Simulation of Ventricular Tachycardia in Photoplethysmogram: Hardware Implementation

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Abstract

This study investigates a hardware implementation of a photoplethysmogram (PPG) simulator designed for testing PPG-based wearable devices, focusing on ventricular tachycardia. The simulator uses beat-to-beat interval series as an input, allowing for the utilization of publicly available ECG databases to simulate PPG signals that include episodes of ventricular tachycardia. To assess the accuracy of the hardware simulator-produced signals compared to actual PPG signals, the PhysioNet/CinC Challenge 2015 training dataset is employed, consisting of 39 recordings with ventricular tachycardia. The simulator-produced signals acquired using a wrist-worn device exhibited an RMS error of 0.72 a.u. for the episodes of ventricular tachycardia and 0.69 a.u. for the entire signal when compared to actual PPG signals. The correlation coefficient for heart rate derived from ECG and simulator-produced PPG signals was 0.98. The hardware simulator is anticipated to be a valuable tool in the development and testing of PPG-based wearable devices intended for monitoring individuals at elevated risk of sudden cardiac death due to ventricular tachycardia.

1. Introduction

Sustained episodes of ventricular tachycardia pose a life-threatening risk and can potentially result in cardiac arrest, especially in the aftermath of a myocardial infarction [1, 2]. Hence, monitoring individuals at heightened risk of sudden cardiac death from ventricular tachycardia may aid in prevention. Nevertheless, the discomfort from using arrhythmia monitoring devices with disposable electrodes makes implantable devices the only option for extended monitoring. As an alternative, photoplethysmography (PPG)-based wearables might be considered due to low cost, wide availability, and minimal risk of infection.

The majority of research has focused on developing techniques for utilizing PPG to identify atrial fibrillation [3–6]. Only a few studies have addressed the detection of premature atrial and/or ventricular beats [7–9], bradycardia, and ventricular tachycardia [10]. Unlike the detection of atrial fibrillation and ectopic beats, which can be detected relying solely on heart rhythm information, the detection of ventricular tachycardia is more demanding. Ventricular tachycardia must be distinguished from nonfatal tachycardias (e.g., atrial tachycardia, AV node tachycardia) and may manifest in the PPG signal by markedly reduced pulsations due to diminished hemodynamics.

Another obstacle hindering the development of ventricular tachycardia detectors is the lack of annotated databases with PPG recordings containing arrhythmia episodes. One way to address this issue is by developing the detectors using a PPG signal simulator that enables control over various signal and arrhythmia properties. Previously, we developed a software simulator, accessible on Physionet, designed to model PPG signals with episodes of paroxysmal atrial fibrillation [11]. Subsequently, we demonstrated that the simulator is equally well-suited for simulating ventricular tachycardia [12]. The main advantage is that publicly available ECG databases containing the annotated episodes of ventricular tachycardia, such as MIT-BIH Malignant Ventricular Arrhythmia or Creighton University Ventricular Tachyarrhythmia, can be employed to generate PPG signal database.

So far, the simulator has only been applicable for developing and testing detectors within a computer environment. In this study, we investigate the hardware implementation of the simulator, which could facilitate the testing of PPG-based wearable devices during their development phase, eliminating the need for patient participation.

2. Methods

2.1. PPG simulation model

The model signal is generated by arranging individual pulses based on the beat-to-beat intervals, resulting in a connected signal where the shape of each pulse is determined by a linear combination of a log-normal and two Gaussian waveforms [11]. To create a connected discrete-time simulated PPG, the separate PPG pulses are placed at the times of heartbeat occurrences.

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2.2. Hardware implementation

The hardware comprises an LPC1768 microcontroller unit that executes the PPG simulator and interfaces with other on-board components, see Fig. 1. A 16-bit digital-to-analog converter, DAC8551, is utilized to convert the digital PPG pulses generated by the simulator into analog signals. The digital-to-analog converter provides control over the amplitude and timing of the PPG pulses when transmitting the analog signal to the current source. The programmable current source, LT3092, is used to regulate the current flow through the LED in accordance with the morphology of the generated PPG pulse. An LED serves as the light source in the hardware setup, selected based on its wavelength and intensity to ensure compatibility with PPG measurements. Additionally, an on-board RGB sensor, TCS34103M, and photodiode, VEMD5010X01, are incorporated to capture the duty cycle and provide additional information, such as the wavelength of the emitted light, from the wearable device under testing.

2.3. Database

Photoplethysmogram signals from the training set of the PhysioNet/CinC Challenge 2015 database, referred to hereafter as actual PPGs, are used to evaluate the accuracy of the signals produced by the hardware simulator. The set contains 294 recordings with life-threatening arrhythmia; however, only recordings meeting the following criteria were included: (1) 5 or more ventricular beats with heart rate higher than 100 bpm; (2) at least one ECG lead eligible for R-peak detection, (3) underlying sinus rhythm. In total, a subset of 39 recordings with the episodes of ventricular tachycardia are used in the study.

