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Development of effective photoplethysmographic measurement techniques: from contact to non-contact and from point to imaging

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Abstract—**This paper provides an overview of the most recent developments in photoplethysmography (PPG). Existing contact point measurement techniques, i.e. pulse oximetry probes, are contrasted with the next generation non-contact and imaging implementations, i.e. non-contact reflection and camera-based PPG. The development of effective PPG monitoring techniques relies on novel approaches to opto-physiological modeling.**

I. Introduction

Photoplethysmography (PPG) is an optical technique for non-invasively measuring changes in blood volume, indirectly based on variations in light intensity passing through, or reflected from, skin tissue through the use of an illumination source and a photodetector [1]. The variation in the detected output is generated by the pulsation of arterial blood within the peripheral vasculature as stimulated by the quasi-periodic cardiac cycle [2].

Among PPG applications, pulse oximetry—the determination of arterial oxygen saturation—is the most widespread thanks to its ability to alert the clinician of the presence of hypoxemia in real time [3]. However, oximeters have a number of factors that lead to inaccurate readings and limit their applicability. The success of solutions to engineering issues of pulse oximetry depends heavily on the validity of the assumptions used to interpret the PPG signals. The principles of operation of current contact PPG is typically described using opto-physiological models based on the Beer-Lambert equation, where the measuring site is treated as a blood-filled cuvette with no scattering effects and the light sources are assumed to be monochromatic [4, 5]. With the increasing availability and accuracy of tissue optical properties in literature, the use of numerical solutions of light propagation in human tissue are providing increasingly valuable insights into such mechanisms.

The use of conventional contact PPG (CPPG) probes, whereby the light source and detector are in contact with the skin surface, means that measurements can only be taken from tissue where the probe fits. Contact probes cannot be used when mechanical isolation is required, such as the case for burned patients in an emergency room. Non-contact PPG (NCPPG) was initially explored by the researchers at Loughborough University [6]. With the growing demand of medical and biomedical environment, a practical NCPPG solution can fulfill this demand for clinical monitoring. There are two different modes of PPG techniques: a) transmission mode, and b) reflection mode. In a biomedical monitoring environment, the non-contact reflection PPG (NRPPG) could be advantageous in settings such as the cardiovascular monitoring of neonates, patients with skin trauma or for large scale population screening with hygiene consideration.

Motivated by general trends towards remote sensing, the desire to reduce physical restrictions and cabling associated with patient monitoring, researchers are working towards non-contact camera-based PPG. An imaging PPG system can result in an alternative functional imaging solution to clinical situations which are currently addressed utilizing magnetic resonance imaging (MRI) and laser Doppler perfusion imaging (LDPI). This technique brings new insights by providing hemodynamic imaging and mapping capability. The functionality of camera-based PPG has been well demonstrated through the visualization of blood perfusion [7,8], where the simultaneous capture of PPG waveforms from a region of interest on the extremities at two wavelengths (660nm and 880nm) was achieved. The 3-D mapping of skin blood microcirculation [9] can be constructed based upon the arterial pulsation extracted from the imaging PPG signals, and a previous study shows the comparability of imaging PPG with conventional contact PPG and the capability of obtaining quality PPG pulsatile signals from a human face [10].

This paper reviews two parts on development of effective photoplethysmographic measurement techniques at Loughborough University: 1). from contact sensing placement to non-contact sensing placement, and 2). from point measurement to imaging measurement.

II. Principles of Photoplethysmographic techniques

A. Contact Sensing Placement to Non-contact Sensing Placement

a. Opto-physiological modeling for pulse oximetry

The contact PPG sensor is used in commercially available pulse oximeters, which use only two wavelengths, an arrangement that assumes the presence of only haemoglobin and oxyhaemoglobin in the blood. The presence of other species of haemoglobin such as carboxyhaemoglobin and methaemoglobin in the blood has been proven to lead to measurement errors [11], as well as other factors such as

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heavy skin pigmentation. It is through a theoretical understanding of the limitations of pulse oximetry as provided by opto-physiological modeling, that possible ways to overcome these can be seen.

