A Wearable Wrist-Band with Compressive Sensing based Ultra-Low Power Photoplethysmography Readout Circuit

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A Wearable Wrist-Band with Compressive Sensing based Ultra-Low Power Photoplethysmography Readout Circuit

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***Abstract*—In this paper, we present our efforts towards packaging of a novel compressive sampling (CS) based ultra-low power photoplethysmography (PPG) application-specific integrated circuit (ASIC) into a wearable form factor. The system comprises of a custom PPG analog front-end circuit, integrated with a digital back-end to enable CS, and a commercial off-the-shelf (COTS) system-on-chip (SoC) for Bluetooth Low Energy (BLE) based wireless data transfer. The ASIC consumes 172 µW power to extract heart rate from the sparse PPG signal where the whole system consumes 1.66 mW power for continuous streaming of heart rate data over the COTS BLE radio. This work presents the first ever demonstration of a CS based PPG ASIC in wrist-band form factors and paves our way towards deploying and evaluating this custom PPG ASIC in future clinical studies. The modular architecture of the wristband platform allows for incorporation of other sensors for future correlated sensing studies between health and environment.**

Keywords*—*photoplethysmography, compressive sampling, heart rate, Bluetooth Low Energy, wearables.

I. Introduction

Wearable physiological monitoring systems have become very popular during the recent years especially after their integration into various consumer products. In these wearable form-factors, continuous heart rate (HR) monitoring and assessing heart rate variability provide critical implications about the user's cardiovascular health status. Among various methods to assess HR, photoplethysmography (PPG) is a simple and compact technique that employs on-skin optical sensing to detect the change of arterial blood volume in the tissue noninvasively.

The recent commercially available fitness tracking and smart watch devices with integrated PPG sensors have targeted user comfort to be one of the most important factors in their design consideration. Despite the fact that clinical PPG systems are generally deployed on the finger, these consumer products have picked wrist in order to reduce movement limitation and deploy larger batteries for longer battery lifetime. Some similar wrist-watch based physiological sensors found in the literature [1], [2] use commercial off-the-shelf (COTS) integrated circuits (ICs), such as AFE4400 from Texas Instruments (TI), for processing raw PPG signal which consume substantial power. These systems have also provided on-board wireless transmission capability to decrease wiring complexity. Power optimization and miniaturization of sensing devices are two of the on-going challenges for system designers to ensure even more comfortable wearability.

Traditionally, any PPG system for monitoring HR comprises of a light source (mostly light emitting diodes, LED), a light-sensing detector (mostly photodiodes, PD), an analog front-end (AFE) circuit to amplify/filter the signal and a processing unit. Among these, the major portion of the power is consumed by the LED itself, even if the duty cycle and frequency of the LED-controlling pulse are pushed to the limits which is usually confined by the Nyquist rate and the bandwidth of the AFE. With an aim to reduce the power consumption further, we had explored compressive sampling (CS) technique before and presented a PPG application specific integrated circuit (ASIC) with a power consumption of 172 µW [3], [4]. This ASIC is capable of extracting the HR data from the compressed PPG signal with an effective sampling rate of only 4 Hz.

Table I displays the results of some attempts to reduce power consumption using other techniques like logarithmic amplifier [6], heart-beat locked loop [8] etc. Among these, compressive sampling is one of the lowest power consuming techniques. However, all these reported specifications are based on benchtop evaluations mostly on the finger and it would be unfair to benchmark these ASICs without clinical validation. The aim of the work presented here is to package our CS PPG ASIC into a miniaturized wearable form-factor in order to evaluate its usability to track HR on the wrist. As the

TABLE I. Comparison of key performance criteria with the State-of-the-Art ASICs

Literature & citation	Our work [4]	TBCAS'08 [5]	TBCAS'10 [6]	TBCAS'15 [7]	ISSCC'18 [8]
Tech (µm)	0.18	0.35	1.50	0.18	0.18
Supply (V)	1.2	2.5	5.0	1.8	3.3
f_s (Hz)	128-4	100	100	165	100,40
LED ctrl.	√	×	×	√	√
Feat. Extr.	HR,HRV	-	SpO_2	-	HR
AFE (µW)	172	600	400	216	27,29
LED (µW)	1200-43	-	4400	1125-120	520-9

