A 1mW Vitals Monitoring System for Asthmatic Patients based on Photoplethysmography

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***Abstract*— The number of patients suffering from asthma, especially in air-polluted regions, has dramatically risen. Thus, the need for a wearable system that is able to monitor the patient's condition is essential. Continuous heart rate (HR) and blood oxygen concentration (SpO_2) measurements are crucial for asthmatic patients. This paper presents a low power, portable/wearable system to measure the HR, heart rate variability (HRV) and SpO_2 based on Photoplethysmography. The analog front end (AFE) consists of a switched capacitor trans-impedance amplifier and switched capacitor low pass filter to realize a high gain (120dB) and low cut off frequency (10Hz) on-chip. The proposed AFE enables a photodiode input DC current up to 20µA with input-referred current noise of 7.3pA/√Hz. The system is implemented in 180nm CMOS with a die area of 2.6mm^2 while consuming 1mW/2.3µW from a 0.5V supply for LED drivers and AFE, respectively. The HR/HRV/SpO_2 extraction processor is implemented on FPGA. The maximum absolute error percentage in HR/SpO_2 measurements from an experiment involving 21 subjects comes out to be <1%.**

Keywords— photoplethysmography (PPG); heart rate; SpO_2; wearable sensor; Asthma

I. INTRODUCTION

Asthma is a chronic infection of the respiratory system that inflames and narrows the airways [1], thus making it difficult to breathe. More than 300 million people are affected by asthma worldwide and it will increase to 400 million by 2025 as announced by the World Health Organization (WHO) [2]. During the last decades, the large growth in the statistics of asthmatic children [2] has levitated asthma as the most prevalent chronic disease of childhood in the United States [2]. Asthma is described by a sequence of recurring symptoms that comprise bronchial hyper-responsiveness, intermittent contraction of the bronchioles and thicker mucus, which make it hard for the patient to expire the air from the lungs. The majority of asthma symptoms are generally treatable with proper medication but in some cases, they may turn critical [1], and therefore accurate and timely diagnosis is essential to deliver the right treatment for the patients, as early intervention appears to prevent permanent airway blockage.

Pathogenesis of asthma has been associated with an anomalous autonomic nervous system (ANS) function [3], mainly with its parasympathetic branch, as parasympathetic nervous system (PSNS) is responsible for controlling the bronchomotor tone and bronchoconstriction functions [4]. Similarly, the absence of sympathetic innervation of airway smooth muscle reflects the PSNS activity responsible for different bronchomotor tone in asthmatics [4]. Since the high frequency (HF) component of heart rate variability (HRV) spectrum is primarily related to PSNS activity, the study of HRV has been utilized to evaluate amplified vagal tone in asthmatic patients [2]. Heart Rate (HR) in beats-per-minutes (bpm) and Oxygen saturation in the blood (SpO_2) are also key metrics in describing moderate asthma (peak expiratory flow rate (PEFR) > 50% to 75% best, $SpO_2 \geqslant 92\%$, HR< 110bpm) from acute severe asthma (PEFR > 33% to 50% best, $SpO_2 \geqslant 92\%$, HR $\geqslant$ 110bpm); and life-threatening asthma (PEFR < 33% best, SpO_2< 92%, bradycardia) [5].

Asthmatic patients are advised to constantly monitor their SpO_2, HR, and HRV to make sure they are within the expected limits. For HR/HRV measurements, the conventional methods are done in a hospital environment using Electrocardiography (ECG)[6],[7]. The SpO_2 can be also monitored at the hospitals/clinics using commercial pulse oximetry. This process requires frequent visits to a doctor or a clinic for measurement of the vitals and can often be cumbersome. A more viable approach is to be able to measure the HR/HRV and SpO_2 at home, in a non-invasive manner where the ease and the comfort of the asthmatic patient are ensured.

