Mechanically Flexiable Biosensor for Detection of Photoplethysography

Kian Davoudi, Moein Shayeganyan, Kouhyar Tavakolian and Bozena Kamisnka
Simon Fraser University, 8888 University Drive, Burnabyr, Canada

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Abstract: Acute cardiovascular failure could be detected by continuous monitoring of electrocardiogram (ECG). While electrode allocation on skin challenges quality of electrocardiograph, beat-to-beat heart rate obtained from photoplethysmography (PPG) could be used to indicate cardiovascular activity as an alternative to heart rate from ECG. In this paper we proposed a mechanically flexible PPG sensor integrated on thin layer of polymer. Mean and standard deviation of beat-to-beat heart rate was obtained from a flexible PPG sensor and compared to the beat-to-beat heart rate obtained from a commercial ECG and PPG devices. The standard deviation of beat-to-beat heart rate from ECG and PPG intervals were analysed by the Bland and Altman analysis. The corresponding 95% limits of agreement were estimated as 0.034 to -0.01 for PPG flexible sensor compared to ECG and as 0.0081 to 0.0037 for PPG flexible sensor compared to PPG commercial device. The good correlation between the measurement results demonstrated capability of our proposed mechanical flexible PPG sensor to be used as practical alternative to ECG for heart rate variability (HRV) analysis.

1 INTRODUCTION

Increasing cost of healthcare and accelerated aging of society incite hospitals and other medical caregivers to look for solutions that are inexpensive, yet maintain proper quality of care. The prospects of efficient remote health and activity monitoring using biosensors have recently gained a lot of interest and stimulate research in the area of wearable electronics. Technological advances in sensors, wireless communication and integrated circuits have brought about small, inexpensive wearable physiological monitors. These devices are usually capable of sensing one or more vital signs, e.g. heart rate, body temperature and blood pressure, then communicating the acquired data to a local or remote processing and interpretation centre.

The conventional wet adhesive Ag/AgCI Electrocardiography (ECG) electrodes are used almost universally in clinical applications. They provide an excellent signal but are cumbersome and irritating for mobile users. The chief advantage of the standard clinical wet electrodes is their strong adhesion to skin. However, their main issue is long-term use problematic for a patient comfort (Chi, 2010). Adhesive wet electrodes stay fixed to specific, clinical standard locations on the body. These standard electrodes not only adhere well to a body, but also are robust, inexpensive and simple.

As an alternative, dry and noncontact electrodes without gel have been introduced to address the comfort issues with the adhesive electrodes. These electrodes offer few advantages for patients with extremely sensitive skin burn units (Griffith, 1979) and neonatal care (Bouwstra, 2009). However the dry electrodes are much more difficult to secure against the skin and they have yet to achieve the acceptance for medical use. They also add cost and complexity in active electrode circuitry. For these technologies to be useful mechanical solutions must be devised to place the dry electrodes in the proper position or an alternative application must be found.

Photoplethysmography (PPG) is a non-invasive method based on the reflection of light from peripheral tissue. PPG waveform contains valuable physiological information, such as respiration and heart rate.

Pressure disturbance induced by the PPG probes placed on a forehead affects quality of a PPG waveform, and leads to the inability to measure heart rate. In addition, due to rounded and optically inhomogeneous surface properties of the skeleton of

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forehead, alternation in forehead PPG sensors changes distribution of the light scattered to tissue; hence, introduces noise in backscattered light reaching PPG sensors (Dresher 2006).

In this paper we proposed a mechanically flexible PPG sensor integrated on plastic substrate. The advantages of our proposed flexible PPG sensor can be summarized as follows:

a. It monitors heart rate without interfering user's daily routine for prolonged period of time.
b. It minimizes the pressure disturbance induced by the placement of a PPG sensor.
c. It requires inexpensive integration methodology and less complex electronic circuitry.

The objective of this paper is to assess feasibility of integrating a PPG sensor on plastic (polymer) for continuous monitoring of heart rate. We have developed an algorithm to extract intervals between successful cardiac cycles from five healthy subjects. The data have been recorded from the proposed flexible PPG sensor in parallel to the ECG and PPG from commercial pulse oximetry and ECG devices. HRV analyses has been conducted in time frequency domain on the PPG waveform acquired from flexible sensor in addition to the ECG and PPG from the commercial devices.