An effective opto-physiological model requires an optimum tradeoff between complexity and applicability. For instance, a simplistic model of an optical bio-monitoring modality might provide a valid description of the general mechanisms involved at the expense of subject-specific parameters, while a complex model might provide a detailed description of contributions from specific parameters but will be a function of a greater number of variables. A novel opto-physiological model for transmission-mode PPG on a human finger has been proposed at Loughborough University [4]. The model considers a set of physiologically significant layers on the measuring site and their corresponding absorption *(μ_a)*, scattering (represented as detector-dependent mean path lengths L_{det}) and pulsatility coefficients *(σ)*, as a function of wavelength *(λ)*, layer *(i)*, longitudinal *(x)* and circumferential *(θ)* positions:

$$\frac{I_{ACpeak}}{I_{DC}}(\lambda,x,\theta)=\exp\left(\sum_{i=1}^{n}\left((1+\sigma(\lambda,i))\mu_a(\lambda,i)L_{det}(\lambda,i,x,\theta)\right)\right) \quad (1)$$

This model is a significant progression from the standard Beer-Lambert Model used for PPG in pulse oximetry, allowing the inclusion of the fast growing tissue optics data available in literature, and providing compatibility with increasingly popular Monte-Carlo solutions to the mathematically rigorous Radiative Transport Equation for optical propagation in tissue.

b. Non-contact Reflection Photoplethysmography(NPPG)

When the PPG measurement techniques move from contact to non-contact sensor placement, we had to consider the situation of non-contact probe configuration. A NRPPG mode based upon the Beer-Lambert law [5] was taken into account to describe the relationship between the light incident on a biological tissue and the light received after transmission through it. This measurement method relies on the varying light absorption due to arterial blood volume change during a cardiac cycle.

Fig. 1 shows a simplified diagram of the light distribution of the NRPPG model. I_i is the light incident on the skin surface, I_0 is the intensity of light penetrating the skin, I_r is the reflected light intensity and I is the intensity of light emerging from the tissue through the banana effect [12]. The light reaching the photodetector can be separated into *static (DC)* and *dynamic (AC)* terms. *DC* consists of the reflected component (I_r) and the static component (from non-variable absorbers, i.e. tissue, bone, static blood etc.), while *AC* represents the variation of light absorption due to arterial blood volume change. *I*, thus, can be expressed as:

$$I = I_o \exp(-\mu_{eff}\, r) = I_o \exp[-\mu_{dynamic}\, d(t) - \mu_{static}\, m] \quad (2)$$

where $\mu_{dynamic}$ is the extinction coefficient of the dynamic component, and μ_{static} is the extinction coefficient of static components. The light path length through the dynamic and static components are *d(t)* and *m*, respectively.

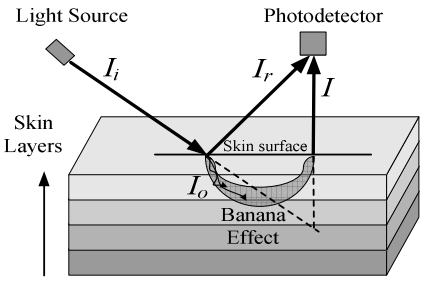

Fig. 1. Simplified representation of light distribution in

The AC term becomes independent of the light source intensity by normalizing the *AC* with respect to the *DC*:

$$\frac{AC}{DC}=\frac{I_o \exp(-\mu_{static}\, m)\,\mu_{dynamic}\, d(t)}{I_o \exp(-\mu_{static}\, m)}=\mu_{dynamic}\, d(t) \quad (3)$$

The amplitude of the normalized *AC* is proportional to the time-variant arterial light path length, which in turn depends on the characteristics of the banana shape. In practice, the reflected light is significant, resulting in an additive term in the denominator of equation (2), the minimization of which results in a maximized dynamic term.

B. Point Measurement to Imaging Measurement

a. Imaging photoplethysmography

Imaging photoplethysmography (IPPG) aims to remotely measure blood perfusion in an area of interest and provides a 3D segment blood perfusion mapping (*x,y,z*) based on extended multi-layered opto-physiological modeling rather than from a point measurement. In this IPPG model, the layered absorption (μ_a), the detector-dependent pulsatile mean path lengths (*r*), as a function of wavelength (λ), layer (*k*), and the detector position are considered. For a specific detector position, the dynamic component *AC(k)* representing specific layer blood perfusion can be derived as:

$$AC(k) = AC \times \frac{\mu_{a,blood}(k,\lambda)\times r(k,t)}{\sum_{i=1}^{n}\mu_{a,blood}(i,\lambda)\times r(i,t)} \quad (4)$$

with an amplitude:

$$ac(i) = ac \times \frac{\mu_{a,blood}(k,\lambda)\times r(k)}{\sum_{i=1}^{n}[\mu_{a,blood}(i,\lambda)\times r(i)]} \quad (5)$$

where *AC* and *ac* are the pulsatile component and corresponding amplitude from the imaging PPG signals. When moving the detector all over the designated tissue, 3-D mapping of segment blood perfusion can be achieved.

