A Robust PPG-based Heart Rate Monitor for Fitness and eHealth Applications

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A Robust PPG-based Heart Rate Monitor for Fitness and eHealth Applications

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***Abstract*—This paper presents the design and implementation of a Heart Rate Monitor (HRM) based on photoplethysmography (PPG). This optical method is non-invasive and provides a convenient means of implementing a low-cost wearable HRM for eHealth applications. Our novel design consists of a light source that emits a modulated or unmodulated light signal through the finger tissue and measures the changes in the reflected light from the arterial blood. These reflections correspond well with the blood volume changes in synchrony with the heartbeat. Here we aim to provide a detailed description of how to design and implement an HRM device based on two approaches: 1) PPG using an unmodulated light source; 2) PPG using a modulated light source. Noise analysis in these devices enabled us to conceptualize and compare their performance and infer which one offers a better choice for optical heart rate monitoring. Based on the measurement results, the latter approach offers superior performance due to better noise cancellation and higher Signal-to-Noise Ratio (SNR) and is therefore robust against movement artefacts, power line noise, flicker noise of electronic components, as well as background environmental light interference.**

Keywords- Heart rate; Photoplethysmography; Filter; Modulated light source; Unmodulated light source

I. INTRODUCTION

Heart Rate (HR) measurement is an important part of bedside and vital sign monitors. It is also an integral part of fitness and wellness equipment. The HR is an established indicator of the overall health status of the human body. HR monitoring by using conventional skin surface electrodes has been available since 1983 [1]. In a clinical setting, the HR is obtained by measuring the R-R intervals in the electrocardiography (ECG) signal acquired with at least three electrodes (with one connected to the right leg- serving as a ground). However, a user-friendly, robust, and wearable device is needed to unobtrusively measure the HR (beats per minute - BPM) during exercise and movement when the ECG signal is susceptible to a large amount of interference from a variety of different sources. To address this need, we propose the design of a novel PPG-based HRM. The PPG signal is comprised of both AC and DC components. The AC part represents a pulsatile physiological waveform attributed to the cardiac rhythm in synchrony with blood volume changes during each cardiac cycle, while the DC component depends on the average blood volume of the arterial and venous blood, structure of the tissue, respiration sympathetic [2], and thermoregulation.

Like other PPG-based heart rate monitors, our system is comprised of a light source to illuminate the skin and a photo detector to measure the reflected light intensity changes as they pass through the skin. There are two PPG measurement operational modes: reflection and transmission. In the reflection mode both the light source and detector are placed on one side of the tissue (skin), while in the transmission mode, the tissue sample is placed between the source and the detector. The intensity of the reflected light changes based on the variations of blood volume during each heartbeat. The photodetector generates an AC electrical signal as it is sensitive to intensity variations of the received light. The generated AC signal represents a pulsatile cardiac waveform. The PPG signal can be recorded from a variety of anatomic locations including the ear, finger, or toe, where blood vessels are readily accessible at the skin surface and the AC heart beat pulses can be easily detected [3]. Different noise reduction methods are reported in the PPG literature to enhance PPG signal quality. Reference [4] used sparse signal recovery algorithms to estimate the spectrum of the PPG signal. A double ring-based PPG sensor is reported by the authors in reference [5]. Their design lowered the influence of ambient light and external force on the sensor. In this work the sensor was attached to the skin tightly and securely. Stockwell transform was used in [6] to filter unwanted noise without prior information about the R-peak position or the reference signal. The authors in reference [7] designed a magnetic earring PPG sensor and used the adaptive noise cancellation method for improving noise performance. In our previous work [8], a PPG device was designed by using a constant infrared light source (Unmodulated Light Source – ULS). In this work, we introduce a design to improve the noise performance of our previous design by using a Modulated Light Source (MLS) instead of a ULS.

The rest of this paper is organized as follows. In Section II, we focus on hardware and software design of the robust PPG HRM and compare the two design approaches using the MLS and ULS, namely the Modulated Light PPG (ML-PPG) and Unmodulated Light PPG (UML-PPG). In Section III we perform a detailed noise analysis of both designs. Measurement results are provided in Section IV, and finally the paper is concluded in Section V.

