Optical difference in the frequency domain to suppress disturbance for wearable electronics

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Abstract: Measurements based on optics offer a wide range of unprecedented opportunities in the biological application due to the noninvasive or non-destructive detection. Wearable skin-like optoelectronic devices, capable of deforming with the human skin, play significant roles in future biomedical engineering such as clinical diagnostics or daily healthcare. However, the detected signals based on light intensity are very sensitive to the light path. The performance degradation of the wearable devices occurs due to device deformation or motion artifact. In this work, we propose the optical difference in the frequency domain of signals for suppressing the disturbance generated by wearable device deformation or motion artifact during the photoplethysmogram (PPG) monitoring. The signal processing is simulated with different input waveforms for analyzing the performance of this method. Then we design and fabricate a wearable optoelectronic device to monitor the PPG signal in the condition of motion artifact and use the optical difference in the frequency domain of signals to suppress irregular disturbance. The proposed method reduced the average error in heart rate estimation from 13.04 beats per minute (bpm) to 3.41 bpm in motion and deformation situations. These consequences open up a new prospect for improving the performance of the wearable optoelectronic devices and precise medical monitoring in the future.

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1. Introduction

Recently, the wearable skin-like devices [1–5] have received increasing attention because of the potential applications in daily healthcare and clinical treatments [6–8]. Such devices can be imperceptible to the skin due to the ultra-thin geometry and low-modulus materials [9,10]. The wearable optoelectronic devices, integrated with the optoelectronics on the soft substrates, have broadly applied to the in-vitro monitor of vital signs and medical imaging systems [11,12]. For example, photoplethysmogram (PPG) [13–15], optically obtained plethysmogram, can be used to detect the change of blood volume in the microvascular bed of tissue for calculating pulse rate, blood oxygen [16,17], and blood pressure [18–21]. However, the waveform related to blood volume is very weak in the PPG signals, and the monitoring of PPG waveform depends on the stability of the optical path in the human tissue or blood [22]. And the optical path in the human tissue is easily impacted by motion artifact, which remains a great challenge for precisely monitoring PPG signals [23,24]. Generally, there are two main affecting factors on the PPG signal, involving the frequency-independent disturbance and frequency-dependent noises [25]. Some complex filtering strategies have been carried out to solve these problems [26], and the bandpass frequency filtering can be adopted for suppressing frequency-dependent noises. The frequency-independent disturbance, generally caused by motion artifact, is still an essential factor limiting the precise measurement of PPG signal. The previous works have reported the techniques for suppressing motion artifact, including spectral estimation [27], adaptive filtering [28,29], independent component analysis [30], empirical mode decomposition [29], wavelet-based denoising [31], or other decomposition models [32–34]. In most of the methods,
the accelerometers are employed to prove the reference signals, and the PPG signals are joined with the acceleration signals to remove the motion artifact. However, the accelerometers can solely measure the movement on the surface, unable to directly predicting the optical path change for skin-like devices, which is the key to influencing the stable PPG signals. For example, the spectrum estimation techniques [35], such as the window functions with different thresholds, can be built to filter the extra frequency components except that of the PPG signal. However, the spectrum of disturbance shares the same frequency bandwidth with a PPG waveform. The adaptive filters [36], which are generally used in active noise canceling (ANC), can effectively suppress the time-independent interferences due to the adaptive change of filter weight [37]. The reference signals are considered to collaborate with adaptive filters to improve the effect of disturbance-reduction [38]. The somatosensory measurement by sensors (e.g., acceleration sensors) can offer the reference signals for removing the interference from the motion and output signals. These techniques are wildly applied in commercial wearable products, such as smart wristbands or watches. Most of the works focus on digital signal processing after the data acquired. In terms of optical detection, the neglect of the optical path results in the low efficiency to suppress irregular disturbance. On another side, the intensity of disturbance signals remains nonlinear relationship to the acceleration or velocity of the human motion, which leads to an incomplete compensation of time-independent disturbance. Also, these techniques are suitable for the traditional rigid devices but no optimization is made to adapt to the wearable devices.