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Dr. Pamula is now with University of Washington and Dr. Valero-Sarmiento is with Intel Corporation.

next stage, these wearable units will be tested during the ongoing clinical studies under the United States National Science Foundation Engineering Research Center for Advanced Self-powered Systems of Integrated Sensors and Technologies (ASSIST) [9]. As these clinical studies will also include the evaluation of other low power ASSIST sensors for correlated sensing of environment and health, we have designed this wearable wrist platform to incorporate a modular and flexible architecture. The following sections describe the hardware and software interfacing techniques we followed for such modularity and present the preliminary evaluation results of our CS PPG ASIC in this package. To the best of our knowledge, this is the first time demonstration of a CS based wrist-worn PPG system in the literature.

II. System Design and Implementation

A. Interfacing the PPG ASIC

The block diagram in (Fig. 1) shows the internal structure and input/output connections of the ASIC. The ASIC is controlled by 16 digital I/O pins for adjusting the gain and bandwidth of the amplifier and filter stages of the AFE, the PD bias cancellation current (I_{DAC}), and the compression ratio (CR) of sampling etc. Moreover, the external LED driver (Fig. 2) has a 4-bit analog switch (TS3A4751, TI, Dallas, TX, USA) which activates the LED and controls the brightness by changing the current (I_{LED}). This sums up to 20 input signals required from an external microcontroller.

The PPG ASIC also has an 8-bit output for the extracted HR data. In order to access all these I/O pins from the microcontroller, we have used three I^2C-based I/O expanders (PCF8574 & PCF8575, TI, Dallas, TX, USA). Unlike the rest of the system which uses 3.3 V power supply, the ASIC was designed to operate at 1.2 V supply voltage (Fig. 3). To exchange the voltage levels, we have used multiple 8-bit voltage translators (NLSV8T244, ON Semiconductor, Denver, CO, USA). The voltage regulation from 3.3 V to 1.2 V is performed by a switching regulator or Buck converter (TPS62122, TI, Dallas, TX, USA) to enhance the power efficiency. However, to keep the supply to the internal analog to digital converter (ADC) in the ASIC less noisy, a separate low dropout (LDO) linear voltage regulator (LT3009, Linear Technology, Milpitas, CA, USA) has also been used.

The $4.0\times2.5\,\text{mm}^2$ ASIC has a total of 84 pins. The use of a standard PLCC package could consume $35.9\times35.9\,\text{mm}^2$ area and would be challenging for accommodating into a wrist-worn form factor. With an aim of miniaturization, we have designed a $12\times12\,\text{mm}^2$ printed circuit board (Fig. 5) to access the necessary pins instead of using the PLCC package and breakout only 40 accessible connections, which is sufficient for system operation. The bare die of the ASIC is attached to the ground plane of the breakout board with silver conductive epoxy (8331, MG Chemicals, Surrey, B.C., Canada) and heat-cured at 65°C for 20 minutes. The chip is then bonded with an ultrasonic/thermosonic wedge-wedge wire bonder using aluminum wire of 1 µm diameter.

B. The Modular PCB Approach

For modularity, all the aforementioned components along with the breakout board are soldered into a circular printed circuit board (PCB) of 38.1 mm diameter. This daughter-board is then plugged onto another PCB (mother-board) of same size (Fig. 5) that incorporates the system on chip (SoC). This allows for the use of the same motherboard with other daughter-boards housing other sensor systems, such as ultra-low power ozone sensor or volatile-organic-compound sensor to assess air quality [9]. As mentioned earlier, this modular architecture is intended for future validation of these sensors in the clinics and to correlate the air quality versus health outcomes.

