.4 mm

	Ach.1	+0.0004 -0.0004	-0.0004 +0.0004
Y	-0.0501	-0.0497	-0.0505
Z	0.0733	0.0729	0.0737
	Ach.2	+0.0004 -0.0004	-0.0004 +0.0004
Y	0.0027	0.0031	0.0023
Z	0.1438	0.1434	0.1442
	Ankle	-0.0004 +0.0004	+0.0004 -0.0004
Y	-0.0102	-0.0106	-0.0098
Z	0.0803	0.0807	0.0799
M_a (N.m)	157.6	158.2 (+0.4%)	157.0 (-0.4%)
F (BW)	10.6	11.1 (+5%)	10.1 (-5%)
d_1 (m)	0.0276	0.0265 (-4%)	0.0287 (+4%)
Δd_1 (m)		-0.0011	+0.0011

M_a = ankle moment
F = Achilles tendon force
d_1 = Achilles tendon moment arm

4.4 Discussion

(i) Ankle Moment

Ankle moment traces obtained in the present study were similar to those presented in the literature. The small dorsi-flexion moment demonstrated at the start of ground contact is consistent with several published studies (Winter, 1983; Reinschmidt and Nigg, 1995). The presence of an initial dorsi-flexion moment for the forefoot striker is in contrast with the unpublished findings concerning ankle joint moments for forefoot strikers, referenced by Reinschmidt and Nigg (1995). Examination of 50 Hz results for this subject indicated that the calculated ankle joint moments were always plantar-flexor moments, suggesting that the apparent dorsi-flexion moment obtained in the present study was the result of using interpolated data for production of plots of mean data. Kinematic data at an increased sampling rate is required for detailed study of joint moments during this phase of stance.

The magnitudes of peak ankle plantar-flexion moment for the rearfoot striker and midfoot striker are to the lower end of the range of values presented in the literature (Reinschmidt and Nigg, 1995). This is likely to be due to the small mass of the subjects, as female distance runners, and the use of a running speed that is towards the slower end of those generally utilised in the literature. The maximum ankle plantar-flexion moment for the forefoot striker is greater than those calculated for the rearfoot striker and the midfoot striker, and is towards the higher end of the range of values presented in the literature. This subject was of similar body mass to the other two subjects of the present study, and ran at the same speed in the running trials. It is therefore suggested that the larger peak ankle plantar-flexion moment was the result of the forefoot ground strike adopted by this runner.

The finding that sagittal plane peak ankle plantar-flexion moment was increased with increasing heel lift for the rearfoot and midfoot strikers conflicts with the results obtained by Reinschmidt and Nigg (1995), who observed no significant differences in this variable with increased heel height. The contrasting results may be due to differences in conditions since Reinschmidt and Nigg (1995) used running shoes with differing heel heights, whereas in the present study the heel height increases were introduced by attaching heel lifts to the barefoot. The running shoes worn by subjects in the study by Reinschmidt and Nigg (1995) are likely to have provided varying degrees of shock absorption with different heel heights, whereas in the present study the intervention was limited to heel height variation between conditions. The running shoes used by Reinschmidt and Nigg (1995) had absolute heel heights ranging from 21 mm to 33 mm, corresponding to heel lifts of between 13 mm and 25 mm, due to the shoe forefoot thickness of 8 mm. The 15 mm heel lifts employed in the present study fell into this range of lift. The cushioning properties of the forefoot of the shoes used by Reinschmidt and Nigg (1995) provided a further difference in conditions compared with the present study.

(ii) Achilles Tendon Force and Angle

The general shape of the Achilles tendon force time histories obtained for all three subjects in the present study were similar to those illustrated in the literature (Burdett, 1982; Komi, 1990). The presence of an initial negative force at ground impact is consistent with the rearfoot strike results of Komi (1990). However, the existence of this negative force for the forefoot striker in the present study is in contrast with the findings of Komi (1990). The use of interpolated data limits the ability to interpret findings for the initial impact phase.

The finding in the present study that there were no significant differences between trials within each condition, indicates that factors such as fatigue, boredom and familiarisation did not influence maximum Achilles tendon force values. This supports the findings of Natrup (1994) who found that, even when running to fatigue, there were no changes in net muscle action about the ankle joint.

Using direct measurement techniques, Komi (1990) obtained a maximum Achilles tendon force of 12.5 BW for a single subject performing running trials. Values for maximum Achilles tendon force obtained using estimation methods presented in the literature range from approximately 6 BW to 10 BW (Burdett, 1982; Scott and Winter, 1990). The maximum Achilles tendon force values obtained in the present study for the rearfoot striker and forefoot striker are within this range of values. However, the maximum Achilles tendon force values estimated for the forefoot striker are considerably higher than any values presented in the literature. It has been noted that direct comparisons of magnitudes of Achilles tendon force between subjects are not possible using the methods of the present study, due to variations in marker placement influencing results. However, the large difference observed for the forefoot striker compared with the other two subjects, is considerably greater than possible differences due to marker placement variation. It is therefore suggested that the large difference in maximum Achilles tendon force for the forefoot striker in the present study is a true difference. This high force value is the combined result of a greater ankle moment for this subject and a smaller moment arm of the Achilles tendon about the ankle joint centre. The relatively large magnitude of maximum Achilles tendon force for a subject using a forefoot ground strike in the present study is in contrast with the findings of Komi (1990). Komi (1990) obtained Achilles tendon force values for a single subject using a rearfoot and a forefoot striking action, and found that similar magnitudes of maximum Achilles tendon force were obtained for these two different ground contact styles. The subject used by Komi (1990) is likely to have had to make a conscious effort to change running style, whereas Subject C in the present study naturally ran using a forefoot ground contacting style. It is speculated that the different loading of the Achilles tendon for a natural forefoot striker in the present study compared with the subject in the Komi (1990) study, may be due to the subject used by Komi (1990) having to make a conscious effort to change running style.

The finding in the present study that raising the heel in relation to the forefoot does not necessarily reduce maximum Achilles tendon force, is contrary to the suggestions of several authors (Clement et al., 1984; Leach et al., 1981). This finding is also in contrast to the results presented in Chapter 3, in which shock absorbing heel lifts were employed. The rearfoot striker of the present study was the same subject as that employed in Chapter 3. The common baseline condition across studies was the barefoot condition. It was therefore possible to compare behaviour relative to the barefoot condition for the rearfoot striking subject across studies. The finding that maximum Achilles tendon force was reduced by the attachment of shock absorbing heel lifts, but increased by the use of firm heel lifts, suggests that the emphasis when attempting to reduce Achilles tendon maximum force should be on additional shock absorption provision, as opposed to increased heel height. It is possible that simultaneous increased shock absorption and heel height are required for optimum reduction of maximum Achilles tendon force. The placement of firm material in a shock absorbing shoe may also reduce maximum force in the Achilles tendon. Although the described result is for a single subject, the possible increase in maximum Achilles tendon force when heel lift is increased indicates that caution must be advised when routinely prescribing heel lifts. The reduction in Achilles tendon injury described in the literature (Fauno et al., 1993; MacLellan and Vyvyan, 1981; Subotnick, 1979) may be due to the additional shock absorbing properties provided by heel lifts prescribed by practitioners, or to the combined influence of raising the heel and providing shock absorption. In contrast to previous investigations, firm heel lifts were attached to the barefoot of the subjects in an attempt to contribute to an understanding of the response of running subjects solely to heel height interventions. At present the application of the results of this study to a clinical setting are limited. However, it is clear that the mechanism by which heel lifts contribute to a reduction in Achilles tendon injury is complex.

An alternative suggested explanation for the discrepancy with the literature is that the common assumption that Achilles tendon injury is related to Achilles tendon force is incorrect. As discussed in Chapter 3, it has been assumed in the present study that the occurrence of Achilles tendon injury is related to the maximum Achilles tendon force that the tendon is subjected to. This has been assumed due to the evidence in the literature that repeated stress of a relatively low magnitude can have an adverse influence on body structures (Radin et al., 1982; Archambault et al., 1995). In Chapter 3, the prescription of heel lifts to treat Achilles tendon injury was suggested to decrease the maximum strain that the tendon is subjected to, and to simultaneously reduce the maximum Achilles tendon force. The appropriateness of the assumption of a linear relationship existing between stress and strain in the Achilles tendon during running has been investigated in the present chapter by simultaneous monitoring of the Achilles tendon force and the angle between the Achilles tendon and the foot. The Achilles tendon angle time histories have provided an illustration of the change in length of the triceps surae muscle-tendon complex due to motion about the ankle. It has been shown that minimum Achilles tendon angle, and thus maximum dorsi-flexion of the ankle joint, occurs at the same percentage of stance as the maximum Achilles tendon force. This finding supports the suggestion that maximum stress and maximum strain in the Achilles tendon occur simultaneously, indicating that the suggested rationales behind the prescription of heel lifts of minimising ankle dorsi flexion, and thus stretching of the muscle-tendon complex, and reducing maximum Achilles tendon force, may both equate to the same thing.

An alternative variable which may be related to the occurrence of Achilles tendon injury is strain rate. It has been found that loading rate of GRF impact peak is more sensitive to changes in running shoes than the magnitude of force (Therrien et al., 1982; Clarke et al., 1983b). Loading rate may be more important than the magnitude of forces on body structures when considering the etiology of injury (Nordin and Frankel, 1980). As a viscoelastic material, tendon is sensitive to strain rate (Hawkins, 1993). However, Ker (1981) found that isolated tendon mechanical properties were not influenced by loading rate, for the range of rates expected during locomotion. Knowledge of the loading rate of the Achilles tendon during ground contact in running and the influence of heel lift on this rate is required.

A further variable which has been associated with the occurrence of Achilles tendon injury is 'excessive' rearfoot movement in the frontal plane (Clement et al., 1984). Winter and Bishop (1992) described how the conflicting rotations at the knee and subtalar joints associated with prolonged subtalar joint pronation may result in the twisting of the Achilles tendon structure. Methods for monitoring of subtalar joint pronation using the measurement of frontal plane rearfoot angle have been described extensively in the literature (Clarke et al., 1983b; Nigg, 1986a). It has been suggested that the maximum range of subtalar joint pronation can be reduced by wearing shoes of increased heel height, although conflicting results have been obtained when this intervention has been applied (Bates et al., 1978; Stacoff and Kaelin, 1983; Clarke et al., 1983b). Further investigation into the possible influence of heel lift on rearfoot motion will provide evidence on the suitability of a heel lift intervention for reduction of subtalar joint pronation.

The use of a single subject approach has allowed detailed study of the behaviour of distinct subjects. The finding that the introduction of firm heel lifts had a differing influence on maximum Achilles tendon force in runners with distinct styles supports the findings of several authors that runners with varying styles are influenced differently by footwear interventions (Therrien et al., 1982; Lees and McCullagh, 1984). Using a single subject approach, the differing response across subjects has highlighted the need for individual assessment. The conclusions of the study regarding the response of subjects demonstrating particular ground contact styles are limited because of the use of only one subject with each of the styles. However, the results of the present study have been used to provide strong contrasting evidence. It has generally been assumed that there is a decrease in Achilles tendon force with increased heel height. Demonstration that this is not the case for a single subject is sufficient to conclude that a decrease in Achilles tendon force with heel height increase cannot be assumed.

The finding that raising the heel did not influence maximum Achilles tendon force for the forefoot striking subject, is not surprising considering that the rearfoot of this subject does not contact the ground during the stance phase of running. This subject had previously been independently prescribed orthotic devices to be placed in the rear of the shoe for treatment of Achilles tendon injury. It was therefore of interest to find that interventions at the rear of the foot did not influence Achilles tendon force or ankle angle for this subject.

(iii) Achilles Tendon Moment Arm

Changes in the length of the Achilles tendon moment arm have been found to contribute to the difference in estimated Achilles tendon forces between conditions. A factor influencing this length is the orientation of the Achilles tendon, which is determined by the ankle and knee joint angles. The influence of joint angle changes on the moment arm of the Achilles tendon has been illustrated in the literature, with conflicting relationships being found. Bobbert et al. (1986) and Rugg et al. (1990) found a linear increase in Achilles tendon moment arm with increasing ankle angle. In contrast, Spoor et al. (1990) found that there was a steady increase in Achilles tendon moment arm up to an ankle angle of approximately 100 degrees, beyond which the moment arm decreased in length with increasing ankle angle. A factor influencing the moment arm variation with ankle joint angle is the location of the Achilles tendon insertion relative to the ankle joint centre. If a rigid link system is assumed, then the influence of joint angle changes on Achilles tendon moment arm is as illustrated in Figure 4.11. Whilst the Achilles tendon insertion is lower than the ankle joint centre, an increase in ankle plantar-flexion results in an increase in Achilles tendon moment arm. The behaviour of the Achilles tendon moment arm if the Achilles tendon insertion moves higher than the ankle joint centre depends on the relative locations of the knee and ankle joints. Maximum moment arm length occurs around the time when the Achilles tendon insertion is on the same horizontal line as the ankle joint centre. If ankle plantar-flexion continues, then the moment arm of the Achilles tendon is reduced with increasing ankle angle. Due to the relatively distant location of the knee joint from the Achilles tendon, changes in knee angle have a relatively small influence on the length of the Achilles tendon moment arm compared with ankle angle changes. The described relationship between ankle angle and Achilles tendon moment arm for a rigid body system is consistent with the findings of Spoor et al. (1990), although data on the relative location of the Achilles tendon insertion and the ankle joint centre were not provided. If the Achilles tendon insertion remains lower than the ankle joint centre, then the Achilles tendon moment arm will show a steady increase with increasing ankle angle, as found by Bobbert et al. (1986) and Rugg et al. (1990).

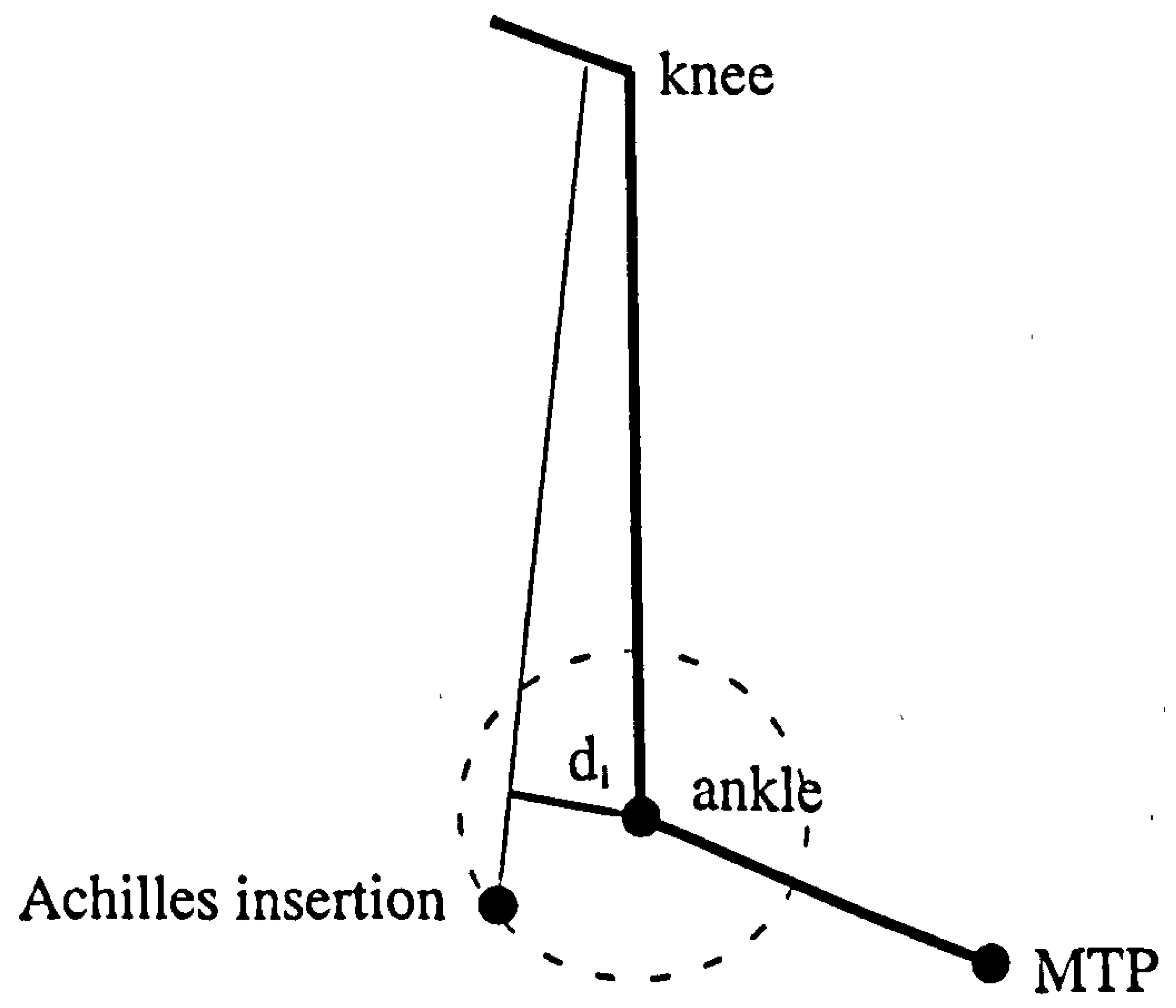


Figure 4.11 Geometry relating Achilles tendon insertion and ankle joint centre

The general increase in moment arm with increasing angle found for the rearfoot striker and the midfoot striker in the present study, supports the findings of Bobbert et al. (1986) and Rugg et al. (1990). However, smaller magnitudes of moment arm of up to 5 mm have been attained using the methods of the present study. Differences of this magnitude are possible between subjects (Burdett, 1982). The difference in shape of the ankle angle-moment arm trace obtained using the methods of the present study and that obtained using the methods of Bobbert et al. (1986) is most marked beyond approximately 90 degrees ankle angle. It is evident, from the data of the present study and the literature, that the steep increase in Achilles tendon moment arm with ankle angle obtained using the Bobbert et al. (1986) data is not appropriate for large ankle plantar-flexion angles. Spoor et al. (1990) stated that moment arm curves cannot always be approximated by straight lines, and that inter-individual differences cannot be represented adequately using segment length alone, as is the case when using the methods of Bobbert et al. (1986). The reduction in moment arm beyond an angle of approximately 90 degrees when using the methods of Burdett (1982) are expected using these methods, since when ankle angle is 90 degrees using the Burdett data, the insertion of the Achilles tendon is level with the ankle joint centre, around the orientation corresponding to maximum Achilles tendon moment arm length. This behaviour is similar to that described by Spoor et al. (1990).

Spoor et al. (1990), and Rugg et al. (1990) found that the variation in Achilles tendon moment arm with ankle angle was similar during the actions of ankle plantar-flexion and ankle dorsi-flexion. The results of the present study are contrary to this, with smaller moment arm values during dorsi-flexion than during plantar-flexion for the same ankle joint angles. Attempting to simulate active conditions, Spoor et al. (1990) applied a constant force when measuring moment arm lengths, and Rugg et al. (1990) instructed subjects to exert a force on the apparatus constraining their foot while measurements were made. The forces applied in these studies are unlikely to be close to those occurring during actual running. During running stance, the resultant GRF changes, as does the point of resultant force application, and thus deformation of soft tissues will vary throughout ground contact. It is suggested that this differing deformation will result in the moment arm lengths varying at different stages of stance, for corresponding ankle angles.

During the impact phase of running, knee flexion occurs with simultaneous ankle dorsi-flexion, whereas during the push-off phase the knee extends while ankle plantar-flexion occurs. At a defined ankle angle during impact, the knee joint will be more flexed than for the same ankle angle during push-off. Spoor et al. (1990) found that the moment arm of the gastrocnemius about the knee joint centre increased during knee extension. It may be, therefore, that knee extension during push-off contributes to an increase in Achilles tendon moment arm compared with the same ankle angle during the impact phase. This suggestion is consistent with the findings of the present study, in which greater Achilles tendon moment arm lengths were obtained during the push-off phase than in the impact phase.

(iv) Evaluation

A sensitivity analysis has revealed that the maximum influence of errors in forceplate measurements of 0.3 BW and in marker point digitisation error of 0.5 BW, resulted in a total maximum possible error in Achilles tendon force of 0.8 BW. The use of mean Achilles tendon force values over 10 trials for each condition reduced the magnitude of possible random error to 0.25 BW. It is noted that this is the error corresponding to the worst possible case, with maximum random error in all variables occurring simultaneously.

The likely error in maximum Achilles tendon force values due to random errors is better represented by use of precision measurements obtained over repeated digitisations. By combining the maximum influence of force plate random error of 0.3 BW with the random digitisation error influence of 0.14 BW, a combined value of 0.44 BW has been obtained. This has been reduced to 0.14 BW by the use of the mean of 10 trials for each condition.

The precision in maximum Achilles tendon force values attained in the present study was compared with that of Chapter 3. Precision in GRF data should be the same between studies since the same apparatus and procedures were used. For digitisation, different systems were employed and different foot image sizes were used. The measurement precision was increased by approximately four times, (from 0.54 BW to 0.14 BW) by the use of the increased foot image size and the new digitising system (Target), compared with the smaller foot image size and the Prisma digitising system. To assess the relative influence on measurement precision of the different digitising system and the increased foot image size, the two different digitising systems were compared using the same larger foot image size. The finding that the use of the different systems for digitisation had no influence on the measurement precision for the increased foot image size, was in contrast with the findings of Tan et al., (1995). These authors found that the Apex based system could reduce variations in digitised data by up to 44% compared with the Prisma system. The absence of a detectable difference in measurement precision by the use of the Apex system compared with the Prisma system in the present study is likely to be due to the large object image size employed. A field of view of 0.8 m was employed in the present study with the foot filling the image, compared with a 10 m field of view in the study by Tan et al., (1995). It is concluded that in order to attain maximum measurement precision in peak Achilles tendon force values, the field of view should be minimised. The level of precision in maximum Achilles tendon force values attained in the present study was adequate for the confident detection of differences between conditions, with effect sizes in the region of 0.5 BW being demonstrated.

The variation of 1.5 BW in maximum Achilles tendon force found for a single subject between testing sessions has been found to be predominantly due to differences in marker placement. Location of the ankle joint marker is very influential when calculating Achilles tendon force, since both the ankle moment and the length of the moment arm of the Achilles tendon will be influenced by its position. This marker was placed on the most prominent part of the lateral malleolus, to represent the location of the ankle joint centre in the sagittal plane. This is consistent with the location generally described in the literature (Scott and Winter, 1990), and is the most reproducible point to locate on the ankle. Any variations in the placement of the ankle marker and Achilles tendon markers will cause a systematic variation in the resulting Achilles tendon force values, and should not therefore influence comparisons of maximum force between conditions for a single subject during a testing session.

A degree of skin movement will influence the position of the ankle joint marker during stance relative to the true ankle joint centre, and the Achilles tendon markers relative to the line of action of the Achilles tendon. Since Achilles tendon angle at maximum Achilles tendon force has been found to be similar across conditions, the amount of skin movement is likely to be of the same magnitude at the time of maximum Achilles tendon force for each condition. Additionally, the maximum Achilles tendon force occurs during the midstance phase when foot movement is at a minimum, and therefore differences in skin movement between conditions are likely to be negligible. Detailed visual observation of the tendon markers during ground contact supported the use of these markers for the estimation of the line of action of the tendon during the middle of the stance phase. For increased confidence in the suitability of the chosen markers and to assess the influence of skin movement on results, a measurement method which can monitor bone and tendon movement simultaneously to skin marker movement is required. Suitable methods may be the use of magnetic resonance imaging (MRI) or ultrasound.

Heel lifts have been attached to the barefoot in the present study in an attempt to isolate the intervention of raising the heel. The influence of extraneous variables, including differing amounts of slippage and friction between the barefoot and the heel lift conditions and the presence of the tape used for attachment, has been assumed to be negligible. The influence of attachment of the lifts to the rearfoot was minimised by using a light micropore tape attached to the skin covering the calcaneus which did not cross the ankle joint. No movement of lifts relative to the foot or slippage during running were reported by the subjects. Additionally, the extraneous effects which have been identified are differences between barefoot and heel lift conditions, indicating that differences detected between the two different heel lift conditions are likely to be due to heel height variation alone.

The assumptions that the triceps surae muscle group is the sole ankle plantar-flexor, and that the triceps surae group acts in the two-dimensional sagittal plane, will introduce errors in Achilles tendon force estimation, influencing the relationship between the estimated Achilles tendon force values and the true forces in the tendon. The validity of these assumptions may vary across conditions for corresponding times in the stance phase. Since maximum Achilles tendon forces have been found to occur at approximately the same percentage of the stance phase for all subjects and all conditions, it has been assumed that any error due to the assumptions will be systematic and will therefore not influence comparisons between conditions for a single subject within a testing session. These assumptions are supported by the finding in the present study that, for each subject, only small differences in Achilles tendon and foot angles occurred across conditions at the time of maximum Achilles tendon force. The assumptions used in the present study should not, therefore, influence the ability of the methods to test the influence of varying heel heights for different subjects and compare these responses between subjects. For a single subject, any trend in results with increased heel height should be reproducible between testing sessions. Findings published in the literature have demonstrated the possible influence of heel lift on movement of the calcaneus in the frontal plane, with negligible lower leg movement being demonstrated (Bates et al., 1978). The relatively large sagittal plane joint angles measured during midstance are unlikely to be influenced significantly by small changes in the frontal plane orientation of the calcaneus. To support this suggestion, monitoring of the amount of rearfoot and lower leg movement in the frontal plane is required for individual subjects.

The comparisons made between heel height conditions in the present study have been limited to the midstance phase, when errors resulting from the use of a relatively low sampling rate of kinematic data (50 Hz) will be minimal, due to the slow rate of change of the measured variables at this stage of stance. This suggestion is supported by the illustration by Komi (1990) that Achilles tendon force varies little over a 0.02 s period around the time of maximum Achilles tendon force. The use of mean force values over 10 trials for each condition has reduced the influence of this source of error on comparisons between heel heights, by reducing the error to similar magnitudes across conditions.

The synchronisation of 50 Hz video data with 1000 Hz forceplate data may be in error by up to ±0.01 s. The vertical component of GRF changes very little around the middle of the stance phase, when maximum Achilles tendon force occurs. For a typical trial, a variation in vertical GRF of 34 N was observed over a 0.02 s period, 0.01 s either side of the time of maximum Achilles tendon force occurrence. This corresponds with a 2% variation in GRF, resulting in a 2% variation in calculated ankle moment. Achilles tendon force may therefore vary by 2%, or approximately 0.2 BW, due to error in synchronisation of video and forceplate data. The influence of synchronisation error has been reduced to approximately 0.06 BW by the use of mean values over 10 trials for each condition.

(v) Response to Questions

No support has been has been provided in the present study for the suggestion that raising the heel by use of heel lifts can reduce maximum Achilles tendon force. In contrast, increases in maximum Achilles tendon force were found for two of the subjects. It has been demonstrated that distinct subjects respond differently to heel lift interventions, with the running style of the forefoot striker being implicated as the likely cause of the different behaviour of this subject compared with the remaining two subjects. Additionally, this subject was found to experience much larger maximum Achilles tendon force values than the rearfoot striker and midfoot striker.

The results of the present study provide only minimal support for the suggestion that the maximum amount of ankle dorsi-flexion can be reduced by the raising of the heel. No changes in peak Achilles tendon angle were demonstrated across conditions for the rearfoot striker and the forefoot striker. The midfoot striker showed an increased Achilles tendon peak angle for both heel lift conditions compared with the barefoot condition. Since the angle was not increased for the 15 mm heel lift compared with the 7.5 mm heel lift, and in fact was reduced, it is suggested that it is not clear whether the difference between the barefoot and heel lift conditions is due to the attachment of a material to the heel, or the raising of the heel relative to the forefoot. Increased control of the heel lift conditions is required before conclusions can be made regarding the influence of heel lift on maximum ankle dorsi-flexion.

The relationship between ankle angle and Achilles tendon moment arm length obtained in the present study has been compared to those presented in the literature. In particular, the methods of Bobbert et al. (1986) and Burdett (1982) have been adopted in the present study. The main difference between moment arm lengths obtained using the methods of the present study and the literature methods was the magnitude of the lengths. Generally, similar shaped plots over stance have been presented. Despite the scaling methods employed using the procedures of Burdett (1982) and Bobbert et al. (1986), it may be that the use of female distance runners as subjects, with small body mass and dimensions, has resulted in the contrast in magnitudes. This is supported by the finding that the maximum Achilles tendon force values calculated using the moment arm lengths of the present study are comparable with those reported in the literature. It is possible that the marker system employed in the present study for representation of Achilles tendon line of action is not appropriate for representation of the tendon location relative to the ankle joint centre. Although this would have a predominantly systematic influence on Achilles tendon forces, the absolute Achilles tendon moment arm lengths would be incorrect. A method of measuring Achilles tendon moment arm lengths for a range of ankle joint angles is required to investigate the validity of marker methods for representation of Achilles tendon line of action. Magnetic resonance imaging has been demonstrated to be suitable for this purpose (Rugg et al., 1990).

The methods developed in Chapter 3 have facilitated the detection of differences in maximum Achilles tendon forces with varying heel height in running. The likely influence of random error on results has been minimised by the use of a large foot image size and a high quality, high resolution frame store. Since the observed effect sizes for the rearfoot and midfoot strikers are larger than the variations resulting from random errors, the calculated level of precision is acceptable for detecting differences between conditions. Although studies comparing absolute Achilles tendon force magnitudes on different occasions may not be carried out, longitudinal studies into changes in response to particular heel lifts are possible. For example, it would be possible to use the methods developed in the present study to test the response of an individual to specific changes in heel lift before and after a month of intense stretching exercises on the triceps-surae muscle group. Increased flexibility may influence the response to particular heel height conditions. The methods developed may, therefore, be useful in a clinical environment where the influence of a particular treatment on maximum Achilles tendon force values for a particular subject is of interest. Further investigation into the influence of heel height variations on maximum Achilles tendon force for a larger number of subjects using the methods of the present study, may reveal trends in response for runners with particular running styles.

CHAPTER 5 EVALUATION USING MAGNETIC RESONANCE IMAGING

5.1 Introduction

The procedures adopted and developed during the course of this research have been evaluated in earlier chapters by the use of sensitivity analyses. The variables influencing ankle moments and Achilles tendon force values have been identified, and the expected error in each of these variables has been quantified. The subsequent influence of these errors on calculated moments, forces and angles has been determined. These procedures have allowed the identification of the confidence with which the respective variables have been estimated for each individual trial. The sensitivity analysis approach has provided a thorough method of quantifying the precision in the estimated variables, but limited knowledge has been gained regarding the accuracy with which variables have been obtained. This has not been of major concern, since the focus of the work has been on the comparison of magnitudes of variables between conditions, rather than obtaining absolute values.

Although absolute ankle moment and Achilles tendon force values are not required for comparison across conditions, it would clearly be beneficial to gain knowledge of these magnitudes. This would facilitate comparison with results presented in the literature, and would allow quantification of the loading of the Achilles tendon in relation to ultimate loading characteristics obtained from isolated tendon specimens. The variables influencing quasi-static ankle moment values have been identified as the GRF resultant force and point of application, and the location of the ankle joint centre. The accuracy with which GRF variables are attained using the force plate of the present study has been discussed and quantified in earlier chapters. With the confidence in GRF data accounted for, the accuracy in the identification of the ankle joint centre of rotation is required in order to obtain an accuracy level for the calculated ankle moment values. Throughout the present research, the most prominent point of the lateral malleolus has been used to represent the location of the ankle joint centre. The actual location of the ankle joint centre of rotation has been reported in the literature using cadaver data (Hicks, 1953), and using scanning techniques such as magnetic resonance imaging (Rugg et al., 1990). Contrasting results have been presented across studies, limiting the possibility of adopting a previously reported location for use in the present study. The location of the ankle joint centre has also been found to vary throughout the range of ankle joint movement (Rugg et al., 1990). The evidence from the literature indicates that the location of the ankle joint centre is variable across subjects, necessitating the individual measurement of this location if accurate results are to be obtained. In the present chapter, magnetic resonance imaging techniques are used to identify the location of the ankle joint centre of rotation for the subject that has been common to all studies of this research. The locations of the ankle joint centre relative to the talus are obtained for distinct angles covering the range of ankle dorsi- and plantar-flexion for this subject. These locations are compared with those obtained using a marker on the skin at the most prominent point of the lateral malleolus.

The Achilles tendon forces obtained using the procedures described in previous chapters are influenced by the accuracy of the ankle moment and the moment arm of the Achilles tendon. It has been clearly demonstrated in the literature that the Achilles tendon moment arm varies according to ankle and knee angles, with the ankle angle being most influential (Burdett, 1982; Rugg et al., 1990). It has been demonstrated that the length of the Achilles tendon moment arm is subject specific (Burdett, 1982; Brand, 1982). In earlier chapters, the line of action of the Achilles tendon has been identified by markers placed on the skin covering the Achilles tendon on the visually identified line of action. The accuracy with which the line of action is identified using this method is investigated in the present chapter using magnetic resonance imaging techniques. The moment arm lengths obtained using this procedure are compared with those obtained using skin markers.

It has been noted in earlier chapters that there is a possibility of skin movement during running affecting the location of the skin markers in relation to the anatomical landmarks they represent. The extent to which skin movement causes the ankle and Achilles tendon markers to move relative to the talus and Achilles tendon respectively, is investigated in the present chapter using magnetic resonance imaging.

The cross-sectional area of the Achilles tendon is required to estimate stress values from Achilles tendon forces. In the present chapter, cross-sectional images of the tendon along its length are presented. These are used to calculate Achilles tendon cross-sectional area, to combine with maximum Achilles tendon force data obtained in earlier chapters for the estimation of maximum Achilles tendon stress in running.

The questions addressed in the present study are:

- Are accurate anatomical data obtained using skin markers ?
- Does skin movement influence the results ?

5.2 Methods

A marker was placed on the most prominent point of the lateral malleolus and two markers on the skin covering the Achilles tendon on the visually identified line of action, simulating the marker placement used in earlier studies. The markers employed were cod-liver oil capsules which showed up clearly on the scan obtained using magnetic resonance imaging (MRI). They were attached using double-sided adhesive tape. The chosen subject was the rearfoot striker common to all chapters of this research. The subject, a rearfoot striker, adopted a prone position with her right foot supported within a padded enclosure. Full range of ankle joint dorsi- and plantar-flexion movement was possible, with support beneath the heel and on the medial and lateral sides of the ankle joint. For the MRI scans, the subject was moved inside the MRI tube, and was initially asked to adopt a relaxed position of the ankle and foot. Scans were obtained for this position in the three orthogonal anatomical planes to ensure the lower extremity was orientated suitably for collection of sagittal, frontal and transverse plane images. Each of the images obtained represented a slice through 10 mm of tissue.

For collection of data over a range of joint angles, the subject was instructed to move the ankle from a position of maximum dorsi-flexion to maximum plantar-flexion in six distinct steps. No measurements were taken of ankle angles during data collection, and thus no direct control was exerted over the absolute joint angles adopted. The subject was required to hold each ankle angle position for approximately one minute, allowing collection of all the required images. There was a period of approximately ten seconds between each ankle position, providing adequate time to take up the next orientation.

Sagittal plane images were used for the calculation of the centre of rotation of the ankle joint and line of action of the Achilles tendon. Images parallel to these and through the skin markers were used to identify marker locations in the sagittal plane. It was assumed that all movement of the ankle joint occurred in this plane (Siegler et al., 1988). Movement was restricted predominantly to this plane by the support provided around the joint during data collection.

Ankle joint centre of rotation was calculated for each ankle rotation using the methods described by Rugg et al. (1990). These methods were adapted from those of Reuleaux (1967), and are illustrated in Figure 5.1. Tracings were made of the outline of the talus and the lower section of the tibia for the sagittal plane image for each distinct ankle angle. Under Position One (maximum dorsi-flexion), a point was marked 12 cm (image distance) proximal to the distal surface of the tibia along the longitudinal axis of the bone (point A). A second point (B) was marked on a line perpendicular to the longitudinal axis through the intersection of the longitudinal axis and the distal articular surface of the tibia, 12 cm from the intersection point. The Achilles tendon line of action was marked as a straight line through the tendon centre, as proposed by Rugg et al., (1990). Reliability of the line of action was assessed by performing five separate repeated identifications of this line. Since it was not possible to compare the chosen lines with the actual line of action, lines were purposefully drawn to encompass the range of orientations considered to be possible for representation of the tendon. The subsequent influence on Achilles tendon moment arm lengths was assessed using RMSD over the five measurements.

To identify the ankle joint centre for each ankle rotation, the talus was superimposed on the previous image and the tibia position for this image was drawn in. The tracing containing points A and B in Position One was superimposed on each tibia position to allow identification of these marker locations on each image, providing points A' and B'. Additionally, the straight line representing the Achilles tendon was traced from the Position One drawing to ensure the consistent placement of this line relative to the tendon. For each rotation, the two points A and A' were joined by a straight line, and the two points B and B' were joined by a straight line. Perpendicular bisectors were identified for each of these lines, and the intersection of these was taken to be the centre of rotation of the joint. This procedure was applied to each rotation, providing individual centres of rotation for each joint rotation. Achilles tendon moment arm lengths were calculated using the perpendicular distance from the ankle joint centre to the line of action of the Achilles tendon for Position Two to Position Six inclusive. Scaling of image distances was achieved using a 10 cm length on each MR image. The magnitudes of changes in ankle angle were obtained by measuring the angle between straight lines joining the point A to the identified centre of rotation on consecutive tibia images (Figure 5.2).

All markings on the tracing were achieved by the use of a sharp pencil and a ruler and protractor. All tracings and measurements of rotation angle, centre of rotation and moment arm were performed three times and mean values used. The size of the images corresponded to 71% actual size. To provide a measure of the reliability of the measured angles, locations and distances, a selected rotation was analysed on five separate occasions and the RMS deviations of these variables were calculated.

For each position, the four corners of the MRI scan were marked on the tracing to allow superimposing on the corresponding image containing the markers, for identification of the ankle and Achilles tendon markers relative to the talus. For the marker images, the Achilles tendon line of action was defined as a straight line through the two Achilles tendon markers, as used in Chapter 3 and Chapter 4 (Figure 5.3). Achilles tendon moment arm lengths were calculated for each of the ankle joint positions using the skin markers, and were compared to the lengths obtained directly from MRI measurements.

Moment arm lengths obtained using a fixed ankle joint centre of rotation were also calculated. This allowed investigation into the influence of using a fixed ankle centre of rotation compared with a moving centre of rotation in calculation of Achilles tendon moment arm lengths. The fixed centre of rotation was calculated for movement between maximum dorsi-flexion and maximum plantar-flexion images, using the procedures already described.

Transverse MRI images were obtained along the length of the tendon for assessment of Achilles tendon cross-sectional area. The amount of skin movement influencing marker location relative to the talus was investigated by identifying the ankle marker location relative to the talus on each sagittal plane image, and superimposing each of the images.

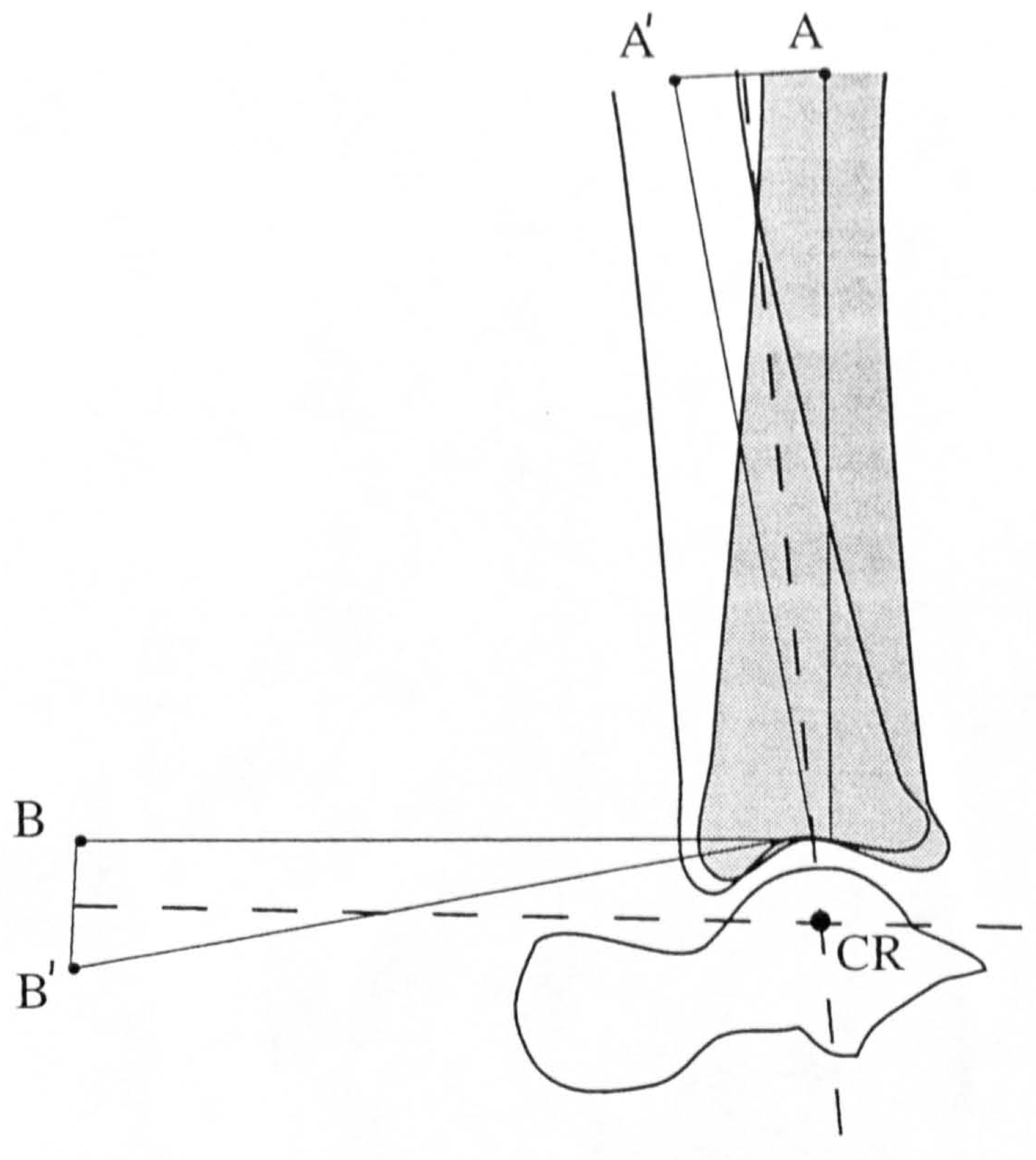


Figure 5.1 Procedure for calculation of centre of rotation (CR) using Reuleaux methods

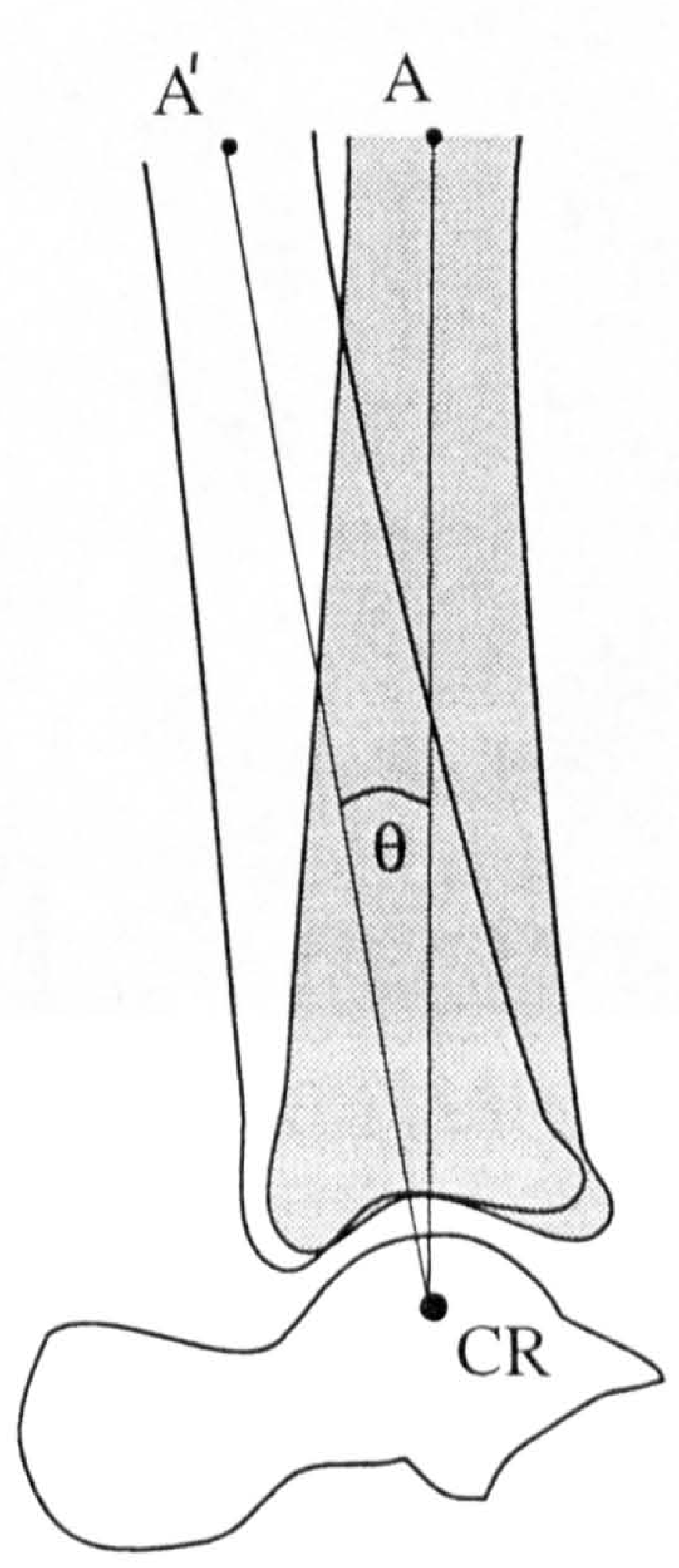


Figure 5.2 Procedure for calculation of rotation angle (θ)

Figure 5.3 Marker placement for Achilles tendon line of action approximation

length.

The fixed centre of rotation was found to be above and anterior to the ankle joint centre identified by the skin marker, by a distance of approximately 6 mm being largely in the anterior direction. The moving centre of rotation was also anterior to the location of the skin marker centre, and was generally above the marker location. The Achilles tendon line of action identified using the skin markers was approximately parallel to the line of action identified using MRI techniques, but was translated by approximately 6 mm in an anterior direction. Over the range of ankle angles used during MRI data collection, the orientation of the lines of action obtained using the skin markers and the MRI methods varied by a maximum of two degrees. The relatively close locations of the Achilles tendon moment arms obtained using each of the methods, resulted in negligible error occurring due to the lines of action not being exactly parallel.

Typical transverse images of the tendon cross-section along its length are illustrated in Figure 5.7. It is evident that the tendon changes shape along its length. The skin thickness was measured as 2.0 mm. The Achilles tendon cross-sectional area was approximately semi-circular at its thickest location at approximately 5 cm proximal to the point of insertion. The area of the tendon at this location was estimated as 77 mm^2, using a measured radius of 7 mm.

Table 5.1 Achilles tendon moment arm lengths obtained using a moving centre of rotation, fixed centre of rotation, and skin marker methods (mm)

position	angle change	moving CR	fixed CR	skin markers
1 - 2	7.7 (0.3)	43.6 (2.0)	43.6 (0.4)	34.1 (0.4)
2 - 3	10.8 (0.8)	46.7 (2.3)	47.4 (0.4)	36.4 (0.0)
3 - 4	6.7 (0.6)	48.5 (2.3)	49.5 (0.8)	39.2 (0.0)
4 - 5	9.2 (0.6)	50.4 (3.2)	51.6 (0.4)	41.8 (0.4)
5 - 6	12.5 (0.5)	51.7 (2.7)	54.8 (1.1)	44.8 (0.0)

Figure 5.4 Sagittal plane magnetic resonance images for each ankle angle

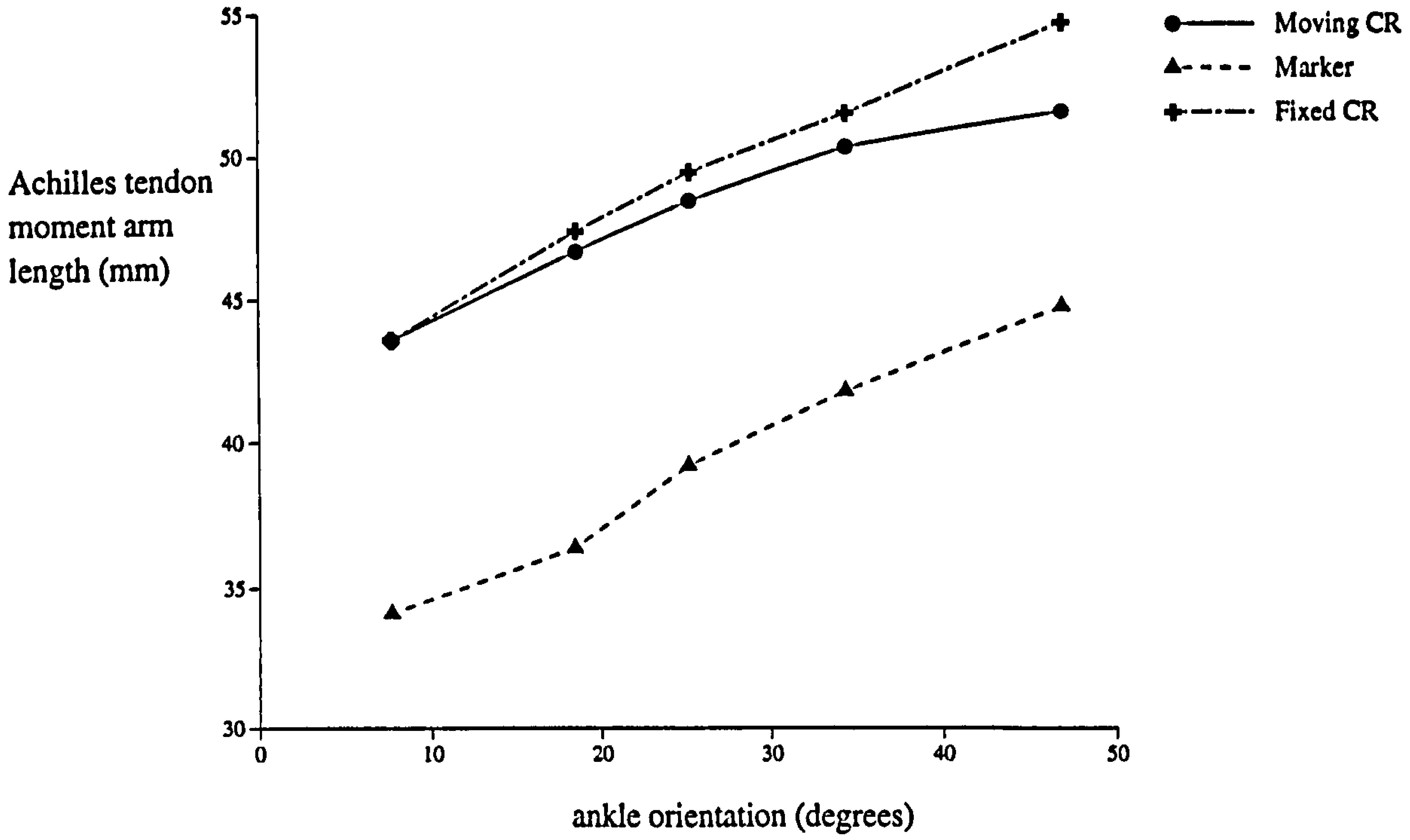


Figure 5.5 Achilles tendon moment arm against ankle orientation using each method

Figure 5.6 Location of fixed and moving centres of rotation relative to the talus

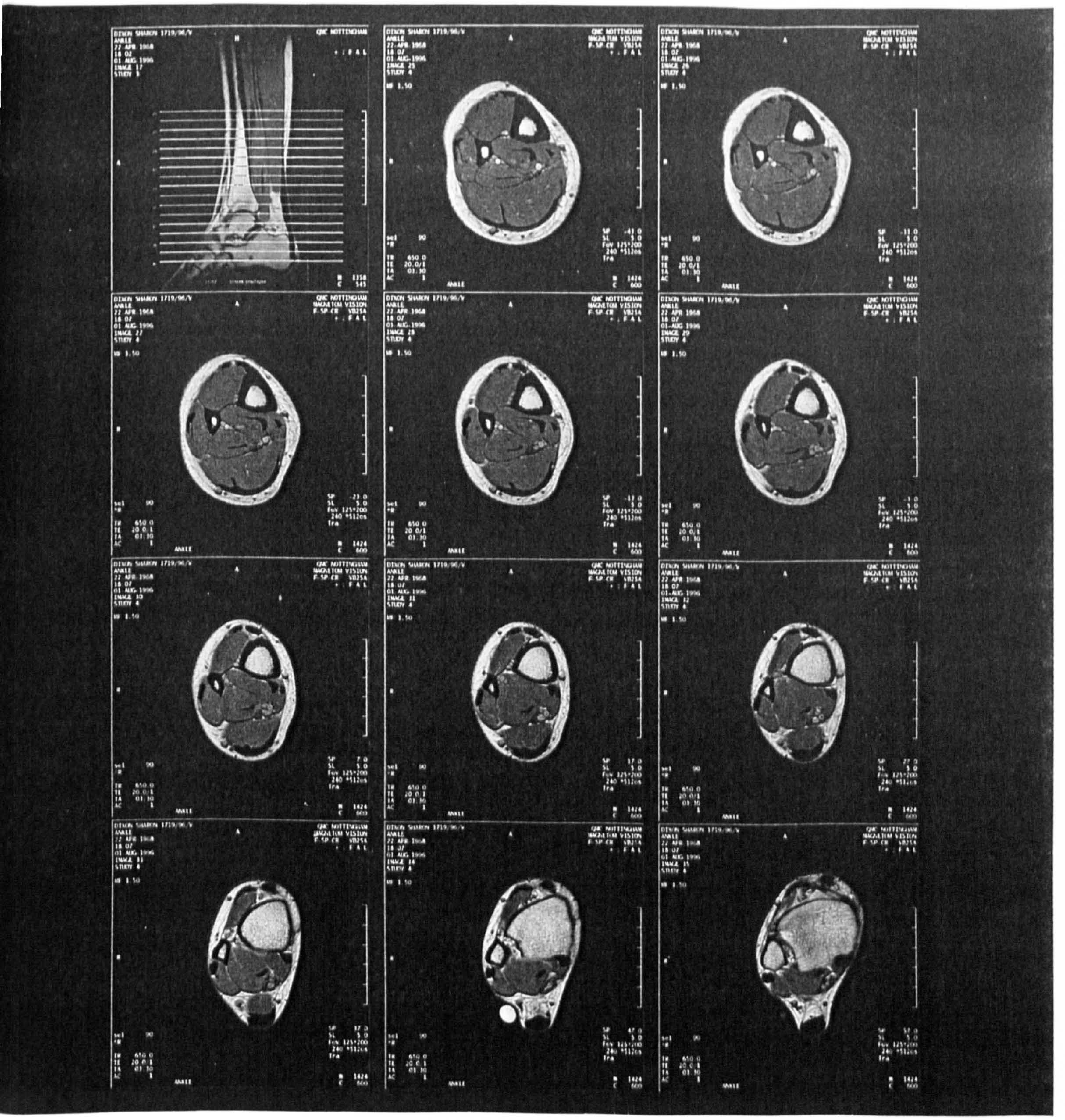


Figure 5.7 Typical transverse images showing the cross-section of the Achilles tendon

5.4 Discussion

The finding that the ankle joint centre of rotation identified using the skin marker has a more posterior location than that calculated using direct measurement from the MRI scans partly explains the discrepancy between Achilles tendon moment arm lengths obtained using the different methods. Since the Achilles tendon line of action is close to the vertical throughout stance, the anterior-posterior difference of approximately 6 mm contributes to the total length difference of, on average, 10.1 mm. The remainder of the difference appears to be due to the approximately 6 mm more anterior line of action of the Achilles tendon identified using the skin markers compared to that obtained using direct measurement from the MR images. Thus, the systematic error in Achilles tendon moment arm when using the skin markers can be attributed to the error in ankle joint centre location, and the error in Achilles tendon line of action.

The influence of these errors on absolute ankle moments and Achilles tendon forces has been investigated using the barefoot results for the rearfoot striker in Chapter 4. For an ankle moment of 159 N.m at the time of maximum Achilles tendon force, a decrease of 6 mm in the length of the moment arm of the GRF, resulting from the more anterior location of the ankle joint centre, results in a decrease in moment to 151 N.m. An increase in Achilles tendon moment arm by 10 mm from 31.9 mm to 41.9 mm, subsequently results in a decrease in estimated maximum Achilles tendon force from the value of 9.3 BW presented in Chapter 4 to 6.7 BW. These findings may be taken into account in future Achilles tendon force calculations for this subject by using a more appropriate ankle joint centre, and by using appropriately placed Achilles tendon markers determined by reference to the MRI scans. The most easily reproducible location to place markers on the skin covering the tendon is on the rear of the leg. It is therefore suggested that the tendon line of action may be monitored during ground contact by the placement of spherical markers on the rear of the lower leg, and translation of the straight line joining these two markers by an appropriate distance in the anterior direction. For the subject studied, the appropriate translation distance may be obtained directly from the MRI scan. For other subjects, it is likely that the geometry of the Achilles tendon is similar to that demonstrated. It is therefore suggested that the most appropriate translation distance for each subject may be obtained by scaling the MRI measurements for the subject studied in this chapter, using calliper measurements of Achilles tendon diameters. The suggested procedure is illustrated in Figure 5.8. At a location approximately 5 cm proximal to the Achilles tendon insertion, the distance from the outer surface of the skin covering the tendon to the approximated line of action was measured on the MRI scan as 3.9 mm. At this location along the tendon, a calliper measurement from the medial to the lateral side of the tendon may be taken for any subject. It is suggested that this distance be used in future work to obtain a scaled subject-specific distance from the skin covering the tendon to the line of action of the tendon.

The ankle joint centre location obtained using both the fixed and moving centres of rotation in the present study differs from those described in the literature. Scott and Winter

(1990) described the ankle joint centre of rotation as being 2.2 cm below the lateral malleolus, but the point of the lateral malleolus that was being referred to was not stated. Hicks (1953), described the location of an ankle joint axis which corresponded to a point of rotation anterior and above that represented by the point on the lateral malleolus in the sagittal plane, in agreement with the axis found in the present study. The contrast in locations of ankle joint centre of rotation found across studies may be due to several factors, including the existence of a moving centre of rotation (Rugg et al., 1990), the differing axes obtained for plantar-flexion and dorsi-flexion (Hicks, 1953), different techniques for calculation of the axis, and differences between subjects. The likelihood that differences in location of the ankle joint centre of rotation exist across subjects indicates that, for accurate determination of ankle moment magnitudes, and Achilles tendon moment arm lengths, individual joint centres must be calculated for each subject. The continued use of the most prominent point of the lateral malleolus to represent the centre of rotation of the ankle joint is recommended if individual anatomical data are not available.

It has been described in earlier chapters, that systematic errors in ankle joint centre and Achilles tendon line of action will not influence comparisons between conditions. It has been demonstrated in the MRI investigation that, although errors occur in the skin marker specifications of the ankle joint centre and the Achilles tendon line of action, the subsequent influence on maximum Achilles tendon force values and Achilles tendon loading rate is predominantly systematic. The deviation from a systematic error when maximum plantar-flexion angle is approached, will not influence comparisons of Achilles tendon loading rate and maximum Achilles tendon force across conditions, since these variables occur when the ankle angle is between a position close to neutral at ground contact and maximum ankle dorsi-flexion. Confidence in the findings of earlier chapters is therefore increased.

It has been demonstrated in the present chapter that the Achilles tendon cross-section changes shape along its length, with an approximately semi-circular shape being evident around the thickest portion of the tendon. The cross-sectional area calculated using a semi-circle can be used to obtain the maximum stress experienced by the tendon in running using the maximum force results obtained in Chapter 4. By assuming that all individuals have similar geometry of the Achilles tendon, stress values may be determined for other subjects by scaling of dimensions using calliper measurements of the tendon.

The marker methods used in earlier chapters have provided Achilles tendon moment arm lengths of approximately 30 mm. When a fixed rather than a moving centre of rotation was employed, the observed 1.0 mm variation in tendon moment arm length corresponded to a difference of approximately 3%. The MRI study has revealed that the marker system appears to under-estimate the length of the moment arm length, with values in the region of 45 mm being obtained using both the moving and the fixed centres of rotation. For this length, a variation of 1.0 mm has an influence on Achilles tendon moment arm of approximately 2%. For a typical ankle moment of 150 N.m and moment arm length of 45 mm, the result on maximum Achilles tendon force of using a fixed versus a moving centre of rotation would be a decrease from 6.2 BW to 6.0 BW. For the quantification of the absolute values of maximum Achilles tendon force, this difference was considered to be negligible. The finding that the difference in moment arm lengths obtained using a fixed and a moving centre of rotation of the ankle joint was negligible, is consistent with the findings of Rugg et al. (1990).

The finding that the variations in Achilles tendon moment arm length obtained over three repeated measurements was not reduced by the use of five repeated measurements supports the use of the mean of three measurements in the present study. Further repeated digitisations do not appear to reduce the influence of random error on resulting moment arm lengths.

The finding in the present study that for each ankle angle the centre of rotation locations identified using the skin marker were identical, indicates that the skin movement over the underlying bone during ankle joint movement has a negligible influence on the ankle joint centre estimated using a skin marker. This result was for a non weight-bearing condition, but provides an initial indication of the appropriateness of a skin marker to identify the ankle joint centre of rotation.

The evaluation of skin marker methods for the prone subject has demonstrated that there is a systematic error in lengths obtained using the procedures of previous chapters. Methods for accounting for this systematic error have been described. It is suggested that the difference in running moment arm lengths and the lengths obtained using skin markers with the subject in a prone position, will be similar to the difference in actual moment arm lengths between these two conditions. MRI scans obtained during running are required to support this suggestion, but facility to perform this form of data collection was not available. It was therefore concluded that the prone MRI data be used for scaling in future work, until dynamic Achilles tendon moment data during running can be obtained.

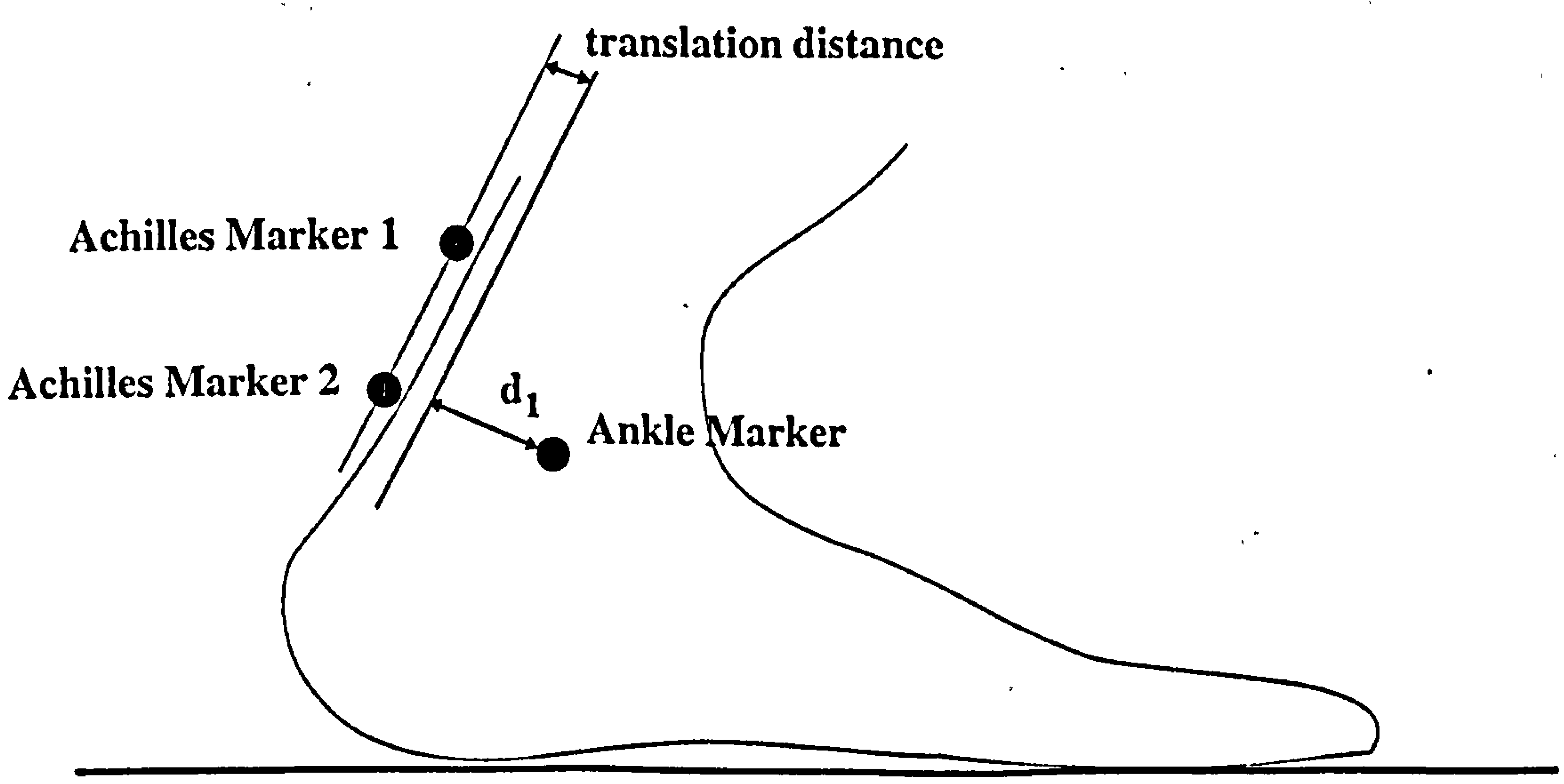


Figure 5.8 Illustration of recommended procedures for the calculation of moment arms

CHAPTER 6 THE INFLUENCE OF HEEL LIFT ON ACHILLES TENDON LOADING IN REARFOOT STRIKERS

6.1 Introduction

In contrast with suggestions in the literature (Clement et al., 1984), it was found in the study of Chapter 4 that, rather than being decreased, maximum Achilles tendon force was increased when the heel was raised relative to the forefoot. The increase in maximum Achilles tendon force observed for the rearfoot striker was in contrast with the decrease found in Chapter 3 when shock absorbing heel lifts were employed. The conditions used in Chapter 3 resulted in both increased heel lift and increased shock absorption, limiting the interpretation of results. In the present chapter, the influence of the controlled variation of shock absorption on maximum Achilles tendon force is investigated for this subject.

In Chapter 4, three runners demonstrating distinctly different running styles were studied. Differences in response across styles were detected, but the generalisation of results to the remainder of the population was limited. In the present chapter, a group of subjects with the same running style have been studied, allowing investigation into possible differences and similarities within the group. A rearfoot strike running style was chosen, since this has been demonstrated to be the technique used by 80% of distance runners (Cavanagh and Lafortune, 1980). Additionally, the results of earlier chapters indicate that maximum Achilles tendon force can be influenced by heel lift variations for subjects with this style. The possibility of the method of attachment of lifts to the barefoot influencing results was highlighted in Chapter 4. In the study of the present chapter, an alternative baseline condition under which the entire foot height is raised is used.

The GRF impact peak immediately following ground contact observed for the majority of runners, has been suggested to be associated with injury to the lower extremity (Miller, 1990). The loading rate of vertical GRF has been suggested to be an indicator of the amount of shock absorption provided beneath the heel (Lees and M^cCullagh, 1984). In Chapter 3, the attachment of shock absorbing heel lifts to the plantar surface of the foot was observed to influence the loading rate of impact force. For each of the interventions applied in the present chapter, GRF variables are presented to indicate whether increased shock absorption has been provided.

Tendon has been shown to be a viscoelastic material, and is therefore sensitive to changes in loading rate (Hawkins, 1993). Mechanical tests on isolated tendon samples have provided contrasting results regarding the influence of loading rate on the properties of tendon (Abrahams, 1967; Ker, 1981). In the present chapter, methods for the estimation of maximum Achilles tendon loading rate during running are developed. The results of an investigation into the influence of heel lift manipulation on this variable are presented.

The evaluation presented in Chapter 5 has revealed a systematic error in the Achilles tendon line of action when markers are placed on the skin covering the Achilles tendon to represent the visually identified line of action. In the present chapter an alternative technique involving the scaling of the MRI anatomical data obtained in Chapter 5 is employed.

In previous chapters, the use of 50 Hz kinematic data has been justified due to the analysis being limited to the midstance phase. An increased sampling rate is required for analysis of Achilles tendon loading rate, since indications are that this variable occurs early in the stance phase. An opto-electronic automatic tracking system is used in the present chapter to collect kinematic data at 120 Hz.

To estimate the stress that the Achilles tendon is subjected to during running, it is necessary to obtain a measure of the cross-sectional area. The use of MRI techniques to obtain transverse images of the Achilles tendon cross-sectional area has been described in Chapter 5. Scaling methods are developed in the present chapter to estimate Achilles tendon cross-sectional area for each subject, allowing the determination of the maximum Achilles tendon stress.

In summary, the main questions addressed in the present chapter are:

- Can a controlled increase in shock absorption reduce the maximum Achilles tendon force ?
- How does heel lift influence the Achilles tendon loading of rearfoot strikers ?

6.2 Methods

(i) General Procedures

Two weeks prior to data collection, prospective subjects performed preliminary barefoot running trials in the laboratory to allow familiarisation with the environment, and for identification of suitable rearfoot striking subjects. Force plate and video data were collected for running trials as described in earlier studies, facilitating the identification of rearfoot striking subjects.

Eight female distance runners demonstrating a rearfoot ground strike were selected for study. One of the subjects (Subject 7) was the rearfoot striker employed in earlier chapters. Seven of the subjects were elite athletes performing in excess of 50 miles per week at the time of data collection, with mean age 23.3 years (SD of 3 years). The one remaining subject was 46 year's old, and was a recreational runner performing approximately 20 miles per week. Before data collection, each subject's height and mass were recorded, and measurement of the Achilles tendon performed using callipers. These measurements were taken from the Achilles tendon of standing subjects at a height of approximately 5 cm above the insertion, since this was the location which was most easily accessible for measurement. At a position along the length of the tendon corresponding to the 5 cm height at which calliper measurements were made, a cross-sectional image of the Achilles tendon was presented in Chapter 5 for Subject 7. The cross-section at this location was approximately semi-circular with radius 7 mm and area 77 mm^2. The calliper measurements obtained in the present chapter for each subject were used to obtain scaled cross-sectional area estimates from the Subject 7 data. The reliability of calliper measurements was assessed by taking repeated measurements for one subject at separate testing sessions.

Subject 7 performed 10 running trials at 3.83 $m.s^{-1}$ under each of six conditions. Conditions were:

A	barefoot
B	firm lift of 7.5 mm thickness attached to the rearfoot and the forefoot
C	as Condition B, with an additional 7.5 mm heel wedge on top of the rearfoot lift
D	as Condition B, with an additional 15 mm heel wedge on top of the rearfoot lift
E	as Condition B, with an additional 6 mm Sorbothane lift on top of both the rearfoot and forefoot lifts
F	as Condition E, with an additional 15 mm heel wedge at the rearfoot

The remaining seven subjects performed 10 running trials at 3.83 $m.s^{-1}$ under conditions A, B, C and D. High density EVA lifts, as used in the Chapter 4 study, were attached to the foot using surgical tape, as previously described. The randomised block design for assigning conditions to trials adopted for the study of Chapter 4 was not considered necessary, due to the finding that no significant differences were identified in the Chapter 4 study within each condition over the testing session. Ten consecutive trials were performed for each condition, with the order of the four conditions randomised separately for each subject.

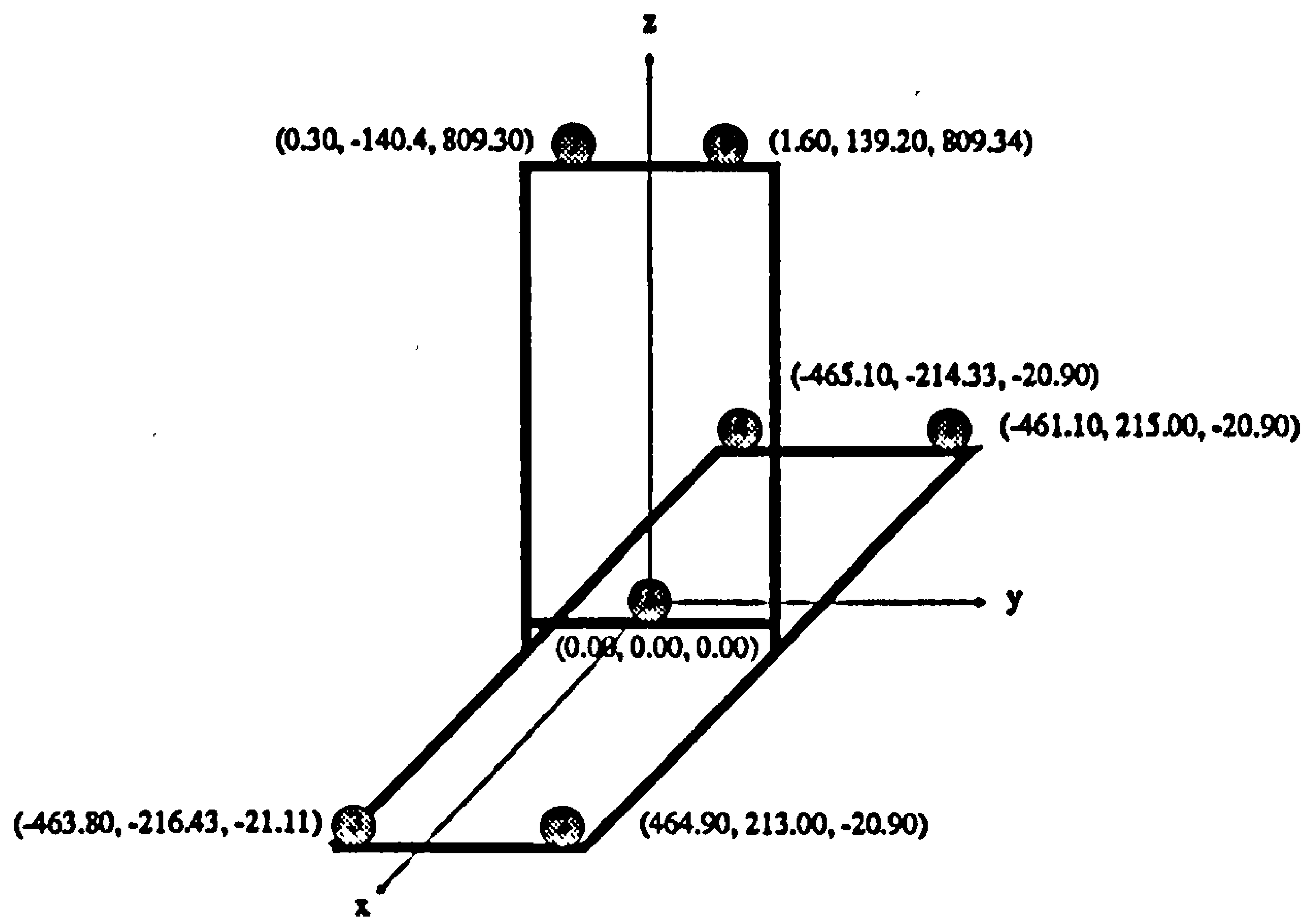
Force plate and sagittal plane video data were collected, and two-dimensional calibration and reconstruction performed, as described in previous chapters. Three-dimensional kinematic data were collected at 120 Hz for each running trial using an automatic opto-electronic tracking unit (MacReflex, Qualisys AB, Sweden). The MacReflex data were obtained using four cameras, positioned as illustrated in Figure 6.3. Data were collected for the ground contact phase and a period of approximately 0.1 s prior to and following ground contact. A calibration frame supplied by the manufacturer with external dimensions of approximately 0.90 x 1.80 x 1.60 m was used to calibrate the volume directly above the force plate (Figure 6.1). These dimensions are defined in the x, y, z directions using a convention with x normal to the plane of progression and parallel to the force plate top surface, y normal to the x-axis and parallel to the line of progression, and z normal to the x and y axes with positive being upwards. The force plate data were provided relative to an origin at the centre of the force plate. The kinematic data obtained using the MacReflex system were provided relative to an origin at Marker 5 of the calibration frame. To combine the force plate and kinematic data for estimation of joint moments, the kinematic data were transformed such that the coordinates were presented relative to an origin at the centre of the force plate. The location of the centre of the force plate was identified during calibration by the placement of markers in threaded bolt holes on the force plate corners. During collection of movement data, markers were placed at three of the four corner points, allowing easy identification of the orientation of the force plate. The midpoint of a line joining markers at diagonally opposite corners was used to locate the force plate centre. The axes orientation and the location of markers on the force plate are illustrated in Figure 6.2.