Another useful strategy to avoid the motion artifact is to fix the optical sensors firmly on the skin by a cuff or other types of bundles, which is usually used by commercial wristbands. But this strategy brings uncomfortableness to the human body due to the stress of the cuff and the irritation on the skin, limiting long-term monitoring. Although the wearable devices emerge as an alternative solution for the long-term monitoring of the vital signs in daily use [39–41], the deformation with the skin will induce huge impacts on the optical path. The false readings caused by shivering or other movements will be more severe on wearable optoelectronics. In conclusion, the suppression of frequency-independent disturbance is still a challenge, especially for wearable devices. The disturbance caused by the motion artifact or deformation is irregular and still exists after the frequency-related filter. Therefore, effective solutions for suppressing irregular disturbance are highly desirable for the biomedical applications of flexible and stretchable optoelectronics.

In this paper, we propose an optical difference in the frequency domain of signals for suppressing time-independent disturbance caused by the motion artifact or deformation of the devices. The simulation demonstrates the efficiency and performance of this strategy with different waveforms and disturbance. Then we fabricate a wearable optoelectronic device integrated with the light sources of different colors and photodetectors for monitoring the PPG signals. The PPG signals with the disturbance of motion artifact and deformation are recorded and then processed by the optical differential method. The results show that the optical difference in the frequency domain can effectively suppress the disturbance and amplify the expected signals. The research opens up a new prospect for improving the performance of wearable optoelectronic devices, especially for biomedical applications.

2. Results

2.1. Principles of the optical difference in the frequency domain

Generally, the PPG signals can derive from the optoelectronics by transmissive absorption (at the fingertip) or reflection (on the forehead) of the light, as shown in Fig. S1. For example, the PPG signals can be obtained through the light-emitting elements with the red and infrared color combined with photodetectors. However, the quality of the PPG signals depends on the stabilization of measuring the optical path. The motion artifact or the deformation of the devices are the two common factors that influence the optical path, triggering unwanted disturbance in a
PPG signal. The optical difference in the frequency domain of signals is proposed here to solve this problem, and the process is illustrated in Fig. 1. The input signal and reference signal are the two PPG signals generated by the light sources with different colors. For example, the signals of red light (620 nm) and infrared light (940 nm) can act as the input and reference signals, respectively. The PPG signals are composed of two components, including DC and AC, as shown in Fig. S2. The AC component is directly attributable to the variation in blood volume caused by the pressure pulse of the cardiac cycle, while the DC component relates to the bulk absorption of the other tissue, such as skin, fat, or water. The intensity of AC is related to the change of blood volume, and its value is much smaller than that of the DC signal. Also, the AC component is easily influenced by the undesired disturbance and will disappear due to the high noise power. To extract the AC component, the optical difference in the frequency domain of signals is proposed here. Firstly, the mean amplitude of the input and reference signals is calibrated by proportional scaling to achieve the approximately equal amplitude of the disturbance of input and reference signals after the calibration. Then the input signals and the reference signals are transferred to the frequency domain, respectively. The transfer methods used here include fast Fourier transform (FFT) and continuous wavelet transform (CWT), and both of them belong to the time-frequency analysis. Actually, according to the type of input signals, the corresponding transform will be selected. For example, the FFT is applied to the frequency analysis for stationary signals, and the CWT provides a flexible time-frequency grid to analyze signals whose spectral change over time.

![Flow chart of the optical difference in the frequency domain.](image)

After the FFT/CWT process, the difference in the input and reference frequency spectrum can be obtained by calculation. The intensity of disturbance will decrease in the difference spectrum due to the differential process. After this step, the optional process can be used to suppress the time-dependent noises. In a window with specific bandwidths of the frequency spectrum, the useless frequency components can be selected to delete. Alternatively, the traditional filtering process can be utilized to suppress high/low-frequency noises after the corresponding inverse transforms (IFFT/ICWT). Finally, the weak period signal in the input signal, such as AC components in the PPG signals, are extracted, and both the irregular disturbance and frequency-dependent noises are suppressed.

