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Remote simultaneous dual wavelength imaging photoplethysmography: a further step towards 3-D mapping of skin blood microcirculation

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ABSTRACT

This paper presents a camera-based imaging photoplethysmographic (PPG) system in the remote detection of PPG signals, which can contribute to construct a 3-D blood pulsation mapping for the assessment of skin blood microcirculation at various vascular depths. Spot measurement and contact sensor have been currently addressed as the primary limitations in the utilization of conventional PPG system. The introduction of the fast digital camera inspires the development of the imaging PPG system to allow ideally non-contact monitoring from a larger field of view and different tissue depths by applying multi-wavelength illumination sources. In the present research, the imaging PPG system has the capability of capturing the PPG waveform at dual wavelengths simultaneously: 660 and 880nm. A selected region of tissue is remotely illuminated by a ring illumination source (RIS) with dual-wavelength resonant cavity light emitting diodes (RCLEDs), and the backscattered photons are captured by a 10-bit CMOS camera at a speed of 21 frames/second for each wavelength. The waveforms from the imaging system exhibit comparable functionality characters with those from the conventional contact PPG sensor in both time domain and frequency domain. The mean amplitude of PPG pulsatile component is extracted from the PPG waveforms for the mapping of blood pulsation in a 3-D format. These results strongly demonstrate the capability of the imaging PPG system in displaying the waveform and the potential in 3-D mapping of blood microcirculation by a non-contact means.

Keywords: photoplethysmography, remote, dual-wavelength imaging, resonant cavity light emitting diode, blood microcirculation.

1. INTRODUCTION

Photoplethysmography (PPG) is an optical bio-monitoring technique for the non-invasive measurement of blood volume changes in vivo¹. Its ease of use and convenience make it an attractive area of research in the biomedical and clinical community². The fundamental modus-operandi of PPG technology is the optical detection of the dynamic cardiovascular pulse-wave, generated by the heart, as it travels throughout the body. Pulse waves obtained from the PPG platform can deliver clinically valuable information for the study of cardiovascular system and skin microcirculation.

1.1 Conventional PPG

The principle of PPG, first used by Hertzman³, employs a small light source and a photosensitive detector (photoelectric cell) applied to the skin. Its biophysical principle is based on the fact that there is a strong contrast in absorption of near infrared (NIR) light between the blood-filled vessels and the ambient static tissue⁴. Hence a photoplethysmographic system with NIR sensors can measure the blood volume change in the tissue layers by registering the attenuation change in the near infrared. However the current PPG system can only monitor the dynamic change of the blood volume at a single site with a volume of a few cubic centimeters. Another obvious disadvantage is that it must be attached to the tissue of subject, which restricts its application to some clinical situation, e.g. wound diagnostics etc..

1.2 The inspiration of imaging PPG

Motivated by trends towards remote sensing in general, the desire to reduce physical restrictions and cabling associated with patient monitoring, researchers have been working towards non-contact camera-based PPG. The imaging PPG

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system can be benefit in removal of potentially injurious wires from magnetic resonance imaging (MRI) machines⁵ and avoiding the time-consuming scanning of laser Doppler perfusion imaging (LDPI)⁶. It can monitor the dynamic change of blood volume on different parts of skin surface simultaneously and then will bring the possible new insights that might come from hemodynamic imaging and mapping.

1.3 The current imaging PPG

The feasibility of a contact camera-based PPG has been well illustrated for the visualization of blood perfusion⁷, which demonstrated the simultaneous capture of PPG waveforms from the extremities at three wavelengths (660 nm, 840 nm and 905nm) in both transmission and reflection modes. One of other research⁸ reports the remote reflection mode capturing of “heartcycle-related” pulsatile variation utilizing a CMOS camera and a ringlight at similar wavelengths (660, 810, and 940 nm). However, simultaneous capture was not achieved, and the arterial pulsation signal was too weak to be recognized. Another research⁹ presents a camera-based system capable of capturing two PPG signals at two different wavelengths simultaneously in a non-contact manner; but the system is difficult to locate the designated position and the high output power of light source requires more safety consideration.

In this paper an imaging PPG system is presented with the capability of capturing PPG signals at two different wavelengths (660nm and 880nm) simultaneously in a remote reflection mode, with a dual-wavelength RCLED ring illumination source (RIS) and a 10-bit CMOS camera at a speed of 21 frames per second (fps) for each wavelength. The results demonstrate the comparability of imaging PPG with a conventional contact PPG instrument and exhibit the system with the capability of obtaining quality PPG pulsatile signals from deep tissue. The 3-D mapping of skin blood microcirculation can be constructed based upon the arterial pulsation extracted from the imaging PPG signals.