For sagittal plane calculations, the yz coordinates obtained from the three-dimensional MacReflex data were used, eliminating error due to projection onto a plane, as occurs when using traditional two-dimensional techniques. The influence of using three-dimensional data to obtain two-dimensional coordinates was assessed theoretically by projecting the MacReflex coordinates onto the yz plane of the force plate, and quantifying the influence on estimated Achilles tendon force values. The sagittal plane was defined as the yz plane of the force plate, resulting in the x coordinate being zero for all calculated two-dimensional locations. During data collection, the video camera was located 5.3 m and 0.78 m from the centre of the force plate, in the x and z directions respectively. The influence of projection on a typical set of marker locations was investigated for a randomly selected trial.

Reflective spherical markers of diameter 10 mm, supplied by the manufacturer of the MacReflex system, were used for identification of body landmarks. These were attached to the skin of the subject using strips of double-sided adhesive tape. Eight body landmarks and three points on the force plate were identified for each movement field.

The following markers were used:

marker number	location
1	hip
2	knee
3	ankle
4	MTP
5	Achilles marker 1
6	Achilles marker 2
7	calcaneus marker 1
8	calcaneus marker 2
9	force plate marker 1: co-ordinates (0.80, 1.20, 0.00)
10	force plate marker 2: co-ordinates (0.80, -1.20, 0.00)
11	force plate marker 3: co-ordinates (-0.80, -1.20, 0.00)



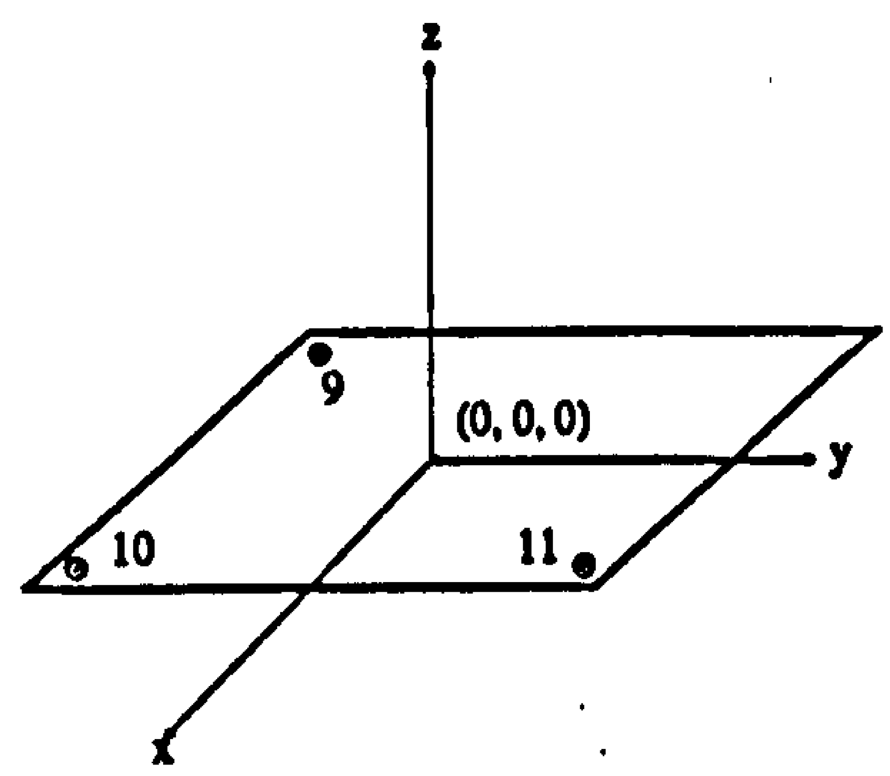


Figure 6.1 Calibration frame for MacReflex system with axes orientation

Figure 6.2 Force plate

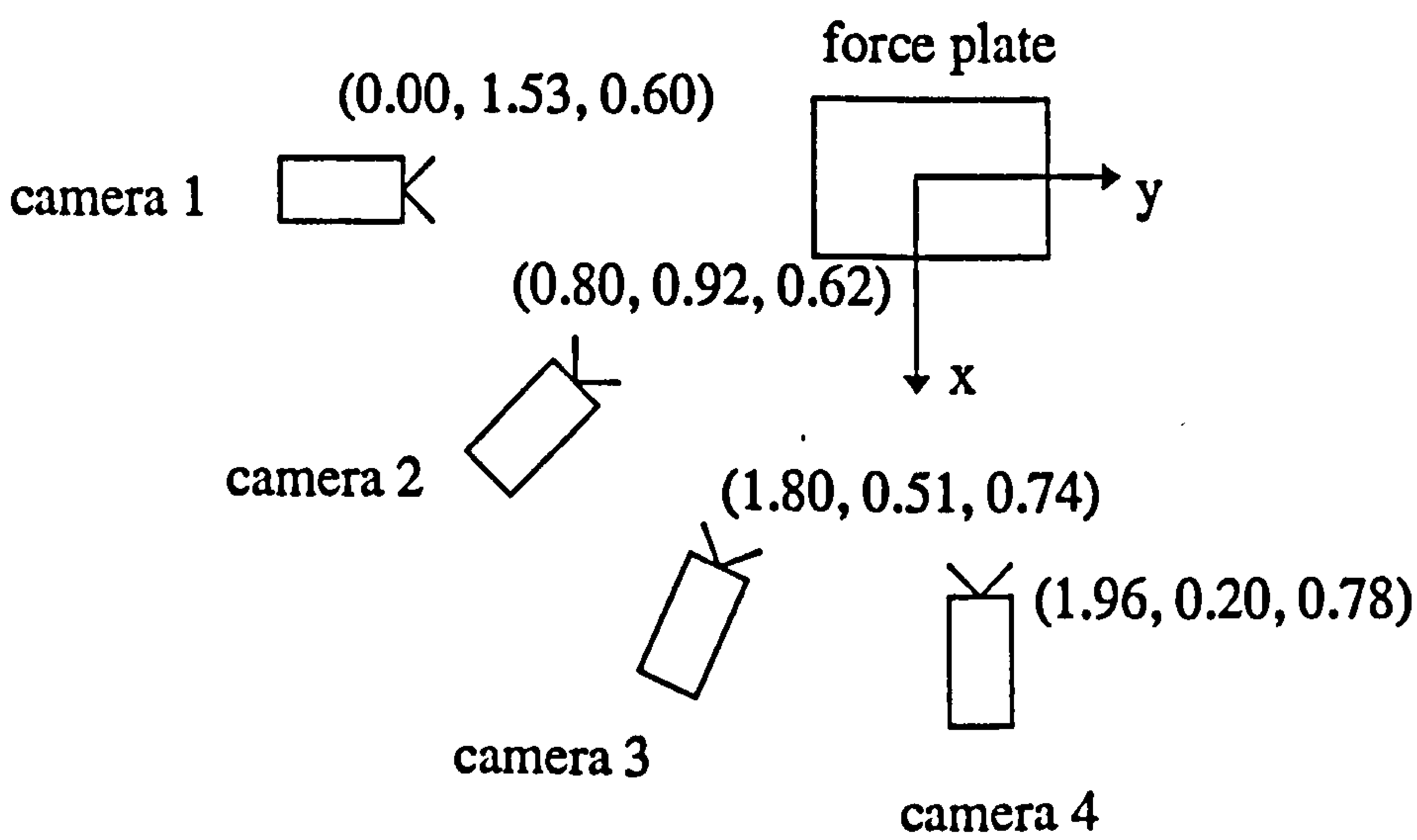


Figure 6.3 A plan view of the MacReflex camera placement

Marker locations are illustrated in Figure 6.4. The hip, knee, ankle and MTP markers were attached to represent the respective joint centres of rotation. The hip marker was placed at the location of the superior border of the greater trochanter, and the knee marker was placed to represent the visually identified location of the knee joint centre. The ankle marker was placed on the most prominent point of the lateral malleolus. The MTP marker was placed to represent the metatarsalphalageal joint centre for the fifth metatarsal. Markers Achilles 1 and Achilles 2 were placed on the rear of the lower leg to represent the line of action of the Achilles tendon as viewed in the frontal plane from the rear. The markers on the calcaneus were placed to form a visually identified vertical line on the skin covering the rear of the calcaneus bone, when the subject was in a relaxed standing position. Marker numbers 9, 10, and 11 were the three markers placed on the corners of the force plate, as described previously.

A standing calibration was performed for each subject under each condition, by recording of kinematic data whilst the subject adopted a standardised position. This position required each subject to stand with their second toe and the calcaneus 2 marker placed on the centre line of the force plate in the anterior-posterior direction (y axis), and the MTP marker placed on a line normal to this through the centre of the force plate (x axis). The repeatability of this standing calibration position was investigated by performing three separate calibrations for one subject. Foot length was defined as the distance in the sagittal plane from the calcaneus 2 marker to the MTP marker when standing in the barefoot calibration position.

The force data and kinematic data were collected simultaneously. Synchronisation of force plate and video data was as described in previous chapters. As the first photocell beam was broken, additional to triggering the start of force plate data collection and an LED in the video field of view, an infrared LED in the field of view of one of the MacReflex cameras was triggered, allowing synchronisation of all data.

1000 Hz kinematic data were obtained by interpolation of the 120 Hz MacReflex kinematic data. Prior to interpolation, the 120 Hz kinematic data were smoothed using a cubic spline (Reinsch, 1967). The closeness of fit between the raw data and the spline was controlled by defining a parameter value, where zero would provide a spline passing through all the raw data points. It was assumed that the force data at zero time corresponded to zero time for the 1000 Hz kinematic data. Since it was possible that the LED was actually triggered within the 120 Hz kinematic data field immediately prior to the identified one, but was not detected because of a short exposure time of 0.25 ms, the influence of synchronisation error was investigated using a sensitivity analysis. The influence of the worst possible synchronisation error of 0.00805 s (1/120 s - 0.00025 s) on estimated variables was investigated.

For each trial, an Excel template spreadsheet was used for synchronisation of MacReflex and force data. All subsequent calculations were carried out in this spreadsheet (Appendix G). For each subject, a summary spreadsheet was set up containing the results for all variables. All statistical analyses and graphs were also contained in this summary spreadsheet (Appendix H).

Joint and segment angular velocities and accelerations were obtained using finite difference methods. A two point finite difference algorithm (Equation 6.1) was used to obtain velocity values (Miller and Nelson, 1973). The influence of using three point (Equation 6.2), five point (Equation 6.3) and nine point (Equation 6.4) finite difference algorithms on acceleration profiles was investigated. The five point and nine point equations used were those provided by Lanczos (1957, cited by Lees, 1980).

$$\text{velocity} = (x_{i+1} - x_{i-1}) / 2t \tag{6.1}$$

$$\text{acceleration} = (x_{i+1} - 2x_i + x_{i-1}) / t^2 \tag{6.2}$$

$$\text{acceleration} = (-x_{i+2} + 16x_{i+1} - 30x_i + 16x_{i-1} - x_{i-2}) / 12t^2 \tag{6.3}$$

$$\text{acceleration} = (4x_{i+4} + 4x_{i+3} + x_{i+2} - 4x_{i+1} - 10x_i - 4x_{i-1} + x_{i-2} + 4x_{i-3} + 4x_{i-4}) / 100t^2 \tag{6.4}$$

t = time interval between two points
n = a number of points
x_i = a defined point
x_{i+n} = is the point $(n*t)$ seconds after x_i
x_{i-n} = is the point $(n*t)$ seconds before x_i

(ii) GRF Calculations

Average vertical GRF traces over ten trials for each subject/condition combination were obtained by interpolation of vertical GRF to provide 100 data points, allowing normalisation across trials using percentage of total stance time. The peak impact forces were identified for each separate trial and the average of the ten peak values for each subject/condition combination was calculated. A similar procedure was used to obtain mean maximum active vertical GRF for each subject/condition combination.

Average loading rate of vertical GRF during impact was calculated for each trial by dividing the impact peak by the time from initial ground contact to impact peak occurrence. Maximum instantaneous GRF loading rate was calculated by consideration of the change in GRF over each consecutive 0.001 s time interval.

Total ground contact time was obtained for each trial using vertical GRF data. Inspection of the vertical GRF data indicated that the vertical component showed a clear increase above a stable threshold when ground contact occurred. During the time when there was no contact with the force plate, the vertical component of GRF did not rise above 14 N. Thus, ground contact was defined as the time during which vertical GRF was 15 N or above.

(iii) Joint Moment Calculations

Two-dimensional, sagittal plane moments about the ankle joint were determined using inverse dynamics procedures (Alexander and Vernon, 1975; Appendix F). The mass of the segments and the location of segment centres of mass were determined using the regression equations derived by Zatsiorsky and Seluyanov (1985), in which subject mass and height were substituted for each individual (Equation 6.5; Equation 6.6). Segmental moments of inertia about transverse axes through segment centres of mass were also calculated using the regression equations provided by Zatsiorsky and Seluyanov (1985; Equation 6.7). The linear velocity and acceleration components in the y and z directions were determined for the segmental centres of mass, using finite difference methods, as previously described.

Ankle joint moments were also calculated using quasi-static methods (Morlock and Nigg, 1988), and were compared with those obtained using inverse dynamics.

Sagittal plane peak joint moment values and times of occurrence were identified, and typical plots obtained for each subject/condition combination. The influence on peak ankle moments of using sets of inertia data provided in the literature by Whitsett (1963), Dempster (1955), Clauser et al. (1969) and Chandler et al. (1975) was investigated.

$$\text{mass} = -0.829 + 0.0077X_1 + 0.0073X_2 \quad (6.5)$$

$$\text{centre of gravity} = 3.767 + 0.065X_1 + 0.033X_2 \quad (6.6)$$

$$\text{transverse moment of inertia} = -97.09 + 0.414X_1 + 0.614X_2 \quad (6.7)$$

X_1 = body mass (kg)
X_2 = body height (cm)

(iv) Achilles Tendon Loading Calculations

The Achilles tendon line of action for use in Achilles tendon force calculations was determined using the tendon skin markers (Figure 6.4). The Achilles tendon line of action was estimated as being parallel to a line through the centre of the two Achilles tendon markers, displaced towards the ankle marker in the yz plane (sagittal plane) by a distance determined by the radius of the Achilles tendon markers (5 mm), and scaled MRI tendon cross-section data obtained for Subject 7 (Chapter 5). The distance measured for Subject 7 from the skin surface to the tendon line of action in the sagittal plane was 3.9 mm (Chapter 5). Tendon calliper measurements for each subject were used to provide scaled translation distances using the Subject 7 data.

Achilles tendon forces were determined throughout the stance phase of running for each trial and subject, using the methods described in Chapter 3. Achilles tendon moment arm and Achilles tendon force time histories were obtained. Maximum Achilles tendon forces were obtained over the 10 trials for each subject/condition combination. Magnitudes of maximum stress that the Achilles tendon was subjected to were obtained for barefoot trials for each subject by division of the maximum Achilles tendon force by the estimated cross-sectional area.

The variables contributing to maximum Achilles tendon force were identified with reference to the hierarchical diagram provided in Figure 4.2. The magnitude of these variables at the time of maximum Achilles tendon force occurrence were identified, allowing investigation into the factors most influential in determination of maximum Achilles tendon force values.

Methods for monitoring of Achilles tendon loading rate were developed using sample Achilles tendon force data obtained in the present study. Instantaneous Achilles tendon loading rates were calculated over each 0.001 s time period. Smoothed Achilles tendon loading rate data were obtained by calculation of the mean Achilles tendon loading rate values obtained over consecutive 0.01 s time periods. Average Achilles tendon loading rate was calculated by division of the maximum Achilles tendon force by the time taken to reach this force, for each subject/condition combination.

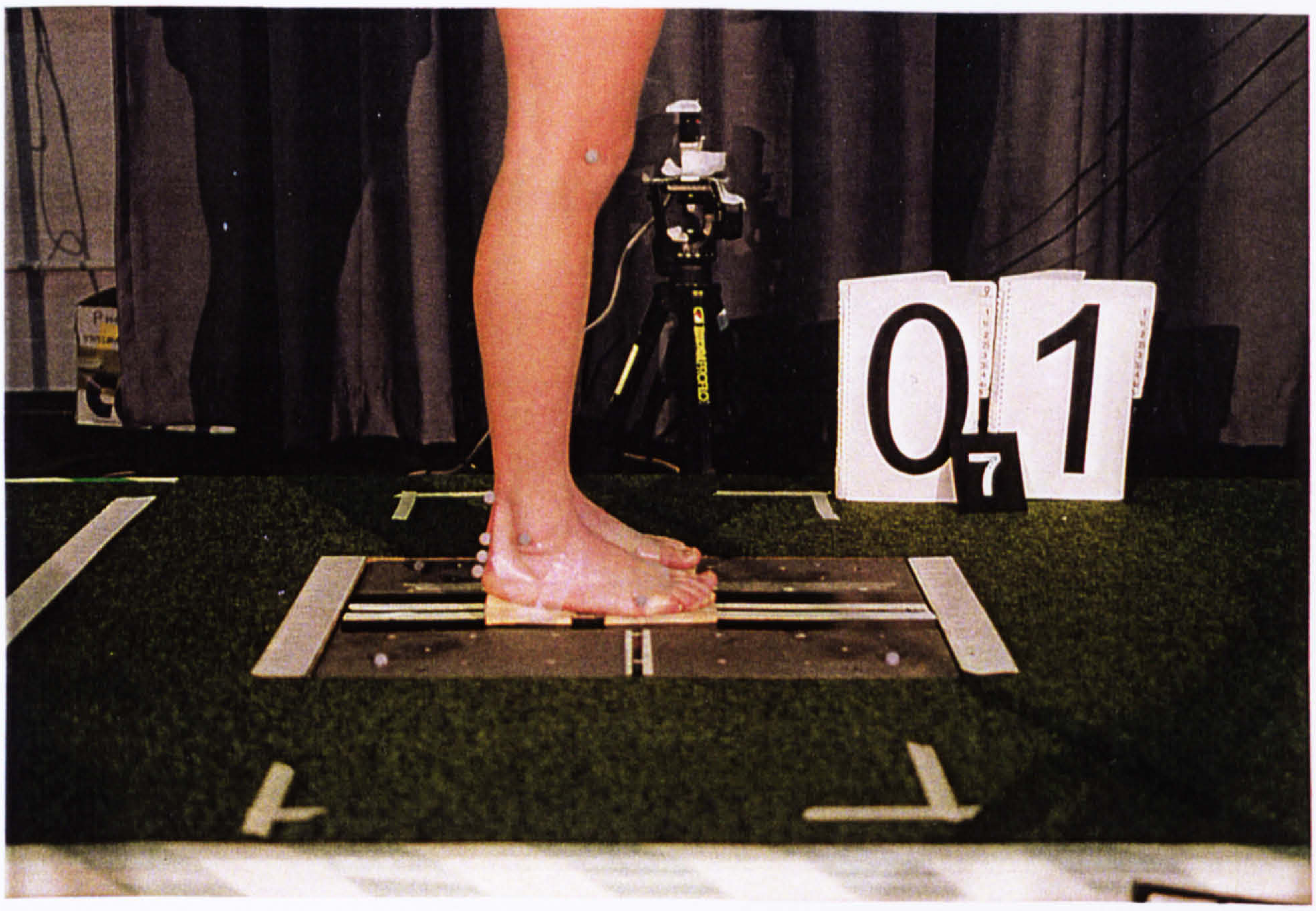


Figure 6.4 Marker locations and attachment of heel lifts

(v) Measurement Precision and Accuracy

The precision attained in estimation of ankle moment and Achilles tendon force variables was investigated. Reliability in locating markers was assessed using RMS variation from the mean estimated coordinates of calibration markers over 10 consecutive fields. The accuracy of three-dimensional coordinates obtained using the MacReflex system was assessed by quantifying the RMS variation from criterion calibration frame data over 10 consecutive fields. Since the calibration frame markers were larger than the markers used on the body, the influence of marker size on the accuracy and reliability levels attained was assessed by additionally locating four markers of the same size as the markers used on the body, placed at known locations on the force plate.

The precision attained in GRF data has been investigated in previous chapters, with a maximum error in peak Achilles tendon forces of 0.3 BW being identified. For quantification of the measurement precision in ankle moment and Achilles tendon loading data, a sensitivity analysis was employed in which the likely errors in GRF and kinematic data were combined. This provided a measure of the confidence in dependent variables for each individual trial. The confidence with which comparisons could be made between conditions was increased by a factor of $10^{1/2}$ by the use of 10 trials for each subject/condition combination. Subsequent confidence intervals for comparisons between conditions were obtained for each variable.

A recommended marker size of between 0.5 % and 18 % of diagonal field of view is recommended in the technical specifications supplied by the manufacturer of the MacReflex system. A measurement resolution of 0.005 % of the field of view is quoted. Sagittal and frontal plane diagonal lengths were calculated for the calibration frame used in the present study, and these were used as diagonal field of view lengths for quantification of recommended marker size and for calculation of the measurement resolution.

A Performance Index (PI) is defined in the technical specification as the length of the diagonal field of view divided by the standard deviation of the noise in the data (Figure 6.5). This PI indicates the amount of noise to be expected in relation to the full measurement scale. A linear relationship is demonstrated between PI and marker size (% diagonal field of view). Thus a greater PI will be expected for the markers on the calibration frame (50 mm diameter) compared with the smaller body markers (10 mm diameter). The measurement resolutions calculated for the markers of different size were compared with the expected increase in PI (Figure 6.5).

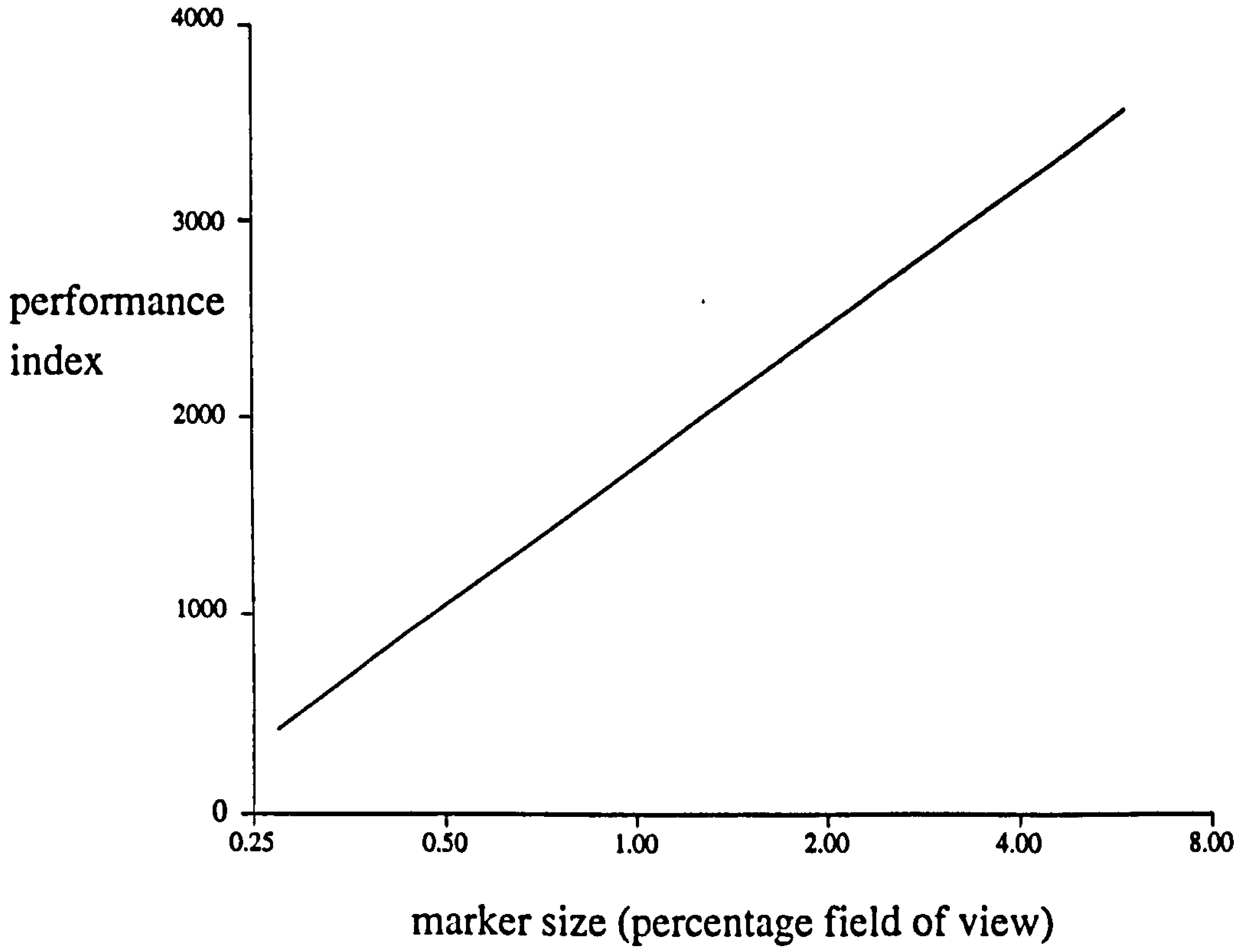


Figure 6.5 Performance index against marker size (MacReflex Technical Specification)

(vi) Data Analysis

Initially results for each of the measured variables were compared between conditions for each individual subject using graphical presentations. Where trends in results were observed, statistical analysis was applied to test for significant differences between conditions. An ANOVA, followed by a post-hoc Tukey test, were used to test for significant differences between conditions for each subject ($p<0.05$). For each subject and variable, comparisons were made between Condition A and Condition B, to investigate the influence of attaching the lifts to the plantar surface of the foot. Condition B was termed the zero heel lift condition since lifts of equal height were attached to the rearfoot and forefoot for this condition. The influence of heel lift was investigated by comparing the 7.5 mm heel lift condition (Condition C) with the zero heel lift condition, and the 15 mm heel lift condition (Condition D) with the zero heel lift condition. Where trends in response to increased heel lift were apparent, the 7.5 mm and 15 mm heel lift results were compared.

For Subject 7, additional comparisons between Condition B, Condition E, and Condition F, allowed investigation of the influence of providing increased shock absorption.

6.3 Results

Each of the subjects were considered separately. Tables of results were provided for each variable, and trends highlighted using graphical methods, where considered appropriate. For each variable, significant differences detected between conditions within subjects were noted ($p<0.05$). For each subject, age, height, mass, foot length and tendon calliper measurement are provided in Table 6.1.

Table 6.1 Mean age, height, mass, foot length and tendon measurement for each subject

	age (years)	height (m)	mass (kg)	heel - MTP (cm)	tendon (mm)
Subject 1	47	1.64	49.5	15.5	19.0
Subject 2	20	1.62	52.3	16.6	20.5
Subject 3	21	1.68	50.8	16.5	19.0
Subject 4	25	1.65	54.7	16.7	17.0
Subject 5	21	1.58	48.9	15.7	21.5
Subject 6	22	1.64	54.8	15.9	18.0
Subject 7	27	1.64	54.1	16.6	19.0
Subject 8	27	1.58	49.8	15.7	20.5
mean (SD)	26 (9)*	1.63 (0.03)	51.9 (2.4)	16.2 (0.5)	19.3 (1.5)

* If omit Subject 1 (n = 7), then mean age = 23, SD = 3

(i) Shock Absorption Conditions

The influence on maximum Achilles tendon force of increased shock absorption (Condition E), and increased heel lift with shock absorption (Condition F) is illustrated in Table 6.2. These conditions were applied for Subject 7 only.

Peak dorsi-flexion moment was found to be reduced for both of the increased shock absorption conditions compared with the zero heel lift and zero shock absorption condition (Condition B). A smaller peak dorsi-flexion moment was observed for the shock absorption with no heel lift condition (Condition E), compared with the increased shock absorption and heel lift condition (Condition F).

The provision of increased shock absorption resulted in an increase in the peak plantar-flexion moment and an increase in the maximum Achilles tendon force compared with the baseline condition.

Table 6.2 Peak ankle moments and Achilles tendon force for Subject 7 (SD in parenthesis)

	Condition B	Condition E	Condition F
peak dorsi-flexion moment (N.m)	22.0 (3.3)	4.8 (4.7)	9.2 (2.5)
peak plantar-flexion moment (N.m)	156.3 (3.6)	162.0 (4.7)	162.7 (2.6)
peak Achilles tendon force (BW)	11.4 (0.5)	11.7 (0.5)	12.0 (0.5)

(ii) Ground Reaction Force

For each subject, time histories of vertical GRF at 1000 Hz were obtained for each of the trials for each condition. It was found that the averaging of normalised vertical GRF data over ten trials for each subject/condition combination, resulted in a smoothing of the peak impact force, due to the slightly different time of occurrence of this peak across trials. This is illustrated by the example provided in Figure 6.6. The average GRF trace clearly does not represent a typical time history of vertical GRF.

Average GRF loading rate and peak instantaneous GRF loading rate during the impact phase are compared across conditions in Table 6.3 for each subject. Also included are the times of occurrence of instantaneous GRF loading rate. It is evident that the trends in loading rate variation with heel lift for each subject are similar using the two different methods of GRF loading rate representation. Example plots of instantaneous GRF loading rate change over time during initial ground contact are provided in Figure 6.7.

Typical vertical GRF traces are provided in Figure 6.8, with two distinct impact peak patterns being illustrated. The mean peak impact forces and times of occurrence, obtained by identification of the peak for each trial, are presented in Table 6.4 for each subject/condition combination. Also provided in this table are the maximum active peak and time of occurrence, total stance time, and instantaneous GRF loading rate for each subject/condition combination. Few significant differences in GRF variables were identified across conditions.

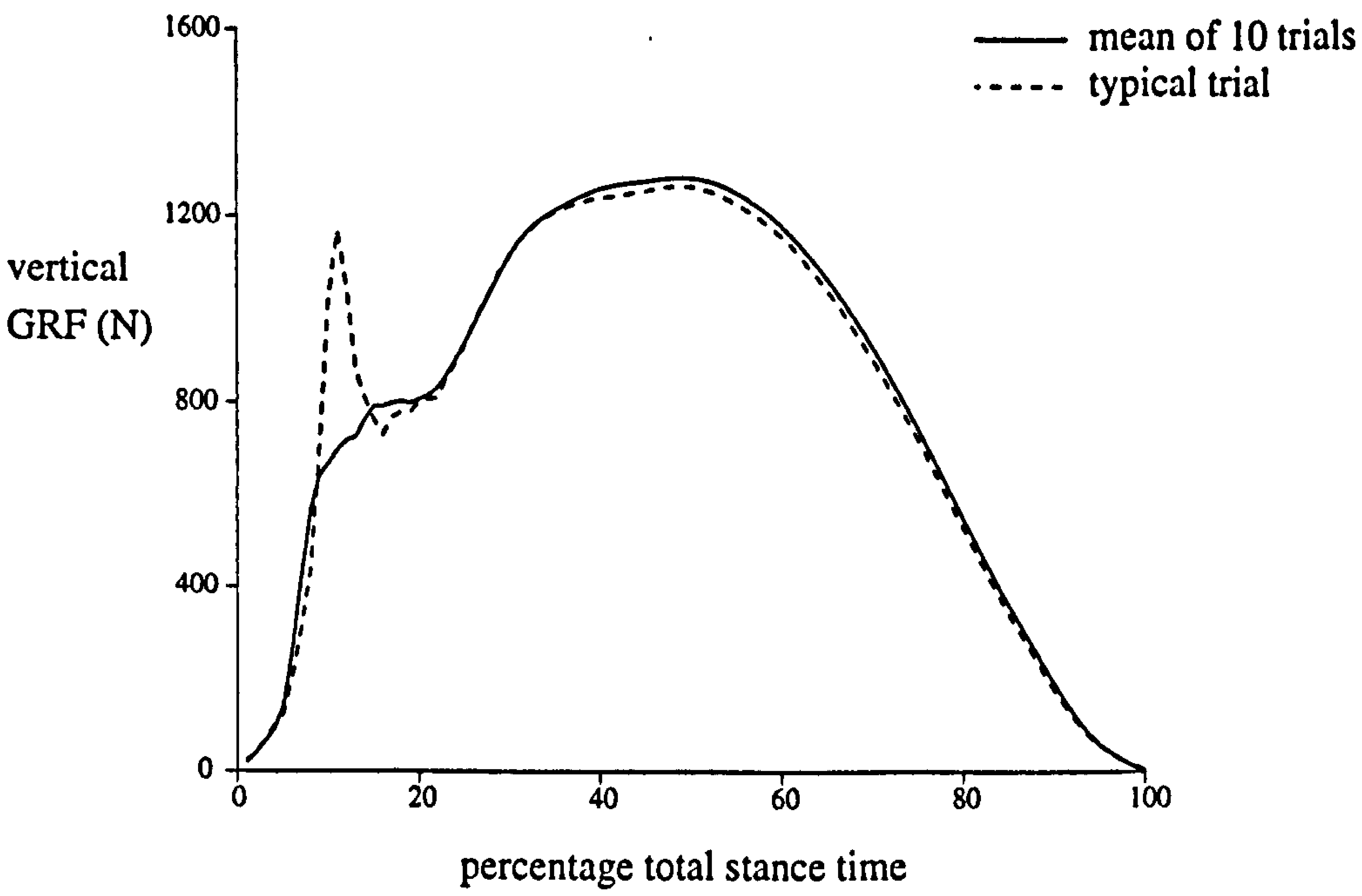


Figure 6.6 Example of the smoothing influence of using mean vertical GRF data

Table 6.3 Average and instantaneous GRF loading rates in BW.s^{-1} (SD), and occurrence of instantaneous GRF loading rate in ms (SD) for each subject/condition combination

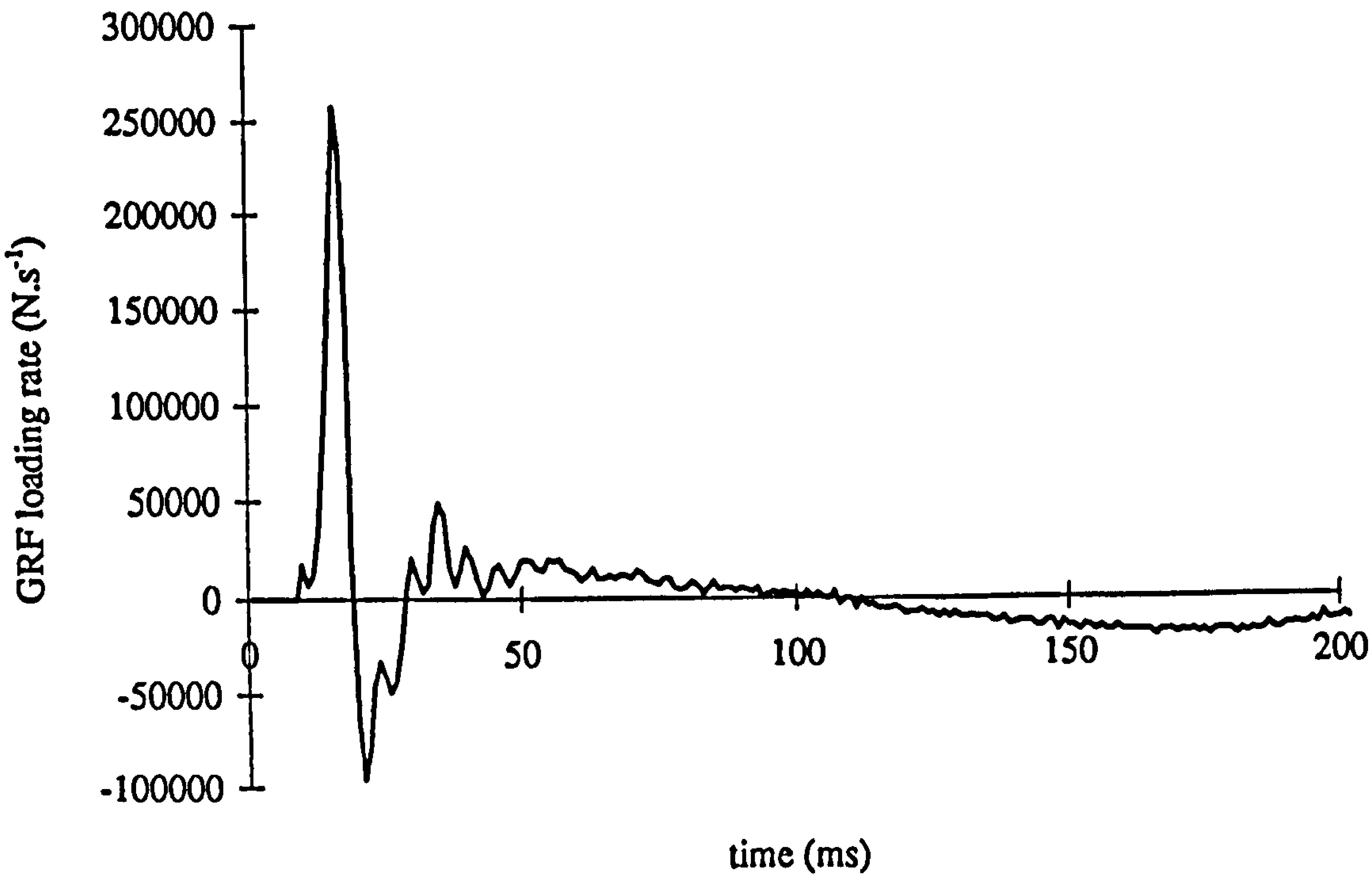
	average rate	instantaneous	instantaneous time
Subject 1			
A	461 (63)	913 (92)	0.005 (0.000)
B	315 (77)	1020 (152)	0.009 (0.001)
C	246 (54)	703 (296)	0.009 (0.003)
D	217 (81)	743 (176)	0.013 (0.004)
Subject 2			
A	422 (57)	881 (81)	0.006 (0.001)
B	237 (38)	756 (84)	0.010 (0.002)
C	153 (20)	524 (55)	0.012 (0.001)
D	149 (15)	601 (38)	0.014 (0.001)
Subject 3			
A	127 (32)	438 (120)	0.019 (0.003)
B	132 (29)	415 (87)	0.018 (0.003)
C	146 (18)	458 (88)	0.016 (0.001)
D	140 (17)	458 (43)	0.015 (0.002)
Subject 4			
A	234 (47)	561 (97)	0.009 (0.002)
B	240 (57)	538 (79)	0.010 (0.003)
C	178 (45)	417 (152)	0.012 (0.003)
D	119 (43)	359 (86)	0.014 (0.002)
Subject 5			
A	105 (58)	212 (169)	0.016 (0.004)
B	98 (30)	172 (183)	0.019 (0.004)
C		398 (108)	0.015 (0.002)
D	77 (29)	286 (70)	0.015 (0.003)
Subject 6			
A	218 (21)	463 (28)	0.005 (0.000)
B	173 (33)	463 (68)	0.011 (0.001)
C	148 (36)	416 (79)	0.012 (0.002)
D	131 (58)	385 (138)	0.012 (0.002)
Subject 7			
A	167 (42)	408 (32)	0.007 (0.001)
B	180 (25)	451 (74)	0.008 (0.003)
C	93 (32)	340 (52)	0.013 (0.001)
D	57 (14)	193 (33)	0.014 (0.002)
Subject 8			
A	110 (50)	330 (207)	0.019 (0.007)
B	115 (40)	347 (160)	0.020 (0.005)
C	124 (26)	407 (97)	0.019 (0.003)
D	140 (31)	447 (105)	0.018 (0.004)

NB. Where values are omitted, no impact peak was observed

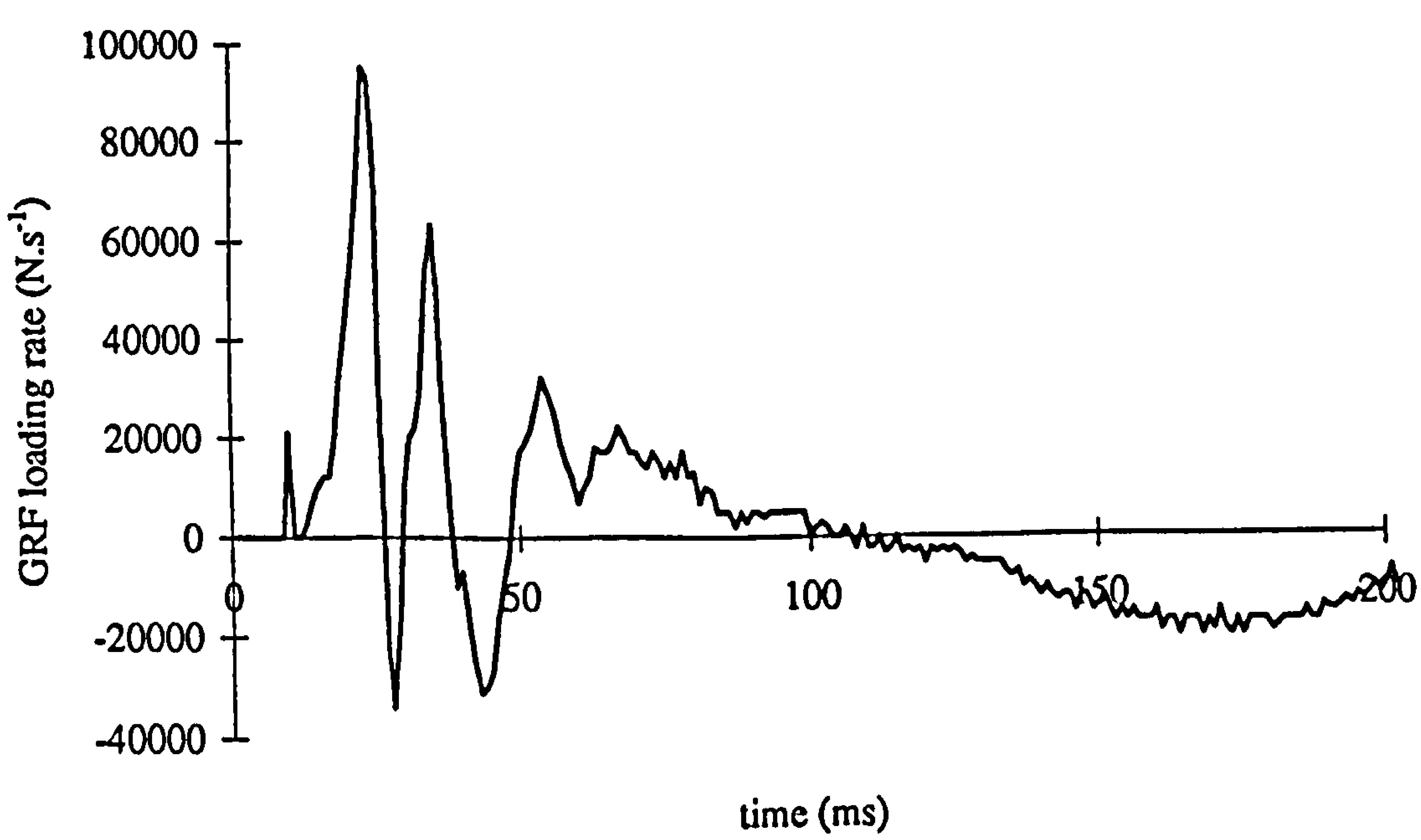
Table 6.4 Mean vertical GRF variables for each subject/condition combination
(A = barefoot, B = zero heel height, C = 7.5 mm heel lift, D = 15 mm heel lift)

	impact peak BW (SD)	peak time % (SD)	active peak BW (SD)	active time % (SD)	stance (ms)	rate ($BW.s^{-1}$)
Subject 1						
A	3.07 (0.25)	3.5 (0.2)	2.82 (0.26)	39.1 (3.1)	192 (8)	913 (92)
B	3.52 (0.45)	6.2 (0.8)	2.90 (0.19)	39.3 (3.1)	180 (12)	1020 (152)
C	2.87 (0.39)	6.4 (0.7)	2.81 (0.16)	41.0 (2.5)	188 (10)	703 (296)
D	2.70 (0.46)	7.3 (2.6)	2.96 (0.26)	39.3 (4.1)	191 (14)	743 (176)
Subject 2						
A	3.06 (0.30)	3.5 (0.3)	2.90 (0.08)	39.5 (0.9)	209 (7)	881 (81)
B	2.68 (0.20)	5.2 (0.7)	2.92 (0.06)	39.2 (0.8)	222 (5)	*756 (84)
C	2.04 (0.16)	6.0 (0.4)	2.82 (0.07)	40.3 (1.1)	227 (4)	*524 (55)
D	2.35 (0.12)	7.0 (0.5)	2.87 (0.05)	40.6 (0.9)	226 (6)	*601 (38)
Subject 3						
A	2.65 (0.31)	9.8 (1.3)	2.61 (0.08)	42.4 (1.3)	219 (6)	438 (120)
B	2.61 (0.17)	9.6 (1.4)	2.69 (0.03)	43.7 (2.1)	214 (6)	415 (87)
C	2.55 (0.32)	8.4 (0.4)	2.71 (0.04)	43.5 (3.8)	209 (13)	458 (88)
D	2.49 (0.20)	8.3 (1.0)	2.67 (0.03)	43.8 (1.4)	216 (5)	458 (43)
Subject 4						
A	2.49 (0.27)	4.8 (0.8)	2.67 (0.07)	41.5 (2.9)	230 (8)	561 (97)
B	2.54 (0.24)	5.0 (1.5)	2.71 (0.08)	40.2 (3.7)	222 (7)	538 (79)
C	2.36 (0.34)	6.1 (1.4)	2.71 (0.11)	41.4 (4.0)	226 (15)	417 (152)
D	*2.10 (0.40)	8.3 (1.6)	2.65 (0.05)	44.3 (4.0)	228 (9)	*359 (86)
Subject 5						
A	1.85 (0.40)	9.0 (2.2)	2.73 (0.07)	40.5 (2.2)	219 (9)	212 (169)
B	1.68 (0.28)	7.1 (1.0)	2.64 (0.11)	40.7 (3.3)	*247 (11)	172 (183)
C			2.80 (0.09)	42.5 (2.9)	*218 (7)	398 (108)
D	1.50 (0.24)	9.4 (2.9)	2.81 (0.11)	43.1 (2.2)	*231 (9)	286 (70)
Subject 6						
A	1.89 (0.10)	4.1 (0.2)	2.53 (0.04)	45.1 (1.1)	210 (4)	463 (28)
B	2.16 (0.24)	5.9 (0.5)	2.55 (0.04)	44.0 (1.5)	215 (7)	463 (68)
C	2.02 (0.25)	6.6 (1.1)	2.56 (0.04)	44.5 (2.2)	216 (5)	416 (79)
D	2.01 (0.48)	8.1 (2.0)	2.58 (0.06)	45.7 (2.6)	212 (10)	385 (138)
Subject 7						
A	1.66 (0.25)	4.9 (0.9)	2.48 (0.08)	43.2 (1.0)	212 (7)	408 (32)
B	1.77 (0.12)	4.6 (0.4)	2.44 (0.07)	40.8 (2.1)	217 (7)	451 (74)
C	*1.47 (0.22)	7.8 (2.5)	2.46 (0.07)	42.3 (3.6)	222 (7)	*340 (52)
D	*1.43 (0.14)	11.4 (2.1)	2.46 (0.05)	44.8 (2.7)	224 (8)	*193 (33)
Subject 8						
A	2.38 (0.43)	11.7 (2.4)	2.64 (0.06)	44.1 (2.7)	203 (7)	330 (207)
B	2.42 (0.32)	11.4 (3.1)	2.74 (0.04)	42.9 (3.6)	201 (6)	347 (160)
C	2.52 (0.17)	10.2 (1.5)	2.68 (0.05)	43.4 (3.7)	204 (6)	407 (97)
D	2.69 (0.16)	10.1 (2.4)	2.77 (0.07)	45.1 (1.1)	200 (9)	447 (105)

*$p<0.05$	impact peak:	Subject 4 B versus D; Subject 7 B versus C, D
	active peak:	none
	stance time:	Subject 5 A versus B; B versus C, D
	loading rate:	Subject 2 A versus B; B versus C, D
		Subject 4 B versus D
		Subject 7 B versus C, D

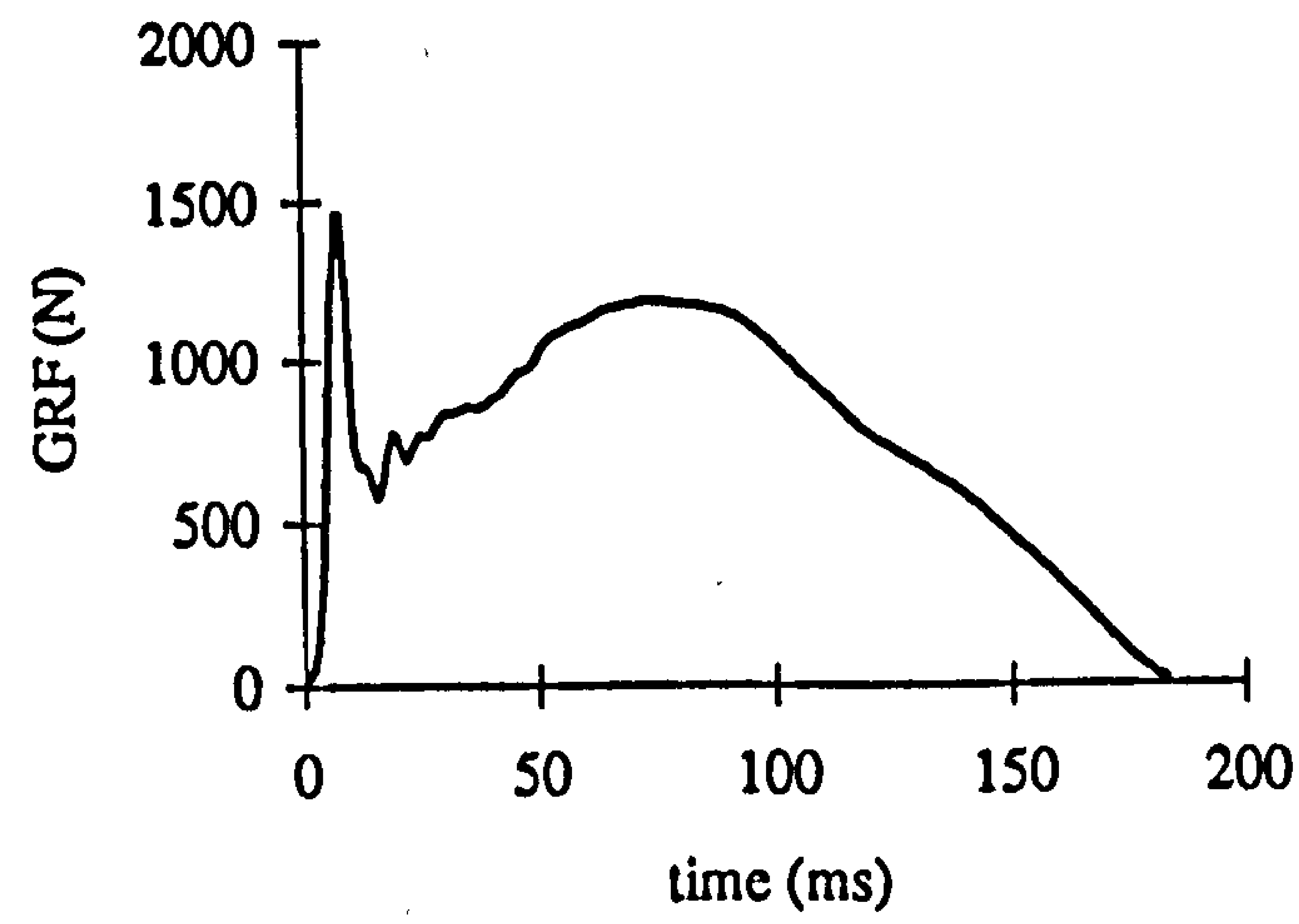


(a) Subject 3

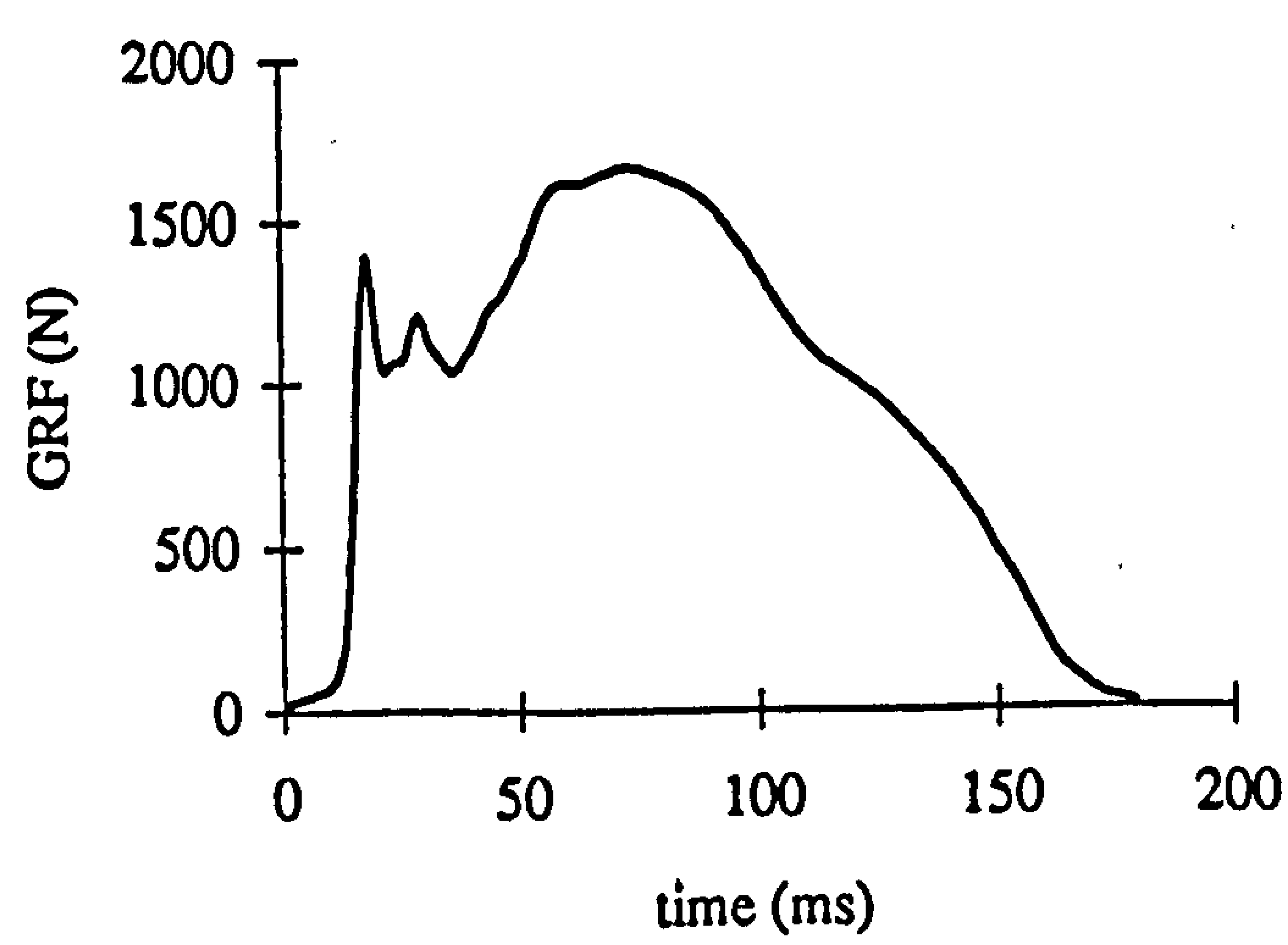


(b) Subject 7

Figure 6.7 Typical instantaneous GRF loading rate plots



(i) Single impact peak



(ii) Double impact peak

Figure 6.8 Typical vertical ground reaction force time histories

(iii) Assessment of Methods

For representation of the level of accuracy attained, RMS deviation in marker co-ordinates from criterion values, calculated over 10 consecutive fields, are presented in Table 6.5 (i, ii), with two sets of typical data being provided. Over the seven markers on the calibration frame, the RMS deviation ranged from 0.06 mm to 0.5 mm. The mean RMS deviation calculated over all seven markers ranged from 0.14 mm to 0.21 mm. Taking these mean values, the accuracy in locating markers on the calibration frame was therefore taken to be in the region of 0.2 mm.

Reliability in locating markers was assessed using RMS deviation from the mean over 10 consecutive fields. RMSD for the seven calibration frame markers and the four markers placed on the force plate were calculated separately, allowing comparison of variation for the two different sizes of marker (Table 6.6 i, ii). It is evident that the variation in measured location for the smaller markers is greater, by two to three times, than the variation for the larger markers on the calibration frame. Over the two typical trials illustrated, the mean variation over the seven calibration frame markers was 0.10 mm, and the mean variation over the four smaller markers was 0.26 mm.

The theoretical influence of calculating marker locations in two dimensions using video data and a two-dimensional DLT procedure is illustrated in Figure 6.9. For the ankle, Achilles 1 and Achilles 2 markers respectively the sample coordinates used to investigate projection influence were: (-27.26, -46.97, 61.44); (-62.74, -59.37, 109.18); (-65.83,-70.42, 78.69). The influence of projection of the z coordinates is illustrated, using the ankle marker as an example. Projection onto the yz plane increased the z-coordinate of the ankle marker from 61.44 mm to 65.12 mm, the Achilles 1 marker from 109.18 mm to 117.03 mm, and the Achilles 2 marker from 78.69 mm to 87.30 mm. This resulted in an increase in the calculated Achilles tendon moment arm from 27.9 mm to 31.3 mm. For the sample data considered, the subsequent influence on estimated Achilles tendon force is a decrease of approximately 1 BW. This corresponds with a 10% influence of projection error on maximum Achilles tendon force value when video data are used compared with three-dimensional MacReflex data. Since the camera was placed such that the optical axis was directed along the visually identified x-axis, it was assumed that the influence of projection onto the yz plane had a negligible influence on y coordinates.

Table 6.5 RMSD from criterion locations for seven calibration frame markers in millimetres, for trial 1 (i) and trial 2 (ii)

(i)

marker	X	Y	Z
1	0.21	0.11	0.09
2	0.21	0.07	0.44
3	0.07	0.21	0.24
4	0.13	0.07	0.14
5	0.12	0.33	0.04
6	0.16	0.10	0.09
7	0.13	0.07	0.11
mean	0.15	0.14	0.17

(ii)

marker	X	Y	Z
1	0.50	0.18	0.47
2	0.26	0.08	0.11
3	0.13	0.10	0.20
4	0.18	0.21	0.35
5	0.08	0.27	0.06
6	0.15	0.11	0.11
7	0.11	0.08	0.15
mean	0.15	0.21	0.20

Table 6.6 RMSD from the mean for seven calibration frame markers, and four small markers in millimetres, for trial 1 (i) and trial 2 (ii)

(i)

marker	X	Y	Z
1	0.33	0.16	0.21
2	0.11	0.07	0.02
3	0.14	0.07	0.10
4	0.10	0.22	0.05
5	0.07	0.16	0.06
6	0.16	0.03	0.02
7	0.11	0.06	0.05
mean	0.15	0.11	0.07
8	0.37	0.12	0.63
9	0.15	0.25	0.60
10	0.26	0.40	0.33
11	0.09	0.25	0.44
mean	0.22	0.25	0.44

(ii)

marker	X	Y	Z
1	0.12	0.04	0.06
2	0.09	0.25	0.08
3	0.04	0.08	0.05
4	0.15	0.08	0.02
5	0.04	0.22	0.07
6	0.06	0.03	0.09
7	0.06	0.01	0.03
mean	0.08	0.10	0.06
8	0.21	0.30	0.30
9	0.22	0.31	0.13
10	0.04	0.30	0.25
11	0.09	0.07	0.31
mean	0.14	0.25	0.25

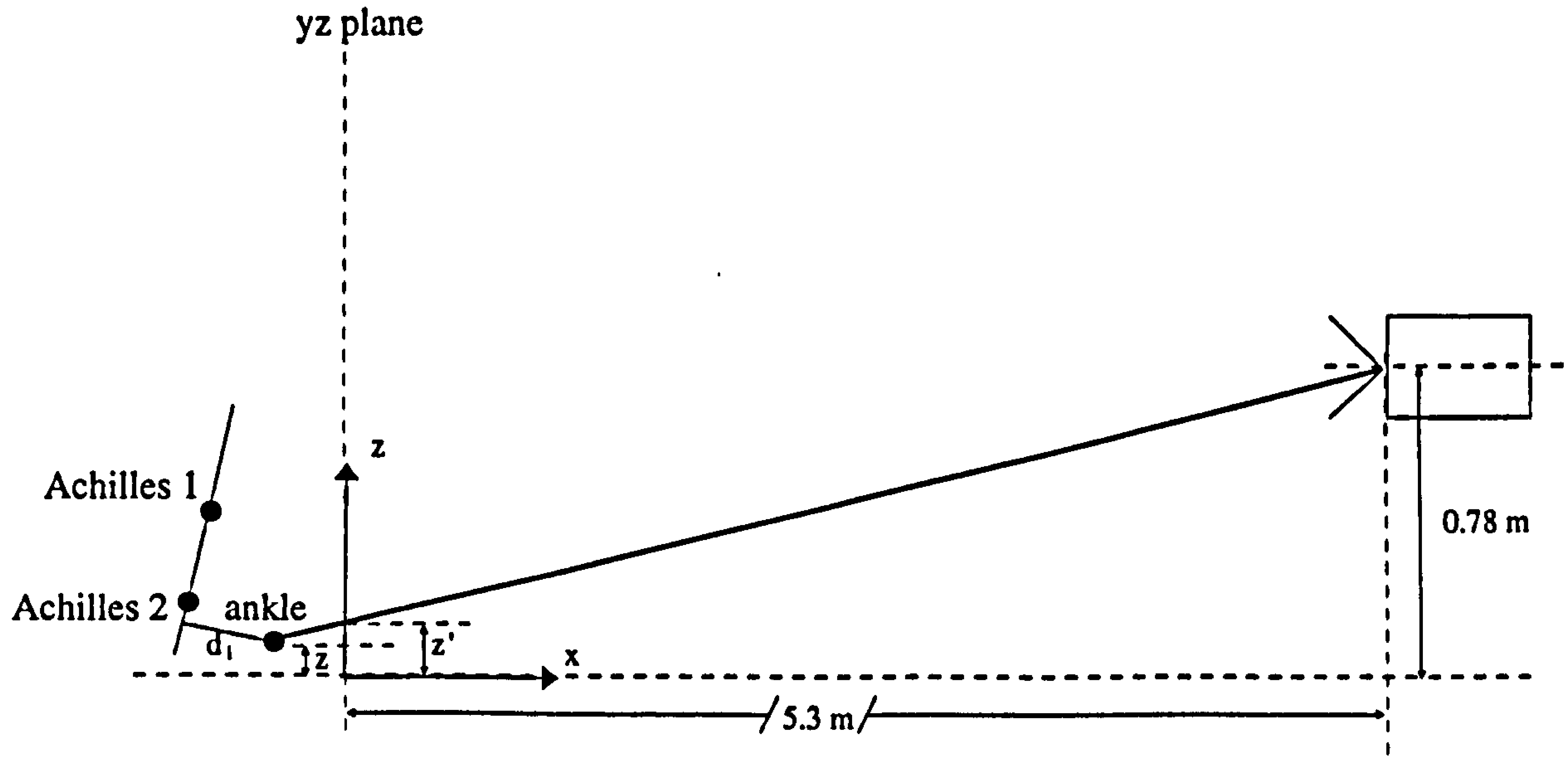


Figure 6.9 Theoretical influence of projection of three-dimensional data onto the yz plane

(x,z) = ankle co-ordinates in xz plane

z' = z co-ordinate of ankle projected onto yz plane

$z' = z + a$

By similar triangles, $a/x = (0.78 - z) / (5.3 - x)$

$a = 0.02726\ (0.78 - 0.06144) / (5.3 + 0.02726)$

$a = 0.00368$ m

$z' = 0.06144 + 0.00368$

$z' = 0.06512$ m

(the influence on Achilles tendon markers is provided in the text)

(iv) Measurement Precision and Accuracy

The repeated calliper measurements of the Achilles tendon for a single subject on three separate occasions, revealed a maximum variation of 0.5 mm. It was assumed that this result was typical for all subjects.

A sagittal plane diagonal of 902 mm, and a frontal plane diagonal of 1246 mm were obtained for the MacReflex data. For a field of view of 1246 mm, the manufacturers' recommended marker size is between 6 mm and 224 mm diameter. Measurement resolutions of 0.05 mm (0.005% of 902 mm) in the yz or sagittal plane, and 0.06 mm in the xz or frontal plane, (0.005% of 1246 mm) were calculated. This measurement resolution, obtained using the system specifications, compares well with the range of 0.06 mm to 0.5 mm accuracy obtained using the data obtained in the present study.

The greater than two fold decrease in standard deviation (RMSD from the mean) when reliability was measured for the body markers compared with the larger calibration frame markers, resulted in a two fold increase in the performance index, defined as field of view divided by standard deviation. The body markers have a 10 mm diameter, corresponding to 0.5% field of view. The calibration frame markers have a 50 mm diameter, corresponding to 2.5% field of view. The PI values corresponding with these relative marker sizes are approximately 1000 and 2000 for the 10 mm and 50 mm markers respectively, demonstrating a two fold increase in reliability, as has been found in the present study.

It was found that the MacReflex system identified the location of the stationary calibration frame markers to within 0.2 mm, and the smaller 10 mm diameter markers to within 0.3 mm. The coordinate data obtained in the present chapter were therefore at least as reliable as data obtained in Chapter 4, in which a RMSD of 0.4 mm was obtained over repeated digitisations. The sensitivity analysis results obtained in Chapter 4 were therefore applied to the results of the present chapter.

Ankle plantar-flexion moments

Reference to the sensitivity analysis of Chapter 4 provided a figure of 3.4% for the random error influence on peak ankle plantar flexion moment. For the peak plantar-flexion ankle moment values in the region of 160 N.m found in the present study, this corresponded with an error of 5.4 N.m. For comparisons made between conditions, the magnitude of this error has been reduced by the use of mean values over 10 trials for each subject/condition combination. The result was a possible random error influence of 1.7 N.m.

Ankle dorsi-flexion moment

The peak dorsi-flexion ankle moment was found to occur within the first 20% of the stance phase, at a time when the moment arm of the resultant GRF about the ankle joint centre was in the region of 0.05 m, compared with 0.13 m in the middle of the stance phase when peak plantar-flexion moment occurs. A variation of 2 mm in the centre of pressure was found to result in a 4% variation in the ankle moment value, and a 0.4 mm variation in the ankle marker location was found to vary the ankle moment by 0.8%. Summing of the possible errors of 1%, 4%, 0.8% for the GRF magnitude, centre of pressure and ankle marker location, respectively, resulted in a maximum possible variation of 5.8%. A typical magnitude of peak dorsi-flexion moment obtained in this chapter was 20 N.m, and thus the maximum random error influence was in the region of 1.2 N.m.

Maximum Achilles tendon force

For comparisons between conditions with maximum Achilles tendon force values in the region of 10 BW, the level of precision was 0.1 BW, or 1% of the force value (Chapter 4). The majority of maximum Achilles tendon force values estimated in the present study were in the region of 10 BW. The influence of 0.4 mm variation in ankle marker location for Subject 6, with force values of around 7 BW, had a comparable influence to that described for a maximum Achilles tendon force of 10 BW. For Subject 8, with maximum Achilles tendon force values of up to 18 BW, it was found that 0.4 mm variation in ankle marker location resulted in a possible change in Achilles tendon force of 4.8 BW for a single trial. In turn, this resulted in a possible error of 1.5 BW in the mean over 10 trials. Thus, confidence due to marker location measurement precision was within 1.5 BW for Subject 8, compared with 0.1 BW for the other subjects. Combined with the influence of force plate random error found in Chapter 4 of 0.3 BW for a single trial, the combined total possible error due to measurement precision was 1.8 BW for Subject 8, and 0.4 BW for the remaining subjects. Over 10 trials, this corresponded with 0.6 BW for Subject 8, and 0.1 BW for the remaining subjects.

Achilles tendon loading rate

Smoothed Achilles tendon loading rate has been calculated by consideration of average rate over 0.01 s time periods. Assuming the time data are reliable, Achilles tendon loading rates have a measurement precision of 10 BWs^{-1} (0.1 BW/0.01 s). This is the level of precision attained when the Achilles tendon force is of a magnitude around 10 BW. At the start of the stance phase, when maximum loading rate was found to occur, the Achilles tendon force was of a lower magnitude, in the region of 0 to 2 BW. The magnitude of the possible error in Achilles tendon force was smaller, and thus the absolute error in loading rate was smaller.

(v) Data Smoothing and Velocity and Acceleration Calculations

Plots of Achilles tendon moment arm over time using different parameter values in the Reinsch cubic spline are illustrated in Figure 6.10. Examination of the raw data plots (Constant = 0) demonstrated that the displacement data contained noise. From the reliability data presented in Section 6.3 (iii), it was known that individual markers could be located to within 0.2 mm. Thus, to protect against over-smoothing of kinematic data, the marker displacement curves selected for use were those which demonstrated maximum smoothing without adjusting any marker locations by more than 0.2 mm, based on visual assessment.

Using these criteria for all markers considered, a parameter of 0.3 was selected for smoothing of all kinematic displacement data used in subsequent calculations.

The influence of using finite difference methods to calculate velocity and acceleration values was assessed, as described by Lees (1980). Velocity profiles obtained using a direct two point method appeared smooth and free of unwanted noise, and were considered to be acceptable for use in comparing joint and marker velocity values between conditions for each subject. An example of a foot segment velocity profile is provided in Figure 6.11.

Accelerations obtained using the two point procedure were found to obtain undesirable noise which would limit the ability to confidently compare acceleration values obtained across conditions (Figure 6.12a). The use of a five point method (Lanczos, 1957) reduced the level of noise, but was still considered to be unacceptable for comparison between conditions, as illustrated in Figure 6.12b. The use of a nine point method (Lanczos, 1957), provided smooth acceleration profiles which were considered to follow the pattern of acceleration variation illustrated by the noisy direct two point method acceptably whilst removing the unwanted noise (Figure 6.12c).

The influences of using the different methods of acceleration calculation on ankle moments and Achilles tendon forces are illustrated in Figures 6.13 and 6.14, respectively.

(vi) Ankle Moment

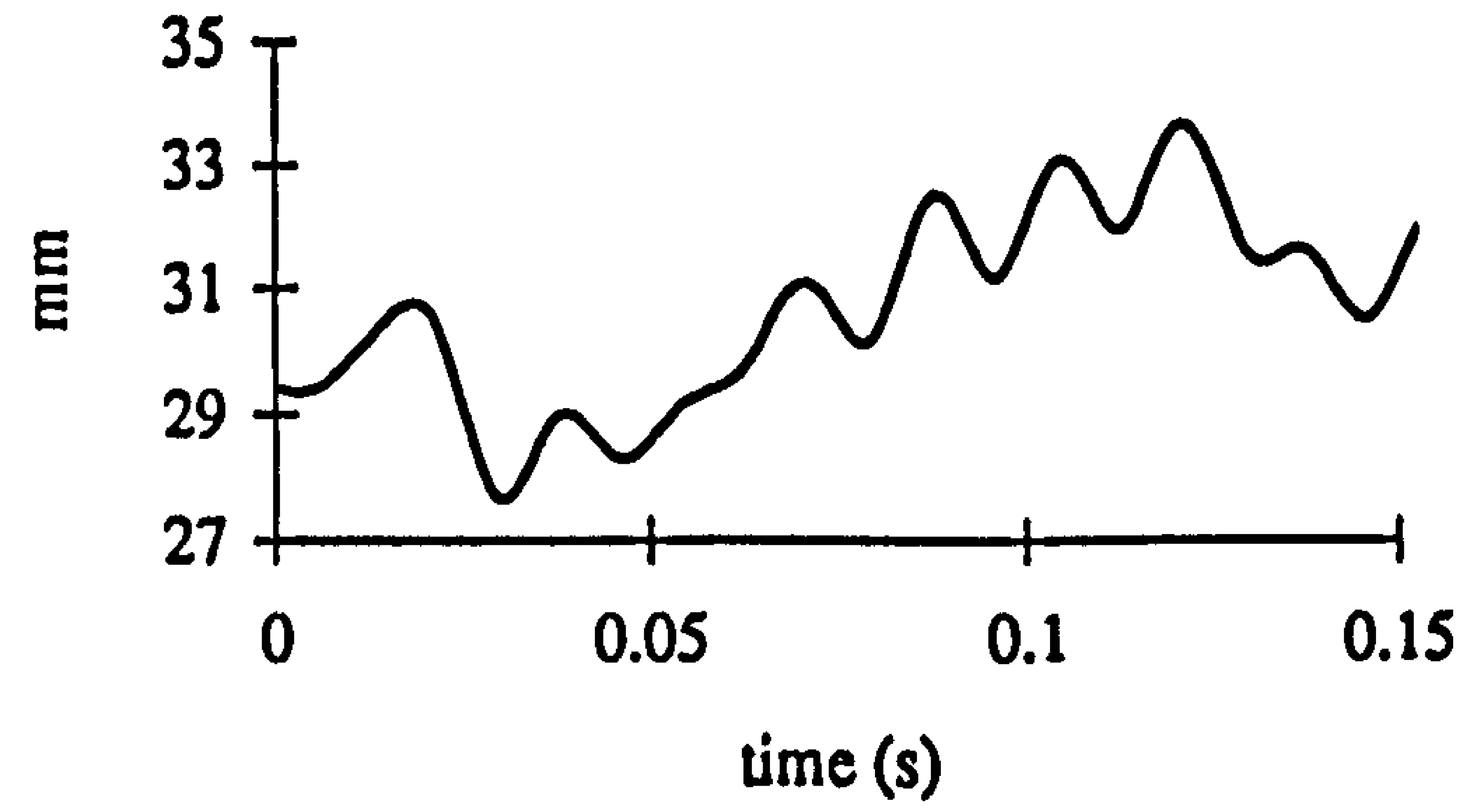
Typical ankle moment time histories for selected subject/condition combinations are provided in Figure 6.15. The use of different sets of inertia data was found to have no influence on maximum ankle plantar-flexion moments or dorsi-flexion moments. Similar time histories were demonstrated across most subjects, with an initial small dorsi-flexion moment occurring, followed by ankle plantar-flexion. For Subject 8, zero dorsi-flexion moment was demonstrated. Peak plantar-flexion and dorsi-flexion ankle moments for each subject/condition combination are provided in Table 6.7 and Table 6.8 respectively, with standard deviation values in parenthesis. The times of occurrence of peak ankle plantar-flexion moment are provided in Table 6.9, as a percentage of total stance time. Variation in peak plantar-flexion moment over repeated trials was found to be larger for Subject 1, than for all other subjects. The percentage of total stance time at which maximum ankle plantar-flexion moment occurred was markedly higher for Subject 1, than for the other seven subjects. For this subject, the peak plantar-flexion moment occurred around 60% of total stance time, compared with between 45% and 55% for the remaining subjects.

