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A highly configurable platform for advanced PPG analysis^{*}

Abstract. Growing interest in healthcare applications has made biomedical engineering one of the fastest growing disciplines in recent years. Photoplethysmography (PPG) is one of these trends since PPG sensors are embedded into many devices like smartwatches or oximeters. Due to the versatility of PPG, it is the technique of choice for the non-invasive monitoring of vital signs such as heart rate, respiratory rate, blood oxygen saturation and blood pressure. However, commercial pulse oximeters often use proprietary data acquisition and visualization techniques, making digital signal processing on raw data difficult or impossible. This paper presents a hardware platform for the exploration of new biological signal analysis based on PPG. The originality of the platform is to expand the number of wavelengths to 8 and to allow to the parameterization of the acquisition by choosing the sample rate. This then allows the PPG sensor to be adapted to the needs of analysis on the whole signal or perform high precision measurements. Much consideration has been given to the accurate data acquisition during sports activities.

Keywords: Photoplethysmography · PPG · Multi-wavelength · Mixed Signal · Heart Rate · SpO₂.

1 Introduction

Photoplethysmography (PPG) is a simple and low-cost optical technique that can be used to detect blood volume changes in the microvascular bed of tissue. It is often used non-invasively to make measurements at the skin surface (Figure 1 [6]). As an optical technique, PPG requires a light source (LED) and a photodetector to operate. The light passes through the tissue and the photodetector detects small changes in light intensity.

When the heart pumps blood, each contraction of the heart causes an increase in the amount of blood in the capillaries in the skin's surface, resulting in more light absorption. The blood then travels back to the heart through the venous network, leading to a decrease of blood volume in the surface capillaries, resulting in less light absorption. As shown on the Figure 1, the signal can be decomposed in two parts, the steady component and pulsating component. Each part contains useful information and is used to compute physiological features. To calculate certain biomarkers, it is necessary to have a precise value for the constant component (DC) and the pulsed component (AC). A particular

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attention should be given to the acquisition of the pulsed component because its amplitude represents only about 5% of the constant component.

PPG signals can be acquired by either transmittance and reflectance. In transmittance mode, the photons leave the LED, pass through the finger and are measured by the photodiode. In reflection mode, the skin and tissues reflect the photons, which are captured by the photodiode placed near the LED. Transmittance mode produces good signals, but can only be used on body sites of limited thickness, such as the fingertips, earlobes or nasal septum, to detect light. The reflection mode does not have this problem and allows measurements to be taken on other body sites such as the wrists, chest, or forehead.

One of the fundamental principles of PPG relies on the higher sensitivity of certain optical wavelengths for blood rather than other tissue components. The Red LED (~ 660 nm) and Infrared (IR) LED (~ 940 nm) are the standard wavelengths used in pulse oximeters for the calculation of oxygen saturation. The green LED wavelength (~ 550 nm) is being widely used in reflection-mode applications, since tissue reflects well at this wavelength. As the evaluation of quantity and concentration using PPG is based on Beer-Lambert law, the more signals with different wavelengths measured, the more accurate and robust the evaluation.

This paper considers the use of PPG in two domains: medical and sports. The medical domain focuses on detecting, tracking and curing diseases, while the sports domain targets training stress tracking and performance enhancement. However, despite the different goals of the two domains, both require measurements of the same physiological features and the same dedication to precision and robustness.

To answer the challenges posed by PPG acquisition, we propose a new hardware sensor that is highly configurable. Section 2 describes the background and the necessity of having a configurable hardware for PPG. The control given to the user allows them to adapt the sensor to the experiment or even to the test subject, to ensure the best result possible. Section 3 presents the platform design and the hardware choices that were made, to achieve the necessary configurability and precision. Section 4 presents the experimental results obtain using this platform. Finally, Section 5 discusses the findings and presents the conclusion.

2 Background

This section presents the background of the state of the art of PPG acquisition, utilization, and limitations, highlighting the benefits of a more configurable platform with better access to raw output data. First, PPG utilization is presented to understand which features are important and how the quality of the output data influences the result of the biomarker calculation. Next, related work on existing platforms are detailed, highlighting the gaps in current commercial offers.

