Smart Photoplethysmographic Sensor for Pulse Wave Registration at Different Vascular Depths

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Abstract—The aim of this paper is to propose a smart optical sensor for cardiovascular activity monitoring at different tissue layers. Photoplethysmography (PPG) is a noninvasive optical technique for monitoring mainly blood volume changes in the examined tissue. However, different important physiological parameters, such as oxygen saturation, heart and breathing rate, dynamics of skin micro-circulation, vasomotion activity etc., can be extracted from the registered PPG signal. The developed sensor consists of 32 light emitting sources with four different wavelengths, which are located to the four different distances from four photo detectors. Compared to the existing sensors, the system enables to select the optimal LED (light emitting diode) and photo detector couple in order to obtain the pulse wave signal from the interested blood vessels with the highest possible signal to noise ratio. In this study, the designed PPG sensor was tested for the pulse wave registration from radial artery. The highest efficiency and signal to noise ratio was achieved using infrared LED (940 nm) and photo-diode pair.

I. INTRODUCTION

Photoplethysmography (PPG) is a noninvasive optical technique for monitoring mainly blood volume changes in the examined tissue. Light from a light source, e.g. LED (light emitting diode), laser, halogen lamp, is emitted to the examined tissue, where it is scattered and absorbed. The transmitted or back-scattered light intensity changes from the tissue can be detected by using photo-diode. This technique has been clinically widely used for example in pulse oximetry systems, where the blood oxygenation rate is calculated based on the simultaneous amplitude measurement of PPG signal on two or more wavelength bands [1]. However, the research and application areas of the PPG technique have been expanding during the recent years. The PPG signal registration and analysis has been used for heart and breathing rate measurement, heart rate variability analysis, pulse transit time, arterial stiffness and vasomotion estimation [2]. PPG sensors are designed mainly for the pulse wave registration from peripheral vascular beds, such as finger, ear lobe, forehead etc. Nevertheless, the pulse wave registration from the artery is needed in order to exclude the influence of the peripheral blood vessels (arterioles, capillaries) and to estimate the stiffness changes of the central arteries or certain segment of artery [3].

Our proposed system consists of 32 LEDs in four different wavelengths and four photo-diodes. Distances between the photo-diodes and LEDs vary to analyze different tissue layers. LEDs can be grouped in order to analyze automatically larger tissue areas without moving the sensor on the skin. The sensor is controlled by the previously proposed miniaturized monitoring device [4]. The designed PPG sensor is tested for the pulse wave registration from radial artery.

In the following section we present background on the PPG measurement and discuss related work together with some possible application areas. In section 3 we describe the methodology that is used to analyze the PPG signal. Section 4 describes the experiments and initial results. Finally section 5 discusses the future work and gives some conclusions.

II. STATE OF THE ART

The alternating current (AC) component in the PPG signal is synchronous with heart cycle and it is related to the heart generated pulse wave [2]. The pulse waveform carries important clinical information about the arterial system, including the micro-circulation of the skin. The detection of the PPG signal from different tissue layers may give a better understanding of the changes in the arterial system [5]. Techniques and applications to obtain the information from deeper tissue layers, such as blood flow monitoring in the tibial anterior muscle [6] [7], foetal oxygen saturation monitoring [8], estimation of pulse wave velocity in larger arteries [9] have been developed.

The light penetration volume-depth in skin depends on the selection of the wavelength [10]. Absorption of the light in the visible and near infrared wavelengths depends mainly on chromophores such as water, hemoglobin, and melanin. There is an ”optical window”, where the light is less absorbed by tissue. Therefore, red and near infrared light can penetrate deeper layers of tissue than shorter (green, blue) or longer (infrared) wavelengths and the absorption of blood is more prevalent. In addition to the absorption, tissue is a highly scattering medium, where the light behaves diffusely. Photons are scattered from cell membranes and organelles. Generally, in shorter wavelengths the light is more scattered than in longer wavelengths. Due to the scattering and absorption properties of the tissue there is possibility to obtain the PPG signal from different tissue layers, which is based on the combination of wavelength and distance between the LED and photo-diode (PD) [11]. In addition, earlier
studies, using extremely short light pulses and time-of-flight analysis, have reported that the distance photons travel in tissue is approximately 4-6 times the distance between the light source and photo detector [12]. Generally, in case of short distance between the LED and photo-diode, and short wavelength (green, blue), the penetration volume-depth is small. On the other hand in case of long distance between the LED and photo-diode, and longer wavelength (near infrared), the penetration volume-depth is larger.