This paper presents a low power, and non-invasive asthma monitoring system integrated on-chip based on Photoplethysmography (PPG) signal. PPG signal is used as a measure of change in the arterial blood volume as the absorption of optical signal in arteries introduces changes in light intensity resulting in the formation of a periodic signal; meanwhile, the absorption of the optical signal in veins, skin, and tissues remains relatively constant for a short measurement period [8],[9]. A red LED and an Infrared (IR) LED, of wavelengths 660nm and 940nm respectively, are used to obtain the PPG signal. The power consumption is a prime concern when using PPG signals because of the high LED currents. This paper presents a low power approach using 0.5V supply and complimentary switching of LED to reduce the total power budget. The proposed system utilizes a switched capacitor trans-impedance amplifier (TIA) to allow high gain (120dB) without using a large feedback resistor.

II. PROPOSED ASTHMA MONITORING SYSTEM

The block diagram of the proposed asthma monitoring system is shown in Fig.1. In the first stage, light from the red LED and IR LEDs is alternated through the finger and onto the photodiode. The photodiode will generate a photocurrent which would be directly proportional to the changes in the light intensity as a result of changes in the arteriole blood volume. This current has a DC component typically in the range of 100nA-10uA with an HR-modulated AC component of amplitude equal to 0.1-2% of the DC component [7]. The amplified output voltage signal will be filtered using a LPF with a cut off frequency of 10Hz to reject out of band noise. The noise-free signal will be digitized using a 10bits SAR

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ADC [7]. The SpO_2, HR, and HRV will be extracted from the digitized signal. In order to allow for a high dynamic range of the PPG analog front end (AFE), the digitized output of the ADC will be used in digital feedback to control the LED brightness and the DC current of the DAC at the input of the TIA to cancel the photodiode's DC current. A current DAC at the input of the PPG AFE is designed to provide a DC current rejection at the input up to 20µA.

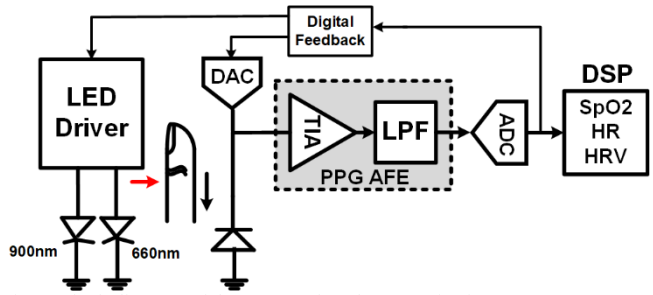

Fig.1: Block diagram of the proposed Asthma monitoring System.

III. LED DRIVERS

For the target application of continuous asthma monitoring system, we aim for minimum power consumption and small form factor. Therefore, the overall system utilizes a single supply voltage of 0.5V. Both red and IR LEDs require a forward voltage of 1.8V and 1.2V, respectively. In order to turn on the LEDs without additional supply voltages, a modified boost converter LED driver based on a design proposed in [7]-[9], is utilized. In this paper, the design is improved to turn on both of the LEDs alternatively at a pulse switching frequency (PSF) of 100Hz. It adopts a clock bootstrapping circuit to reduce the overall size of the design and yet achieve high efficiency [7]. The duty cycle (DuC) of the LED will be adjusted using Pulse Density Modulation (PDM) which will be controlled through the digital feedback. This will help in reducing the LED power consumption and avoid turning on LEDs for unnecessary longer periods. The DuC can be adjusted from 1.25%-25% from a PSF of 100Hz.

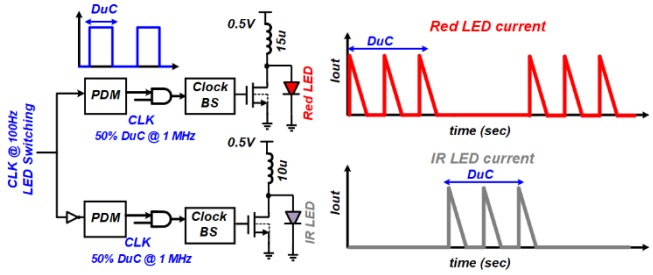

Fig. 2: Schematics of Red and IR LED Drivers.

