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Use of bi-level pulsed frequency-division excitation for improving blood oxygen saturation precision

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ABSTRACT

The measurement of blood oxygen saturation (SpO_2) plays an increasingly key role in medical monitoring and health detection. In order to improve SpO_2 precision, this paper proposes a new method, namely bi-level pulsed frequency-division excitation (BPFDM), which uses 200 Hz and 400 Hz bi-level pulse to excite two LEDs and then utilizes the digital demodulation algorithm to synchronously obtain dual-wavelength photoplethysmogram (PPG) signals. The principle of BPFDM is theoretically analysed, and the effectiveness is illustrated by the contrast experiments among time-division method (TDM), frequency-division (FDM) and BPFDM. The experimental results show that BPFDM gets the highest SNR and the least root mean square error (RMSE) of SpO_2% value. Compared with TDM and FDM, the SNR of BPFDM increases by 11 dB and 5 dB, and the RMSE of BPFDM reduces by 58% and 25%, respectively. So, BPFDM can significantly improve the SNR of PPG signal and SpO_2 precision under the same hardware and software conditions, which is valuable for the application in pulse oximeter.

1. Introduction

SpO_2 is a critical physiological parameter to evaluate blood oxygen-carrying capacity and reflect the level of human health, playing an increasingly key role in medical monitoring and health detection [1,2]. With the rapid development of Internet of things, SpO_2 has widely applied not only in clinical trials, but also in home health monitoring [3,4]. Therefore, obtaining high precision SpO_2 without the complicated analog circuit and the high cost is of importance for many scientific researches and practical applications.

The traditional method [5–8] of measuring SpO_2 is R value method, in which the LED extinction method normally is time-division method (TDM). The schematic diagram is shown in Fig. 1. Two identical current sources with multi-channel analog switch are used to drive R and IR LEDs in turn to obtain dual-wavelength PPG signals. AC-DC component values of dual-wavelength PPG signals are separated by the cumbersome filter and amplifier circuits, which are used to calculate R value. SpO_2 is calculated by the formula $SpO_2 = aR + b$ [9] derived from Beer–Lambert law. On the one hand, one main source of the error in this calibration model is caused by the fact that the Beer–Lambert law does not take into consideration the scattering effect of light in the tissues [10], on the other hand, the four AC-DC amplifier circuits can't maintain consistency of magnification, which causes the measurement error, meanwhile, also increases the complexity and volume of analog circuits.

To improve SpO_2 precision, Bao et al. [11] propose a new method of modeling SpO_2 with 4 variables (the peak and valley of logarithm dual-wavelength PPG), but in the way of LED excitation, they remain to adopt TDM, which carries PPG signal in 1/4 sampling time and can't make full use of the whole sampling time. In addition, some researchers also use multiple wavelengths of light sources, such as three-wavelength sources [12], five-wavelength sources [12], eight-wavelength sources [13], etc. to improve the precision of SpO_2. However, when the number of wavelengths is counted in the pulse oximeter, it is more impractical to be implemented in hardware [10].

In 2004, Li et al. propose dynamic spectrum [2,3,14,15] (DS) theory, which overcomes the principled error of the traditional R value and reduces or eliminates the influence of the individual differences (skin tissue, muscles tissue, subcutaneous tissue) and measuring conditions. It can refer to the publications of Li [2].

With the development of digital signal processing, Zhou et al. propose digital lock-in algorithm and parameter settings in multi-channel

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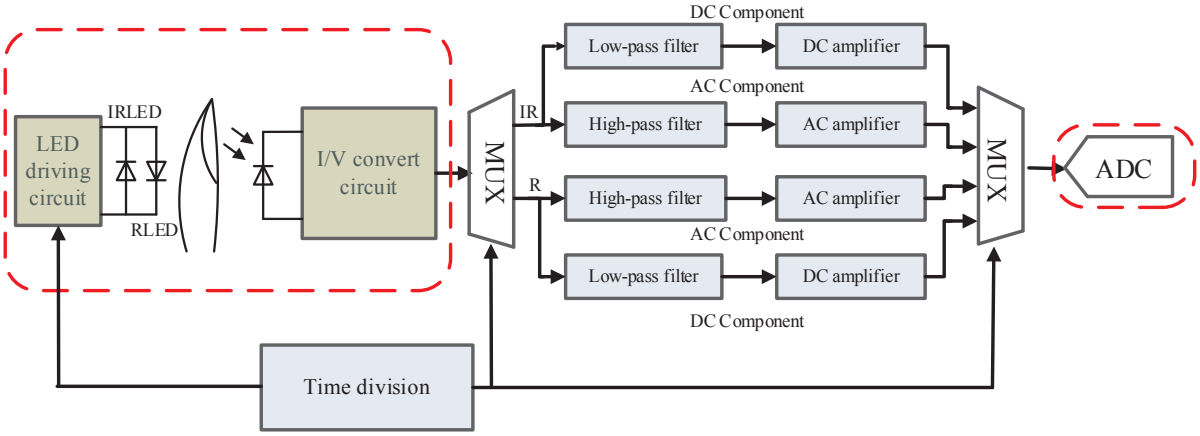

Fig. 1. The PPG signal detection of the time-division method.

sensor signal detection which demodulate a multi-channel composite signal with a high-speed computationally efficient [16]. Furthermore, the digital lock-in algorithm combined with DS can get better SNR, which has been verified in this publications of Zhou [3]. However, the carrier signals in digital lock-in algorithm they used are sine signal which is hard to generate by MCU.