2 METHODOLOGY

2.1 Principal of Operation

PPG assessment is available through the commercial pulse oximetry devices. The estimation of pulse oximetry requires use of two wavelengths, red and infrared (IR) (~640nm, ~940 nm respectively). Red Light Emitting Diode (LED) is being used for estimation of oxygenated hemoglobin and IR LED is being used for estimation of deoxygenated hemoglobin.

The conventional PPG sensors are available in two models. In the first model, light emitter is located opposite side of photoreceptor. In this case, the emitted light passes through skin tissue and a photoreceptor on the other side of tissue detects portion of the light passed through. This type of PPG sensor requires a mechanical clip to secure location of photodiode, for example on a finger and ear. In the seconds model, LED and photo detector are placed side by side on top of a tissue. In this case, the photo detector observes the reflected light in the same planar surface as the LED and sensor is attached using a tap against the skin (Dresher R. 2006). Our sensor follows principles of the second model.

2.2 Design of a PPG Circuit

Since a PPG waveform can be obtained from one LED, we have avoided the excessive use of LEDs for our application, and LED (Kings light 530 nm) has been chosen as the primary source of light. To detect the reflected light, a photodiode (APDS-9008) was utilized. To amplify the output waveform of the photodiode, an operational amplifier (MCP6001) was used. In addition, passive components with small foot-print (0603Metric) were chosen for filtering the high frequency noise from light sensor.

We have developed an electric circuit to obtain a PPG waveform from the APSD-9008 photodiode. As demonstrated in Figure 1, an electronic circuit consists of five sections: light, photodiode, high pass filter and active amplification. The circuit has been designed and simulated using Eagle software.

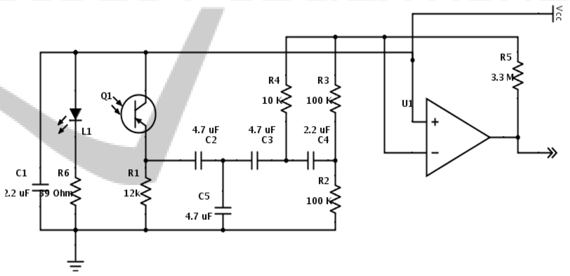

Figure 1: The schematic of a PPG circuit obtained with the Cad soft Eagle 6.5.

2.3 Fabrication Process

For fabrication of PCB sketch on flexible materials, two components were used; polymer (IJ-220, Novacentrix) and conductive silver ink (JS-25HV, Novacentrix). Diamatix inkjet printing technology (Fujifilm, DMP-2831) was used to fabricate the PCB sketch on a polymer.

To print JS-B25HV silver ink on IJ-220 PET substrate (figure 2.a), the jetting waveform was adjusted to single size drops, and 6 nuzzles were used during printing process. Because the JS-B25HV is a water-based ink, each sample was also cured for 60 minutes at 100^{oc} in an oven. The following setting were adjusted on Diamatix to pattern conductive lines on IJ-220:

- Resolution of drops = 20 um
- Temperature of Surface of the plate = 42^{oc}
- Temperature of cartridge = 28^{oc}

As demonstrated in figure 2.b electronic components of the PPG circuit were manually attached on the flexible substrate using silver epoxy (8331, MG Chemicals). Curing time of the silver epoxy was reported 20 minutes by manufacture. However during testing and evaluation phase, we noted a high frequency noise in the PPG signal that was introduced by the silver epoxy. Hence, PCB layers were cured for 24 hours at room temperature where consequently the noise was disappeared.

Figure 2 illustrates the flexible characteristic of our mechanical sensor after integration of electrical components on the IJ-220. The flexibility and simplicity are the main advantages of our sensor in providing comfort for prolong usage on skin and within accessory or special clothing.

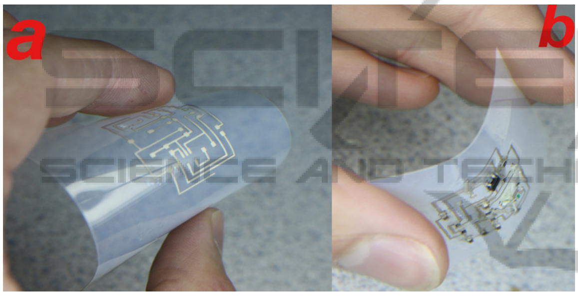

Figure 2: A: demonstration of the PPG sketch fabricated on IJ-220. B: Integration of electrical component of the PPG circuit on metalized layer of polymer.