Peak plantar-flexion ankle moments for each condition are illustrated graphically in Figure 6.17, with separate plots for each individual subject. For all subjects, the raising of the rearfoot and forefoot simultaneously resulted in an increase in the magnitude of the peak moment, compared with the barefoot condition. For five of the eight subjects, this increase was significant ($p<0.05$). That is, raising of the foot tended to cause an increase in the maximum ankle plantar-flexion moment.

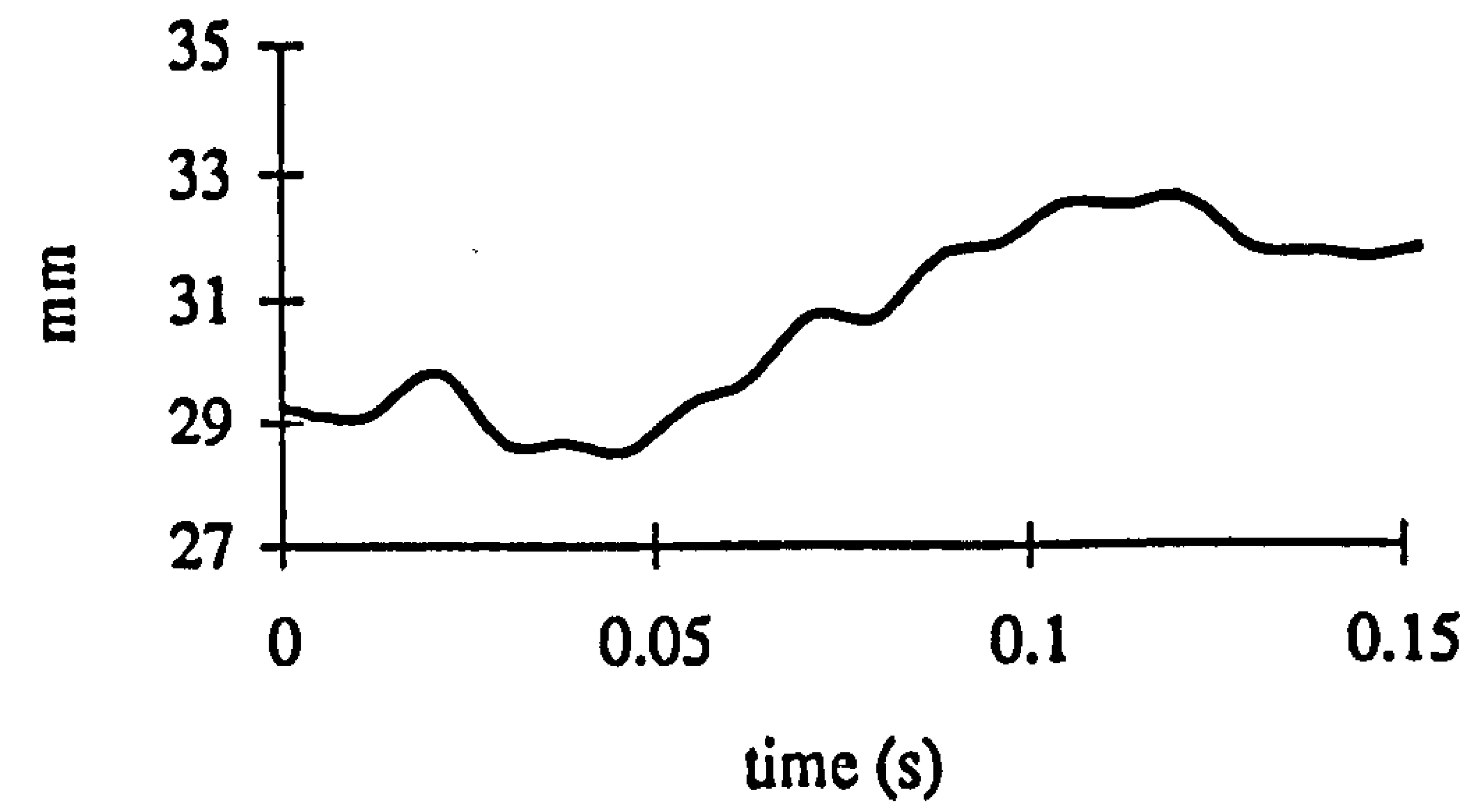
Comparison across the heel lift conditions highlighted differing responses across subjects. Both significant increases and significant decreases were demonstrated, but in general no difference in maximum ankle plantar-flexion moment occurred across heel lift conditions. All effect sizes which were found to be significant were considerably greater than the magnitude of error expected due to random variation.

Examination of the maximum dorsi-flexion ankle moment values revealed no clear patterns of variation in the change in this moment across conditions (Table 6.8). Although differences were found, few of these were significant. Full sets of results for dorsi-flexion moments were only available for five of the eight subjects, due to a dorsi-flexion moment of zero for some or all of the conditions for three subjects. Both significant increases and significant decreases in maximum ankle dorsi-flexion moment were demonstrated.

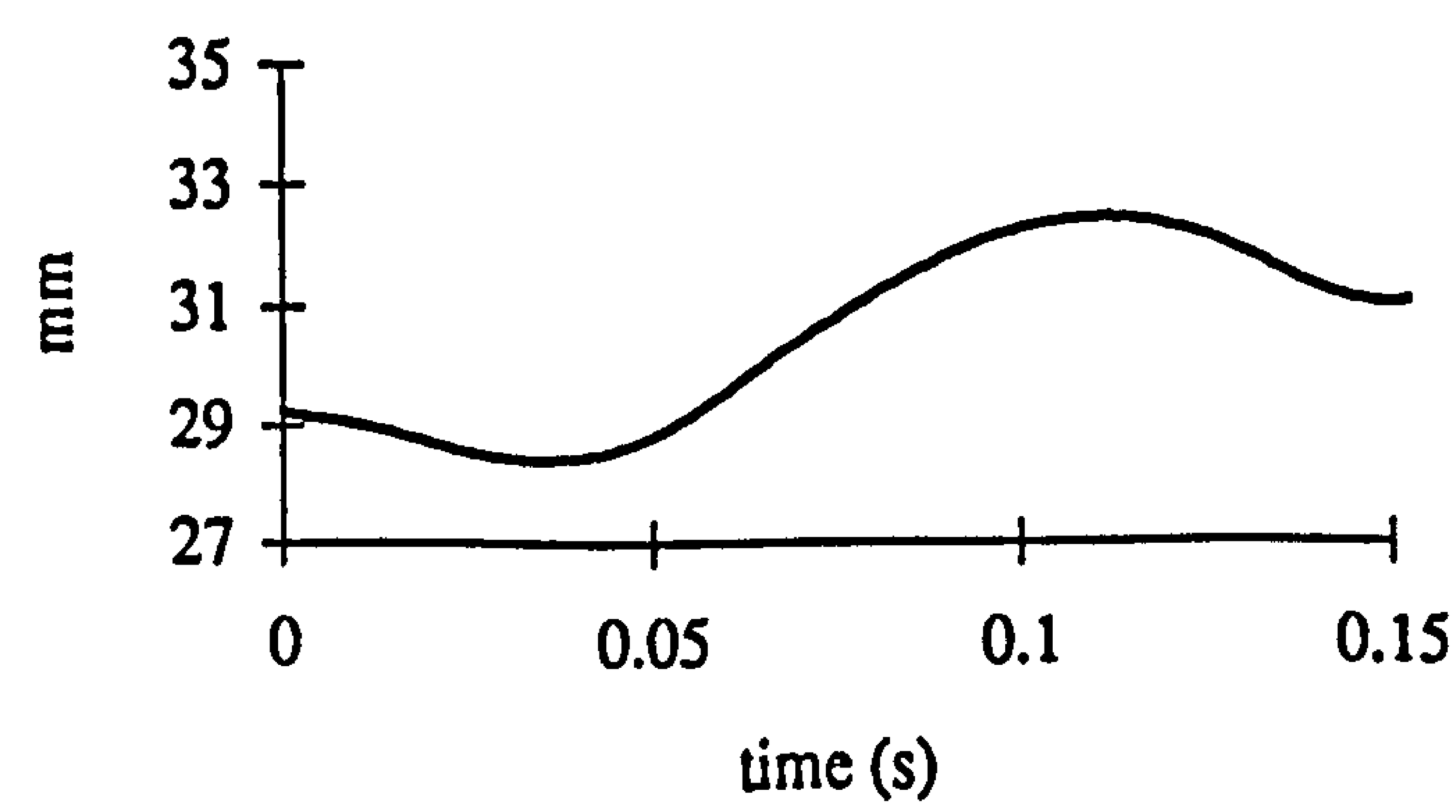
The attachment of lifts to the rearfoot and the forefoot resulted in a decrease in the time of occurrence of maximum ankle plantar-flexion moment for all but Subject 1 (Table 6.9). Compared with zero heel lift, both heel lift conditions caused an increase in this occurrence time for all but Subject 1. Since no significant differences in total stance time were found across heel lift conditions (Table 6.4), this also indicated an increase in the absolute time of occurrence of maximum ankle plantar-flexion moment with increased heel lift. Thus, for all seven of the elite distance runners used as subjects in the present study, heel lift intervention resulted in an increase in the time of occurrence of maximum ankle plantar-flexion moment as a percentage total stance time, and in absolute time. The heel lift which resulted in the most increase was not consistent across subjects, highlighting the subject specific response to heel lift manipulation.



(i) Constant = 0



(ii) Constant = 0.2



(iii) Constant = 1

Figure 6.10 The influence of smoothing parameters on Achilles moment arm (Subject 3)

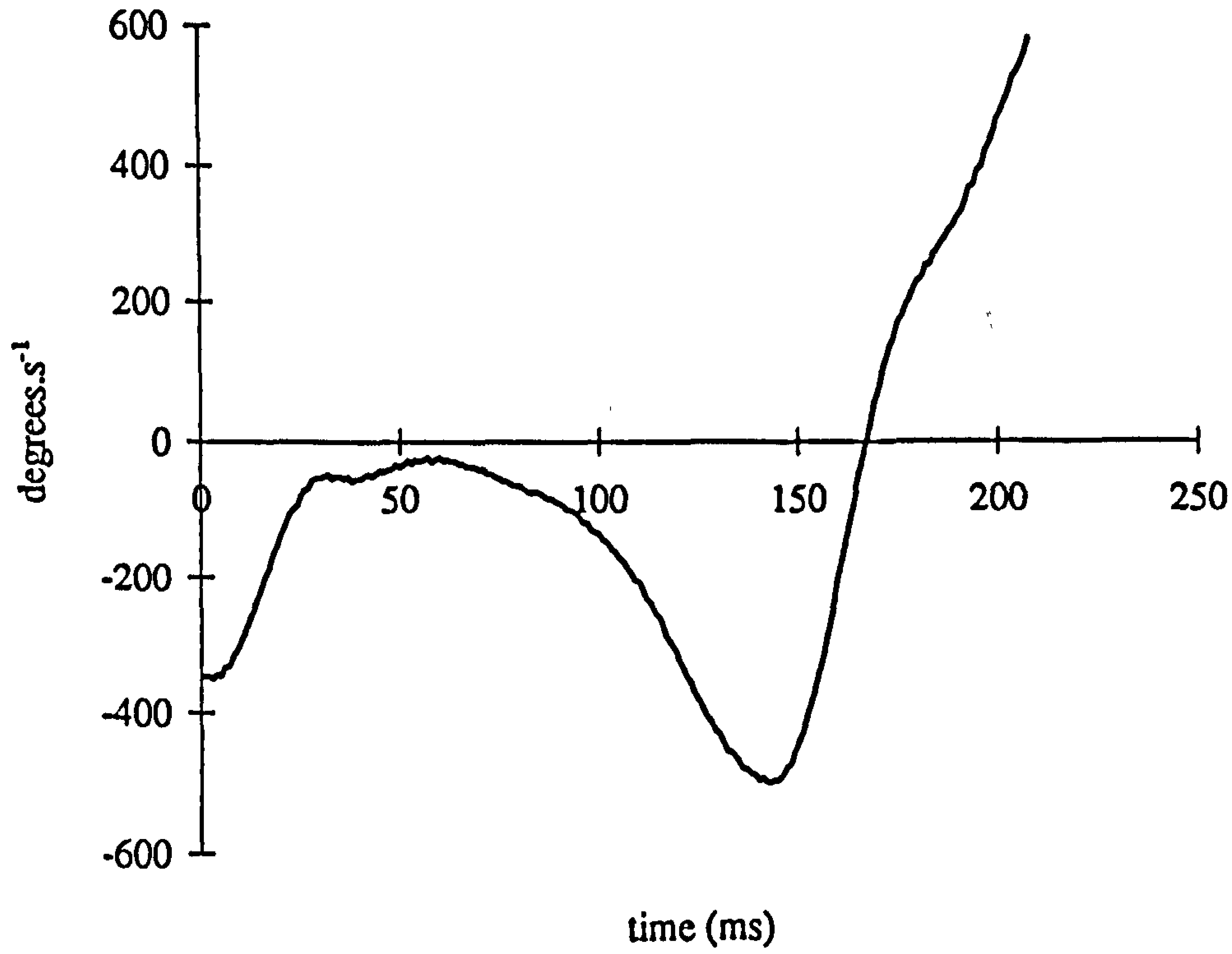
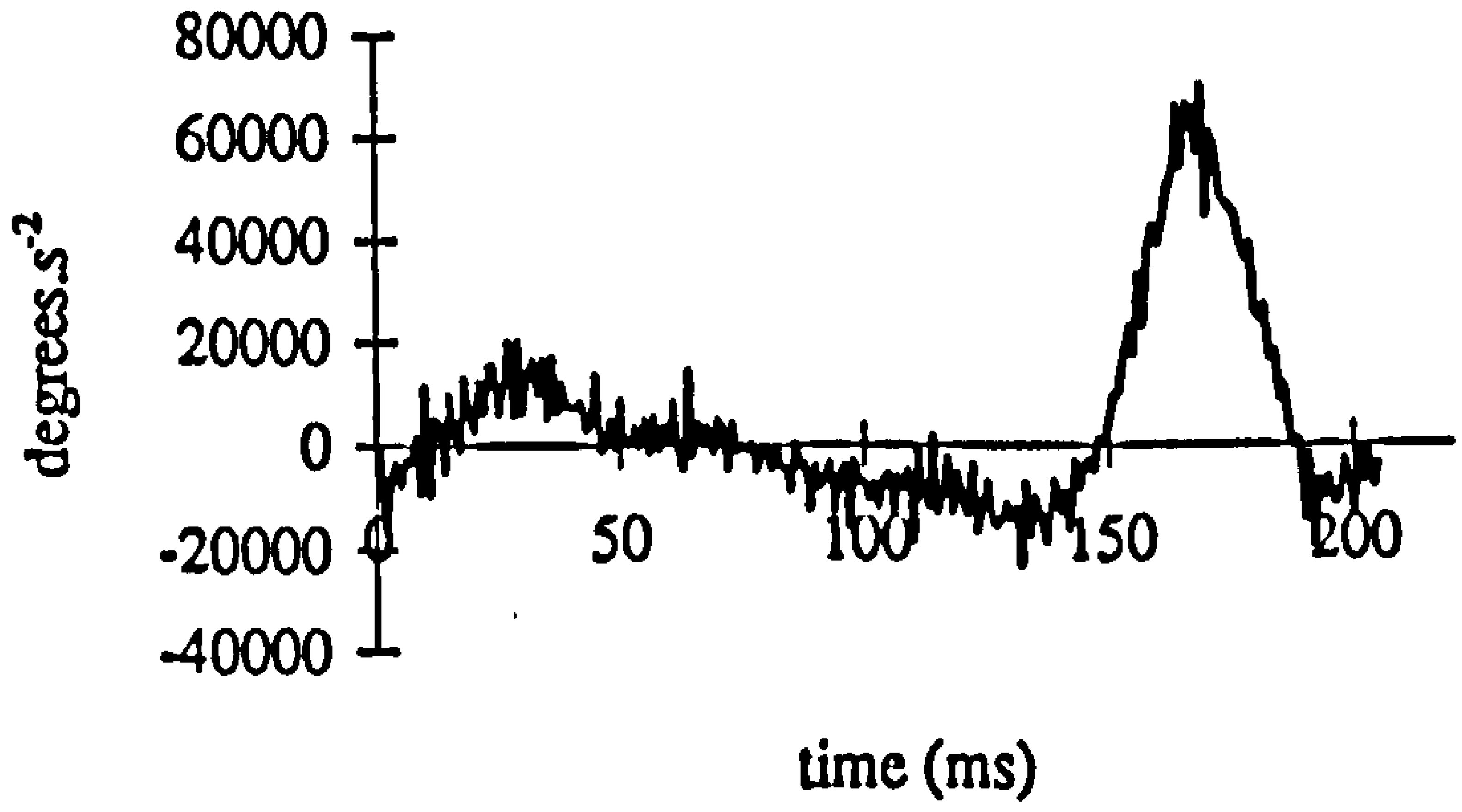
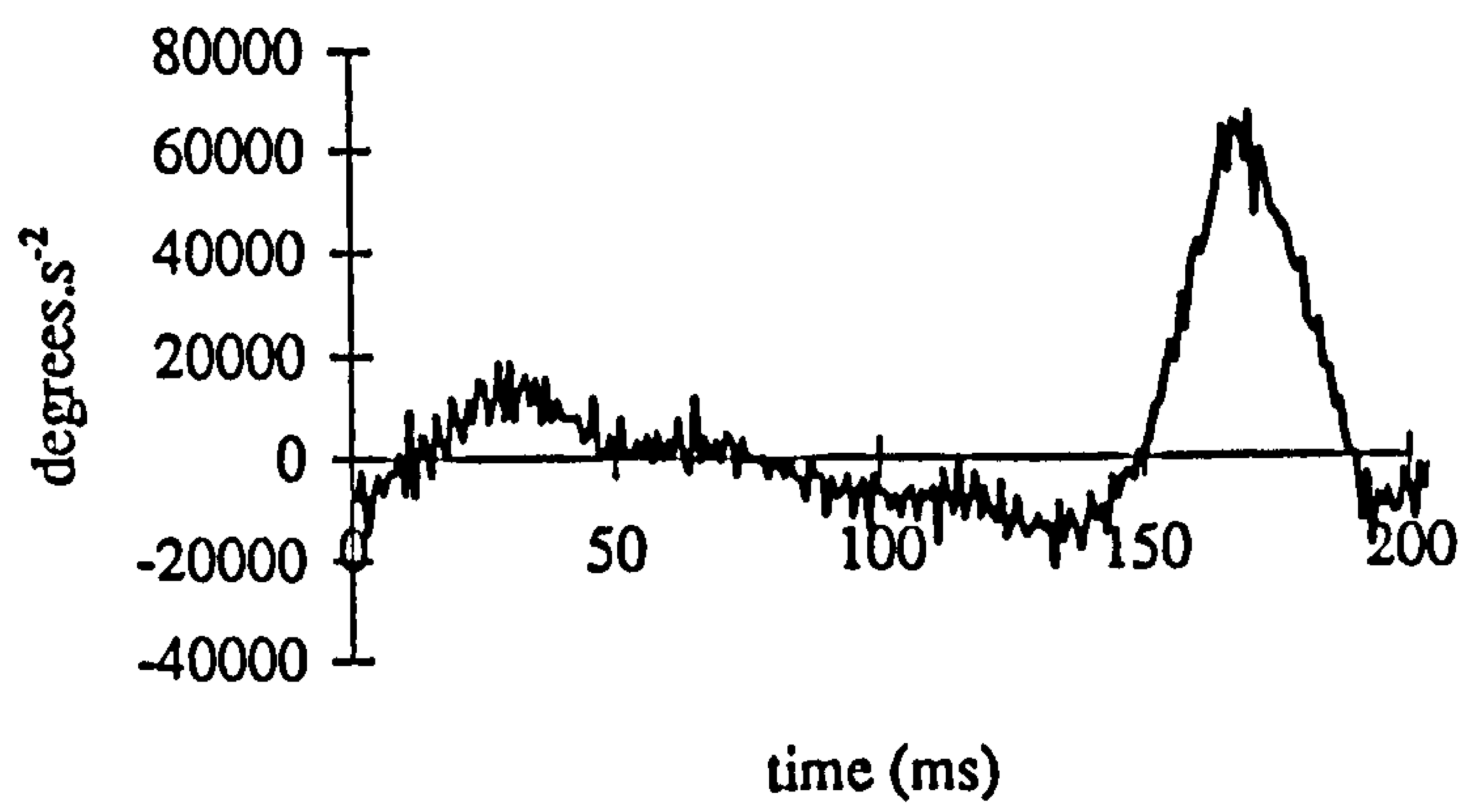


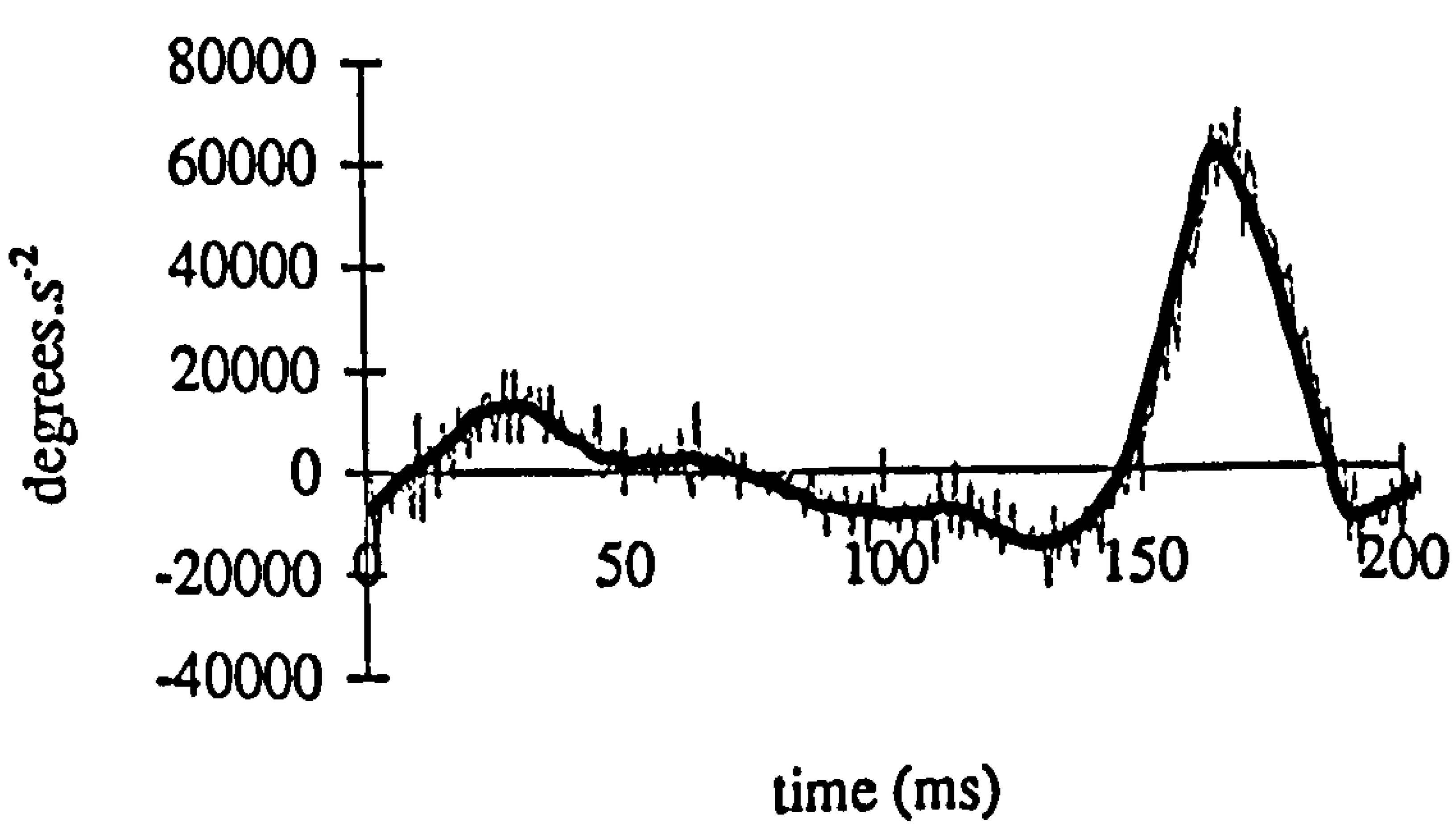
Figure 6.11 Angular velocity of foot segment using a two point procedure (Subject 7)



(i) Two point procedure

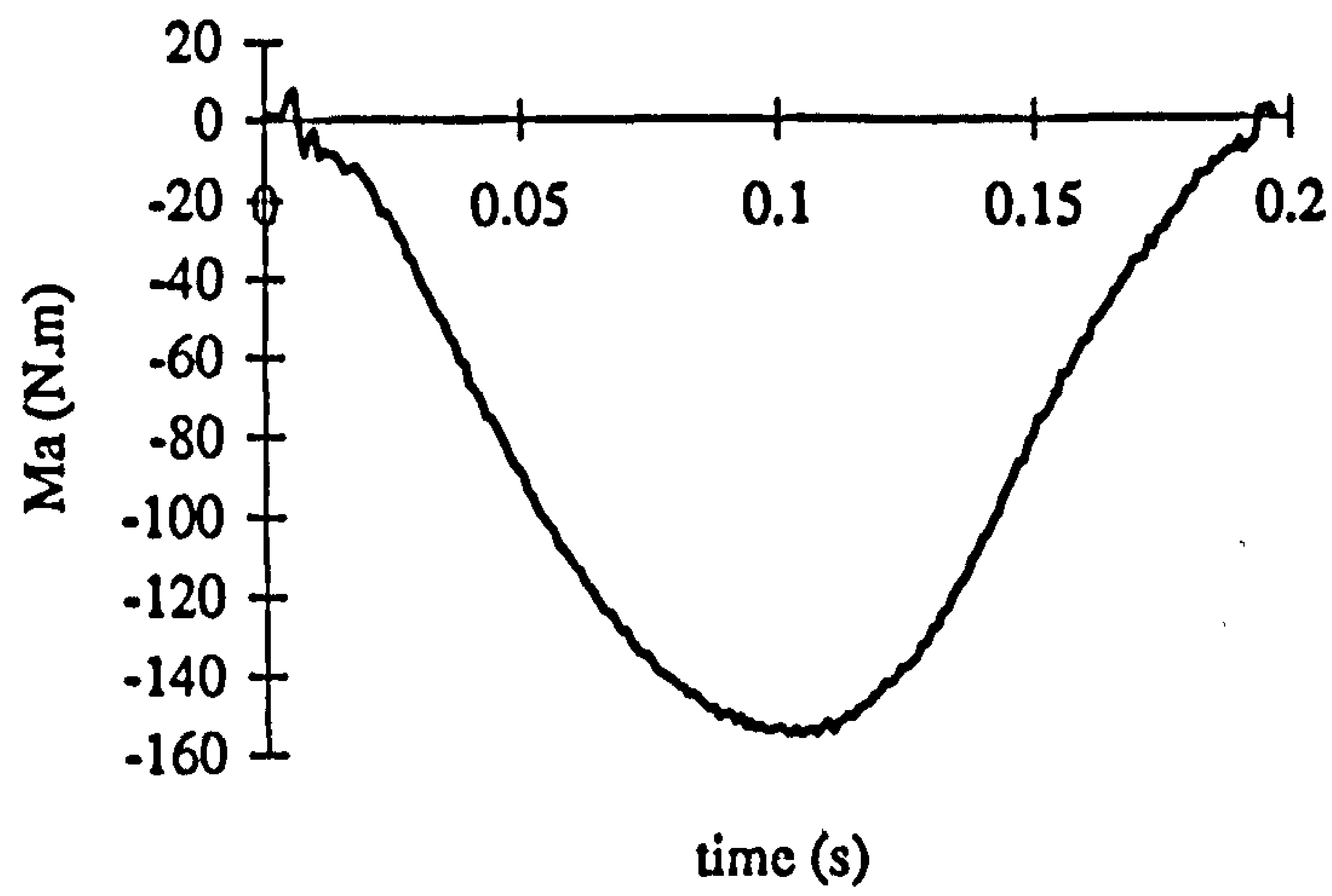


(ii) Five point procedure

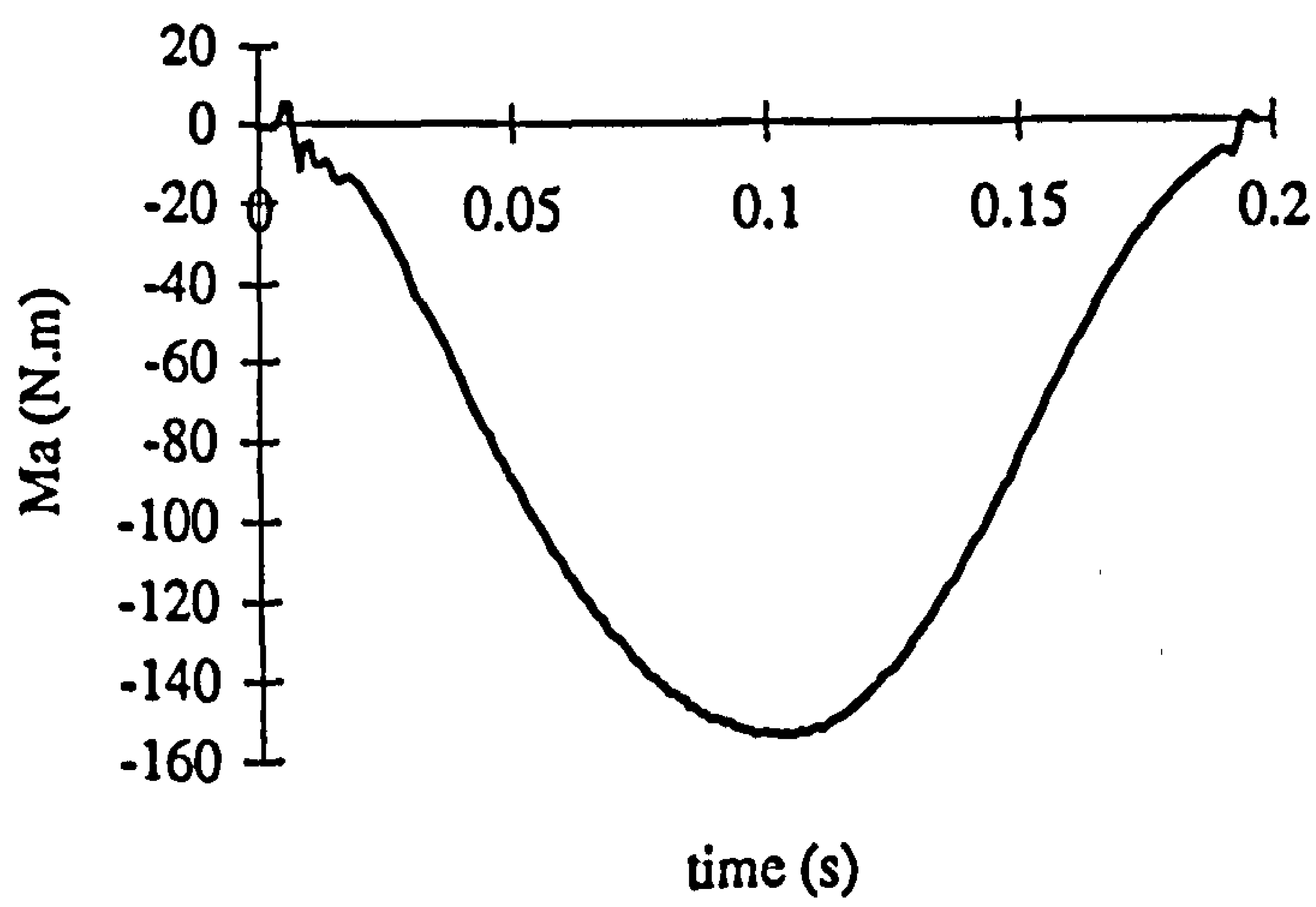


(iii) Nine point procedure

Figure 6.12 The influence of using different algorithms for acceleration calculation

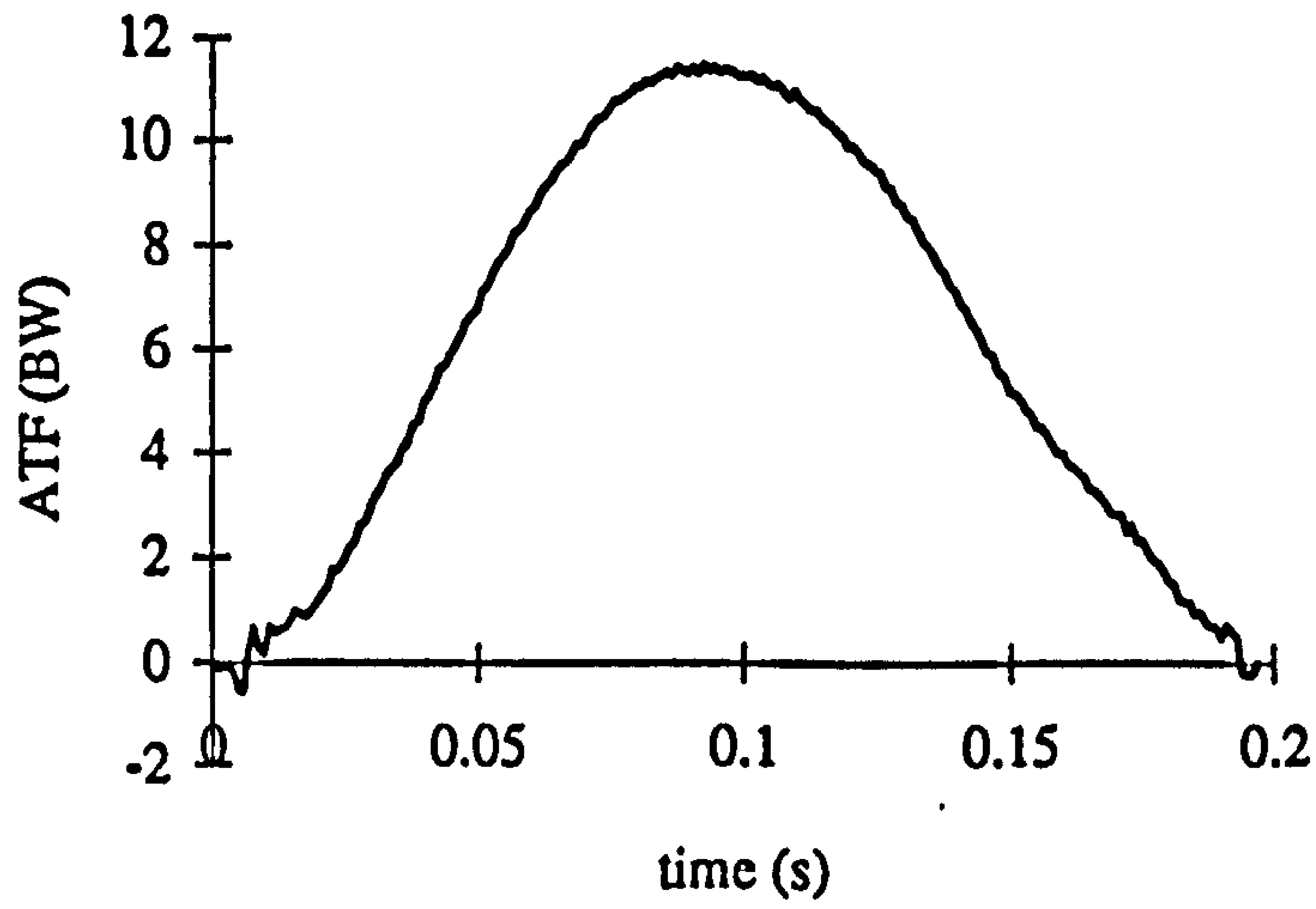


(i) Inverse dynamics

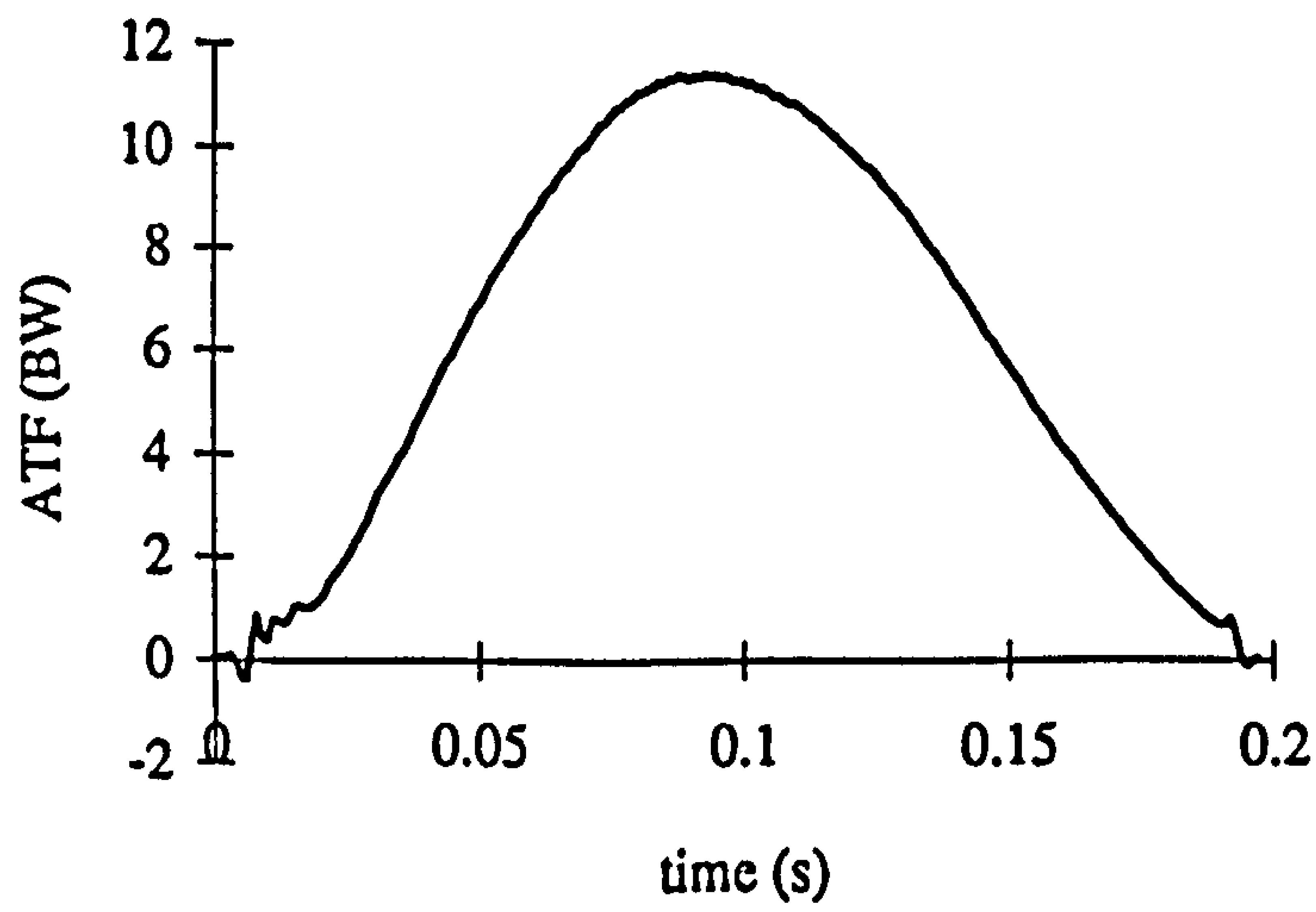


(ii) Quasi-static

Figure 6.13 Ankle moments (i) inverse dynamics (ii) and quasi-static

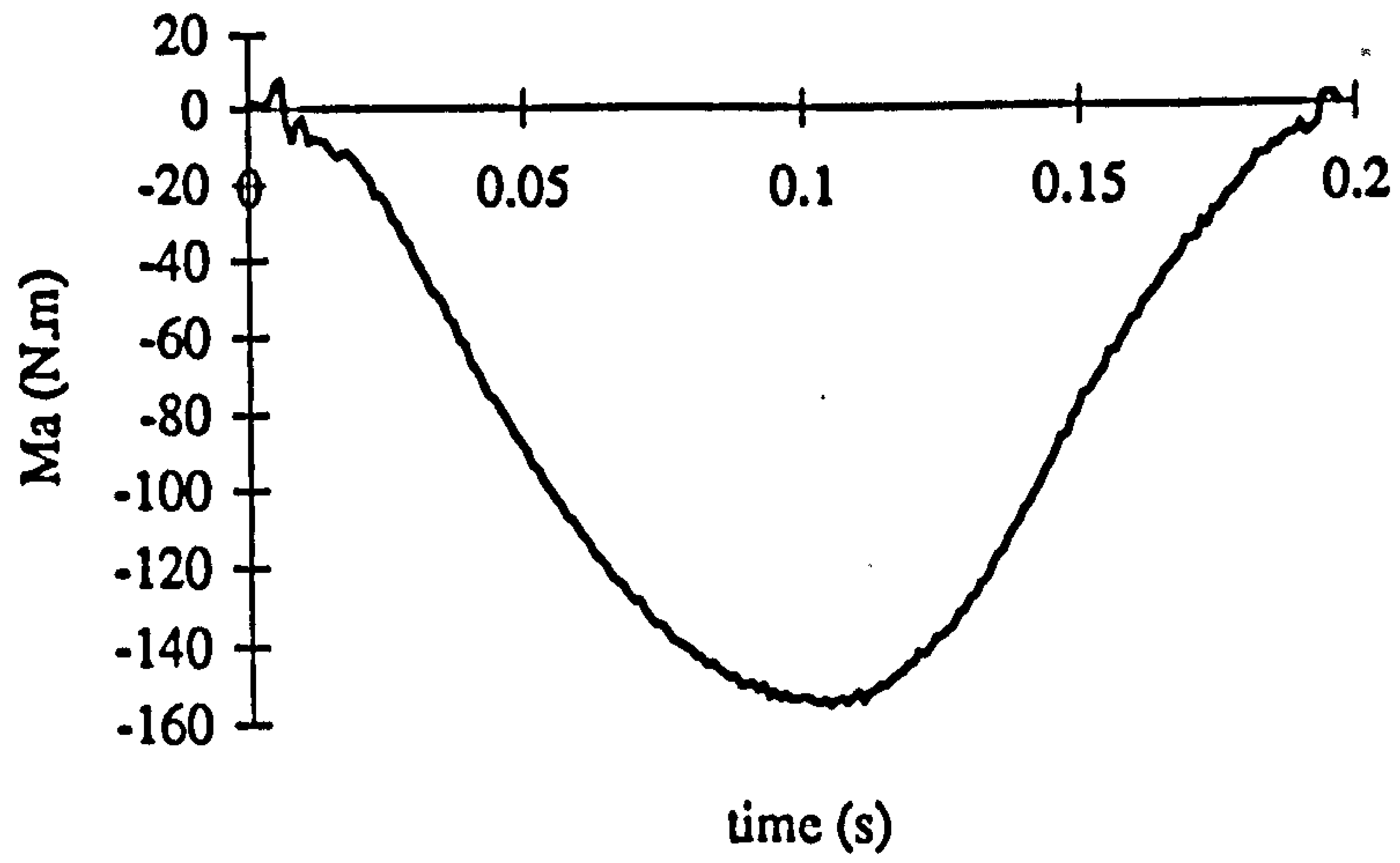


(i) Inverse dynamics

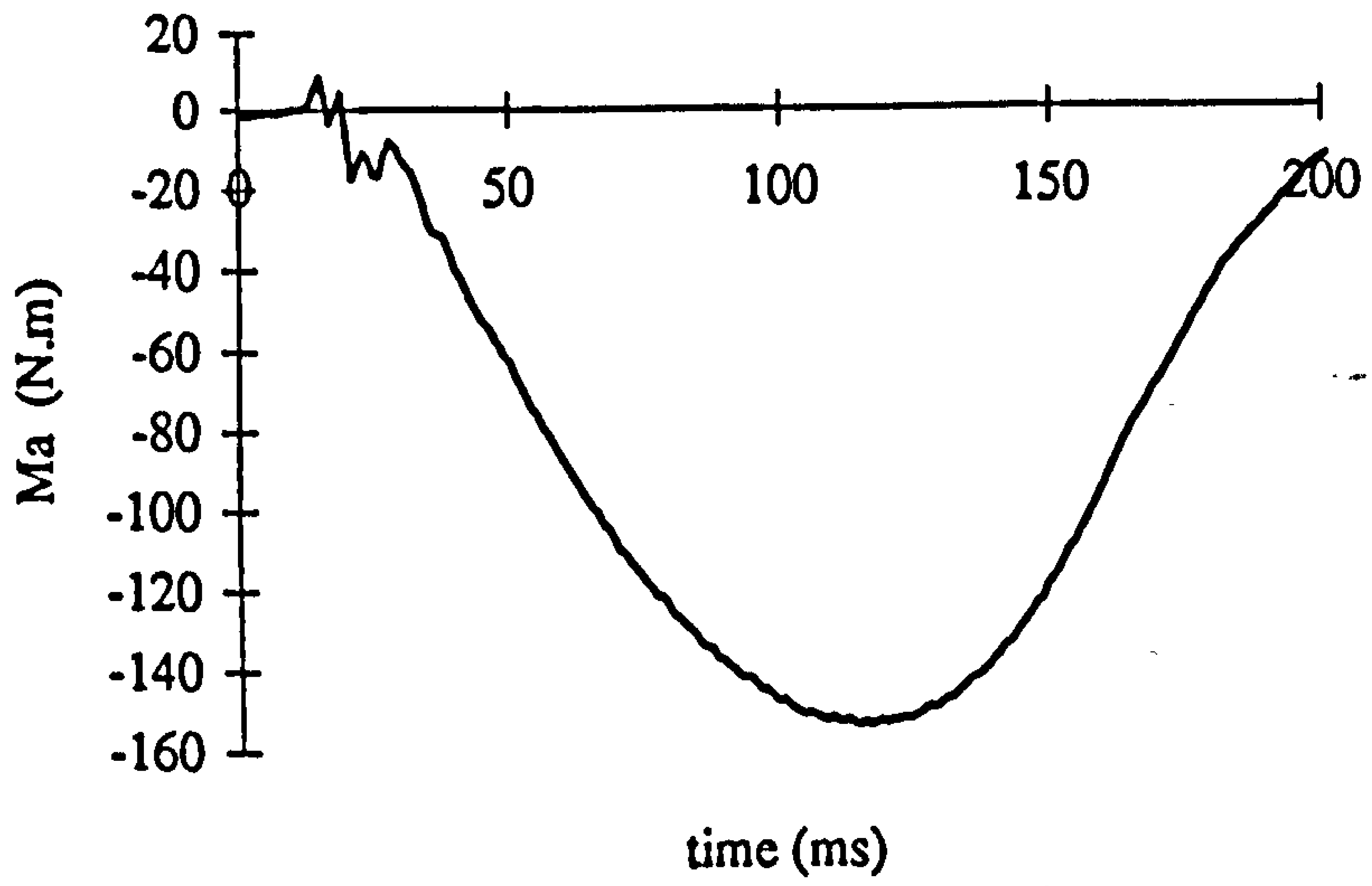


(ii) Quasi-static

Figure 6.14 Achilles tendon force using inverse dynamcs (i) and quasi-static (ii) methods



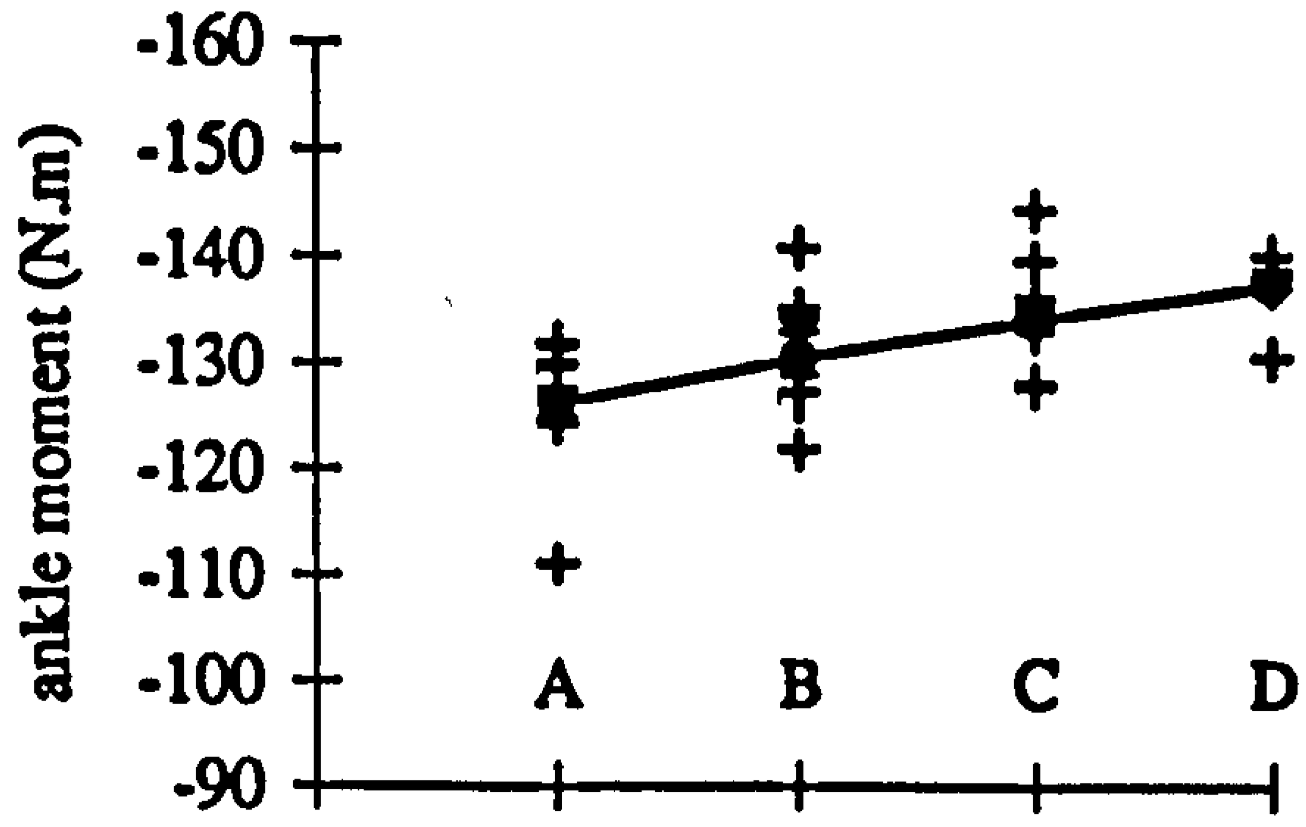
(i) Subject 3



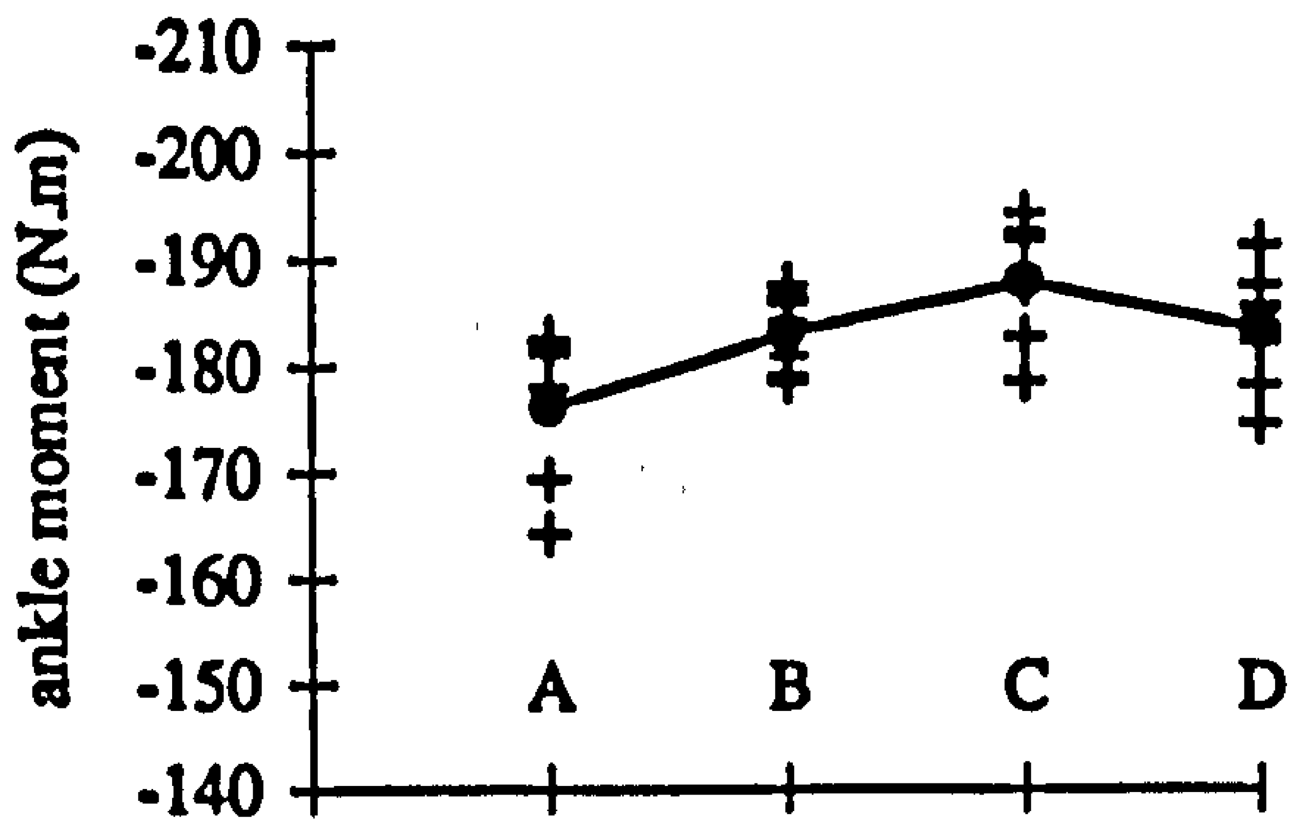
(ii) Subject 5

Figure 6.15 Typical ankle moment time histories

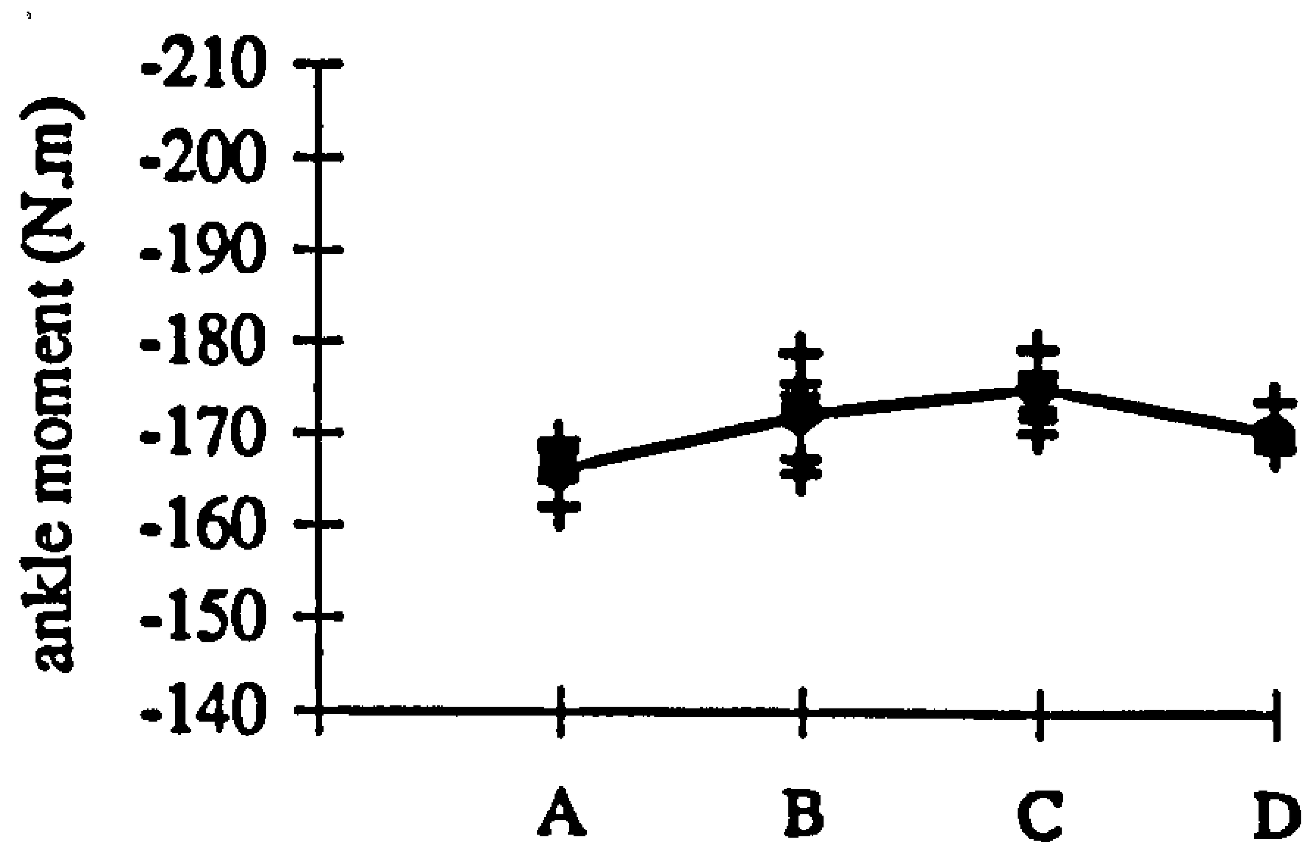
A - barefoot B - zero heel lift C - 7.5 mm heel lift D - 15 mm heel lift



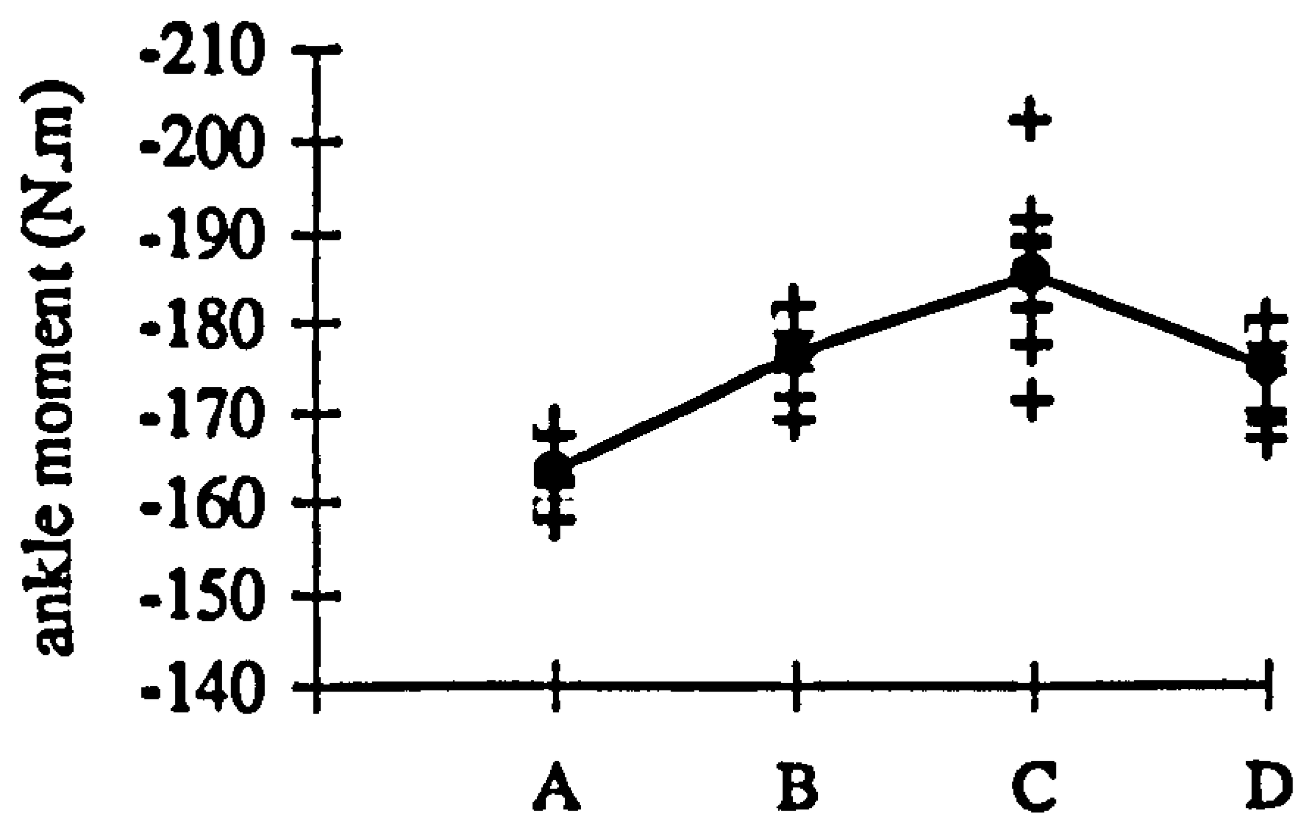
(i) Subject 1



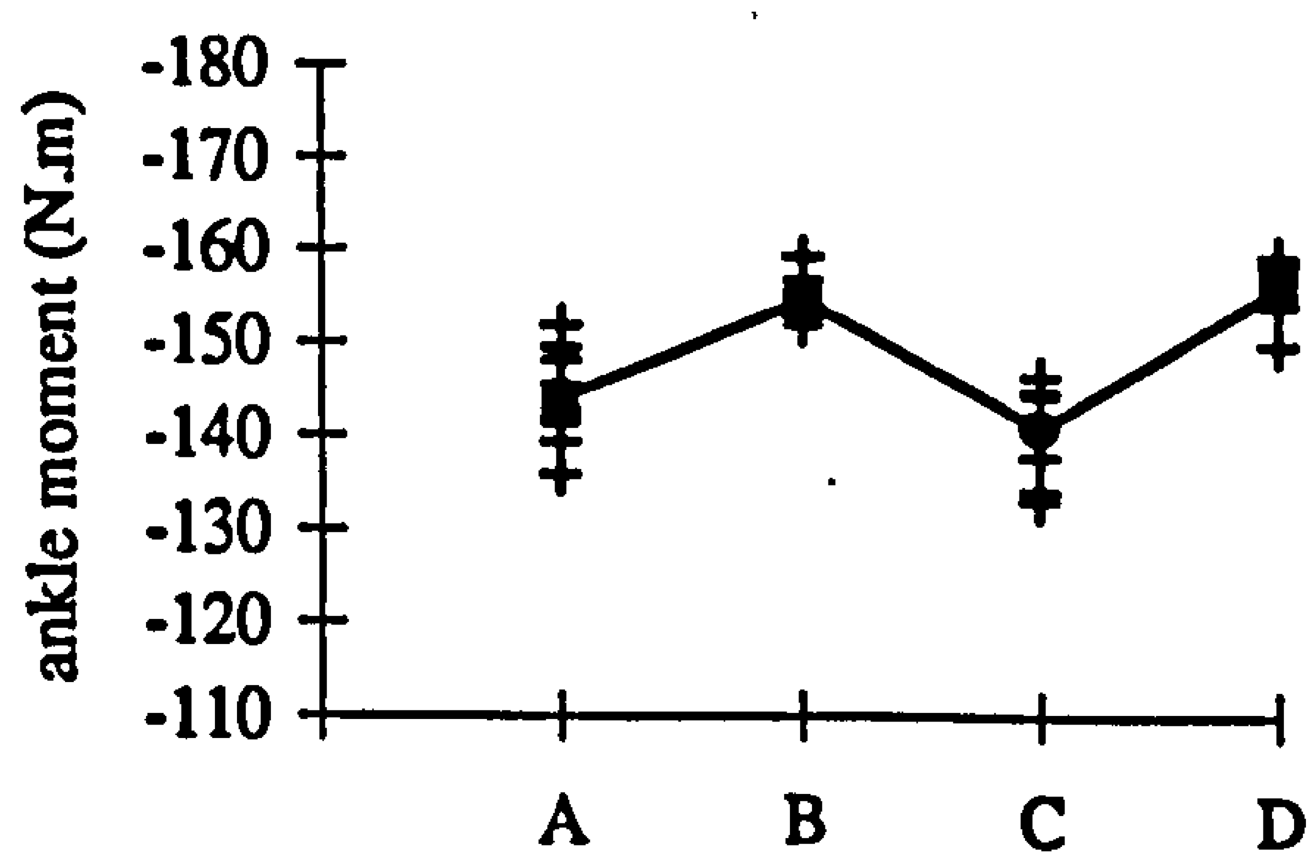
(ii) Subject 2



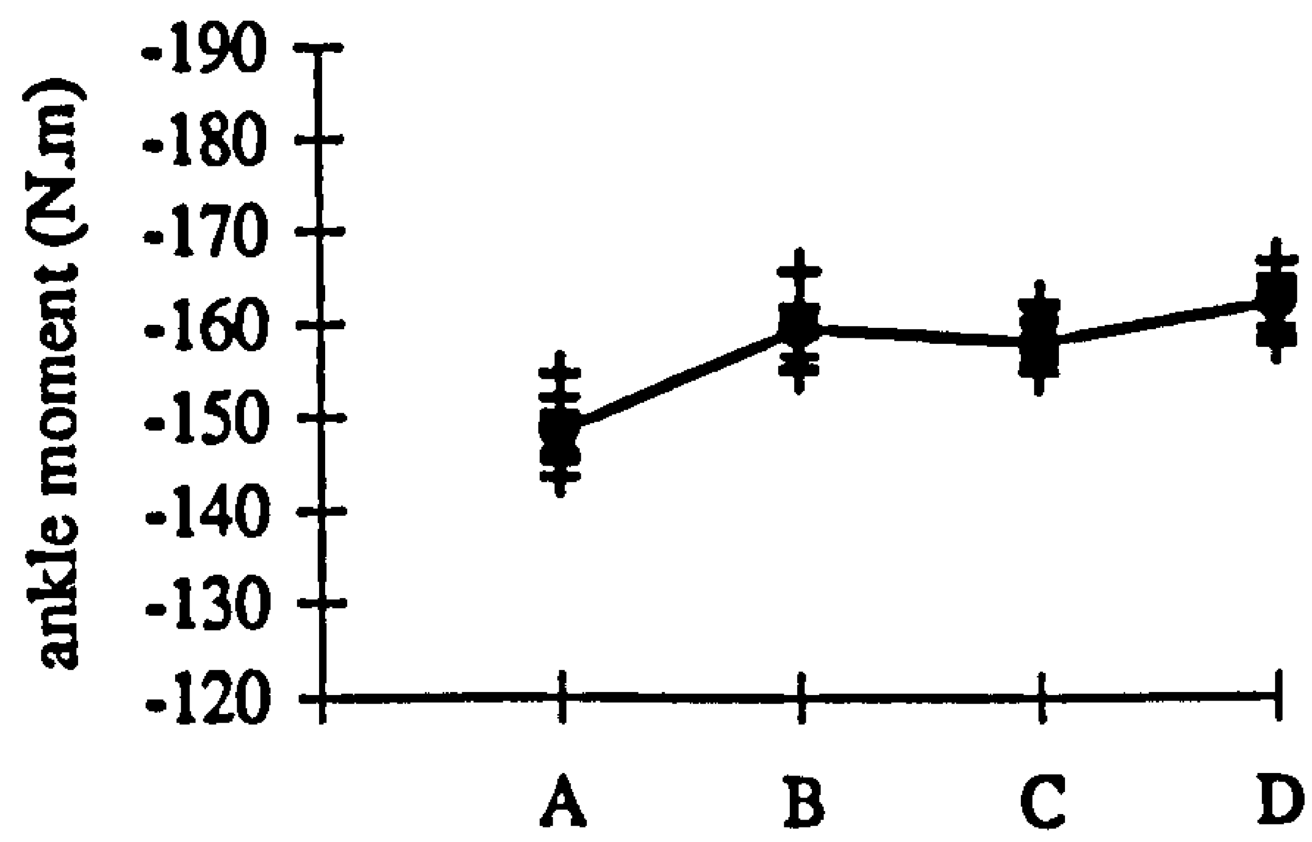
(iii) Subject 3



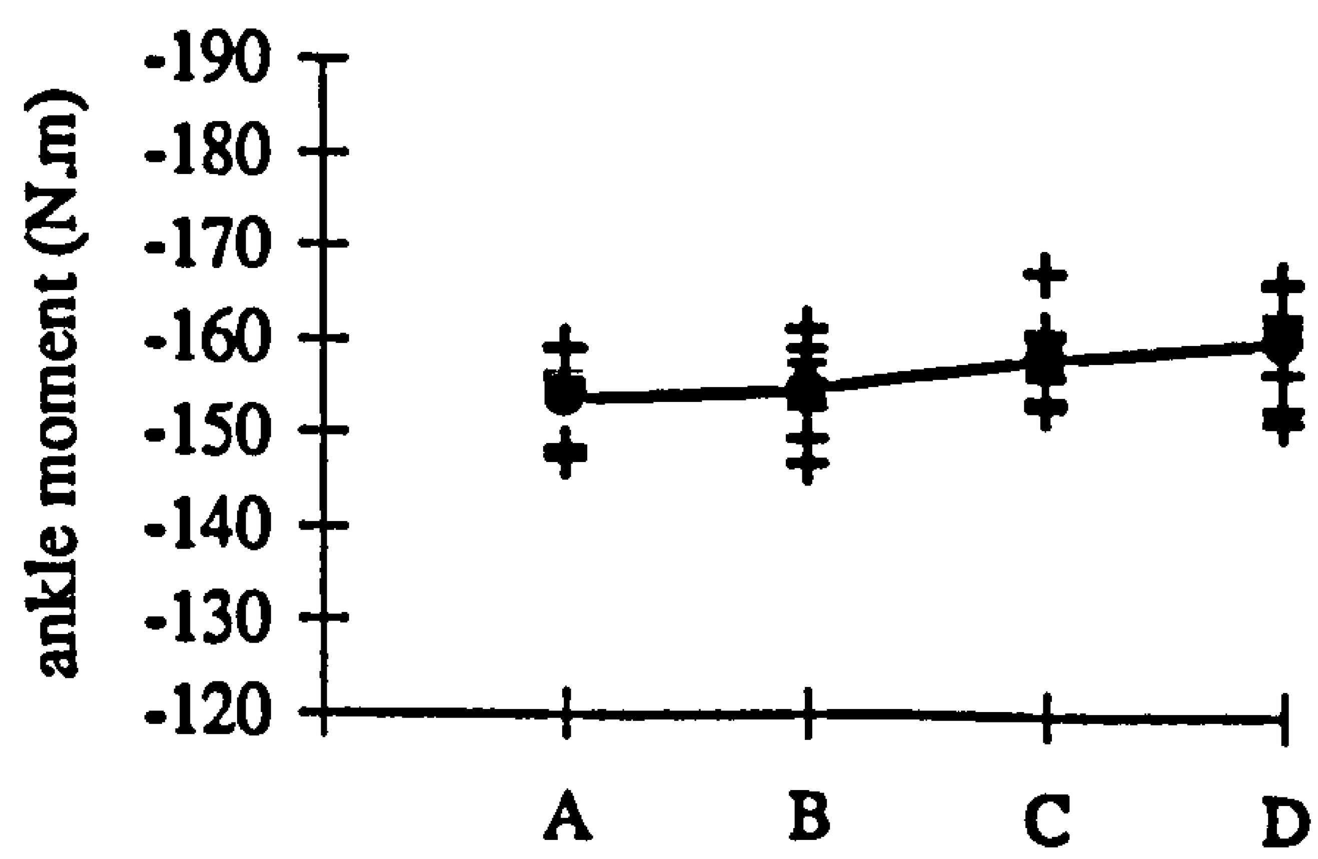
(iv) Subject 4



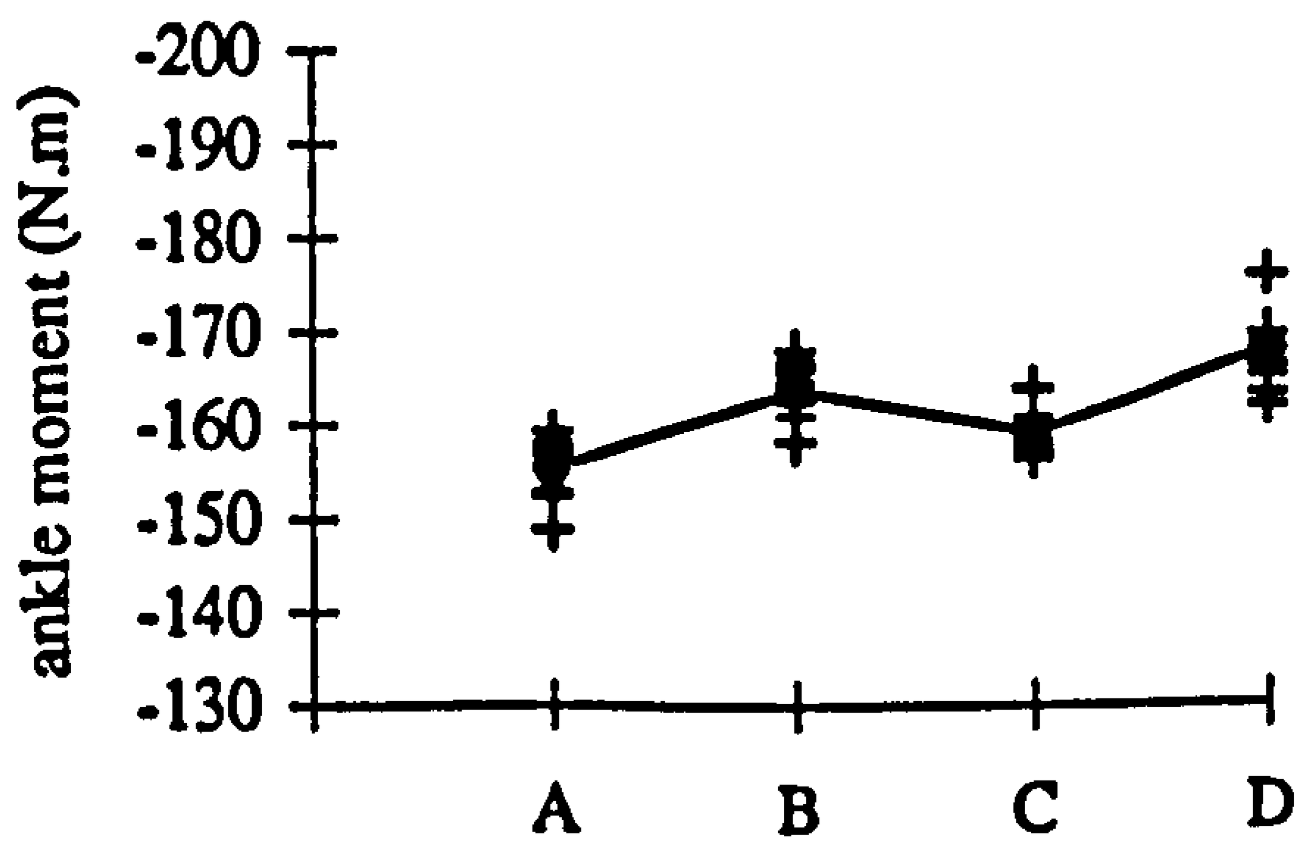
(v) Subject 5



(vi) Subject 6



(vii) Subject 7



(viii) Subject 8

mean value
+ Individual values

Figure 6.16 Peak ankle plantar-flexion moment for each subject/condition

Table 6.7 Mean peak plantar-flexion ankle moment magnitude for each subject/ condition combination in N.m (SD)
(A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

Condition	A	B	C	D
Subject 1*	126.3 (6.8)	130.5 (5.7)	131.4 (9.9)	137.2 (3.4)
Subject 2*	177.5 (7.2)	182.8 (4.4)	187.9 (5.1)	183.5 (6.1)
Subject 3*	164.9 (6.6)	172.6 (4.0)	176.1 (3.0)	170.7 (1.5)
Subject 4*	163.7 (3.7)	177.1 (4.1)	186.5 (8.7)	175.4 (4.1)
Subject 5*	144.8 (4.9)	155.5 (2.3)	142.5 (5.4)	157.2 (2.8)
Subject 6*	149.1 (3.2)	159.9 (3.1)	158.6 (2.9)	163.1 (2.8)
Subject 7	154.3 (3.8)	156.3 (3.6)	159.2 (4.0)	160.8 (4.7)
Subject 8*	156.1 (3.1)	164.6 (2.9)	159.6 (2.0)	168.4 (4.4)

*$p<0.05$ Subject 1: B versus D
Subject 2: A versus B
Subject 3: A versus B
Subject 4: B versus C; C versus D
Subject 5: A versus B; B versus C
Subject 6: A versus B; C versus D
Subject 7: none
Subject 8: A versus B; B versus C; B versus D; C versus D

Table 6.8 Mean peak ankle dorsi-flexion moment magnitude for each subject/condition combination in N.m (SD)
(A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

Condition	A	B	C	D
Subject 1*	26.8 (5.7)	28.3 (9.8)	22.5 (5.7)	31.2 (8.1)
Subject 2	21.7 (5.3)	19.7 (3.2)	22.0 (3.8)	17.3 (4.3)
Subject 3				3.1 (1.4)
Subject 4*	5.4 (2.9)	7.9 (6.1)	3.8 (2.4)	16.4 (5.0)
Subject 5*		3.4 (2.8)	13.9 (4.8)	13.5 (2.9)
Subject 6*	26.6 (3.6)	6.5 (6.0)	10.7 (4.5)	9.1 (8.2)
Subject 7*	12.7 (3.4)	22.0 (3.3)	17.5 (3.3)	14.2 (3.4)
Subject 8				

NB. Where no value is provided, no dorsi-flexion moment occurred

*$p<0.05$ Subject 1: C versus D
Subject 2: none
Subject 3: none
Subject 4: B versus D; C versus D
Subject 5: B versus C; B versus D
Subject 6: A versus B
Subject 7: A versus B; B versus C; B versus D
Subject 8: none

Table 6.9 Mean time of occurrence of maximum ankle plantar-flexion moment for each subject/condition combination (percentage total stance time)
(A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

Condition	A	B	C	D
Subject 1	60.4	60.6	58.5	57.8
Subject 2	52.6	50.9	51.1	52.2
Subject 3	47.0	45.3	46.9	48.5
Subject 4	50.0	47.7	47.8	50.9
Subject 5	50.2	44.1	59.6	53.7
Subject 6	51.0	47.0	48.1	46.7
Subject 7	50.5	48.4	50.0	52.2
Subject 8	49.8	46.3	47.5	47.0

(vii) Maximum Achilles Tendon Force

The tendon calliper measurement obtained for Subject 7 was 19 mm (Table 6.1). The distance from the skin covering the posterior aspect of the tendon to the line of action was 3.9 mm (Chapter 5). A scaling factor of (3.9/19) was therefore used with the calliper measurement for each subject to obtain Achilles tendon line of action displacement distances.