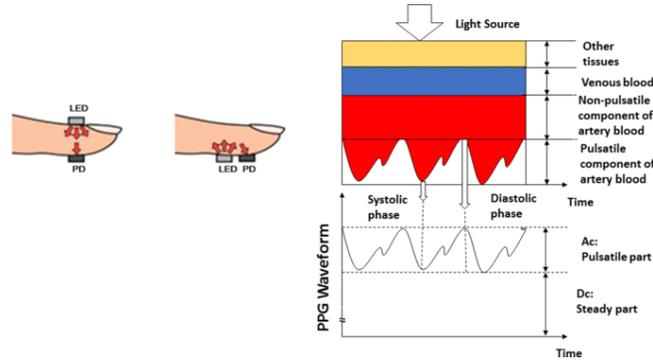


Fig. 1: Principle of the PPG (from [6]).

2.1 PPG analysis

Heart Rate is the frequency of the beating of the heart. It is a good indication of the physical condition and provides valuable information, in particular in a sporting context. Heart rate monitoring for fitness is valuable for athletes to monitor their workload. The blood volume change detected by the PPG is correlated to the heart rate. The pulse rate is effectively equivalent to the heart rate and can be calculated using PPG [2, 3, 8].

In order to do so, different techniques can be used. Most simply, the peaks and troughs of the signal are detected and used to deduce the pulse rate [8]. A technique using the frequency domain can also be used [7]. The accuracy of the pulsating part (sampling rate and quantization) of the signal has an overarching importance for this feature.

Heart rate variability quantifies the variation in time between each heartbeat. This parameter is tracked primarily in sports applications as it changes during and after exercise. The Heart Rate Variation (HRV) feature can be used to analyse the stress the body experiences during training and to gain insight into physiological recovery after training. Information regarding the extent to which the body recovers after training provides useful data for the personalization of sports training, training loads, and recovery time. HRV monitoring of athletes is frequently applied to the prevention and diagnosis of overtraining syndrome [4].

As well as HR, HRV computation accuracy is tied to the pulsating component of the captured signal. High sample rate and high quantization accuracy of the signal are necessary for this metric.

Oxygenation and SpO2 The blood oxygenation can be deduced from the PPG signal, as the oxygenated hemoglobin has a different absorption to the deoxygenated hemoglobin [9] when they are exposed to multiple wavelengths, as illustrated on Fig. 2.

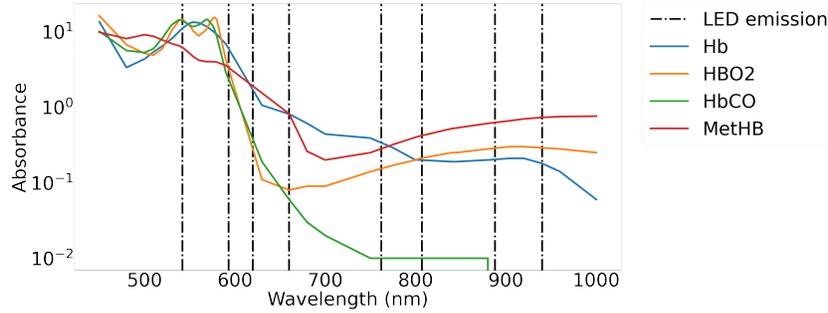


Fig. 2: Absorbance of different haemoglobins and wavelength of the LEDs chosen for the platform.

Carboxyhemoglobin and Methemoglobin are often neglected, as their concentration in the blood under normal circumstance is very low to non-existent. The use of absorption of light for calculation of SpO₂ levels makes use of the Beer-Lambert Law. Using this law with additional wavelengths enable the detection of those types of hemoglobin.

Breathing The effect of breathing can be considered as noise or as another signal, depending on the PPG analysis to be performed. Indeed, respiration and synaptic nerve activity influence the final signal, making some analyses difficult to perform. However, at the same time, it is an interesting biomarker that can be extracted. Physiological effects related to respiratory activity and the cardiac cycle induce several modulations of the PPG. For this metric, accurate quantization of the signal and its sampling rate are a key of accurate computation.

2.2 Related work

This section surveys some available devices used to extract and study PPG signals. Several technologies are categorized depending on the intended use. Whilst most of these devices allow the visualization, storage, and extraction of parameters from the PPG signal, they offer limited possibilities for additional data extraction and novel signal processing algorithms.

Platforms that allow experimental analysis should be able to make comparisons of different algorithms. Commercial solutions generally do not disclose the details of their hardware and software. From our perspective, existing PPG devices can be categorized in packaged solutions and evaluation systems.