ECG and PPG signals were sampled at 250 Hz with 12-bit precision. ECGs were used for extracting RR interval series, which served as an input to the simulation model. PPGs were filtered with a bandpass zero-phase finite impulse response filter with cut-off frequencies at 0.5 Hz and 15 Hz. Each recording was trimmed into 30-second segments, ensuring the inclusion of at least one tachycardia episode. From each PPG signal, a single pulse template with distinctive morphology, free of noise and artifacts, was extracted and employed for estimating the model parameters.

2.4. Performance evaluation

A wrist-worn device developed at Biomedical Engineering Institute of Kaunas University of Technology was employed to capture a simulator-produced PPG signal at a sampling frequency of 100 Hz. To facilitate the comparison, the actual PPG signal was resampled to 100 Hz.

The adequacy of the PPG signal produced by the simulator is assessed by calculating the RMS error E between the actual and acquired PPG signals, both during episodes of ventricular tachycardia and throughout the entire signal. E was calculated for each recording and subsequently averaged to derive the overall performance measure, denoted as $\overline{E}$. Cross-correlation was employed to align the actual and simulator-produced signals. Before computing E, both signals were normalized to the unit amplitude of the first PPG pulse.

An association between heart rates calculated for actual and simulator-produced signals is represented using a scatter plot. The relation is assessed using linear regression and presented as a Pearson correlation coefficient.

3. Results

An example of an actual PPG signal and a hardware simulator-produced PPG signal during an episode of ventricular tachycardia is presented in Fig. 2. The ventricular tachycardia episode, as captured by the wrist-worn device, closely resembles the actual PPG signal.

Fig. 3 displays the individual RMS error E between actual and simulator-produced PPG signals, along with the average RMS error $\overline{E}$, which is 0.69 a.u. for the entire signal and 0.72 a.u. for tachycardia episodes only.

Fig. 4 shows the association between heart rates calculated from simulator-produced PPG, actual PPG, and ECG signals. The correlation coefficient for heart rate derived from ECG and simulator-produced PPG signals, as well as for heart rate derived from actual and simulator-produced PPG signals, are 0.98 and 0.81, respectively.

Fig. 5 sheds a light on the source of the large disparities between the heart rates derived from actual and simulator-produced PPG signals. In certain instances, the PPG quality is poor, with pulses during ventricular tachycardia barely recognizable, thus resulting in large E.

4. Discussion

This paper evaluated the simulation of PPG signals with ventricular tachycardia by integrating the phenomenological PPG model into a hardware simulator. This research is driven by the widespread use of personal wearable devices equipped to capture PPG, presenting an opportunity to monitor individuals at a heightened risk of ventricular tachycardia. Notably, approximately 50% of cardiovascular deaths are linked to sudden cardiac death due to life-threatening arrhythmias [2]. Consequently, the developed hardware simulator could offer benefits in the development of wearable devices for monitoring various patient groups, including those with acute coronary syndrome or structural heart disease.

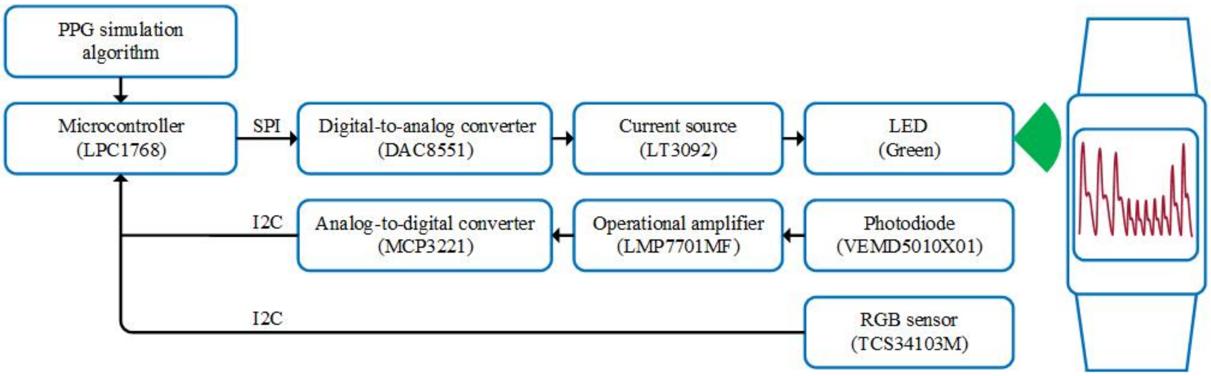

Figure 1. Block diagram of a hardware PPG simulator.