Figure 2.b exhibits the location of the LED and the photodiode on our flexible mechanical PPG sensor. Clearly, transmitter and receiver are located on a same planar surface, and reflection mode is the operation principle. In this design, all pressure induced by the attachment of a sensor to the skin is endured by the backside of the sensor. Hence, the minimal disturbance is interfered into the photo diode. This fact establishes the second main advantage of our proposed PPG sensor, which is minimizing the pressure disturbance included in the available commercial PPG sensors such as Nonin 8600 pulse oximetry.

2.4 Experimental Setup

Five healthy subjects were voluntarily chosen to install our fabricated sensor, Nonin 8600, and Biopac ECG100c on them. The selected subjects were non-smokers, aged between 22-32, males, and without any known physiological diseases. They were asked to sit on a chair and breathed at normal pace.

The PPG and ECG signals were obtained from two sensors: flexible PPG sensor and Nonin 8600. The ECG was also obtained from Biopac ECG100C. Two PPG sensors were located side by side on a forehead of each subject, and the Biopac electrodes were attached to the chest. One minute of the PPG and ECG signals was recorded by NI 9205 NI DAQ at 1kHz sampling rate and stored on a personal computer.

3 SIGNAL PROCESSING

The changes in a PPG waveform arise from the variation in path-length between source and detector (Schäfer 2012). The typical waveform of a PPG cycle can be divided into two parts: the anacrotic phase and the catacrotic phase. Anacrotic phase is the rising part of the pulse due to systole, which happens shortly after QRS complex in ECG. Catacrotic phase corresponds to the cardiac diastole and often contains a secondary peak so called dicrotic north, an effect diminishing with aging and increasing arterial stiffness (Allen 2007).

In this study, we have developed an algorithm to identify anacrotic pitch of the PPG waveform. An interval between anacrotic peaks was identified as a full cardiac cycle, and time difference between pair of cardiac cycle was measured as beat-to-beat heart rate. We have used this automated algorithm to detect beat-to-beat heart rate from the PPG waveform obtained by the flexible and the commercial PPG sensors. The initial PPG waveform from both sensors and ECG was fragmented into different frames. Initial constants such as maximum and minimum expected peak to valley values, maximum window size, and maximum or minimum window change were set. The algorithm actively allocated a variable sized window to each frame. A peak detector was used to measure the peaks, valleys and the associated time index values, within each window frame. The size of the window was then varied based on initial window size, interval between the peaks and pulse width to frame ratio. The peaks, valleys and their time index of each window were padded in to a vector. Hence, the beat-to-beat heart rate was obtained by differentiating consecutive time index values. The initial ECG waveform was also fragmented into different frames. Since QRS height is larger than anacrotic pitch in PPG waveform, the initial constants were adjusted differently than in PPG. The rest of the algorithm remained the same as in the case of PPG. Hence, the peaks of R wave of ECG were automatically obtained.

As demonstrated in Figure 3, while in the commercial pulse oximeter devices, a high-pass filter is applied to remove the respiration component of the PPG signal by the manufacturer of device (Karnel, 2013), the AC component of the signal corresponding to the respiration induced intensity variation was contained in our study (Nilsson, 2003).

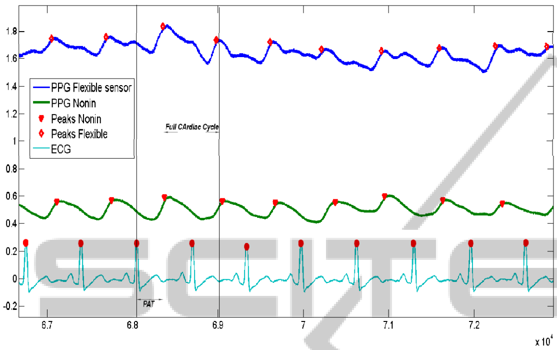

Figure 3: Demonstration of PPG waveform by flexible and commercial PPG sensor and ECG waveform.

In the Figure 3 the PPG waveform obtained by flexible PPG sensor demonstrates variation in base line of the PPG waveform, which corresponds to the respiration component of a PPG signal. The red dots correspond to the occurrence of anacrotic pitch in PPG waveform. The PPG obtained from the commercial pulse oximeter was high-passed filtered by the manufacturer. The red dots correspond to the anacrotic pitch were detected by the automated algorithm as described in section 3. This algorithm was also applied to the ECG waveform and the red dots on ECG waveform correspond to the occurrence of each successful cardiac cycle extracted automatically.