Packaged PPG devices Nowadays, activity trackers and smartwatches embed PPG sensors to perform HR analysis and SpO₂ measurements. The raw data measured by these devices are generally not accessible by the user; the user only has access to the compiled data (mean, max, or min) or even proprietary

analysis including sleep analysis or stress level. These devices are dedicated to consumer applications and do not allow for advanced analysis required for research projects. Despite the acceptable accuracy of the displayed metrics (such as HR, HRV, SpO₂, RR, etc...), further analysis is impossible due to inaccessibility of the raw data [5].

PPG Evaluation/Development Kits Some chip manufacturers and companies propose kits for developing projects and applications using physiological sensors.

As an example of dedicated company, BITalino proposes a handy plug and play system that allows many physiological measurements [1]. While these kinds of devices are useful to develop new projects and applications in a limited time, the performance of sensors are limited to commercial sensor performance, i.e. 500Hz sample rate with 2 or 3 wavelengths. Moreover, they do not allow tuning of all the different building blocks needed for PPG signal acquisition and analysis.

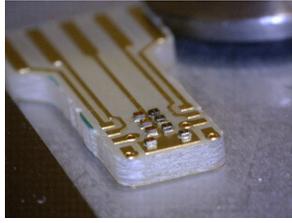
The MAX30101 Evaluation kit (EV kit) from Maxim Integrated allows the evaluation of the MAX30101, which is an integrated pulse oximetry and heart-rate monitor integrated circuit. The MAX30101 includes internal LEDs (green, red and IR), a photodetector together with low-noise electronics including ambient light rejection and digital filtering. The MAX30101 EV kit monitors and stores the recorded PPG signals while allowing the user to modify the sampling rate, LED currents, and pulse widths. This EV kit board allows the user to fully exploit the MAX30101, which is a 3-wavelengths sensor.

Similarly, the AFE44x0SpO₂ EVM from Texas Instruments is another kit intended for evaluating AFE4400 and AFE4490 devices. The AFE4490 is a complete analog front-end solution targeted for pulse-oximeter applications. The device consists of a low-noise receiver channel, a LED transmit section, and diagnostics for sensor and LED fault detection. The software of the AFE44x0SpO₂ EVM includes a GUI that allows the configuration of the I-V amplifier, the ambient light cancellation DAC, the analog filtering, the ADC sampling and LED pulse & intensity, among many other parameters [29]. This is a very complete evaluation system with highly configurable capabilities that unfortunately only allows two wavelengths.

To the authors' knowledge, there are no platforms that allow the exploration of multiple signals, higher frequency and higher resolution with respect to the quality of the PPG acquisition.

3 Platform design

The proposed platform is designed to tackle limits of current devices for PPG analysis. It extends current platforms by proposing multiple wavelengths (more than 3), higher frequency, higher resolution while maintaining a 'nomadic' device. The device should capture and store raw data, as well as making embedded signal processing possible with a general purpose processor.



(a) Transmittance pattern before encapsulation.



(b) Reflectance pattern encapsulated in a protective resin.

Fig. 3: The two first sensors manufactured for this study.

The platform is based on a PYNQ-Z1 board, incorporating a Xilinx Zynq SoC. The integrated ARM processor, equipped with an operating system, serves as an ideal sandbox for signal processing algorithm research. Leveraging the FPGA within the Zynq platform facilitates the migration of the acquisition process to dedicated hardware. This ensures the reliability of the acquisition while avoiding any adverse impact on the ARM processor. A custom electronic board complement the PYNQ-Z1 to control the physical sensor.

The platform uses an original physical sensor presented in Section 3.1, custom electronic design for LED power and signal acquisition explained in Section 3.2 and finally an integrated logic design as coordinator detailed in Section 3.3.

3.1 Physical sensor design

One of the primary goals of the advanced PPG sensor is to provide measurements with more wavelengths than the typical green/red/IR. Consequently, a new sensor has been designed with 8 LEDs allowing 8 wavelengths to be studied. It is based on non-encapsulated LEDs ($300 \times 300 \mu\text{m}$), with the objective to minimize the sensor size. The sensor is covered with a protective resin.

With the constraints of size, market availability and price, the following 8 LEDs wavelengths have been chosen. A red LED operating at 660 nm and an IR LED operating at 940 nm allow the comparison with standard oximeters. LEDs operating at 542 nm, 593 nm, 620 nm, 762 nm, 807 nm, 888 nm are selected to estimate Carboxyhemoglobin and Methemoglobin.