III. Development of Effective Opto-Physiological Monitoring Techniques

A. Validation of a Monte Carlo Platform for Opto-Physiological Modeling

An accurate 3D anatomical model (Zygote Media Group, USA) of a male adult finger was used to numerically resolve the radiative transfer theory equation for the case of a point source LED light source at 633 nm and 850 nm, placed near the standard position found in commercial probes. The simulation results were used to distinguish localized light intensities that vary due to the cardiovascular cycle from those that are static, thus providing a means to construct static and dynamic intensity distributions mapped to the surface of the finger. Sensor responses were generated by scanning a square bucket of arbitrary size in 0.1 mm steps from -1 to 1 mm longitudinally and from 0 to 360º circumferentially.

For validation of Monte Carlo simulation results, a CMOS camera (Mightex, USA) was mounted on a rotating platform that allowed the camera to maintain a constant distance and alignment with respect to the subject's finger (Fig.2). Sets of 200 frames were captured at 40 fps for both red and infrared illumination, between 90° and 270° in 9° steps, where 180° is the standard transmittance mode sensor position with respect to the light source. For each of the sets, the mean peak-to-peak intensity and mean intensity was determined from windows of size 10x10 pixels, resulting in two additional frames representing the AC and DC intensity distributions of that set. Finally, the AC and DC frames were trimmed and concatenated to form continuous intensity distributions from 90° to 270°.

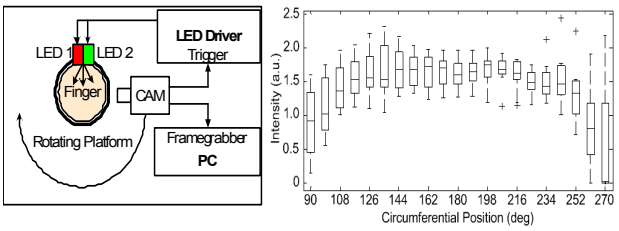

Fig.2. Validation platform for Monte Carlo virtual environment (left) and box & whisker plot of ratio of ratios distributions for a 10 subject study (right).

A study was performed on this validation platform, in which the fingers of 10 subjects were mapped and the resultant distributions were used to construct ratio of ratios distributions. The statistical analysis of these, shown in Fig. 2, indicates an even distribution of dynamic signal strength 120º - 240º and suggests a tendency for stronger signals in the vicinity of ±40º locations at either side of the 180º mark, as predicted in simulation.

B. Non-contact reflection PPG Photonics Engineering

The fundamental concepts underlying the design of the NRPPG system were the minimization of direct coupling, the concentration and even distribution of source irradiance for improved signal-to-noise ratio and reduced sensitivity to motion artifact, and the subsequent detection of light constrained to an area with the aforementioned light source characteristics. A vertical cavity surface-emitting LEDs (VCSEL, PH85-F1P1S2, mode: 10 mw, 850 nm, Beam Divergence: 30º, Roithner LaserTechnik, Austria) with an inherently narrow beam divergence and a high coherence, was used as an illumination source and was positioned to illuminate the tissue area at a changeable angle. A high-speed Silicon PIN photodiode (type: S5821-03, mode: 320-1100 nm, relative sensitivity: 20º, Hamamatsu Photonics UK Limited) with a high quantum efficiency (QE) over the spectral range was used as a photodetector in the engineering setup, positioned perpendicular to the tissue surface at a distance of 5cm. The photodiode was fitted with a lens which constrains its viewing angle to 10º. This restricts the amount of probed tissue, thus ensuring that the measuring site is within a homogeneously illuminated area, as shown in Fig. 2.

The quality of the signal is determined by the current of light source and the recording distance, which is related to the banana shape effect. The clean signal with a small variation of the amplitude is detected in the saturation range, where the photodetetor is easily saturated as a result of using a relative large light source. The clean signal with a large variation of the amplitude is detected in the optimal range, where is well revealed in the banana shape effect and the signal amplitude can be regulated by the light source driving current. In the low signal range, the quality of signal is sensitive to the interference, such as ambient light. The engineering setup corresponding the recording distance is within the range of 15 centimeters.

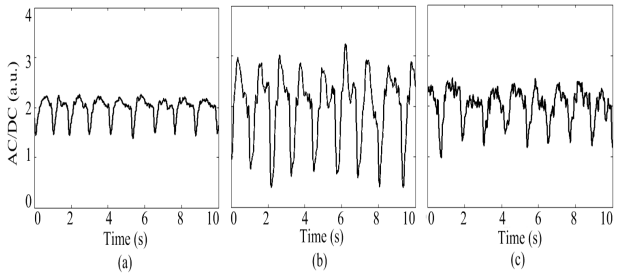

Fig. 3. Representative PPG signals captured from NRPPG photonics engineering setup: (a) Saturation range, (b) Optimal range and (c) Low signal range, respectively.