II. The PPG-HRM Design

A simplified block diagram of the UML-PPG and ML-PPG heart rate monitoring devices is shown in Figs. 1a and 1b, respectively. These HRM devices are realized by using signal conditioning in both analog and digital domains. The pulsatile activity of the heart is picked up by discrete ICs implemented on a breadboard and the processed signals are transferred to a computer for further signal processing and display. What follows is a description of the hardware and software design of PPG-HRM devices.

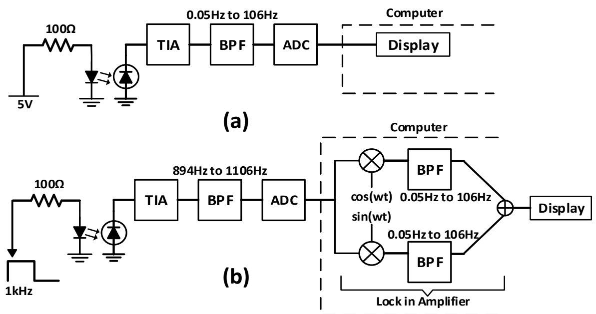


Figure 1. Block diagram of the PPG-HRM devices: (a) UML-PPG design, (b) ML-PPG design

A. *Hardware Design*

In both designs, a PPG sensor comprised of a low-cost LED and a matching photodetector operating at red light wavelength serve as reference. The photodiode converts the reflected light into a proportional electrical signal and works in zero bias or reversed bias. In our design, the photodiode operates in the reversed bias mode. This mode of operation results in a faster response compared to the zero bias mode. The reason for the better response time is due to the fact that the reversed biasing increases the depletion area of the photodiode leading to a reduction of its junction capacitance. In turn, this reduction in the junction capacitance makes the photodiode respond faster.

The LED in the UML-PPG design (Fig. 1a) is supplied by a constant 5 Volt source, while the ML-PPG design (Fig. 1b) is supplied by a 1 kHz clock signal to convert the PPG signal to the higher frequency of 1 kHz. This frequency is adequate to push the PPG signal out of all of the potential interfering frequencies such as: the 50/60 Hz power line noise, flicker noise of electronic components, and environmental light noise, and is different from the frequency of all known noise sources. In addition, using this modulation frequency results in power saving in the clock generator circuit compared to using a higher modulation frequency. For these reasons, a 1 KHz clock signal is applied to the LED. The LED shines red light through the finger and the photodetector picks up the variable reflections of light. In the reflection mode, both the LED and the photodiode are placed side by side to monitor the reflected light from the finger's pulsatile blood (arterial). The reflected waveform detected by the photodiode is made up of a non-pulsatile DC component and a pulsatile AC component. The AC part of the PPG signal is indicative of arterial blood volume increases during the systolic phase of the cardiac cycle (Fig. 2a) and decreases during the diastolic phase of cardiac cycle (Fig. 2b). Therefore, the amount of absorbed red light is increased and decreased, respectively.

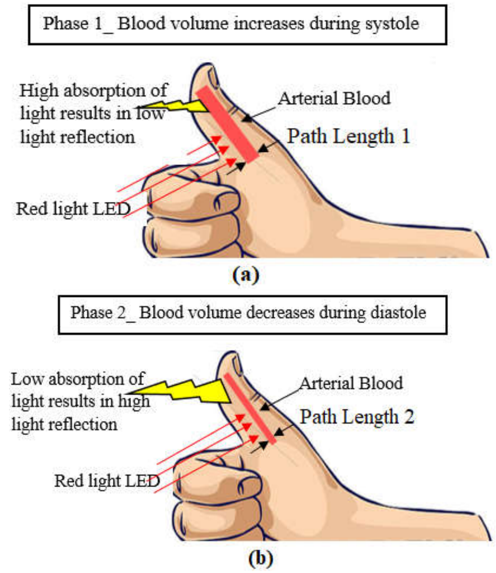


Figure 2. Arterial blood volume changes during: (a) systolic phase, (b)diastolic phase