### 2.2. Simulating analysis

To explore the performance of the optical difference in the frequency domain, we first set the input signal as the sine waveform with random noises and simulate the optical difference in the frequency domain with a homemade program. Then we observe the waveform obtained by each step in Fig. 1. The simulating tool is a homemade program operated on MATLAB software (Ver. R2018b). In this simulation, the sine waveform can simulate the AC in the PPG signal monitored by the light source with a specific color. Due to the similarity between the no-regularity and
randomness, we use the random values added on the AC waveform to simulate the irregular disturbance by the change of the optical path. The random values are generated by the \textit{rand} function built in the MATLAB. Compared with the disturbance by the motion artifact, the add-on random values represent severe signal disturbance due to the real-time property. Both the AC waveform and random values are recorded by a 1kHz sampling rate, which is corresponding to the 0 Hz to 500 Hz spectrum in the frequency domain. The input signal ($S_i$) and the reference signal ($S_r$) can be written as the following equation:

\[
S_i(t) = A_i(t)T(t) + B_iN(t) \tag{1}
\]

\[
S_r(t) = A_r(t)T(t) + B_rN(t). \tag{2}
\]

$A_i$, $A_r$, $B_i$, $B_r$ are the intensity of AC signals and disturbance in $S_i$ and $S_r$, respectively. $T(t)$ is the period functions of AC signals. Here, $A_i$, $r$ and $B_i$, $r$ are the intensity of the AC signal and disturbance, respectively.

Figures 2(a) and 2(b) illustrate the AC waveform only and the waveform of $S_i$ in the time domain, respectively. The AC signal is generated by the standard sine function with 1 Hz frequency and unit intensity. $S_i$ can be expressed as:

\[
S_i = \sin(2\pi t) + 50 \times (\text{rand1}(-1, 1) + \text{rand2}(-1, 1)) \tag{3}
\]

The \textit{rand1}(-1,1) and \textit{rand2}(-1,1) are the random numbers from -1 to 1 by different turns, which can simulate the irregular disturbance. We define the output signal as $S_o$. The intensity of the disturbance is 50 times as large as the AC signal to show the performance of the optical difference. The signal-to-noise ratio (SNR) of $S_i$ is -39 dB, which can simulate the PPG signals with extreme disturbance. The AC cannot be extracted by traditional filtering methods due to a low SNR and the randomness of disturbance. A bandpass filter is an efficient tool for suppressing noises. However, it is invalidated to filter the irregular disturbance due to its frequency independence. Figure 2(c) shows the waveform of $S_i$ filtered by a bandpass filter with a 0.5–2 Hz frequency response. Although the noises with low and high frequency are suppressed effectively, the standard sine waveform of AC still can not be observed. The random noises fail to correspond to a specific frequency, and the irregular disturbance cannot be suppressed by the signal processing in the time domain.

To utilize the optical difference in the frequency domain, we introduce another signal $S_r$ to simulate the reference signal. In the experiment, $S_r$ is the PPG signal of the infrared light with 940 nm wavelength. $S_r$ is generated as follows:

\[
S_r = 0.7 \sin(2\pi t) + 50 \times (\text{rand1}(-1, 1) + \text{rand2}(-1, 1)) \tag{4}
\]

Then $S_i$ is processed by the optical difference in the frequency domain. This method can be used to extract the weak AC signals and suppress random noises. The waveform of $S_o$ is the same as the AC signal, as shown in Fig. 2(d).

Figure 2(e) illustrates the comparison of the frequency spectrum of $S_i$ and $S_o$. The frequency spectrum of $S_i$ contains various power density of noises from 0 Hz to 50 Hz, and the power density of the AC signal is less than that of the other frequency items. The spectrum of $S_o$ has the exclusive peak corresponding to 1 Hz, which equals to the frequency of the input AC waveform. The disappearance of the other frequency power represents the disturbance is entirely suppressed.