Three sensor patterns consisting of 8 LEDs covering wavelengths from 542 to 940 nm were manufactured. The first sensor pattern on Figure 3a was designed for use in transmittance, so contained 8 LEDs, with the photodiode on another PCB. The second sensor pattern on Figure 3b was for use in reflectance, so the 8 LEDs surrounded a photodiode. The third sensor pattern is also used for reflectance, but the distance between the LEDs and the photodiode is greater, in order to be able to detect light that has penetrated deeper into the tissue. The reflectance mode allows the platform to be used in a wider variety of body sites compared to the transmittance mode.

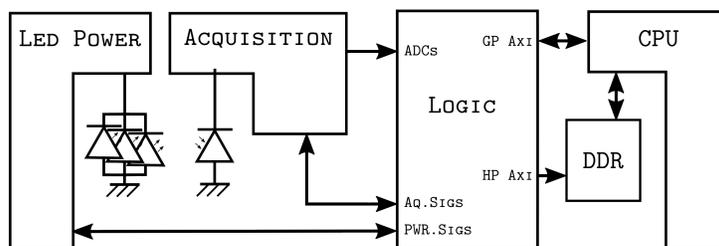


Fig. 4: Global schematic of the system.

3.2 Electronic design

To enable sample measurements using these sensors, two specific electronic circuits have been designed. The primary goal of the first board is to provide stable and accurate currents to the LEDs. The second board performs fine measurements of the received current in the photodiode. These two circuits are connected to the LEDs, photodiode and logic as shown in the Figure 4.

LED Power The first electronic circuit drives the current in a specific LED using 2 signals. The first signal is the *enable* signal. This signal is composed of one wire per LED and is set to a logic 1 (here 3.3V) which causes the selected LED to light up. The second signal is the *power* signal. This signal is a Pulse Width Modulation (PWM) signal that control the amount of current to drive into the LED. The Figure 5 represents the circuit design for a specific LED.

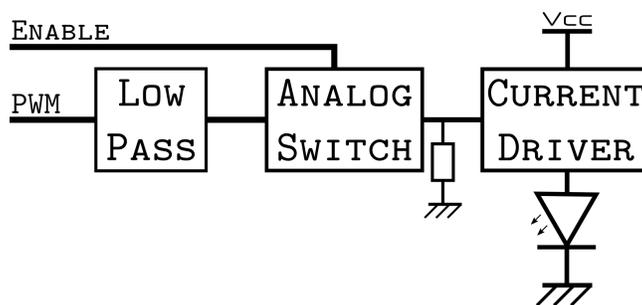


Fig. 5: Electronic LED power stage for one LED.

In this circuit, the *power* signal passes inside a low pass filter (an RC design) that transform the PWM into an analog signal. The analog switch injects this signal in the current driver based on the level of the enable signal. The analog switch chosen is a HCT4066-based circuit. A pull down resistor set the output of

the switch to ground when the switch is disconnected. Finally, the current driver section operate the current regulation. The schematic is based on an Operational Amplifier (OpAmp) and a transistor (TLC271CP and BC847CW). This circuit is duplicated 8 times in the complete circuit to drive the 8 LEDs independently.

Acquisition The light emitted by the LED is then attenuated or reflected by the tissue of the subject and captured by the photodiode. Figure 6 shows the schematic of the capture circuit.

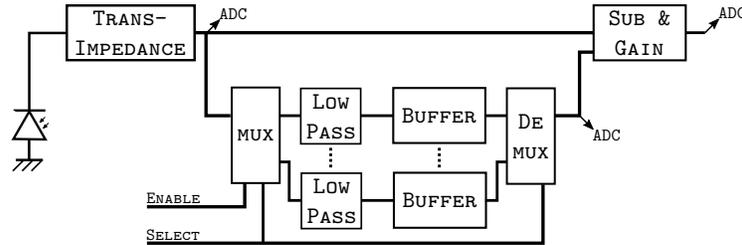


Fig. 6: Photodiode stage for data acquisition.

This schematic has 2 types of input signals and 3 ADC outputs. The ADC engines of the targeted architecture can select one of these 3 signals and selectively measure the three inputs. The 2 inputs are a 3-bit *select* that specify which LED is currently measured and a 1-bit *enable* signal that specify if the device is in calibration mode.

The first block is a transimpedance block composed of AD8021AR OpAmp that converts the current delivered by the photodiode to a voltage. This first voltage can be measured (after a level adjuster) to one of the ADC pins of the Zynq chip. This voltage can be used to evaluate the ambient noise during a capture.