4 RESULTS

According to Table 1, the mean and standard deviation of beat-to-beat heart rate were calculated from the PPG obtained by the flexible sensor, the PPG obtained by the commercial pulse oximetry and the ECG for each participant. In Table 2 the difference between the means of heart rate from two PPG sensors were compared to the ECG. This comparison was fulfilled by estimating the mean square error (MSE) for means of PPG from the flexible sensor and the means of ECG, in parallel to the means of PPG from the commercial device to means of the ECG.

Table 1: Mean and standard deviation of beat-to-beat heart rate from five subjects.

	Mean ECG	SD ECG	Mean PPG Nonin	SD PPG Nonin	Mean PPG flex	SD PPG flex
#1	0.09259	0.00655	0.09254	0.00998	0.09268	0.00826
#2	0.08656	0.00756	0.086849	0.0144	0.08881	0.01490
#3	0.08763	0.07656	0.087754	0.00806	0.08794	0.01054
#4	0.08658	0.00735	0.08637	0.00654	0.09244	0.02437
#5	0.09318	0.00777	0.09391	0.00690	0.10599	0.03662

Table 2: Mean square error of means of beat-to-beat heart rate from PPG flexible to ECG and PPG commercial device to ECG.

Subjects	MSE PPG commercial to ECG	MSE PPG flexible to ECG
#1	$13*10^{-10}$	$4.4*10^{-9}$
#2	$4*10^{-8}$	$2.5*10^{-6}$
#3	$7.1*10^{-9}$	$4.9*10^{-8}$
#4	$2.23*10^{-8}$	$1.7*10^{-5}$
#5	$2.679*10^{-7}$	$8.2*10^{-5}$

Furthermore, as demonstrated in Table 3 and Figure 4, the degree of agreement between the PPG from the flexible sensor to PPG from the commercial oximeter and ECG was assessed using Bland and Altman analysis. This analysis indicated the mean of Standard Deviation (SD) ratio and corresponding 95% limits of agreement, 0.034 to -0.010 for PPG flexible compared to ECG and 0.0081 to 0.0037 for PPG flexible to PPG commercial.

Table 3: Bland and Altman analysis of SD of beat-to-beat heart rate.

	PPG Flex to PPG nonin	PPG Flex to ECG	PPG Nonin to ECG
95% upper limit	0.03632	0.03482	0.008131
Mean	0	0	0
95% lower limit	-0.01679	-0.01089	-0.003736

Degree of similarity between bit-to-bit heart rates obtained from flexible PPG devices was compared to the heat rates obtained from commercial PPG device and ECG, as demonstrated in figure 5. This analysis demonstrates strong similarity between two PPG signals.

Figure 6 presents the MSE between two sets of measurements for all 5 subjects. The first measurement is MSE between the mean of the heart rate recorded by the commercial available pulse oximetry and the one of Biopac ECG, as the benchmark. The second measurement is MSE

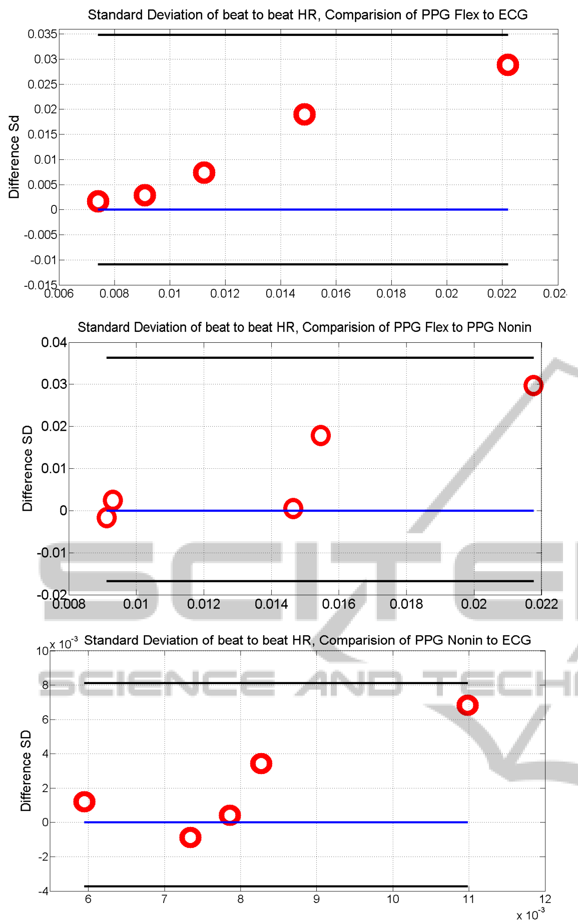

Figure 4: Bland and Altman analysis, comparison of SD of beat-to-beat heart rate from PPG flexible to ECG, PPG flexible to commercial PPG and PPG from commercial device to ECG.

between the mean of the heart rate estimated by our flexible PPG sensor, and the one of the Biopac ECG. In this figure, the horizontal line indicates the MSE for each subject, and the vertical lines reveal the MSE standard deviation between all subjects.