C. Configuration of Imaging Photoplethysmography

The IPPG system as shown in Fig. 4, a custom built dual-wavelength RCLED ring light source (λ1: 650 nm, λ2: 870 nm) with a parabolic reflector (DIA: 18 cm, O.L: 6.5 cm, B&Q, UK) to provide a collimated and uniform light, was mounted around the camera lens. The ring light consisted of 20 RCLEDs: 10 with a peak wavelength of 650 nm (TRC650SMD0603, WelTek Co. Ltd., Taiwan), each emitting 1.0 mW power at a forward current of 20 mA, and 10 with a peak wavelength of 870 nm (TRC870SMD0603, WelTek Co. Ltd., Taiwan) emitting 1.3 mW power individually. The arrangement of the RCLEDs is shown in

the lower inset (Fig. 4). A control circuit with a microcontroller (PIC16F876A, MicroChip Inc., USA) alternately powered each wavelength group of RCLEDs, such that the light output duration for each wavelength was constant. The relative illumination timings are illustrated in the upper inset, where the camera was triggered every time there was a switch of wavelength. The frame rate was set at 30 fps (15 fps for each wavelength) that was sufficient to recover the shape of the PPG arterial waveforms. We observed the arbitrary areas of the skin surface, from a few square millimeters to several square centimeters. The camera was mounted on an optical bench with the lens positioned approximately 13 cm away from the human face (Fig. 4).

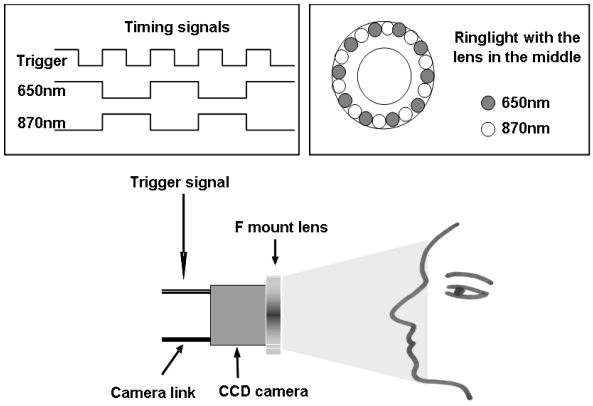

Fig. 4. Illustration of the IPPG system. The depicting illumination and the imaging geometry (bottom); the relative timing signals (upper right); and the RCLED arrangement of the dual-wavelength ring light (upper left).

The PPG information from one wavelength, such as heart rate, is clearly observed in the Fig. 5(a). The time-frequency spectrum for these pulsatile components derived from the IPPG signals was presented in Fig. 5(b), for a time window of 10seconds (150 frames). The fundamental frequency (i.e. HR) and the 2nd through 3rd harmonics can be identified in Fig. 5(b), indicating that both the HR and the shape of the plethysmogram were determined.

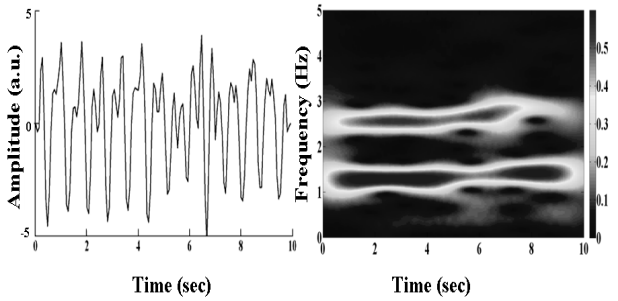

Fig. 5. (a) is the AC component from 870 nm wavelength (left) and (b) is the corresponding joint time-frequency diagram (right).

IV. DISCUSSION

The Monte Carlo validation approach to opto-physiological modeling relies on a process of fine-tuning the accuracy of characterization for both simulation and empirical validation. The approach demonstrates its functionality at a qualitative level and provides a path towards increasingly comprehensive descriptions of the mechanisms involved in PPG.

In the NRPPG engineering setup, the light source and the photodetector are placed on the same side of skin surface. The geometrical consideration in a such reflection PPG sensor is to determine the optimal distance between the light source and the photodetector. The technique used in non-contact measurement not only considered the above principles but also the combination of the incident angle and the illumination distance, because these two geometrical parameters are to decide the optimal range which determines the quality of PPG signal captured in the setup.

The IPPG technique performs comparably with a conventional contact device for HR measurements to acquire PPG waveforms. Although the camera is more susceptible to ambient light and motion artifact than a conventional contact probe, the potential of a non-contact imaging system to measure pulse rate and display PPG waveforms by non-contact means is only limited by the current configuration of the system.

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