Examples of time histories of Achilles tendon force during ground contact are provided in Figure 6.17. Figure 6.18 illustrates the changes in magnitude of maximum Achilles tendon force over the four conditions, with each subject presented individually. Magnitude and standard deviation for maximum Achilles tendon force for each subject and each condition are provided in Table 6.10 in bodyweight units (BW). For all subjects there was an increase in maximum Achilles tendon force in response to the raising of the entire foot compared with the barefoot condition, with this increase being significant for four of the eight subjects.

It was found that the introduction of a heel lift can influence maximum Achilles tendon forces. Four subjects demonstrated a significant decrease in maximum Achilles tendon force for one of the heel lift conditions compared with zero heel lift, whilst one subject showed a significant increase. For all but one subject, there was a decrease or no change in maximum Achilles tendon force for the 15 mm heel lift condition compared with zero heel lift.

The times of occurrence of the maximum Achilles tendon forces are provided as a percentage of total stance time and in milliseconds in Table 6.11, for each subject/condition combination. With the exception of Subject 1, maximum Achilles tendon force was found to occur between 44% and 53% of the total stance time for all subjects across conditions. For Subject 1, the time of occurrence of maximum Achilles tendon force was later than observed for the remaining subjects. For the seven other subjects, the attachment of lifts to the rearfoot and forefoot resulted in a decrease in the time of occurrence of maximum Achilles tendon force, compared with the barefoot condition, with four of these differences being significant. There was a trend for the time of occurrence of maximum Achilles tendon force, as a percentage of total stance time, to be increased with the use of a heel lift. Subject 3 did not demonstrate this pattern, but the variations seen for this subject across conditions were small (1%). In contrast with the results for these seven subjects, Subject 1 showed an increase in the time of occurrence of maximum Achilles tendon force when the rearfoot and the forefoot were raised compared with the barefoot condition, and a decrease when heel lifts were introduced compared with zero heel lift.

The magnitude of Achilles tendon moment arm, ankle moment, GRF and GRF moment arm at the time of maximum Achilles tendon force are provided in Table 6.12. For all subjects, the ankle moment at the time of maximum Achilles tendon force was increased for the zero heel lift condition compared with the barefoot condition, contributing to the observed increases in maximum Achilles tendon force. These increases in ankle moment were generally the combined result of increases in both GRF magnitude and moment arm. Both increases and decreases in the Achilles tendon moment arm length were demonstrated for the lift condition compared with the barefoot condition.

Comparison of the changes in ankle moment at maximum Achilles tendon force across heel lift conditions with the changes in maximum Achilles tendon force, demonstrated that all but two subjects showed the same patterns in these two variables, indicating that the observed changes in maximum Achilles tendon force with heel lift variation were contributed to by changes in ankle moment magnitude. In general, the changes in Achilles tendon moment arm with heel lift variation were found to contribute to the observed changes in maximum Achilles tendon force value. Thus, the results of this study indicate that changes in maximum Achilles tendon force occurring with heel lift variation are generally contributed to by changes in both ankle joint moment and Achilles tendon moment arm length.

For Subject 7, a calliper tendon measurement of 19 mm was obtained at the location corresponding to an Achilles tendon cross-sectional area of 77 mm^2. Using tendon calliper measurements to provide scaled cross-sectional area values for each subject, cross-sectional areas ranged from 58 mm^2 to 104 mm^2. The maximum tendon stress values calculated for barefoot running ranged from 49×10^6 $N.m^{-2}$ to 91×10^6 $N.m^{-2}$ (Table 6.10). A correlation coefficient of 0.78 was obtained between Achilles tendon force and Achilles tendon stress for the barefoot condition across subjects.

Table 6.10 Mean maximum Achilles tendon force for each condition and subject in BW (SD), and [maximum stress for barefoot condition ($N.m^{-2}$)] (A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

Condition	A	[max stress]	B	C	D
Subject 1	9.8 (0.7)	[62]	10.2 (0.4)	11.0 (1.0)	9.6 (0.6)
Subject 2*	11.4 (0.8)	[63]	13.8 (1.1)	12.2 (0.6)	12.2 (1.0)
Subject 3	9.8 (0.5)	[63]	9.9 (0.4)	10.1 (1.5)	9.7 (0.6)
Subject 4*	7.3 (0.3)	[67]	8.0 (0.7)	8.4 (0.6)	7.9 (0.4)
Subject 5*	12.2 (0.9)	[56]	13.7 (1.2)	11.5 (0.8)	12.9 (1.6)
Subject 6*	6.1 (0.2)	[49]	6.8 (0.3)	6.5 (0.4)	7.0 (0.3)
Subject 7*	9.9 (0.5)	[68]	10.0 (0.5)	10.1 (0.4)	10.5 (0.4)
Subject 8*	17.4 (1.5)	[91]	17.4 (2.9)	17.2 (2.0)	15.7 (0.8)

*$p<0.05$ Subject 1: none
Subject 2: A versus B; B versus C; B versus D
Subject 3: none
Subject 4: A versus B
Subject 5: A versus B; B versus C; C versus D
Subject 6: A versus B; B versus C; C versus D
Subject 7: B versus D
Subject 8: B versus D; C versus D

Table 6.11 Time of occurrence of maximum Achilles tendon force for each condition and subject in milliseconds and as percentage total stance time (SD) (A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

Condition	A	B	C	D
Subject 1*	112 (10)	109 (9)	105 (7)	111 (7)
	58.5 (3.3)	60.5 (3.5)	55.6 (3.4)	57.9 (3.2)
Subject 2	107 (6)	111 (3)	115 (5)	115 (7)
	51.2 (3.3)	50.1 (0.8)	50.8 (1.7)	50.7 (1.8)
Subject 3	98 (3)	94 (4)	91 (6)	93 (9)
	44.7 (1.4)	44.0 (2.0)	43.4 (4.4)	43.0 (3.8)
Subject 4*	112 (5)	102 (5)	106 (6)	111 (9)
	48.6 (2.4)	46.0 (1.8)	46.7 (3.0)	48.9 (2.5)
Subject 5*	105 (8)	115 (9)	113 (8)	122 (6)
	48.1 (2.2)	46.7 (3.0)	51.7 (3.5)	52.8 (3.8)
Subject 6*	104 (3)	100 (6)	104 (6)	100 (10)
	49.7 (1.1)	46.7 (1.4)	48.3 (2.0)	46.2 (2.9)
Subject 7*	106 (6)	103 (5)	108 (6)	112 (8)
	49.8 (1.3)	47.4 (1.4)	48.7 (1.1)	50.2 (2.2)
Subject 8*	95 (8)	83 (13)	95 (5)	90 (5)
	47.0 (2.5)	41.2 (5.8)	46.5 (1.9)	45.1 (2.4)

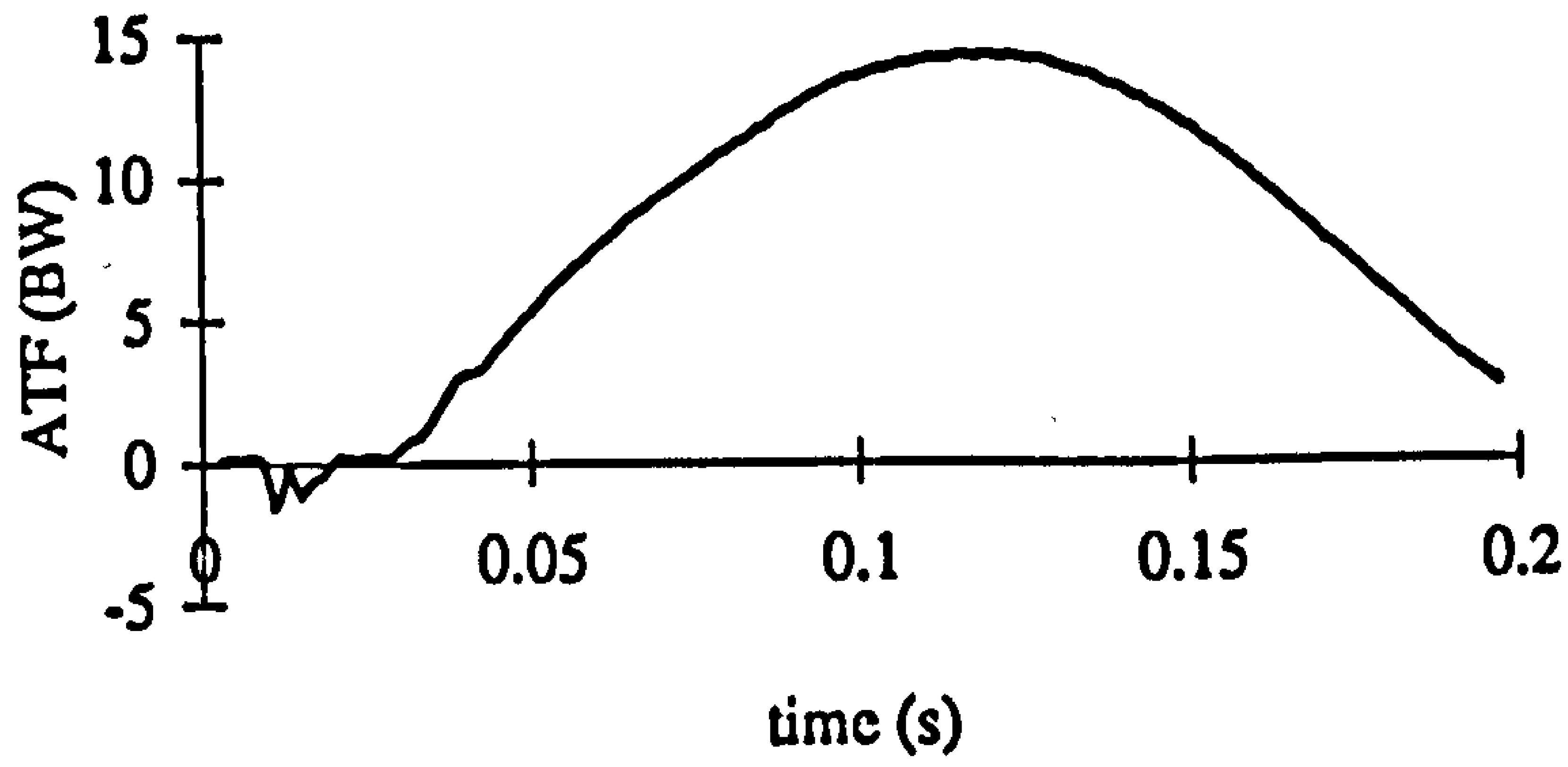
*$p<0.05$ (percentage total stance time)

Subject 1: B versus C
Subject 2: none
Subject 3: none
Subject 4: A versus B
Subject 5: B versus C; B versus D
Subject 6: A versus B
Subject 7: A versus B; B versus D
Subject 8: A versus B; B versus C

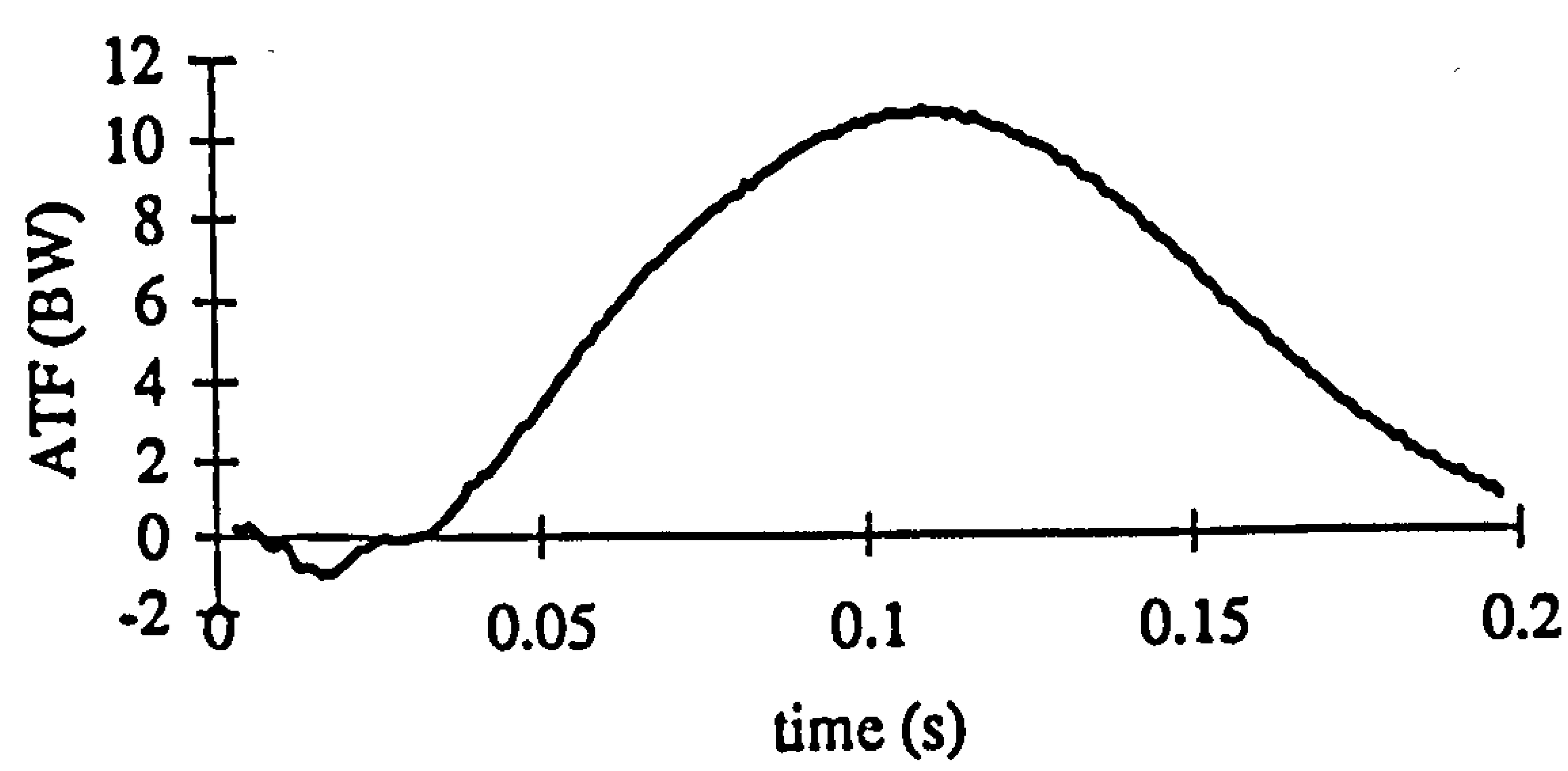
Table 6.12 Maximum Achilles tendon force (F) and contributing factors of Achilles tendon moment arm (d_1), ankle moment (M_a), resultant GRF (R), and moment arm (d_2) (A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

	Condition	A	B	C	D
Subject 1					
	F_{max} (BW)	9.8	10.2	11.0	9.6
	d_1 (mm)	24.6	25.4	24.3	27.6
	M_a (N.m)	116.1	125.5	130.1	128.9
	R (N)	888	893	925	
	d_2 (m)	0.1308	0.1406	0.1407	
Subject 2					
	F_{max} (BW)	11.4	13.8	12.2	12.2
	d_1 (mm)	30.2	25.9	29.7	29.3
	M_a (N.m)	176.4	182.6	185.8	183.0
	R (N)	1239	1214	1210	1219
	d_2 (m)	0.1424	0.1504	0.1536	0.1501
Subject 3					
	F_{max} (BW)	9.8	9.9	10.1	9.7
	d_1 (mm)	33.4	34.9	34.4	34.6
	M_a (N.m)	162.9	171.3	172.9	167.5
	R (N)	1170	1205	1191	1138
	d_2 (m)	0.1392	0.1422	0.1452	0.1472
Subject 4					
	F_{max} (BW)	7.3	8.0	8.4	7.9
	d_1 (mm)	41.5	41.2	41.3	41.1
	M_a (N.m)	163.1	176.0	185.1	174.4
	R (N)	1215	1265	1243	1199
	d_2 (m)	0.1342	0.1391	0.1489	0.1455
Subject 5					
	F_{max} (BW)	12.2	13.7	11.5	12.9
	d_1 (mm)	24.5	23.3	25.8	25.2
	M_a (N.m)	143.3	152.8	140.4	155.6
	R (N)	1107	1138	1037	1095
	d_2 (m)	0.1294	0.1343	0.1354	0.1421
Subject 6					
	F_{max} (BW)	6.1	6.8	6.5	7.0
	d_1 (mm)	45.4	43.5	45.6	43.3
	M_a (N.m)	148.0	159.4	158.0	162.4
	R (N)	1183	1199	1183	1185
	d_2 (m)	0.1251	0.1329	0.1336	0.1371
Subject 7					
	F_{max} (BW)	9.9	10.0	10.1	10.5
	d_1 (mm)	29.3	29.2	29.4	28.5
	M_a (N.m)	153.3	154.2	157.3	159.2
	R (N)	1151	1114	1112	1090
	d_2 (m)	0.1332	0.1384	0.1415	0.1461
Subject 8					
	F_{max} (BW)	17.4	17.4	17.2	15.7
	d_1 (mm)	18.1	18.5	18.8	21.8
	M_a (N.m)	153.5	157.3	158.3	166.6
	R (N)				1182
	d_2 (m)				0.140

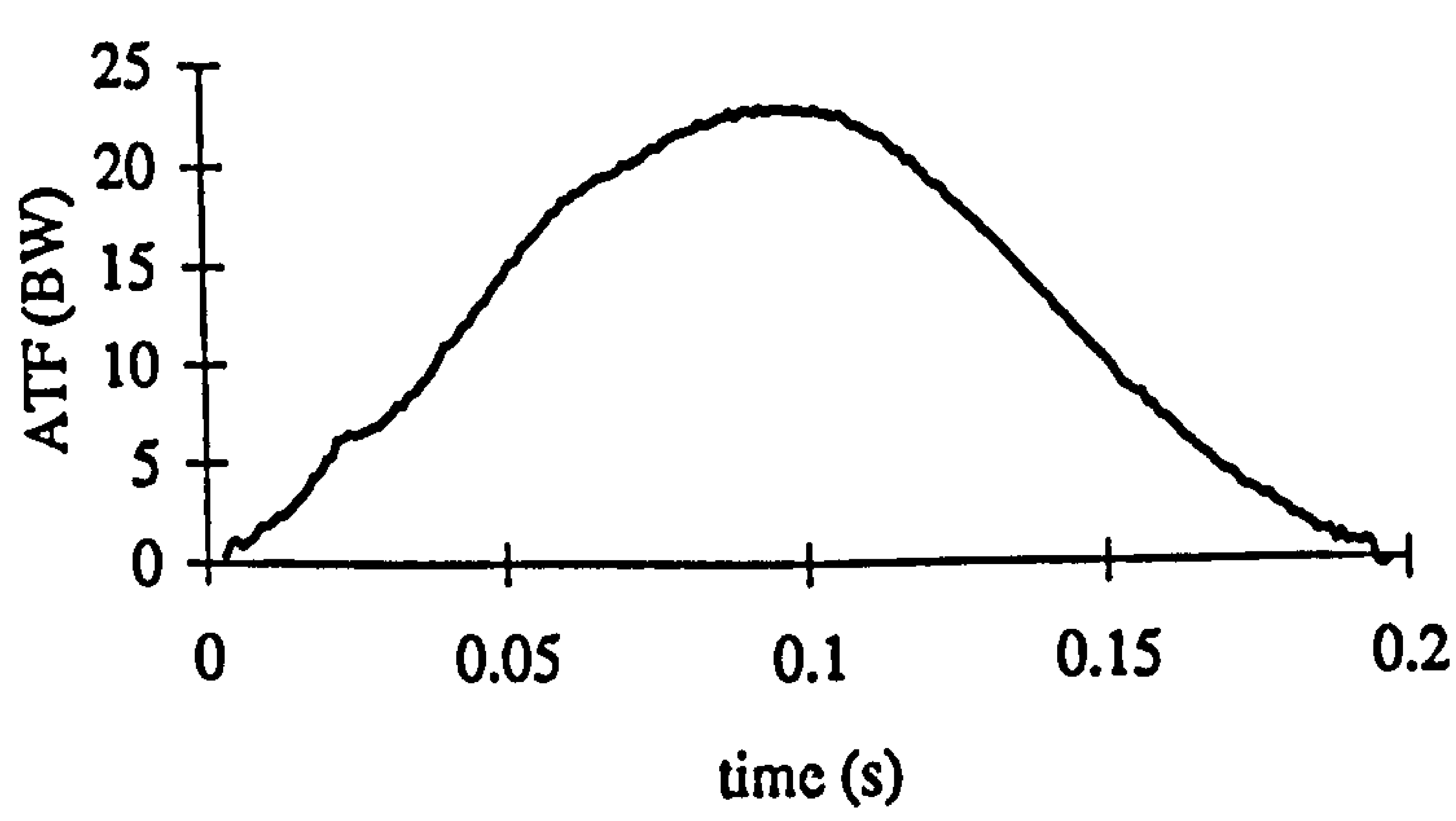
F_{max} maximum Achilles tendon force
d_1 Achilles tendon moment arm
M_a ankle moment
R resultant GRF
d_2 GRF moment arm



(i) Subject 3



(ii) Subject 5



(iii) Subject 8

Figure 6.17 Typical Achilles tendon force time histories

(A - barefoot B - zero heel lift C - 7.5 mm heel lift D - 15 mm heel lift)

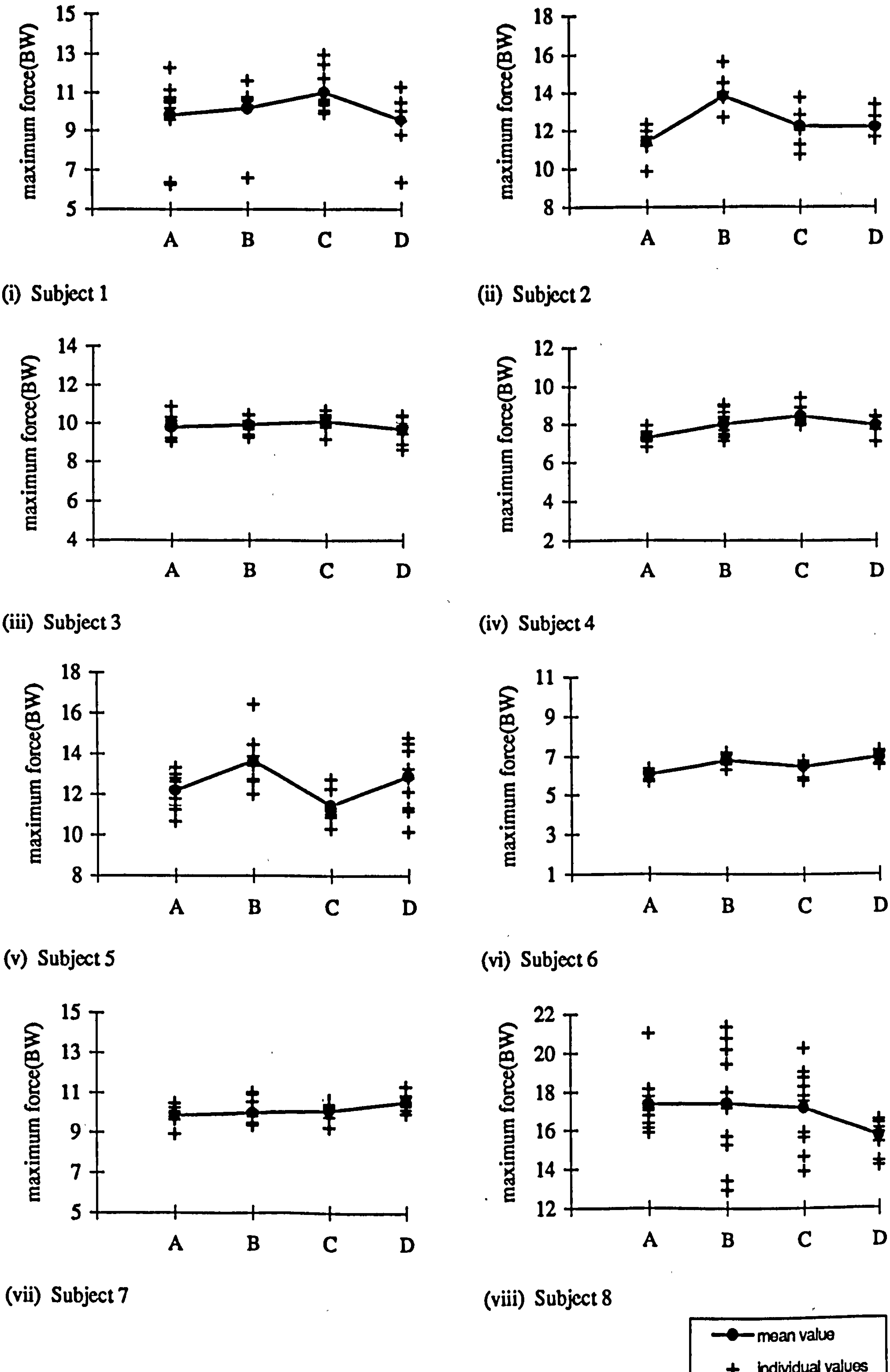


Figure 6.18 Maximum Achilles tendon force for each subject/condition

(viii) Achilles Tendon Loading Rate

At initial ground contact the Achilles tendon is generally not loaded, as illustrated by the theoretical negative force values at the start of ground contact for most subjects. Within the first 0.02 s of ground contact the Achilles tendon force has generally become positive. Examination of the time history of this variable during ground contact for all subjects and conditions revealed that, following the initial negative force, there is a sharp increase in Achilles tendon force (Figure 6.17). Consideration of the instantaneous Achilles tendon loading rate highlighted very large fluctuations, particularly during initial ground contact (Table 6.13). The comparison of ankle moment values obtained during stance using quasi-static and inverse dynamics methods revealed that, even during the impact phase, similar profiles were obtained, indicating that the observed fluctuations were not caused by errors in the acceleration data.

The further investigation of variables influencing Achilles tendon loading rate values revealed that profiles of Achilles tendon moment arm and vertical GRF were smooth (Figure 6.19). However, the profile for the GRF moment arm was shown to clearly contain noise at both the beginning and end of the stance phase (Figure 6.20). This noise can be attributed to the relatively low level of accuracy in centre of pressure data when the GRF is below a defined threshold, as illustrated in the trace for the centre of pressure (Figure 6.20). It was found that confidence in centre of pressure data was reduced whilst the vertical GRF was below 50 N. This figure is the manufacturers' defined threshold of 1% of the maximum range in vertical GRF, which was set at 5000 N for force data collection in the present study. To reduce the influence of centre of pressure errors on estimation of maximum Achilles tendon loading rate, the previously defined ground contact criteria of 15 N or above was replaced by 50 N, for calculation of this variable. This corresponds with 1% of the 5000 N maximum range in vertical GRF. Examination of selected data sets from each subject indicated that this generally involved omitting between the first four and the first six Achilles tendon force results at 1000 Hz. Thus, to ensure 50 N had been surpassed before Achilles tendon loading rate was calculated, estimations of this variable were started at a time 0.006 s beyond the zero force data reading. Sample data are provided in Table 6.13 to illustrate this procedure.

Despite the omission of early Achilles tendon force values from loading rate calculations, it was found that large fluctuations remained in the instantaneous Achilles tendon force loading rate data. For example, it can be seen in Table 6.13 that a change in instantaneous Achilles tendon loading rate from -201.8 $BW.s^{-1}$ to 320.46 $BW.s^{-1}$ occurred over a 0.001 s period. These loading rate changes were the result of changes in Achilles tendon force from 0.996 BW to 0.795 BW and from 0.795 BW to 1.115 BW, over time periods of 0.001 s. These changes are beyond the range of confidence in estimated Achilles tendon force values obtained when considering precision attained in the present study in Achilles tendon force measurement (0.4 BW). Observation of changes in instantaneous Achilles tendon loading rate later in the stance phase also revealed large fluctuations. Again these loading rate changes were found to be the result of changes in Achilles tendon force which were beyond the range of confidence in estimated Achilles tendon force values.

Values for maximum smoothed Achilles tendon loading rate for each subject/condition combination are provided in Table 6.14, and time of occurrence in Table 6.15. With the exception of Subject 3, there was an increase in Achilles tendon loading rate with the introduction of rearfoot and forefoot lifts compared with the barefoot condition, with this difference being significant for two subjects. For the different heel lift conditions, varying responses were found across subjects. This varied response, together with the lack of detection of significant differences, indicated that there was no clear trend in Achilles tendon loading rate response to heel lift intervention. However, for each subject, the observed changes in maximum smoothed loading rate of Achilles tendon force were generally in the same direction as the changes found in maximum Achilles tendon force.

Average Achilles tendon loading rates, obtained by division of the maximum Achilles tendon force values by the time of occurrence of this force in seconds, are presented in Table 6.16, in $BW.s^{-1}$. When heel lifts were attached to the rearfoot and the forefoot, there was an increase in average Achilles tendon loading rate for all subjects. The use of an increased heel lift had a varied influence across subjects. However, at least one of the heel lift conditions resulted in a decrease in average loading rate of Achilles tendon force, for all subjects. For all but Subject 7, the direction of change of the average Achilles tendon loading rate was in the same direction as the observed change in maximum Achilles tendon force.

Table 6.13 Calculation of Achilles tendon loading rate: sample data

Row	time (s)	ATF (BW)	Rate ($BW.s^{-1}$)	F_z (N)
12	0.000	0.0002	-0.13	17
13	0.001	-0.478	-477.80	29
14	0.002	-0.487	-9.60	43
15	0.003	-0.212	275.42	63
16	0.004	0.806	1018.20	83
17	0.005	1.077	270.67	102
18	0.006	0.996	-80.67	119
19	0.007	0.795	-201.80	136
20	0.008	1.115	320.46	158

Table 6.14 Mean maximum Achilles tendon loading rates for each subject and condition in $BW.s^{-1}$ (SD)
(A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

Condition	A	B	C	D
Subject 1	245 (62)	258 (39)	265 (31)	241 (40)
Subject 2	246 (17)	280 (21)	242 (30)	254 (34)
Subject 3	225 (33)	202 (22)	216 (20)	216 (23)
Subject 4*	134 (10)	151 (9)	161 (17)	158 (15)
Subject 5*	242 (32)	291 (29)	226 (28)	286 (49)
Subject 6*	119 (11)	130 (8)	120 (10)	141 (13)
Subject 7	198 (12)	208 (14)	215 (7)	221 (13)
Subject 8	416 (47)	491 (138)	397 (29)	413 (86)

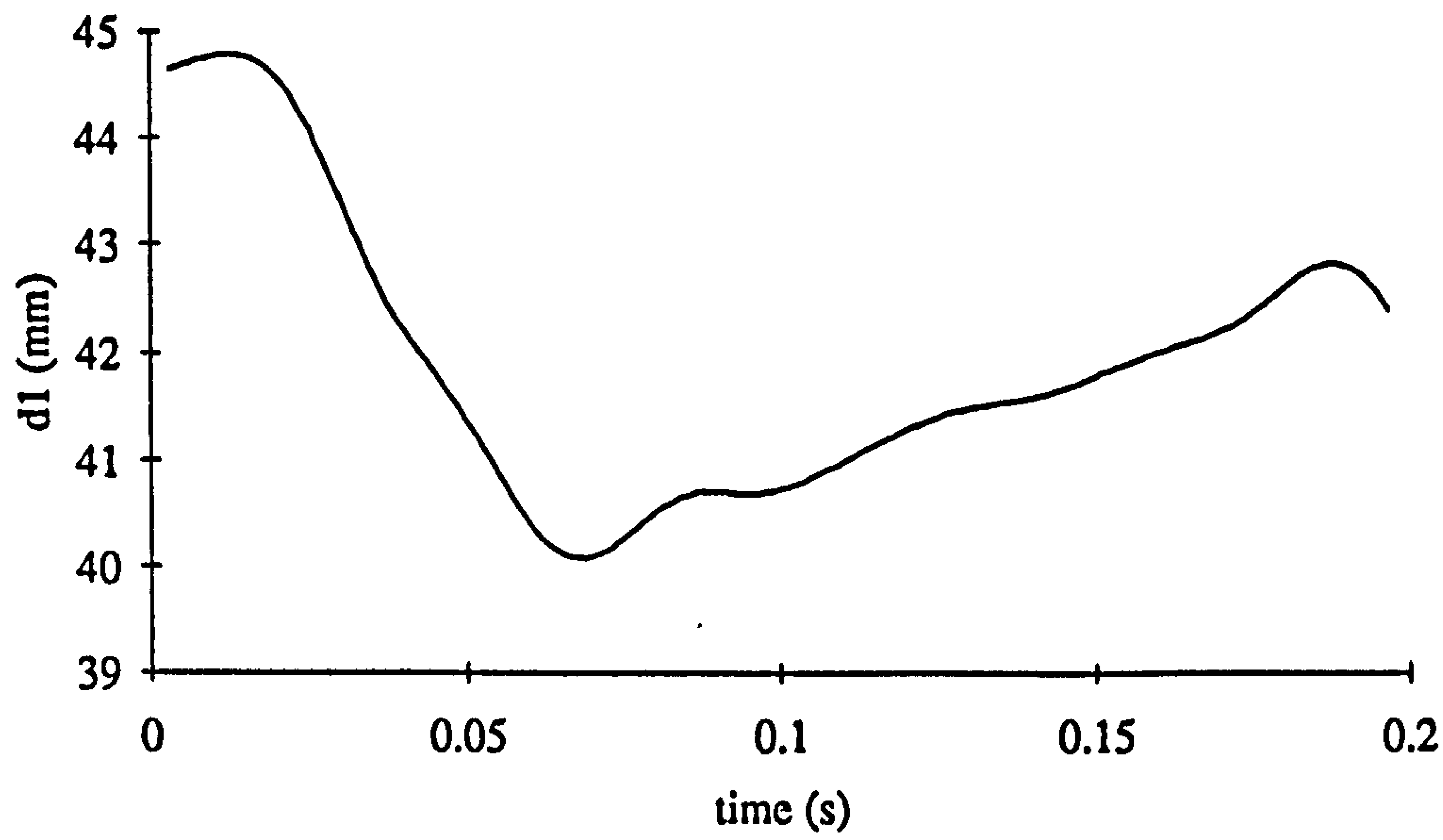
*$p<0.05$ Subject 1: none
Subject 2: none
Subject 3: none
Subject 4: A versus B
Subject 5: A versus B; B versus C; C versus D
Subject 6: C versus D
Subject 7: none
Subject 8: none

Table 6.15 Mean time of occurrence of maximum Achilles tendon loading rate for each subject and condition in ms (SD)
(A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

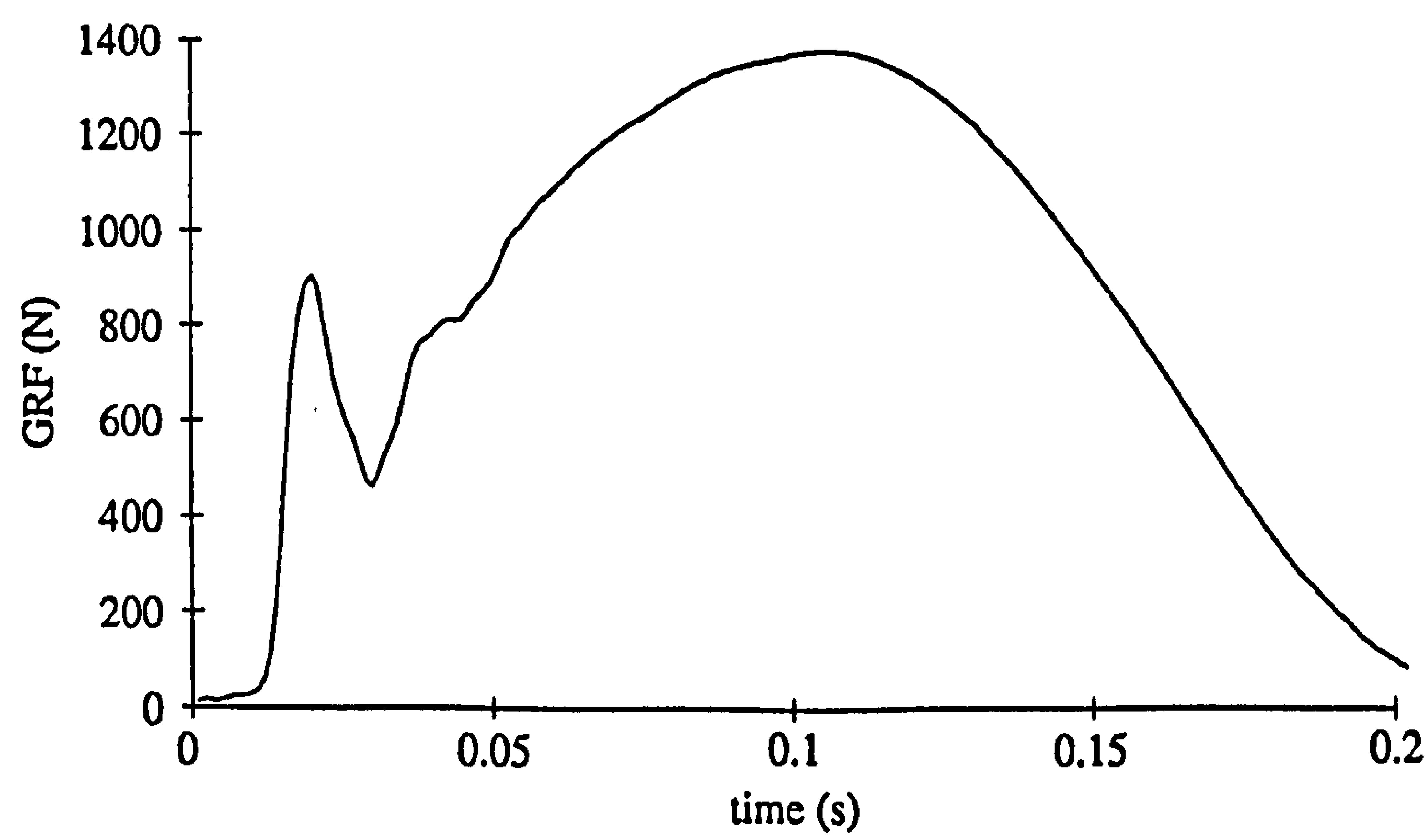
Condition	A	B	C	D
Subject 1	34 (16)	47 (6)	46 (9)	45 (15)
Subject 2	49 (13)	46 (5)	49 (8)	48 (6)
Subject 3	20 (5)	25 (9)	32 (9)	23 (9)
Subject 4	47 (3)	30 (7)	29 (7)	36 (8)
Subject 5	47 (16)	42 (19)	64 (10)	60 (12)
Subject 6	29 (17)	43 (7)	40 (14)	38 (12)
Subject 7	50 (8)	49 (7)	50 (8)	60 (5)
Subject 8	32 (15)	37 (9)	22 (11)	18 (4)

Table 6.16 Average Achilles tendon loading rate for each subject and condition ($BW.s^{-1}$)
(A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

Condition	A	B	C	D
Subject 1	88	94	105	86
Subject 2	91	124	106	106
Subject 3	100	105	111	104
Subject 4	65	78	79	71
Subject 5	116	119	102	106
Subject 6	59	68	63	70
Subject 7	93	97	94	94
Subject 8	183	210	181	174



(a)



(b)

Figure 6.19 Example traces of Achilles tendon moment arm (a) and GRF (b)

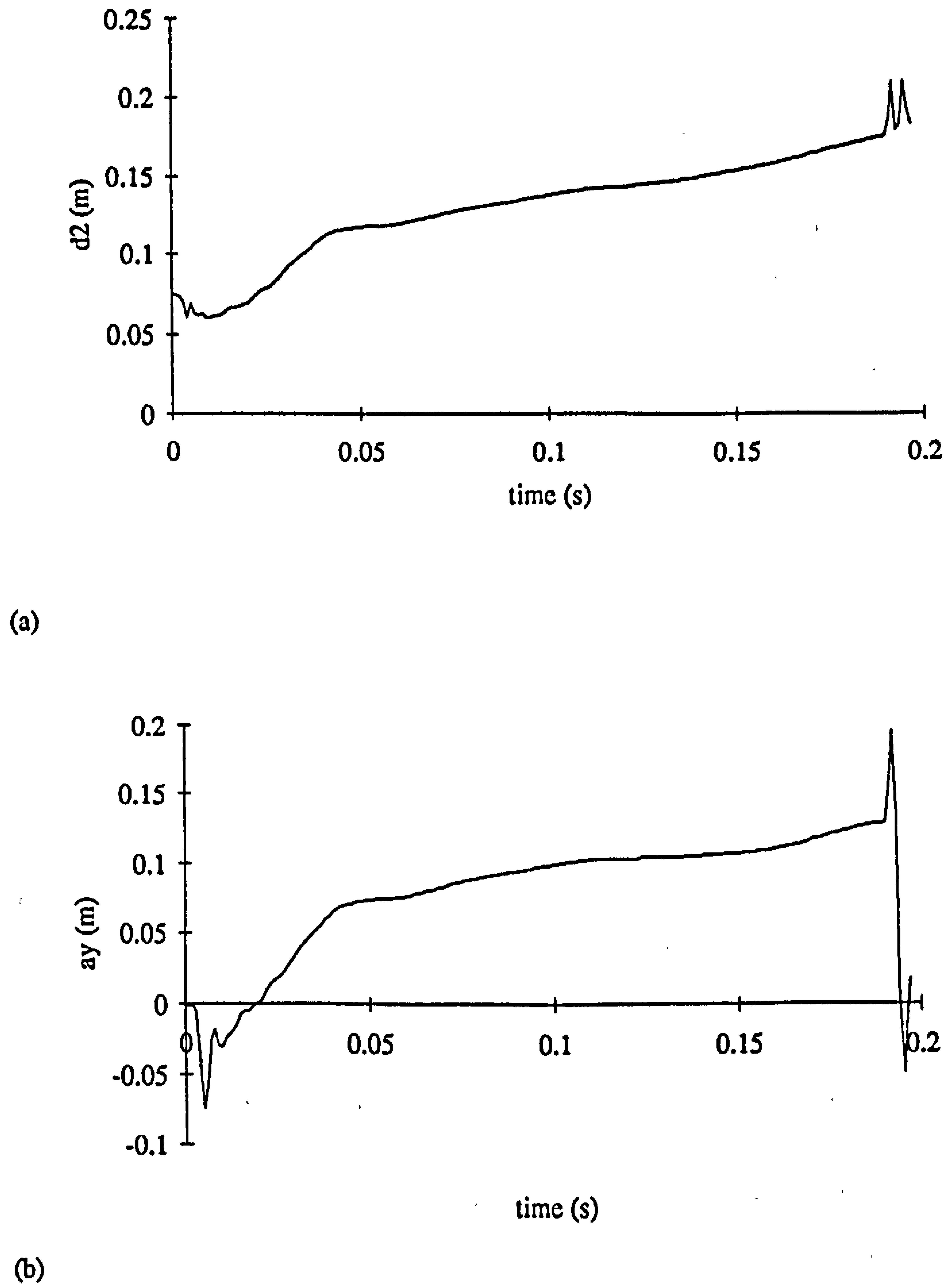


Figure 6.20 Sample traces of GRF moment arm (a) and centre of pressure (b)

6.4 Discussion

(i) Shock Absorption Conditions

The finding in the present chapter that the maximum ankle plantar-flexion moment and maximum Achilles tendon force have been increased by the introduction of increased shock absorption, is in contrast with the findings for this subject in Chapter 3. In Chapter 3, a shock absorbing heel lift was attached to the plantar surface of the foot providing both increased shock absorption and heel lift, whereas in the present chapter attempts have been made to control the condition such that only shock absorption has been varied. This has involved the attachment of Sorbothane to both the rearfoot and the forefoot, in addition to the attachment of a firm material to achieve the baseline required. The subject reported that the Sorbothane conditions used in the present chapter were particularly inhibiting due, it was felt, to the heaviness of the Sorbthane, especially that attached to the forefoot. For Condition E, under which attempts had been made to vary only the shock absorption, an increased mass at the rearfoot of 43 g and at the forefoot of 108 g had been applied. Although this was less than the mass of the conventional running shoe worn for training by this subject (260 g), the mass was distributed with a large proportion at the front of the foot.

For the 15 mm heel lift condition (Condition D) the total mass attached to the rear of the foot was 19.5 g, and to the forefoot was 30 g. No comments on the weight of these lifts were made by the subject. It was concluded from the findings for Subject 7 that it was not possible to successfully manipulate shock absorption in the same manner that heel lift was varied. Thus, for the remaining subjects, only heel lift manipulation was carried out.

(ii) GRF

The vertical GRF traces obtained for each subject/condition combination have characteristic impact and active peaks, as described in the literature (Cavanagh and Lafortune, 1980; Nigg, 1983). Two of the three basic types of vertical GRF trace during heel strike running as described by Nigg (1983) are represented. For most subjects, the GRF trace demonstrated an impact peak followed by an active peak. For a small number of subject/ condition combinations, a double impact peak was evident. Nigg (1983) suggested that this less common double impact peak may be due to a forefoot to rearfoot landing, or a landing from the lateral to the medial side of the heel. Magnitudes of impact peak and active peak obtained in the present study are comparable with those presented in the literature for barefoot running (Frederick, 1983).

The average GRF loading rates calculated in the present study are to the lower end of the range of values presented in the literature for barefoot running with a rearfoot strike (Bourassa and Therrien, 1981; Therrien et al., 1982). The instantaneous GRF loading rate was chosen for comparisons between conditions because this variable was available for all subjects. The shape of the GRF loading rate profile presented in the present study is comparable with that provided by Lees and Haynes (1995), with a peak value occurring within the first 20 ms of stance. The peak instantaneous GRF loading rates obtained in the present study were in general greater than those presented in the literature (Nigg et al., 1987; Nigg and Bahlsen, 1988). However, the published values referenced were for subjects wearing running shoes. Only one subject in the present study demonstrated a significant reduction in the instantaneous GRF loading rate when the lift material was attached to the foot. Since GRF loading rate has generally been found to be decreased by the wearing of shoes with increased shock absorption properties (Therrien et al., 1982; Nigg et al., 1987), this result indicated that the lifts did not provide a noticeable amount of shock absorption. This supports the use of the high density EVA material used in this study for the controlled manipulation of foot and ankle geometry.

GRF data were used to monitor total stance time in the present study. Stance times obtained were comparable with those presented in the literature (Munro et al., 1987). The wearing of different footwear, or running in footwear compared to the running barefoot, have generally been found not to influence the total stance time (Nigg and Bahlsen, 1988). Although the conditions employed in the present study were different to those more normally utilised in running studies, the results were in agreement with literature findings, with few differences in total stance time being detected across conditions. Any observed changes in other temporal variables were therefore not the result of differences in total stance time.

(iii) Ankle Moments

The maximum ankle plantar-flexion moments recorded in the present study were comparable with those presented in the literature (Winter, 1980; Reinschmidt and Nigg, 1995). The inertia data of Zatsiorsky and Seluyanov (1985) were chosen for the inverse dynamics ankle moment calculations, since the regression equations provided by these authors were derived using measurements from students of physical education, indicating these subjects were young, active individuals. All but Subject 1 of the present study were young (28 or younger), and all were distance runners and thus could be categorised as active.

The finding in the present study that the ankle moments obtained using quasi-static calculations and inverse dynamics were identical to within 0.1 N.m suggests that the use of the inertia data provided by other authors will not have influenced results. This suggestion was supported when different inertia data sets were used in moment calculations, and no differences in results were observed. The small mass and moment of inertia of the foot segment, and the small accelerations of this segment during ground contact will have contributed to the negligible influence of inertia data.

The results of the present study demonstrate that increases in heel lift can have an influence on the magnitude of the maximum ankle plantar-flexion moment, indicating an effect on loading of the structures associated with the ankle joint. However, no clear trend has been demonstrated in this variable with increased heel lift. The absence of a trend in maximum ankle plantar-flexion moment with increased heel lift across subjects is in agreement with the findings of Reinschmidt and Nigg (1995). These authors highlighted the differing responses across subjects, as has been demonstrated in the present study. Reinschmidt and Nigg (1995) used five different amounts of heel lift ranging from 13 mm to

25 mm lift relative to the forefoot, compared with two heel lifts of 7.5 mm and 15 mm in the present study. No baseline condition was employed by these authors. Observation of their data over the five heel lifts, and the five subjects, revealed that fluctuations in maximum ankle plantar-flexion moment across heel lift conditions existed for all subjects. Despite the smaller range of heel lifts used in the present study, the general findings of varied response to heel lift across subjects, and the possibility of maximum ankle plantar-flexion moment being increased, are in agreement with those of Reinschmidt and Nigg (1995). Predicting the influence of a heel lift of a defined height on ankle plantar-flexion moment for the running population of rearfoot strikers is clearly not possible using the results of the present study, although it may be possible to identify an optimum heel lift for reduction of maximum ankle plantar-flexion moment for an individual subject.

The observation in the present study that maximum ankle plantar-flexion moment occurred later in stance when heel lift was introduced is difficult to interpret, since varied changes in magnitude of maximum ankle plantar-flexion moment with heel lift occurred across subjects. It is possible that the later occurrence time contributes to a decreased average Achilles tendon loading rate when a heel lift is used. Changes in this variable across conditions are discussed in Part (v) of the present section.

The collection of 120 Hz kinematic data in the present study, which was smoothed and interpolated to 1000 Hz, has provided a clear illustration of the dorsi-flexion moment at the beginning of the stance phase. This is in contrast with the findings in previous chapters in which kinematic data were collected at 50 Hz. The general occurrence of a peak dorsi-flexion ankle moment within the first 20% of the stance phase is consistent with the findings presented in the literature (Winter, 1984; Reinschmidt and Nigg, 1995). In contrast with the findings of Reinschmidt and Nigg (1995), no trends in dorsi-flexion maximum moment or time of occurrence were observed in the present study. In fact, only one of the eight subjects of the present study showed a significant decrease for heel lift conditions compared with zero heel lift, in comparison with the significant decrease observed across subjects by Reinschmidt and Nigg (1995). This contrast between studies may be due to the different conditions. The attachment of heel lifts to the rear of the foot may have a vastly different influence on mechanics around the impact phase than the rear part of a running shoe. During the midstance phase, the foot is firmly on the ground and direct comparisons between studies may be more appropriate for this stage of stance.

It has been observed that Subject 1 demonstrated clearly different behaviour than the remaining subjects. In particular, smaller values of maximum ankle plantar-flexion moment were demonstrated, and these occurred at a later time in the stance phase. Additionally, this subject showed a decrease in the time of occurrence of maximum ankle plantar-flexion moment with increased heel lift. It has been noted that this subject is older than the other subjects (46 compared with a mean age of 23 for the remaining subjects), and also is not as well trained as the remaining subjects. Differences in mechanical properties of muscle and tendon due to age may have caused the different behaviour observed for this subject (Yamada, 1970).

(iv) Achilles Tendon Force

The time histories of Achilles tendon force were similar to those provided in the literature (Burdett, 1982; Komi, 1990), and those presented in previous chapters. For all but Subject 8, the magnitudes of maximum Achilles tendon force were comparable to those presented in the literature. Subject 8 demonstrated relatively large maximum Achilles tendon force values, which were comparable with the values obtained or the forefoot striker in Chapter 4, despite the GRF and kinematic evidence supporting the classification of this subject as a rearfoot striker. Maximum Achilles tendon stress values obtained in the present study, ranging from 49×10^6 N.m^{-2} to 91×10^6 N.m^{-2}, are comparable with those presented in the literature. Ker (1981) provided a typical Achilles tendon stress during running of 53×10^6 N.m^{-2}, whilst Komi (1990) calculated a value of 111×10^6 N.m^{-2}. It was observed in the present study that those subjects demonstrating the largest values of maximum Achilles tendon force, namely Subject 2, Subject 5 and Subject 8 also were found to have the largest values for cross-sectional area of the Achilles tendon. The influence of high maximum force values on maximum stresses for these subjects was therefore reduced by the larger cross-sectional areas. The correlation coefficient of 0.78 demonstrated that in general a higher maximum Achilles tendon force indicated a higher maximum Achilles tendon stress.

The maximum Achilles tendon stresses obtained in the present study were within the range of ultimate stress values for tendon (Abrahams, 1967; Bennett et al., 1986), indicating that the stress experienced by the Achilles tendon in running is close to the maximum stress that this structure can sustain. This suggestion is consistent with findings in the literature (Scott and Winter, 1990; Komi, 1990). The Achilles tendon cross-sectional area was approximated in the present study by a semi-circle using MRI data for Subject 7. Observation of the MRI image (Figure 5.7) demonstrated that this simplification was likely to result in a small under-estimation of the area, causing a greater tendon stress than would have actually been experienced in this area of the tendon. In the present study, it has been assumed that the triceps surae muscle group is the sole contributor to ankle plantar-flexion. This was justified since the aim of the study was primarily to compare Achilles tendon values across conditions, rather than to obtain absolute magnitudes. For speculation on the relative stress magnitudes obtained in the present study with ultimate stress values, consideration was made of the influence of assumptions. Scott and Winter (1990) described how the triceps surae muscle group contributes at least 85% of the total plantar-flexion moment. This value was obtained using relative physiological cross-sectional areas, with additional reference to the relative moment arm lengths. The maximum Achilles tendon stress for Subject 7 would have been reduced from 68×10^6 N.m^{-2} to 58×10^6 N.m^{-2} by the use of 85% contribution to total moment, a value that remains within the range of ultimate stress of tendon. These findings support suggestions in the literature that the tendon specimens used to obtain ultimate stress values do not represent the dynamic loading capacity of in-vivo tendons (Scott and Winter, 1990).

Reinschmidt and Nigg (1995) suggested that changes in maximum ankle plantar-flexion moment indicate changes in maximum Achilles tendon force. In the present study, it has been shown that decreases in Achilles tendon force are not always accompanied by decreases in ankle moment, demonstrating that it may not be appropriate to use ankle moment variations to indicate a change in Achilles tendon force. This is consistent with the findings of Chapter 4.

Despite the common running style adopted by each of the subjects in the present study, differing maximum Achilles tendon force responses to heel lift intervention were observed across subjects. This finding highlights the difficulty in predicting the response of an individual to heel lift variation. Limited support is provided for the suggestion that the introduction of a heel lift reduces the magnitude of maximum Achilles tendon force.

Despite the different responses to heel height variation across subjects, four of the subjects showed a significant decrease in maximum Achilles tendon force for one or both of the heel lift conditions, compared with zero heel lift, whereas only one subject demonstrated a significant increase in maximum Achilles tendon for the heel lift conditions. This suggests that, for selected subjects, a heel lift can be employed to reduce the maximum force that the Achilles tendon is subjected to. The significant increase for one subject demonstrates the difficulty in making generalisations with regard to heel lift prescription. If the occurrence of Achilles tendon injury is associated with the maximum force, as has been suggested in the literature, then for some individuals the use of heel lift may have an adverse influence on this injury. The finding that the degree of heel lift required for a significant reduction of Achilles tendon force differs between subjects, further highlights the importance of individual assessment.

Despite the clearly lower magnitudes of maximum ankle plantar-flexion moment observed for Subject 1, compared with the remaining seven subjects, no obvious differences were found in maximum Achilles tendon force values. However, the finding that the maximum Achilles tendon force occurred later in the stance phase for this subject, is in agreement with the indications of the ankle plantar-flexion moment results. The inconsistencies in magnitude of force results for this subject demonstrate, as previously discussed, the unsuitability of using ankle plantar-flexion moment to indicate Achilles tendon force differences. However, from the results for all eight subjects studied, it seems that indications of temporal differences in Achilles tendon force between conditions may be made using ankle moment values.

Clear differences were found in maximum Achilles tendon force magnitudes between subjects, with maximum forces ranging from 6.1 BW (Subject 6) to 17.4 BW (Subject 8). Despite the inherent systematic error in estimation of Achilles tendon forces, it was possible to make limited comparisons between subjects by quantifying the maximum systematic error and the random variation, to obtain confidence limits in force values. A source of systematic error in Achilles tendon force estimations using the methods of the present study will be differing placement of ankle and Achilles tendon markers between subjects in relation to the true anatomical locations. Systematic errors may also occur as a result of the assumptions made regarding the relative contribution of the triceps surae muscle group to ankle plantar-flexion, and the two-dimensional action of this muscle group. However, it was assumed that the error resulting from these assumptions was similar across subjects. Measurement errors were assumed to be random, and have been quantified earlier. The influence of these errors on comparisons was reduced by the use of mean data over 10 trials, decreasing the magnitude of the error and causing similar magnitudes across conditions. Thus, the main source of error limiting the ability to make comparisons of Achilles tendon forces between subjects was the marker placement. By quantifying the potential error resulting from marker placement variation, confidence limits for making comparisons between subjects were obtained for the present study.

In Chapter 4, marker placement was found to have a maximum influence of 1.5 BW on maximum Achilles tendon force for barefoot running, for a subject demonstrating maximum force values ranging from 9.3 BW to 10.8 BW. In the present study, the ankle marker was placed using the same criteria as in Chapter 4. Achilles tendon markers, however, were placed on the rear of the leg, rather than on the lateral side as described in Chapter 4. Since with both of these methods the aim was to place the markers to best approximate the line of action of the tendon, it is expected that the resulting estimated Achilles tendon forces will be similar for both marker sets, and thus the magnitude of systematic error resulting from marker placement will be similar. This is supported by the finding that the estimated maximum Achilles tendon force obtained for the same subject when employed in the present study was 9.9 BW, a value within 1.5 BW of those obtained using the previous methods. Additionally, for two subjects in the present study new sets of markers were placed during separate testing sessions to investigate the influence of marker placement on estimated Achilles tendon force values. For both subjects, differences between testing sessions were less than 1.5 BW.

It has therefore been demonstrated that the maximum influence of marker placement variation on estimated maximum Achilles tendon force was ±1.5 BW. Thus, differences between subjects of more than 3 BW were likely to be true differences in magnitudes of maximum Achilles tendon force. Examination of results for the barefoot condition revealed that the smaller maximum Achilles tendon force values observed for Subject 6 were likely to indicate that the absolute values for this subject were less than those obtained for all but Subject 4. Similarly, the relatively large values estimated for Subject 8 were taken to indicate large absolute maximum Achilles tendon forces for this subject compared with the remaining subjects.

It is of interest that Subject 8 has experienced a recurring Achilles tendon injury over the past three years, since commencing a new training program with increased weekly mileage.

(v) Achilles Tendon Loading Rate

As previously discussed in Chapter 4, the rate of loading of the Achilles tendon may be of significance when considering the etiology of Achilles tendon injury (Komi, 1990). It has been demonstrated that the Achilles tendon forces estimated in the present study were not sufficiently reliable for calculation of instantaneous Achilles tendon loading rates using time periods of 0.001 s. Gallagher (1994) estimated instantaneous Achilles tendon force loading rates using raw cine-film kinematic data at 250 Hz, and found that acceptable reliability was attained in Achilles tendon force values to allow the calculation of Achilles tendon loading rates. The use of 1000 Hz force data together with smoothed and interpolated 120 Hz kinematic data has highlighted large variations in Achilles tendon force values immediately following impact, which were not detected at 250 Hz.

To obtain reliable Achilles tendon loading rate values over stance, a method of smoothing the Achilles tendon loading data was developed using average values over 0.01 s intervals. Magnitudes of maximum Achilles tendon loading rate calculated using these methods ranged from 119 $BW.s^{-1}$ to 491 $BW.s^{-1}$, comparing well with ranges provided by Gallagher (1994) of 215 $BW.s^{-1}$ to 366 $BW.s^{-1}$, and Komi (1990) of 67 $BW.s^{-1}$ to 267 $BW.s^{-1}$. The maximum loading rate was found to occur within the first 18 ms to 64 ms of ground contact for each subject, corresponding to within approximately 10% to 30% of total stance time. These results are comparable with those of Gallagher (1994) who found that maximum instantaneous Achilles tendon force loading rates occurred within 10% to 30% of total stance time. The introduction of heel lifts was found to have a variable influence on the maximum smoothed Achilles tendon loading rate across subjects, again demonstrating the importance of assessing each subject individually.

In the study carried out by Gallagher (1994) a single subject was employed. This same subject was studied in the present investigation (Subject 7), allowing direct comparison of Achilles tendon loading results between studies for this subject for the barefoot condition. Gallagher (1994) obtained a maximum Achilles tendon loading rate of 215 $BW.s^{-1}$, and a time of occurrence of this variable of 13% total stance time, for barefoot running for a single running trial at 3.9 $m.s^{-1}$. In the present study, these figures for Subject 7 were 198 $BW.s^{-1}$ and 25%, respectively. The magnitude of Achilles tendon loading rate magnitude is comparable across the two studies, with the small difference being accounted for by factors such as natural variation between trials. The later occurrence of this variable may be attributed, at least in part, to the different ground contact criteria used in the two studies, with 50 N being used by Gallagher (1994), compared with 15 N in the present study. Observation of GRF 1000 Hz data revealed that this may account for up to 5 ms difference, equating to approximately 2.5% of stance time. The use of different time intervals over which the maximum loading rate was calculated may also have contributed to the different values obtained in the two studies. A final consideration is the use of only one trial for the results presented by Gallagher (1994), since natural variation will exist between trials. This is illustrated by the result that, for a 95% confidence interval, the mean time of occurrence calculated in the present study over 10 trials may have been in error by up to 6 ms, or 3%.

The average Achilles tendon loading rates obtained in the present study ranged from 59 $BW.s^{-1}$ to 210 $BW.s^{-1}$ compared with 87 $BW.s^{-1}$ to 107 $BW.s^{-1}$ obtained by Gallagher (1994). This variable was found to be influenced more consistently across subjects than other Achilles tendon loading variables, with a decrease generally being demonstrated with increased heel lift. The consistently later occurrence of maximum Achilles tendon force is the common factor across subjects contributing to this reduction in the average Achilles tendon loading rate.

In general, it has been demonstrated in the present chapter that the observed changes in maximum and average Achilles tendon loading rate with heel lift manipulation are in the same direction as the changes in maximum Achilles tendon force and stress. This supports the suggestion of Abrahams (1967) that an increased rate of tendon loading increases the stress experienced by the tendon. This finding is also consistent with the behaviour of a viscoelastic material (Hawkins, 1993).

(vi) General Findings

The choice of smoothing and differentiation procedures for use in the present study was made by careful consideration of the study requirements, as recommended by Wold (1974). The main requirement was to obtain interpolated 1000 Hz displacement data from 120 Hz MacReflex data. The cubic spline provided by Reinsch (1967) was chosen following recommendations in the literature (Zernicke et al., 1976; McLaughlin et al., 1977). A quintic spline was not employed due to the possibility of erratic behaviour of interpolated data, as highlighted in the literature (Yeadon, 1984; van den Bogert and Glossop, 1996).

Although, the displacement data were subsequently differentiated to obtain accelerations for use in inverse dynamics calculations, the smoothing procedure was not chosen according to that most appropriate for obtaining acceleration data since the calculated acceleration values were unlikely to have a marked influence on resultant ankle joint moments. This was due to the small magnitude of acceleration of the foot segment during ground contact, and the small mass and moment of inertia of this segment (Morlock and Nigg, 1988). The suggestion that foot acceleration values have a negligible influence on resultant ankle moments has been supported in the present study. Limitations of using cubic spline techniques have been highlighted when accelerations are required, due to the endpoint restrictions placed on the second derivative (Wood and Jennings, 1979). In the present chapter, finite difference techniques were chosen for calculation of velocity and acceleration values, avoiding the possibility of endpoint restrictions influencing results. These techniques were easily incorporated in the spreadsheet design employed for calculations in the present chapter. The availability of additional data points either side of ground contact facilitated the use of nine point procedures without the loss of data at each end of the period of interest. The use of the two point and nine point methods provided acceptable velocity and acceleration data, respectively, as has been demonstrated by Lees (1980).

In general, it was found that for all variables the attachment of lifts to the rearfoot and the forefoot to raise the height of the foot without introducing a heel lift, had a common influence across subjects. The use of this baseline condition, rather than making heel lift comparisons with the barefoot conditions, is clearly necessary. By using this condition, the possibility of methods of attachment of lifts having an influence on results has been limited to a systematic effect. The attachment of material to the rearfoot may also have acted to increase the lever arm about the subtalar joint in the frontal plane, as has been demonstrated for increased heel flare of running shoes (Nigg, 1986). The use of the described baseline condition has maintained a constant heel width across heel lift conditions, and thus has limited changes in the described lever arm to being due to heel height variation, rather than heel width.

The Achilles tendon loading results obtained in the present study have demonstrated that generally there is a varied response across subjects to changes in heel lift. The only response that can be predicted with reasonable confidence is the timing of maximum Achilles tendon force occurrence, with a later occurrence being found with increased heel lift, resulting in a smaller magnitude of average Achilles tendon loading rate. It is suggested that the mechanism by which an increased heel lift acts to reduce the incidence of Achilles tendon injury is by reducing the average loading rate of the tendon, by causing the tendon to be subjected to maximum force later in the stance phase. In order to investigate this suggestion, clinical evidence on the occurrence of Achilles tendon injury must be combined with estimations of Achilles tendon loading, using methods such as those developed in the present study.

CHAPTER 7 THE INFLUENCE OF HEEL LIFT ON LOWER EXTREMITY KINEMATICS IN REARFOOT STRIKERS

7.1 Introduction

It has been demonstrated that the forces experienced by the Achilles tendon during running are in the linear region of the stress-strain curve (Ker et al., 1988). This indicates that there is a linear relationship between Achilles tendon stress and Achilles tendon strain in running. It is therefore expected that maximum stress and maximum strain of the Achilles tendon in running occur simultaneously. The amount of stress experienced by the Achilles tendon has been represented in the previous chapter by estimation of Achilles tendon force and Achilles tendon cross-sectional area. The amount of strain experienced by the Achilles tendon will be dependent on the relative stiffness of the tendon and muscle, and the applied force (Alexander and Bennet-Clark, 1977). Changes in strain can be represented using ankle and knee angles (Grieve et al., 1978; Bobbert et al., 1986). The relative contribution of the Achilles tendon to length changes of the muscle-tendon complex will vary depending on the type of muscle contraction and the activity being performed (Bobbert et al., 1986). However, the long and compliant properties of the Achilles tendon indicate the potential of this tendon to contribute predominantly to the overall stretch of the muscle-tendon complex (Caldwell, 1995). An observed increase in ankle dorsi-flexion or knee extension therefore appears to indicate an increase in Achilles tendon strain. In the present chapter, ankle and knee angles are used to indicate changes in Achilles tendon strain across conditions.

In the Chapter 4 investigation, ankle angle was defined as the angle between the Achilles tendon line of action and a straight line joining the ankle joint centre to the MTP joint centre, using the assumption that the Achilles tendon line of action is parallel to the lower leg throughout the stance phase of running. Limitations in using this assumption were highlighted. In the present chapter, lower extremity joint angles are represented using angle definitions from the literature (Milliron and Cavanagh, 1990).

Sagittal plane ankle and knee angles have been found to be influenced by changes in the design characteristics of running shoes, with the influence of varied shock absorption being studied in particular (Frederick et al., 1983; Clarke et al., 1983a). It has been speculated that increased heel lift results in a reduction in the amount of maximum ankle dorsi-flexion, contributing to a decrease in Achilles tendon strain (Clement et al., 1984). There does not appear to be any literature evidence to support this suggestion. In the present study, ankle and knee angles are presented for the different heel lift conditions described in the previous chapter. These data were collected simultaneously to the collection of Achilles tendon force data in the previous study, allowing comparison of Achilles tendon force variation and joint angle variation with increased heel lift.

As outlined in Chapter 4, movement of the calcaneus relative to the lower leg in the frontal plane, commonly referred to as rearfoot movement, has been found to be influenced by footwear interventions (Nigg, 1986). Maximum eversion of the rearfoot has been associated with lower extremity injuries, including injury of the Achilles tendon (Clement et al., 1984; Winter and Bishop, 1992; Pagliano, 1987a). It has been demonstrated in selected studies that increased heel lift in running shoes can reduce the amount of rearfoot eversion (Stacoff and Kaelin, 1983). However, Clarke et al. (1983b) found no change in maximum eversion with increased heel lift. In the present chapter, the results of an investigation into the influence of heel height variation on rearfoot movement are presented.

The aim of the present study is to address the questions:

- Are sagittal plane joint angles influenced by heel lift variations ?
- Can joint angles be employed to adequately represent Achilles tendon strain in running ?
- Is maximum rearfoot eversion reduced by the use of heel lifts ?

7.2 Methods

Smoothed marker displacement data obtained in Chapter 6 were used to determine lower extremity kinematics for the eight subjects studied.

(i) Sagittal Plane Calculations

The smoothed marker displacement data were used to monitor sagittal plane segment and joint angular displacement and velocity values. The markers at the MTP, ankle, knee and hip were used to define segment end points. Angle measurement conventions in the sagittal plane were as described by Milliron and Cavanagh (1990), as has been illustrated in Figure 2.11. The thigh segment was represented by a straight line joining the hip and knee markers, and the lower leg by a straight line joining the knee and ankle markers. The foot segment was represented by a straight line joining the ankle and MTP markers. These lower extremity segments were assumed to constitute a rigid link system with segments joined by hinge joints.