2. METHODS

2.1 Instrumentation

The main components of the imaging PPG system are depicted in Fig. 1. The camera (MC1311, Mikrottron GmbH, Germany) with a 1.25” CMOS sensor has a maximum resolution of 1280x1024 pixels with square pixels measuring 12 μ m. The pixels are encoded using 10 bits, making it eligible to detect the weak arterial pulsation. With the powerful image capture and control software developed in Labview (National Instruments Co., USA), the camera delivers quality images, and the frame rate can be as high as 32 fps in a full resolution and up to 4852 fps using a region of interest (RoI) mode. The camera was connected to a personal computer (PC) via the full Camera Link®. The captured frames was directly streamed into the PC. The system also used an industry-standard C-Mount zoom lens (Focal length: 8mm, F: 1.4, Ø: 2/3”, M0814-MP, Computar, Japan). Depending upon the lens, it was possible to observe arbitrary parts of the skin surface, from a few square millimeters to several square centimeters as requested in actual measurement. The camera was mounted on a tripod with the lens facing approximately 5 cm from the tissue surface under investigation.

The imaging PPG system presented here firstly introduced resonant cavity light emitting diode (RCLED) as the illumination source. The RCLED is a kind of high-speed light emitting diode (LED) normally designed for fiber optical communication. It also brings advantages in the acquisition of better signal to noise ratio (SNR), as it has a much narrower bandwidth of spectrum superior to conventional LEDs. A custom built dual-wavelength RCLED RIS (λ_1 : 660 nm, λ_2 :880 nm) with an elliptical reflector (DIA: 14cm, O.L: 5cm, B&Q Co. UK) outside to collimate and uniform the light distribution, was mounted around the camera lens. The RIS was composed of 20 RCLEDs—10 with a peak wavelength of 660 nm (TRL-2D20-060, Truelight Co., Taiwan) and 10 with a peak wavelength of 880 nm (TRL-2D20-080, Truelight Co., Taiwan), emitting 0.8 mW output power at a forward current of 10mA individually. The arrangement of the 660 and 880 nm RCLEDs is depicted in the lower inset as shown in Fig. 1. A control circuit with a microcontroller (PIC16F876A, MicroChip Inc., USA) alternately energized each wavelength group of RCLEDs, such that the RIS alternately produced equal duration pulses of diffuse illumination centered at 660 and 880 nm. The relative timing of the signal is illustrated in the upper inset as shown in Fig. 1. As each wavelength RCLEDs on the RIS were being energized, the camera received a trigger signal and began integration of a frame. The frame rate was set to be 21 fps for each wavelength that was sufficient to recover the waveform of PPG arterial pulsation.

Fig. 1. Illustration of the non-contact imaging PPG system, depicting illumination and imaging geometry, relative timing signals (upper inset), and RCLED arrangement of the dual-wavelength RCLED RIS (lower inset).

2.2 Signal processing

The captured frames was processed offline in Labview, and the pixel value from each frame was saved in the format of a Matlab (The Math Work Inc., USA) 3-D matrix: Rows $\times$ columns $\times$ framenumber.. The matrix was split into two sequences along the frame number: the brighter (higher pixel value) one illuminated by 660nm illumination and the other by 880nm, because of the higher camera quantum efficiency (QE) at 660nm. The process is illustrated in Fig. 2. Each frame was divided into boxes or groups of adjacent pixels of interested dimensions, such as 20 $\times$ 20 pixels. The average value of the pixels in each box was tracked from frame to frame as depicted in Fig. 2(a). This process was performed in both sequences of frames. Where backscattered light emerging from a box passed through an artery, a PPG signal was detected in the time-varying mean value of that box as shown in Fig. 2(b). This process yielded two multiplexed PPG signals, captured at two wavelengths simultaneously.

After the filtration of noise and the DC component in the captured PPG signals, a characteristic PPG AC pulsatile waveform usually dependent upon heart rate was derived in both time domain and frequency domain. While collecting the PPG AC signals from all the regions in the sequence of frames, the mean amplitude from each pulse of individual region over the measurement period was derived to describe regional blood pulsation situation.

Fig. 2. Illustrated the procedures of imaging processing (a) and signal acquisition (b).

2.3 Experimental protocol

A healthy female subject participated in this study and the experimental protocol was performed in a dark room. The subject was requested to seat in an upright position with her right arm resting on a cushioned bench. A conventional Dolphin 3311V transmission mode pulse oximetry probe (Dolphin Medical, UK) controlled by DISCO4 PPG system (Dialog Devices Ltd, UK) was connected to the subject's right index finger. The camera with the lens surrounded by the RIS was positioned and focused on the subject's volar side of the arm, and the palm close to the wrist was in the preview window of the camera. The camera was configured to capture 420 frames at a resolution of 640x512 pixels. The output power of the RIS for each wavelength reached to 8mW by setting the forward current as 10mA for each RCLED. A trigger signal was sent simultaneously to the camera system and the conventional PPG data acquisition system, initiating a 10s capture of data by both systems.