This concept can be understood by the Beer-Lambert Law of Absorbance and could be applied to a modeled blood vessel [9] as follows:

$$A=\varepsilon(\lambda).l.c \qquad (1)$$

, where A is the absorbance, ε is the absorption coefficient of the species in the material sample at specific wavelength in lit/(mmol.cm), l is the length of optical path through the blood vessel in cm, and c is the concentration of species in the sample in mmol/lit. It can be understood that the quantity of light absorbed by the blood substance directly relates to the length of the optical path (l). The length of optical path changes with respect to the time during cardiac cycle. Fig. 2 shows that Path Length 1 during systole is greater than Path Length 2 during diastole resulting in a higher absorption and consequently lower reflection.

Variable reflection of light generates an alternating current (AC) signal in the photodiode with small changes in amplitude. This AC current is then amplified and converted to an AC voltage by a Trans-impedance Amplifier (TIA). The TIA in Fig. 1(a) is followed by a band pass filter (BPF) with a bandwidth of 0.05Hz to 106Hz, which is flat over 0.5Hz-40Hz for monitoring purposes. However, the BPF used in the second circuit (Fig. 1b) has a bandwidth of 894Hz to 1106Hz because the desired bandwidths of shifted PPG signal are up-converted to a frequency close to 1 KHz.

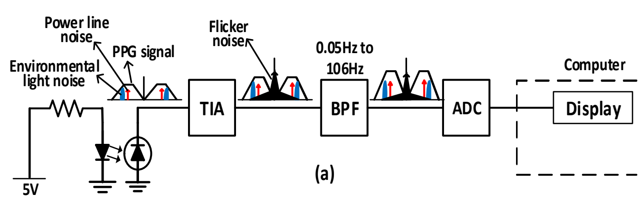
B. *Software Design*

Our second design includes one more step which requires software implementation. As the desired signal is up- converted to a 1 kHz frequency, it is necessary to return the PPG signal back to its original frequency for monitoring.

This can be realized by using a Lock-in Amplifier (LIA). This quadratic structure comprises two mixers driven by sine and cosine waves in order to convert the signal down to a base band frequency which is then followed by a band pass filter with bandwidth of 0.05Hz to 106Hz in order to detect the down-converted PPG signal (Fig. 1b). The reason for using both sine and cosine driven mixers is to remove the phase dependency of the LIA's output [10]. Digital implementation of the LIA overcomes either the gain/phase mismatch between the two paths as well as any temperature dependency. The digital implementation of the LIA is achieved by applying a mathematical algorithm to the digitized signal in MATLAB software. Analog Discovery 2 is used to serve as an interface between the designed hardware and computer sampling as well as in transferring the PPG signal from the output of designed hardware to the computer for more signal processing.

III. Noise Analysis

The frequency spectrum of the desired PPG signal as well as the undesired components at every node for both structures are shown in Fig. 3. Flicker noise (which has a greater value at a lower frequency) of electronic components and power line noise are displayed by black arrowheads at a frequency of around 0 Hz and with red color at 50/60Hz, respectively. Environmental light interference is displayed in the desired PPG bandwidth by blue color. In the ML-PPG design, the PPG signal is up-converted to a higher frequency of around 1 kHz and is far from noise interference sources. Therefore, the BPF in Fig. 3b preceding the ADC has the ability to reject unwanted low frequency noise significantly and to remove the power line noise as well as environmental light noise. The down-converter mixer in the digital domain takes the PPG signal back to the base band frequency and then the desired PPG bandwidth is detected by a BPF in the digital domain. The PPG signal is thus displayed without any contamination by any form of noise in its desired bandwidth. In contrast, the desired bandwidth of the PPG signal in Fig. 3a (the UML design) is polluted by unavoidable noise sources.



