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Motion-compensated non-contact detection of heart rate

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ABSTRACT

A new non-contact heart rate detection method based on the dual-wavelength technique is proposed and demonstrated experimentally. It is a well-known fact that the differences in the circuits of two detection modules result in different responses of two modules for motion artifacts. This poses a great challenge to compensate the motion artifacts during measurements. In order to circumvent this problem, we have proposed the amplitude spectrum and phase spectrum adaptive filter. Comparing with the time-domain adaptive filter and independent component analysis, the amplitude spectrum and phase spectrum adaptive filter can suppress the interference caused by the two circuit differences and effectively compensate the motion artifacts. To make the device is much compact and portable, a photoelectric probe is designed. The measurement distance is from several centimeters up to several meters. Moreover, the data obtained by using this non-contact detection system is compared with those of the conventional finger blood volume pulse (BVP) sensor by simultaneously measuring the heart rate of the subject. The data obtained from the proposed non-contact system are consistent and comparable with that of the BVP sensor.

1. Introduction

With the day-to-day increase in health concerns among human population, home health care is gaining significant importance. Heart rate detection is one of the most frequent examinations performed in health care monitoring. Regular and non-invasive assessments of heart rate are of great importance for the biomedical and clinical community.

Conventionally, a contact pulse oximeter is being widely used in routine and critical clinical applications. However, this may not be suitable under situations for damaged skin or when unconstrained movement is required. Moreover, it has been demonstrated that the spring-loaded clips in conventional contact photoplethysmography (PPG) finger sensor probes tend to affect the waveform of PPG signals because of the contact force between the sensor and the measurement site [1].

The potential way to overcome this issue is through non-contact measurements. To this end, several researchers have developed non-contact measurement techniques based on PPG imaging [2–7] and photodiodes [8]. In particular, photodiodes offer the advantage of faster, cheaper, and high-dynamic acquisition of PPG signal from the skin tissues, which make it an attractive area of research in the biomedical and clinical community. For instance, Shi et al. have demonstrated a non-contact PPG using infrared LED illumination and a simple photodiode [8]. However, motion artifact has always been the major factor affecting the precise measurement of heart rate, limiting the range of applications for the system. Driven by this limitation, several researchers have been working to circumvent the problem.

Lately, several methods have been proposed to remove the motion artifact from biomedical signals, such as blind source separation [5], independent component analysis (ICA) [9], single-channel ICA (SCICA) [6,10], and time-domain adaptive filter [11]. Of the different methods, ICA cannot be used for determining the order of the independent components [12]. Therefore, blind source separation [5], ICA [9], and SCICA [6,10] are often considered suitable for use in systems with small motion. The measurement becomes inaccurate when the motion is larger or the motion frequency is close to heart rate. To this end, Cennini et al. [11] introduced a dual-wavelength technique to effectively reduce the motion artifact in case of systems with larger motion. In this technique, the accuracy of measurement is affected by the interferences caused by the circuit differences. However, the existing methods do not seem to effectively solve the above mentioned problem.

In the present study, we have considered the circuit differences in the two detection modules and proposed an amplitude spectrum and phase spectrum adaptive filter. At the same time, the

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amplitude spectrum and phase spectrum adaptive filter is compared with time-domain adaptive filter and ICA. To make the measurement device is much compact and portable, a photoelectric probe is designed and the light source and collection system are integrated together. The measurement distance is from several centimeters up to several meters.

2. Theory and methods

In the previous study reported by Cennini et al. [11], it was assumed that the light power variations caused by motion of the skin and the blood volume pulses (BVPs) are additive. And they used the time-domain adaptive filter and the dual-wavelength to reduce the influence of motion. However, they did not consider interferences caused by the two circuit differences which would affect the accuracy.

In this article, we also use the dual-wavelength technique. In appendix A, the effect of motion artifact on the signal received by the photodiode is analyzed and the signal model is established. As mentioned in Appendix A, the signal model is shown as Eq. (1).

$$P = y \cdot \Delta l + z \cdot \Delta R \tag{1}$$

where y and z are constants that are different for different sources of light. Δl can represent the pulse signal and ΔR can represent the motion signal.

By the dual-wavelength technique, we can get two signals simultaneously. One of the two signals can be considered to include the pulse signal and the motion artifact (S_{P+M}) and the other signal is proportional to the motion artifact (S_M). Fig. 1 shows the absorption spectra of HbO_2 and Hb. As is shown in Fig. 1, the molar extinction coefficient of blue is much higher than the infrared molar extinction coefficient. The S_{P+M} signal can be obtained with the blue light and the S_M signal can be obtained with the infrared light. The blue light of wavelength 470 nm and infrared light of wavelength 940 nm are selected.