Generally, the tested signals are not simple sine waveform, and their intensity or period will change radically with time. Here, a waveform with frequency conversion is utilized to demonstrate the performance of the optical difference in the frequency domain. We simulate a waveform with changeable frequency along the timeline, as shown in Fig. 3(a). The AC signal is composed of three types of the waveform, including the line without a period, the frequency...
Fig. 2. Suppression of frequency-independent disturbance and the extraction of sine waveform by the optical difference in the frequency domain. (a) – (d) The waveform in the time domain at different steps. From (a) to (d): AC signal with 1 Hz frequency; input signal ($S_i$); $S_i$ filtered by a bandpass filter with 2 Hz cutoff frequency; Output signal ($S_o$) processed by the difference in the frequency domain. (e) The frequency spectrum of $S_i$ (top) and $S_o$ (bottom).

with 0.5 Hz, and 1 Hz. In the frequency domain, the items of the frequency will be changed at different time points. $S_i$ is generated by the following equation:

$$S_i = \begin{cases} 
\sin(2\pi t) & (10 \leq t < 20) \\
\sin(\pi t) & (25 \leq t < 45) \\
0 & (t = \text{other}) 
\end{cases} + 10 * (\text{rand1}(-1, 1) + \text{rand2}(-1, 1)).$$  \hspace{1cm} (5)

Two groups of random noises simulate the disturbance, and the intensity of the noises is ten times as large as the AC signal. Equation (6) simulates $S_r$ used for the optical difference:

$$S_r = 0.7 * S_i + 9(\text{rand1}(-1, 1) + \text{rand2}(-1, 1)).$$  \hspace{1cm} (6)

The random noises of the $S_r$ are 0.9 times that of $S_i$, and the influence of such an unequal situation will be discussed later.

Figure 3(b) illustrates the distribution of the frequency over time. The figure has two bright bars with 1 Hz ($10 \leq t < 20$) and 2 Hz ($25 \leq t < 45$), respectively, corresponding to the frequency spectrum of the AC signal. The power density of random noise appears on the power spectrum of
Fig. 3. Optical difference by CWT. (a) the waveform of the AC signal and (b) corresponding frequency distribution at each step. From top to bottom: AC signal, the input signal ($S_i$); the output signal ($S_o$).

$S_i$. The remarkable frequency components of the $S_o$ are corresponding to $S_i$, which is 1 Hz and 2Hz at different times. The waveform of $S_o$ in the time domain is also similar to the AC signal as expected. We also simulate the AC signal modulated by linearity, catenary, second harmonic, logarithm. The optical difference also suppresses irregular disturbance, as shown in Fig. S3.

The performance of this method will be influenced by the difference between $B_i$ and $B_r$ in Eqs. (1) and (2). Figure 4(a) shows the $S_o$ is calculated in the condition of 0% and 20% difference between $B_i$ and $B_r$. The AC waveform ($t \cdot \sin(2\pi t)$) can be extracted, while the waveform of $S_o$ will have a little distortion in the condition of a 20% difference between $B_i$ and $B_r$. We analyze the waveform of $S_o$ in terms of the differences between $B_i$ and $B_r$ from 0% to 20%. The intensity of $S_o$ should be increased by the same proportion with a period of 1 s. However, the difference between $B_i$ and $B_r$ influences the distribution of the intensity. There will be a 5% error when the difference is 20% (Fig. 4(b)). The period of the signals with a different difference in the disturbance is also calculated. Compared with the standard period, the average signals period’s absolute error is 2%, with a 20% difference, as shown in Fig. 4(c). These errors are acceptable due to their small values. The difference between $A_i$ and $A_r$ will not influence the waveform of the $S_o$, as shown in Fig. S4.

