# Development of a Smartphone-Based Pulse Oximeter with Adaptive SNR/Power Balancing

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***Abstract*— Millions worldwide suffer from diseases that exhibit early warnings signs that can be detected by standard clinical-grade diagnostic tools. Unfortunately, such tools are often prohibitively expensive to the developing world leading to inadequate healthcare and high mortality rates. To address this problem, a smartphone-based pulse oximeter is presented that interfaces with the phone through the audio jack, enabling point-of-care measurements of heart rate (HR) and oxygen saturation ($SpO_2$). The device is designed to utilize existing phone resources (e.g., the processor, battery, and memory) resulting in a more portable and inexpensive diagnostic tool than standalone equivalents. By adaptively tuning the LED driving signal, the device is less dependent on phone-specific audio jack properties than prior audio jack-based work making it universally compatible with all smartphones. We demonstrate that the pulse oximeter can adaptively optimize the signal-to-noise ratio (SNR) within the power constraints of a mobile phone (< 10mW) while maintaining high accuracy (HR error < 3.4% and $SpO_2$ error < 3.7%) against a clinical grade instrument.**

## I. Introduction

In developing countries, lack of adequate diagnostic and healthcare tools results in a substantial loss of life each year [1]. Illnesses such as asthma, chronic obstructive pulmonary disease, sleep apnea, sudden infant death syndrome, and cardiovascular disease impact the quality of life for billions of people and cost billions of dollars annually to treat [1], often inadequately in many settings. While many of these illnesses can be diagnosed and treated, lack of access to appropriate preventative healthcare technology contribute to maternal and neonatal health problems. In response, community health workers are deployed as frontline providers equipped with low-cost health technologies that can have significant impact on maternal and neonatal health as well as overall health of the general population [2]. Hence, there is a desire to develop sensitive, yet portable, inexpensive diagnostic devices. With the processing power and widespread availability of mobile phones and smartphones, there is a tremendous opportunity to meet this need. Currently there are 7.4 billion mobile connected devices (including 1.76 billion smartphones) in use worldwide [3]. With the processing power available on these portable devices, there is an opportunity to transform a mobile phone into highly specialized medical devices by augmenting it with additional sensors.

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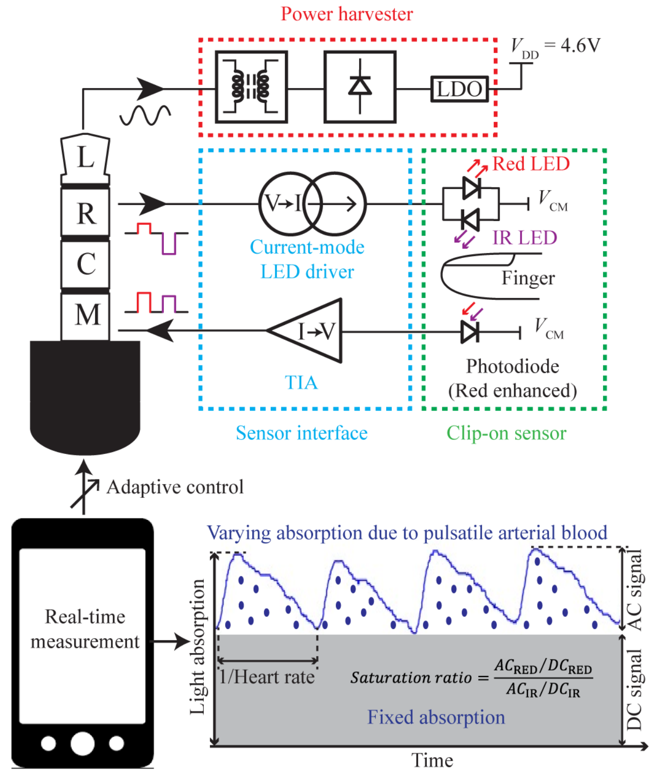


Fig. 1. Overview of the proposed mobile phone-based pulse oximeter.

Pulse oximetry is a non-invasive method to monitor arterial blood oxygenation by measuring the relative absorption of red (absorbed by deoxygenated blood) and infrared (absorbed by oxygenated blood) light. Pulse oxygenation is an important vital sign used in intensive care units, operating theaters, clinical screenings, and checkups. Several researchers have adopted pulse oximeters to mobile phones [4]–[6]. For example, reflection pulse oximetry integrated with a smartphone achieved high accuracy and compact size; however, it required opening the phone back-cover (unfeasible on many phones) and manual wiring since it did not utilize a standard port [6]. Others have used the audio jack, available on nearly all phones, for transmission pulse oximetry where the light emitting diodes (LEDs) were driven directly by the audio outputs [4], [5]. However, the variation (e.g., voltage swing and output impedance) between different makes and models of smartphones limits the utility and measurements quality.