Evidently, a consistency can be noticed between the two error bars. That is, the MSE of the 5th subject is greatest in two sets of measurements, and the best estimation in both sets belongs to subject number 1. This consistency validates the heart rate variability measured by our fabricated flexible PPG sensor. In addition, the very small MSE between the heart rate means between the flexible PPG sensor and the Biopac ECG establishes the high accuracy of our fabricated sensor. Even though this MSE is higher than the one between the commercial PPG and the Biopac ECG, but our sensor benefits from a lower cost and less complexity. Also, this higher error can be inferred to the lower accurate laboratory facilities available in academics compared the enhanced high accuracy fabrication facilities in industry.

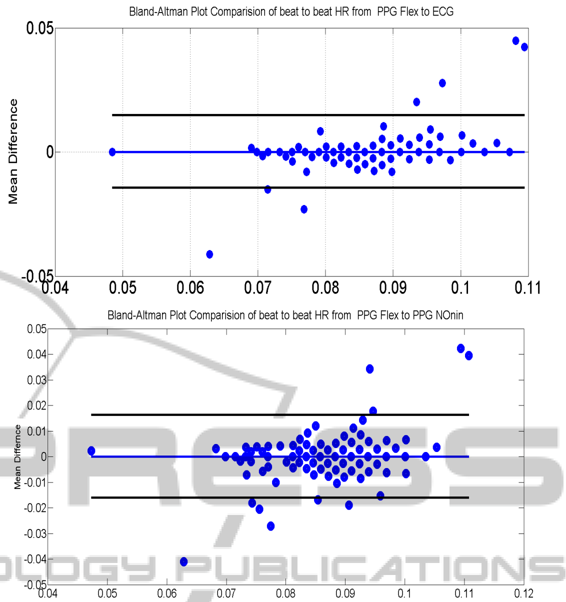

Figure 5: Bland and Altman analysis between heart rate obtained from PPG flexible to PPG commercial and ECG as well as PPG from commercial device to ECG for subject number 3.

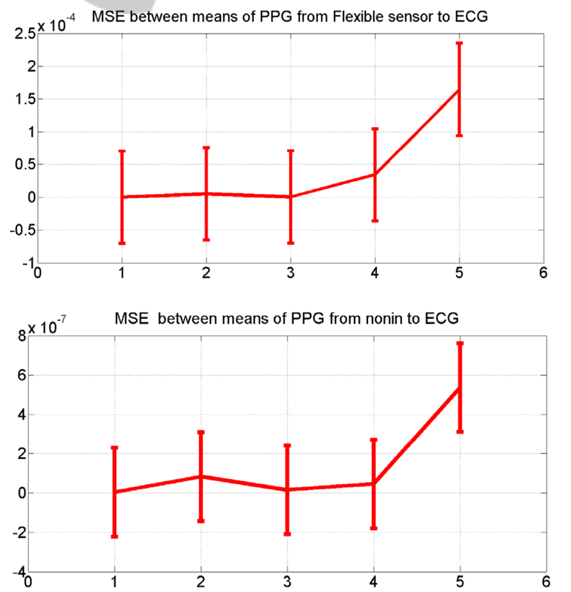

Figure 6: MSE between means flexible PPG sensor of heart rate from five subjects to ECG, in parallel to MSE between means of heart rate from commercial PPG to ECG.

5 CONCLUSIONS

The proposed flexible PPG sensor has been studied as an alternative solution to the commercial PPG sensor and the ECG device for continuous monitoring of heart rate. Standard deviation and mean of intervals between cardiac cycles were analysed. Accuracy of this sensor and the algorithm were analysed by Bland and Altman analysis. The MSE between the mean values of beat-to-beat heart rates from flexible PPG sensor and those of Biopac ECG was compared to those between the commercial PPG and Biopac ECG. The results confirm feasibility of obtaining heart rate from the flexible PPG sensor for analysis of heart rate variability.

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