2.3. Experimental results

An important application of the optical difference in the frequency domain is for monitoring human PPG signals with wearable devices. The scattering and reflecting indexes of the blood vessel depend on the colors of the light sources. The photodetectors near the light source can detect the optical signal reflecting the volume of blood, and the intensity of the AC and DC
components depends on the color of incident lights, as shown in Fig. 5(a). The DC signal relates to the human tissue, and the change of the optical path by deformation mainly distorts the DC signals. The AC waveform is the key to calculating pulse rate and other biomedical parameters, and similar to the AC waveform, as mentioned in the simulating section.

Here we fabricate a wearable device integrated with optoelectronics for monitoring the PPG signals, as shown in Figs. 5(b) and 5(c). The device is integrated with two light-emitting elements with the center wavelengths of 660 nm (red) and 940 nm (infrared), respectively, and four photodetectors distributed around the light sources to detect the PPG signals. Four photodetectors are placed 0.5 cm away from the emitting elements, which is a parallel connection for amplifying the PPG signals. These optoelectronics are operated by an analog front end to monitor the PPG signals at different times. The time-division is used to record the PPG from red and infrared simultaneously. The sample rate is set as 100 Hz, which can record the sufficient information of PPG signal as well as filter unnecessary high-frequency noises. The 160 nm thick metal (Au/Cr) encapsulated by polyimide serves as the electrical interconnects for power and signal transmission. The thickness of the wearable device is around 17 µm (Fig. 5(d)) and packaged by two semipermeable polyurethane films without any irritating contact for the skin. In the experiments, the wearable device is coated with an optical filter dye to limit the ambient light. The device enables comfortable and imperceptible wearing due to its great softness and ultra-thinness, suggesting its great potential application for long-term monitoring without any irritation on the skin.

However, the deformation still will introduce unwanted disturbance to the PPG signals. Figure 5(e) shows the PPG signals acquired during the deformation, and the waveform processed by the optical difference in the frequency domain. In 0-10 seconds, we acquire the standard PPG waveform without the disturbance of deformation. The period of the waveform, which represents the AC component, equals the pulse rate. There are only some high-frequency noises, which can be filtered by the digital high-pass filter. Then we stretch the device on the skin, and an obvious disturbance occurs at 10 s point. The AC waveform also becomes irregular. The stretching strain on the device is eliminated at about 17 s point, and another huge DC distortion occurs. The disturbance lasts from 10 s to 27 s, and the waveform of AC components cannot be distinguished. Then we set the red and infrared PPG signals as $S_i$ and $S_r$, respectively, and calculate the $S_o$ by the optical difference in the frequency domain. Compared with the original PPG signal, the disturbance of the output signal is suppressed, and we can obtain the period and other AC components. The pulse rate is around 85 beat per minute (bpm), which can be calculated by the period of the differential signals (Fig. 5(f)).

Through a frequency window combined with the optical difference, the specific frequencies can be selected for filtering useless information. Figure 6(a) shows a typical PPG signal with
complex interferences, including device deformation, motion artifact, high-frequency noises, and power-line interference. The PPG signals are acquired under different conditions, including without interferences (0 s – 60 s) and with interference (60 s – 300 s). Figures 6(b) and 6(c) show the clean waveform and the disturbed waveform, which are corresponding to the segments in Fig. 6(a) marked in dashed boxes. The deformation will cause substantial fluctuation of the DC component, whereas the sudden arm motion results in the sharp spurious signal. Both of them lead to interference AC waveform. Besides, the AC waveform is very weak compared with the DC variation, and a slight change of the DC waveform will cover the AC waveform. Figure 6(d) shows the PPG signals processed by the traditional bandpass filter. The cutoff frequency of the bandpass filter is 0.3 Hz and 3 Hz, and the magnitude response is illustrated in Fig. S5. The bandpass filter removes noise with high frequency and outside of the frequency band of interest. However, irregular interferences still influence the filtered waveform. Figure 6(e) shows the PPG