The motherboard contains a Bluetooth Low Energy (BLE) radio transceiver (Bluegiga BLE113, Silicon Labs, Austin, TX, USA) controlled by an SoC (CC2541, TI, Dallas, TX, USA). It is powered by a 3.7 V rechargeable Li-ion Polymer battery which is regulated to 3.3 V. This board also houses

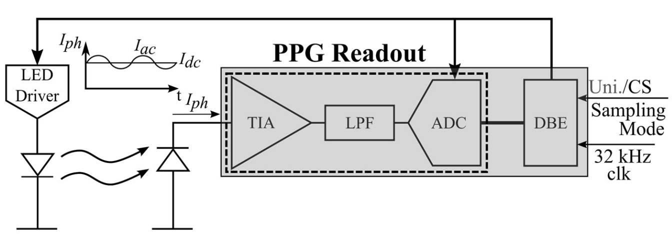

Fig. 1. Block diagram of the PPG ASIC showing connections to external components for I/O exchange [10]

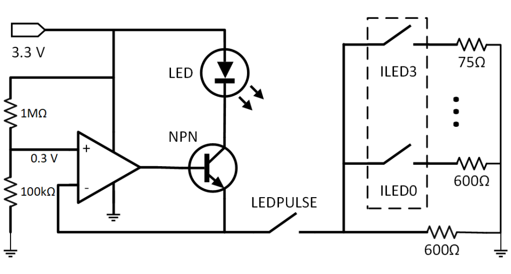

Fig. 2. Schematic of the LED driver circuit which is designed to drive upto 8 mA current with a resolution of 0.5 mA.

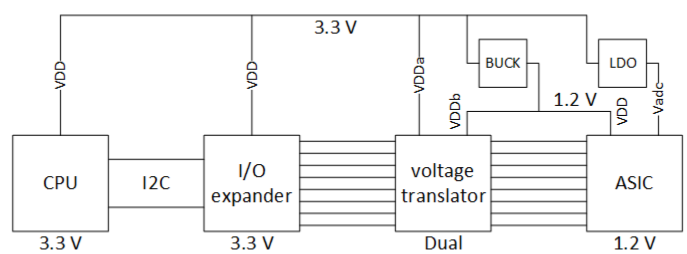

Fig. 3. Schematic of I/O conditioning circuitry along with supply voltage regulation topology.

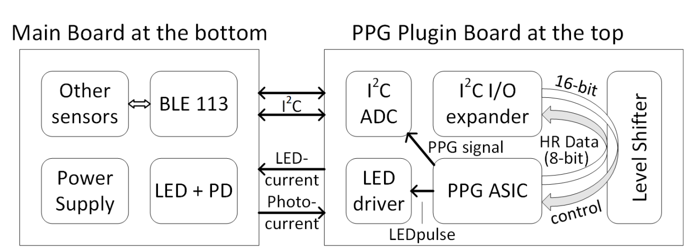

Fig. 4. Diagram of the whole system showing inter-connectivity among the major blocks of the system.

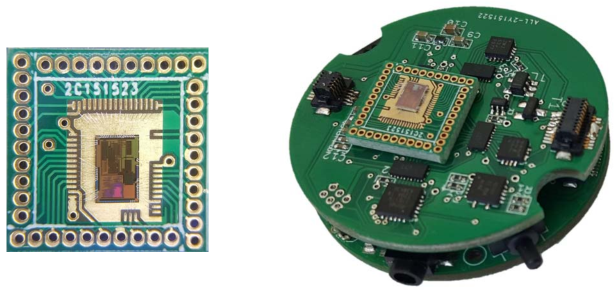

Fig. 5. Picture of the three printed circuit boards assembled together with the breakout board carrying the ASIC (also enlarged on the left), and the plugin (daughter) board on top of the mother-board.

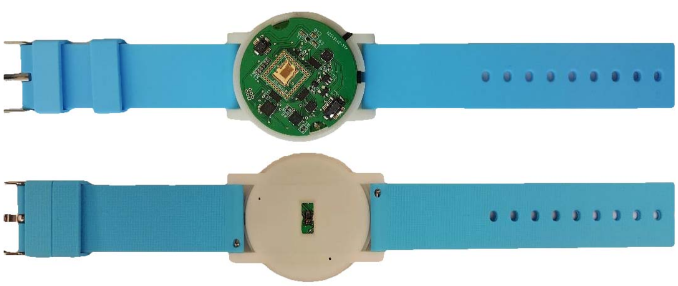

Fig. 6. Picture of front and backside of the wrist watch showing the placement of the BioMon Sensor in the bottom.

other COTS integrated circuits including memory, ADC etc. to accommodate the broader range of requirements for various plugin sensor boards.