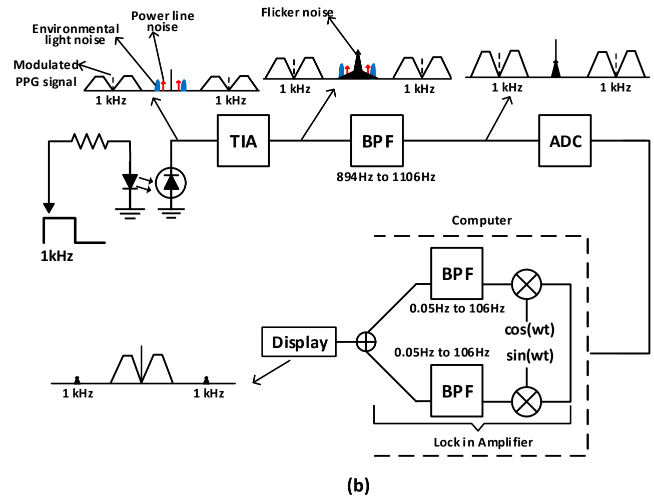


Figure 3. Noise performance of PPG device (a) with an unmodulated light source – UML, (b) with a modulated light source - ML.

IV. Circuit Implementation and Measurement Results

Based on the block diagrams shown in Fig. 1, the hardware parts for both PPG devices designed with the unmodulated and modulated light sources are shown in Figs. 4a and 4b, respectively. There are two band pass filters for better noise rejection and the gain of both structures is the same (current-to-voltage gain of 75x10^{6}). These circuits are implemented by using LMC6484 Op-Amps. The prototype circuits were implemented by using discrete electronic components connected on a breadboard. Fig. 5 shows the implemented PPG devices using a modulated light source (ML).

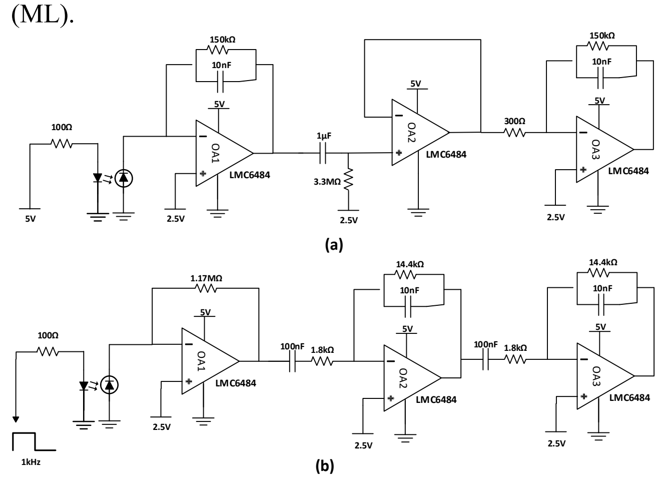


Figure 4. Schematics of PPG device (a) UML design, (b) ML design

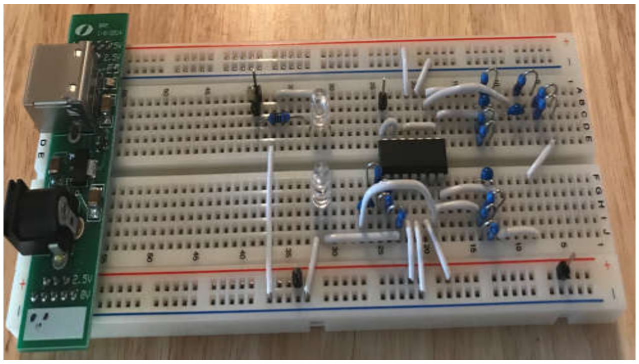


Figure 5. The Implemented PPG device with a modulated light source (ML-PPG design)

Fig. 6 shows the frequency spectrums of the PPG signal at the output of the BPF preceding the ADC based on the Fast Fourier Transform (FFT). The design with the modulated light source (Fig. 4b) has a clean response with power line noise less than -100dBV, a flicker noise less than -50dBV at zero frequency and a flicker noise less than -100dBV at other frequencies. Moreover, these noise sources are far away from the desired bandwidth of the PPG signal. In contrast, the PPG–UML device (Fig. 4a) suffers from unwanted environmental light interferences, power line noise of -18dBV, and the undesirable presence of flicker noise in the desired PPG bandwidth which contaminates the desired signal.