PPG sensor development for the signal registration from the different penetration volume-depths, has been described earlier [6] [7] [13]. The advantage of our proposed sensor is to combine PDs and LEDs with different wavelengths into groups so that they can be driven independently. The distance between the LEDs and PDs and selection of the wavelength in proposed smart PPG sensor has been made based on the previously mentioned studies.

III. SENSOR ARCHITECTURE

The proposed sensor architecture is part of the previously developed system. The architecture and the functionality of the system has been discussed in [4] and [14]. Smart PPG sensor consists of LED and photo-diode array with control logic and optical measurement functionality.

A. Sensor

Figure 1 depicts the architecture of the optical sensor module. There are four independent LED and PD groups, and two independent channels. A channel means that all signals that are measured in this particular group are connected with one particular analog front-end (AFE) module. In total there are two identical AFEs integrated into one AFE module that are working in parallel. Each group, separated with red borderline, has one PD, green (G), red (R) and two infra-red (IR) LED emitters.

![Fig. 1. Structure of optical sensor array](image1)

Four different wavelengths in each group are used. Green LED 560 nm, red LED 660 nm, inner infrared LED (IRn-1 and IRn-3) 880 nm, and outer infrared LED (IRn-2 and IRn-4) 940 nm. All vertical and horizontal distances between LEDs and PD-s are based on the studies, mentioned in state of the art.

1) Light Source Driver: Figure 2 depicts the hardware block diagram of the smart PPG sensor. All digital control signals are marked with dotted line. Multiplexing of control signal is done by using serial in to parallel out shift registers that drive analogue switches. Each shift register controls two switches and each switch in turn two LED pairs. A LED pair means that there are two LEDs, one LED anode is connected to another’s cathode so that they can be turned on alternately. With current configuration we can drive independently up to 16 LEDs per one channel. In Figure 2 an emitter-pair is drawn as emitter box. For simplicity there are four two emitter boxes drawn and inside each one there are two emitters. Depending on the used analogue switch, there can be any number of LED pairs inside each box. The switching of optical sensor is initiated by the data processing module (a). Real switching of optical sensor is performed by the integrated Control Logic (b). The current for each LED is digitally controlled with 8-bit accuracy. Current configuration allows to set up to 100 mA LED current for each LED independently by the AFE.

![Fig. 2. Hardware block diagram](image2)

All four PD-s are driven by one analogue switch. For each channel there are two PD-s. Solid blue lines on the Figure 2 mark analogue signals. From the PD, an analogue signal goes directly to the AFE module.

2) Communication: Device is controlled by the user interface via USB connection. For better sensor management we have developed a Python based graphical user interface (GUI) that allows to set individually the current of each LED, feedback resistors and capacitors, to view the received signals and save the raw data into the file. From Analogue front-end we receive 6 signals: LED1, LED2, LED1 ambient, LED2 ambient, LED1-LED1 ambient and LED2-LED2 ambient. All signals are 22-bit long. Automatic ambient measurement and cancellation is done by the AFE.

B. Logic

1) Driving phases: The LED array driving process has two main phases. At first, the array has to be calibrated which is mandatory to start the measurement process. Calibration process analyses the acquired signal and determines LED groups that have the best signal quality.

For the calibration we group two LEDs into one group. In Figure 1 LEDs IRn-1 and IRn-2 forms one group, Rn-1 and Gn-1 second, Rn-2 and Gn-2 third, and IRn-3 and
IRn-4 fourth group. Same grouping methodology is defined in each group and on both channels. Altogether we get 16 LED groups. Each group is switched on and off for a short period of time with different pre-defined configurations.

Calibration with each group is started by setting the LED current to 100mA and amplification with the feedback resistor to the maximum level. If the signal strength goes into saturation, amplification is decreased until the AC component has the maximum value and DC component is not in the saturation. Based on the AC and DC component, we calculate the efficiency. At first, a received photo-current is calculated:

\[ I_p = \frac{V_{out}}{2 \cdot R_f}, \]  

where \( I_p \) is photo-current, \( V_{out} \) is photo-voltage analog-digital conversion (ADC) value divided by 22-bits, \( R_f \) is feedback resistor of the amplifier. With that equation we can calculate photo-current for AC and DC component. Efficiency is calculated with the following formula:

\[ \gamma_{eff} = \frac{I_{AC}}{I_{DC}}, \]  

where \( \gamma_{eff} \) is the efficiency, \( I_{AC} \) is photo-current of AC component and \( I_{DC} \) is photo-current of DC component.