To maximize the signal precision, the signal is amplified. However, since the pulsating level of the signal is significantly lower than the steady component, the choice has been made to subtract a fixed offset value from the signal. This fixed offset value is determined at the start of the capture by taking the average value of the signal for each LED.

This operation is performed through the first analog Mux (74HCT4051) that allows the input signal to charge a low pass RC filter when the *calibration* signal is set. The signal is then buffered using an TLC271 OpAmp used as voltage follower, thus considerably increasing the discharge time of the filter capacitor. After calibration, DC-offset can be measured using the second analog Mux that outputs to an ADC.

Finally, the full signal and the offset are subtracted then amplified using an TLC271 OpAmp in differential amplification mode. Resistors are chosen accord-

ingly to obtain the following output voltage:

$$V_{out} = g * (V_{avg} - V_{in}) + 0.5 \quad (1)$$

Choosing an offset of 0.5V allows maximum signal variation, as the ADC has a range of 0-1 V.

3.3 Logic

The analog/digital interface is controlled by the FPGA, to operate the sensor's electronic and the ADC. The signal capture is not processor dependent and the FPGA guarantees synchronicity of the capture operation as well as less processing on the processor. The logic system is described in Figure 7.

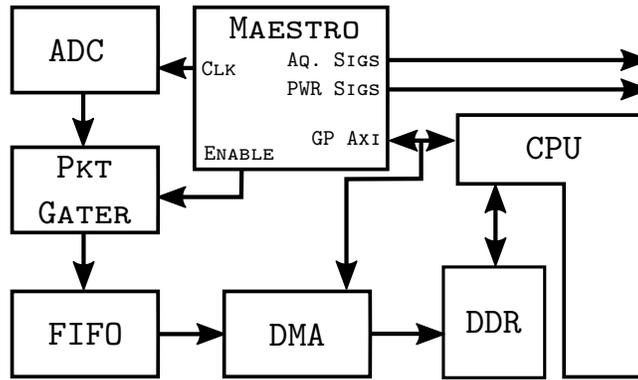


Fig. 7: Schematic of the FPGA design.

A global state machine called Maestro handles the capture process. It outputs the signals for the LED and photodiode electronics and pilots the capture of the ADC. The LEDs are switched on and off sequentially, with a negligible delay between them, so that the signals from the 8 wavelengths can be measured independently.

The ADC has a 12-bit resolution, but it outputs a 16-bit result. The least significant 4 bits in the result are not meaningful and can be used to reduce the quantization noise by averaging. The 16 bits result is transmitted to the user.

Data management The processor communicates with the Maestro through a general purpose AXI interface, by writing in specific registers. Samples are retrieved directly through the DDR.

When performing the capture, the ADC outputs its captured data through an AXI Stream interface. This stream is then gated to allow the sample to flow only when a capture is performed. This is necessary to both prevent the ADC

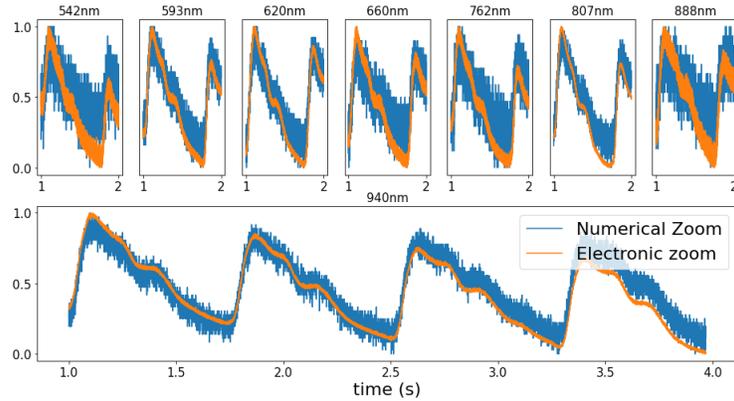


Fig. 8: Comparison of precision between complete and AC normalized signals.

to output a sample at startup when it is not configured correctly, but also for debugging (performing dry runs).

When the capture starts, samples are buffered in a FIFO before being copied to the DDR using a DMA. When triggered by the processor, the DMA writes packets of samples to a given address in the DDR. High performance AXI ports of the Zynq architecture are used to achieve a high throughput.

Highly configurable The use of an FPGA also provides a wide range of sampling frequencies. The platform can currently support sampling frequencies ranging from 200 Hz to 10 kHz, each sample comprising a sample for the 8 LEDs. In addition, various acquisition parameters can be configured.