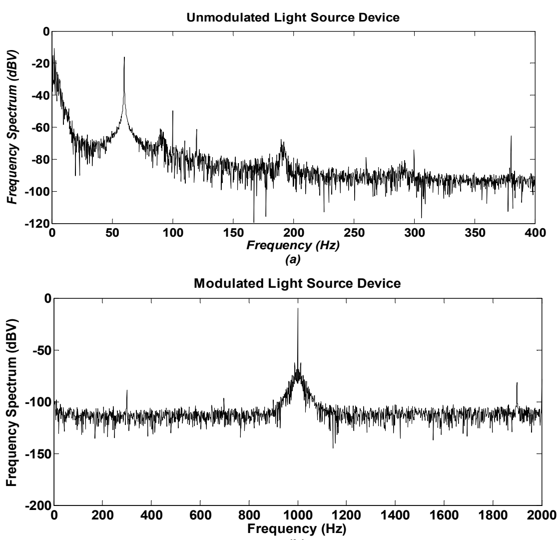


Figure 6. The frequency spectrum (a) at output of OA3 in the UML-PPG design, (b) at output of OA3 in the ML-PPG design

The output signal of the PPG devices at their last stages are shown in Fig. 7 in time domain. As we expected, the PPG device with the unmodulated light source produced inferior performance both in shape and noise level of the PPG signal (Fig. 7a). However, the second structure tracks the detailed variation of the PPG signal well resulting in clearly defined PPG signals suitable for robust heart rate monitoring (Fig. 7b).

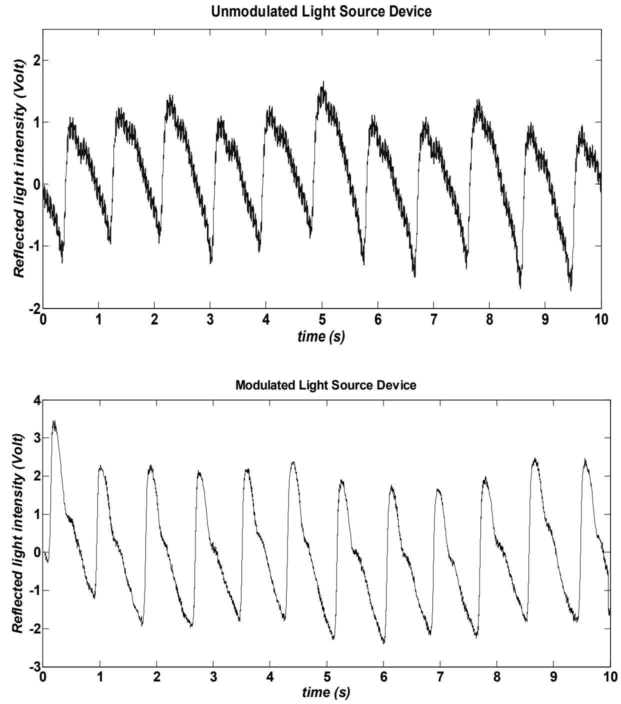


Figure 7. The blood pulsation signals captured by the PPG devices: (a) the UML-PPG design, (b) the ML-PPG design

V. CONCLUSION

A low-cost, small, and easy to implement PPG-based HRM monitoring system with two designs was presented and discussed. We showed that optical sensors with robust designs could provide user-friendly wearable HRM devices suitable for a variety of applications including fitness and eHealth. We proposed two different approaches for the design of PPG-based HRM devices, one based on a modulated light source (ML) and the other based on an unmodulated light source (UML). Noise analysis of both designs was carried out and the performance of these devices in terms of their noise rejection ability was compared. The PPG device with a modulated light source showed a clean frequency spectrum with impressive noise rejection, while the PPG sensor with an unmodulated light source was subjected to significant interferences from environmental light noise, power line noise, and flicker noise from the electronic components in the desired PPG bandwidth.

ACKNOWLEDGMENT

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