After all groups are toggled once with their own best settings, signal quality analysis follows to detect the presence of pulse wave. The group with the highest amplitude of AC component will be chosen automatically to start the continuous measurement process. If there are signals with identical quality from more than one group we can redefine groups and repeat the same process to find only those LEDs that give the best signal for our needs and group these into one group that will be used during the analysis.

2) Configuration of light source driver: The AFE module is capable of generating up to 5 kHz pulse repetition frequency (PRF). Each period includes two times ambient and LED sampling. The sample rate is four times PRF, up to 20 kHz. For pulse wave detection the common sampling rate is 250 Hz and up but using higher sampling rate it is possible to use built in hardware averaging functionality that gives even better signal quality. In our current configuration, we are using sampling rate of 500Hz and no averaging.

Experiments were performed on different days during few hours on both days, but on the same test person. The signals were recorded from the left hand radial artery. Efficiency was calculated as described in the previous section. Table I depicts the relative signal efficiency for each LED and photo-diode pair. The optopair with highest efficiency on each vertical group is colored. Red color marks infrared, orange red and light green marks green LED. Efficiency more than 1% is considered usually as a good signal. The bigger the efficiency number the better signal to noise ratio we get.

The signal with the highest efficiency is received with the LEDs that have the longest wavelength, marked with red. Comparing the left and right side, the signal with highest efficiency is on the right side because radial artery is more close to the surface of the skin on the wrist side. As it can be seen from Table I, there are also some differences between measurements on different days. However, it is visible, that the results are repeatable and the radial artery can be detected under certain optopair.

Fig. 3 depicts the build-up of the sensor module. Module has a connector for external system connection, that is built on the flex part. All control logic is placed on the rigid part as it helps to increase the mechanical reliability because the rigid does not bend. All optics are on the flex part as it touches directly the skin and needs to be bent accordingly. All electronics, including LEDs and photo-diodes, is poured into the medical silicone to minimize the effect of the skin.

Rigid and flex parts have 4-layer design to suppress the noise and increase the stiffness to the appropriate level. Extra care has been taken with the signal line routing of the detectors. As the length of the whole sensor part is 138 mm, there is a risk for increased noise. For that reason all detector lines are routed on the middle layer and also surrounded with shielding traces.

IV. Experiments and Results

During the real experiments we have got results that verify out expectations about obtaining the best pulse wave signal from the radial artery only from the LED and photo-diode pair with the highest efficiency, that is calculated using formula (2). Measurements were performed by placing the sensor on the wrist, as depicted on figure 4, and fastening it using bending strap.

Fig. 4. Sensor placement on the wrist
For the reference we have also measured noise level of photo-diode by shutting down LED driving part of the AFE module and putting the sensor to the dark. The average noise is 0.256 mV and it is not dependent on the feedback resistor in Eq. (1).

Figure 5 depicts the results of one LED pair. Upper part describes the signal measured with IR2-3 and below IR2-4. Both signals have already ambient subtracted and LED current is calculated based on the Eq. (1) and (2). As this figure belongs to the measurements made on the second day, it correlates well with the Table I bottom part. Efficiency values in that table show also that IR2-4 has slightly better efficiency compared to IR2-3, 0.69% and 0.48% accordingly.

V. FUTURE WORK AND CONCLUSIONS

Currently all measurements were performed manually by switching between LED pairs, setting feedback resistor, capacitor and LED current values manually. This task will be automated in the future because this is the calibration task that needs to be performed before each measurement. During that time there should be no sensor placement changes nor any other disturbances that may change the environment conditions, which impacts heavily signal quality.

LED and photo-diode control logic is currently separated from AFE with micro-connector. To increase the physical reliability and noise immunity, AFE and LED/PD driving can be bundled into one board to reduce the physical dimensions of the module twice. This makes the placement of the sensor to the human wrist more comfortable and faster.

It is possible to decrease the ambient noise by increasing ambient cancellation current and decreasing LED duty cycle. It is also possible to enable the second stage ambient cancellation amplification to increase the noise immunity and get more noise free bits.

In this paper, the architecture and driving possibilities of smart PPG sensor is presented. We have designed the first prototype of smart PPG sensor. It has 32 independent emitters and 4 independent detectors that can be grouped and driven individually. With the current configuration the system is flexible enough to perform measurements by grouping emitters into different groups or driving all of them individually. First experiments show that it is possible to use developed sensor for registration of pulse wave from radial artery in more comfortable way and faster. According to our results the efficiency variation ca 1.46% between different optopairs and different emitter wavelengths shows clearly the usability of proposed sensor.

REFERENCES