Foot angle was defined as the angle between the foot segment and the horizontal. Ankle angle was defined as the angle between the foot segment and the lower leg segment. Knee angle was defined as the angle between the thigh segment and the lower leg. The joint angles of the ankle and knee were defined such that an increase in angle indicated a flexion of the joint. A thigh angle was defined as the angle between the thigh segment and the vertical. Angle time histories were obtained during ground contact for each of the defined angles. Following examination of typical time histories for each of the angles, selected parameters were chosen to characterise the patterns of angle variation during stance. Thigh, knee, ankle and foot angles at the time of initial ground contact, and maximum knee flexion and ankle dorsi-flexion angles were determined. Times of occurrence of each of these variables were obtained in milliseconds.

For confident comparison of sagittal plane angles across conditions, it was necessary to investigate the reliability of angle data. Of the measured angles, the foot angle was dependent on markers in closest proximity to one another. Thus, the level of precision attained in foot angle measurements was assessed, and it was assumed that the precision in measurement of the remaining angles was at least the level attained for this segment. It was demonstrated in Chapter 6 that the marker coordinates obtained in the present study were reliable to within 0.4 mm. The typical dimensions illustrated in Figure 7.3 were utilised to investigate the influence of ±0.4 mm error in marker locations. For comparisons between conditions for each subject, it was also necessary to consider the influence of movement out of the sagittal plane. The medial-lateral orientation of the foot may vary across conditions as a result of ab/adduction and in/eversion movements. The magnitude of the difference in medial-lateral foot orientation between conditions was assessed by monitoring the relative coordinates of the ankle and MTP markers in the xy plane, for selected trials for each subject. An angle y was defined as the angle between a straight line joining the ankle and MTP markers and a line through the ankle marker and parallel to the x-axis (Figure 7.1). The influence of a five degree abduction and a 10 degree eversion of the subtalar joint on comparisons made in the sagittal plane were investigated.

The standing calibration described in Chapter 6 was used for investigation of the influence of heel lifts on standing joint angles. These angles were subsequently compared with the joint angles at impact and during the middle of the stance phase when the foot was flat on the ground. The influence of possible subject movement during collection of standing data was investigated by comparison of variation in body marker locations with variation in the identified locations of stationary markers placed on the force plate. Reliability of angle data was investigated by use of a sensitivity analysis, using the same expected variations in marker locations as described in Chapter 6.

Joint and segment angular velocities were obtained using a two point finite difference algorithm, as previously described (Equation 6.1; Miller and Nelson, 1973). Force data were used to indicate the start of the ground contact phase. The possibility of errors in the synchronisation of force and kinematic data has been described in previous chapters. To ensure that impact kinematic variables were measured prior to actual ground contact, the five fields immediately preceding ground impact, as identified by the force plate data, were not considered in calculation of impact variables. Joint and segment angles at ground impact were calculated using displacement over the period 0.005 s immediately preceding the five discarded fields.

In Chapter 6, it was found that the observed changes in maximum Achilles tendon force were predominantly the result of variations in the moment arm of GRF and the moment arm of the Achilles tendon. The moment arm of the Achilles tendon is influenced by the ankle and the knee joint angles. The magnitudes of the ankle and knee joint angles were therefore obtained for the time of maximum Achilles tendon force, providing an indication of the relative contribution of joint angle differences to observed changes in Achilles tendon moment arm.

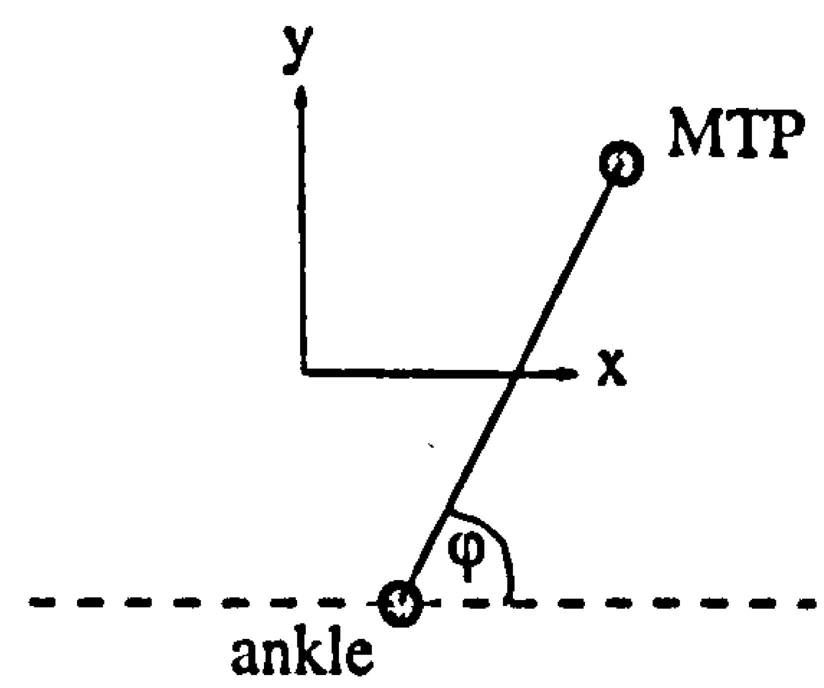


Figure 7.1 Identification of the medial-lateral foot orientation (view in transverse plane)

(ii) Frontal Plane Calculations

The conventions of Clarke et al. (1983b) were used to monitor movement of the calcaneus relative to the lower leg in the frontal plane (Figure 7.2), termed the rearfoot angle. Marker locations were described and illustrated in Chapter 5 (Figure 6.4). Frontal plane coordinates were obtained by extraction of the xz coordinates from three-dimensional marker coordinates. The Achilles 1 and Achilles 2 markers were joined by a straight line representing the Achilles tendon orientation in the frontal plane. The calcaneus 1 and calcaneus 2 markers were joined to represent the orientation of the calcaneus in the frontal plane. The angle between the lower leg and the vertical was termed ϕ, and the angle between the calcaneus and the vertical γ. The difference between these two angles was termed the rearfoot angle (η), as described by Clarke et al. (1983b). Also conforming to the conventions of Clarke et al. (1983b), angle definition was such that a positive rearfoot angle indicated supination, and a negative angle pronation. Following the recommendations of Hamill et al. (1994), the rearfoot variables presented were limited to those associated with maximum rearfoot angle. Maximum lower leg angle, calcaneal angle and rearfoot angle were recorded for each trial, together with the time of occurrence of maximum rearfoot angle in milliseconds. The reliability of rearfoot angle measurements was assessed by monitoring the rearfoot and contributing angles for standing calibration trials, as for the sagittal plane reliability investigation. To separate the variability due to the data collection system and the variability due to subject movement during calibration, the variability in stationary markers on the forceplate surface was quantified using RMS deviation over 10 consecutive fields.

For those subjects demonstrating a reduction in maximum rearfoot angle, the time of occurrence of this angle was compared with the timing of the maximum knee flexion angle. This allowed the investigation of the suggestion in the literature that a reduction in the maximum rearfoot angle reduces the discrepancy in timing of these two variables.

A sensitivity analysis was performed to investigate the influence of typical marker random error of 0.4 mm on calculated rearfoot angles. A maximum rearfoot angle of ten degrees, and a typical distance of 30 mm between the two calcaneus markers were assumed, to investigate the influence of measurement error on the calcaneal angle (β, Figure 7.3). For the lower leg angle (α, Figure 7.3) a maximum value of 3 degrees, and a typical distance between Achilles tendon markers of 60 mm, were used. The resultant influence on rearfoot angle measurements was calculated as the sum of the errors in calcaneal and lower leg angles.

The influence of rearfoot movement on sagittal plane angles was assessed by investigation of the influence of a ten degree variation in rearfoot angle on sagittal plane foot angle. The typical dimensions illustrated in Figure 7.3 were utilised.

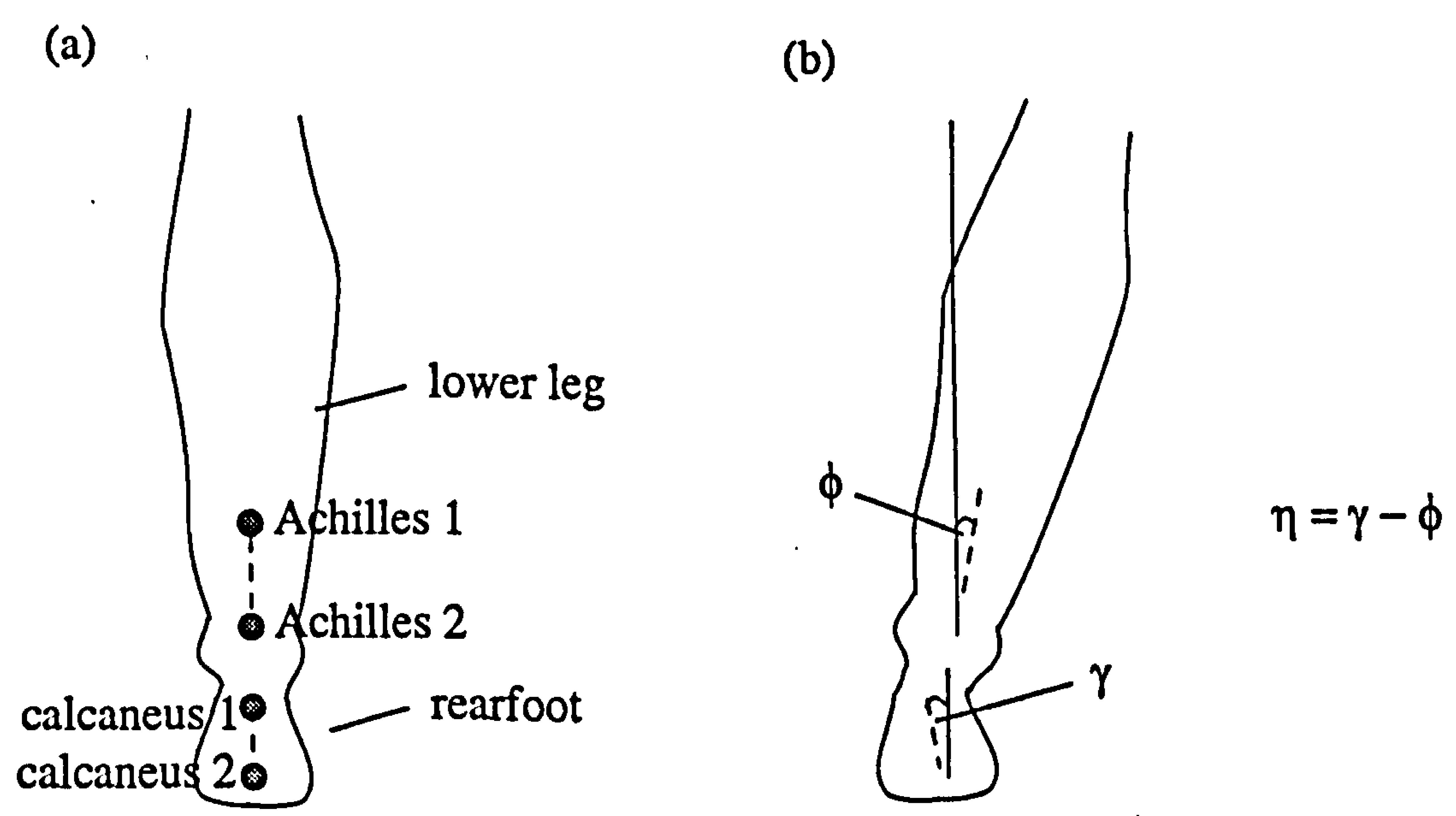


Figure 7.2 (a) Marker placement and definition of lower leg and calcaneus (right limb)
(b) Lower leg and calcaneal angles (ϕ, γ), contributing to rearfoot angle (η)

7.3 Results

(i) Sagittal Plane Joint Angles

Standing calibration

Standing angles for the foot segment, the thigh segment and the ankle and knee joints are provided in Table 7.1, for each subject/condition combination. Mean angle values obtained over 10 consecutive fields are provided in degrees, with standard deviations in parenthesis. Plots of foot angle against heel lift for each condition are provided in Figure 7.4, for each subject separately. For all subjects, the inclination of the foot segment to the horizontal was increased as heel lift was increased from zero heel lift to 7.5 mm heel lift, and from 7.5 mm heel lift to 15 mm heel lift. Negative foot angle values indicate that the line representing the foot is inclined with a negative slope. Thus increased inclination indicates an increased contribution to ankle plantar-flexion. Across subjects, differences in foot angle between zero heel lift and 7.5 mm heel lift were found to range from 0.96 degrees to 3.78 degrees. Differences between 7.5 mm heel lift and 15 mm heel lift ranged from 0.21 degrees to 3.66 degrees. The influence of a 5 degree medial rotation of the foot in the transverse plane on foot angles is illustrated in Figure 7.3 (a, b). For these typical dimensions, the foot angle increased from 33.7 degrees to 33.8 degrees.

The influence of heel height increase without an increase in heel lift can be seen by comparing zero heel lift (Condition B) with the barefoot condition (Condition A). Differing responses were found to occur across subjects, with some demonstrating similar foot angles for both conditions, some showing an increase in foot inclination with the horizontal and some showing a decrease. Compared with the changes in foot angle observed when heel lift was varied, these angle changes were small.

The standard deviation values presented in Table 7.1 indicate the amount of variation in measured marker location during collection of the standing calibration data. This variation will be the sum of variation in the measuring procedures and movement of the subject over the 10 fields (0.08 s) of data recording. To evaluate the contribution of these two variations, RMSD values for the markers on the force plate were calculated and compared with RMSD values for body markers when standing, for two randomly selected calibration trials involving two different subjects (Table 6.2). For the two trials analysed, the RMSD values ranged from 0.05 mm to 0.94 mm. Maximum RMSD values in the three orthogonal axes directions were 0.80 mm (X), 0.94 mm (Y), and 0.60 mm (Z). The force plate markers demonstrated similar RMSD values to the body markers, indicating that the variation in calculated marker locations during calibration was predominantly the result of variation in the measuring system, as opposed to movement of the subject during data collection, for those trials analysed. This result was assumed to be true for the remaining subjects.

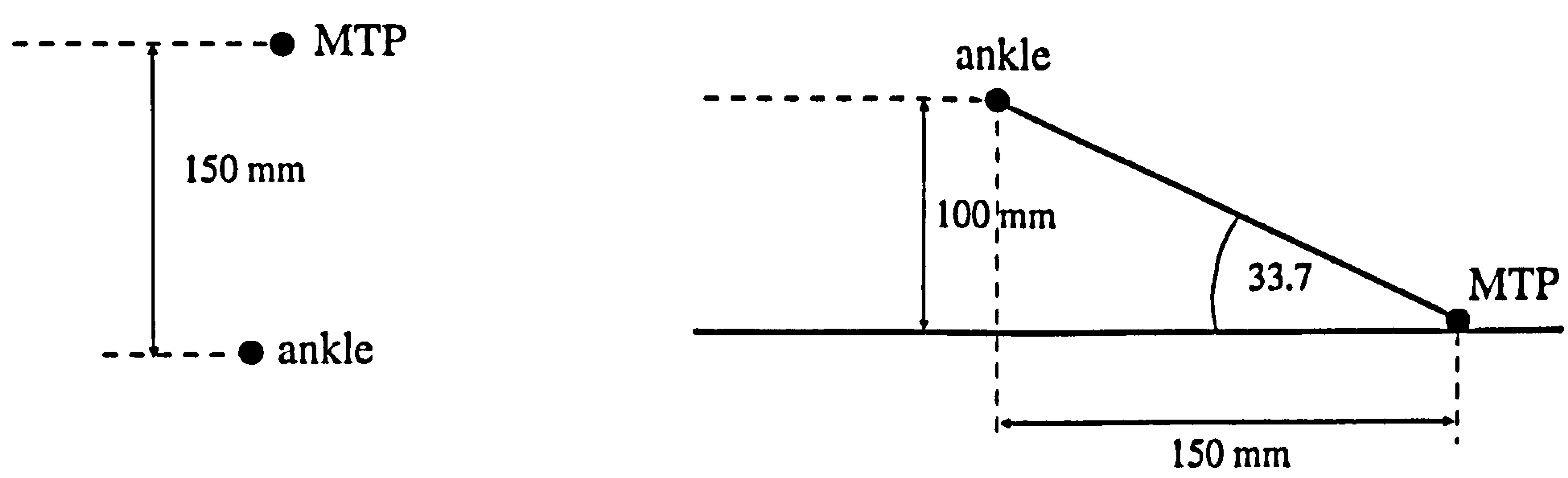


Figure 7.3 (a) Transverse and sagittal plane dimensions with foot in standardised position

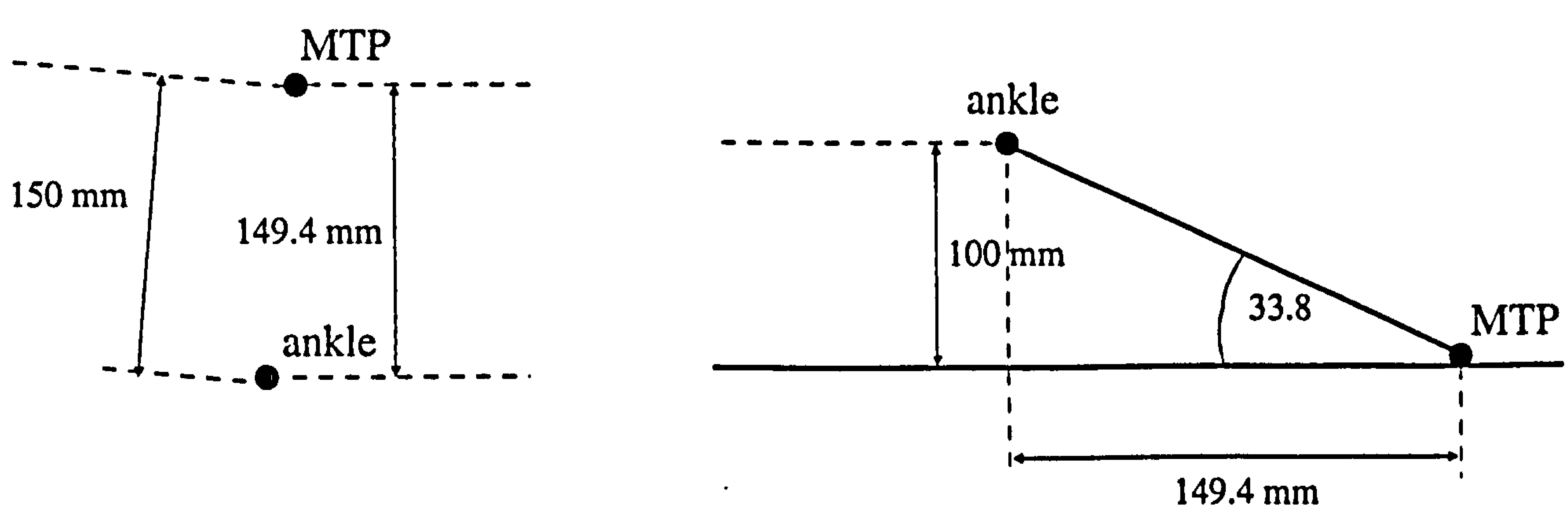


Figure 7.3 (b) Transverse and sagittal plane dimensions with 5 degrees medial rotation

Table 7.1 Mean standing angles for each subject/condition combination in degrees (SD)
(A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

Subject 1	foot angle	ankle angle	knee angle	thigh angle
A	-23.21 (0.04)	76.69 (0.12)	6.38 (0.01)	-3.51 (0.03)
B	-23.16 (0.06)	77.84 (0.11)	7.74 (0.13)	-3.26 (0.02)
C	-23.84 (0.04)	77.17 (0.12)	7.87 (0.10)	-3.14 (0.01)
D	-24.45 (0.08)	73.94 (0.09)	5.16 (0.04)	-3.24 (0.01)
Subject 2				
A				
B	-21.76 (0.07)	73.33 (0.11)	1.58 (0.02)	-3.93 (0.04)
C	-22.20 (0.03)	73.30 (0.07)	0.82 (0.04)	-4.26 (0.04)
D	-26.53 (0.07)	67.19 (0.11)	-0.99 (0.04)	-4.72 (0.02)
Subject 3				
A	-20.45 (0.03)	76.55 (0.03)	5.93 (0.03)	-1.07 (0.05)
B	-17.11 (0.30)	79.65 (0.31)	4.97 (0.04)	-1.79 (0.04)
C	-20.76 (0.34)	75.81 (0.38)	6.97 (0.03)	0.40 (0.02)
D	-23.06 (0.10)	73.52 (0.12)	6.32 (0.06)	-0.27 (0.01)
Subject 4				
A	-21.50 (0.14)	77.21 (0.20)	5.22 (0.06)	-3.50 (0.01)
B	-20.95 (0.21)	76.69 (0.24)	4.70 (0.03)	-2.94 (0.02)
C	-23.08 (0.23)	74.47 (0.15)	4.49 (0.06)	-3.06 (0.03)
D	-25.37 (0.12)	71.88 (0.20)	5.08 (0.05)	-2.17 (0.04)
Subject 5				
A	-26.80 (0.08)	76.87 (0.13)	10.48 (0.11)	-3.19 (0.02)
B	-23.98 (0.23)	77.86 (0.26)	12.07 (0.03)	0.23 (0.06)
C	-27.17 (0.15)	75.02 (0.16)	10.28 (0.02)	-1.91 (0.02)
D	-28.94 (0.28)	74.30 (0.33)	11.21 (0.03)	-2.02 (0.02)
Subject 6				
A	-28.61 (0.14)	66.06 (0.16)	0.74 (0.06)	-3.93 (0.01)
B	-28.46 (0.20)	67.53 (0.25)	1.29 (0.09)	-4.70 (0.01)
C	-31.28 (0.15)	67.70 (0.17)	1.92 (0.02)	-7.07 (0.02)
D	-31.42 (0.35)	64.82 (0.41)	1.73 (0.04)	-4.50 (0.03)
Subject 7				
A	-21.83 (0.08)	79.03 (0.13)	9.17 (0.05)	-1.70 (0.02)
B	-21.94 (0.02)	79.14 (0.12)	9.49 (0.12)	-1.59 (0.03)
C	-23.80 (0.16)	75.57 (0.24)	8.61 (0.08)	-0.76 (0.01)
D	-27.59 (0.12)	74.39 (0.19)	12.54 (0.10)	0.56 (0.03)
Subject 8				
A	-19.45 (0.11)	88.88 (0.07)	21.71 (0.05)	3.38 (0.01)
B	-18.39 (0.35)	85.70 (0.28)	18.61 (0.08)	4.52 (0.01)
C	-20.56 (0.09)	84.35 (0.09)	17.97 (0.06)	3.06 (0.05)
D	-21.65 (0.27)	81.20 (0.22)	16.80 (0.12)	3.95 (0.10)

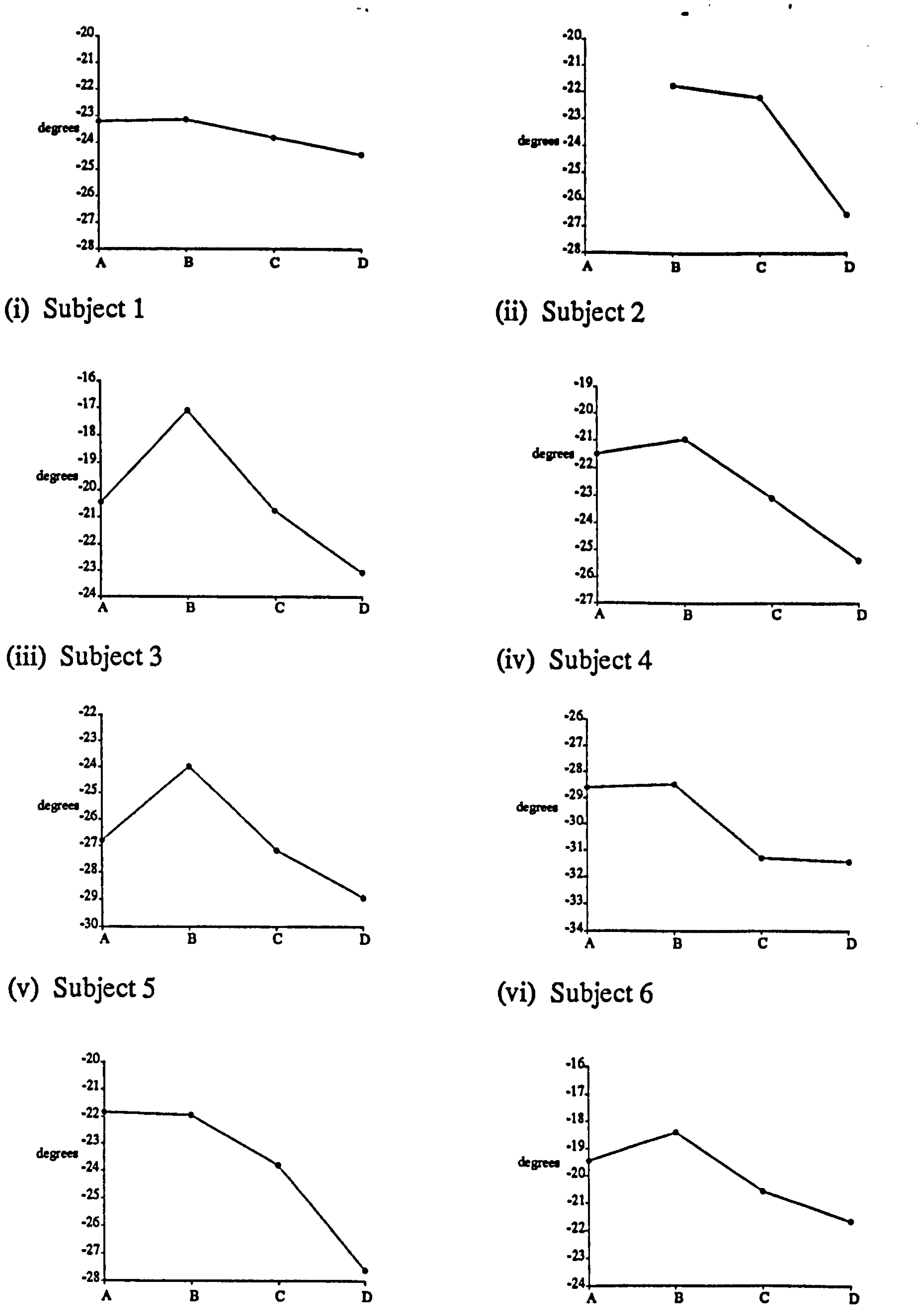
Table 7.2 Standard deviation of markers (mm) during calibration for trial 1 (i) and trial 2 (ii)

(i)

marker	X	Y	Z
hip	0.05	0.35	0.08
knee	0.32	0.56	0.60
ankle	0.10	0.10	0.11
Achilles 1	0.15	0.14	0.22
Achilles 2	0.45	0.17	0.20
calcaneus 1	0.31	0.10	0.23
calcaneus 2	0.57	0.33	0.33
MTP	0.40	0.26	0.35
fp1	0.14	0.34	0.09
fp2	0.34	0.34	0.07
fp3	0.14	0.34	0.09

(ii)

marker	X	Y	Z
hip	0.18	0.19	0.18
knee	0.15	0.17	0.23
ankle	0.13	0.24	0.34
Achilles 1	0.24	0.35	0.31
Achilles 2	0.64	0.11	0.22
calcaneus 1	0.47	0.26	0.32
calcaneus 2	0.80	0.80	0.24
MTP	0.57	0.23	0.21
fp1	0.27	0.94	0.19
fp2	0.39	0.30	0.20
fp3	0.39	0.30	0.20



(i) Subject 1

(ii) Subject 2

(iii) Subject 3

(iv) Subject 4

(v) Subject 5

(vi) Subject 6

(vii) Subject 7

(viii) Subject 8

A: barefoot
B: zero heel lift
C: 7.5 mm lift
D: 15 mm lift

Figure 7.4 Foot angle for each condition for each subject

The presence of heel lifts was found to influence the orientation of the entire lower extremity, not only the foot segment (Table 7.1). After the foot angle, the ankle angle was influenced most severely by the changes in condition. This is as expected, given the fact that the ankle angle was dependent on the foot orientation and the orientation of the lower leg. The changes in ankle angle with heel lift are illustrated more clearly in Table 7.3, in which differences in ankle angle between conditions are presented. Changes in knee and thigh angles were also evident across conditions, although these angle changes were generally smaller than the changes observed in foot and ankle angles (Table 7.1).

Table 7.3 Changes in ankle angle (degrees) with heel lift increase

	7.5 mm lift	15 mm lift	difference
Subject 1	-0.96	-5.61	-4.65
Subject 2	-1.51	-7.52	-6.01
Subject 3	-3.67	-3.87	-0.20
Subject 4	-1.35	-4.07	-2.72
Subject 5	-3.43	-3.77	-0.34
Subject 6	-0.32	-5.02	-4.70
Subject 7	-3.92	-3.83	+0.09
Subject 8	-0.68	-6.42	-5.74

The column headed '7.5 mm' provides the difference in foot angle between zero heel lift and 7.5 mm lift. The column headed '15 mm' provides the difference between zero heel lift and 15 mm heel lift. The final column gives the difference in foot angle between the 7.5 mm and 15 mm heel lift conditions.

Characterisation of angle plots

Similar plots of foot, ankle, knee and hip angles during ground contact were obtained for each subject. Typical examples are provided in Figure 7.5. After examination, selected parameters of these plots were chosen to describe the time histories.

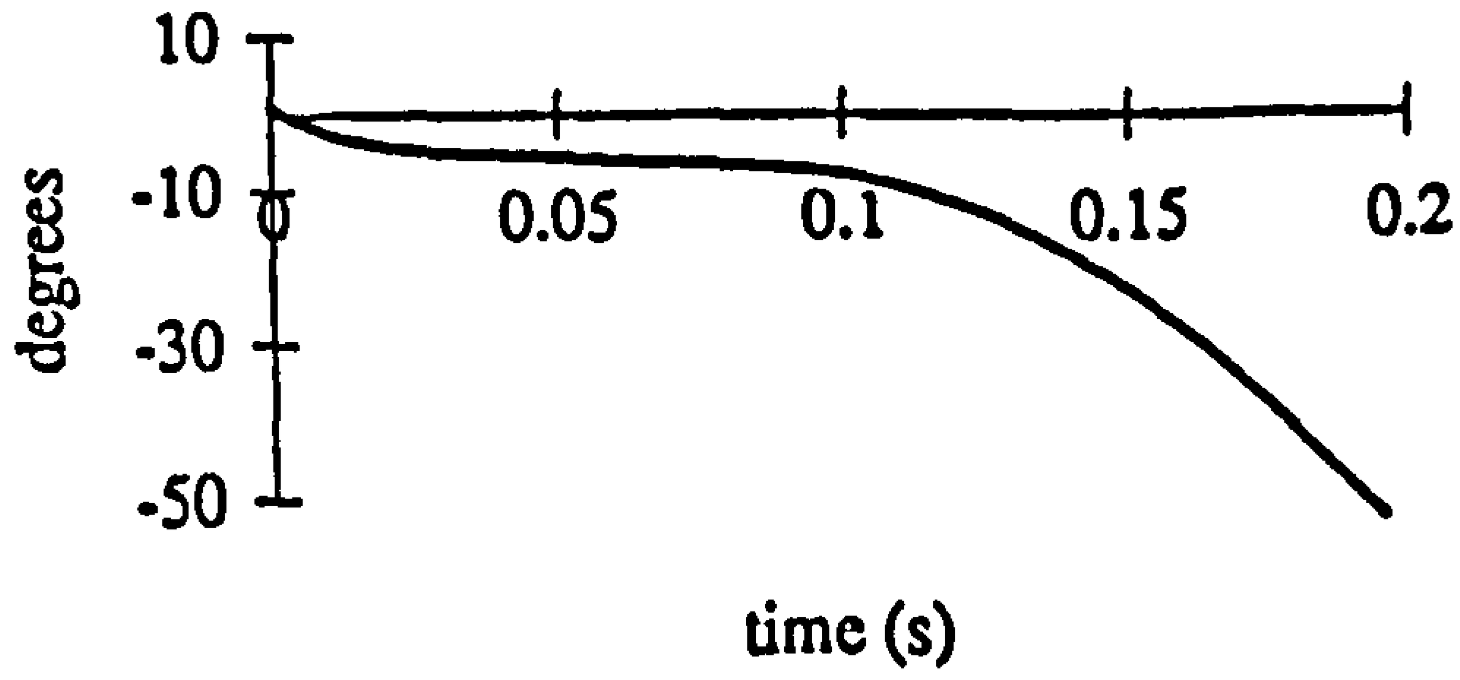
Three distinct patterns were identified across trials for the foot angle traces immediately following impact. An example of each of these these is illustrated in Figure 7.6. For some trials, the impact angle was maintained for approximately the first 30% of ground contact. For some trials, the foot segment showed an initial anti-clockwise rotation, contributing to an increased ankle joint dorsi-flexion. A peak angle was reached within the first 20% of stance, beyond which the foot rotated in the opposite direction. For the remaining trials, the foot angle demonstrated an increase in magnitude throughout ground contact, contributing to plantar-flexion of the ankle joint. For all trials, beyond approximately 30% of the stance phase the foot angle increased in magnitude, indicating an increase in the inclination of the foot to the horizontal which contributed increasingly to plantar-flexion of the ankle joint.

Since the changes in foot angle during approximately the first 30% of stance were minimal, these changes were not quantified. The foot trace was characterised by the impact angle and the angle when the foot was flat. This 'flat foot' angle was defined as the angle at the time when the foot angle time history demonstrated a minimum slope, indicating that momentarily there was little, if any, change in this angle. This angle represented the orientation of the foot relative to the ground during midstance when the foot was flat and in a position determined predominantly by the presence of the heel lifts. The difference between foot impact angle and standing angle was calculated for each subject and condition to investigate the possibility of impact angle being related to the perception of the 'flat foot' standing angle.

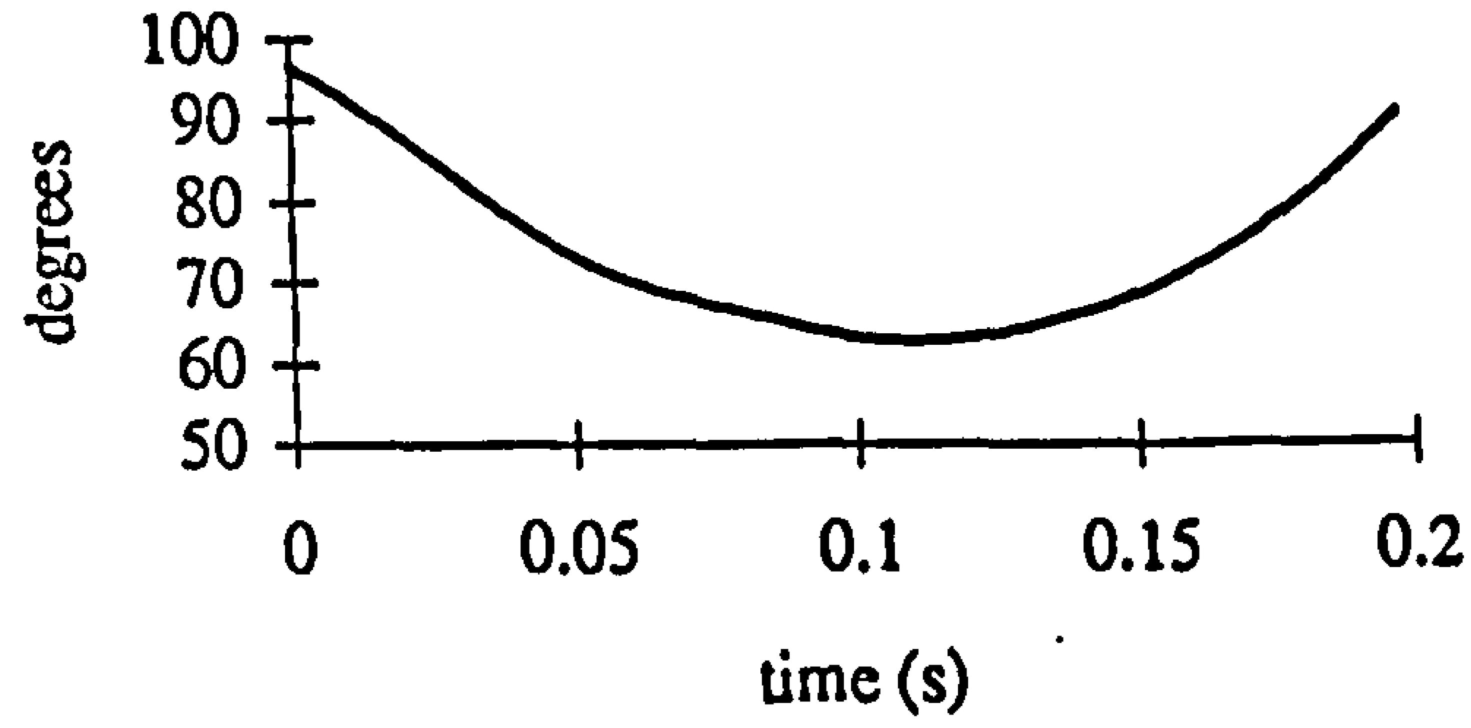
For all subjects, ankle angle was found to increase steadily up to a maximum around 50% of the stance phase, and to show a steady decrease in magnitude beyond this maximum (Figure 7.5). The ankle angle for each trial was characterised by the magnitude of the angle at impact, and the maximum angle. The time of occurrence of maximum ankle angle in milliseconds was also identified.

Similar shaped knee angle traces were observed across subjects (Figure 7.5). The knee angle showed a rise up to a maximum at between 30% and 45% of total stance time. A steady decrease in angle then occurred to the end of stance. The initial knee angle at impact, the maximum knee angle, and the time of maximum knee angle were the variables used to characterise the knee angle plots for each trial.

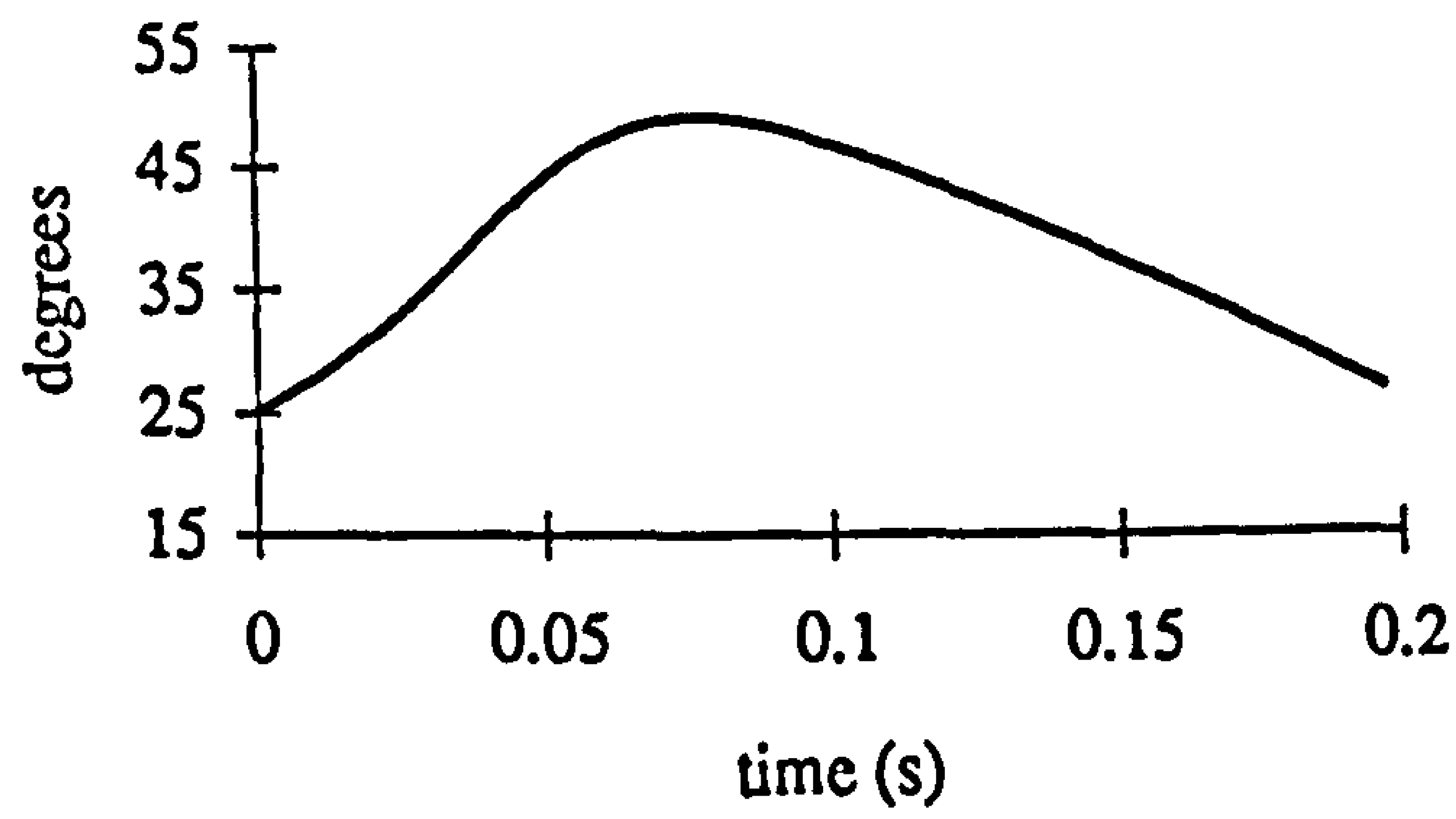
The thigh angle demonstrated minimal variation over approximately the first 30% of stance, followed by a steady decrease in magnitude (Figure 7.5). The timing of the start of the decrease in this angle generally coincided with the maximum knee angle. The time histories of thigh angle were characterised by the impact angle alone.



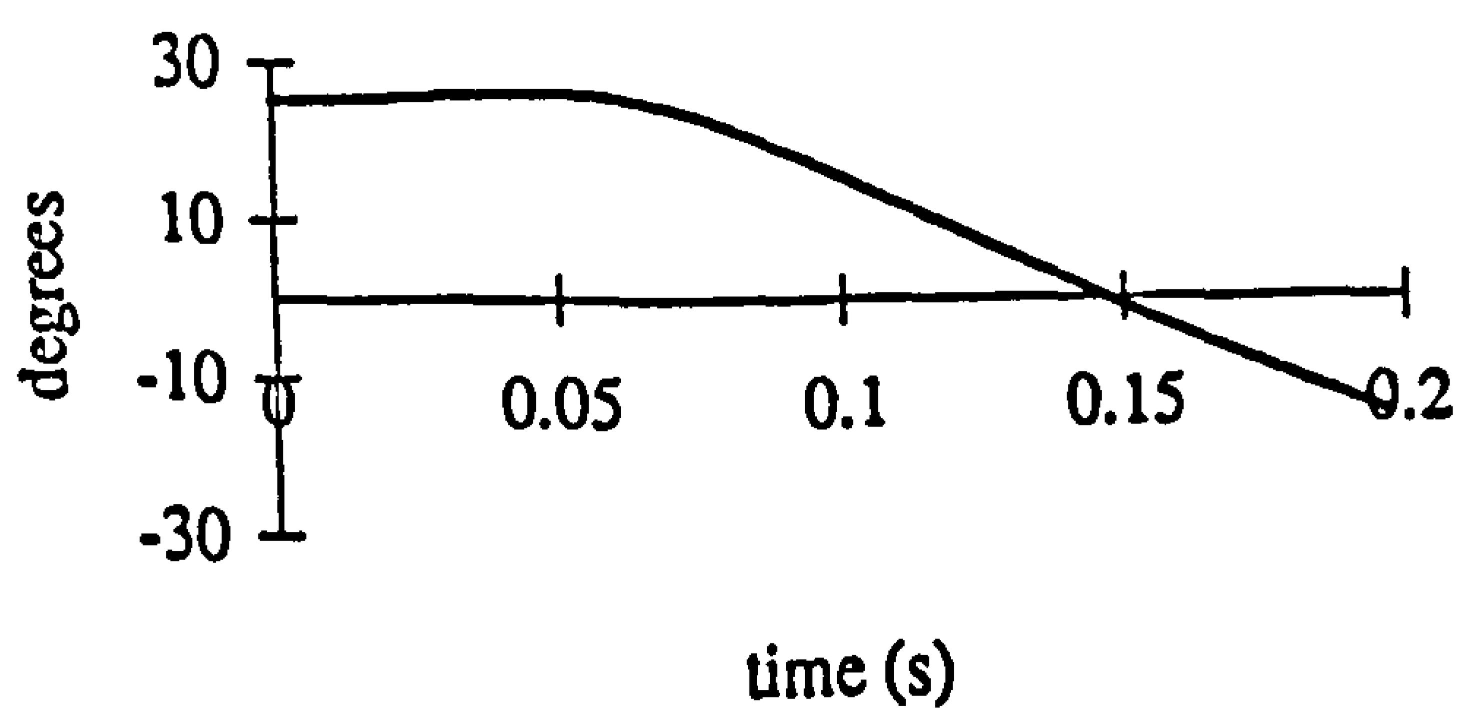
(i) Foot angle



(ii) Ankle angle

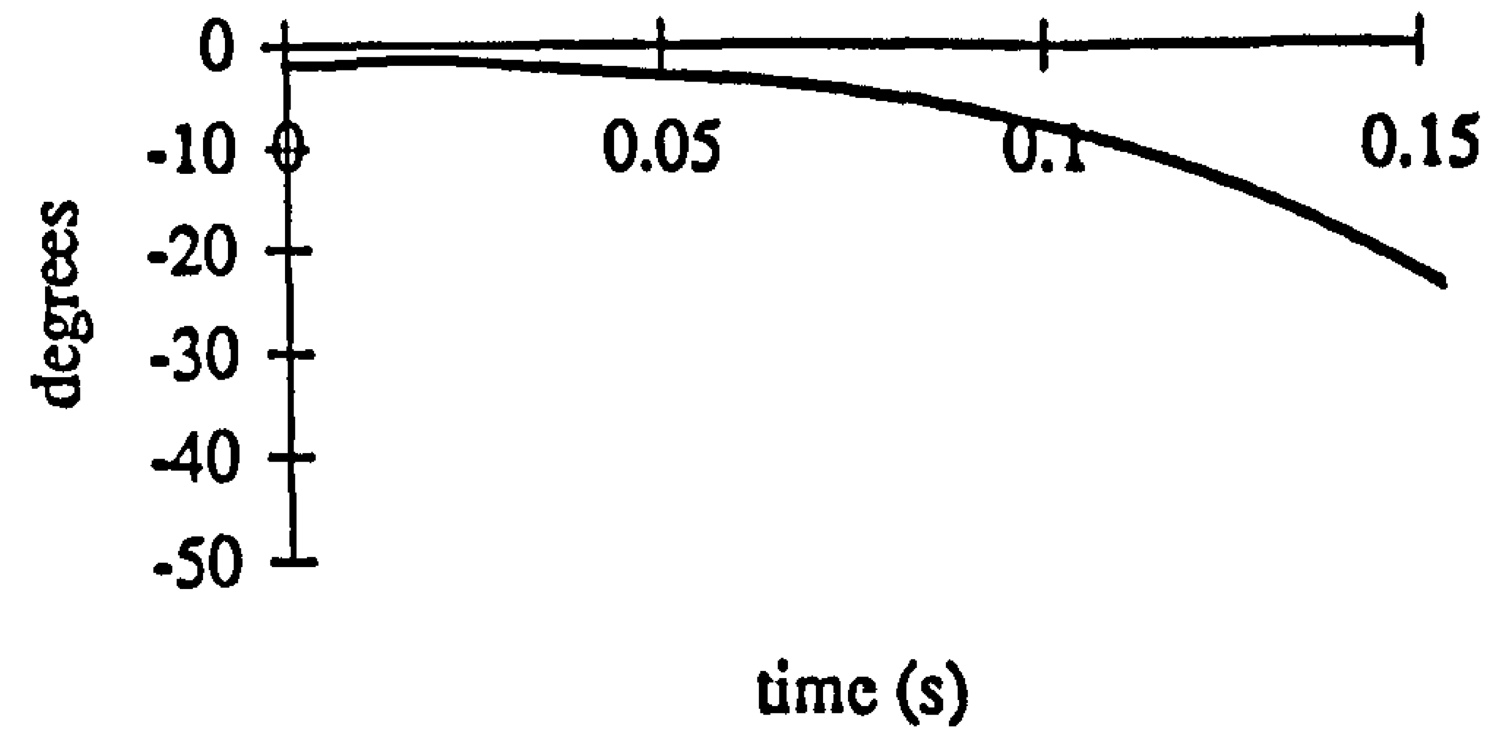


(iii) Knee angle

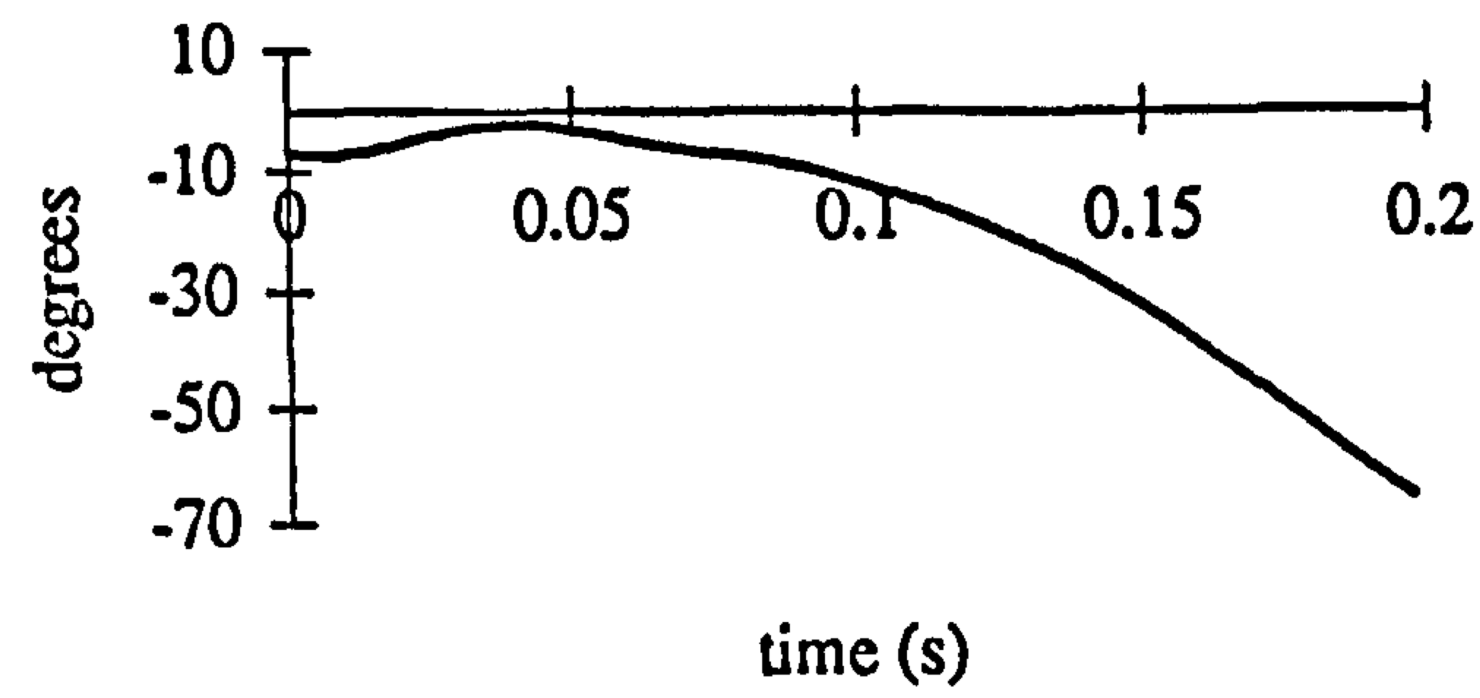


(iv) Thigh angle

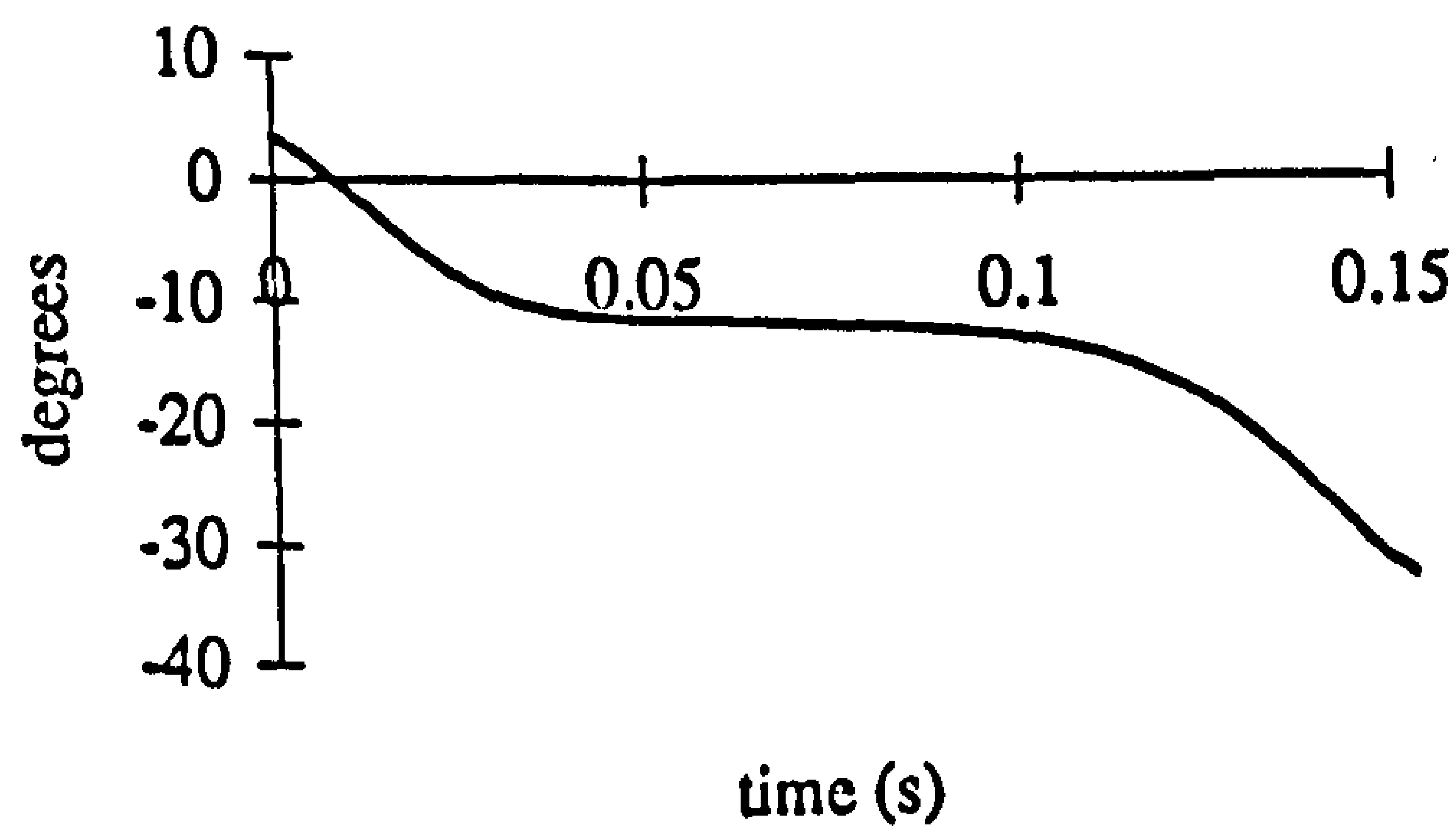
Figure 7.5 Typical angle time histories for the foot, ankle, knee and thigh (Subject 7)



(i) Subject 3



(ii) Subject 4



(iii) Subject 8

Figure 7.6 Characteristic foot time histories

Evaluation of methods

Consideration of the influence of ±0.4 mm on foot angle using the dimensions illustrated in Figure 7.3 revealed that calculated foot angles were reliable to within 0.2 degrees. For each subject, the medial-lateral orientation of the foot during midstance did not differ across conditions by more than five degrees. It was demonstrated previously that changes in the medial-lateral orientation of the foot of five degrees in the transverse plane resulted in a variation in foot angle of 0.1 degree, indicating that ab/adduction of the foot did not influence comparisons between conditions by more than 0.1 degree.

For the typical dimensions illustrated in Figure 7.3, a frontal plane rotation of the foot segment relative to the lower leg of five degrees will result in a 0.4 mm variation in the relative z-coordinates of the MTP and ankle markers. Changes in the x-coordinates will not influence sagittal plane angles. The subsequent influence of the described five degree rotation on sagittal plane foot angle was found to be 0.1 degrees. Inversion-eversion movement of the calcaneus out of the sagittal plane was therefore assumed to have a negligible influence on sagittal plane joint angle comparisons across conditions.

Impact angles

Foot angles at ground impact are provided in Table 7.4. These angles were negative throughout stance, as expected given the foot segment representation as a straight line from the ankle to the MTP marker. An increase in magnitude indicated an increase in the inclination of the foot to the horizontal, and thus an increase in the contribution of foot angle to ankle joint plantar-flexion. Investigation of the influence of attaching lifts to the rearfoot and forefoot revealed that six subjects demonstrated a decreased foot impact angle for the zero heel lift condition compared with the barefoot condition, with two of these decreases being significant. This decrease in angle indicated an increased contribution of the foot segment to ankle dorsi-flexion. The remaining two subjects showed small increases in foot impact angle, which were found not to be significant.

Heel lift influence on impact angles was assessed by comparison of zero heel lift with the increased heel lift conditions. For the 7.5 mm heel lift condition compared with zero heel lift, an increase in magnitude of foot contact angle occurred for six subjects, with all of these increases being significant ($p<0.05$). The exceptions were Subject 7 with a small difference between conditions which was not significant, and Subject 5 who demonstrated a significant decrease in magnitude of foot angle for the heel lift conditions. With the exception of Subject 5, all subjects demonstrated an increase in magnitude of foot angle at initial ground contact for the 15 mm heel lift condition compared with zero heel lift. This difference was found to be significant for three of the subjects. Foot angle at impact was, therefore, generally found to have a greater magnitude when heel lift was increased, indicating an increased contribution to ankle plantar-flexion (or decreased contribution to dorsi-flexion) at impact.

Ankle angles at impact are presented in Table 7.5. For all but Subject 1, who demonstrated a small decrease which was not significant, the ankle angle at impact was increased by the attachment of lifts to the rearfoot and forefoot compared with the barefoot condition, with three of these increases being significant. This increase in ankle angle indicated a more dorsi-flexed orientation of the ankle at ground contact.

For all but Subject 5, a decrease in ankle angle at impact was observed for both heel lift conditions compared with zero heel lift. Of these decreases, for all but Subject 3, one or both were significant ($p<0.05$). Subject 5 demonstrated a significant increase in ankle impact angle for both heel lift conditions compared with the zero heel lift condition ($p<0.05$). The decreased angle observed for seven of the eight subjects indicated that for increased heel lift conditions the ankle joint was more plantar-flexed (or less dorsi-flexed) at impact. The variation in ankle angle at impact across conditions is illustrated in Figure 7.7, for each subject separately.

Changes in shank impact angle across conditions were shown to be relatively small, and were varied across subjects, indicating that the main contribution to ankle angle changes was made by variations in the inclination of the foot (Appendix I). The relative contribution of foot and shank orientation to resultant ankle angles varied across subjects.

For all subjects, the knee angle at impact was increased by the attachment of lifts to rearfoot and forefoot, indicating that the knee was more flexed on ground contact for this condition compared with barefoot (Appendix I). Variations in knee angle between heel lift conditions were small and showed no clear trends across subjects.

Table 7.6 provides the difference between foot angles at impact and the standing calibration angles. Standing calibration angles were taken as neutral angles (zero degrees), and negative foot angles indicated that the plantar surface of the heel was lower than the plantar surface of the forefoot. Figure 7.8 illustrates the calculation of the difference between foot standing calibration angle and impact angle with difference = $(\theta_1 - \theta_0)$. Negative angles were therefore expected for the rearfoot strikers employed in the present study, and indicated a contribution of the foot angle to increased ankle dorsi-flexion relative to the neutral. For most subjects, negative differences between impact and standing angles were demonstrated, supporting the definition of these subjects as rearfoot strikers. For a small number of subject/ condition combinations positive differences were found, indicating that the forefoot was lower at impact than in the standing calibration position. For all of these cases the positive angles were small, indicating that the foot was close to neutral on impact with the ground. With the exception of Subject 5, for each subject there was around four degrees maximum variation in the $(\theta_1 - \theta_0)$ difference across conditions. Subject 5 demonstrated a very large variation across conditions, with a pronounced rearfoot strike when heel lift was introduced.

Table 7.4 Foot angle at impact for each subject/condition combination in degrees (SD)
(A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

Condition	A	B	C	D
Subject 1*	-11.9 (2.2)	-12.2 (4.3)	-16.8 (2.2)	-13.9 (4.0)
Subject 2*	-12.4 (1.3)	-12.8 (0.8)	-16.7 (2.6)	-14.7 (1.6)
Subject 3*	-21.6 (1.0)	-19.3 (1.0)	-20.9 (0.9)	-21.5 (0.8)
Subject 4*	-19.4 (1.7)	-18.5 (1.5)	-21.9 (2.0)	-20.6 (2.6)
Subject 5*	-26.1 (3.2)	-25.3 (3.6)	-14.0 (1.9)	-13.9 (1.9)
Subject 6*	-22.7 (1.5)	-22.1 (3.0)	-26.6 (2.1)	-28.2 (2.0)
Subject 7*	-13.1 (2.4)	-12.2 (1.9)	-12.0 (2.7)	-16.0 (3.6)
Subject 8*	-18.6 (2.8)	-16.5 (1.9)	-21.2 (1.6)	-23.7 (2.4)

*$p<0.05$ Subject 1: B versus C
Subject 2: B versus C; C versus D
Subject 3: A versus B; B versus C; B versus D
Subject 4: B versus C
Subject 5: B versus C; B versus D
Subject 6: B versus C; B versus D
Subject 7: B versus D; C versus D
Subject 8: A versus B; B versus C; B versus D

Table 7.5 Ankle angle at impact for each subject/condition combination in degrees (SD) (A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

Condition	A	B	C	D
Subject 1*	78.1 (0.8)	78.0 (2.0)	74.5 (1.3)	77.7 (1.4)
Subject 2*	73.4 (0.8)	74.0 (0.8)	69.6 (1.2)	70.9 ()
Subject 3*	65.4 (1.4)	68.8 (1.3)	67.4 (1.8)	67.2 (1.7)
Subject 4*	65.8 (1.2)	68.8 (1.1)	65.9 (1.8)	65.1 (0.8)
Subject 5*	65.1 (4.5)	62.2 (4.4)	70.9 (1.0)	71.0 (1.6)
Subject 6*	63.5 (0.9)	66.3 (1.8)	60.7 (1.3)	60.1 (1.8)
Subject 7*	77.8 (1.2)	79.0 (0.9)	78.2 (1.1)	74.2 (1.1)
Subject 8*	69.3 (4.8)	72.0 (2.4)	68.8 (2.0)	64.5 (3.8)

*$p<0.05$ Subject 1: B versus C; C versus D
Subject 2: B versus C; B versus D
Subject 3: A versus B
Subject 4: A versus B; B versus C; B versus D
Subject 5: B versus C; B versus D
Subject 6: A versus B; B versus C; B versus D
Subject 7: B versus D; C versus D
Subject 8: B versus C; B versus D

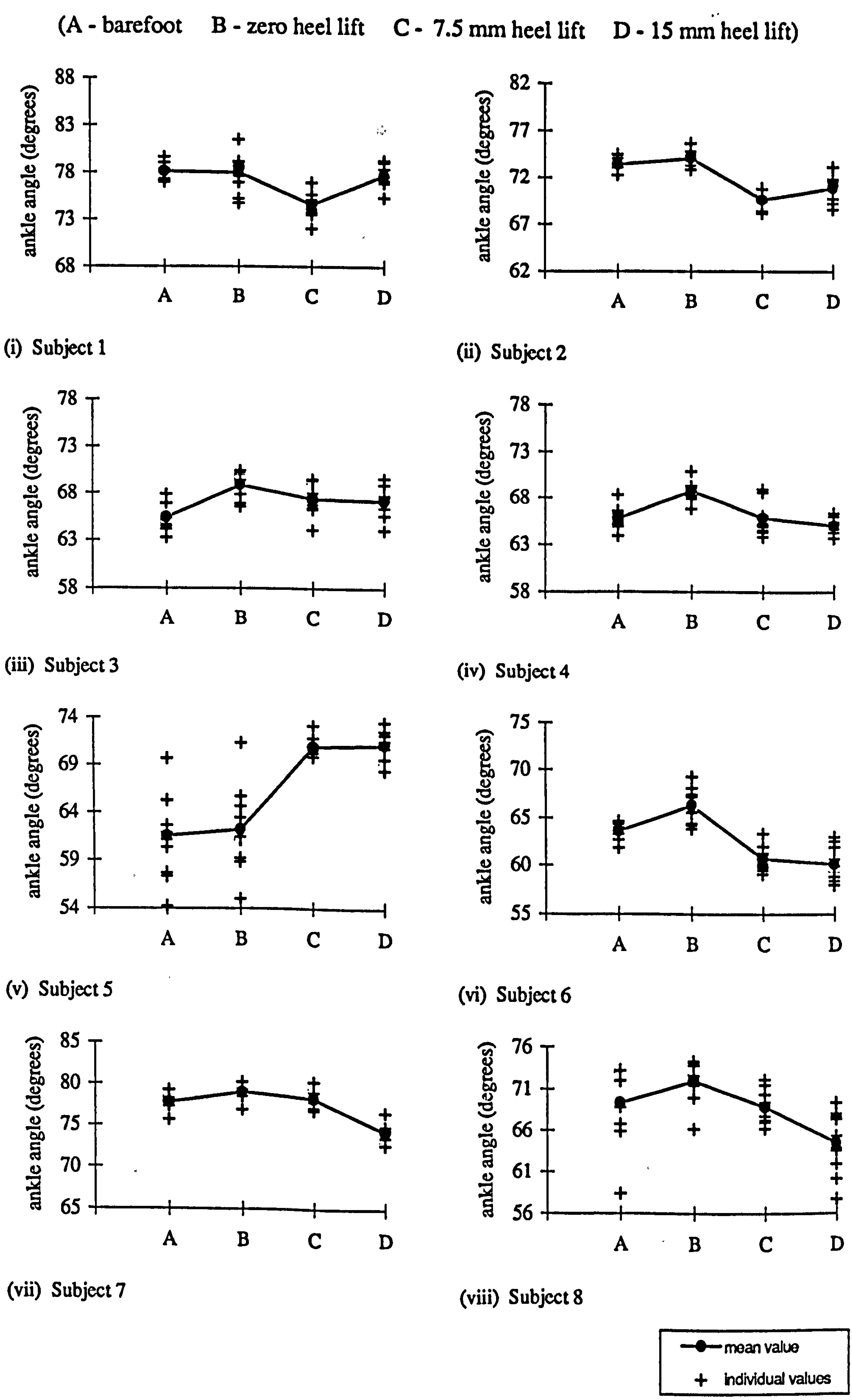
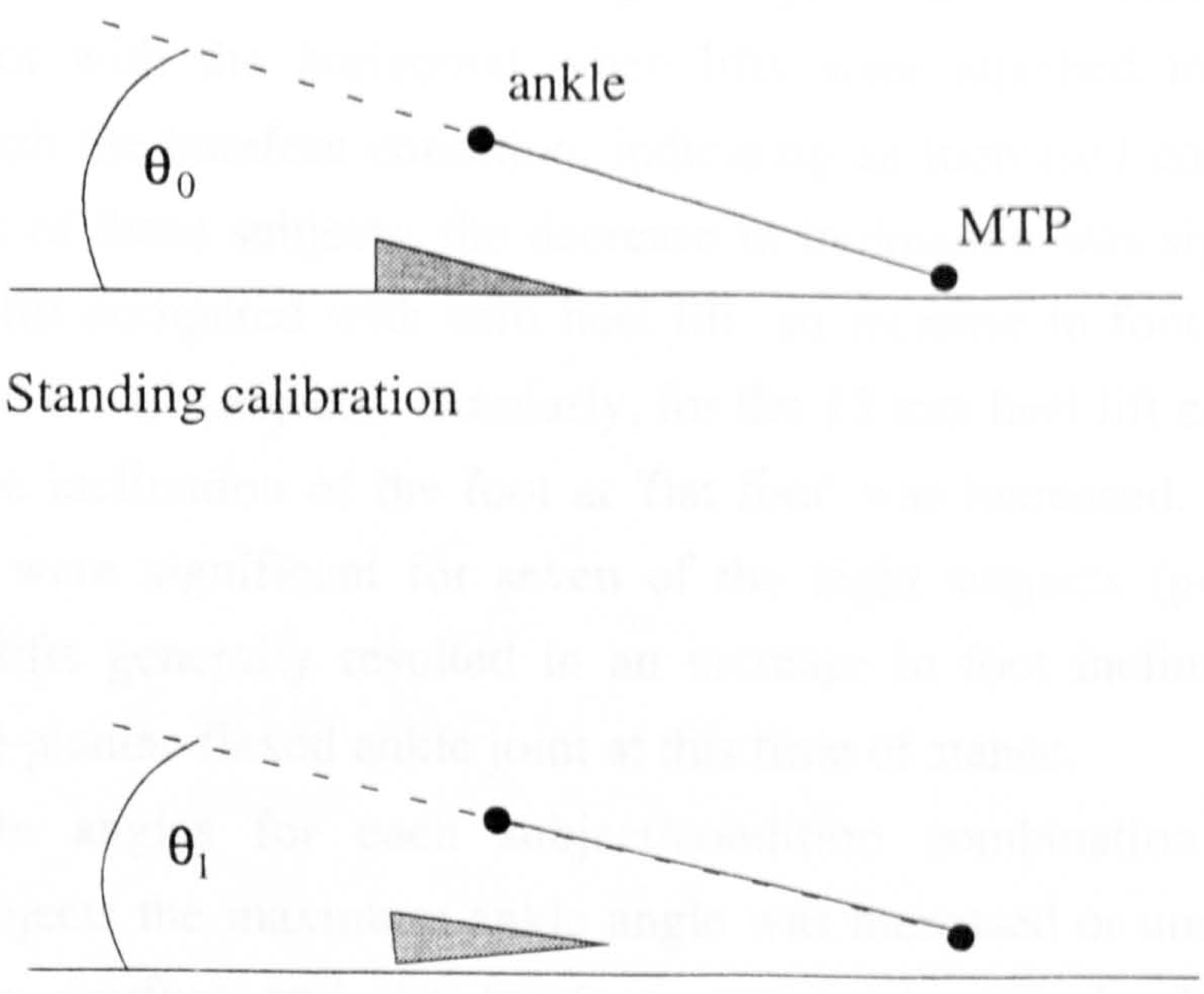


Figure 7.7 Impact ankle angles for each subject/condition combination



(i) Standing calibration

(ii) Impact angle

Figure 7.8 Illustration of foot calibration angle (θ_0) and impact angle (θ_1), with difference = (θ_1 - θ_0)

Table 7.6 Impact angles relative to standing calibration angles in degrees (θ_1 - θ_0)
(A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

Condition	A	B	C	D
Subject 1	-11.3	-11.0	-7.0	-10.6
Subject 2	-11.7	-9.0	-5.3	-11.8
Subject 3	+1.1	+2.2	+0.1	-1.6
Subject 4	-2.1	-2.5	-1.2	-4.8
Subject 5	-0.7	+1.3	-13.2	-15.1
Subject 6	-5.9	-6.4	-4.7	-3.2
Subject 7	-8.7	-9.7	-11.6	-11.6
Subject 8	-0.9	-1.9	+0.6	+2.0

Sagittal plane joint angles during stance

Table 7.7 provides the magnitudes of 'flat foot' angle for each subject/condition combination. It was found that seven of the eight subjects demonstrated a decrease in the inclination of the foot with the horizontal when lifts were attached to the rearfoot and forefoot, compared with the barefoot condition, indicating an increased contribution to ankle dorsi-flexion. For six of these subjects, the decrease in inclination was significant ($p<0.05$). For the 7.5 mm heel lift compared with zero heel lift, an increase in foot inclination at 'flat foot' was demonstrated for all subjects. Similarly, for the 15 mm heel lift condition compared with zero heel lift, the inclination of the foot at 'flat foot' was increased. For both heel lift conditions, increases were significant for seven of the eight subjects ($p<0.05$). Thus, the introduction of heel lifts generally resulted in an increase in foot inclination at 'flat foot', contributing to a more plantar-flexed ankle joint at this time of stance.

Maximum ankle angles for each subject/condition combination are provided in Table 7.8. For all subjects the maximum ankle angle was increased or unchanged when lifts were attached to the rearfoot and the forefoot compared with the barefoot condition, indicating an increase in maximum ankle dorsi-flexion. For four of the eight subjects this increase was significant ($p<0.05$). For the 7.5 mm heel lift compared with the zero heel lift condition, there was a decrease in ankle joint maximum angle for all subjects, indicating decreased ankle joint dorsi-flexion. This decrease was significant for all but two subjects ($p<0.05$). For the 15 mm heel lift condition compared with zero heel lift, maximum ankle angle was also decreased for all subjects, with this decrease being significant for all but one subject. In general, the trend was for the maximum ankle angle to decrease with heel lift increase, although only 3 subjects showed significant decreases when comparing the 7.5 mm and 15 mm heel lift conditions. The variation in maximum ankle angle across conditions is illustrated in Figure 7.9, for each subject separately.

Times of occurrence of maximum ankle angle for each subject/condition combination are provided in Table 7.9, with times in milliseconds (ms) from initial ground contact. The attachment of lifts to the rearfoot and the forefoot resulted in a reduction in the time to maximum ankle angle for six of the eight subjects, with two of these decreases in time being significant ($p<0.05$). The remaining two subjects showed small increases in occurrence time which were not significant. For the 7.5 mm heel lift condition compared with zero heel lift, there was an increase in the time of occurrence of maximum ankle angle for seven of the eight subjects, with the remaining subject (Subject 2) showing no change in this variable. Two of these increases were found to be significant. For the 15 mm heel lift condition compared with zero heel lift, there was an increase in the time to maximum ankle angle for seven of the eight subjects, with four of these increases being significant. The remaining subject showed a small decrease in time of occurrence which was not significant. There is a clear trend for the maximum ankle angle to occur at an increased time after initial ground contact for the heel lift conditions compared with zero heel lift, as illustrated in Figure 7.10.

Maximum knee angles for each subject/condition combination showed both increases and decreases across subjects in response to the attachment of lifts to the rearfoot and the forefoot, with any changes in angle generally being small and no significant changes being detected (Appendix I). For all subjects, maximum knee angle was increased for the 7.5 mm heel lift, compared with zero heel lift, although this increase was small (generally less than one degree), and only significant for one subject. For all but one subject, maximum knee angle was increased for the 15 mm heel lift compared with zero heel lift, although none of these increases were significant. In general similar maximum knee angles were seen for the 7.5 mm and 15 mm heel lift conditions. Thus, in general, the raising of the heel relative to the forefoot resulted in a small increase in the magnitude of the maximum knee angle, indicating increased knee flexion. This trend is illustrated in Figure 7.11.