IV. PPG READOUT CIRCUIT

The conventional TIA requires a large resistor (several to tens of MΩ's) to be placed in the feedback of the Op-amp to achieve high impedance gain [10]. These large resistors will consume a large die area and increase the thermal noise. Therefore, in the proposed PPG readout circuit, the photodiode current will be amplified using a switched capacitor based TIA (SC-TIA)[11]. However, the SC-TIA is prone to undesirable effects such as charge injection and switching noise. The implemented SC-TIA utilizes the principle of correlated double sampling (CDS) to reject the undesired 1/f noise and amplifier offset. The SC-TIA generates an output voltage inversely related to the feedback capacitor value. It also depends on the total input charge that flows through during the designed clock period. The SC-TIA also includes a sample and hold (S&H) circuit at the output to hold the output voltage until the following clock cycle. The SC-TIA contains three different switches operated at three phases of the clock signal Φ1, Φ2, and Φ3. The capacitors C_f, C_{out}, and the clock cycle T are selected based on the desired gain of SC-TIA.

The SC-TIA has two modes of operation, reset mode and amplification mode. In the rest mode: the switches operated by Φ1 and Φ2 are closed. Thus, V1 is considered as a virtual ground while the V2 is connected to an actual ground. Thus, both capacitors C_f and C_{out} have zero charges. On the other hand, the amplification mode is divided into three phases:

a) Φ1 is open while Φ2 is closed: During this phase, the photodiode current will start to be integrated on capacitor C_f, leading to an output voltage V1 proportional to the input current. During this phase, V2 remains at 0 since Φ2 is still connected to ground. Moreover, the offset, as well as flicker noise, are stored on C_{out}.

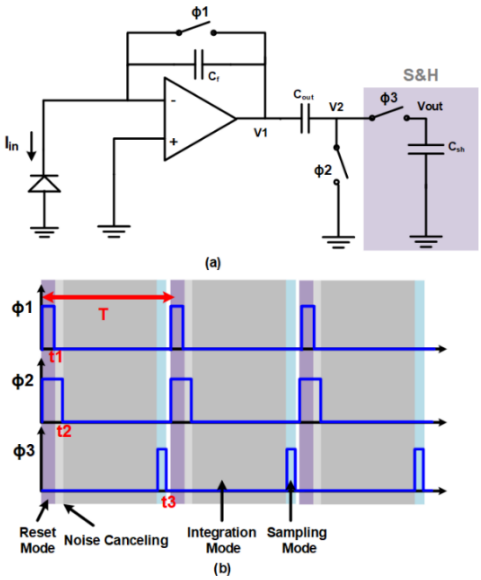

Fig. 3: (a) schematics of SC-TIA, and (b) Timings diagram of SC-TIA.

b) Both Φ1 and Φ2 are open: The input current continues to be integrated on C_f. Since Φ2 is open now, the voltage V2 is not connected to ground anymore; therefore, it follows the value of V1, after subtracting the offset and flicker noise stored at C_{out} following the mechanism of CDS to cancel 1/f noise and circuit offset.

c) S&H operation (Φ3 is closed): During this phase, where Φ3 is closed for a short period of time, the S&H circuit samples V2 and the output V_{out} is held until the next time Φ3 closes.

At the end of each clock period, the output voltage of SC-TIA is given by the following expression [11]:

$$V_{out}(nT) = \frac{1}{C_f}\int_{t_2+(n-1)T}^{t_3+(n-1)T} I_{in}\,dt \qquad (1)$$

In this design, C_f is selected to be 1.2pF with a clock of 1MHz.

In order to realize a LPF on-chip with a low cutoff frequency (10 Hz), a SC stray insensitive LPF is utilized. The

resistor is implemented using SC circuit, which allows for a large resistor without actually placing it on the chip. The schematics of the SC stay-insensitive LPF is shown in Fig. 4.

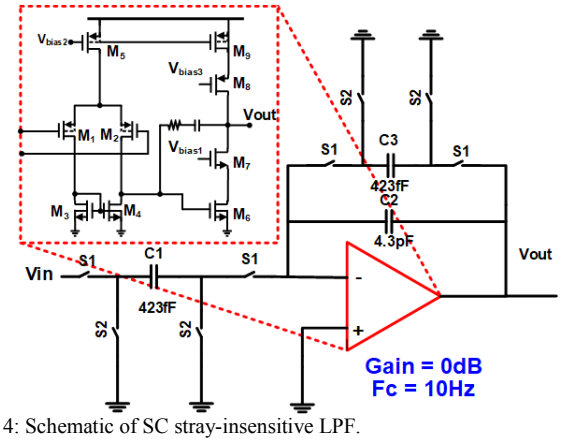

Fig. 4: Schematic of SC stray-insensitive LPF.