Fig. 5. Wearable optoelectronic device for demonstrating the optical difference in the frequency domain. (a) Principle of detecting PPG signals by the wearable device. (b) Image of the wearable device integrated with two light sources (660 nm and 940 nm) and photodetectors. (c) Zoom-in image to show the device in detail. 1-4: the location of four photodetectors. (d) SEM image of the semiconductor used in the wearable device. The thickness and width are 16.97 µm and 1 mm, respectively. (e) The PPG signals with moving artifacts (top) and processed by the optical difference in the frequency domain (bottom). (f) The period and pulse rate calculated by the output signal.
signals processed by the optical difference in the frequency of signals. Another benefit of this method is the direct filter by the frequency window due to the signal processing in the frequency domain. Due to the use of a frequency window, only 0.3–3 Hz frequency participates in the IFFT process. This process is high efficiency due to avoid the complex filtering strategy in the time domain. The waveform processed by the optical difference method results in a clean PPG signal, and the peaks or valleys on the waveform can be used to calculate accurate heart rate (Fig. 5(f)).

The frequency spectrum of the signal processed by a bandpass filter and differential signal (R-IR) is shown in Figs. 7(a) and 7(b). Compared to the spectrum obtained by the bandpass filter, the amplitude near 1.3 Hz in the differential signal is amplified, and the other frequency components are suppressed. The maximum point appears at 1.25 Hz, which is the same as that of the R(IR) frequency spectrum.

The effectiveness of our method in terms of heart rate measurement is shown in Fig. 7(c). The heart rates are obtained by an optoelectronic sensor mounted on the forehead as the golden standard (Fig. S5), the original signal after bandpass filtering, and the processed signal with the proposed method, respectively. The heart rates from the PPG signals are obtained by peak

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**Fig. 6.** Waveform of the PPG signals measured by the wearable optoelectronic device. (a) Clean PPG signals (0 s – 60 s) and PPG signals influenced by device deformation and motion artifact (60 s – 300 s). (b) and (c) Waveform of 16 s – 26 s and 260 s – 270 s in (a), marked in blue and red dashed line, respectively. (d) and (e) PPG signals processed by a bandpass filter and the optical difference in the frequency domain of signal, respectively. Inset: PPG waveform in a frame marked by the red dashed line.
detection and calculating the time difference of the two continuous peaks. The three heart rates are very close in the first 60 s due to the clean PPG waveform. Then the heart rates obtained by the bandpass filter are inaccuracy because of the motion artifact and deformation of the flexible sensor, while the one obtained using the proposed method is still close to the standard value. Correspondingly, the mean absolute error (MAE) and the standard deviation of the absolute error (SDAE) overall processed waveform are computed and compared. We achieve a MAE ± SDAE of 3.41 ± 0.42 bpm for records waveform processed by the optical difference in the frequency domain, reflecting the improvement compared to the 13.04 ± 2.02 bpm of the bandpass filtering results.

3. Discussion

The PPG signals are the primary determinants of cardiovascular morbidity and mortality in individuals. Compared with previous technologies for filtering the noises in the PPG signal, the optical difference in the frequency domain of signals has the advantages of filtering the frequency-dependent noises as well as the irregular disturbance. The method can be applied to current hardware with minor changes. Thus, the low cost of this method enhances an immeasurable commercial value. The in-vitro experiments demonstrate the performance of the optical difference in the frequency domain for clinical application. In this work, the wearable device is adapted to monitor PPG signals in the condition of complex disturbance, and the AC component is extracted from the distorting PPG signals. The experiments demonstrate that the optical difference is totally adjusted to wearable devices for preventing the disturbance generated by deformation. The accuracy of the method is also demonstrated by the comparison of the frequency spectrum.

Fig. 7. (a) and (b) Frequency spectrum of the PPG signals processed by a bandpass filter and optical difference in the frequency domain, respectively. (c) Heart rates obtained by PPG signals without motion artifact, original signal with bandpass filtering, and the processed signal using the optical difference method, corresponding to Fig. 6.
The time-independent noises or disturbance introduce uncertain terms on the frequency spectrum, which may share the same frequency with pulse rate. As a result, the disturbance still exists after a complex filtering process. In this work, the filter starts with the propagation of the light in human tissue. Two signals with different light sources as the references are firstly recorded to suppress the disturbance. A heart rate calculation produces low MAE errors of $3.41 \pm 0.42$ bpm, which shows a high performance compared with the traditional bandpass filter.