The attachment of lifts to the rearfoot and forefoot resulted in a decrease in the time to maximum knee angle for five of the eight subjects, with two of these decreases being significant (Appendix I). One subject showed no change in the time to maximum knee angle. Two subjects showed small increases which were not significant. For the 7.5 mm heel lift condition compared with zero heel lift, there was an increase in the time to maximum knee angle for seven of the eight subjects, with only one of these increases being significant. The remaining showed a small decrease which was not significant. For the 15 mm heel lift condition compared with zero heel lift, there was an increase in the time to maximum knee angle for seven of the eight subjects, with the remaining subject showing no change in this variable. Two of the increases observed were found to be significant. For the increase in heel lift from 7.5 mm to 15 mm, the time to maximum knee angle was increased for six of the eight subjects, with none of these increases being found to be significant. Despite only a small number of significant differences being detected, there was clearly a trend for maximum knee angle to occur at a later time beyond initial ground contact for the heel lift conditions compared with zero heel lift. Generally, this increase in time was found to be greater for the larger heel lift.

Table 7.7 'Flat foot' angle in degrees (SD)
(A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

	A	B	C	D
Subject 1*	-26.7 (1.2)	-24.4 (1.4)	-28.2 (0.5)	-28.3 (0.6)
Subject 2*	-24.2 (0.3)	-24.0 (0.4)	-26.4 (0.7)	-26.0 (0.8)
Subject 3*	-20.8 (0.9)	-18.4 (0.8)	-20.7 (1.0)	-23.4 (0.8)
Subject 4*	-23.3 (0.8)	-19.7 (1.0)	-26.0 (1.1)	-27.8 (0.8)
Subject 5*	-29.6 (1.2)	-26.6 (1.4)	-30.8 (1.6)	-32.8 (1.9)
Subject 6*	-29.5 (0.5)	-25.3 (2.8)	-29.2 (1.9)	-28.3 (3.6)
Subject 7*	-24.0 (0.4)	-23.2 (0.9)	-25.4 (0.3)	-29.2 (0.4)
Subject 8*	-17.7 (2.5)	-19.3 (1.3)	-19.9 (2.1)	-22.5 (1.7)

*$p<0.05$ Subject 1: A versus B; B versus C; B versus D
Subject 2: B versus C; B versus D
Subject 3: A versus B; B versus C; B versus D; C versus D
Subject 4: A versus B; B versus C; B versus D; C versus D
Subject 5: A versus B; B versus C; B versus D; C versus D
Subject 6: A versus B; B versus C
Subject 7: A versus B; B versus C; B versus D; C versus D
Subject 8: B versus D; C versus D

Table 7.8 Maximum ankle angle in degrees (SD)
(A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

	A	B	C	D
Subject 1*	94.7 (1.5)	94.7 (1.6)	91.9 (1.3)	90.9 (2.1)
Subject 2*	94.1 (0.6)	95.7 (0.9)	92.5 (1.4)	93.0 (1.3)
Subject 3*	97.4 (1.5)	100.3 (0.6)	98.5 (1.3)	95.1 (1.2)
Subject 4*	97.5 (1.2)	99.9 (2.2)	94.5 (2.6)	92.6 (1.7)
Subject 5*	91.3 (1.0)	94.0 (0.6)	93.2 (2.3)	91.6 (1.9)
Subject 6*	89.6 (1.2)	91.9 (3.2)	89.4 (1.9)	88.4 (2.2)
Subject 7*	97.7 (1.0)	97.8 (0.6)	95.8 (1.0)	93.6 (1.0)
Subject 8*	100.0 (0.9)	100.5 (1.8)	97.5 (0.9)	98.5 (1.0)

*$p<0.05$ Subject 1: B versus C; B versus D
Subject 2: A versus B; B versus C; B versus D
Subject 3: A versus B; B versus C; B versus D; C versus D
Subject 4: A versus B; B versus C; B versus D
Subject 5: A versus B; B versus D
Subject 6: C versus D
Subject 7: B versus C; B versus D; C versus D
Subject 8: B versus C; B versus D

Table 7.9 Times of occurrence of maximum ankle angle in ms (SD)
(A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

	A	B	C	D
Subject 1	107 (8)	102 (9)	103 (4)	103 (10)
Subject 2	103 (3)	108 (2)	108 (6)	110 (7)
Subject 3*	103 (5)	94 (6)	99 (6)	104 (4)
Subject 4*	110 (5)	102 (3)	105 (8)	112 (8)
Subject 5*	102 (6)	104 (6)	128 (8)	123 (4)
Subject 6	104 (5)	93 (10)	103 (4)	92 (16)
Subject 7*	101 (5)	100 (8)	108 (5)	113 (8)
Subject 8	88 (12)	83 (6)	89 (7)	87 (7)

*$p<0.05$ Subject 1: none
Subject 2: none
Subject 3: A versus B; B versus D
Subject 4: A versus B; B versus D
Subject 5: B versus C; B versus D
Subject 6: none
Subject 7: B versus C; B versus D
Subject 8: none

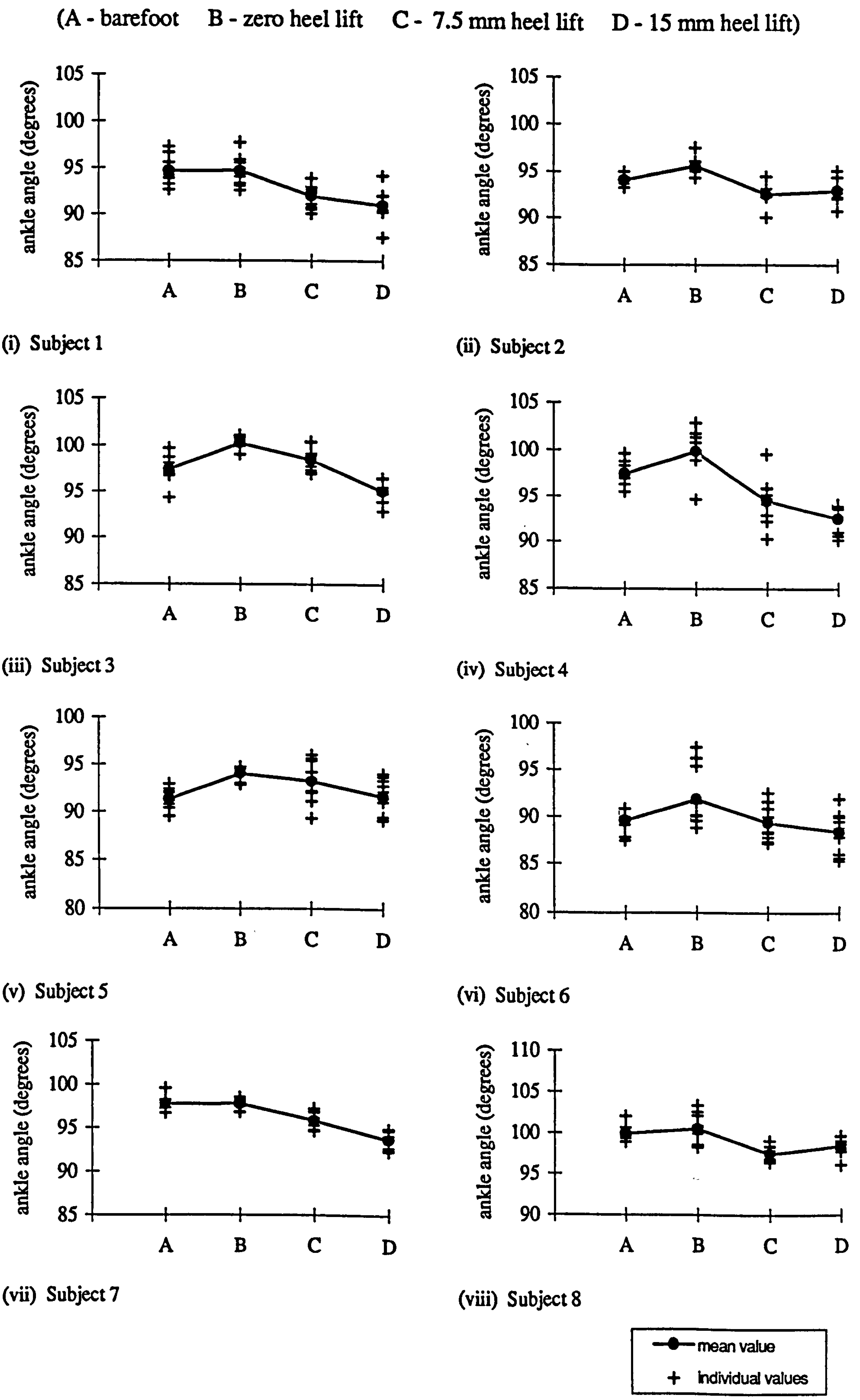


Figure 7.9 Maximum ankle angle for each subject/condition combination

(A - barefoot   B - zero heel lift   C - 7.5 mm heel lift   D - 15 mm heel lift)

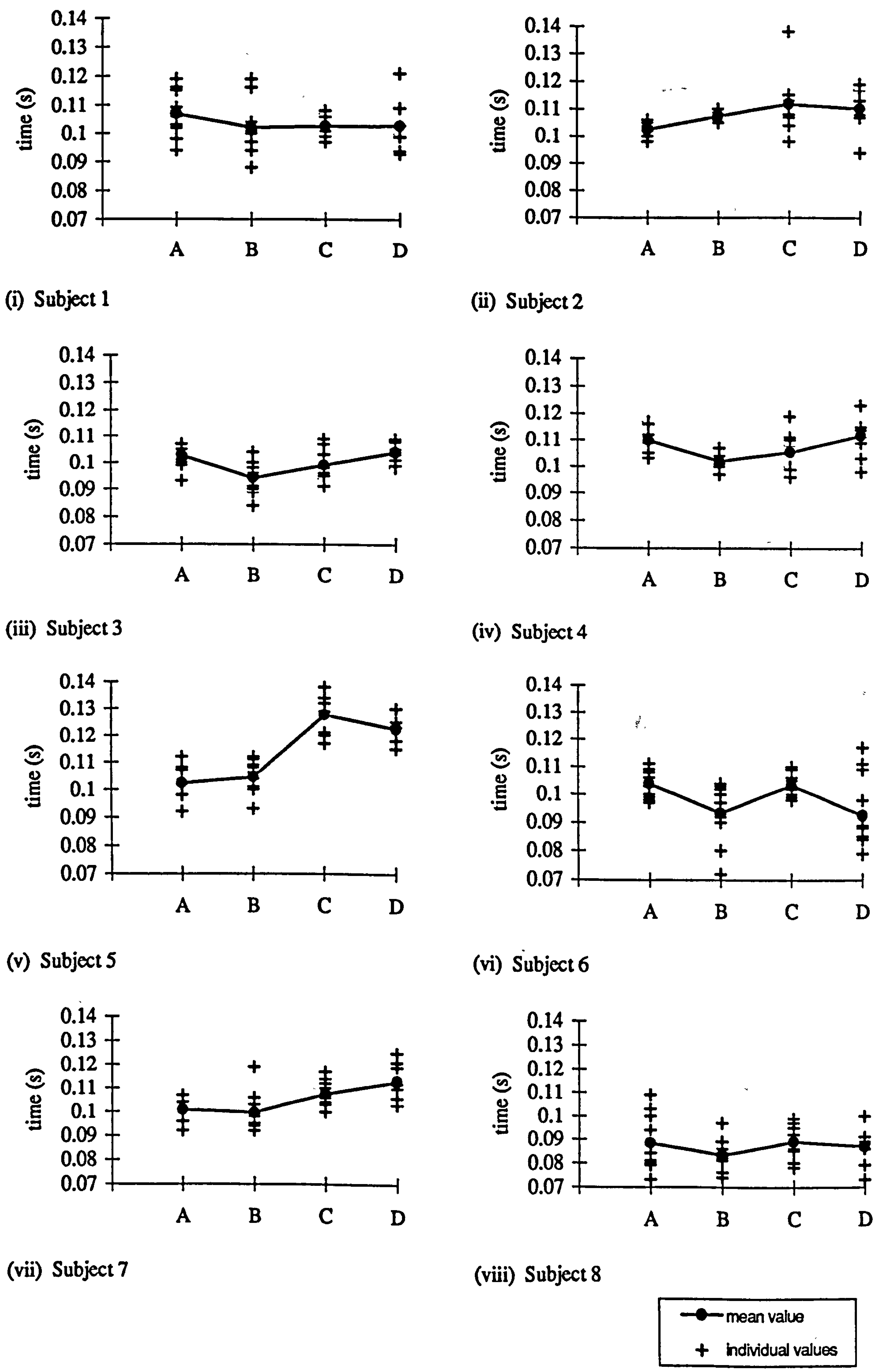


Figure 7.10 Time of peak ankle angle for each subject/condition

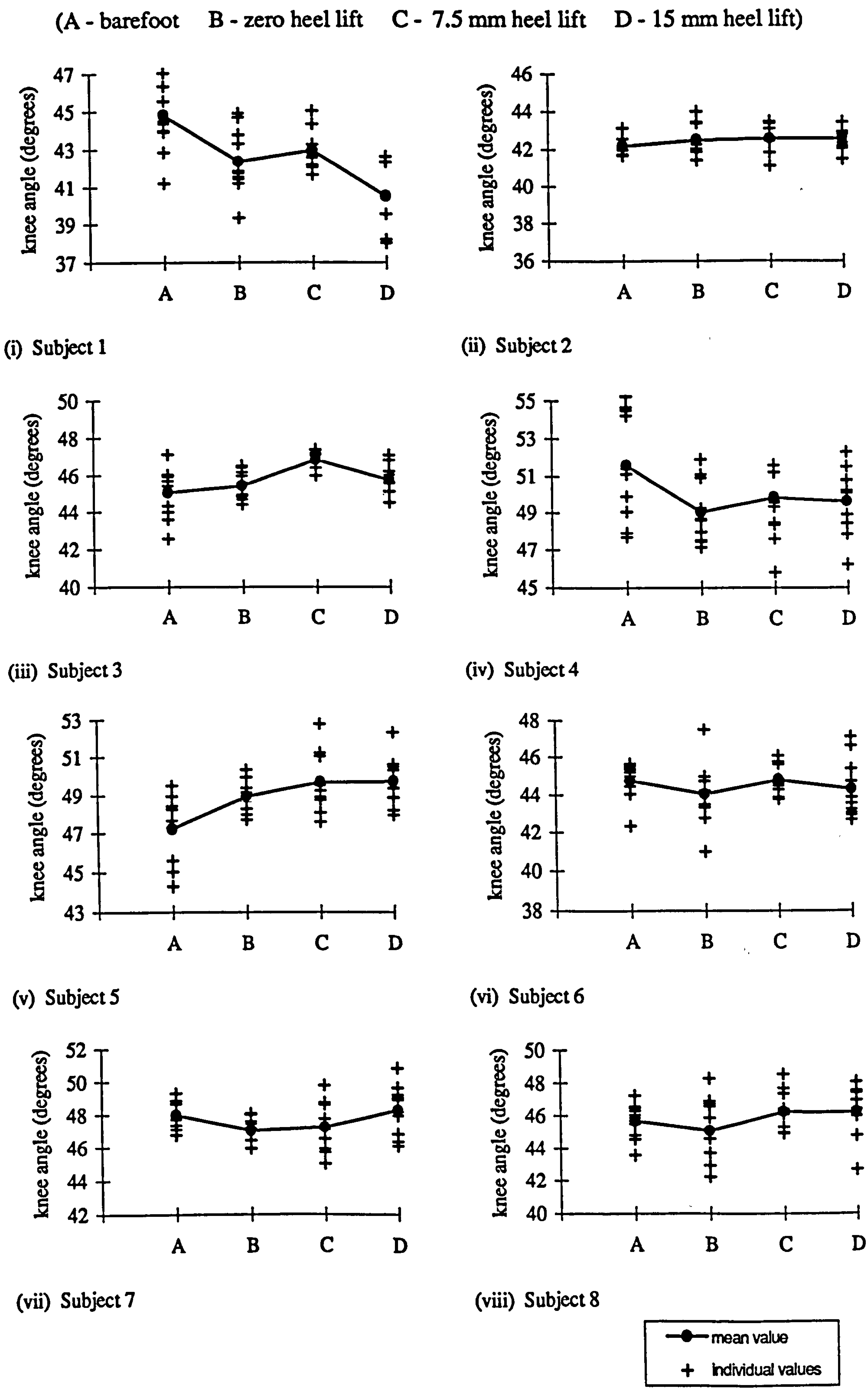


Figure 7.11 Maximum knee angle for each subject/condition combination

When lifts were attached to the rearfoot and the forefoot there was an increase in maximum ankle dorsi-flexion velocity for six of the eight subjects, with four of these increases being significant (Appendix I). For the two subjects demonstrating decreases in maximum dorsi-flexion velocity, the differences were not significant. Across the different heel lift conditions, varied responses were found across subjects. For the 7.5 mm heel lift condition compared with zero heel lift, five subjects demonstrated a decreased maximum ankle dorsi-flexion velocity, with three of these decreases being significant. The remaining three subjects showed an increased angular velocity, with one of these increases being significant. For the 15 mm heel lift condition compared with zero heel lift, three of the eight subjects showed a decrease in maximum dorsi-flexion velocity, with all three of these decreases being significant. Four subjects showed an increased dorsi-flexion velocity, with only one of these increases being significant. One subject demonstrated no change in this variable for the 15 mm heel lift condition compared with zero heel lift.

The attachment of lifts to the rearfoot and the forefoot resulted in a decrease in the maximum knee flexion velocity for six of the eight subjects (Appendix I). This decrease was only significant for one subject. Generally, only small changes in maximum knee flexion velocity were observed for both heel lift conditions compared with zero heel lift. Both increases and decreases were observed, with the only significant difference detected being an increase for the 15 mm heel lift condition compared with zero heel lift for one of the subjects. Thus, the evidence of the present study indicates that the raising of the heel relative to the forefoot has little detectable influence on maximum knee flexion velocity.

Joint angles at maximum Achilles tendon force

Ankle and knee joint angles at the time of maximum Achilles tendon force are provided in Table 7.10 and Table 7.11, respectively. As the magnitude of maximum Achilles tendon force was increased for the zero heel lift condition compared with the barefoot condition, ankle angle at this time was increased for five subjects. The remaining three subjects showed negligible variation in ankle angle across these two conditions. For most subjects, a decrease in knee angle at this time was observed. Thus, there was a trend for increased ankle dorsi-flexion and increased knee extension with increased maximum Achilles tendon force, for comparisons between these two conditions.

Compared with zero heel lift, both of the increased heel lift conditions caused a decrease in ankle angle at maximum Achilles tendon force for all subjects. This indicated a decrease in dorsi-flexion of the ankle joint at this time. Both increases and decreases in knee angle were demonstrated across the group of subjects when heel lift was introduced. For all but one of the subjects, there was a decrease in ankle angle at maximum Achilles tendon force for the 15 mm heel lift condition compared with the 7.5 mm heel lift condition. A trend has therefore been demonstrated for increased heel lift to cause a decrease in the dorsi-flexion of the ankle joint at the time of maximum Achilles tendon force. This was in contrast with a variable maximum Achilles tendon force response (Table 6.10). No trends were observed in knee angle at the time of maximum Achilles tendon force.

Table 7.10 Ankle angle (degrees) at the time of maximum Achilles tendon force (A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

	A	B	C	D
Subject 1	94.5	94.4	91.8	90.6
Subject 2	93.8	95.6	92.3	92.9
Subject 3	97.3	100.1	97.9	94.5
Subject 4	97.4	99.8	94.5	92.5
Subject 5	91.9	93.8	93.1	91.5
Subject 6	89.5	91.4	89.3	87.9
Subject 7	97.6	97.5	95.8	93.5
Subject 8	99.6	99.6	97.2	98.2

Table 7.11 Knee angle (degrees) at the time of maximum Achilles tendon force (A = barefoot, B = zero heel lift, C = 7.5 mm heel lift, D = 15 mm heel lift)

	A	B	C	D
Subject 1	36.9	32.6	35.3	32.3
Subject 2	37.0	37.1	36.6	36.8
Subject 3	41.2	41.2	43.3	42.7
Subject 4	51.1	48.0	48.0	48.1
Subject 5	44.4	41.0	47.0	39.8
Subject 6	40.2	39.1	39.7	39.5
Subject 7	43.1	42.5	42.2	43.2
Subject 8	41.7	42.2	41.0	42.2

(ii) Frontal Plane Movement

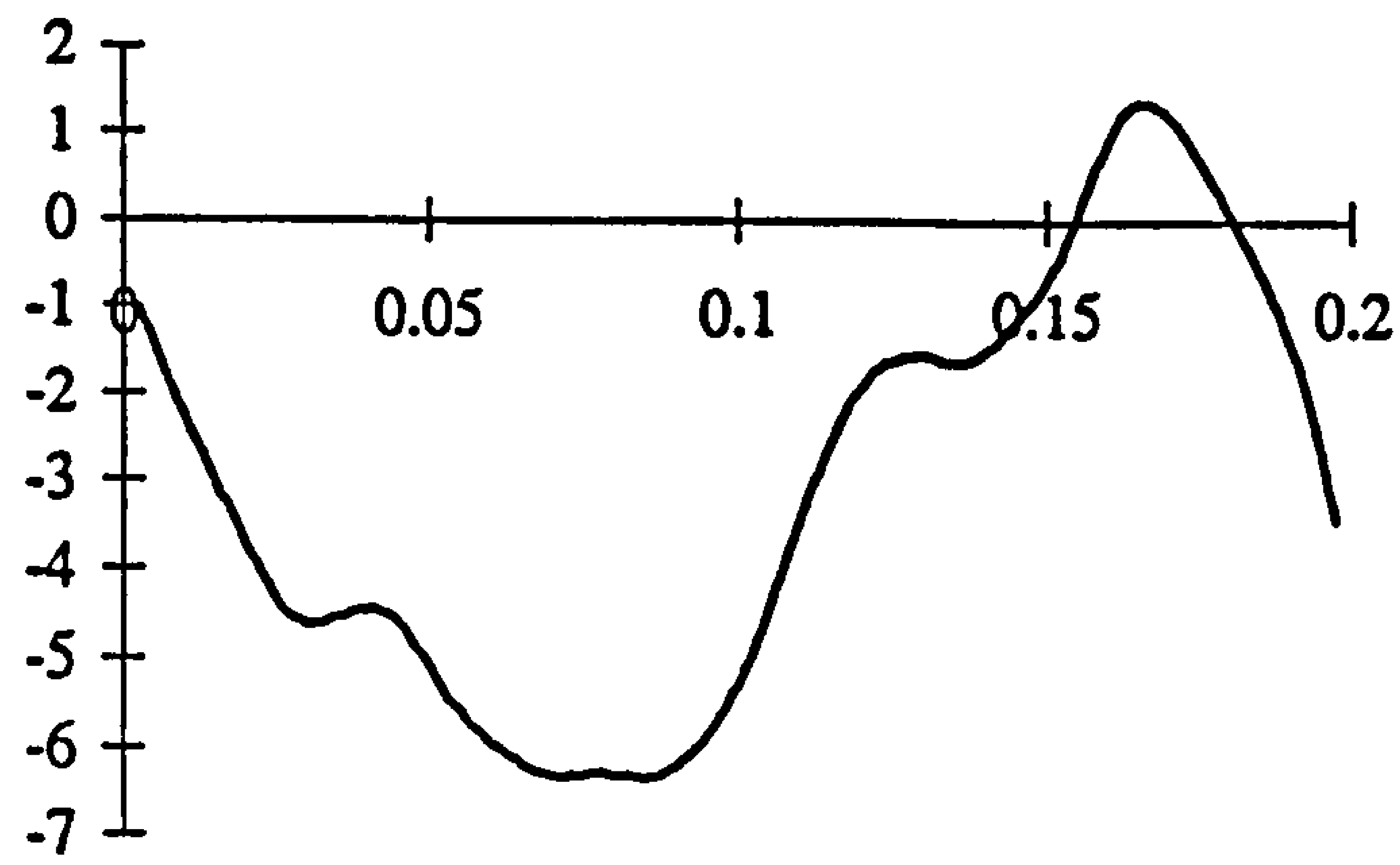
A sensitivity analysis revealed that, for a maximum rearfoot angle of 10 degrees, and a typical distance of 30 mm between the two calcaneus markers, a maximum error of 1.5 degrees occurred in the calcaneal angle (γ, Figure 7.2). For a lower leg angle of 3 degrees (ϕ, Figure 7.2), and a typical distance between Achilles tendon markers of 60 mm, a maximum error of 0.7 degrees was found to occur. Thus, for typical orientations and marker locations for the present study, an error of up to 2.2 degrees in calculated rearfoot angles may occur due to random error in marker coordinates.

Typical rearfoot angle time histories are illustrated in Figure 7.12. For some trials, the close proximity of the calcaneus markers resulted in these markers interfering with each other and sometimes being recorded as one single marker. In particular, Subject 3 rearfoot angle data showed unrealistic variation across trials, with angles being obtained that were clearly not possible. Thus, the rearfoot angle data for this subject were discarded. Rearfoot angles obtained for the remaining seven subjects are provided in Table 7.12. As illustrated by the standard deviation values, for several subjects there was a large amount of variation in rearfoot angle measurements obtained across trials for each condition.

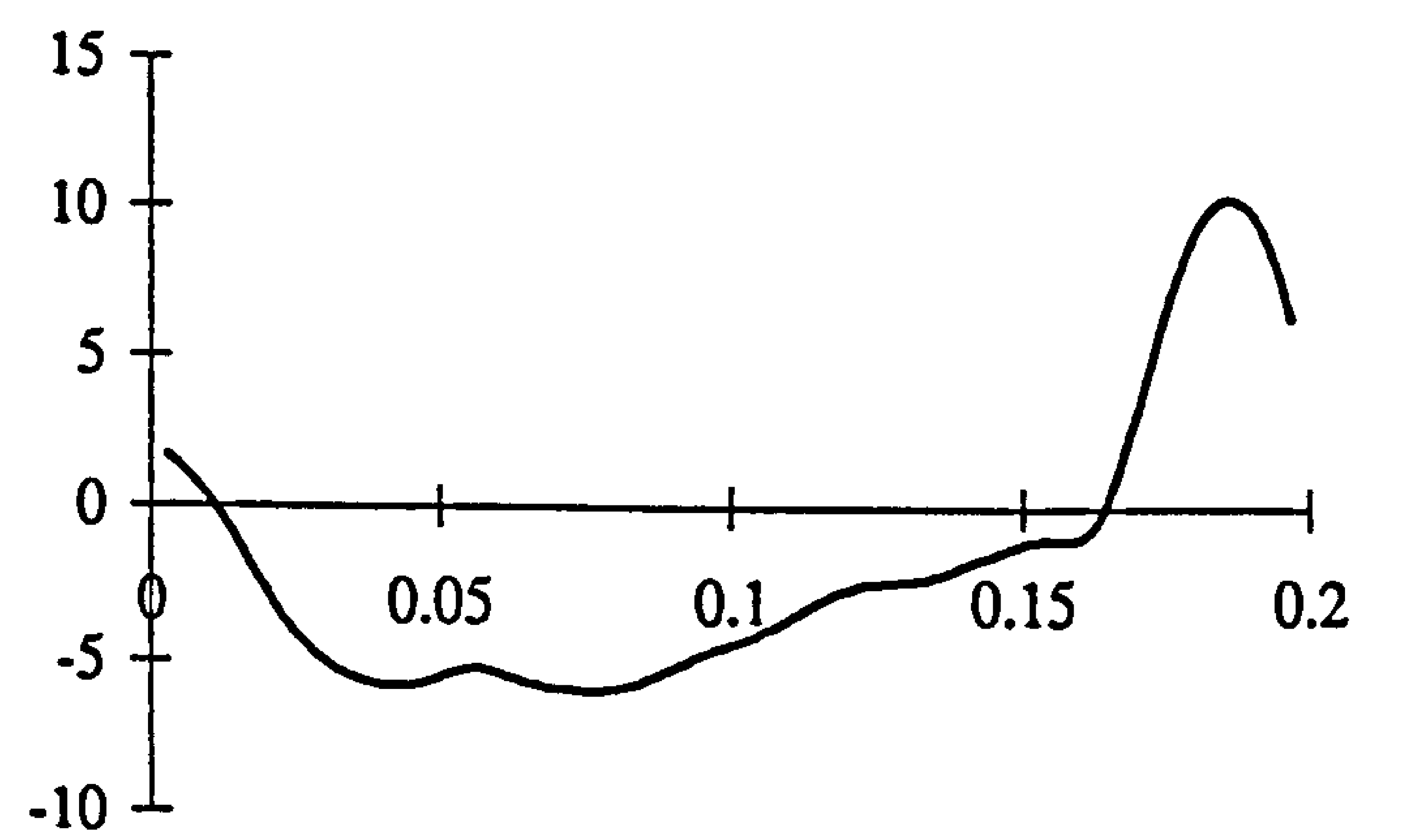
The attachment of lifts to the rearfoot and the forefoot resulted in an increase in the peak rearfoot angle for five of the seven subjects, although none of these increases were found to be significant. The two remaining subjects showed a decreased angle, with one of these decreases being significant. For the 7.5 mm heel lift condition compared with zero heel lift, there was an increase in the peak rearfoot angle for five of the seven subjects, with three of these increases being significant. For the two remaining subjects the peak rearfoot angle was decreased, with one of these decreases being significant. For the 15 mm heel lift condition compared with zero heel lift, there was an increase in peak rearfoot angle for four subjects, and a decrease for the remaining three subjects. One of the increases was significant, whereas all three of the decreases were significant. Varied maximum rearfoot angle response to heel lift variation has therefore been demonstrated across subjects.

The most reliable data, represented using standard deviation as a percentage of barefoot angle, were demonstrated for Subject 6 and Subject 7. These two subjects showed contrasting behaviour across heel lift conditions, as illustrated in Figure 7.13. Subject 6 demonstrated an increase in peak rearfoot angle with the attachment of lifts to the rearfoot and the forefoot compared with the barefoot conditions, whereas Subject 7 showed a decrease. Subject 6 showed an approximately linear decrease in rearfoot angle with increased heel lift, whereas Subject 7 showed an approximately linear increase. The changes observed with heel lift for these two subjects were found to be significant.

Maximum variations in calcaneal and lower leg angles of 4.1 degrees and 3.6 degrees respectively for Subject 6, and 0.9 degrees and 3.2 degrees respectively for Subject 7, occurred across conditions (Appendix I).



(i) Subject 4



(ii) Subject 7

Figure 7.12 Typical rearfoot angle time histories

In general, an increased heel lift delayed the time of occurrence of the maximum rearfoot angle (Appendix I). For the two subjects demonstrating a reduced rearfoot angle with heel lift, the timing of peak knee and rearfoot angles were compared. Subject 6 and Subject 8 demonstrated a significant reduction in maximum rearfoot angle with increased heel lift. Examination of the relative timing of the maximum rearfoot angle and the maximum knee flexion angle for these two subjects revealed that Subject 6 showed a later occurrence of maximum rearfoot angle relative to knee angle with increased heel lift, whereas Subject 8 showed similar timings of these two variables across conditions.

Table 7.12 Peak rearfoot angle (degrees) for each condition

	A	B	C	D
Subject 1*	-6.9 (2.1)	-7.1 (1.3)	-10.1 (1.6)	8.2 (5.9)
Subject 2	-1.6 (2.3)	-11.8 (4.2)	-12.2 (3.1)	-16.6 (3.1)
Subject 3				
Subject 4	-4.3 (2.6)	-4.5 (3.4)	-6.7 (5.1)	-6.6 (4.0)
Subject 5*	-22.1 (8.1)	-19.6 (3.7)	-34.3 (3.0)	-20.5 (3.4)
Subject 6*	-7.1 (0.7)	-8.1 (1.0)	-6.0 (0.6)	-5.2 (1.1)
Subject 7*	-4.3 (0.2)	-1.3 (1.1)	-3.3 (1.6)	-4.3 (0.4)
Subject 8*	-4.5 (2.4)	-5.9 (6.0)	-2.6 (3.0)	2.3 (6.6)

*$p<0.05$ Subject 1: B versus C; B versus D; C versus D
Subject 2: none
Subject 3: none
Subject 4: none
Subject 5: B versus C; C versus D
Subject 6: B versus C; B versus D
Subject 7: A versus B; B versus C; B versus D
Subject 8: B versus D

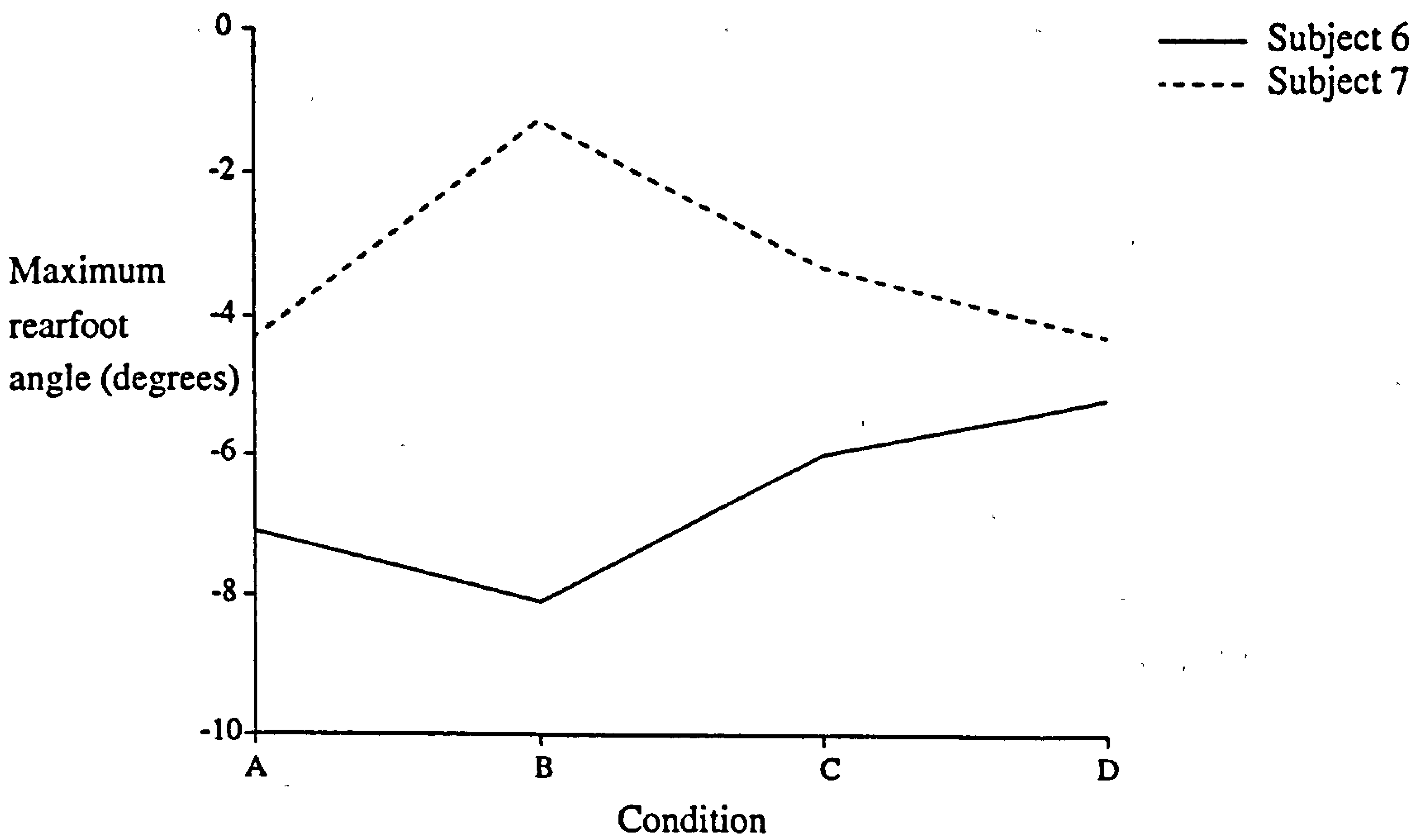


Figure 7.13 Changes in maximum rearfoot angle with heel lift for Subject 6 and Subject 7

7.4 Discussion

(i) Evaluation of Methods

It has been assumed that movement about the ankle, knee and hip joints occurs predominantly in the sagittal plane and that, for each subject, violations of this assumption have a systematic influence on calculated joint angles, and therefore do not influence comparisons between conditions. The use of this assumption has been supported by the finding that differences in the amount of movement out of the sagittal plane across conditions have a negligible influence on results for each subject. For a typical set of standing calibration data, a five degree medial-lateral movement out of the sagittal plane was found to cause only a 0.1 degree error in estimated foot angle. This magnitude of error is smaller than the level of precision attained in sagittal plane angles, and thus can be considered to be negligible. A five degree ab/aduction of the foot is typical for the stance phase of running, as demonstrated by Holden et al. (1986), who found that five subjects running at the same speed as used in the present study showed a mean change in stance phase foot ab/adduction angle across subjects of 4.4 degrees. The absolute ab/aduction value is not of concern, since joint angles have been compared across conditions using single subject analyses. The findings of the present study regarding the negligible influence of movement out of the sagittal plane on measured angles supports the findings of Arebald et al. (1990) and Soutas-Little et al. (1987), who found that two-dimesional sagittal plane data corresponded well with three-dimensional results.

For the two subjects considered in detail for rearfoot movement analysis, a maximum variation in calcaneal angle across conditions of less than five degrees was found. A five degree frontal plane rotation of the foot segment was found to result in less than 0.1 degrees variation in sagittal plane foot angle. The demonstration that there is negligible error in sagittal plane joint angles when typical movement occurs in the transverse and frontal planes has supported the use of two-dimensional analyses.

(ii) Standing Calibration

Observation of the standing calibration data across subjects has revealed that the change in foot angle magnitude across conditions was variable across subjects. Despite the use of the same lifts for each subject, the absolute amount the heel was raised was dependent on the individual. This highlights the importance of using a single subject approach. For the zero heel lift condition, lifts of identical thickness were attached to the rearfoot and the forefoot, indicating that no change in foot angle would be expected when comparing these two conditions. The differing response of subjects to the raising of the entire foot may be the result of differing distributions of pressure across the plantar surface of the foot. Although a firm lift material has been used, this may be deformed by a small amount. The degree of deformation of the rearfoot and forefoot lift was dependent on the distribution of pressure in the anterior-posterior direction when standing.

As the foot is a deformable segment comprising numerous complex articulations, it was expected that the exact change in foot angle due to the introduction of heel lift would not be predictable, even during standing. Since the lifts did not span the total length of the foot, it was not expected that the inclination of the foot would be increased by an amount corresponding to the taper of the lift. Calculation of the expected angle change for Subject 7 for a 7.5 mm heel lift using heel to MTP distance to represent foot length, and the assumption that the foot and the lift were rigid, provided a theoretical difference of two degrees, compared with the measured 2.5 degrees. Measurement errors may have contributed to the observed discrepancy of 0.5 degree, but could not alone account for a difference of this magnitude. Inconsistency in the position of attachment of the lift to the plantar surface of the foot may have resulted in the heel not being raised by the expected amount. For a rigid foot segment and heel lift, the change in foot angle when moving from the zero heel lift to the 7.5 mm heel lift would be comparable with the change when moving from the 7.5 mm heel lift to the 15 mm heel lift. Also the increase in foot angle when lifts were introduced would be greater for those subjects with smaller heel to MTP distances than for those with larger distances. The standing angles measured in this study demonstrate that the foot did not behave as a rigid segment. Factors contributing to the differing responses to heel lift across subjects and conditions may therefore have included the lift being deformed by the subject, the rigid foot assumption not being appropriate, and the heel not being lifted by the amount stated.

Direct comparison of joint angle magnitudes between subjects were not made, and thus standardised marker placement and calibration of joint angles, as suggested by Clarke et al., (1983b), were not necessary. It has been assumed that the relaxed standing position for each condition is of relevance when considering the response of an individual to a heel lift intervention. No constraints were placed on the subject when performing running trials, and thus comparison with relaxed 'natural' standing stance was considered to be more appropriate than comparison with a standardised position.

(iii) Impact Angles

Clear trends have been demonstrated in impact angle response to heel lift manipulation across subjects. An adaptation has been demonstrated whereby the ankle angle adopted at impact was less for increased heel lift conditions, causing reduced dorsi-flexion of the ankle joint at this time. The possibility of kinematic adjustments prior to ground contact has been highlighted by Bobbert et al. (1992). These authors described the necessity of selection of initial conditions before ground contact to allow control of the impact, since there is inadequate time to respond to an impact once it has occurred. The possibilities of selection of geometry of the body, and muscular activation levels prior to touchdown were described. Clarke et al., (1983a) provided experimental evidence to support this suggestion, with changes in ankle angle at impact being demonstrated when the amount of shock absorption provided by a shoe was varied. Frederick (1986) described changes in maximum knee flexion velocity with varied shock absorption properties of a shoe. In the present study, further evidence has been found to indicate that kinematic adaptations to footwear occur prior to ground contact. It has been demonstrated that adaptations occur in response to geometrical differences, in this case heel lift, as well as changes in material properties.

Bobbert et al. (1992) described how it would be of interest to know by what criteria the initial conditions are selected. The evidence provided in the present study indicates that a mechanism was operating by which the range of ankle joint movement from impact to maximum ankle angle remained reasonably constant across conditions. There was a trend for a reduced ankle dorsi-flexion for increased heel lift heel lift conditions. If a constant impact ankle angle had been maintained across conditions, then the change in ankle angle from impact to peak dorsi-flexion would have been greater for the condition with heel lift. It is therefore suggested that the trend for reduced ankle dorsi-flexion at impact for the increased heel lift conditions was an adaptation contributing to a reduction in the required ankle angle change. This reduction contributed to a decrease in the rate of change of ankle angle, compared with the rate of change if impact ankle angle had remained constant across conditions. Since rate of change, and thus loading rate of soft tissue around the ankle joint, has been associated with injury occurrence, the change in ankle impact ankle may have been a compensatory adaptation to inadequate heel lift.

For the majority of subjects, the difference calculated between foot angles during standing and angles at impact indicated a rearfoot strike, supporting the GRF evidence and visual observation of ground contact. For a small number of subject/condition combinations the foot angle difference indicated that a forefoot strike occurred. However, this was not supported by the GRF data which were characteristic of a rearfoot strike. This may be explained by the possible compression of the heel lift during standing, resulting in standing angles smaller than if solid heel lifts had been used. Impact angle definitions resulted in angles being calculated using data prior to ground contact, and thus no compression of lifts will have occurred when these angles were calculated. A small amount of compression of the lifts during standing calibration may therefore account for the apparent forefoot strike, when in fact the heel of the subject contacted the ground before the forefoot.

(iv) Angles during Stance

Foot, ankle, knee and thigh angle time histories presented for the subjects used in the present study were similar to those described in the literature (Milliron and Cavanagh, 1990). Angle plots for each joint were characterised using selected parameters, allowing comparison across conditions for each individual subject.

Direct comparison of impact angles and peak angles across subjects and studies was limited by differences in marker placement. Ankle angles obtained in the present study were generally smaller than those provided in the literature (presented in Chapter 2), due to the use of the ankle marker as the proximal end of the foot segment in the present study, as opposed to a marker on the heel. Knee impact and peak angles obtained in the present study demonstrated magnitudes that were comparable with those presented in the literature (Chapter 2).

Direct comparisons can be made of the total joint movement from impact to peak angle, since these should not have been influenced by the marker placement. Total ankle joint movement ranged from 16.6 degrees to 32.0 degrees in the present study, compared with 23.3 degrees (Nilsson et al., 1985) and 41.0 degrees (Williams, 1985) presented in the literature. Total knee joint movement in the present study ranged from 23.4 degrees to 28.8 degrees, comparable with literature values ranging from 15 degrees (Sinning and Forsyth, 1970) to 32.6 degrees (Williams, 1980).

Despite the rearfoot strike demonstrated for all subjects in the present study, the foot was generally found to be close to flat on impact with the ground ground, resulting in only a small amount of rotation immediately following impact. This is consistent with descriptions in the literature (Milliron and Cavanagh, 1990). In general, the foot rotated in a direction contributing to increased ankle plantar-flexion throughout stance, with a period of 'flat foot' during the first 50% of ground contact. Due to flexion of the knee joint and rotation of the lower leg immediately following impact, dorsi-flexion of the ankle joint was found to occur up to the middle of the stance phase, as has been described by Milliron and Cavanagh (1990). Beyond this time, the rotation of the foot combined with knee extension resulted in plantar-flexion of the ankle joint.

The 'flat foot' angle defined in the present study was considered to be of interest as it represented the orientation of the foot relative to the ground during midstance when the foot was flat and in a position determined predominantly by the presence of the heel lifts. The comparison of foot angles during this 'flat foot' stage of stance with standing calibration angles was made to investigate whether standing angles could feasibly be used to predict stance angles. If these angles clearly differed, then the common use of standing angles by practitioners to indicate desired angle changes during the stance phase of running may not be appropriate. As would logically be expected, the trend of 'flat foot' angle change with increased heel lift was the same as the trend demonstrated for standing angles. This supports the use of relative sagittal plane standing angles for different footwear conditions for prediction of angle differences during midstance, here termed the 'flat foot' angle. However, there is no evidence provided in the present study to support the use of absolute foot angles during standing for prediction of absolute midstance angles during running.

Heel lifts were found to generally have a consistent influence on lower extremity sagittal plane kinematics. Generalisations concerning the kinematic response of rearfoot strikers to heel lift interventions were therefore possible. The increased 'flat foot' angle and subsequent decrease in peak ankle angle found in the present study with increased heel lift was consistent across subjects. Since the foot has been shown to rotate to contribute to increased plantar-flexion of the ankle joint throughout the majority of the stance phase, then the increased 'flat foot' angle for increased heel lift appears to indicate an increase in the total rotation of the foot from initial ground contact to 'flat foot'. In order to attain this increased angle within the same amount of time for the increased heel lift conditions, it would have been necessary for the average foot angular velocity to be increased. This appears to have been averted by a combination of an increased time from ground impact to peak ankle angle for the increased heel lift conditions compared with zero heel lift, and an increase in foot inclination at impact. This is consistent with the earlier suggestion that there was a compensatory response to heel lift whereby the total range of movement was reasonably constant across conditions.

Studies of sagittal plane kinematic responses to footwear interventions described in the literature have generally focused on the responses to changes in shoe cushioning, as opposed to shoe geometry variations (Frederick, 1983; Clarke et al., 1983a). It is, therefore, not possible to directly compare the trends identified in sagittal plane kinematics with heel lift variation in the present study, with those described in the literature. Frederick (1983) found that there was an increased knee flexion velocity with decreased shock absorption of running shoes, suggesting that this result indicated a compensation resulting from reduced cushioning provided by the shoe. In the present study, no trend in maximum knee flexion velocity was demonstrated across conditions. This finding supports the use of the firm heel lifts described in the present study for providing a controlled increase in heel lift, whilst not changing the amount of shock absorption by a noticeable amount.

(v) Relationship between Achilles Tendon Moment Arm and Ankle Angle

Achilles tendon forces estimated in the present study were dependent on ankle moment and Achilles tendon moment arm. With increased heel lift, both increases and decreases in Achilles tendon moment arm length were found with the decreased ankle dorsi-flexion observed across subjects. This finding is in contrast to literature suggestions that there is a linear increase in Achilles tendon moment arm length with increased ankle plantar-flexion (Grieve et al., 1978; Bobbert et al., 1986). The significance of the location of the Achilles tendon insertion relative to the ankle joint centre has been described in Chapter 4 of this research. If a rigid link system is assumed, then whilst the Achilles tendon insertion is lower than the ankle joint centre, decreased ankle angle (ankle plantar-flexion) will result in an increase in the moment arm of the Achilles tendon. If ankle plantar-flexion continues beyond a position such that the Achilles tendon insertion moves above the ankle joint centre, then a point is reached at which the Achilles tendon moment arm length will start to decrease.

A heel lift was expected to raise the height of the Achilles tendon insertion relative to the ankle joint centre. For five of the subjects in the present study, at the time of maximum Achilles tendon force occurrence, the moment arm was found to increase for the 7.5 mm heel lift condition compared with zero heel lift and then to decrease for the 15 mm heel lift. A possible explanation for this is the higher position of the Achilles tendon insertion for the condition with the greater heel lift, as supported by the increased inclination of the foot to the horizontal at maximum Achilles tendon force. One subject demonstrated an increase in the length of the Achilles tendon moment arm with increased heel lift, which can also be explained using the described mechanism, with this subject having the Achilles tendon insertion lower than the ankle joint centre for all conditions. One of the remaining subjects showed negligible variation in Achilles tendon moment arm, which may have been the result of small differences in ankle and knee angles across conditions for this subject. Finally, one subject was found to have a decreased Achilles tendon moment arm when heel lift was increased from zero to 7.5 mm, and an increased moment arm when heel lift was further increased to 15 mm. This variation in Achilles tendon moment arm could not be explained using the mechanism described. A possible explanation may be that changes in knee angle across conditions contributed to the observed variation in Achilles tendon moment arm. Alternatively, the rigid link model which has been used to illustrate the mechanism by which Achilles tendon moment arm changes with ankle and knee angle may not be appropriate.

It has therefore been demonstrated that, for seven of the eight subjects employed in the present study, the observed variation in Achilles tendon moment arm with ankle angle can be explained using a rigid link model. The observed variation in Achilles tendon moment arm during ground contact for subjects in the present study was similar to that found by Burdett (1982) and Spoor et al., (1990).

(vi) Relationship between Kinematics and Achilles Tendon Loading

If there is a linear relationship between Achilles tendon stress and strain in running, as has been indicated by Ker et al. (1988), then it is expected that maximum Achilles tendon force and maximum stretch of the Achilles tendon will occur simultaneously. Even if the relationship is not linear, the maximum values would be expected to occur at the same time since, unless ultimate values have been reached, an increase in stress is always accompanied by an increase in strain (Abrahams, 1967).. As discussed previously, the amount of stretch of the triceps surae group is influenced primarily by ankle angle. It has been concluded from the relevant literature that the increase in overall length of the muscle-tendon complex of the triceps suare group in running will be contributed to primarily by the tendon (Herzog and Loitz, 1995; Caldwell, 1995). Ankle angle has therefore been used to indicate changes in Achilles tendon strain. On average, for the barefoot condition, the maximum Achilles tendon forces have been found to occur 2.75 ms later than the peak ankle angles, corresponding to a difference of approximately 1% of total stance time. The similar timing of these variables supports the suggestion that maximum stress and maximum strain of the Achilles tendon occur simultaneously during running. In further support of this suggestion, the time from impact to maximum ankle angle and maximum Achilles tendon force occurrence showed the same trends across conditions, with a consistently shorter time for the zero heel lift condition compared with barefoot, and consistently greater time for the increased heel lift conditions compared with zero heel lift.

Trends in magnitudes of peak ankle dorsi-flexion angle and maximum Achilles tendon force were expected to be consistent across conditions if Achilles tendon strain values had been predicted adequately using ankle angles. Comparison of the zero heel lift condition with the barefoot condition supported this suggestion, with an increased ankle dorsi-flexion and increased maximum Achilles tendon force observed for all subjects. When an increased heel lift was introduced, the general trend observed in the present study was for a reduction in the amount of dorsi-flexion of the ankle joint, suggesting that there was a reduction in the stretch of the Achilles tendon, and thus a reduction in Achilles tendon force. For all but one subject, a decrease or negligible change in maximum Achilles tendon force occurred for the 15 mm heel lift compared with zero heel lift, in agreement with the change indicated by the observed ankle angle changes. However, the influence of the 7.5 mm heel lift on maximum Achilles tendon force was varied across subjects, despite the ankle angle changes being consistent.

The possible influence of changes in knee angle on maximum Achilles tendon strain was also considered. Although no trends in magnitude of knee angle were observed across conditions, similar trends were observed across subjects in the time taken to attain maximum knee angle. A later occurrence of maximum knee angle resulted in an increase in the knee angle at the time of maximum ankle dorsi-flexion with increased heel lift, indicating a greater knee flexion. This was consistent with the observed trend for an increased knee angle at the time of maximum Achilles tendon force for an increased heel lift. The trends in timing of maximum knee angle observed in the present study contributed to length changes of the triceps surae group in the same direction as those indicated by the ankle angle changes.

The discrepancies observed for some subjects between changes in estimated maximum Achilles tendon force and changes indicated from joint angles illustrates the limitations of using joint angles alone to indicate changes in loading of the tendon. The assumption in the present study that a change in the overall length of the triceps-surae group indicated a proportional change in the length of the Achilles tendon, was made based on findings in the literature regarding the potential contribution of the Achilles tendon to overall length changes (van Ingen Schenau, 1984; Bobbert et al., 1986; Caldwell, 1995). The conclusions of these authors have been based on mathematical models of the muscle-tendon complex, using input parameters which have been obtained from measurements of isolated muscle and tendon samples. Herzog and Loitz (1995) described how experimental evidence regarding the relative contributions to overall length of the muscle-tendon is outstanding. Zajac (1989) described how the properties of a muscle-tendon complex can be defined using the ratio of tendon slack length (length at zero force) to muscle fibre length. This ratio was shown to determine the force-length relation of the muscle-tendon complex, with a higher ratio indicating a more compliant complex. The human triceps surae muscle-tendon complex was described as being compliant. The isometric force-length relation for a compliant muscle-tendon complex is illustrated in Figure 7.14. Zajac (1989) described how compliant tendons can cause the overall length of the muscle-tendon complex to lengthen while the associated fibres shorten, or shorten while the muscle fibres lengthen, due to the effects of tendon length and velocity. For a muscle-tendon complex length providing a ratio of less than *a* (Figure 7.14), an increase in length is associated with an increase in force, and a decrease in length is associated with a decrease in force. For a ratio of between *a* and *b* (Figure 7.14), the opposite relationship between force and length is demonstrated. Therefore, an increase in heel lift which reduces the maximum overall length of the muscle-tendon complex and has a length which is contained within *a* and *b* length ratios, will correspond to a decrease in force. During eccentric contractions, as experienced by the triceps surae group during the generation of Achilles tendon maximum force, the force is not influenced greatly by changes in velocity (Gregor, 1993). The application of the isometric relationship illustrated in Figure 7.14 for explanation of observations in the present study was therefore justified. Methods have recently been developed for the measurement of dynamic relative length changes during locomotion using ultrasound techniques. Using these methods, Hoffer et al. (1989) found that the length changes of the muscle and tendon of the cat gastrocnemius in walking were not in phase.

It appears possible that the length of the muscle-tendon complex in running may be reduced whilst the Achilles tendon does not change in length, or even experiences an increase in length. Thus, the reduction in muscle-tendon overall length indicated by an increased ankle dorsi-flexion that was observed for some subjects with increased heel lift, did not necessarily indicate a reduction in Achilles tendon strain. This suggested mechanism may explain how the Achilles tendon maximum force was increased for some subjects, whilst the overall length change of the triceps surae group had apparently been reduced. The described mechanism can explain the apparent discrepancy in strain estimated using force and kinematic data, but direct measurement is required to support this suggestion. The use of the techniques described by Hoffer et al. (1989) in the analysis of human running would allow the investigation of the relative contributions to overall length changes of the triceps surae muscle-tendon complex. The explanation provided in the present study for increased Achilles tendon force with decreased length of the muscle-tendon complex could therefore be investigated.

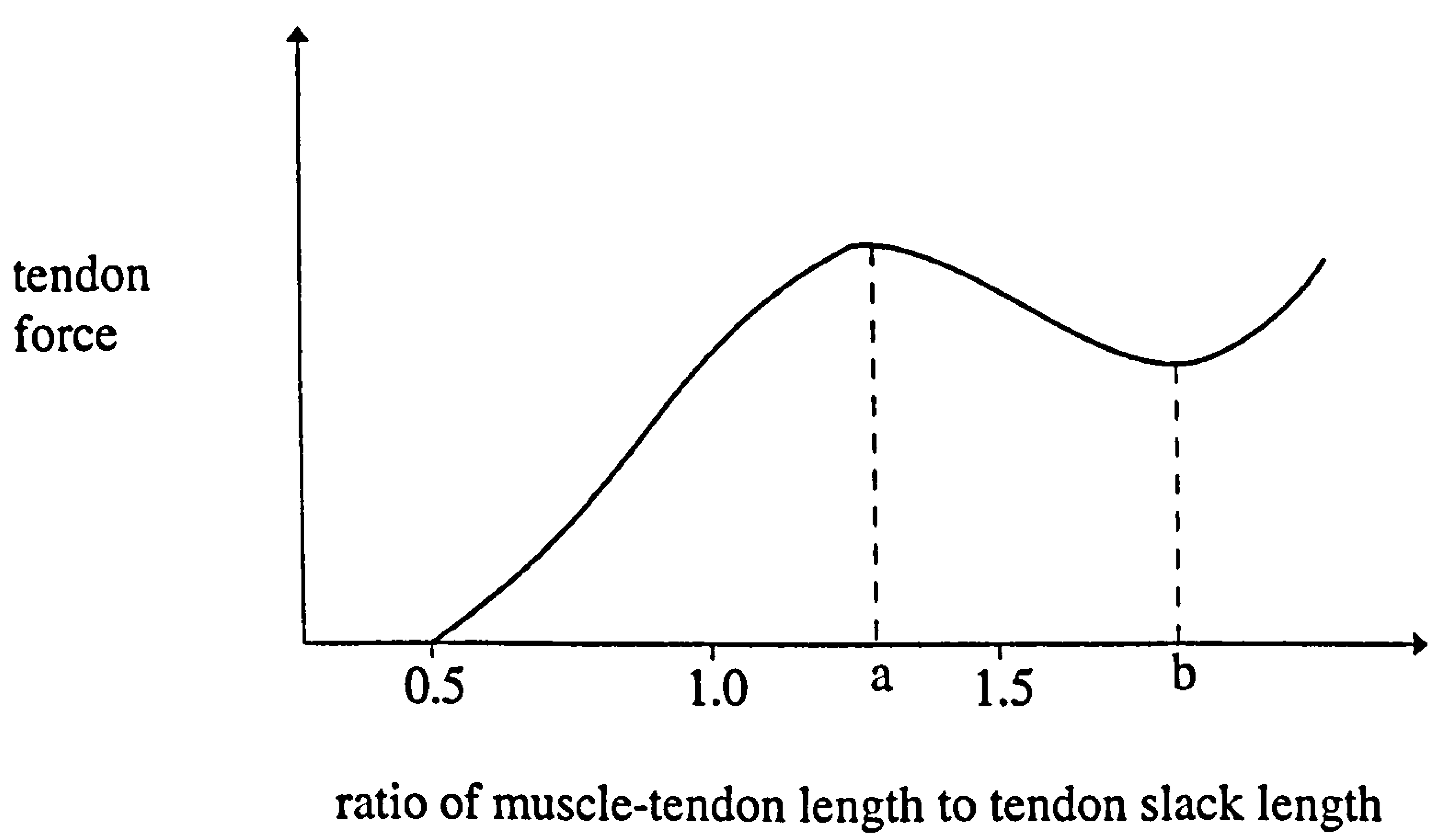


Figure 7.14 Isometric force-length relation of a compliant muscle-tendon complex

The exact change in length of the Achilles tendon was not of concern in the present study, since the aim was to compare across heel lift conditions rather than to identify the amount of tendon strain. Methods have been presented in the literature for estimation of the overall length of the muscle-tendon complex of the triceps surae using ankle and knee joint angles (Grieve et al., 1978). It was not deemed necessary to estimate this length in the present study as the direction of length change was considered to be adequately represented by the observed angle differences, and would not be changed by the the estimation of absolute lengths.

Achilles tendon strain values can also be estimated using Achilles tendon stresses and the assumption that the relationship between Achilles tendon stress and strain is linear (Ker et al., 1988). Achilles tendon stiffness values provided in the literature have been in the region of 250 $kN.m^{-1}$. Thus a typical maximum force of 5400 N obtained in the present study will stretch the tendon by 21.6 mm. This length is similar to the length change of 18 mm provided for running by Alexander and Bennet-Clark (1977). Ker et al. (1988) have provided a typical Achilles tendon length of 350 mm. Thus an increase in length of 21.6 mm corresponds to a strain of 6.2%.

The rate of loading of the tendon during running may influence the mechanical behaviour, resulting in a different response to a particular force than that recorded during isolated tests. In particular, it has been found by some investigators that the stiffness of tendon is increased at increased loading rates (Abrahams, 1967). This results in the possibility of increased stress in the tendon, without an increase in the amount of strain experienced. However, the majority of recent studies have indicated that loading rate of the Achilles tendon does not influence mechanical behaviour for the rates of loading expected during human running (Ker et al., 1988; Zajac, 1989).

Discrepancies have been identified in the present study between the trends of loading of the Achilles tendon predicted using joint angles, and those obtained by estimation of Achilles tendon forces. The investigation of apparent discrepancies has highlighted the possibility of the length change of the tendon not being predictable using changes in length of the muscle-tendon complex, due to differences in relative stiffness, and differences in the relative contribution of muscle fibres and tendon to overall length changes. It therefore appears that kinematic variations alone are not sufficient to provide an indication of the influence of heel lift interventions on Achilles tendon loading. A method for estimation of Achilles tendon force, as has been used in the present study, is required for an understanding of loading response. The force time history is sufficient for representation of the loading of the tendon, since strain can be predicted using estimated stress values. These results may be combined with kinematic data if investigation of the relative contributions of muscle and tendon to changes in the overall length of the muscle-tendon complex is required.

(vii) Rearfoot Movement

The general shape of the typical rearfoot angle time histories presented in the present study was similar to those illustrated in the literature (Clarke et al., 1983b). However, beyond approximately 70% of total stance, the rearfoot angle decreased in magnitude at a rate much greater than that usually demonstrated. This pattern, which was observed across all subjects, was attributed to the use of three-dimensional data to obtain frontal plane coordinates in the present study, in contrast with literature conventions which have involved the projection of markers onto a two-dimensional plane. Peak rearfoot angles obtained in the present study occurred within the first 50% of total stance, and thus were not influenced by the sudden change in calculated angle.

It is of interest that six of the seven subjects showed consistent patterns of either increased peak rearfoot angle for both heel lift conditions, or decreased peak rearfoot angle for both heel lift conditions. Four subjects showed increased peak angle for both heel lift conditions, with three of these subjects having at least one significant increase. Thus, the results of the present study have demonstrated that for some subjects there was an increase in the maximum rearfoot movement when the heel was raised relative to the forefoot. This is contrary to the suggestions in the literature that increased heel lift contributes to a decrease in the peak rearfoot angle (Stacoff and Kaelin, 1983; Bates et al., 1978).

The reduction of rearfoot movement has been suggested as a possible mechanism by which by the introduction of heel lifts may act to reduce the incidence of Achilles tendon injury. Bates et al. (1978) stated that, since dorsi-flexion is a component of pronation, the reduced dorsi-flexion of the ankle joint resulting from an increased heel lift, will reduce the amount of pronation. The finding in the present research that rearfoot movement may be increased with heel lift, despite a reduction in the amount of ankle dorsi-flexion, supports findings in the literature that dorsi- / plantar-flexion occurs almost exclusively at the ankle joint, and is thus relatively independent of rotations at the subtalar joint (Siegler et al., 1988). The subtalar joint axis has been demonstrated to vary across subjects, and throughout joint motion (Engsberg, 1987). It does not seem that surprising therefore that variations in the motion about this joint were found to occur across subjects in response to the introduction of a heel lift. It has been demonstrated in the present study that individual assessment of runners is necessary due to the varied response across subjects.

A suggested mechanism by which 'excessive' pronation has been claimed to contribute to injury of the Achilles tendon is by causing conflicting rotations at the proximal and distal ends of the tendon as a result of maximum knee flexion angle occurring earlier than maximum pronation angle (Clement et al., 1984; Smart et al., 1980). In the present study, trends have been demonstrated for both maximum knee flexion angle and maximum rearfoot angle to occur later in the stance phase. For the two subjects found to exhibit a decreased maximum rearfoot angle in the present study, a comparison of timing of these two variables indicated that one subject showed a later time of maximum rearfoot angle relative to knee angle with increased heel lift. This result contradicts the literature suggestion than a reduction in pronation will prevent a delay in the occurrence of this variable in relation to maximum knee flexion (Pagliano, 1987a). For the other subject, similar timing in the maximum knee flexion and rearfoot angles was demonstrated across conditions. Thus, no evidence has been provided in the present study to support the suggestion that a reduction in pronation will limit the conflicting rotations at the proximal and distal ends of the tendon. The finding that changes in conditions could result in the maximum rearfoot angle occurring prior to maximum knee flexion was consistent with the findings of Hamill et al. (1992). If differences in the timing of maximum knee flexion and maximum rearfoot angle result in conflicting rotations at either end of the tendon, then discrepancies in either direction may be related to Achilles tendon injury occurrence.

CHAPTER 8 SUMMARY AND GENERAL DISCUSSION

8.1 Summary of Studies

The purpose of the studies presented in this research has been to contribute to an increased knowledge and understanding of the loading of the Achilles tendon in running. In particular, the influence of heel lift manipulation on Achilles tendon loading in elite female middle distance runners has been investigated. In the first study, presented in Chapter 3, the main question addressed was whether heel lift manipulation could influence maximum Achilles tendon force in running. The subsequent study, described in Chapter 4, involved a controlled investigation of the influence of an isolated heel lift intervention on three runners demonstrating distinctly different running styles. The main question addressed was whether raising the heel with a firm heel lift could reduce maximum Achilles tendon force. In Chapter 5 the accuracy with which anatomical measurements had been obtained was investigated using magnetic resonance imaging techniques. The question addressed was whether the anatomical data required for the estimation of Achilles tendon forces could be obtained adequately using skin markers. In the study described in Chapter 6, the main question investigated was how did heel lift influence the maximum magnitude and loading rate of Achilles tendon force in rearfoot strikers. The use of eight runners all with a rearfoot ground strike provided an indication of the consistency of responses. The question addressed in Chapter 7 was whether lower extremity kinematics were influenced by heel lift intervention in rearfoot strikers. Both sagittal plane and frontal plane kinematics were analysed.

The aspects of Achilles tendon loading chosen for study were selected by reference to suggestions in the literature regarding the etiology of Achilles tendon injury. The maximum stress and strain experienced by the Achilles tendon were studied, following the association of the repeated loading of the tendon in running with the occurrence of this injury (eg. Archambault, 1995). Rearfoot movement was monitored following suggestions that Achilles tendon injury may be the result of high localised stresses caused by twisting of the tendon as a result of excessive rearfoot movement (eg. Smart et al., 1980). The peak magnitude and loading rate of vertical ground reaction force were determined due to the association of decreases in these variables with a reduced incidence of Achilles tendon injury (eg. MacLellan, 1984).

Heel lift manipulation was chosen for investigation following a review of the literature regarding the interventions that have been associated with a reduced incidence of Achilles tendon injury. It has been suggested that inadequate heel lift increases the force and stretch experienced by the Achilles tendon (Clement et al., 1984). It has also been suggested that an increase in heel lift may contribute to a decrease in localised stresses in the Achilles tendon by reducing the maximum amount of pronation (Bates et al., 1978). Clinical findings have demonstrated that Achilles tendon injuries can be treated successfully by the raising of the heel relative to the forefoot (MacLellan and Vyvyan, 1980; Grisogono, 1989). This intervention is prescribed frequently by practitioners, but the mechanism by which it is successful has not been documented.

The main aim of this research was therefore to investigate the influence of heel lift on selected aspects of Achilles tendon loading. Knowledge of the factors influenced by increased heel lift should provide an insight into the mechanism by which Achilles tendon injuries occur.

8.2 General Discussion and Future Research

As suggested by Reboussin and Morgan (1996), at each stage of this research, reassessment was made of the interaction between the scientific question, the experimental design, and the statistical analysis. For example, the choice of dependent variables monitored in each study was influenced by the findings of the previous investigation, effecting the main question that was addressed. Through the course of the research, the procedures for the manipulation of heel lift were refined to reduce the possible influence of extraneous variables. The number of running trials required and the level of statistical significance were varied in response to previous results, to ensure that an adequate statistical power was attained.

The main focus of this research has been the comparison of Achilles tendon loading across conditions. The 'comparison technique', described by Nigg and Bobbert (1990), has been successfully employed for the detection of the statistically significant differences in aspects of Achilles tendon loading with heel lift manipulation. The use of this approach has increased the available knowledge regarding the factors influencing Achilles tendon loading. Statistical significance has provided a measure of the confidence in observed differences in loading, but the clinical significance of these differences is not known. To facilitate the identification of clinically significant differences, knowledge is required on the absolute loading of the tendon.

In addition to making comparisons between conditions, the loading of the Achilles tendon relative to ultimate stress and strain values has been investigated in this research. Nigg and Bobbert (1990) highlighted the limitations of this 'cause-effect' approach, describing the errors in estimating internal loading and in obtaining mechanical properties of isolated tendon specimens. However, by quantifying the potential influence of errors on results, indications of the loading of the Achilles tendon in running relative to its ultimate strength have been obtained. The errors in measurements taken from isolated tendon samples have been acknowledged by consideration of the range of values presented in the literature. It has been found that the stress and strain experienced by the Achilles tendon in running are close to the maximum that this structure can sustain.

In general, tendon has been found to be stronger than the muscle to which it is attached (Nordin and Frankel, 1980; Ker et al., 1988). Ker et al. (1988) described how most tendons have a safety factor of at least four, that is the stress that the muscle can exert on the tendon is less than four times the tendon ultimate stress. The Achilles tendon has been found to experience high loads relative to other tendons, but not to have a markedly increased strength to reflect this requirement (Ker et al., 1988). The strength of a tendon is influenced by the size of its cross-sectional area. Ker et al. (1988) described the likelihood of metabolic cost influencing the size of a tendon, with tendons in the distal parts of the leg being as thin as possible so as to minimise the cost associated with locomotion. Additionally, the potential of the Achilles tendon to act as a spring during locomotion has been demonstrated (Alexander and Bennet-Clark, 1977; Ker et al., 1987). This function will be enhanced by an increased compliance, consistent with a relatively thin tendon (Shorten, 1987). The Achilles tendon therefore appears to have developed to be as thin as possible, minimising mass and facilitating extension of the tendon during locomotion. The result has been a tendon that functions close to its ultimate stress and strain.

The finding that the Achilles tendon operates close to its ultimate strength during running has implications regarding the occurrence of Achilles tendon injury. On a five mile run an individual will take approximately 5000 steps, resulting in each Achilles tendon being loaded 2500 times by an amount close to its ultimate strength. In general, this loading does not result in injury, and any microdamage appears to be repaired sufficiently for repeated runs to be performed, often twice daily, without an accumulation of damage. The development of accurate methods of quantifying the absolute loading of the tendon and the threshold above which injury may occur may facilitate the detection of clinically significant differences in loading. This approach could be employed in longitudinal studies in which the Achilles tendon injury status and loading are regularly monitored for the investigation of loading associated with injury occurrence.

For some subjects in the present research, the Achilles tendon strain predicted using estimated Achilles tendon force values behaved in a different manner to that indicated by joint angles. It was speculated that this may have been the result of varying contributions of tendon and muscle fibres to overall length changes of the triceps surae complex. The use of methods such as those employed by Hoffer et al. (1989), in which the lengths of both components of the muscle-tendon complex are continuously monitored, would allow the investigation of this suggestion. Since a consistent behaviour was observed in the overall length of the muscle-tendon complex with increased heel lift, heel lift may contribute to a reduction in Achilles tendon injury occurrence by reducing this overall length. Different contributions across subjects may reflect the ability of the Achilles tendon of an individual to sustain loads of a particular magnitude, relative to the strength of the associated muscle.

The analysis of rearfoot movement in the present study has demonstrated that heel lifts have a varying influence on rearfoot motion across subjects. It has been found that a reduction in dorsi-flexion does not necessarily indicate a reduction in the eversion of the subtalar joint. This has not been demonstrated previously and highlights the importance of individual assessment of runners when taking steps aimed at reducing pronation. The suggestions made by Bates et al. (1978) concerning a direct relationship between dorsi-flexion and eversion appear to have been made without consideration of the anatomical relationship between these joint movements.

Differing explanations have been provided in the literature regarding the mechanism by which excessive pronation may contribute to Achilles tendon injury occurrence. Clement et al. (1984) described a 'whipping action' or 'bowstring' effect observed in the Achilles tendon using slow motion, high speed cinematography. This was attributed to excessive pronation of the subtalar joint acting to move the Achilles tendon medially during midstance, resulting in microtears in the tendon and subsequent injury. To investigate this suggested mechanism thoroughly, methods of quantifying this movement directly would be preferable. Clement et al. (1984) also suggested that excessive pronation may cause conflicting rotations at the proximal and distal ends of the tendon, resulting from motions about the knee joint and subtalar joint being out of phase. These conflicting rotations were associated with high localised stress in the tendon. Limiting the amount of pronation was suggested to contribute to increased synchronisation of knee flexion and subtalar joint pronation. No evidence has been found in the present research to support this suggestion.

Factors other than those studied in the present research have been associated with Achilles tendon injury occurrence. MacLellan has categorically stated that the use of shock absorbing heel lifts is successful in the treatment of Achilles tendon injury. In contrast, Kvist (1994) presented a thorough description of the available literature relating to the etiology and treatment of Achilles tendon injury, including 358 related references, and stated that the role of shock absorbing heel lifts in the treatment of Achilles tendon injury was not known. In the present research, shock absorbing heel lifts attached to the plantar surface of the foot have been demonstrated to reduce the magnitude and loading rate of the vertical GRF for a single subject. A decrease in the maximum Achilles tendon force was also observed for this condition. Although it has been demonstrated that the Achilles tendon is not loaded on impact with the ground, MacLellan (1984) has suggested that there are shear stresses developed between the tendon and surrounding soft tissue upon impact. Modelling techniques are required to represent the tendon in detail, and to investigate the influence of variations in shock absorption on shear forces.

Following a study investigating the influence of running shoe heel height on ankle moments, Reinschmidt and Nigg (1995) suggested that single subject studies may help to explain why heel lift is used clinically in the treatment of Achilles tendon injury. The use of a single subject approach throughout the present research has illustrated the varied response across subjects in some aspects of Achilles tendon loading with heel lift manipulation. This behaviour may have been overlooked if a traditional group design had been employed. It has additionally been demonstrated that a single subject design can be used for a group of subjects to detect consistent behaviour, whilst not overlooking the response of individuals. This approach is therefore recommended for the future study of mechanisms of injury.

Reinschmidt and Nigg (1995) also stated that further research is required into the effect of changes in heel height on variables other than the sagittal plane ankle moment. In the present research ankle moments have been employed to estimate Achilles tendon forces. It has been clearly demonstrated throughout the studies that ankle moment alone is not sufficient for representation of Achilles tendon loading. In addition to Achilles tendon force estimation, sagittal and frontal plane kinematics have been analysed. Increased insight into the influence of heel lift manipulation on Achilles tendon loading has therefore been gained.