3. RESULTS AND DISCUSSIONS

3.1 The waveform of PPG

The results from the remote imaging PPG system and conventional contact PPG system are presented in Fig. 3. The mean pixel values in a RoI of dimensions 20x20 pixels against time are plotted in Fig. 3. It needs attention that the camera corresponds to the received light intensity rather than the absorbed light intensity, thus the derived waveforms have been inverted so that their vertical axes correspond to light absorption which is directly proportional to the peripheral arterial pressure wave form. For the signals from the non-contact imaging system, the systolic peaks and diastolic troughs are clearly recognized in accordant to the contact PPG signal. The non-contact imaging PPG system performs even more sensitive to the dicrotic notch which can reflect the closure of the aortic valve. However, originated from the same heartbeat, different shape and amplitude of the non-contact PPG pulsation signals are observed from the outputs illuminated by two different wavelength 660nm and 880nm. The phenomena indicates the different penetration depths depending on the emitter wavelength, and also the distinct blood microcirculation at various vascular layers in tissue. The Fourier spectra of the three PPG AC signals in Fig. 3 are plotted in Fig. 4. The pulsatile component is clearly visible in all three spectra at 1.4 Hz 2.

Fig. 3. The comparison of the non-contact imaging PPG and the conventional PPG signals in time domain.

Fig. 4. The comparison of the Fourier spectra of non-contact and conventional PPG signals corresponding to Fig. 3.

3.2 3-D Mapping of blood pulsation

A conceptual 3-D mapping of the blood microcirculation is described based upon the mean pulsatile amplitude, which commonly illustrates the average pulsation strength. After the collection of the PPG AC signals from all the regions in the individual wavelength set of frames, the mean AC amplitude of individual regions over the measurement period is derived and plotted in a 3-D format to simulate the situation of the blood pulsation in the illuminated palm. The 3-D plot and the contour plot from 880nm illumination as shown in Fig. 5, indicate a visibly stronger pulsation in the finger site than in the palm because of the abundant superficial vascular distribution in finger.

Fig. 5. The mean AC amplitude of individual regions from 880nm frame sequence is plotted in 3-D and contour formats to simulate the situation of the blood microcirculation of the illuminated palm.

The radiation mean penetration depth for 660nm is about 600um reaching to the papillary plexus layer and 1100um for 880nm to the dermis layer ^{10, 11}. Based upon the same approach above, the 3-D mapping of distinct blood microcirculation at two vascular layers is shown in Fig. 6.

Fig. 6. The mean AC amplitude of individual regions from 660nm and 880nm frame sequence is plotted in a 3-D format to simulate the blood microcirculation of illuminated palm in two different vascular layers.

4. CONCLUSIONS

This study has demonstrated the feasibility and capability of the remote camera-based imaging PPG system to detect PPG pulsation signals. Although the camera is more susceptible to ambient light and motion artifact than a conventional contact probe, the PPG pulsatile signals captured by the imaging system have still shown a strong comparability with those captured by the conventional PPG sensor in both time domain and frequency domain. Based on the capability of the imaging PPG in capturing the waveform from multi-wavelength simultaneously and monitoring at a larger area, the 3-D mapping of blood microcirculation by a non-contact means is in prospect.

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REFERENCES

- ¹ A. B. Hertzman, "The blood supply of various skin areas as estimated by the photoelectric plethysmography," *AM. J. Physiol.* **124**, 329-340 (1938).
- ² J. Allen, "Photoplethysmography and its application in clinical physiological measurement," *Physiol. Meas.* **28**, R1–R39 (2007).
- ³ A. B. Hertzman, "Photoelectric plethysmograph of the fingers and toes in man," *Proc. Soc. Exp. Biol. Med.* **37**, 529-534 (1937).
- ⁴ A. V. Challoner, "Photoelectric plethysmography for estimating cutaneous blood flow," *Non-Invasive Physiological Measurements*, **1**, ed P Rolfe (London: Academic), 125–151 (1979).
- ⁵ T. R. Brown, B. Goldstein, and J. Little, "Severe burns resulting from magnetic resonance imaging with cardiopulmonary monitoring. Risks and relevant safety precautions," *Am. J. Phys. Med. Rehabil.* **72(3)**, 166-167 (1993).
- ⁶ K. Wardell and G.E. Nilsson, "Duplex Laser Doppler Perfusion Imaging," *Microvasc. Res.* **52**, 171-182 (1996).
- ⁷ J. Zheng, and S. Hu, "The preliminary investigation of imaging photoplethysmographic system", *J. of Physics: Conference Series* **85** 012031(2007).
- ⁸ F. P. Wieringga, F. Mastik, and A. F. W. Van Der Steen, "Contactless Multiple Wavelength Photoplethysmographic Imaging: A First Step Toward 'SpO2 Camera' Technology," *Ann. Biomed. Eng.* **33(8)**, 1034–1041 (2005).
- ⁹ K. Humphreys and T. Ward, "Noncontact simultaneous dual wavelength photoplethysmography: A further step toward noncontact pulse oximetry," *Rev. Sci. Instrum.* **78**, 044304 (2007).
- ¹⁰ E. Kohen, R. Santus, and J. G. Hirschberg, *Photobiology* (Academic, 1995), p. 308
- ¹¹ J. L. Reuss, "Multilayer modeling of reflectance pulse oximetry," *IEEE Trans. on Biomed. Eng.* **52(2)**, 153-159 (2005).