Combining the previous studies with the characteristics of PPG signal detection, in this paper, a new method is proposed using bi-level pulsed frequency-division excitation and digital demodulation algorithm to improve the precision of SpO_2, which combines the advantage of frequency-division modulation and digital signal processes. So-called bi-level is high and low brightness of LEDs, so it can always carry PPG signal in the whole sampling time, however, the TDM and FDM can only carry PPG signal in 1/4 and 1/2 sampling time. So-called pulsed is square-wave signals easy to be generated by MCU. And in this paper, a corresponding digital demodulation algorithm is proposed at the same time, simple and effective.

2. Principle and theory

2.1. The SpO_2 calculation based on dynamic spectrum

Dynamic spectrum [2,3] (DS) is the spectrum composed by ΔOD^{λ_i}, where ΔOD^{λ_i} represents the difference of maximal and minimal absorbance during one cardiac cycle corresponding to the wavelength λ_i (i = 1,2,3…N). As shown in Fig. 2, because of the characteristics of abundant capillaries under the skin, absorbance of the tissue would cyclically vary with the perfusion of blood to the dermis and subcutaneous tissue. Supposing a light beam irradiates vertically down and

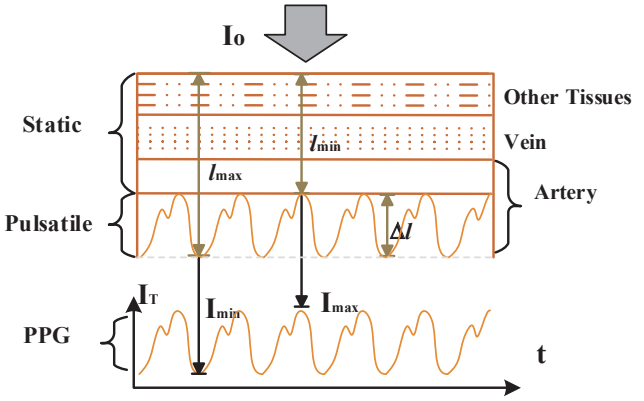

Fig. 2. The generation of the PPG signal and a simplified model of tissue. The PPG waveform consists of two parts: AC (pulsatile) and DC (baseline), so we can regard the tissue as a combination of a pulsatile part and a "static" part [9,17].

transmits through the tissues of fingertips (from the nail to the finger pad). In each cardiac cycles, the heart pumps blood to the periphery. During diastole, the transmitted light will reach its maximum intensity with the decrease of blood flow in the arteries and arterioles. Whereas, when heart contracts, due to the increasing of absorbance of the tissue, the transmitted light intensity would be the lowest. Theoretically, when measuring the PPG, the subcutaneous tissue could be simplified two parts: the pulsatile part and the static part. The absorbance of all the "static" tissue (mainly the venous blood and other tissues) remains constant during the measurement process. i.e., the pulsatile part of arterial blood is the only factor which causes the circular change of absorbance of tissues. Therefore, we can acquire the absorption information of arterial blood by measuring the alternating component of the PPG signal: subtracting the background absorption from the total absorption. The DS method which can eliminate the individual discrepancies of static tissues is more like a kind of ex *vivo* measurement of blood components [17].

DS theory is based on the modified Lambert-Beer's Law (MLBL) [18]. In Eqs. (1) and (2), I_o and I_T represent the incident light intensity and the transmitted light intensity, respectively. The maximum and minimum transmitted light intensities are represented by l_{max} and I_{min} which correspond, respectively, to the minimum optical path length $l_{\min}$ and maximum optical path length $I_{\max}$. $\alpha_j^{\lambda I}$ is component j molecular extinction coefficient at wavelength λ_i, c_j is the concentration of component j, and l represents the optical path length. L is the differential path length factor (DPF), which represents the path length lengthening caused by scattering and G depicts the scattering loss. L and G are considered invariable throughout the measurement. n is the number of components, Δl represents the optical path length of pulsating blood layer.

$$OD^{\lambda_i} = \ln\left(\frac{I_o^{\lambda_i}}{I_T^{\lambda_i}}\right) = \ln(I_o^{\lambda_i}) - \ln(I_T^{\lambda_i}) = \sum_{j=i}^{n} \alpha_j^{\lambda_i} c_j l L + G \tag{1}$$

$$\begin{aligned} \Delta OD^{\lambda_i} &= \ln(I_{\max}^{\lambda_i}) - \ln(I_{\min}^{\lambda_i}) = \ln\left(\frac{I_o^{\lambda_i}}{I_{\min}^{\lambda_i}}\right) - \ln\left(\frac{I_o^{\lambda_i}}{I_{\max}^{\lambda_i}}\right) \\ &= \left(\sum_{j=i}^{n} \alpha_j^{\lambda_i} c_j l_{\max} L + G\right) - \left(\sum_{j=i}^{n} \alpha_j^{\lambda_i} c_j l_{\min} L + G\right) \\ &= \sum_{j=i}^{n} \alpha_j^{\lambda_i} c_j L (l_{\max} - l_{\min}) \\ &= \sum_{j=i}^{n} \alpha_j^{\lambda_i} c_j L \Delta l \end{aligned} \tag{2}$$

SpO_2 is the oxygenated hemoglobin (HbO_2) as a percentage of total hemoglobin (Hb). Usually, the measurement of SpO_2 uses dual-wavelength sources (λ_1 and λ_2). When calculating SpO_2, we consider only the influence of HbO_2 and Hb components. So, according to Eq. (2), we