As mentioned before, the power in the LEDs can be specified separately for each LED. A software on the processor perform an exhaustive search for the best power value in the LED for each LED, in order to optimize dynamic range during acquisition. This can be used to adapt the platform to the experiment environment but also to the experiment subject.

The delay between turning on the LED and the measurement can be changed. There is a compromise to find between the settling time and fastness. Similarly, it is possible to configure the delay between turning off the previous LED and turning on the next one. The numbers of LEDs used for an experiment can be chosen. Any combination that use between 1 and 8 LEDs are allowed. The number of packets used for the calibration of the offset and the number of samples per packet can also be configured.

4 Evaluation

Several experiments were performed in order to evaluate the proposed platform. This section presents 2 of them with an evaluation of the acquired signal quality and a first biomarker extraction.

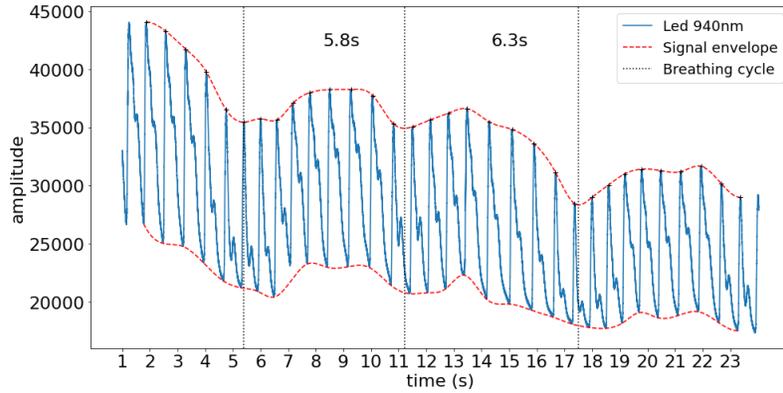


Fig. 9: Breathing cycle measurement using PPG.

The first experiment illustrates the added value of the AC extraction, i.e. the removal of the signal offset before acquisition. The advanced photodetector was used to capture a signal for 4 seconds with a sampling rate of 2000 Hz. The complete signal and the AC signal, both unfiltered, were normalized between 0 and 1 to compare them. The Figure 8 represents the result for the signals, with of focus on the 940 nm LED. The AC (removing the DC before acquisition) is less noisy than the full signal (removing the DC after the acquisition). The average variance on windows of 40 samples is 0.066 versus 0.476 .

The sum of the variances of the signals on windows with a size of 40 samples were calculated. The results are given in the table 1 where V_n is the variance of the digital zoom and V_e is the variance of the electronic zoom and λ the wavelength in nm.

Table 1: Comparison of variance between complete and AC normalized signals.

λ	542	593	620	660	762	807	888	940
V_n	0.73	0.31	0.29	0.46	0.63	0.21	0.71	0.47
V_e	0.11	0.04	0.04	0.07	0.06	0.03	0.13	0.05

For the second experiment, the subject was asked to repeat breathing cycles consisting of 3 seconds of inspiration followed by 3 seconds of expiration. We can observe the acquired AC signal on Figure 9. The envelope of the signal has a periodicity of about 6 seconds, which corresponds to the experimental conditions. The breathing is thus easily extracted from the signal obtained with the sensor. The same acquisition without offset removal would be much more difficult to interpret.

5 Conclusion

In conclusion, this paper has centered on the development of an extensively configurable PPG sensor with accessible raw output data. This feature enhances flexibility in scientific investigations within the PPG domain. The direct access to raw data facilitates seamless comparison of research outcomes among scholars. Notably, our sensor empowers researchers to personalize critical parameters, such as the sample rate and the utilisation of multiple LEDs (up to 8). This flexibility opens avenues for exploring novel methods of calculating biomarkers.

Moreover, the adaptability of this sensor extends the possibilities for devising energy-efficient hardware strategies. By adjusting parameters based on the specific requirements and reducing energy consumption during periods of low information, novel approaches to energy-aware hardware design can be explored. Furthermore, the portability of this platform, coupled with its compatibility with a diverse array of body sites, makes it a versatile tool for both educational and research purposes.

Future work will explore the use of this new sensor and platform in a clinical trial, associated with a battery of other sensors. This trial will help validate the performance of the proposed platform. It will also enable new research into biomarker extraction from the acquired PPG signals.

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