The parasitic insensitive integrator controls the sampling capacitor by four switches and two non-overlapping clocks. In phase 1, when S1 switches are closed, the capacitor C1 charges to the Vin and the charge will be also transferred to the capacitor C2, which still has the charge stored in the previous cycle. In phase 2, when S1 switches are open while S2 switches are closed, the capacitor C1 is now discharged since both terminals are connected to ground. The parasitic capacitances at both the input and near to C2 will be charged and discharged to ground, thus not causing any charge transfer output. The circuit is therefore insensitive to parasitic capacitances. By shunting the integrating capacitor with an SC-based resistance, a damped behavior is achieved to provide a finite DC gain. The ratio of C1/C3 is designed to be 1 such that no DC gain will be provided.

The filtered signal will be converted to digital using a 10bits SAR ADC.

V. SpO2/HR/HRV Extraction Processor

The digitized PPG signal is then sent to the vitals extractions processor on FPGA. Fig. 5 shows the flow chart of the operations in the vital extraction processor. Initially, the signal is filtered to remove noise and baseline drift encountered in the PPG signal by applying wavelet transformation. In order to calculate the HR, two methods can be employed: frequency domain versus temporal domain [6]. In this work, we have used the time-domain method of calculating HR through peak detection. The computed number of peaks are then multiplied by 60 and divided by the time duration to get the value for the HR.

For the SpO_2 concentration, the method of ratios of ratios (ROR) is used for the calculation. First, a peak detection mechanism is employed that finds all the available peaks within both the signals, from the red LED and the IR LED respectively. Once the peaks have been detected, the mean of the diastolic and systolic peaks is calculated. The following formula is used for the calculation of ROR, which is a ratio of the difference and sum of these respective peaks [12]:

$$ROR = \left(\frac{RED_{AC}}{RED_{DC}}\right) \Big/ \left(\frac{IR_{AC}}{IR_{DC}}\right) \quad (2)$$

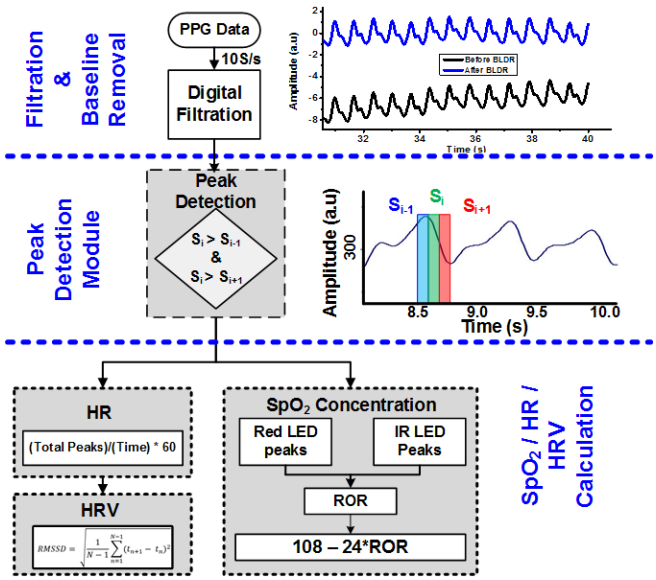

Fig. 5: Flow chart of the vitals extraction process.

Using the ROR calculated above, the following formula is applied to calculate the oxygen concentration in the blood;

$$SpO_2 = C1 - C2\ x\ ROR \quad (3)$$

Where C1 and C2 are factory-calibrated constants, depends on the photodiode used in the circuit. In this system, C1 and C2 are selected to be 108 and 24, respectively. The HRV is computed by finding the root mean square of successive differences (RMSSD) [7]:

$$RMSSD = \sqrt{\frac{1}{N-1}\sum_{n=1}^{N-1}(t_{n+1} - t_n)^2} \quad (4)$$

Where, t_n is the duration between two successive heartbeats. N is the number of all included heartbeats.