The relationship between sagittal plane shoe geometry and joint angles has not previously been investigated, with earlier studies being directed at the influence of changes in shock absorption characteristics of a shoe (Clarke et al., 1983a; Frederick et al., 1983). The present study has supported the suggestion that adaptations due to changes in footwear can occur prior to ground contact. The likelihood of adaptation in preparation for impact with the ground highlights the need to include this phase of running in the analysis of the responses of individuals to variations in shoe design characteristics.

In the present research, the controlled study of the influence of isolated heel lift intervention has been stimulated by the limited understanding of human behaviour that has been gained in running shoe studies. The results of the present study have indicated that there may be an optimum heel lift for each subject for the minimisation of maximum Achilles tendon force. However, the aim has not been to determine the optimum heel lift for a particular subject, since heel lift is only one design characteristic of a running shoe. Factors such as the material properties and frontal plane geometry of the shoe are also likely to influence each of the measured variables, making it impossible to state the most appropriate amount of heel lift. The aim of the studies performed in the present research has been to increase the understanding of the influence that manipulating heel lift can have on Achilles tendon loading. The investigation of running shoe influences would be a future development of the present work, but alone would not provide insight into the influence of isolated interventions on Achilles tendon loading.

In addition to the possible adverse effect of increased heel lift on Achilles tendon loading demonstrated in the present research, the loading of other structures of the foot may also be increased by the introduction of heel lift. The increased plantar-flexion observed with heel lift at ground contact and during midstance is likely to have contributed to an increased loading of the forefoot. This suggestion has been supported in studies investigating the influence of wearing high-heeled shoes in walking (Soames and Clark, 1985; Snow and Williams, 1994). The determination of Achilles tendon loading using estimation techniques has been facilitated by the predominantly two-dimensional forces experienced by this tendon, and by the fact that the associated muscle group is the dominant ankle plantar-flexor. Three-dimensional studies are likely to be required for estimation of the loading of other structures of the lower extremity, and difficulties in solving the distribution problem are likely to occur. To the present authors knowledge, the influence of heel lift on internal loading of the foot has not previously been investigated. It is therefore suggested that future investigation of the influence of heel lift includes the consideration of the loading of internal structures additional to the Achilles tendon.

In the study described in Chapter 4, differences in responses across subjects were attributed to different running styles. In the studies described in Chapter 6 and 7, all subjects were rearfoot strikers. It has been suggested that the differences in behaviour across these subjects may have been the result of variations in anatomy, injury status, level of flexibility, and strength. These variables are generally quantified in the clinical assessment of injured runners, but have tended not to be considered in biomechanical investigations. It is suggested that future studies should include the quantification of these measures, facilitating the identification of clinical variables influencing the response to interventions.

8.3 Practical Implications

Investigation of the influence of orthotic intervention on the medial-lateral stability of the rearfoot during midstance has been widespread (Smith et al., 1986; Nawoczenski et al., 1995). The use of orthotic devices with medial support has been demonstrated to reduce the amount of eversion of the calcaneus relative to the lower leg (Bates et al., 1979; Smith et al., 1986). In the present research, it has been demonstrated that an increased heel lift can be used to increase the angle that the foot makes with the ground during midstance. The significance of this finding was that the increased foot angle contributed to a decrease in the maximum amount of ankle dorsi-flexion. Although this result would appear to be predictable, it has not previously been demonstrated, and could not be assumed due to the possibility of changes in the orientation of the lower leg. Although a reduction in maximum ankle dorsi-flexion has been found to be a consistent response across subjects, the absolute influence of a lift of defined dimensions on foot and ankle angles was not found to be predictable. Thus, the identification of an appropriate amount of heel lift to attain specified joint angles appears to require an element of trial and error.

The consistent reduction in the maximum amount of ankle dorsi-flexion with increased heel lift demonstrated across subjects in the present research, has shown that the maximum overall length of the muscle-tendon complex during stance can be reduced using this intervention. Thus, if an intervention is required to reduce this length then the results of the present study indicate that the use of a heel lift is recommended. The apparent difference in the relative contributions of the Achilles tendon and the associated muscle fibres to this overall change in length may be due to factors such as individual anatomy, injury status, level of flexibility, and strength differences. It has been beyond the scope of the present research to quantify these primarily clinical measures.

An important characteristic highlighted by the results of the studies in this research has been the varied response in Achilles tendon loading across subjects, supporting findings in the literature that subjects demonstrate varied responses to identical interventions (Therrien et al., 1982; Lees and McCullagh, 1984). The evaluation of injured athletes by podiatrists has become widespread at all competitive levels. In the clinical assessment of individuals, static weight bearing or nonweight-bearing measurements are routinely taken, based on the theory of Root et al. (1971) concerning the requirement of a neutral foot during midstance. These measures have been used for identification of possible anatomical factors related to the injury occurrence (Hlavac, 1977). The inappropriateness of utilising static measures to predict dynamic movements has been demonstrated in the literature (Hamill et al., 1989; Knutzen and

Price, 1994). Dynamic assessment during walking or running has recently become more widespread, possibly due to the relatively easy measurement of frontal plane rearfoot movement (Cornwall and McPoil, 1995). Despite many practitioners describing the successful treatment of injuries using orthoses, Kilmartin and Wallace (1995) have highlighted the fact that no research has demonstrated the advantage of a particular intervention on the treatment of a defined injury. Although excessive movement of the foot and ankle are likely to be related to injury occurrence, the cause of an injury will be the excessive loading of the structure. An approach which involves consideration of the loading of the injured structure would therefore appear more appropriate than the routine assessment of movement patterns alone.

Clinical procedures utilising this approach have recently been proposed. McPoil and Hunt (1995) described the use of a 'tissue stress model' for the evaluation and treatment of foot and ankle disorders. Four steps were recommended using this model. Step 1 involved the identification of the tissues being excessively stressed based on subject history, symptoms and other subjective information. Step 2 involved the application of controlled stresses to these tissues, using weight-bearing and nonweight-bearing tests. Step 3 involved the use of these findings to identify the activities possibly causing this loading. Step 4 required the prescription of a management procedure, including rest, orthoses, and physiotherapy treatments such as the use of ultrasound.

It is suggested that the ideal treatment protocol would involve the development of this approach, utilising detailed biomechanical analyses in addition to the described clinical procedures. Following the clinical identification of the structure under stress (Step 1 and Step 2), and the identification of the likely contributing factors to injury occurrence (Step 3), the identification of suitable management procedures (Step 4) could include biomechanical measurements. In particular, the estimation of stress and strain experienced by the structure, and the investigation of the most appropriate intervention for reduction of this loading are recommended. Estimation methods such as those developed in the present research could be employed to achieve this aim.

The evidence provided in the present research has highlighted the difficulty in the recommendation of common treatments for all individuals. For the elite performer, the design of personalised footwear may be realistic. However, the vast majority of the running population are unlikely to obtain running shoes specific to their requirements. Several running shoe companies have used criteria such as foot type to categorise runners to determine the most important shoe design characteristics for these different groups. In general runners are required to identify whether they have characteristics such as a 'flat foot' or a 'high arch' (Saucony Ltd., Puma Ltd.). Runners may be required to know whether they exhibit 'excessive' pronation or supination, for indentifation of appropriate footwear (Nike Inc., Reebok Ltd.). This approach is limited due to the difficulties in predicting dynamic foot behaviour using static measurements, and due to the general reliance on individual runners or untrained shop assistants to identify foot types and running styles. One running shoe company (adidas (UK) Ltd.) has taken the step of introducing a system to measure the pressure distribution between the plantar surface of the foot and the ground during running in selected running shoe shops. The aim has been to utilise these systems to identify the most appropriate shoe for the customer. With a running population that is searching for any intervention likely to increase performance or reduce the likelihood of injury occurrence, the commercial benefits of such an approach are likely to be considerable. Although differences in plantar pressure distribution have been demonstrated with variations in rearfoot movement (eg. Milani et al., 1995), changes in the distribution of pressure do not necessarily represent differences in foot movement. Advances in biomechanical measures can clearly be employed for the recommendation of running shoes most suitable for an individual. The amount of biomechanical knowledge supporting the systems available at present is limited. The use of methods such as those developed in the present research to provide information on the influence of shoes on the loading of internal structures would be ideal.

Cavanagh et al. (1985) described an approach to the biomechanical profiling of elite distance runners. These authors recommended the development of a database of normative biomechanical measurements and the provision of a biomechanical screening service to athletes. It is of interest to assess the extent to which a database is available for practitioners and biomechanists in Great Britain at present, and what level of service has been made available to elite athletes. Data are provided in the literature on factors including the amount of rearfoot movement measured for athletes (Clarke et al., 1983b), typical GRF patterns (Cavanagh and Lafortune, 1980, and joint angles during running (Milliron and Cavanagh, 1990), but a database of typical values in the non-injured population is not available. The service available to injured athletes is limited to the clinical assessment of static and dynamic movement patterns. This situation is in contrast to the physiological services available to the British elite athlete, with extensive field and laboratory testing available, and the development of a database of physiological measures by the British Olympic Association, through testing of elite athletes at the British Olympic Medical Centre. The development of a database and the provision of a routine service for elite athletes are required. It is speculated that these could be incorporated within the structure of the proposed British Academy of Sport. The routine detailed biomechanical study of elite athletes attending the Academy would facilitate the development of a database, providing extensive information to the researcher and the practitioner. The treatment of injured athletes could therefore involve the use of the combined experience of practitioners and biomechanists, together with the database information.

8.4 Summary of Results

The studies presented in this research have demonstrated the potential of an estimation method for investigation of the influence of footwear interventions on Achilles tendon loading. In response to the main question addressed in the initial study, it has been demonstrated that maximum Achilles tendon force can be influenced by heel lift manipulation. Subsequent investigation of the influence of a controlled isolated heel lift has demonstrated that in some cases maximum Achilles tendon force increased when the heel was raised. The possibility of an adverse influence of raising the heel has therefore been highlighted.

A varied response to heel lift intervention with running style has demonstrated the importance of the identification of individual style when treating Achilles tendon injury. The subsequent study of a group of rearfoot strikers has demonstrated that running style alone is not sufficient for identification of the influence of heel lift on the loading of the Achilles tendon. The importance of individual assessment has therefore been highlighted.

The consistent reduction in ankle dorsi-flexion at impact with increased heel lift has highlighted the adaptative kinematic response to this intervention. The common reduction in average loading rate with heel lift has indicated that the rate of change of stress and strain are influenced by heel lift manipulation. Since heel lift has been associated with a reduction in the incidence of Achilles tendon injury, the possibility of loading rate being implicated in this injury occurrence has been highlighted.

REFERENCES

Abdel-Aziz, Y.I. and Karara, H.M. (1971). Direct linear transformation from comparator coordinates into object space coordinates in close-range photogrammetry. In *Proceedings of the ASP Symposium on Close Range Photogrammetry*, 1-18. American Society of Photogrammetry, Falls, Church.

Abrahams, M. (1967). Mechanical behaviour of tendon in-vitro. *Medical and Biological Engineering* **5**, 433-443.

Alexander, R.McN. and Vernon, A. (1975). The dimensions of the knee and ankle muscles and the forces they exert. *Journal of Human Movement Studies* **1**, 115-123.

Alexander, R.McN. and Bennet-Clark (1977). Storage of elastic strain energy in muscle and other tissues. *Nature* **265**, 114-117.

Alexander, R.McN., Maloiy, G.M.O., Ker, R.F., Jayes, A.S. and Warui, C.N. (1982). The role of tendon elasticity in the locomotion of the camel. *Journal of Zoology* **198**, 293-313.

Andrews, J.G. (1984). On the specification of joint configurations and motions. *Journal of Biomechanics* **17**, 155-158.

Angulo, R.M. and Dapena, J. (1992). Comparison of film and video techniques for estimating three-dimensional coordinates within a large field. *International Journal of Sport Biomechanics* **8**, 145-151.

Archambault, J.M., Preston Wiley, J. and Bray, R.C. (1995). Exercise loading of tendons and overuse injuries. *Sports Medicine* **20**, 77-89.

Arebald, M., Nigg, B.M., Ekstrand, J., Olsson, K.O. and Ekstrom, H. (1990). Three-dimensional measurement of rearfoot motion during running. *Journal of Biomechanics* **23**, 933-940.

Bates, B.T., Osternig, L.R., Mason, B. and James, S.L. (1978). Lower extremity function during the support phase of running. In E. Asmussen and K. Jorgensen (Eds.). *Biomechanics* **VI-B**, 30-39. University Park Press, Baltimore.

Bates, B.T., Osternig, L.R., Mason, B. and James, S.L. (1979). Foot orthotic devices to modify selected aspects of lower extremity mechanics. *American Journal of Sports Medicine* **7**, 338-342.

Bates, B.T., Devita, P. and Kinoshita, H. (1983a). The effect of intra-individual variability on sample size. In B.M. Nigg and B.A. Kerr (Eds.), *Biomechanical aspects of sport shoes and playing surfaces*, 191-198. Calgary, AB: University of Calgary.

Bates, B.T., Osternig, L.R., Sawhill, J.A. and James, S.L. (1983b). An assessment of subject of variability, subject-shoe interaction and the evaluation of running shoes using GRF data. *Journal of Biomechanics* **16**, 181-191.

Bates, B.T. (1989). Comment on 'the influence of running velocity and midsole hardness on external impact forces in heel-toe running.' *Journal of Biomechanics* **22**, 963-965.

Bates, B.T., Dufek, J.S. and Davis, H.P. (1992). The effect of trial size on statistical power. *Medicine and Science in Sports and Exercise* **24**, 1059-1065.

Bates, B.T. (1996). Single subject methodology: an alternative approach. *Medicine and Science in Sports and Exercise* **28**, 631-643.

Bauman, W. (1981). On mechanical loads on the body during sports activities. In A. Morecki, K. Fidelus, K. Kedzior and A. Wit (Eds.). *Biomechanics* **VII-B**, 79-87. University Park Press, Baltimore.

Benedict, J.V., Walker, L.B. and Harris, E.H. (1968). Stress-strain characteristics and tensile strength of unembalmbed human tendon. *Journal of Biomechanics* **1**, 53-63.

Bennett, M.B., Ker, R.F., Dimery, N.J. and Alexander, R.McN. (1986). Mechanical properties of various mammalian tendons. *Journal of Zoology* **209**, 537-548.

Blake, R.L. and Ferguson, H. (1992). Extrinsic rearfoot posts. *Journal of the American Podiatric Medical Association* **82**, 202-207.

Bobbert, M.F., Huijing, P.A. and Van Ingen Schenau, G.J. (1986). A model of the human triceps surae muscle-tendon complex applied to jumping. *Journal of Biomechanics* **19**, 887-898.

Bobbert, M.F., Ettema, G.C. and Huijing, P.A. (1990a). *European Journal of Applied Physiology* **61**, 323-329.

Bobbert, M.F. and Van Ingen Schenau, G.J. (1990b). Mechanical output about the ankle joint in isokinetic plantar flexion and jumping. *Medicine and Science in Sports and Exercise* **22**, 660-668.

Bobbert, M.F. and Van Ingen Schenau, G.J. (1990c). Isokinetic plantar flexion: experimental results and model calculations. *Journal of Biomechanics* **23**, 105-119.

Bobbert, M.F., Yeadon, M.R. and Nigg, B.M. (1992). Mechanical analysis of the landing phase in heel-toe running. *Journal of Biomechanics* **25**, 223-234.

van den Bogert, A.J. and Glossop, (1996). Discussion on *BIOMECH-L.*

Bojsen-Moller, F. (1983). Biomechanical effects of shock absorbing heels in walking. In B.M. Nigg and B.A. Kerr (Eds.), *Biomechanical aspects of sport shoes and playing surfaces*, 73-81. Calgary, AB: University of Calgary.

Bourassa, P. and Therrien, R.G. (1981). A force platform for the evaluation of human locomotion and posture. *Biomechanics* **VIII-A**, pp. 582-588.

Brand, R.A., Crownshield, R.D., Wittstock, C.E., Pedersen, D.R. and Clark, C.R. (1982). A model of the lower extremity muscular anatomy. *ASME Journal of Biomechanical Engineering*, **104**, 304-310.

Braune, W. and Fischer, C. (1889). *The Centre of Gravity of the Human Body as related to the Equipment of the German Infantry.* S. Herzel Verlag, Leipzig.

Brewster, R.C., Chao, E.Y. and Stauffer, R.N. (1974). Force analysis of the ankle joint during the stance phase of gait. *27th A.C.E.M.B.* Alliance for Engineers, Philadelphia. Cited by Proctor and Paul (1982).

Bryan, A.J. (1987). Single-subject designs for evaluation of sport psychology interventions. *The Sport Psychologist* **1**, 283-292.

Buczek, F.L. and Cavanagh, P.R. (1990). Stance phase knee and ankle kinematics and kinetics during level and downhill running. *Medicine and Science in Sports and*

Exercise **22**, 669-677.

Burdett, R.G. (1982). Forces predicted at the ankle during running. *Medicine and Science in Sports and Exercise* **14**, 308-316.

Butler, D.L., Grood, E.S., Noyes, F.R., Zernicke, R.F. and Brackett, K. (1984). Effects of structure and strain measurement technique on the material properties of young human tendon and fascia. *Journal of Biomechanics* **17**, 579-596.

Caldwell, G.E. (1995). Tendon elasticity and relative length: effects on the Hill two-component muscle model. *Journal of Biomechanics* **11**, 1-24.

Campbell, M.J. and Machin, D. (1990). *Medical Statistics. A Commonsense Approach.* John Wiley and Sons Ltd.

Caster, B.L., Chen, F.C., Dufek, J.S. and Bates, B.T. (1994). Single subject design in biomechanics research: justification of statistical assumptions. *Conference Proceedings, Eighth Biennial Canadian Society for Biomechanics*, Calgary, 294-295.

Cavanagh, P.R. and Gregor, R.J. (1975). Knee joint torque during the swing phase of normal treadmill walking. *Journal of Biomechanics* **8**, 337-344.

Cavanagh, P.R. and Lafortune, M.A. (1980). Ground reaction forces in distance running. *Journal of Biomechanics* **13**, 397-406.

Cavanagh, P.R., Valiant, G.A. and Misevich, K.W. (1984). Biological aspects of modelling shoe/surface interaction during running. In E.C.Frederick (Ed). *Sports Shoes and Playing Surfaces* 24-47. Human Kinetics Publishers.

Cavanagh, P.R., Andrew, G.C., Kram, R., Rodgers, M.M., Sanderson, D.J. and Hennig, E. (1985). An approach to biomechanical profiling of elite distance runners. *International Journal of Sport Biomechanics* **1**, 36-62.

Cavanagh, P.R. (1987). The biomechanics of lower extremity action in distance running. *Foot and Ankle* **7**, 197-217.

Challis, J.H. (1991). *Estimating individual muscle forces in human movement.* Unpublished Ph.D. Thesis. Loughborough University.

Challis, J.H. and Kerwin, D.G. (1992). Accuracy assessment and control point configuration when using the DLT for photogrammetry. *Journal of Biomechanics* **25**, 1053-1058.

Challis, J.H. and Kerwin, D.G. (1994). Determining individual muscle forces during maximal activity: Model development, parameter determination, and validation. *Human Movement Science* **13**, 29-61.

Challis, J.H. and Kerwin, D.G. (1996). Quantification of the uncertainties in resultant joint moments computed in a dynamic activity. *Journal of Sports Sciences* **14**, 219-231.

Chandler, R.F., Clauser, C.E., McConville, J.T., Reynolds, H.M. and Young, J.W. (1975). Investigation of the inertial properties of the human body. *AMRL Technical Report*, 74-137. Wright Patterson Air Force Base, Ohio.

Clark, J.E., Scott, S.G. and Mingle, M. (1989). Viscoelastic shoe insoles: their use in aerobic dancing. *Archives of Physical and Medical Rehabilitation* **70**, 37-40.

Clarke, T.E., Lafortune, M.A., Williams, K.R. and Cavanagh, P.R. (1980). The relationship between centre of pressure and rearfoot movement in distance running. *Medicine and*

Science in Sports and Exercise **12**, 192.

Clarke, T.E., Frederick, E.C. and Cooper, L.B. (1983a). Biomechanical measurement of running shoe cushioning properties. In B.M. Nigg and B.A. Kerr (Eds.), *Biomechanical aspects of sport shoes and playing surfaces*, 25-33. Calgary, AB: University of Calgary.

Clarke, T.E., Frederick, E.C. and Hamill, C.L. (1983b). The effects of shoe design parameters on rearfoot control in running. *Medicine and Science in Sports and Exercise* **15**, 376-381.

Clarke, T.E., Frederick, E.C. and Hamill, C.L. (1984). The study of rearfoot movement in running. In E.C. Frederick (Ed). *Sports Shoes and Playing Surfaces* 166-189. Human Kinetics Publishers.

Clauser, C.E., McConville, J.T. and Young, J.W. (1969). Weight, volume, and centre of mass of segments of the human body. *AMRL Technical Report*, 69-70. Wright Patterson Air Force Base, Ohio.

Clement, D.B., Taunton, J.E., Smart, G.W. (1984). Achilles tendinitis and peritendinitis: etiology and treatment. *The American Journal of Sports Medicine* **12**, 179-184.

Cohen, J. (1969, 1990). *Statistical Power Analysis for the Behavioural Sciences.* Academic Press, Inc.

Cohen, L. and Holliday, M. (1982). *Statistics for Social Scientists.* Harper and Row, Publishers, London.

Cole, G.K., Nigg, B.M., Fick, G.H. and Morlock, M.M. (1995). Internal loading of the foot and ankle during impact in running. *Journal of Applied Biomechanics* **11**, 25-46.

Cornwall, M.W. and McPoil, T.G. (1992). Effect of rearfoot posts in reducing forefoot forces. A single subject design. *Journal of the American Podiatric Medical Association* **82**,371-374.

Cox, D.R. (1968). *Planning of Experiments.* John Wiley and Sons, Inc.

Cureton, T.K. (1939). Elementary principles and techniques of cinematographic analysis as aids in athletic research. *Research Quarterly* **10**, 3-23.

Davidson, R.G and Taunton, J.E. (1987). Achilles tendinitis. *Medicine and Sport Science* **23**, 71-79.

Dempster, W.T. (1955). Space requirements of the seated operator. *WADC Technical Report,* 55-59. Aerospace Medical Research Laboratory, Wright-Patterson Air Force Base, Ohio.

Denoth, J. (1986). Load on the locomotor system and modelling. In B.M. Nigg (Ed.) *Biomechanics of running shoes*, pp. 63-116. Champaign, IL: Human Kinetics.

DeVita, P. and Bates, B.T. Intraday reliability of ground reaction force data. *Human Movement Science* **7**, 73-85.

Dickinson, J.A., Cook, S.D. and Leinhardt, T.M. (1985). The measurement of shock waves following heel strike while running. *Journal of Biomechanics* **18**, 415-422.

Dillman, C.J. (1975). Kinematic analyses of running. In *Exercise and Sports Science Reviews* **3**, 193-218.

Dimery, N.J., Ker, R.F. and Alexander, R.McN. (1986). Elastic properties of the feet of deer. *Journal of Zoology* **208**, 161-169.

Doolittle, T.L. (1971). Errors in linear measurement with cinematographical analysis. *Kinesiology Review,* 32-38.

Dufek, J.S. and Bates, B.T. (1990). The evaluation and prediction of impact forces during landings. *Medicine and Science in Sports and Exercise* **22**, 370-377.

Dufek, J.S., Bates, B.T., Stergiou, N. and James, R.C. (1995). Interactive effects between group and single-subject response patterns. *Human Movement Science* **14**, 301-323.

Edgington, E.S. (1967). Statistical inference from N=1 experiments. *Journal of Psychology* 65, 195-199.

Edington, C.J., Frederick, E.C. and Cavanagh, P.R. (1990). In P.R. Cavanagh (Ed.), *Biomechanics of Distance Running*. pp. 135-164. Human Kinetics Publishers, Champaign, IL.

Ellis, D.G. (1969). Cross-sectional area measurements for tendon specimens: A comparison of several methods. *Journal of Biomechanics* **2**, 175-186.

Eng, J.J. and Pierrynowski, M.R. (1994). The effect of soft foot orthotics on three-dimensional lower limb kinematics during walking and running. *Physical Therapy* **74**, 836-844.

Engsberg, J.R. (1987). A biomechanical analysis of the talocalcaneal joint - in vitro. *Journal of Biomechanics* **20**, 429-442.

Fauno, P., Kalund, S., Andreasen, I. and Jorgensen, U. (1993). Soreness in lower extremities and back is reduced by use of shock absorbing heel inserts. *International Journal of Sports Medicine* **14**, 288-290.

Franks, B.D. and Huck, S.W. (1986). Why does everyone use the 0.05 significance level ? *Research Quarterly* **57**, 245-249.

Frederick, E.C. (1983). Measuring the effects of shoes and surfaces on the economy of locomotion. In B.M. Nigg and B.A. Kerr (Eds.) *Biomechanical Aspects of Sports Shoes and Playing Surfaces* pp. 93-114.

Frederick, E.C. (1986). Kinematically mediated effects of sport shoe design: a review. *Journal of Sports Sciences* **4**, 169-184.

Fukashiro, S., Komi, P.V., Jarvinen, M. and Miyashita, M. (1993). Comparison between the directly measured achilles tendon force and the tendon force calculated from the ankle moment during vertical jumps. *Clinical Biomechanics* **8**, 25-30.

Gerritsen, K.G.M., van den Bogert, A.J. and Nigg, B.M. (1995). Direct dynamics simulation of the impact phase in heel - toe running. *Journal of Biomechanics* **28**, 661-668.

Gray, H. (1858). *Gray's Anatomy.* T. Pickering Pick and R. Howden (Eds.) Chancellor Press.

Gregor, R.J., Roy, R.R., Whiting, W.C., Hodgson, J.A. and Edgerton, V.R. (1988). Force-velocity potentiation in cat soleus muscle during treadmill locomotion. *Journal of Biomechanics* **21**, 721-732.

Gregor, R.J., Komi, P.V., Browning, R.C. and Jarvinen. (1991). A comparison of the triceps surae and residual muscle moments at the ankle during cycling. *Journal of Biomechanics* **24**, 287-297.

Gregor, R.J. (1993). Skeletal muscle mechanics and movement. In M.D. Grabiner (Ed.) *Current Issues in Biomechanics*, pp. 123-150.

Gregor, R.J. and Abelew, T.A. (1994). Tendon force measurements in musculoskeletal biomechanics. *Sports Science Review* **3**, 8-33.

Grieve, D.W., Pheasant, S. and Cavanagh, P.R. (1978). Prediction of gastronemius length from knee and ankle joint posture. Asmussen, E. and Jorgensen, K. (Eds). *Biomechanics* **VI-A**, 405-412. University Park Press, Baltimore.

Groh, H. and Baumann, W. (19) Joint and muscle forces acting in the leg during gait. *Biomechanics* , 328-333.

Grisogono, V. (1989). Physiotherapy treatment for Achilles tendon injuries. *Physiotherapy* **75**, 562-572.

Grood, E.S. and Suntay, W.J. (1983). A joint coordinate system for the clinical description of three-dimensional motions: Application to the knee. *Journal of Biomechanical Engineering* **105**, 136-144.

Gross, M.T. (1995). Lower quarter screening for skeletal malalignment - suggestions for orthotics and shoewear. Journal of Orthopaedic Sports Physical Therapy 21, 389-405.

Hamill, J., Freedson, P.S., Boda, W. and Reichsman, F. (1988). Effects of shoe type on cardiorespiratory responses and rearfoot motion during treadmill running. *Medicine and Science in Sports and Exercise* **20**, 515-521.

Hamill, J., Bates, B.T., Knutzen, K. and Kirkpatrick, G. (1989). Relationship between selected static and dynamic lower extremity measures. *Clinical Biomechanics* **4**, 217-225.

Hamill, J., Bates, B.T. and Holt, K.G. (1992). Timing of lower extremity joint actions during treadmill running. *Medicine and Science in Sports and Exercise* **24**, 807-813.

Hamill, J., Milliron, M.J. and Healy, J.A. (1994). Stability and rearfoot motion testing: a proposed standard. Proceedings, Symposium on the Biomechanics of Functional Footwear, pp. 12-13. Eighth Biennial Conference, Canadian Society for Biomechanics, Calgary.

Hanavan, E.P. (1964). A mathematical model of the human body. *AMRL Technical Report*, 64-102. Wright-Patterson Air Force Base, Ohio.

Harrison, R.N., Lees, A., McCullagh, P.J.J and Rowe, W.B. (1987). Bioengineering analysis of muscle and joint forces acting in the human leg during running. *Biomechanics* **X-B**, 855-862.

Hatze, H. (1975). A new method for the simultaneous measurement of the moment of inertia, the damping coefficient and the location of the centre of mass of a body segment in situ. *European Journal of Applied Physiology* **34**, 217-226.

Hatze, H. (1980). A mathematical model for the computational determination of parameter values of anthropomorphic segments. *Journal of Biomechanics* **13**, 833-843.

Hawkins, D. (1993). Ligament biomechanics. In M.D. Grabiner (Ed.) *Current Issues in*

Biomechanics, pp. 123-150.

Herrick, W.C., Kingsbury, H.B. and Lou, D.Y.S. (1978). A study of the normal range of strain, strain rate and stiffness of tendon. *Journal of Biomedical Materials Research* **12**, 877-894.

Herzog, W. and Loitz, B. (1995). Biomaterials: Tendon. In W. Herzog and B.M. Nigg (Eds.) *Biomechanics of the musculoskeletal system,* pp. 133-153.

Herzog, W. (1995). Biomaterials: Muscle. In W. Herzog and B.M. Nigg (Eds.) *Biomechanics of the musculoskeletal system,* pp. 154-190.

Hicks, J.H. (1953) The mechanics of the foot I. The joints. *Journal of Anatomy* **87**, 345-357.

Hlavac, H.F. (1977). *The Foot Book.* World Publications.

Hof, A.L., Geelen, A. and van den Berg, J. (1983). Calf muscle moment, work and efficiency in level walking; role of series elasticity. *Journal of Biomechanics* **16**, 523-537.

Hoffer, J.A., Caputi, A.A., Pose, I.E. and Griffiths, R.I. (1989). Role of muscle activity and load on the relationship between muscle spindle length and whole muscle length in the freely moving cat. *Progress in Brain Research* **80**, 75-85.

Holden, J.P., Cavanagh, P.R., Williams, K.R. and Bednarski, K.N. (1986). Foot angles during walking and running. *Biomechanics* **IX-A**, 451-456.

Hooley, C.J, McCrum, N.G. and Cohen, R.E. (1980). The viscoelastic deformation of tendon. *Journal of Biomechanics* **13**, 521-528.

Hoy, M.G., Zajac, F.E. and Gordon, M.E. (1990). A musculoskeletal model of the human lower extremity: the effect of muscle, tendon, and moment arm on the moment-angle relationship of musculotendon actuators at the hip, knee and ankle. *Journal of Biomechanics* **23**, 157-169.

van Ingen Schenau, G.J. (1984). An alternative view of the concept of utilisation of elastic energy in human movement. *Human Movement Science* **3**, 301-336.

Inman, V.T. (1947). Functional aspects on the abductor muscles of the hip. *Journal of Bone and Joint Surgery* **29A**, 607-619.

Inman, V.T. (1976). *The Joints of the Ankle.* Baltimore, Williams and Wilkins.

James, S.L., Bates, B.T. and Osternig, L.R. (1978). Injuries to runners. *American Journal of Sports Medicine* **6**, 40-50.

James, S.L. and Jones, D.C. (1990). Biomechanical aspects of distance running injuries. In P.R. Cavanagh (Ed.) *Biomechanics of Distance Running,* pp. 249-270.

Johansson, C. (1986). Injuries in elite orienteers. *American Journal of Sports Medicine* **14**, 410-452.

Kaelin, X., Denoth, J., Stacoff, A. and Stuessi, E. (1985). Cushioning during running - material tests contra subject tests. In S.M. Perren and E. Schneider (Eds) *Biomechanics: Current Interdisciplinary Research,* pp. 645-650. Martinus Nijhoff, Dordrecht.

Ker, R.F. (1981). Dynamic properties of the plantaris tendon of sheep (ovis aries). *Journal of Experimental Biology* **93**, 283-302.

Ker, R.F., Bennett, M.B., Bibby, S.R., Kester, R.C. and Alexander, R.McN. (1987). The spring in the arch of the human foot. *Nature* **325**, 147-149.

Ker, R.F., Alexander, McN.R. and Bennett, M.B. (1988). Why are mammalian tendons so thick? *Journal of Zoology* **216**, 309-324.

Kerwin, D.G. (1994). Video and cinefilm digitisation accuracy. In J. Watkins (Ed.) *Proceedings of the Biomechanics Section of the British Association of Sport and Exercise Sciences* **19**, 21-24.

Kilmartin, T.E. and Wallace, W. (1994). The scientific basis for the use of biomechanical foot orthoses in the treatment of lower limb sports injuries - a review of the literature. *British Journal of Sports Medicine* **28**, 180-184.

Kinoshita, H., Bates, B.T. and DeVita, P. (1985). Intertrial variability for selected running gait parameters. In D. Winter, R. Norman, R. Wells, K.Hayes and A. Patla. (Eds.) *Biomechanics* **IX-A**, pp. 499-502.

Knutzen, K.M. and Price, A. (1994). Lower extremity static and dynamic relationships with rearfoot motion in gait. *Journal of the American Podiatric Medical Association* **84**, 171-180.

Komi, P.V. (1990). Relevance of in-vivo force measurements to human biomechanics. *Journal of Biomechanics* **23**, 23-34.

Kratochwill, T.R. (1978). *Single Subject Research: Strategies for Evaluating Change.* London, Academic Press.

Krissoff, W.B. and Ferris, W.D. (1979). Runners' injuries. *Physician and Sportsmedicine* **7**, 55-64.

Kvist, M. (1994). Achilles tendon injuries in athletes. *Sports Medicine* **18**, 173-201.

Lanczos, C. (1957). *Applied Analysis.* New York: Pitman.

Leach, R.E., James, S., and Wasilewski, S. (1981). Achilles tendinitis. *The American Journal of Sports Medicine* **9**, 93-98.

Lees, A. (1980). An optimised film analysis method based on finite difference techniques. *Journal of Human Movement Studies* **6**, 165-180.

Lees, A. and McCullagh, P.J. (1984). A preliminary investigation into the shock absorbency of running shoes and shoe inserts. *Journal of Human Movement Studies* **10**, 95-106.

Lees, A. and Haynes, A. (1995). The effect of speed on dynamic load rate in running. *Journal of Sports Science* , p. 9.

Lesh, M.D., Masour, J.M. and Simon, S.R. (1979). A gait analysis subsystem for smoothing and differentiation of human motion data. *Journal of Biomechanical Engineering* **101**, 205-212.

Light, L.H., MacLellan, G.E. and Klenerman, L. (1980). Skeletal muscle transients on heel strike in normal walking with different footwear. *Journal of Biomechanics* **13**, 477-480.

Lippold, O.C.J. (1952). The relation between integrated action potentials in a human muscle and its isometric tension. *Journal of Physiology* **117**, 492-499.

Lowdon, A., Bader, D.L. and Mowat, A.G. (1984). The effect of heel pads on the treatment of Achilles tendinitis: A double blind trial. *The American Journal of Sports Medicine* **12**, 431-435.

Luethi, S.M. and Stacoff, A. (1987). The influence of the shoe on foot mechanics in running.

Medical Sport Science **25**, 72-85.

Lysholm, J. and Wiklander, J. (1987). Injuries in runners. *American Journal of Sports Medicine* **15**, 168-171.

MacLellan, G.E. and Vyvyan, B. (1981). Management of pain beneath the heel and Achilles tendonitis with visco-elastic heel inserts. *British Journal of Sports Medicine* **15**,117-121.

MacLellan, G.E. (1984). Skeletal strike transients, measurement, implications, and modification by footwear. In E.C. Frederick (Ed). *Sports Shoes and Playing Surfaces* 76-86. Human Kinetics Publishers.

Malsen, B.A and Ackland, T.R. (1994). Radiographic study of skin displacement errors in the foot and ankle during standing. *Clinical Biomechanics* **9**, 291-296.

Marti, B., Vader, J.P. and Minder, C.E. (1988). On the epidemiology of running injuries. The 1984 Bern Grand-Prix study. *American Journal of Sports Medicine* **16**, 285-294.

Matthews, L.S. and Ellis, D. (1968). Viscoelastic properties of cat tendon: effects of time after death and preservation by freezing. *Journal of Biomechanics* **1**, 65-71.

McLaughlin, T.M., Dillman, C.J. and Lardner, T.J. (1977). Biomechanical analysis with cubic spline functions. *Research Quarterly* **48**, 569-581.

van Mechelen, W. (1992). Epidemiology of running injuries. *Sports Medicine* **14**, 320-335.

Messier, S.P. and Pittala, K.A. (1988). Etiological factors associated with selected running injuries. *Medicine and Science in Sports and Exercise* **20**, 501-505.

Milani, T.L., Schnabel, G. and Hennig, E.M. (1995). Rearfoot motion and pressure distribution patterns during running in shoes with varus and valgus wedges. *Journal of Applied Biomechanics* **11**, 177-187.

Miller, D.I. and Nelson, R.C. (1973). Data smoothing in biomechanics. In D.I. Miller and Nelson, R.C. (Eds.) *Biomechanics of Sport*, 243-248.

Miller, D.I. (1978). Biomechanics of running - what should the future hold ? *Canadian Journal of Applied Sports Sciences* **3**, 229-236.

Miller, D.I. (1990). Ground reaction forces in distance running. In P.R. Cavanagh (Ed.), *Biomechanics of Distance Running*, 203-224. Human Kinetics Books.

Milliron, M.J. and Cavanagh, P.R. (1990). Sagittal plane kinematics of the lower extremity during distance running. In P.R. Cavanagh (Ed.), *Biomechanics of Distance Running*. 65-105. Human Kinetics Publishers, Champaign, IL.

Morgan, D.L., Proske, U. and Warren, D. (1978). Measurements of muscle stiffness and the mechanism of elastic storage of energy in hopping kangeroos. *Journal of Physiology* **282**, 253-261.

Morlock, M.M. and Nigg, B.M. (1988). Dynamic and quasi-static models of the foot. In: de Groot (Ed) *Biomechanics* **XI-A**, , 410-416. Amsterdam: Free University Press.

Morlock, M.M. and Nigg, B.M. (1991). Theoretical and practical results on the influence of the representation of the foot for the estimation of internal forces with models. *Clinical Biomechanics* **6**, 3-13.

McPoil, T.G. and Cornwall, M.W. (1991). Rigid versus soft foot orthoses. *Journal of the American Podiatric Medical Association* **81**, 638-642.

McPoil, T.G. and Hunt, G.C. (1995). Evaluation and management of foot and ankle disorders: present problems and future directions. *Journal of Sports Physical Therapy* **21**, 381-388.

Munro, C.F., Miller, D.I. and Fuglevand, A.J. (1987). Ground reaction forces in running: a reexamination. *Journal of Biomechanics* **20**, 147-155.

Murray, M.P., Guten, G.N., Baldwin, J.M. and Gardner, G.M. (1976). A comparison of plantar-flexion torque with and without the triceps surae. *Acta Orthopedia Scandinavia* **47**, 122-124.

Natrup, J., Peikenkamp, K. and Nicol, K. (1994). The influence of fatigue on the internal load during long distance running. Abstract, *Second World Congress of Biomechanics, Amsterdam.* L. Blankevoort and J.G.M. Kooloos (Eds), 375.

Nawoczenski, D.A., Cook, T.M. and Saltzman, C.L. (1995). The effect of foot orthotics on three-dimensional kinematics of the leg and rearfoot during running. *Journal of Sports Physical Therapy* **21**, 317-327.

Nicholl, J.P., Coleman, P. and Williams, B.T. (1991). *A national study of the epidemiology of exercise-related injury and illness.* The Sports Council, London.

Nigg, B.M. (1983). External force measurements with sport shoes and playing surfaces. In B.M. Nigg and B.A. Kerr (Eds.), Biomechanical aspects of sport shoes and playing surfaces, 143-151. Calgary, AB: University of Calgary.

Nigg, B.M. (Ed.) (1986a) *Biomechanics of running shoes.* Champaign, IL: Human Kinetics.

Nigg, B.M. (1986b). Biomechanical aspects of running. In B.M. Nigg (Ed.) *Biomechanics of running shoes*, pp. 1-26. Champaign, IL: Human Kinetics.

Nigg, B.M., Bahlsen, H.A., Luethi, S.M. and Stokes, S. (1987). The influence of running velocity and midsole hardness on external impact forces in heel-toe running. *Journal of Biomechanics* **20**, 951-959.

Nigg, B.M. and Morlock, M.M. (1987). The influence of lateral heel flare of running shoes on pronation and impact forces. *Medicine and Science in Sports and Exercise* **19**, 294-302.

Nigg, B.M and Bahlsen, H.A. (1988). Influence of heel flare and midsole construction on pronation, supination, and impact forces for heel-toe running. *International Journal of Sport Biomechanics* , 205-219.

Nigg, B.M., Herzog, W. and Read, L.J. (1988). Effect of viscoelastic shoe insoles on vertical impact forces in heel-toe running. *The American Journal of Sports Medicine* **16**, 70-76.

Nigg, B.M. and Bobbert (1990). On the potential of various approaches in load analysis to reduce the frequency of sports injuries. *Journal of Biomechanics* **23**, 3-12.

Nigg, B.M. (1994). In W. Herzog and B.M. Nigg (Eds.) *Biomechanics of the musculoskeletal system*, pp.

Nigg, B.M., Cole, G.K. and Bruggemann P. (1995). Impact forces in heel-toe running. *Journal of Applied Biomechanics* **11**, 407-432.

Nilsson, J., Thorstensson, A and Halbertsma, J. (1985). Changes in leg movements and muscle activity with speed of locomotion and mode of progression in humans.

Acta Physiologica Scandinavica 123, 457-475.

Nordin, M. and Frankel, V.H. (1980). Biomechanics of collagenous tissues. In V.H. Frankel and Nordin, M. (Eds.) *Basic Biomechanics of the Skeletal System.* Lea and Febiger, Philadelphia.

Oppenheim, A.V., Willsky, A.S. and Young, I.T. (1983). *Signals and Sytems.* Englewood Cliffs, New Jersey: Prentice-Hall.

Pagliano, J. (1987a). Management of the pronated and supinated foot. *Medicine and Sport Science* **23**, 155-160.

Pagliano, J. (1987b). Soft tissue injuries of the foot tendinous lesions. *Medicine and Sport Science* **23**, 161-168.

Paul, I.L., Munro, M.B., Abernethy, P.J., Simon, S.R., Radin, E.L, Rose, R.M. (1978). Musculo-skeletal shock absorption: relative contribution of bone and soft tissues at various frequencies. *Journal of Biomechanics* **11**, 237-239.

Pezzack, J.C., Norman, R.W. and Winter, D.A. (1977). An assessment of derivative determining techniques used for motion analysis. *Journal of Biomechanics* **10**, 377-382.

Platt, D., Wilson, A.M., Timbs, A., Wright, I.M. and Goodship, A.E. (1994). Novel force transducer for the measurement of tendon force in vivo. *Journal of Biomechanics* **27**, 1489-1493.

Procter, P. and Paul, J.P. (1982). Ankle joint biomechanics. *Journal of Biomechanics* **15**, 627-634.

Proske, U. and Morgan, D.L. (1984). Stiffness of cat soleus muscle and tendon during activation of part of muscle. *Journal of Neurophysiology* **52**, 459-468.

Proske, U. and Morgan, D.L. (1987). Tendon stiffness: methods of measurement and significance for the control of movement. A review. *Journal of Biomechanics* **20**, 75-82.

Puddu, G., Ippolito, E. and Postacchini, F. (1976). A classification of Achilles tendon disease. *American Journal of Sports Medicine* **4**, 146-150.

Rack, P.M.H. and Westbury, D.R. (1984). Elastic properties of the cat soleus tendon and their functional importance. *Journal of Physiology* **347**, 479-495.

Radin, E.L., Orr, R.B., Kelman, J.L., Paul, I.L. and Rose, R.M. (1982). Effect of prolonged walking on concrete on the knees of sheep. *Journal of Biomechanics* **15**, 487-492.

Radin, E.L., Yang, K.H., Riegger, C., Kish, V.L. and O'Conner, J.J. (1991). Relationship between lower limb dynamics and knee joint pain. *Journal of Orthopedic Research* **9**, 398-405.

Reboussin, D.M. and Morgan, T.M. (1996). Statistical considerations in the use and analysis of single-subject designs. *Medicine and Science in Sports and Exercise* **28**, 639-644.

Reinsch, C.H. (1967). Smoothing by spline functions. *Numerische Mathematik* **10**, 177-183.

Reinschmidt, C. and Nigg, B.M. (1995). Influence of heel height on ankle joint moments in running. *Medicine and Science in Sports and Exercise* **27**, 410-416.

Reuleaux, F. (1875). Theoritische kinematik: Grundzuge einer Theorie des Maschinenwesens. F. Vieweg, Braunschweig; MacMillan, London (1963). The Kinematics of Machinery: Outline of a theory of Machines. (Translated by Kennedy, A.B.W.), pp. 56-70. Dover, New York. Cited by Rugg et al. (1990).

Rigby, B.J., Hirai, N., Spikes, J.D. and Eyring, H. (1959). The mechanical properties of rat tail tendon. Journal of Physiology **43**, 59-86.

Root, M.L., Orien, W.P. and Weed, J.H. (1971). *Normal and Abnormal Function of the Foot.* Los Angeles, CA: Clinical Biomechanics Corporation. Cited in Cornwall and McPoil and Cornwall (1995).

Rugg, S.G., Gregor, R.J., Mandelbaum, B.R. and Chius, L. (1990). In vivo moment arm calculations at the ankle joint using magnetic resonance imaging (MRI). *Journal of Biomechanics* **23**, 495-501.

SATRA (1991). Bulletin. Northants., UK.

Schechtman, H. and Bader, D.L. (1994). Dynamic characterisation of human tendons. *Proceedings of the Institution of Mechanical Engineers,* **208**, 241-245.

Schwerdt, H., Constantinesco, A. and Chambron, J. (1980). Dynamic behaviour of the human tendon in vitro. *Journal of Biomechanics* **13**, 913-922.

Scott, S.H. and Winter, D.A. (1990). Internal forces at chronic running injury sites. *Medicine and Science in Sports and Exercise* **22**, 357-369.

Scott, S.H. and Winter, D.A. (1993). Biomechanical model of the human foot: kinematics and kinetics during the stance phase of walking. *Journal of Biomechanics* **26**, 1091-1104.

Seireg, A. and Arvikar, R.J. (1973). A mathematical model for evaluation of forces in lower extremities of the musculo-skeletal system. *Journal of Biomechanics* **6**, 313-326.

Shorten, M.R. (1987). Muscle elasticity in human performance. *Medical Sport Science* **25**, 1-18.

Siegler, S., Chen, J. and Schneck, C.C. (1988). The three-dimensional kinematics of the human ankle and subtalar joints. *Biomechanics* **XI-A**, 417-422.

Sinning, W.E. and Forsyth, H.L. (1970). Lower Limb actions while running at different velocities. Medicine and Science in Sports and Exercise 2, 28-34.

Smart, G.W., Taunton, J.E. and Clement, D.B. (1980). Achilles tendon disorders in runners - a review. *Medicine and Science in Sports and Exercise* **12**, 231-243.

Smith, A.J. (1975). Estimates of muscle and joint forces at the knee and ankle during a jumping activity. *Journal of Human Movement Studies* **1**, 78-86.

Smith, L., Clarke, T., Hamill, C. and Santopietro, F. (1986). The effects of soft and semi-rigid orthoses upon rearfoot movement in running. *Medicine and Science in Sports and Exercise* **18**, p. 171.

Smith, R.E. (1988). The logic and design of case study research. *The Sport Psychologist* **2**, 1-12.

Snel, J.G., Dellman, N.J., Heerkens, Y.F. and van Ingen Schenau, G.J. (1985). Shock absorbing characteristics of running shoes during actual running. In D.A. Winter, R.W. Norman, R.P. Wells, K.C.Hayes, and A.E. Patla (Eds.) *Biomechanics* **IX-B**, pp. 133-138.

Snow, R.E, and Williams, K.R. (1994). High heeled shoes: their effect on centre of mass position, posture, three-dimensional kinematics, rearfoot motion, and ground reaction forces. *Archives of Physical Medicine and Rehabilitation* **75**, 568-576.

Soames, R.W. and Clark, C. (1985). Heel height induced changes in metatarsal loading patterns during gait. In *Biomechanics* **IX-A**, pp. 446-450.

Soutas-Little, R.W., Beavis, G.C., Verstraete, M.C., Markus, T.L. (1987). Analysis of foot motion during running using a joint co-ordinate system. *Medicine and Science in Sports and Exercise* **19**, 285-293.

Spoor, C.W., van Leeuwen, J.l., Meskers, C.G.M., Titulaer, A.F. and Huson, A. (1990). estimation of instantaneous moment arms of lower-leg muscles. *Journal of Biomechanics* **23**, 1247-1259.

Stacoff, A. and Kaelin, X. (1983). Pronation and sportshoe design. In B.M. Nigg and B.A. Kerr (Eds.) *Biomechanical aspects of sport shoes and playing surfaces*, 143-151. Calgary, AB: University of Calgary.

Stacoff, A., Denoth, J., Kaelin, X. and Stuessi, E. (1988). Running injuries and shoe construction: some possible relationships. *International Journal of Sport Biomechanics* **4**, 342-357.

Stacoff, A., Reinschmidt, C. and Stussi, E. (1992). The movement of the heel within a running shoe. *Medicine and Science in Sports and Exercise* **24**, 695-701.

Stijnen, V.V., Williams, E.J., Spaepen, A.J., Peeraer, L. and van Leemputte, M. (). A modified release method for measuring the moment of inertia of the limbs.

Subotnick, S.I. (1979). *The Running Foot Doctor.* World Publications. Mt. View, Canada.

Subotnick, S.I. and Sisney, P. (1986). Treatment of Achilles tendinopathy in the athlete. *Journal of the American Podiatric Medical Association* **76**, 552-557.

Svensson, O.K. (1987). *On quantification of muscular load during standing work.* Unpublished Ph.D. Dissertation.

Tan, J., Kerwin, D.G and Yeadon, M.R. (1995). Evaluation of the Apex video digitisation system. In J. Watkins (Ed.) *Proceedings of the Biomechanics Section of the British Association of Sport and Exercise Sciences* **20**, 21-24.

Teeple, J.B. (1968). A biomechanical analysis of running patterns of college women. Unpublished master's thesis, The Pennsylvania State University, University Park. Cited by Milliron and Cavanagh (1990).

Therrien, R.G., Cote, S., de al Durantaye, M. and Trudeau, F. (1982). Influence of footwear and running technique on te dynamics of foot-ground contact in jogging. Proceedings of *International Symposium of Biomechanics of Sports*, 355-359.

Tomaro, J.E. and Butterfield, S.L. (1995). Biomechanical treatment of traumatic foot and ankle injuries with the use of foot orthotics. *Journal of Orthopaedic Sports Physical Therapy* **21**, 373-380.

Vaughan, C.L. (1982). Smoothing and differentiation of displacement-time data: An application of splines and digital filtering. *International Journal of Biomedical Computing* **13**, 375-385.

Vaughan, C.L. (1984). Biomechanics of running gait. *CRC Critical Reviews of Biomedical*

Engineering **12**, 1-48.

Vincent, W.J. (1994). Analysis of variance with repeated measures. In *Statistics in Kinesiology* pp. 167-183.

Walmsley, B. and Proske, U. (1981). Comparison of stiffness of soleus and medial gastrocnemius muscles in cats. *Journal of Neurophysiology* **46**, 250-259.

Weib, W and Weib, A. (1994). Determining the number of steps needed for reaching continual long distance running pattern. Abstract, *Second World Congress of Biomechanics, Amsterdam.* L. Blankevoort and J.G.M. Kooloos (Eds), p. 256.

Whitsett, C.E. (1963). Some dynamic response characteristics of weightless man. *AMRL TR-63-18,* Wright-Patterson Air Force Base, Ohio.

Williams, K.R. (1980). *A biomechanical evaluation of distance running efficiency.* Unpublished Ph.D. thesis. The Pennsylvania State University, University Park. Cited by Milliron and Cavanagh (1990).

Williams, K.R. (1985). Biomechanics of running. *Exercise and Sports Science Reviews* **13**, 389-441.

Williams, K.R., Snow, R. and Agruss, C. (1991). Changes in distance running kinematics with fatigue. *International Journal of Sport Biomechanics* **7**, 138-162.

Williams, J.G.P. (1993). Achilles tendon lesions in sport. *Sports Medicine* **16**, 216-220.

Winter, D.A., Sidwell, H.G. and Hobson, D.A. (1974). Measurement and reduction of noise in kinematics of locomotion. *Journal of Biomechanics* **7**, 157-159.

Winter, D.A. (1980). Overall principle of lower limb support during stance phase of gait. *Journal of Biomechanics* **13**, 923-927.

Winter, D.A. (1983). Moments of force and mechanical power in jogging. *Journal of Biomechanics* **16**, 91-97.

Winter, D.A. (1984). Kinematic and kinetic patterns in human gait: variability and compensating effects. *Human Movement Science* **3**, 51-76.

Winter, D.A. (1991). *Biomechanics and Motor Control of Human Movement.* Champaign, IL. Human Kinetics.

Winter, D.A. and Bishop, P.J. (1992). Lower extremity injury. *Sports Medicine* **14**, 149-156.

Wold, S. (1974). Spline functions in data analysis. *Technometrics* **16**, 1-11.

Woltring, H.J. (1986). A Fortran package for generalised, cross-validatory spline smoothing and differentiation. *Advances in Engineering Software* **8**, 104-113.

Woltring, H.J. (1992) One hundred years of photogrammetry in biolocomotion. In Cappozzo, A., Marchetti, M. and Tosi, V. (Eds.) *Biolocomotion: a century of research using moving pictures* 199-225.

Wood, G.A. (1975). Computer models for human motion analysis. *Australian Journal of Health, Physical Education, and Recreation* **76**, 35-42.

Wood, G.A. and Jennings, L.S. (1979). On the use of spline functions for data smoothing. *Journal of Biomechanics* **12**, 477-479.

Wood, G.A. (1982). Data smoothing and differentiation procedures in biomechanics. *Exercise and Sport Sciences Reviews* **10**, 308-357.

Yamada, H. (1970). *Strength of Biological Materials.* Robert E. Krieger Publishing Company, Huntington, New York.

Yeadon, M.R. (1981). Twisting Somersaults. *Unpublished Ph.D. Thesis.* Loughborough University.

Yeadon, M.R. (1989). The simulation of aerial movement - II. A mathematical inertia model of the human body. *Journal of Biomechanics* **23**, 67-74.

Yeadon, M.R. and Challis, J.H. (1994). The future of performance-related sports biomechanics research. *Journal of Biomechanics* **12**, 3-32.

Zaichkowsky, L.D. (1980). Single case experimental designs and sport psychology research. In C.H. Nadeau, W.R. Halliwell, K.M. Newell, and G.C. Roberts (Eds.) *Psychology of motor behaviour and Sport* - 1979, 171-179. Champaign, IL: Human Kinetics.

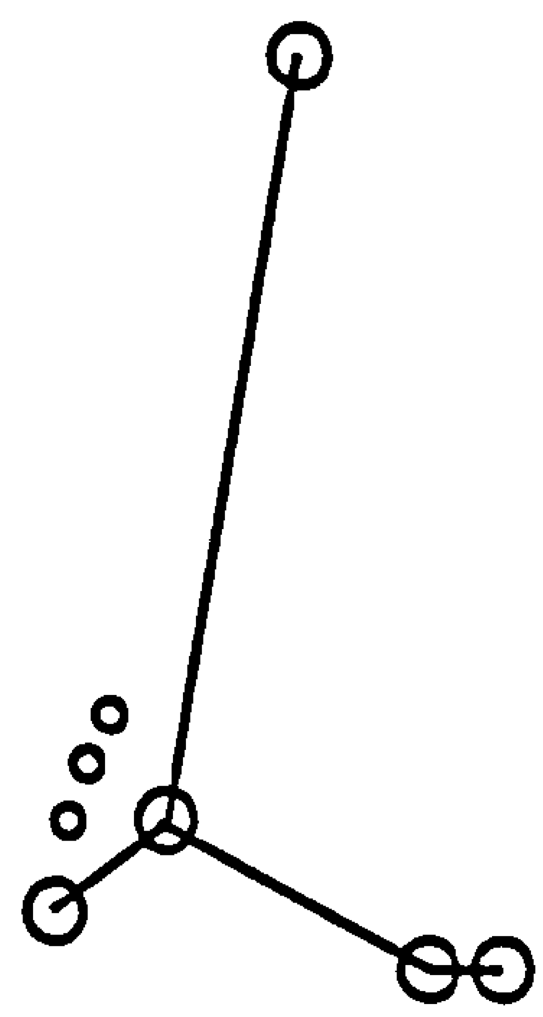
Zajac, F.E. (1989). Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. In *Critical Reviews in Biomedical Engineering* **17**, pp. 359-411.

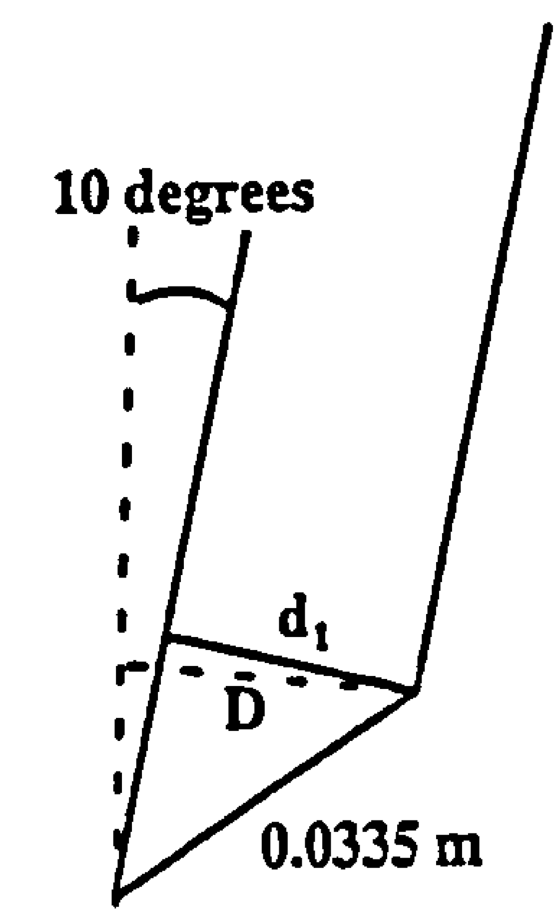
Zatsiorsky, V. and Seluyanov, V. (1985). Estimation of the mass and inertia characteristics of the human body by means of the best predictive regression equations. In *Biomechanics* **IX-B**, pp. 233-239.

Zernicke, R.F., Caldwell, G. and Roberts, E.M. (1976). Fitting biomechanical data with cubic spline functions. *Research Quarterly* **47**, 9-20.

APPENDICES

Appendix A Error if 10 degree variation in tendon line of action (Burdett, 1982)





Sample dimensions:
$d_1 = 0.028$ m
ankle to heel marker (tendon insertion) = 0.0335 m

a = angle between Achilles tendon and line from ankle to Achilles insertion
a_{10} = same angle with 10 degree deviation of Achilles tendon

sin a = 0.0280/0.0335; a=56.7 degrees
a_{10} = (56.7 + 10) = 66.7 degrees
D = 0.0335*sin 66.7; D = 0.0308
difference in Achilles tendon moment arm length = 2.8 mm

Appendix B Sample questionnaire

ACHILLES TENDON INJURY

Sex: male / female   Age: _____

(1) What is the highest representative level you have competed at ?
club / county / area / international

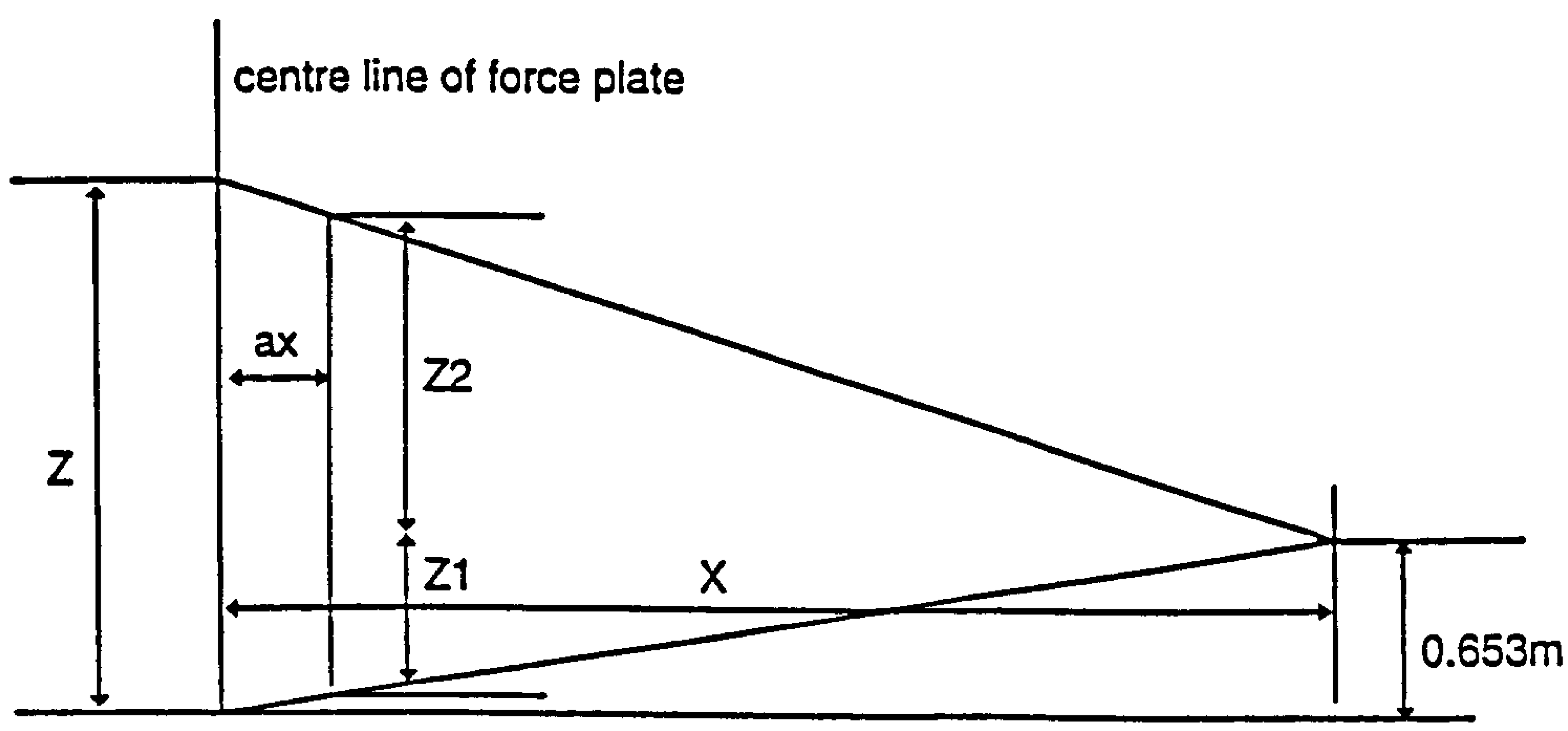
(2) Have you ever had an Achilles tendon injury ?   yes / no

(3) If yes to (2), did this injury cause you to miss one week or more of your usual training ?   yes / no

(4) Did you have this injury within the last three years ?   yes / no

Appendix C Correction of 2D data using centre of pressure location

Side view:



height of camera = 0.653m
Z = height of calibration plane
X = horizontal distance from camera lens to calibration plane
ax = x coordinate of centre of pressure
Z1, Z2 define vertical co-ordinate of required point

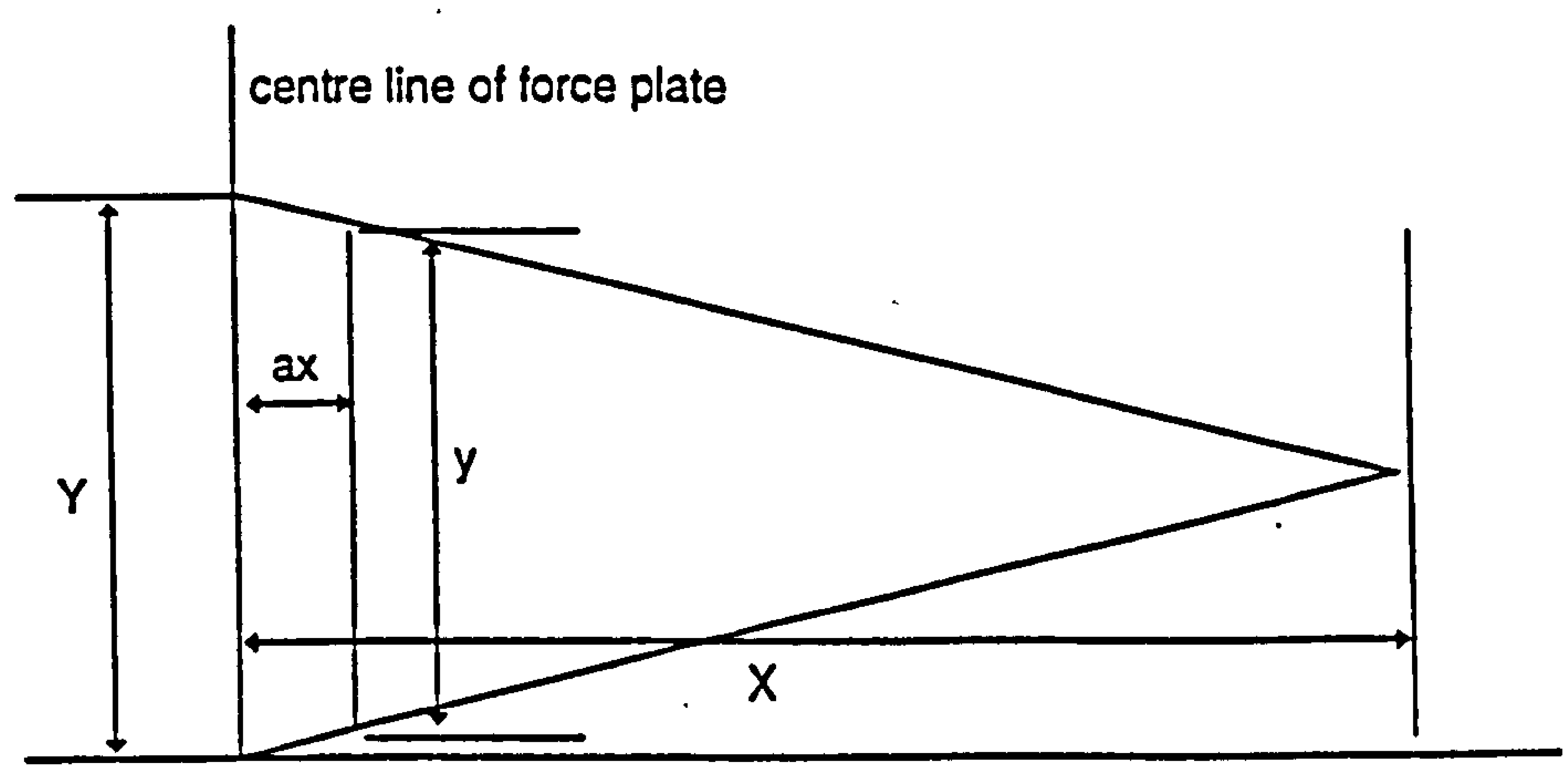
$$X / 0.653 = (X-ax) / Z1$$

$$Z1 = 0.653.(X-ax) / X$$

$$X / (Z-0.653) = (X-ax) / Z2$$

$$Z2 = (Z-0.653).(X-ax) / X$$

Plan view:



Y = width of calibration plane

X = horizontal distance from camera lens to calibration plane

ax = X coordinate of centre of pressure

y = horizontal distance in plane of centre of pressure

$$2X / Y = (2X - 2ax) / y$$

$$y = Y.(X-ax) / X$$

Co-ordinates (X,Y) corrected for centre of pressure location = (x,y)