The blue light signal power (S_{P+M}) and the infrared light signal power (S_M) are expressed as Eqs. (2) and (3), respectively.

$$S_{P+M} = P_b = y_b \cdot \Delta l + z_b \cdot \Delta R \tag{2}$$

$$S_M = P_i = y_i \cdot \Delta l + z_i \cdot \Delta R \tag{3}$$

In the actual measurement process, there are always small differences between the two circuits, such as the response nuances between the two photodiodes and the nuances of the circuit components. These differences will make the two signals are not completely consistent. However it is complex to describe the small differences between the two circuits. Here, we simply assume that there is a response time delay interval between the two signals, which corresponds to the circuit differences of the two detection modules. The time delay interval is represented as the variable Δt. Accordingly, the blue and infrared signals are adjusted as Eqs. (4) and (5).

$$P_b = y_b \cdot \Delta l(t) + z_b \cdot \Delta R(t) \tag{4}$$

$$P_i = y_i \cdot \Delta l(t - \Delta t) + z_i \cdot \Delta R(t - \Delta t) \tag{5}$$

When Δt is very big, there will be severe dislocation in the time domain between the two signals. In this case the accuracy of the time-domain adaptive filter [11] will be reduced, even time-domain adaptive filter will have no effect. So the new method is needed.

As is well known, the Fourier transform has the time shift property [13]. In simple terms, if the Fourier transform of $x(t)$ is $X(\omega)$, then the Fourier transform of $x(t - t_0)$ is $X(\omega)e^{-j\omega t_0}$. The time delay does not affect the amplitude spectrum of the Fourier transform, rather introduces only the phase difference into the phase spectrum. Through the Fourier transform of these two signals, the amplitude spectra corresponding to the signal intensity and the phase spectra corresponding to the signal phase changes.

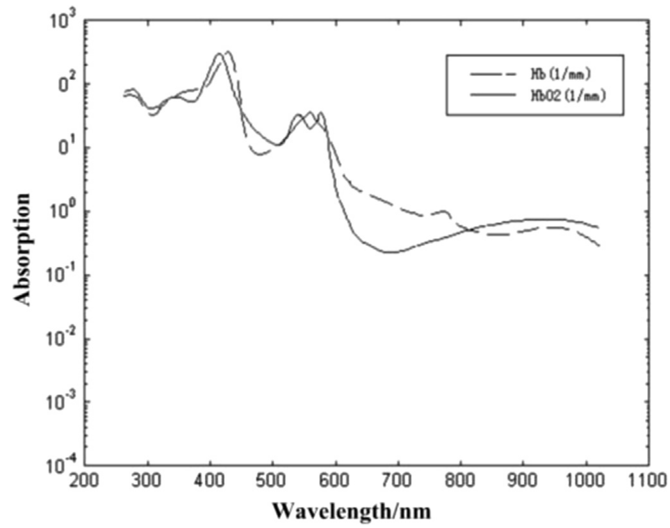

Fig. 1. Absorption spectra of HbO_2 and Hb.

To compensate the motion artifact intensity corresponding to the amplitude spectra, the amplitude spectra adaptive filter is used. While the phase spectra adaptive filter is used to calibrate the phase.

The process is shown in Eqs. (6) and (7):

$$m = |P_b(\omega)| - k_1 \cdot |P_i(\omega)| \tag{6}$$

$$n = \theta\left[P_b(\omega)\right] - k_2 \cdot \theta[P_i(\omega)] \tag{7}$$

where $P_b(\omega)$ and $P_i(\omega)$ are the Fourier transform of the blue signal and the infrared signal, respectively, $|P_b(\omega)|$ and $|P_i(\omega)|$ are the amplitude spectra of blue signal and infrared signal, respectively, $\theta[P_b(\omega)]$ and $\theta[P_i(\omega)]$ are the phase spectra of the blue signal and infrared signal, respectively, and k_1 and k_2 are obtained through the adaptive echo cancelation. Here, we use the variable step size least mean square adaptive filter algorithm [14–18].

The Fourier transform of the compensated signal is shown in Eq. (8).

$$P(\omega) = m \cdot \exp(j \cdot n) \tag{8}$$

Through the inverse Fourier transform, we obtain the new signal that compensates the motion artifact. We called this method amplitude spectrum and phase spectrum adaptive filter algorithm.

In order to determine the heart rate, a Fast Fourier Transform of the new signal is calculated. As is well known, the person's heart rate is 30–180 bpm (beat per minute), it corresponds to 0.5–3 Hz in the frequency domain. Then the heart rate is obtained which is the maximum frequency spectrum value in the selected frequency range.

3. Experimental setup and description of the study

3.1. Instruments

Fig. 2 shows the schematic illustration of the non-contact heart rate system proposed in this study. The photoelectric probe is designed to emit light and collect light from an assigned region of