All components required for the PPG ASIC are placed in the plugin board (Fig. 4) except the LED and PD which is placed underneath the motherboard to access the tissue through a rectangular window cut from the 3D printed casing (Fig. 6). The current system uses a multi-LED and single-PD package (SFH7050 BioMon Sensor, OSRAM Opto Semiconductors, Regensburg, Germany) in order to reduce area and allow possible expansion to pulse oximetry in the future.

III. Results and Discussions

To test the system, we ensured a tight attachment between the tissue and the Biomon sensor. The single-ended design of the amplifier makes the output bias level sensitive to the applied pressure or motion artifacts. A closed-loop control in the embedded software protects the PPG signal from going into saturation by sampling it with an external ADC IC (ADS7142, TI, Dallas, TX, USA) and adjusting I_{LED} and I_{DAC}. However, as the microcontroller needs to wake up every time to process the sampling, bias adjustment is more power consuming and can be repeated only at a certain interval.

Once the bias level is stabilized, the controller shifts into the low power mode by enabling CS and disabling ADC sampling. During CS operation mode, the ASIC utilizes Lomb-Scargle periodogram to estimate power spectral density (PSD) of the non-uniformly sampled PPG signal and extracts the heart rate once in every 4 seconds [4]. Higher CR makes the waveform difficult to be visually recognized; although the ASIC still can produce HR data up to 30x compression. The extracted HR is read via I^2C, accumulated and transferred after each 40 seconds via BLE.

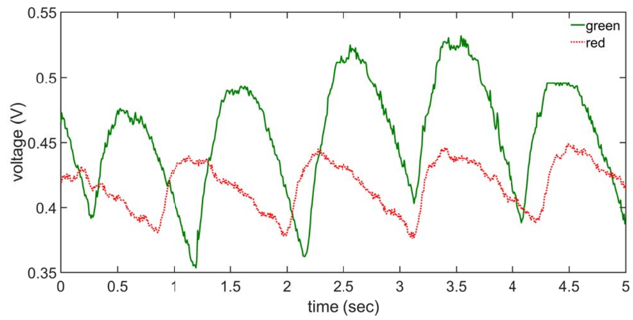

Fig. 7. Comparison of signal quality of two uniformly sampled PPG signals collected at different wavelengths.

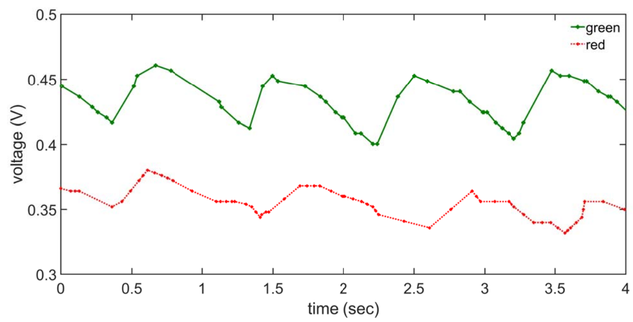

Fig. 8. Two PPG signals collected at different wavelengths with an 8x compression ratio to illustrate the randomly selected sampling pulses.

In our earlier work, we used a red (660 nm) LED for evaluation purposes following general trend of routine clinical applications [4]. However, the green light is absorbed more in total by both oxygenated and deoxygenated hemoglobin, thereby demonstrating the least influence from motion artifacts when used in reflection mode PPG [11]. Hence, we have also evaluated the quality of the PPG signal using a green (525 nm) LED. Both the examples of PPG signals, collected with and without CS enabled, shows that, the amplitude of the PPG wave recorded with green light is higher as illustrated in both Fig. 7 and Fig. 8. The green wavelength is also less prone to bias sensitivity, due to its inherent immunity from motion artifacts.

For validation purpose, the PPG signal (recorded with red LED) is compared with the simultaneous ECG signal of the subject obtained by placing electrodes on the chest. The ECG signals were amplified using an AC amplifier (Extracellular Differential AC Amplifier Model 3500, A-M Systems, Sequim, WA, USA). Although it is visually apparent from Fig. 9, frequency domain analysis of both signals (Fig. 10) double-confirms the matching of the HR.