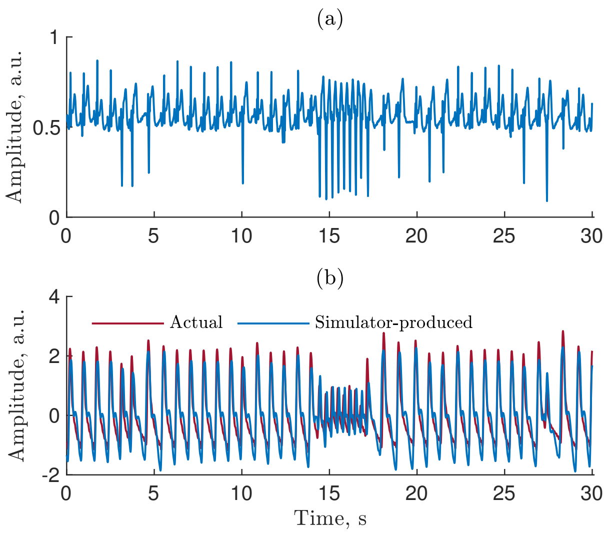

Figure 2. An example of (a) an ECG and (b) actual and simulator-produced PPG signals with an episode of ventricular tachycardia (recording v837lr). Ventricular tachycardia starts at the 14^{th} second and stops at the 17^{th} second.

The present hardware simulator is not the sole solution for evaluating PPG-based wearable devices. Notable examples are PPG simulators manufactured by WhaleTeq Co., Ltd, exemplified by the HRS200, HRS100+ or AECG100 device supplemented by PPG-2R-880/PPG-2R-940 reflectance PPG module. These simulators replay prerecorded data or customized waveforms repeatedly and enables the adjustment of key PPG waveform characteristics, including the systolic and diastolic peaks, as well as the dicrotic notch. However, the present approach to generating PPGs with tachycardia (and other rhythms such as premature beats, bradycardia, atrial fibrillation) offers other unique features such as generating a synthetic signal in a beat-to-beat manner by only supplying heart rate information, ability to change pulse morphology and rhythm type.

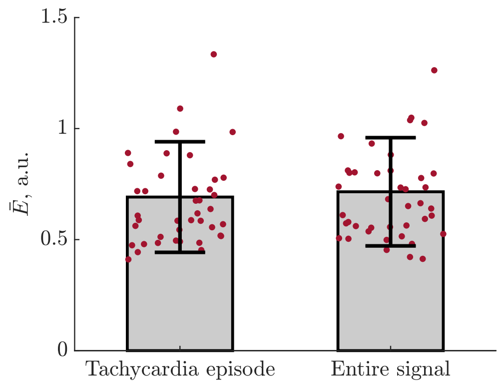

Figure 3. The average RMS error $\overline{E}$ between actual and simulator-produced PPG signals. Red dots represent individual E. Error bars show mean $\pm$ s.d.

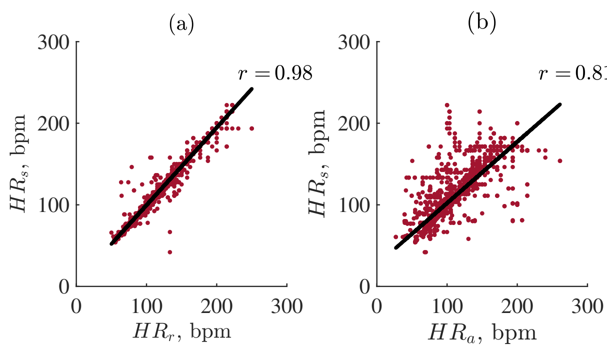

Figure 4. A scatter plot for associations between heart rates calculated for (a) ECG and simulator-produced PPG signals and (b) actual PPG and simulator-produced PPG signals. Here HR_r, HR_a, and HR_s represent the heart rates obtained from ECG, actual PPG, and simulator-produced PPG signals, respectively.

5. Conclusions

The present study demonstrates that, in most cases, PPG signals produced by the hardware simulator accurately

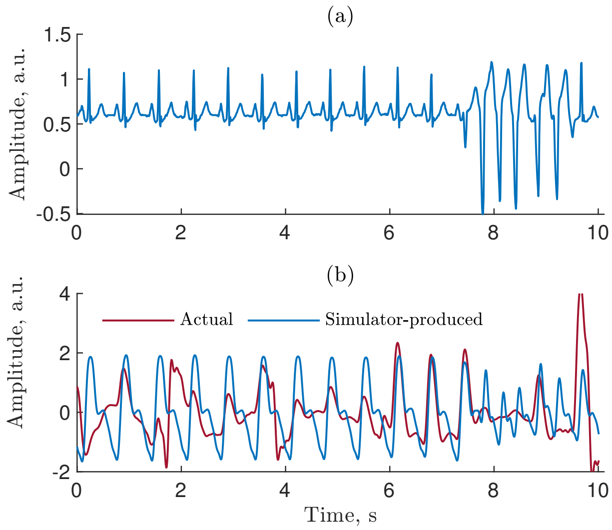

Figure 5. An example of low-quality PPG with barely recognizable pulses during ventricular tachycardia, leading to discrepancies in the derived heart rates between actual and simulator-produced PPG signals (recording v632sr). Ventricular tachycardia starts approximately at the 7^{th} second.

replicate episodes of ventricular tachycardia. The hardware simulator is expected to serve as a valuable tool in both the development and testing phases of PPG-based wearable devices intended for monitoring of individuals at elevated risk of sudden cardiac death due to ventricular tachycardia.

Acknowledgements

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