VI. Measurement Results and Discussion

In order to validate the performance of the proposed system, the HR and SpO_2 values were calculated for 21 volunteers including people suffering from bradycardia, a low resting heart rate, and tachycardia, a high resting heart rate. Fig. 6 shows the measured red and IR PPG signals (overlaid on the same time scale) for a female subject (age 22 years) along with the measured vitals using the proposed system. To demonstrate the accuracy of the proposed model, the measured values for the HR and SpO_2 concentration are compared with the values from the conventional pulse oximeter used in local clinics and hospitals. As shown in Fig.7(a) and Fig.7(b), the maximum error margin between the actual and measured values of HR and SpO_2 is <1%. The SC-TIA provides a gain of 120dB and bandwidth of 25 kHz. The measured input-referred current noise of the SC-TIA is 7.3pA/√Hz for a bandwidth of 10Hz which reduces the input-referred noise by 89% compared to design in [7].

The asthma monitoring system is realized in CMOS 180nm, with an active chip area of 2.6mm^2 for the signal acquisition. The system also utilizes 2 off-chip inductors for LED drivers. Fig. 8 shows the die photo of the LED drivers and AFE of the system along with the performance summary table. The complete system is powered from 0.5V supply

consuming 1mW/2.3µW for LED drivers and AFE, respectively. The comparison of the state of the art-work is shown in Table 1. The proposed system consumes less power compared to [13] and [14] while providing 3 vitals measurements.

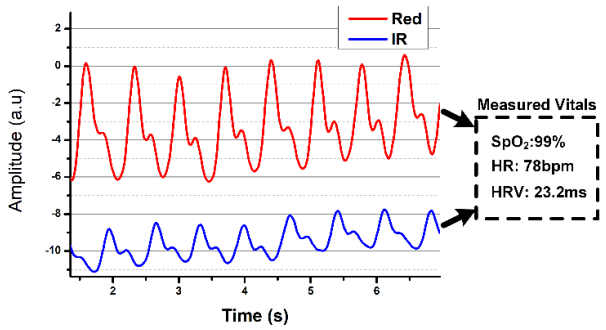

Fig. 6: Measured IR and Red LED PPG signal and extracted vitals.

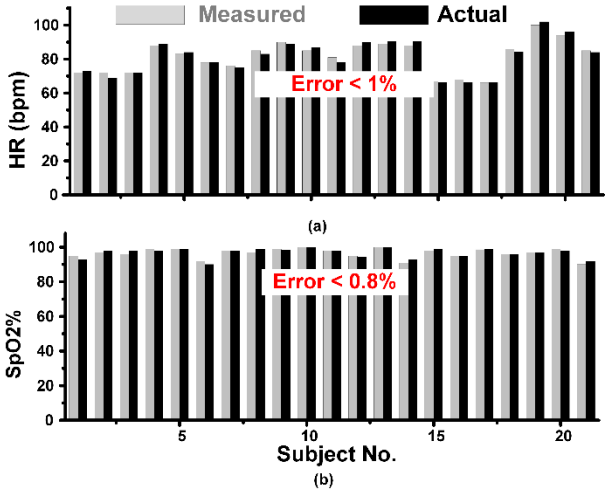

Fig. 7: Measured and real vitals for 21 subjects (a) HR, and (b) SpO_2.

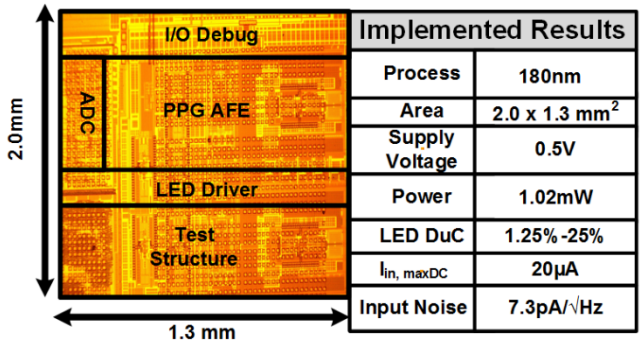

Fig. 8: Die Photo of the Asthma monitoring System and Summary table.

VII. Conclusion

A non-invasive PPG based asthma monitoring system for measuring HR/HRV and SpO2 is implemented. It consists of an AFE on-chip, combined with digital processing on-FPGA where the signal is processed in order to get the values of the required vitals. The mean absolute error for values calculated using 21 subjects come out to be <1% hence our calculated values are quite close to the actual ones.