Appendix D BASIC program for ankle moment and Achilles tendon force calculation

Appendix: Program for calculation of quasi-static moments and Achilles tendon forces

```
10REM program to read force and video data from spooled data files
20REM synchronise and calculate quasi-static ankle moments
30REM and Achilles tendon forces
40REM
50REM OSCLI"URD :0" (ALL DATA ON HARD DISC FOR TEST RUNS)
60REM
70REM dimensions all defined
80REM
90DIM Fz(51):DIM Fy(51):DIM Fx(51):DIM ay(51):DIM ax(51)
100DIM Y(51,12):DIM Z(51,12):DIM d(51,5):DIM S$(51):DIM T(51)
110DIM dA(51):DIM FA(51):DIM Fjt(51,10)
120DIM Th(51):DIM Al(51):DIM Fljt(51,10):DIM Fsjt(51,10)
130DIM Q(51,10):DIM Mjt(51,10):DIM R(51,10):DIM P(51,10)
140DIM FA2(51):DIM d2(51):DIM G(51)
150DIM FA3(51):DIM d3(51):DIM GRAD(51):DIM Yint(51):DIM Zint(51)
160DIM A(51):DIM C(51):DIM U(51)
170DIM d4(51):DIM FA4(51):DIM GRAD2(51):DIM Yint2(51):DIM Zint2(51)
180DIM d5(51):DIM FA5(51):DIM GRAD5(51):DIM Yint5(51):DIM Zint5(51)
190REM
200PROCloadforcedata
210PROCloaddigitdata
220REM PROCscaling
230PROCdistcalc
240PROCmommtp
250PROCachilles
260PROCachilles2
270REM PROCachilles3
280PROCachilles4
290PROCachilles5
300REMPROCmean_d
310PROCdatasummary
320REMPROCscaldata
330OSCLI"DIR IDEFS::IDEDisc4.$.BASIC.ACH"
340END
```

```
350DEFPROCloaddigitdata
360REM read spooled digitised data file
370REM OSCLI"DIR adfs::Foot.$"
380REM OSCLI"*."
390OSCLI"DIR IDEFS::IDEDisc4.$.BASIC.ACH.DLT2"
400INPUT "filename"; f$
410in%=OPENIN(f$)
420REPEAT:in$=GET$#in%:UNTIL in$<>CHR$(0)
430F=VAL(MID$(in$,10,3)):P=VAL(MID$(in$,24,2))
440REPEAT:in$=GET$#in%:UNTIL in$<>CHR$(0)
450REM read co-ords of each joint centre from DLT2D file
460REM
470FOR N=1 TO F
480 FOR A = 1 TO P
490REPEAT:in$=GET$#in%:UNTIL in$<>CHR$(0)
500 Y(N,A)=VAL(MID$(in$,1,8))*((5.91-ax(N))/5.91)
510 Z(N,A)=VAL(MID$(in$,12,8))*((5.91-ax(N))/5.91)
520PRINT Y(N,A)
530REPEAT:in$=GET$#in%:UNTIL in$<>CHR$(0)
540 REM
550 NEXT A
560REPEAT:in$=GET$#in%:UNTIL in$<>CHR$(0)
570REPEAT:in$=GET$#in%:UNTIL in$<>CHR$(0)
580NEXT N
590PRINT F ; P
600CLOSE#in%
610ENDPROC
620DEFPROCscaling
630REM change origin of co-ords. and scale
640Mv = 0.000854
650Mh = 0.000825
660FOR N=Nst TO Nend
670FOR A=3 TO 10
680Y(N,A) = Mh*(Y(N,A) - Y(0,4))*((5.91 - ax(N))/5.91)
690Z(N,A) = Mv*(Z(N,A) - Z(0,4) + (0.0295/Mv))*((5.91 - ax(N))/5.91)
700REM
710NEXT A
720NEXT N
730FOR A=3 TO 10
740Y(0,A) = Mh*(Y(0,A) - Y(0,4))*((5.91 - ax(N))/5.91)
750Z(0,A) = Mv*(Z(0,A) - Z(0,4) + (0.0295/Mv))*((5.91 - ax(N))/5.91)
```

```
760Y(51,A) = Mh*(Y(51,A) - Y(0,4))*((5.91 - ax(N))/5.91)
770Z(51,A) = Mv*(Z(51,A) - Z(0,4) + (0.0295/Mv))*((5.91 - ax(N))/5.91)
780NEXT A
790REM above takes ax position of CP into account
800REM
810
820ENDPROC
821REM
880 DEFPROCloadforcedata
890 REM read data from force file
900REM OSCLI"DIR adfs::BASIC1.$"
910OSCLI"DIR IDEFS::IDEDisc4.$.BASIC.ACH.FORCE"
920INPUT "force data filename"; f2$
930in2%=OPENIN(f2$)
940REM
950REPEAT:in2$=GET$#in2%:UNTIL in2$<>CHR$(0)
960REM PRINTin2$
970REPEAT:in2$=GET$#in2%:UNTIL in2$<>CHR$(0)
980REPEAT:in2$=GET$#in2%:UNTIL in2$<>CHR$(0)
990REM PRINTin2$
1000N=0:REM field no. referring to force data
1010REM each line of force data read until vertical force (Fz) is
1020REM greater than 50N.
1030REM
1040N=N + 1
1050REM
1060REM
1070REPEAT:in2$=GET$#in2%:UNTIL in2$<>CHR$(0)
1080REPEAT:in2$=GET$#in2%:UNTIL in2$<>CHR$(0)
1090REM PRINT in2$
1100Fz(N)=VAL(MID$(in2$,21,8))
1110Fx(N)=VAL(MID$(in2$,42,7))
1120Fy(N)=VAL(MID$(in2$,31,8))
1130ay(N)=VAL(MID$(in2$,63,6))
1140ax(N)=VAL(MID$(in2$,54,6))
1150T(N)=VAL(MID$(in2$,15,5))
1160REM
1170IF Fz(N)<50 THEN GOTO 1040
1180REM
1190REM read data for fields in contact with the force plate
1200Fz(1)=Fz(N):Fx(1)=Fx(N):Fy(1)=Fy(N):ay(1)=ay(N):ax(1)=ax(N)
```

```
1210N=1
1220N=N+1
1230REPEAT:in2$=GET$#in2%:UNTIL in2$<>CHR$(0)
1240REPEAT:in2$=GET$#in2%:UNTIL in2$<>CHR$(0)
1250Fz(N)=VAL(MID$(in2$,21,4)):Fy(N)=VAL(MID$(in2$,31,8))
1260Fx(N)=VAL(MID$(in2$,42,7))
1270ay(N)=VAL(MID$(in2$,63,6))
1280ax(N)=VAL(MID$(in2$,54,6))
1290REM
1300REM
1310REM continue until vertical force less than 50N.
1320IF Fz(N)>=50 THEN GOTO 1220
1330REMNend=N-1
1340ENDPROC
1350 DEFPROCdistcalc
1360REM distance calculations:
1370REM temp. scaling
1380REMtb = ((Y(0,7)-Y(0,5))^2 + (Z(0,7)-Z(0,5))^2)^0.5
1390REMM = (0.602-0.0305)/tb
1400REM calculation of segment lengths
1410REM joint centre to joint centre
1420FOR N = 1 TO F
1430REM PRINT "FIELD NO.="; N
1440 d(N,1)=(((Z(N,1)-Z(N,2))^2)+((Y(N,1)-Y(N,2))^2))^0.5
1450 REM ANKLE TO HEEL
1460 d(N,2)=(((Z(N,2)-Z(N,3))^2)+((Y(N,2)-Y(N,3))^2))^0.5
1470 REM ANKLE TO MTP
1480 d(N,3)=(((Z(N,2)-Z(N,4))^2)+((Y(N,2)-Y(N,4))^2))^0.5
1490 REM MTP TO TOE
1500 d(N,4)=(((Z(N,5)-Z(N,4))^2)+((Y(N,5)-Y(N,4))^2))^0.5
1510 REM KNEE TO HEEL
1520 d(N,5)=(((Z(N,1)-Z(N,3))^2)+((Y(N,1)-Y(N,3))^2))^0.5
1530REM PRINT "distance from knee to ankle:";d(N,1)
1540REM PRINT "distance from ankle to heel:";d(N,2)
1550REM PRINT "distance from ankle to mtp:";d(N,3)
1560REM PRINT "distance from mtp to toe:";d(N,4)
1570REM PRINT "distance from knee to heel:";d(N,5)
1580 NEXT N
1590REM
1600ENDPROC
1610REM
```

```
1620DEFPROCdatasummary
1630REM
1640OSCLI"DIR IDEFS::IDEDisc4.$.BASIC.ACH.OUTPUT"
1650REM
1660file$=LEFT$(f$,6)
1670OP$=file$ +"_OP2"
1680CLS:PRINT"Spooling data to file "; OP$
1690OSCLI"SPOOL "+OP$
1700PRINT"FILE: ";OP$
1710PRINT"Field No.(N)   ay(m)   Fz(N)   Fy(N)  MTP(Nm) Ankle(Nm) Knee Moment"
1720FOR N=1 TO F
1720PRINTN;TAB(16)ay(N);TAB(25)Fz(N);TAB(33)Fy(N);TAB(39)Mjt(N,4);TAB(48)Mj-
t(N,2);TAB(58)Mjt(N,1)
1740NEXT N
1750PRINT
1760PRINT"Field No.(N) Achilles1(N)  Achilles2(N)  Achilles4(N)  Achilles5(N)"
1770FOR N=1 TO F
PRINT N;TAB(15)FA(N);TAB(30)FA2(N);TAB(42)FA4(N);TAB(58)FA5(N)
1790NEXT N
1800PRINT
1810PRINT"Field No.(N) Achilles1(BW)  Achilles2(BW)  Achilles4(BW)  Achilles5(BW)"
1820BW=540
1830FOR N=1 TO F
1840PRINTN;TAB(15)FA(N)/BW;TAB(30)FA2(N)/BW;TAB(42)FA4(N)/
BW;TAB(60)FA5(N)/BW
1850NEXT N
1860PRINT
1870REM print mean force throughout contact
1880PRINT"MEAN";TAB(15)MEANF/BW;TAB(30)MEANF2/BW;TAB(42)MEANF4/
BW;TAB(60)MEANF5/BW
1890PRINT
1900PRINT "LEG LENGTH    MOM ARM1    MOM ARM2    MOM ARM4    MOM
ARM5"
1910PRINT
1920FOR N=1 TO F
1930PRINT d(N,3);TAB(15)dA(N);TAB(30)d2(N);TAB(42)d4(N);TAB(57)d5(N)
1940NEXT N
1950PRINT
1960REMPRINT"MEAN";TAB(15)MEANdA;TAB(30)MEANd2;TAB(42)MEANd4;TAB-
(57)MEANd5
1970
```

```
1980OSCLI"SPOOL "
1990REMOSCLI"SETTYPE "OP$" &FFD"
2000OSCLI"DIR &"
2010ENDPROC
2020REM
2030
2040
2050DEFPROCmommtp
2060REM temp scaling
2070REM Moments about joint centres
2080G=0
2090REM G = gradient of plate in digitised field
2100REMG=(Z(0,4)-Z(0,3))/(Y(0,4)-Y(0,3))
2110A=ATN(G)
2120REM one point on plate:
2130REM
2140REM YP1=Y(0,3) + (0.0295)*SINA
2150REM ZP1=Z(0,3) - (0.0295)*COSA
2160REM equation of line representing plate:
2170REM Z = G*Y
2180REM horizontal distance from centre of force plate to joint centre (Q)
2190REM YC,ZC rep. centre of plate
2200REM YC=0
2210REM ZC=0
2220REM
2230
2240REM calculation of point of intersection with plate of line
2250REM normal to plate through joint centre
2260REM
2270FOR N=1 TO F
2280FOR A=1 TO P
2290REM YI=(1/((G^2)+1))*(Y(N,A) + G*(Z(N,A)))
2300REM ZI=G*(YI)
2310 REM
2320 REM P = normal distance from joint centre to plate
2330 REMP(N,A)=(((YI-Y(N,A))^2 + (ZI-Z(N,A))^2)^0.5)
2340 REM Q = horizontal distance from plate centre to (YI,ZI)
2350 REMQ(N,A) = (((YI)^2 + (ZI)^2)^0.5)
2360 REM consideration of direction of Q
2370 REMIF YI<0 THEN
2380 REM Q(N,A) = -Q(N,A)
```

```
2390 ENDIF
2400 REM moment about joint centre
2410 REMMjt(N,A)=(ay(N)-Q(N,A))*Fz(N) + (P(N,A)*Fy(N))
2420 REM
2430 REM
2440 REM
2450 Mjt(N,A)=(ay(N)-Y(N,A))*Fz(N) + (Z(N,A)*Fy(N))
2460 REM
2470 NEXT A
2480NEXT N
2490ENDPROC
2500REM calculation of Achilles tendon force (FA)
2510DEFPROCachilles
2520FOR N=1 TO F
2530REM dA(N)= perpendicular distance from lower leg to Achilles line
        2540dA(N)=(((Z(N,1)-Z(N,2))/d(N,1))*(Y(N,2)-Y(N,3)-(((Y(N,1)-Y(N,2))/(Z(N,1)-
Z(N,2)))*(Z(N,2)-Z(N,3)))))
2550REM consideration of moments at joint 2
2560FA(N)=Mjt(N,2)/dA(N)
2570REM calculation of bone-to-bone force at ankle joint
2580REM angle between lower leg and horizontal (Th)
2590Th(N)=ATN((Z(N,1)-Z(N,2))/d(N,1))
2600REM angle between resultant external force and horizontal (Al)
2610
2620REMAl(N)=ATN(Fz(N)/Fy(N))
2630REM force component along longitudinal axis of lower leg (Fljt)
2640REMlong=(((Fz(N))^2+(Fy(N))^2)^0.5)*(COS(PI-(Th(N)-Al(N))))
2650REM modulus of longitudinal component
2660REMLONG=(long^2)^0.5
2670REMFljt(N,4)=LONG - FA(N)
2680REM
2690REMFsjt(N,4)=(((Fz(N))^2+(Fy(N))^2)^0.5)*((SIN(PI-(Th(N)-Al(N)))))
2700REMFjt(N,4)=((Fljt(N,4))^2+(Fsjt(N,4))^2)^0.5
2710NEXT N
2720REM calc of mean force throughout contact
2730LET S=0
2740FOR N=1 TO F
2750S=S + FA(N)
2760NEXT N
2770MEANF=S/(F)
2780ENDPROC
```

```
2790DEFPROCachilles2
2800FOR N=1 TO F
2810G(N)=ACS(((d(N,5))^2+(d(N,1))^2-(d(N,2))^2)/(2*d(N,5)*d(N,1)))
2820d2(N)=(SIN(G(N))*d(N,1))
2830FA2(N)=Mjt(N,2)/d2(N)
2840NEXT N
2850REM calc of mean force throughout contact
2860LET S2=0
2870FOR N=1 TO F
2880S2=S2 + FA2(N)
2890NEXT N
2900MEANF2=S2/(F)
2910ENDPROC
2920DEFPROCachilles3
2930REM
2940FOR N=1 TO F
2950C(N)=ACS(((d(N,3))^2+(329.6^2)-(113.2^2))/(2*d(N,3)*329.6))
2960U(N)=ATN((Z(N,2)-Z(N,4))/(Y(N,4)-Y(N,2)))
2970A(N)=U(N)-C(N)
2980e=(Y(N,4)-Y(N,2))/COSA(N)
2990f=((Y(N,4)-Y(N,2))/e)*330
3000g=f*TANA(N)
3010Y(N,11)=Y(N,4)-f
3020Z(N,11)=Z(N,4)+g
3030Y(N,12)=Y(N,1)-88
3040Z(N,12)=Z(N,1)
3050GRAD(N)=(Z(N,12)-Z(N,11))/(Y(N,12)-Y(N,11))
3060Yint(N)=(Y(N,4)+Y(N,11)*((GRAD(N))^2)+(GRAD(N)*Z(N,4))-(GRAD(N)*Z(N,11)))/((GRAD(N)^2)+1)
3070Zint(N)=(Yint(N)*GRAD(N))-(Y(N,11)*GRAD(N))+Z(N,11)
3080d3(N)=(((Y(N,4)-Yint(N))^2+(Z(N,4)-Zint(N))^2)^0.5)
3090FA3(N)=(Mjt(N,4))/d3(N)
3100NEXT N
3110REM calc of mean force throughout contact
3120LET S3=0
3130FOR N=1 TO F
3140S3=S3 + FA3(N)
3150NEXT N
3160MEANF3=S3/(F)
3170ENDPROC
3180REM
```

```
3190DEFPROCachilles4
3200FOR N=1 TO F
3210IF Y(N,6)=Y(N,8) THEN
3220d4(N) = Y(N,2) - Y(N,6)
3230REM
3240ELSE
3250GRAD2(N)=(Z(N,6)-Z(N,8))/(Y(N,6)-Y(N,8))
3260Yint2(N)=(Y(N,2)+Y(N,8)*((GRAD2(N))^2)+(GRAD2(N)*Z(N,2))-(GRAD2(N)*Z(N,8)))/((GRAD2(N)^2)+1)
3270Zint2(N)=(Yint2(N)*GRAD2(N))-(Y(N,8)*GRAD2(N))+Z(N,8)
3280d4(N)=(((Y(N,2)-Yint2(N))^2+(Z(N,2)-Zint2(N))^2)^0.5)
3290ENDIF
3300FA4(N)=(Mjt(N,2))/d4(N)
3310NEXT N
3320REM calc of mean force throughout contact
3330LET S4=0
3340FOR N=1 TO F
3350S4=S4 + FA4(N)
3360NEXT N
3370MEANF4=S4/(F)
3380ENDPROC
3390REM
3400DEFPROCachilles5
3410FOR N = 1 TO F
3420Y=0.26*(Y(N,2)-Y(N,3))
3430Z=0.26*(Z(N,2)-Z(N,3))
3440Y(N,9)=Y(N,3)+Y
3450Z(N,9)=Z(N,3)+Z
3460Y(N,10)=Y(N,1)- 0.04667
3470Z(N,10)=Z(N,1)
3480GRAD5(N)=(Z(N,10)-Z(N,9))/(Y(N,10)-Y(N,9))
3490Yint5(N)=(Y(N,2)+Y(N,9)*((GRAD5(N))^2)+(GRAD5(N)*Z(N,2))-(GRAD5(N)*Z(N,9)))/((GRAD5(N)^2)+1)
3500Zint5(N)=(Yint5(N)*GRAD5(N))-(Y(N,9)*GRAD5(N))+Z(N,9)
3510d5(N)=(((Y(N,2)-Yint5(N))^2+(Z(N,2)-Zint5(N))^2)^0.5)
3520FA5(N)=(Mjt(N,2))/d5(N)
3530NEXT N
3540REM calc of mean force throughout contact
3550LET S5=0
3560FOR N=1 TO F
3570S5=S5 + FA5(N)
```

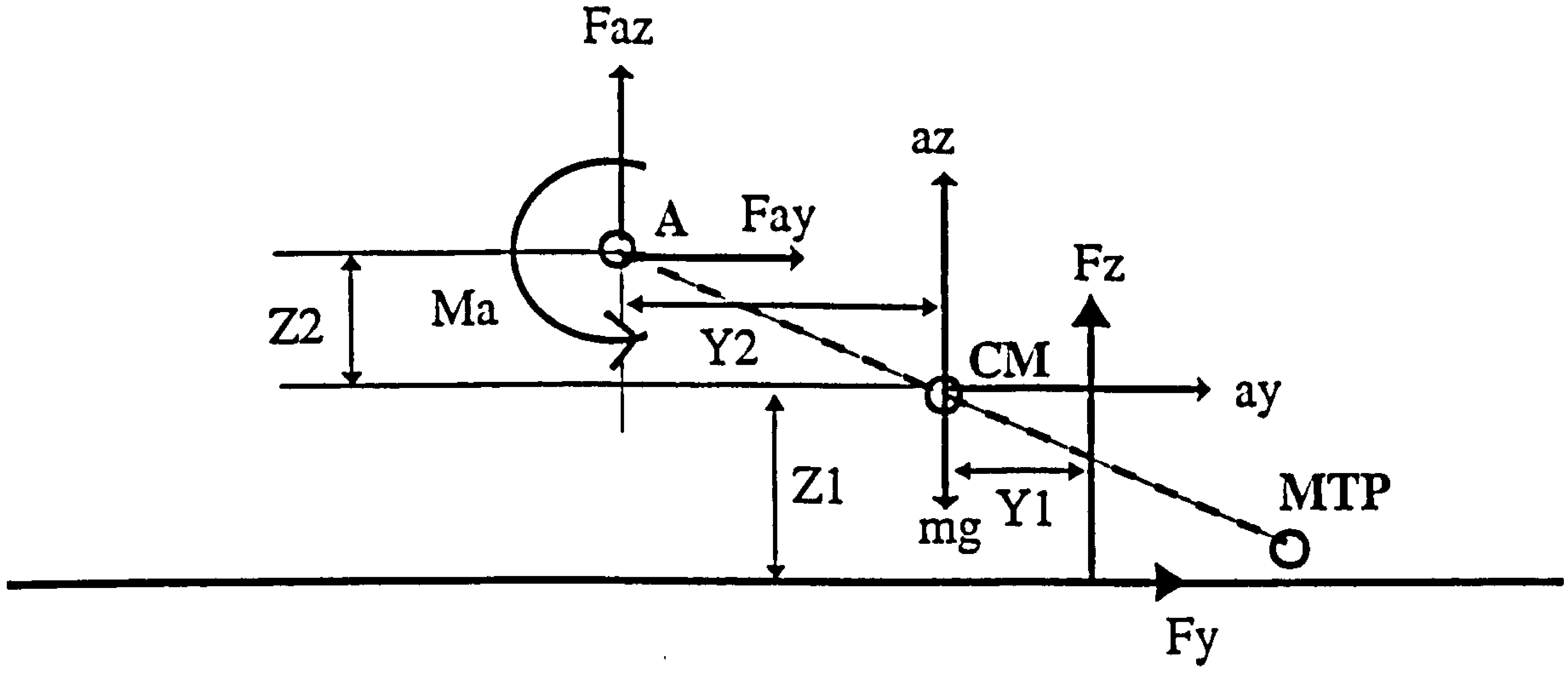
```
3580NEXT N
3590MEANF5=S5/(F)
3600ENDPROC
3610REM
3620DEFPROCscaldata
3630REM
3640OSCLI"DIR IDEFS::IDEDisc4.$.BASIC.ACH.OUTPUT"
3650REM
3660file2$=LEFT$(f$,6)
3670OP$=file2$ +"_SC"
3680CLS:PRINT"Spooling data to file "; OP$
3690OSCLI"SPOOL "+OP$
3700PRINT"FILE: ";OP$
3710FOR N=1 TO F
3720PRINT N
3730FOR A=3 TO 10
3740@%=&00408:PRINT Y(N,A);",”Z(N,A)
3750NEXT A
3760NEXT N
3770@%=10
3780OSCLI"SPOOL "
3790REMOSCLI"SETTYPE "OP$" &FFD"
3800OSCLI"DIR &"
3810ENDPROC
3820REM
3830DEFPROCmean_d
3840FOR N=Nst TO Nend
3850REM
3860REM calc of mean tendon moment arm throughout contact
3870LET ADD=0
3880FOR N=Nst TO Nend
3890ADD=ADD + dA(N)
3900NEXT N
3910MEANdA=ADD/(Nend-Nst+1)
3920LET ADD2=0
3930FOR N=Nst TO Nend
3940ADD2=ADD2 + d2(N)
3950NEXT N
3960MEANd2=ADD2/(Nend-Nst+1)
3970LET ADD4=0
3980FOR N=Nst TO Nend
```

```
3990ADD4=ADD4 + d4(N)
4000NEXT N
4010MEANd4=ADD4/(Nend-Nst+1)
4020LET ADD5=0
4030FOR N=Nst TO Nend
4040ADD5=ADD5 + d5(N)
4050NEXT N
4060MEANd5=ADD5/(Nend-Nst+1)
4070ENDPROC
```

Appendix E ANOVA for data from Section 3.3 (iii)

source of variation	sum of squares	d.f.	variance	F
between groups	30.5	5	6.1	6.9
within groups	44.7	51	0.88	
total	75.2	56		

Appendix F



CM = centre of mass; A = proximal end of segment; MTP = metatarsalphalageal joint

F_{hori}, F_{vert} = resultant force in the horizontal and vertical directions respectively

F_{ay}, F_{az} = joint reaction forces in the horizontal and vertical directions respectively

F_y, F_z = ground reaction forces in the horizontal and vertical directions respectively

m = segment mass

g = acceleration due to gravity

M_a = moment about the ankle joint

a_y, a_z = horizontal and vertical components of acceleration of the segment centre of mass

Y_1, Y_2, Z_1, Z_2 = distances

I = moment of inertia of the segment

α = angular acceleration of the segment

Inverse Dynamics Calculations

$\Sigma F_{hori} = m.a_y$

$F_{ay} + F_y = m.a_y$

$\Sigma F_{vert} = m.a_z$

$F_{az} + F_z - mg = m.a_z$

$\Sigma Moments = I.\alpha$

$M_a + F_y.Z_1 + F_z.Y_1 - F_{ay}.Z_2 - F_{az}.Y_2 = I.\alpha$

Appendix G Template spreadsheet containing a sample set of data

	1	2	3	4	5	6	7	8	9	10	11
1	FIELD	hip			knee			ankle			mtpp
2	100.32	203.1	-754.17	123.43	26.46	-411.77	51.38	-2.77	-15.38	24.06	-110.13
3	100.44	199.79	-753.77	123.46	23	-411.53	51.25	-4.59	-14.62	23.74	-111.68
4	100.56	196.47	-753.39	123.49	19.57	-411.25	51.13	-6.35	-13.85	23.42	-113.17
5	100.68	193.15	-753.02	123.51	16.15	-410.95	51.02	-8.07	-13.08	23.1	-114.6
6	100.8	189.81	-752.67	123.53	12.76	-410.61	50.91	-9.75	-12.29	22.78	-115.98
7	100.92	186.47	-752.32	123.54	9.4	-410.24	50.81	-11.38	-11.51	22.46	-117.3
8	101.04	183.11	-751.99	123.55	6.05	-409.83	50.72	-12.97	-10.72	22.13	-118.56
9	101.16	179.75	-751.67	123.55	2.73	-409.39	50.63	-14.51	-9.94	21.81	-119.76
10	101.28	176.38	-751.36	123.55	-0.58	-408.92	50.55	-16.01	-9.15	21.49	-120.91
11	101.4	173	-751.05	123.54	-3.86	-408.41	50.48	-17.46	-8.38	21.17	-122
12	101.52	169.61	-750.75	123.53	-7.13	-407.87	50.42	-18.86	-7.61	20.85	-123.04
13	101.64	166.21	-750.45	123.52	-10.38	-407.3	50.37	-20.21	-6.86	20.53	-124.03
14	101.76	162.8	-750.15	123.51	-13.62	-406.69	50.33	-21.52	-6.12	20.22	-124.96
15	101.88	159.38	-749.85	123.5	-16.84	-406.06	50.29	-22.77	-5.4	19.9	-125.83
16	102	155.95	-749.54	123.48	-20.05	-405.39	50.27	-23.97	-4.7	19.59	-126.66
17	102.12	152.52	-749.23	123.47	-23.25	-404.7	50.27	-25.11	-4.03	19.28	-127.43
18	102.24	149.08	-748.92	123.47	-26.43	-403.97	50.27	-26.21	-3.38	18.97	-128.15
19	102.36	145.62	-748.59	123.46	-29.61	-403.22	50.29	-27.24	-2.76	18.66	-128.82
20	102.48	142.17	-748.26	123.46	-32.78	-402.45	50.31	-28.23	-2.17	18.35	-129.45
21	102.6	138.7	-747.92	123.47	-35.95	-401.66	50.35	-29.16	-1.61	18.05	-130.02
22	102.72	135.24	-747.57	123.48	-39.11	-400.85	50.4	-30.03	-1.09	17.76	-130.55
23	102.84	131.77	-747.21	123.49	-42.27	-400.03	50.45	-30.86	-0.6	17.47	-131.04
24	102.96	128.29	-746.83	123.52	-45.42	-399.19	50.52	-31.63	-0.15	17.18	-131.48
25	103.08	124.81	-746.44	123.55	-48.58	-398.34	50.6	-32.34	0.27	16.9	-131.88
26	103.2	121.34	-746.04	123.6	-51.74	-397.48	50.68	-33.01	0.64	16.62	-132.24
27	103.32	117.86	-745.63	123.64	-54.89	-396.62	50.77	-33.62	0.97	16.35	-132.56
28	103.44	114.38	-745.2	123.7	-58.05	-395.76	50.86	-34.19	1.27	16.08	-132.85
29	103.56	110.9	-744.77	123.76	-61.2	-394.89	50.96	-34.71	1.53	15.82	-133.1
30	103.68	107.42	-744.32	123.83	-64.35	-394.04	51.05	-35.18	1.75	15.57	-133.32
31	103.8	103.94	-743.86	123.91	-67.5	-393.19	51.15	-35.62	1.93	15.32	-133.52
32	103.92	100.46	-743.38	123.99	-70.65	-392.35	51.24	-36.01	2.08	15.08	-133.68
33	104.04	96.99	-742.9	124.07	-73.79	-391.52	51.33	-36.36	2.19	14.85	-133.83
34	104.16	93.52	-742.41	124.16	-76.93	-390.71	51.42	-36.68	2.27	14.62	-133.95
35	104.28	90.05	-741.91	124.25	-80.07	-389.91	51.49	-36.97	2.32	14.4	-134.05
36	104.4	86.59	-741.41	124.34	-83.2	-389.14	51.56	-37.22	2.34	14.19	-134.13
37	104.52	83.14	-740.9	124.43	-86.34	-388.38	51.61	-37.45	2.33	13.98	-134.2
38	104.64	79.69	-740.38	124.51	-89.47	-387.64	51.66	-37.66	2.31	13.78	-134.25
39	104.76	76.25	-739.87	124.6	-92.61	-386.91	51.69	-37.84	2.26	13.59	-134.3
40	104.88	72.82	-739.35	124.68	-95.75	-386.21	51.71	-38.01	2.2	13.4	-134.33

	12	13	14	15	16	17	18	19	20	21	22
1			ach1			ach2			calc1		
2	33.94	36.29	56.45	-47.17	-6.8	53.21	-7.14	-5.4	53.53	10.97	-7.62
3	35.11	35.86	54.53	-46.69	-7.1	51.44	-6.74	-5.71	51.85	11.35	-7.91
4	36.29	35.43	52.65	-46.2	-7.4	49.72	-6.34	-6.03	50.23	11.71	-8.21
5	37.45	35	50.81	-45.72	-7.7	48.04	-5.94	-6.34	48.67	12.07	-8.5
6	38.6	34.59	49.02	-45.23	-8	46.41	-5.56	-6.65	47.15	12.42	-8.8
7	39.74	34.19	47.28	-44.75	-8.29	44.82	-5.18	-6.96	45.68	12.75	-9.08
8	40.85	33.8	45.57	-44.27	-8.59	43.28	-4.82	-7.28	44.26	13.06	-9.37
9	41.93	33.43	43.91	-43.79	-8.88	41.79	-4.47	-7.6	42.89	13.36	-9.66
10	42.99	33.07	42.29	-43.33	-9.17	40.34	-4.14	-7.92	41.57	13.64	-9.94
11	44.01	32.72	40.72	-42.87	-9.46	38.93	-3.83	-8.23	40.3	13.9	-10.21
12	45	32.39	39.19	-42.43	-9.75	37.58	-3.53	-8.55	39.08	14.14	-10.49
13	45.95	32.08	37.7	-42	-10.05	36.27	-3.25	-8.87	37.9	14.36	-10.76
14	46.85	31.78	36.25	-41.59	-10.34	35.01	-2.99	-9.18	36.78	14.57	-11.02
15	47.71	31.5	34.85	-41.2	-10.64	33.8	-2.74	-9.49	35.7	14.75	-11.29
16	48.51	31.23	33.48	-40.83	-10.94	32.64	-2.52	-9.8	34.67	14.91	-11.55
17	49.27	30.98	32.16	-40.5	-11.25	31.53	-2.32	-10.1	33.69	15.05	-11.8
18	49.97	30.74	30.88	-40.18	-11.56	30.47	-2.15	-10.39	32.76	15.16	-12.05
19	50.62	30.52	29.64	-39.9	-11.87	29.46	-1.99	-10.69	31.87	15.26	-12.3
20	51.21	30.32	28.44	-39.64	-12.18	28.5	-1.86	-10.98	31.03	15.33	-12.54
21	51.76	30.13	27.28	-39.42	-12.5	27.58	-1.76	-11.26	30.23	15.39	-12.78
22	52.25	29.95	26.15	-39.22	-12.81	26.72	-1.68	-11.55	29.48	15.42	-13.02
23	52.69	29.79	25.07	-39.06	-13.12	25.9	-1.62	-11.83	28.77	15.44	-13.25
24	53.08	29.65	24.01	-38.93	-13.42	25.13	-1.59	-12.12	28.11	15.43	-13.48
25	53.43	29.51	23	-38.83	-13.72	24.41	-1.59	-12.4	27.49	15.41	-13.7
26	53.72	29.39	22.01	-38.77	-14.02	23.73	-1.61	-12.69	26.9	15.37	-13.92
27	53.97	29.29	21.06	-38.74	-14.3	23.09	-1.66	-12.97	26.36	15.31	-14.14
28	54.18	29.19	20.14	-38.74	-14.58	22.5	-1.73	-13.25	25.85	15.25	-14.35
29	54.35	29.11	19.25	-38.78	-14.84	21.94	-1.81	-13.52	25.38	15.16	-14.56
30	54.49	29.04	18.39	-38.85	-15.1	21.42	-1.92	-13.78	24.94	15.07	-14.76
31	54.6	28.98	17.55	-38.95	-15.34	20.93	-2.03	-14.04	24.54	14.96	-14.95
32	54.68	28.93	16.74	-39.09	-15.56	20.47	-2.16	-14.28	24.16	14.85	-15.14
33	54.74	28.89	15.95	-39.26	-15.77	20.05	-2.3	-14.51	23.81	14.73	-15.33
34	54.78	28.87	15.19	-39.47	-15.97	19.65	-2.44	-14.73	23.49	14.6	-15.51
35	54.8	28.85	14.45	-39.7	-16.15	19.27	-2.59	-14.93	23.2	14.47	-15.68
36	54.8	28.84	13.73	-39.96	-16.31	18.92	-2.75	-15.12	22.92	14.33	-15.85
37	54.8	28.84	13.03	-40.25	-16.46	18.58	-2.92	-15.29	22.67	14.19	-16.01
38	54.79	28.84	12.34	-40.56	-16.59	18.27	-3.09	-15.45	22.44	14.04	-16.16
39	54.77	28.85	11.68	-40.88	-16.71	17.97	-3.27	-15.59	22.23	13.89	-16.31
40	54.75	28.86	11.03	-41.22	-16.81	17.68	-3.46	-15.71	22.03	13.74	-16.46

	23	24	25	26	27	28	29	30	31	32	33
1	calc2			fp1			fp2			fp3	
2	50.4	34.03	-7.17	216.52	74.6	144.64	212.62	75.01	-175.95	-262.9	72.27
3	48.87	34.41	-7.46	216.54	74.6	144.63	212.64	75.01	-175.95	-262.9	72.27
4	47.4	34.78	-7.74	216.55	74.6	144.63	212.65	75	-175.94	-262.9	72.27
5	45.99	35.14	-8.02	216.56	74.6	144.63	212.67	75	-175.94	-262.9	72.27
6	44.63	35.47	-8.29	216.57	74.6	144.62	212.68	75	-175.94	-262.9	72.27
7	43.32	35.79	-8.55	216.58	74.6	144.62	212.68	74.99	-175.93	-262.9	72.27
8	42.07	36.09	-8.8	216.58	74.6	144.62	212.68	74.99	-175.93	-262.89	72.27
9	40.88	36.37	-9.04	216.58	74.6	144.62	212.67	74.98	-175.93	-262.89	72.27
10	39.74	36.63	-9.27	216.57	74.6	144.62	212.66	74.98	-175.92	-262.89	72.27
11	38.65	36.86	-9.48	216.57	74.6	144.62	212.64	74.98	-175.92	-262.89	72.27
12	37.61	37.07	-9.69	216.56	74.6	144.61	212.61	74.97	-175.92	-262.88	72.27
13	36.62	37.26	-9.89	216.55	74.6	144.61	212.59	74.97	-175.92	-262.88	72.28
14	35.68	37.43	-10.09	216.54	74.6	144.61	212.57	74.97	-175.91	-262.88	72.28
15	34.79	37.56	-10.27	216.54	74.6	144.61	212.56	74.96	-175.91	-262.87	72.28
16	33.95	37.68	-10.45	216.54	74.6	144.61	212.55	74.96	-175.91	-262.87	72.28
17	33.16	37.76	-10.62	216.55	74.6	144.61	212.55	74.95	-175.9	-262.86	72.28
18	32.42	37.83	-10.79	216.56	74.6	144.62	212.56	74.94	-175.9	-262.86	72.28
19	31.72	37.87	-10.96	216.58	74.6	144.62	212.57	74.94	-175.89	-262.85	72.28
20	31.06	37.88	-11.12	216.6	74.6	144.62	212.58	74.93	-175.89	-262.85	72.28
21	30.45	37.88	-11.28	216.61	74.6	144.62	212.59	74.92	-175.89	-262.84	72.28
22	29.88	37.86	-11.44	216.63	74.6	144.62	212.61	74.92	-175.88	-262.84	72.29
23	29.35	37.82	-11.59	216.65	74.6	144.63	212.61	74.91	-175.88	-262.83	72.29
24	28.86	37.76	-11.75	216.66	74.6	144.63	212.62	74.9	-175.87	-262.83	72.29
25	28.4	37.69	-11.91	216.66	74.6	144.63	212.61	74.9	-175.86	-262.83	72.29
26	27.98	37.61	-12.07	216.66	74.6	144.64	212.6	74.89	-175.86	-262.82	72.29
27	27.6	37.51	-12.23	216.66	74.6	144.65	212.59	74.89	-175.85	-262.82	72.29
28	27.25	37.4	-12.39	216.65	74.6	144.65	212.57	74.88	-175.85	-262.82	72.3
29	26.93	37.29	-12.54	216.64	74.6	144.66	212.56	74.88	-175.84	-262.82	72.3
30	26.64	37.18	-12.7	216.63	74.6	144.67	212.54	74.87	-175.83	-262.82	72.3
31	26.37	37.05	-12.85	216.63	74.6	144.67	212.53	74.87	-175.82	-262.82	72.3
32	26.13	36.93	-13	216.62	74.6	144.68	212.53	74.87	-175.81	-262.83	72.3
33	25.92	36.81	-13.14	216.62	74.6	144.69	212.53	74.86	-175.8	-262.83	72.3
34	25.72	36.7	-13.28	216.63	74.6	144.7	212.55	74.86	-175.79	-262.84	72.3
35	25.55	36.58	-13.42	216.64	74.6	144.71	212.57	74.86	-175.78	-262.84	72.3
36	25.39	36.47	-13.55	216.65	74.6	144.72	212.59	74.86	-175.77	-262.85	72.3
37	25.25	36.36	-13.68	216.66	74.6	144.73	212.62	74.86	-175.76	-262.85	72.3
38	25.13	36.25	-13.8	216.67	74.6	144.74	212.65	74.86	-175.75	-262.86	72.3
39	25.02	36.14	-13.92	216.68	74.6	144.75	212.67	74.87	-175.73	-262.86	72.3
40	24.92	36.04	-14.05	216.68	74.6	144.76	212.69	74.87	-175.72	-262.87	72.29

	34	35	36	37	38	39
1		move_x	move_y	move_z	hip	
2	150.58	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
3	150.6	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
4	150.61	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
5	150.63	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
6	150.64	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
7	150.65	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
8	150.65	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
9	150.66	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
10	150.65	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
11	150.65	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
12	150.64	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
13	150.64	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
14	150.63	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
15	150.63	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
16	150.63	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
17	150.64	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
18	150.64	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
19	150.65	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
20	150.66	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
21	150.67	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
22	150.68	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
23	150.69	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
24	150.7	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
25	150.7	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
26	150.69	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
27	150.69	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
28	150.68	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
29	150.68	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
30	150.67	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
31	150.67	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
32	150.67	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
33	150.68	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
34	150.69	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
35	150.7	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
36	150.72	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
37	150.74	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
38	150.75	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
39	150.77	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36
40	150.78	=-0.5*(C34+C31)	=(0.5*(C32+C29))	=(0.5*(C33+C30))	=C4+C35	=-C2+C36

	40	41	42	43	44	45	46
1		knee			ankle		
2	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
3	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
4	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
5	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
6	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
7	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
8	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
9	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
10	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
11	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
12	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
13	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
14	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
15	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
16	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
17	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
18	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
19	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
20	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
21	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
22	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
23	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
24	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
25	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
26	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
27	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
28	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
29	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
30	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
31	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
32	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
33	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
34	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
35	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
36	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
37	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
38	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
39	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37
40	=-C3+C37	=C7+C35	=-C5+C36	=-C6+C37	=C10+C35	=-C8+C36	=-C9+C37

	47	48	49	50	51	52	53
1	mtp			ach1			ach2
2	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
3	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
4	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
5	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
6	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
7	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
8	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
9	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
10	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
11	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
12	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
13	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
14	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
15	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
16	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
17	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
18	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
19	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
20	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
21	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
22	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
23	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
24	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
25	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
26	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
27	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
28	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
29	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
30	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
31	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
32	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
33	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
34	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
35	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
36	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
37	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
38	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
39	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35
40	=C13+C35	=-C11+C36	=-C12+C37	=C16+C35	=-C14+C36	=-C15+C37	=C19+C35

	61	62	63	64	65	66	67
1		fp1			fp2		
2	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
3	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
4	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
5	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
6	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
7	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
8	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
9	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
10	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
11	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
12	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
13	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
14	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
15	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
16	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
17	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
18	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
19	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
20	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
21	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
22	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
23	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
24	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
25	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
26	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
27	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
28	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
29	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
30	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
31	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
32	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
33	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
34	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
35	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
36	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
37	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
38	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
39	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37
40	=-C24+C37	=C28+C35	=-C26+C36	=-C27+C37	=C31+C35	=-C29+C36	=-C30+C37

	68	69	70	71	72
1	fp3			m1	c1
2	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
3	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
4	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
5	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
6	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
7	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
8	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
9	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
10	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
11	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
12	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
13	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
14	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
15	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
16	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
17	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
18	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
19	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
20	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
21	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
22	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
23	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
24	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
25	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
26	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
27	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
28	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
29	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
30	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
31	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
32	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
33	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
34	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
35	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
36	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
37	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
38	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
39	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)
40	=C34+C35	=-C32+C36	=-C33+C37	=(C52-C55)/(C51-C54	=C52-(C71*C51)

	73	74	75	76
1	m2	c2	yint	zint
2	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
3	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
4	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
5	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
6	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
7	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
8	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
9	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
10	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
11	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
12	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
13	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
14	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
15	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
16	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
17	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
18	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
19	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
20	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
21	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
22	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
23	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
24	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
25	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
26	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
27	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
28	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
29	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
30	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
31	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
32	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
33	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
34	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
35	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
36	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
37	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
38	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
39	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72
40	=(C54-C51)/(C52-C55	=C46-(C73*C45)	=(C74-C72)/(C71-C73	=(C71*C75)+C72

	77	78	79
1	d1	l.leg	deg.
2	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
3	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
4	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
5	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
6	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
7	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
8	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
9	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
10	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
11	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
12	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
13	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
14	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
15	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
16	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
17	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
18	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
19	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
20	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
21	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
22	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
23	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
24	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
25	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
26	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
27	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
28	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(18C/3.14)
29	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
30	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
31	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
32	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
33	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
34	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
35	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
36	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
37	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
38	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
39	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)
40	=SQRT((C45-C75)^2+(C46-C76)^2)-1	=ATAN((C50-C53)/(C52-C55)	=C78*(180/3.14)

	80	81	82	83	84	85	86	87
1	calc	deg.	rf_ang	Fz	Fy	Fx	ay	ax
2	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79					
3	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79					
4	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79					
5	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79					
6	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79					
7	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79					
8	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79					
9	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79					
10	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79					
11	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79					
12	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	14	0	-2	-0.001	0.001
13	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	17	0	-1	-0.001	0.001
14	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	17	0	-1	-0.001	0.001
15	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	14	0	-1	-0.001	0.001
16	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	17	0	-1	-0.001	0.001
17	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	19	-1	-1	-0.001	0.001
18	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	24	-2	-2	-0.001	0.001
19	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	24	-4	-1	-0.001	0.001
20	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	26	-4	0	-0.001	0.001
21	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	31	-5	0	-0.001	0.001
22	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	39	-8	0	-0.002	0.002
23	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	61	-14	0	-0.019	0.025
24	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	122	-26	0	-0.044	0.057
25	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	231	-41	0	-0.033	0.04
26	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	390	-55	0	-0.003	0
27	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	563	-66	-3	0.007	-0.009
28	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	720	-77	-9	0.001	0.003
29	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	830	-89	-15	0.003	0.008
30	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	886	-106	-26	0.014	0.002
31	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	903	-125	-42	0.024	-0.001
32	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	876	-136	-58	0.027	0.001
33	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	815	-133	-64	0.025	0.004
34	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	739	-114	-53	0.023	0.005
35	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	671	-88	-25	0.024	0.003
36	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	625	-70	14	0.027	-0.001
37	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	590	-60	49	0.028	-0.006
38	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	559	-52	69	0.031	-0.01
39	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	515	-43	69	0.038	-0.012
40	=ATAN((C56-C59)/(C58-C61)	=C80*(180/3.14)	=C81-C79	473	-36	53	0.048	-0.012

	88	89	90	91	92
1	Fy.z	Fz.'y'	Ma(qs)	ATF	BW
2	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
3	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
4	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
5	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
6	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
7	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
8	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
9	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
10	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
11	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
12	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
13	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
14	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
15	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
16	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
17	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
18	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
19	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
20	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
21	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
22	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
23	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
24	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
25	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
26	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
27	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
28	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
29	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
30	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
31	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
32	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
33	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
34	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
35	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
36	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
37	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
38	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
39	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)
40	=(C84*C46)/1000	=(C86-(C45/1000))*C8	=C88+C89	=(C90/C77)*1000	=C91/(54.8*9.8)

	93	94	95
1	thigh	deg.	shank
2	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
3	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
4	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
5	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
6	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
7	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
8	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
9	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
10	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
11	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
12	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
13	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
14	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
15	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
16	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
17	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
18	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
19	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
20	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
21	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
22	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
23	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
24	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
25	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
26	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
27	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
28	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
29	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
30	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
31	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
32	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
33	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
34	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
35	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
36	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
37	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
38	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
39	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)
40	=ATAN((C42-C39)/(C40-C43)	=C93*(180/3.14)	=ATAN((C42-C45)/(C43-C46)

	96	97	98	99
1	deg.	knee	foot	deg.
2	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
3	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
4	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
5	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
6	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
7	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
8	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
9	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
10	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
11	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
12	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
13	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
14	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
15	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
16	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
17	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
18	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
19	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
20	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
21	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
22	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
23	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
24	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
25	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
26	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
27	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
28	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
29	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
30	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
31	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
32	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
33	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
34	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
35	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
36	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
37	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
38	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
39	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)
40	=C95*(180/3.14)	=C94+C96	=ATAN((C49-C46)/(C48-C45)	=C98*(180/3.14)

	100	101	102	103
1	ankle	ank_w	rad.s-1	knee_w
2	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
3	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
4	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
5	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
6	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
7	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
8	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
9	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
10	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
11	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
12	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
13	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
14	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
15	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
16	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
17	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
18	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
19	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
20	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
21	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
22	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
23	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
24	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
25	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
26	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
27	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
28	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
29	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
30	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
31	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
32	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
33	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
34	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
35	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
36	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
37	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
38	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
39	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000
40	=90+(C99+C96)	=(RC(-1)-R(-1)C(-1))*1000	=C101*(3.14/180)	=(RC97-R(-1)C97)*1000

	104	105	106	107
1	rad.s-1	thi_w	rad.s-1	foot_w
2	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
3	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
4	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
5	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
6	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
7	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
8	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
9	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
10	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
11	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
12	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
13	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
14	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
15	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
16	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
17	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
18	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
19	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
20	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
21	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
22	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
23	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
24	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
25	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
26	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
27	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
28	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
29	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
30	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
31	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
32	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
33	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
34	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
35	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
36	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
37	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
38	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
39	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000
40	=C103*(3.14/180)	=(RC94-R(-1)C94)*1000	=C105*(3.14/180)	=(RC99-R(-1)C99)*1000

	108
1	rad.s-1
2	=C107*(3.14/180)
3	=C107*(3.14/180)
4	=C107*(3.14/180)
5	=C107*(3.14/180)
6	=C107*(3.14/180)
7	=C107*(3.14/180)
8	=C107*(3.14/180)
9	=C107*(3.14/180)
10	=C107*(3.14/180)
11	=C107*(3.14/180)
12	=C107*(3.14/180)
13	=C107*(3.14/180)
14	=C107*(3.14/180)
15	=C107*(3.14/180)
16	=C107*(3.14/180)
17	=C107*(3.14/180)
18	=C107*(3.14/180)
19	=C107*(3.14/180)
20	=C107*(3.14/180)
21	=C107*(3.14/180)
22	=C107*(3.14/180)
23	=C107*(3.14/180)
24	=C107*(3.14/180)
25	=C107*(3.14/180)
26	=C107*(3.14/180)
27	=C107*(3.14/180)
28	=C107*(3.14/180)
29	=C107*(3.14/180)
30	=C107*(3.14/180)
31	=C107*(3.14/180)
32	=C107*(3.14/180)
33	=C107*(3.14/180)
34	=C107*(3.14/180)
35	=C107*(3.14/180)
36	=C107*(3.14/180)
37	=C107*(3.14/180)
38	=C107*(3.14/180)
39	=C107*(3.14/180)
40	=C107*(3.14/180)

	109
1	foot_a
2	
3	
4	
5	
6	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
7	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
8	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
9	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
10	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
11	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
12	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
13	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
14	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
15	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
16	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
17	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
18	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
19	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
20	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
21	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
22	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
23	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
24	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
25	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
26	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
27	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
28	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
29	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
30	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
31	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
32	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
33	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
34	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
35	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
36	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
37	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
38	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
39	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C
40	=((4*R(-4)C99) + (4*R(-3)C99) + (R(-2)C99) - (4*R(-1)C99) - (10*RC99) - (4*R(1)C99) + (R(2)C

	110	111	112
1	rad.s-2	y_cm	z_cm
2	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
3	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
4	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
5	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
6	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
7	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
8	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
9	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
10	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
11	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
12	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
13	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
14	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
15	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
16	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
17	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
18	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
19	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
20	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
21	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
22	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
23	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
24	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
25	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
26	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
27	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
28	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
29	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
30	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
31	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
32	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
33	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
34	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
35	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
36	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
37	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
38	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
39	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000
40	=C109*(3.14/180)	=(C45+0.45*(C48-C45))/1000	=(C49+0.55*(C46-C49))/1000

	113	114
1	y_vel	z_vel
2	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
3	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
4	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
5	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
6	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
7	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
8	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
9	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
10	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
11	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
12	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
13	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
14	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
15	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
16	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
17	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
18	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
19	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
20	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
21	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
22	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
23	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
24	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
25	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
26	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
27	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
28	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
29	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
30	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
31	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
32	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
33	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
34	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
35	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
36	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
37	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
38	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
39	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000
40	=(RC111-R(-1)C111)*1000	=(RC112-R(-1)C112)*1000

	115
1	y_acc
2	
3	
4	
5	
6	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
7	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
8	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
9	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
10	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
11	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
12	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
13	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
14	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
15	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
16	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
17	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
18	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
19	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
20	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
21	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
22	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
23	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
24	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
25	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
26	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
27	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
28	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
29	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
30	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
31	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
32	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
33	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
34	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
35	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
36	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
37	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
38	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
39	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +
40	=((4*R(-4)C111) + (4*R(-3)C111) + (R(-2)C111) - (4*R(-1)C111) - (10*RC111) - (4*R(1)C111) +

	116
1	z_acc
2	
3	
4	
5	
6	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
7	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
8	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
9	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
10	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
11	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
12	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
13	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
14	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
15	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
16	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
17	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
18	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
19	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
20	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
21	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
22	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
23	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
24	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
25	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
26	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
27	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
28	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
29	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
30	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
31	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
32	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
33	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
34	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
35	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
36	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
37	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
38	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
39	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +
40	=((4*R(-4)C112) + (4*R(-3)C112) + (R(-2)C112) - (4*R(-1)C112) - (10*RC112) - (4*R(1)C112) +

	117	118	119	120
1	force_y	force_z	comp1	comp2
2	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
3	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
4	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
5	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
6	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
7	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
8	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
9	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
10	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
11	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
12	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
13	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
14	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
15	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
16	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
17	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
18	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
19	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
20	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
21	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
22	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
23	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
24	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
25	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
26	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
27	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
28	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
29	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
30	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
31	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
32	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
33	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
34	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
35	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
36	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
37	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
38	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
39	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112
40	=(0.747*C115)-C84	=(0.747*C116)-C83+(0.747*9.8)	=0.0083*C110	=C84*C112

	121	122	123
1	comp3	comp4	comp5
2	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
3	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
4	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
5	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
6	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
7	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
8	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
9	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
10	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
11	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
12	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
13	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
14	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
15	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
16	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
17	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
18	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
19	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
20	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
21	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
22	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
23	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
24	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
25	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
26	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
27	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
28	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
29	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
30	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
31	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
32	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
33	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
34	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
35	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
36	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
37	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
38	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
39	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)
40	=(C83*(C86-C111))	=C117*((C46/1000)-C112	=C118*(C111-(C45/1000)

	124	125	126	127
1	*Ma (ID)*	Ma_diff	heel_y	heel_z
2	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
3	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
4	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
5	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
6	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
7	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
8	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
9	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
10	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
11	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
12	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
13	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
14	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
15	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
16	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
17	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
18	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
19	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
20	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
21	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
22	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
23	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
24	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
25	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
26	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
27	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
28	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
29	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
30	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
31	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
32	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
33	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
34	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
35	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
36	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
37	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
38	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
39	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)
40	=C119-C120-C121+C122+C123	=C90+C124	=(RC60-R(-1)C60)	=(RC61-R(-1)C61)

	128	129	130	131
1	heel_v	ATF	BW	ATF_diff
2	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
3	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
4	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
5	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
6	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
7	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
8	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
9	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
10	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
11	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
12	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
13	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
14	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
15	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
16	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
17	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
18	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
19	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
20	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
21	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
22	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
23	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
24	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
25	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
26	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
27	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
28	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
29	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
30	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
31	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
32	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
33	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
34	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
35	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
36	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
37	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
38	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
39	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92
40	=SQRT(C126^2+C127^2)	=-C124/(C77/1000)	=C129/(54.8*9.8)	=C130-C92

	132	133	134	135
1	ATF_rate	N.s-1	ank_cp	mtp_y
2	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
3	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
4	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
5	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
6	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
7	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
8	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
9	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
10	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
11	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
12	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
13	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
14	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
15	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
16	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
17	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
18	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
19	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
20	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
21	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
22	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
23	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
24	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
25	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
26	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
27	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
28	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
29	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
30	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
31	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
32	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
33	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
34	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
35	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
36	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
37	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
38	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
39	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)
40	=(RC130-R(-1)C130)*1000	=(RC129-R(-1)C129)*1000	=C86-(C45/1000)	=C86-(C48/1000)

	136	137	138
1	mtp_z	Mmtp	sh_cmy
2	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
3	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
4	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
5	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
6	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
7	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
8	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
9	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
10	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
11	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
12	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
13	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
14	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
15	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
16	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
17	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
18	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
19	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
20	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
21	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
22	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
23	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
24	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
25	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
26	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
27	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
28	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
29	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
30	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
31	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
32	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
33	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
34	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
35	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
36	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
37	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
38	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
39	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45
40	=C49/1000	=(C135*C83)+(C136*C84	=C45/1000+(0.629/1000)*(C42-C45

	139	140	141
1	sh_cmz	sh_vy	sh_vz
2	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
3	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
4	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
5	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
6	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
7	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
8	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
9	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
10	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
11	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
12	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
13	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
14	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
15	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
16	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
17	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
18	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
19	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
20	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
21	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
22	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
23	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
24	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
25	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
26	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
27	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
28	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
29	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
30	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
31	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
32	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
33	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
34	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
35	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
36	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
37	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
38	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
39	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000
40	=(0.629/1000)*(C43-C46)+(C46/1000	=(RC138-R(-1)C138)*1000	=(RC139-R(-1)C139)*1000

	142
1	sh_accy
2	
3	
4	
5	
6	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
7	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
8	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
9	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
10	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
11	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
12	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
13	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
14	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
15	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
16	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
17	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
18	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
19	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
20	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
21	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
22	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
23	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
24	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
25	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
26	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
27	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
28	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
29	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
30	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
31	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
32	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
33	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
34	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
35	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
36	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
37	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
38	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
39	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +
40	=((4*R(-4)C138) + (4*R(-3)C138) + (R(-2)C138) - (4*R(-1)C138) - (10*RC138) - (4*R(1)C138) +

	143
1	sh_accz
2	
3	
4	
5	
6	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
7	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
8	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
9	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
10	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
11	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
12	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
13	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
14	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
15	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
16	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
17	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
18	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
19	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
20	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
21	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
22	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
23	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
24	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
25	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
26	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
27	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
28	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
29	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
30	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
31	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
32	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
33	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
34	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
35	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
36	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
37	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
38	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
39	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +
40	=((4*R(-4)C139) + (4*R(-3)C139) + (R(-2)C139) - (4*R(-1)C139) - (10*RC139) - (4*R(1)C139) +

	144	145
1	fky	fkz
2	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
3	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
4	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
5	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
6	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
7	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
8	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
9	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
10	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
11	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
12	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
13	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
14	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
15	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
16	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
17	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
18	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
19	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
20	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
21	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
22	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
23	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
24	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
25	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
26	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
27	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
28	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
29	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
30	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
31	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
32	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
33	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
34	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
35	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
36	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
37	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
38	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
39	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)
40	=((0.043*49.8)*C142)+C117	=C118+((0.043*49.8)*9.8)+((0.043*49.8)*C143)

	146	147	148
1	comp1	comp2	comp3
2	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
3	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
4	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
5	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
6	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
7	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
8	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
9	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
10	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
11	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
12	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
13	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
14	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
15	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
16	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
17	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
18	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
19	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
20	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
21	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
22	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
23	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
24	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
25	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
26	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
27	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
28	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
29	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
30	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
31	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
32	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
33	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
34	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
35	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
36	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
37	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
38	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
39	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145
40	=(C139-(C46/1000))*C117	=(C138-(C45/1000))*C118	=((C42/1000)-C138)*C145

	149	150	151
1	comp4	shank_w	rad.s-1
2	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
3	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
4	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
5	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
6	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
7	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
8	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
9	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
10	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
11	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
12	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
13	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
14	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
15	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
16	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
17	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
18	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
19	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
20	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
21	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
22	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
23	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
24	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
25	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
26	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
27	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
28	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
29	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
30	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
31	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
32	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
33	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
34	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
35	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
36	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
37	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
38	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
39	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)
40	=((C43/1000)-C139)*C144	=(RC96-R(-1)C96)*1000	=C150*(3.14/180)

	152
1	shank_a
2	
3	
4	
5	
6	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
7	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
8	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
9	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
10	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
11	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
12	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
13	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
14	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
15	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
16	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
17	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
18	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
19	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
20	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
21	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
22	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
23	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
24	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
25	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
26	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
27	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
28	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
29	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
30	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
31	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
32	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
33	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
34	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
35	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
36	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
37	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
38	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
39	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C
40	=((4*R(-4)C95) + (4*R(-3)C95) + (R(-2)C95) - (4*R(-1)C95) - (10*RC95) - (4*R(1)C95) + (R(2)C

	153	154	155	156
1	comp5	Mk (ID)	knee_y	knee_z
2	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
3	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
4	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
5	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
6	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
7	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
8	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
9	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
10	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
11	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
12	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
13	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
14	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
15	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
16	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
17	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
18	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
19	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
20	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
21	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
22	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
23	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
24	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
25	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
26	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
27	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
28	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
29	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
30	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
31	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
32	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
33	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
34	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
35	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
36	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
37	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
38	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
39	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000
40	=0.0504*C152	=C124+C146-C147-C148+C149+C153	=C86-(C42/1000)	=C43/1000

	157	158	159	160
1	Mk (qs)	Mk_diff	percent.	th_cmy
2	=(C155*C83)+(C156*C84)			=(0.628/1000)*(C39-C42)
3	=(C155*C83)+(C156*C84)			=(0.628/1000)*(C39-C42)
4	=(C155*C83)+(C156*C84)			=(0.628/1000)*(C39-C42)
5	=(C155*C83)+(C156*C84)			=(0.628/1000)*(C39-C42)
6	=(C155*C83)+(C156*C84)			=(0.628/1000)*(C39-C42)
7	=(C155*C83)+(C156*C84)			=(0.628/1000)*(C39-C42)
8	=(C155*C83)+(C156*C84)			=(0.628/1000)*(C39-C42)
9	=(C155*C83)+(C156*C84)			=(0.628/1000)*(C39-C42)
10	=(C155*C83)+(C156*C84)			=(0.628/1000)*(C39-C42)
11	=(C155*C83)+(C156*C84)			=(0.628/1000)*(C39-C42)
12	=(C155*C83)+(C156*C84)			=(0.628/1000)*(C39-C42)
13	=(C155*C83)+(C156*C84)			=(0.628/1000)*(C39-C42)
14	=(C155*C83)+(C156*C84)			=(0.628/1000)*(C39-C42)
15	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
16	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
17	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
18	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
19	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
20	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
21	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
22	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
23	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
24	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
25	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
26	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
27	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
28	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
29	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
30	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
31	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
32	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
33	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
34	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
35	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
36	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
37	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
38	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
39	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)
40	=(C155*C83)+(C156*C84)	=C154+C157	=(C158/C157)*100	=(0.628/1000)*(C39-C42)

	161	162	163
1	th_cmz	th_vy	th_vz
2	=(0.628/1000)*(C40-C43)		
3	=(0.628/1000)*(C40-C43)		
4	=(0.628/1000)*(C40-C43)		
5	=(0.628/1000)*(C40-C43)		
6	=(0.628/1000)*(C40-C43)		
7	=(0.628/1000)*(C40-C43)		
8	=(0.628/1000)*(C40-C43)		
9	=(0.628/1000)*(C40-C43)		
10	=(0.628/1000)*(C40-C43)		
11	=(0.628/1000)*(C40-C43)		
12	=(0.628/1000)*(C40-C43)		
13	=(0.628/1000)*(C40-C43)		
14	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
15	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
16	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
17	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
18	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
19	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
20	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
21	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
22	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
23	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
24	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
25	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
26	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
27	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
28	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
29	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
30	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
31	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
32	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
33	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
34	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
35	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
36	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
37	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
38	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
39	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000
40	=(0.628/1000)*(C40-C43)	=(RC160-R(-1)C160)*1000	=(RC161-R(-1)C161)*1000

	164
1	th_accy
2	
3	
4	
5	
6	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
7	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
8	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
9	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
10	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
11	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
12	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
13	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
14	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
15	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
16	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
17	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
18	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
19	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
20	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
21	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
22	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
23	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
24	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
25	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
26	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
27	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
28	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
29	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
30	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
31	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
32	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
33	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
34	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
35	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
36	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
37	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
38	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
39	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +
40	=((4*R(-4)C160) + (4*R(-3)C160) + (R(-2)C160) - (4*R(-1)C160) - (10*RC160) - (4*R(1)C160) +

	165
1	th_accz
2	
3	
4	
5	
6	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
7	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
8	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
9	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
10	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
11	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
12	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
13	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
14	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
15	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
16	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
17	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
18	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
19	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
20	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
21	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
22	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
23	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
24	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
25	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
26	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
27	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
28	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
29	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
30	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
31	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
32	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
33	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
34	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
35	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
36	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
37	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
38	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
39	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +
40	=((4*R(-4)C161) + (4*R(-3)C161) + (R(-2)C161) - (4*R(-1)C161) - (10*RC161) - (4*R(1)C161) +

	166	167
1	fhy	fhz
2		
3		
4		
5		
6		
7		
8		
9		
10		
11		
12		
13		
14		
15	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
16	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
17	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
18	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
19	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
20	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
21	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
22	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
23	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
24	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
25	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
26	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
27	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
28	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
29	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
30	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
31	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
32	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
33	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
34	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
35	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
36	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
37	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
38	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
39	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)
40	=((0.103*49.8)*C164)+C144	=((0.103*49.8)*C165)+C145+((0.103*49.8)*9.8)

	168	169	170
1	comp1	comp2	comp3
2			
3			
4			
5			
6			
7			
8			
9			
10			
11			
12			
13			
14			
15	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
16	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
17	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
18	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
19	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
20	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
21	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
22	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
23	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
24	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
25	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
26	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
27	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
28	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
29	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
30	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
31	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
32	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
33	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
34	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
35	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
36	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
37	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
38	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
39	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167
40	=(C161-(C43/1000))*C144	=(C160-(C42/1000))*C145	=((C39/1000)-C160)*C167

	171
1	comp4
2	
3	
4	
5	
6	
7	
8	
9	
10	
11	
12	
13	
14	
15	=((C40/1000)-C161)*C16
16	=((C40/1000)-C161)*C16
17	=((C40/1000)-C161)*C16
18	=((C40/1000)-C161)*C16
19	=((C40/1000)-C161)*C16
20	=((C40/1000)-C161)*C16
21	=((C40/1000)-C161)*C16
22	=((C40/1000)-C161)*C16
23	=((C40/1000)-C161)*C16
24	=((C40/1000)-C161)*C16
25	=((C40/1000)-C161)*C16
26	=((C40/1000)-C161)*C16
27	=((C40/1000)-C161)*C16
28	=((C40/1000)-C161)*C16
29	=((C40/1000)-C161)*C16
30	=((C40/1000)-C161)*C16
31	=((C40/1000)-C161)*C16
32	=((C40/1000)-C161)*C16
33	=((C40/1000)-C161)*C16
34	=((C40/1000)-C161)*C16
35	=((C40/1000)-C161)*C16
36	=((C40/1000)-C161)*C16
37	=((C40/1000)-C161)*C16
38	=((C40/1000)-C161)*C16
39	=((C40/1000)-C161)*C16
40	=((C40/1000)-C161)*C16

	172
1	th_a
2	
3	
4	
5	
6	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
7	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
8	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
9	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
10	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
11	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
12	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
13	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
14	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
15	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
16	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
17	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
18	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
19	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
20	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
21	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
22	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
23	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
24	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
25	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
26	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
27	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
28	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
29	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
30	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
31	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
32	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
33	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
34	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
35	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
36	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
37	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
38	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
39	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C
40	=((4*R(-4)C93) + (4*R(-3)C93) + (R(-2)C93) - (4*R(-1)C93) - (10*RC93) - (4*R(1)C93) + (R(2)C

	173	174	175	176
1	comp5	Mh (ID)	hip_y	hip_z
2			=C86-(C39/1000)	=C40/1000
3			=C86-(C39/1000)	=C40/1000
4			=C86-(C39/1000)	=C40/1000
5			=C86-(C39/1000)	=C40/1000
6			=C86-(C39/1000)	=C40/1000
7			=C86-(C39/1000)	=C40/1000
8			=C86-(C39/1000)	=C40/1000
9			=C86-(C39/1000)	=C40/1000
10			=C86-(C39/1000)	=C40/1000
11			=C86-(C39/1000)	=C40/1000
12			=C86-(C39/1000)	=C40/1000
13			=C86-(C39/1000)	=C40/1000
14			=C86-(C39/1000)	=C40/1000
15	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
16	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
17	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
18	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
19	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
20	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
21	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
22	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
23	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
24	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
25	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
26	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
27	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
28	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
29	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
30	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
31	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
32	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
33	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
34	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
35	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
36	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
37	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
38	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
39	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000
40	=0.1052*C172	=C154+C168-C169-C170+C171+C173	=C86-(C39/1000)	=C40/1000

	177	178	179	180	181
1	Mh (qs)	Mh_diff	percent	TIME	ATF(ID)
2	=(C175*C83)+(C176*C84			0	=C130
3	=(C175*C83)+(C176*C84			0	=C130
4	=(C175*C83)+(C176*C84			0	=C130
5	=(C175*C83)+(C176*C84			0	=C130
6	=(C175*C83)+(C176*C84			0	=C130
7	=(C175*C83)+(C176*C84			0	=C130
8	=(C175*C83)+(C176*C84			0	=C130
9	=(C175*C83)+(C176*C84			0	=C130
10	=(C175*C83)+(C176*C84			0	=C130
11	=(C175*C83)+(C176*C84			0	=C130
12	=(C175*C83)+(C176*C84			0	=C130
13	=(C175*C83)+(C176*C84			0	=C130
14	=(C175*C83)+(C176*C84			=R(-1)C +0.001	=C130
15	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
16	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
17	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
18	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
19	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
20	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
21	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
22	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
23	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
24	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
25	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
26	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
27	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
28	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
29	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
30	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
31	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
32	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
33	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
34	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
35	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
36	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
37	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
38	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
39	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130
40	=(C175*C83)+(C176*C84	=C177+C174	=(C178/C177)*100	=R(-1)C +0.001	=C130

	182	183	184	185	186	187	188	189
1	N	RATE	N.s-1	Ma(ID)	d1(mm)	Mk(ID)	Mh(ID)	heelv(z)
2	=C129			=C124	=C77	=C154	=C174	=C127
3	=C129			=C124	=C77	=C154	=C174	=C127
4	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
5	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
6	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
7	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
8	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
9	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
10	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
11	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
12	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
13	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
14	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
15	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
16	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
17	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
18	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
19	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
20	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
21	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
22	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
23	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
24	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
25	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
26	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
27	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
28	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
29	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
30	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
31	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
32	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
33	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
34	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
35	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
36	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
37	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
38	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
39	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127
40	=C129	=C132	=C133	=C124	=C77	=C154	=C174	=C127

	190	191	192	193	194	195	196	197
1	an_ang	kn_ang	th_ang	an_w	kn_w	th_w	cal_an	ll_ang
2	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
3	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
4	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
5	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
6	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
7	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
8	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
9	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
10	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
11	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
12	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
13	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
14	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
15	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
16	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
17	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
18	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
19	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
20	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
21	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
22	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
23	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
24	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
25	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
26	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
27	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
28	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
29	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
30	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
31	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
32	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
33	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
34	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
35	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
36	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
37	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
38	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
39	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79
40	=C100	=C97	=C94	=C102	=C104	=C106	=C81	=C79

	198	199	200	201	202
1	rf_ang	ft_ang	sh_ang	Mmtp	d2
2	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
3	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
4	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
5	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
6	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
7	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
8	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
9	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
10	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
11	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
12	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
13	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
14	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
15	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
16	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
17	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
18	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
19	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
20	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
21	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
22	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
23	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
24	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
25	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
26	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
27	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
28	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
29	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
30	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
31	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
32	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
33	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
34	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
35	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
36	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
37	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
38	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
39	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^
40	=C82	=C99	=C96	=C137	=SQRT((C86-(C45/1000))^2 +(C46/1000)^

	203
1	GRF lr
2	=(RC83 - R(-1)C83)*1000
3	=(RC83 - R(-1)C83)*1000
4	=(RC83 - R(-1)C83)*1000
5	=(RC83 - R(-1)C83)*1000
6	=(RC83 - R(-1)C83)*1000
7	=(RC83 - R(-1)C83)*1000
8	=(RC83 - R(-1)C83)*1000
9	=(RC83 - R(-1)C83)*1000
10	=(RC83 - R(-1)C83)*1000
11	=(RC83 - R(-1)C83)*1000
12	=(RC83 - R(-1)C83)*1000
13	=(RC83 - R(-1)C83)*1000
14	=(RC83 - R(-1)C83)*1000
15	=(RC83 - R(-1)C83)*1000
16	=(RC83 - R(-1)C83)*1000
17	=(RC83 - R(-1)C83)*1000
18	=(RC83 - R(-1)C83)*1000
19	=(RC83 - R(-1)C83)*1000
20	=(RC83 - R(-1)C83)*1000
21	=(RC83 - R(-1)C83)*1000
22	=(RC83 - R(-1)C83)*1000
23	=(RC83 - R(-1)C83)*1000
24	=(RC83 - R(-1)C83)*1000
25	=(RC83 - R(-1)C83)*1000
26	=(RC83 - R(-1)C83)*1000
27	=(RC83 - R(-1)C83)*1000
28	=(RC83 - R(-1)C83)*1000
29	=(RC83 - R(-1)C83)*1000
30	=(RC83 - R(-1)C83)*1000
31	=(RC83 - R(-1)C83)*1000
32	=(RC83 - R(-1)C83)*1000
33	=(RC83 - R(-1)C83)*1000
34	=(RC83 - R(-1)C83)*1000
35	=(RC83 - R(-1)C83)*1000
36	=(RC83 - R(-1)C83)*1000
37	=(RC83 - R(-1)C83)*1000
38	=(RC83 - R(-1)C83)*1000
39	=(RC83 - R(-1)C83)*1000
40	=(RC83 - R(-1)C83)*1000

Appendix H Sample statistics in Excel spreadsheet

Anova: Single Factor (time max ankle angle)

SUMMARY

Groups	Count	Sum	Average	Variance
Column 1	10	1.04	0.104	2.73E-05
Column 2	10	0.933	0.0933	0.00011
Column 3	10	1.034	0.1034	1.65E-05
Column 4	10	0.924	0.0924	0.000267

ANOVA

Source of Variation	SS	df	MS	F	P-value	F crit
Between Groups	0.001183	3	0.000394	3.7549	0.01915	2.866265447
Within Groups	0.003781	36	0.000105			
Total	0.004964	39				

Appendix I

Shank angle at impact for each subject/condition combination in degrees (SD)

Condition	A	B	C	D
Subject 1	1.0 (2.2)	1.1 (2.4)	1.2 (1.7)	2.5 (3.7)
Subject 2	-2.9 (1.3)	-2.1 (1.0)	-2.6 (2.4)	-3.0 (1.5)
Subject 3	-1.7 (1.3)	-0.5 (1.3)	-0.3 (1.3)	-0.1 (1.2)
Subject 4	-4.9 (1.5)	-2.8 (1.5)	-2.2 (3.2)	-4.4 (3.1)
Subject 5	-1.3 (2.1)	-1.4 (2.9)	-4.3 (1.9)	-4.3 (2.2)
Subject 6	-2.7 (1.0)	-0.6 (2.1)	-1.8 (1.1)	-0.7 (2.6)
Subject 7	2.0 (2.0)	2.2 (1.9)	1.2 (1.9)	1.1 (2.8)
Subject 8	-0.8 (3.1)	-0.2 (2.2)	1.2 (1.8)	-0.5 (2.8)

Knee angle at impact for each subject/condition combination in degrees (SD)

Condition	A	B	C	D
Subject 1	21.4 (1.8)	22.3 (2.7)	24.3 (2.6)	20.6 (2.5)
Subject 2	17.0 (0.8)	17.9 (1.1)	17.8 (2.8)	17.0 (1.0)
Subject 3	21.0 (1.9)	22.3 (1.2)	23.3 (1.7)	23.9 (1.5)
Subject 4	22.8 (3.2)	23.8 (1.7)	24.6 (4.3)	22.5 (3.0)
Subject 5	20.9 (2.4)	22.9 (3.2)	19.8 (2.8)	19.2 (2.5)
Subject 6	21.4 (1.0)	24.2 (2.4)	24.2 (1.2)	25.5 (2.0)
Subject 7	23.2 (2.6)	23.7 (2.3)	24.2 (2.8)	24.8 (3.3)
Subject 8	16.9 (3.6)	17.5 (2.6)	19.6 (2.9)	17.1 (3.8)

Maximum knee angle in degrees (SD)

	A	B	C	D
Subject 1	44.8 (2.0)	42.4 (1.7)	43.0 (1.0)	40.5 (4.4)
Subject 2	42.2 (0.6)	42.5 (0.9)	42.6 (0.8)	42.6 (0.6)
Subject 3*	45.1 (1.4)	45.4 (0.8)	46.8 (0.6)	45.8 (0.9)
Subject 4	51.6 (2.9)	49.0 (1:7)	49.8 (3.0)	49.6 (1.8)
Subject 5	47.2 (1.8)	48.9 (0.9)	49.7 (1.7)	49.7 (1.4)
Subject 6	44.8 (1.0)	44.0 (1.7)	44.8 (0.8)	44.3 (1.6)
Subject 7	48.0 (1.0)	47.0 (0.8)	47.2 (1.5)	48.2 (1.5)
Subject 8	45.7 (1.1)	45.1 (2.1)	46.3 (1.3)	46.2 (1.6)

*$p<0.05$ Subject 1: none
Subject 2: none
Subject 3: B versus C
Subject 4: none
Subject 5: none
Subject 6: none
Subject 7: none
Subject 8: none

Time of occurrence of maximum knee angle in ms (SD)

	A	B	C	D
Subject 1	64 (3)	62 (5)	65 (3)	67 (8)
Subject 2	60 (3)	63 (2)	64 (7)	65 (6)
Subject 3	67 (4)	64 (3)	65 (3)	65 (4)
Subject 4*	99 (12)	75 (17)	73 (13)	82 (15)
Subject 5*	71 (7)	71 (5)	91 (13)	92 (15)
Subject 6	63 (3)	62 (4)	67 (3)	62 (7)
Subject 7*	68 (5)	69 (8)	73 (5)	76 (5)
Subject 8*	71 (4)	66 (4)	67 (3)	67 (3)

*p<0.05 Subject 1: none
Subject 2: none
Subject 3: none
Subject 4: A versus B
Subject 5: B versus C; B versus D
Subject 6: none
Subject 7: B versus D
Subject 8: A versus B

Maximum ankle dorsi-flexion velocity (SD)

	A	B	C	D
Subject 1*	5.9 (0.7)	6.7 (0.3)	6.0 (0.5)	6.8 (0.6)
Subject 2*	6.8 (0.3)	7.3 (0.3)	7.7 (0.5)	7.3 (0.3)
Subject 3*	9.2 (0.5)	8.8 (0.5)	8.5 (0.9)	7.7 (0.5)
Subject 4*	8.1 (0.5)	9.2 (0.4)	8.1 (1.2)	8.1 (0.4)
Subject 5*	8.2 (1.9)	8.6 (1.1)	6.2 (0.3)	6.7 (0.4)
Subject 6	7.3 (0.5)	8.4 (1.3)	7.9 (1.3)	9.7 (2.0)
Subject 7*	5.9 (0.2)	6.3 (0.3)	6.6 (0.3)	6.4 (0.4)
Subject 8*	9.8 (1.7)	8.4 (1.0)	9.1 (1.5)	10.9 (2.0)

*$p<0.05$ Subject 1: A versus B; B versus C
Subject 2: A versus B; B versus C; C versus D
Subject 3: B versus D; C versus D
Subject 4: A versus B; B versus C; B versus D
Subject 5: B versus C; B versus D
Subject 6: none
Subject 7: A versus B
Subject 8: B versus D

Maximum knee flexion velocity (SD)

	A	B	C	D
Subject 1	9.1 (0.8)	9.5 (1.4)	8.6 (1.6)	9.5 (1.2)
Subject 2	10.7 (0.6)	10.0 (0.4)	10.2 (0.5)	10.4 (0.5)
Subject 3	8.2 (0.9)	8.0 (1.0)	8.0 (0.7)	7.8 (0.5)
Subject 4	11.3 (1.2)	11.0 (1.2)	10.9 (1.2)	10.7 (1.1)
Subject 5*	9.3 (0.9)	9.2 (1.5)	10.2 (0.8)	11.3 (1.0)
Subject 6*	9.2 (0.6)	7.9 (1.1)	7.5 (0.4)	7.9 (0.9)
Subject 7	8.4 (0.9)	8.0 (0.9)	7.8 (0.7)	7.6 (1.1)
Subject 8	9.5 (1.1)	9.6 (1.5)	10.0 (0.9)	10.1(1.3)

*$p<0.05$ Subject 1: none
Subject 2: none
Subject 3: none
Subject 4: none
Subject 5: B versus D
Subject 6: A versus B
Subject 7: none
Subject 8: none

Time of occurrence of maximum rearfoot angle (ms)

	A	B	C	D
Subject 1	93 (5)	78 (23)	89 (5)	76 (20)
Subject 2	99 (26)	72 (31)	93 (16)	88 (21)
Subject 4	46 (20)	39 (19)	44 (17)	47 (22)
Subject 5	35 (5)	72 (5)	90 (26)	100 (22)
Subject 6	73 (5)	28 (9)	65 (22)	74 (34)
Subject 7	102 (35)	73 (33)	64 (29)	56 (25)
Subject 8	56 (11)	67 (54)	66 (36)	63 (52)

Maximum lower leg angles for Subject 6 and Subject 7 in degrees (SD)

	A	B	C	D
Subject 6	4.4 (1.0)	2.3 (2.0)	0.8 (0.8)	2.1 (1.0)
Subject 7	1.3 (0.6)	3.8 (0.7)	4.1 (0.7)	6.0 (0.6)

Minimum calcaneal angles for Subject 6 and Subject 7 in degrees (SD)

	A	B	C	D
Subject 6	-3.2 (0.8)	-7.3 (1.7)	-5.9 (0.6)	-4.5 (1.4)
Subject 7	1.3 (0.6)	0.9 (1.4)	0.4 (1.6)	1.1 (0.8)