The power consumption of the system is measured by a source meter (Model 2450, Keithley Instruments, Cleveland, OH, USA). For demonstration purpose, the bias adjustment part was repeated for all the four CR modes and the average power is shown in Fig. 11. After the Bluetooth connection is established, the system switches into the lowest power mode by transmitting 10 bytes after every 40 seconds and consumes 1.66 mW power (estimated lifetime of one week with a 290 mAh battery). The distribution of total power consumed by all the components in the plugin board and the BLE113 is also estimated. Although the SoC is configured

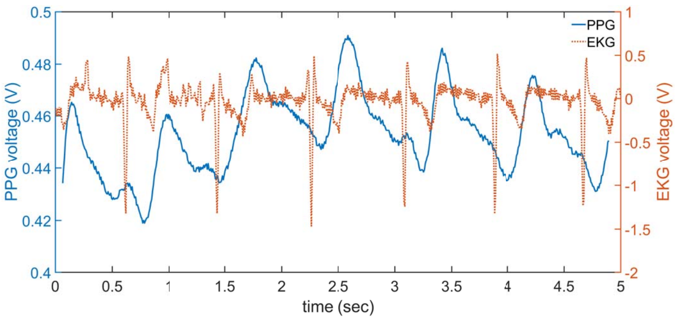

Fig. 9. Benchmarking the PPG signal with the ECG signal (with 10,000x amplification). The period of heart pulses matches quite well with the R-R interval.

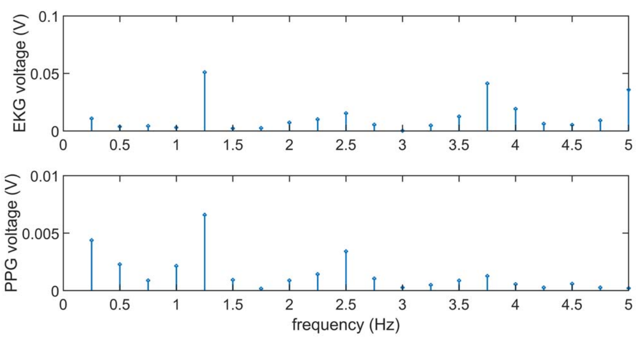

Fig. 10. FFT of EKG and PPG signal showing a HR of 1.25 Hz (75 bpm). The PPG signal shows significant low-frequency components which is mainly due to the slope in bias voltage.

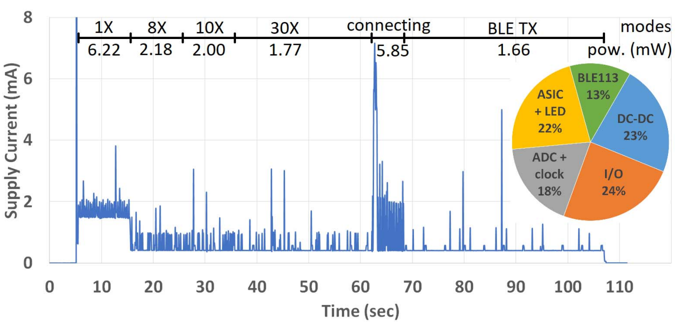

Fig. 11. Transient plot of supply current at 3.7 V input and average power consumption at different modes of operation. Distribution of power consumption among the different components on the boards during the transmission phase at CR = 30x is also included.

to be in sleep mode as long as possible to keep the power consumption minimum, the I/O conditioning circuitry uses up to one-fourth of the total power. This can be avoided in the future by including a communication protocol like SPI or I^2C within the ASIC to access programmable registers. Although there is an ADC in the ASIC, the converted values are not accessible currently, requiring the external ADC IC. Inclusion of such gateway with the ASIC would result in much simpler embedded system with less area and power consumption.

IV. Conclusion

In this paper, we have demonstrated the first ever integration of a novel compressive sampling based ultra-low power PPG ASIC [4] into a wearable wrist-band platform aimed for future clinical evaluation. The ASIC provides an analog front end circuit for PPG signal acquisition and heart rate extraction. The system miniaturization effort towards a wearable form-factor is achieved along with no compromise in the performance of the ASIC. The consumed power was 1.66 mW and can be further reduced in the next iteration of the ASIC design by replacing the COTS SoC with a more application specific SoC and optimizing the ASIC for the targeted application. Overall, we were able to demonstrate a potential wrist-worn system as an efficient platform for future evaluation of the CS based PPG technique through the *in vivo* clinical studies at ASSIST Center.

Acknowledgment

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