The limitations of the current techniques for PPG monitoring are the requirement of immobility for avoiding motion artifact while the devices (e.g., smartwatch) working. These devices for in-vitro measurements are inaccurate due to the displacement of the skin. The optical difference method is an efficient solution for improving device performance in motion. In human tests, the device accuracy is demonstrated by monitoring multi-vital signs during walking with motion and deformation. The optical difference represents another route for improving the anti-disturbance ability.

4. Conclusion

In summary, the optical difference in the frequency domain of signals is proposed for filtering the interference caused by motion artifact or wearable optoelectronic device deformation during the optical measurements. This method is demonstrated by the simulation of various waveforms with large random noises. The FFT and Wavelet are used to calculate the differential signals, and the time-independent inference is effectively suppressed. The difference between the inferences of the input and reference signal is discussed to analyze its influence on the final results. The useful information can be obtained, indicating the possibility of monitoring PPG signals during body motion or wearable optoelectronic device deformation. To demonstrate such an application in biomedical engineering, we fabricate a wearable device integrated with the light sources of red and infrared color and photodetectors. We monitor the PPG signal during the device deforming and process the signal with the optical differential method. The results show that the method can effectively suppress interference and noises as well as amplify the pulse rate signal. The results open up new recognition of optical devices and provide the theoretical support for precise measurements using the wearable optoelectronic device.

5. Method

5.1. PPG signal processing

A theoretical explanation of the PPG signal is based upon Beer-Lambert Law, which can be expressed as:

$$ I_{out} = I_{in} e^{c_0 \varepsilon_0 L_0} e^{c_1 \varepsilon_1 L_1} = I_{DC} e^{c_1 \varepsilon_1 L_1}. \quad (7) $$

$I_{out}$, $I_{in}$, and $I_{DC}$ represent the light intensity that the photodetector receives, the light intensity of the LED, and the tissue absorbed intensity, respectively. The $c_{0,1}$, $\varepsilon_{0,1}$, and $L_{0,1}$ are the average substance concentration, index of light absorption, and light path in the blood and other human tissue. The changes in blood volume determine the value of the term $e^{c_1 \varepsilon_1 L_1}$, which is corresponding to the AC signal. The light path will be changed if the motion artifact occurs, and there will be a term $L_{0,1}$ added on the light path.

$$ I_{out} = I_{in} e^{c_0 \varepsilon_0 (L_0 + \Delta L_0)} e^{c_1 \varepsilon_1 (L_1 + \Delta L_1)} = I_{in} e^{c_0 \varepsilon_0 L_0} e^{c_1 \varepsilon_1 L_1} e^{c_0 \varepsilon_0 \Delta L_0 + c_1 \varepsilon_1 \Delta L_1} \quad (8) $$

The last term on the right side represents the intensity change caused by the motion artifact. Then we use the logarithmic function to generate:

$$ \ln(I_{out}) = \ln(I_{in}) + c_0 \varepsilon_0 L_0 + c_1 \varepsilon_1 L_1 + (c_0 \varepsilon_0 \Delta L_0 + c_1 \varepsilon_1 \Delta L_1). \quad (9) $$

The different blood volume results in a change of $L_1$, which is a time dependence parameter. The motion artifact will cause the change of $\Delta L_{0,1}$, and the irregular change introduces a random
disturbance. The other terms are constant due to the unchanged volume of human tissue. Equation (9) corresponds to our base model, and the disturbance uses an additive expression.

\[ S(t) = A(t)T(t) + BN(t) \]  

(10)

**Funding**

National Natural Science Foundation of China (11320101001, 11625207); National Basic Research Program of China (2015CB351904).

**Disclosures**

The authors declare no conflicts of interest.

See Supplement 1 for supporting content.

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