Table 1: Comparison with the state-of-art- work for PPG SoCs.

	ASSCC'13 [8]	TBIOCAS'15 [14]	TBIOCAS'17 [13]	BioCAS'18 [7]	This Work
Technology	180nm	180nm	180nm	65nm	180nm
Supply	0.5V	1.8V	1.2V	0.5V	0.5V
DC Current Cancellation	4µA	100µA	10µA	4µA	20µA
Integrated Noise (RTI)/√Hz	-	600pA	486pA	68pA	7.3pA
Vital Extraction	HR	-	HR/HRV	HR/HRV	SpO_2/HR/ HRV
Power (Readout)	4µW	216µW	172µW	3µW	2.3µW
Power (LED Driver)	2.66mW	1.1mW	1.2mW	1.1mW	1mW

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References

[1] C. M. Ionescu, "The human respiratory system," in The Human Respiratory System, pp. 13–22, Springer, 2013.

[2] J. Milagro, *et. al*, "Nocturnal Heart Rate Variability spectrum characterization in Preschool Children with Asthmatic Symptoms," in *IEEE J.Bio. Med. Health and Info.*, vol. 22, no. 5, pp. 1332 – 1340, Sep. 2018.

[3] D. Caudri, *et. al*,"Predicting the long-term prognosis of children with symptoms suggestive of asthma at preschool age," *J Allergy Clin Immunol*, vol. 124, no. 5, pp. 903–910, 2009.

[4] M. Lewis, A. Short, and K. Lewis, "Autonomic nervous system control of the cardiovascular and respiratory systems in asthma," *Respir Med*, vol. 100, no. 10, pp. 1688–1705, 2006.

[5] E. J Welsh, K. M Kew, and R. Carr," Pulse oximeters to self‐monitor oxygen saturation levels, as part of a personalized asthma action plan for people with asthma," in *The Cochrane Library*, issue 3, Sep. 2015.

[6] S. Abubakar, *et al.*, "A Wearable Auto-Patient Adaptive ECG Processor for Shockable Cardiac Arrhythmia," *IEEE Asian Solid-State Circuits Conf. (ASSCC),* Nov. 2018, pp. 267-268.

[7] W. Saadeh, *et al.,* "A 0.5V PPG-based Heart Rate and Variability Detection System,"*IEEE Bio. Cir. & Sys. (BioCAS),* pp.1-4, Oct. 2018.

[8] W. Saadeh, *et al.*, "A > 89% efficient LED driver with 0.5V supply voltage for applications requiring low average current," *IEEE Asian Solid-State Circuits Conference (A-SSCC)*, Nov. 2013, pp.273-276.

[9] W. Saadeh, *et al.,* "High efficiency led driver circuit," *US Patent Application*, US20140225524A1.

[10] A. Hina, *et al.*, "A Single LED Photoplethysmography-Based Noninvasive Glucose Monitoring Prototype System," *IEEE International Symposium on Circuits and Systems (ISCAS)*, May. 2019, pp. 1-5

[11] Y. Tang, *et al*, "An ultra-low noise Switched-Capacitor Transimpedance Amplifier for parallel Scanning Tunneling Microscopy," *IEEE SENSORS*, pp. 1-4, Oct. 2012.

[12] D. Shao *et al.,* "Noncontact Monitoring of Blood Oxygen Saturation Using Camera and Dual-Wavelength Imaging System," *in IEEE Trans. on Bio. Eng.*, vol. 63, no. 6, pp. 1091-1098, June 2016.

[13] V. R. Pamula *et al.*, "A 172 µW compressively sampled photoplethysmographic (PPG) readout ASIC with heart rate estimation directly from compressively sampled data," *IEEE Trans. Biomed. Cir. Syst.*, vol. 11, no. 3, pp. 487-496, Jun. 2017.

[14] E. S. Winokur, *et al.,* "A low-power, dual wavelength photoplethysmography(PPG) SoC with static and time-varying interferer removal," *IEEE Trans. Biomed. Circuits Syst.*, vol. 9, no. 4, pp. 581–589, Aug. 2015.