To address the limitations in the prior work, we propose an audio jack-based pulse oximeter with a power harvesting circuit and locally regulated supply, thereby eliminating the dependence on the audio jack output characteristics and the need for an external battery (Fig. 1). A current-mode LED driver linearly modulates the LED current and light intensity

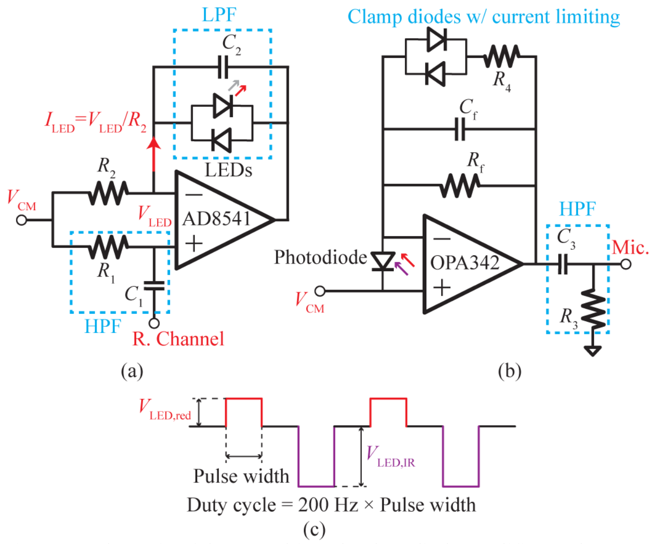


Fig. 2. a) Schematic of the LED driver, b) photodiode amplifier, and c) LED control waveform (from the right audio channel).

enabling precise control of the signal quality. To obtain an optimal signal quality from different host devices, an adaptive SNR/power tuning method is proposed afforded by the abundant computation ability of the mobile phone. All signal processing and adaptive control is performed by the mobile phone. A user interface (UI) was developed to graphically display measurement results. The hardware design is detailed in the next section.

## II. Circuit Architecture

A block diagram of the mobile phone-based pulse oximeter is shown in Fig. 1. The device connects to its host via the audio jack and interfaces with most standard finger clip pulse oximeter sensors. A bi-directional pulsed current alternatively turns on the red and infrared (IR) LEDs at a controlled light intensity. The photodiode, mounted on the other side of the finger clip, converts the transmitted light into current, which is amplified and sent back to the host device. To be compatible with multiple host devices, the circuit features a local regulated supply voltage and a current-mode LED driver, such that the circuit performance as well as the light intensity have less dependence on the host device properties (e.g., voltage swing, output impedance and current limit). Since the audio outputs are AC coupled, the circuit cannot extract DC power directly. Instead, a power harvester is used, which boosts, rectifies, and regulates an AC tone (14 kHz) on the left audio channel to a fixed 4.6 V local supply voltage ($V_{DD}$) [7]. This supply voltage ensures enough headroom for the LED driver. It has been shown that >10 mW can be harvested from all phones tested using this circuit [7].

The LED driver consists of an op-amp with the LEDs in feedback (Fig. 2a). Bi-directional voltage pulses are generated by the smartphone and sent out through the right audio output where they are level shifted to a common-mode voltage ($V_{CM}$=2.3 V) through an *RC* high-pass filter (HPF) with a cutoff frequency <1 Hz. The pulse voltage, $V_{LED}$, is converted into a current, $I_{LED}$, via $R_2$, as $I_{LED}=V_{LED}/R_2$, to drive the LEDs. By adjusting $V_{LED}$, both $I_{LED}$ and subsequently the light intensity are linearly changed to control the SNR. Since tissue has a higher transmission of red light, $V_{LED}$ is designed to be asymmetric (Fig. 2c) to maintain similar signal swing at the TIA output. The pulse repetition rate is set to 200 Hz, sufficiently higher than the typical HR (<3 Hz), thereby ensuring good accuracy when restoring the photoplethysmogram (PPG) signal. A feedback capacitor $C_2$ serves as a low-pass filter (LPF) to attenuate 14 kHz interference coupled in from the power harvesting circuit. In between each red and IR reading, a measurement of the ambient lighting is also taken and used for background suppression (described later). Since LEDs typically require >1 mA of bias current, the op-amps in the LED driver and common-mode voltage generation circuit should have high output current, but low quiescent current for such a low-power design. We used Analog Devices AD8541 for both due to the low static current (45 µA) and high output current (10 mA).

The photodiode that receives the transmitted light is biased between two inputs of the op-amp in a zero-bias scheme (Fig. 2b). Compared to the commonly used reverse-biased configuration, this configuration has lower dark current to save output voltage headroom and lower noise at the expense of longer settling time, which is not an issue for low frequency (200 Hz) operation. Clamp diodes and a current limiting resistor are added to protect the phone by preventing an over-voltage condition when a finger is not inside the sensor and the light is unblocked. Since the microphone channel has its own DC bias voltage within the audio jack, the TIA output is AC-coupled to it via a HPF. A Texas Instruments OPA342 was chosen for low input current noise, low input bias current, and low quiescent current (150 µA).

## III. Adaptive LED Control

By utilizing the phone to generate the driving waveform and to record the transmission light intensity, we propose a closed-loop system where the phone can adaptively adjust the LED driving signal. This extra degree of freedom allows the phone to manage the power consumption and received SNR dynamically. This is especially useful when the environment changes, such as for long-term monitoring. It also allows the device to interface with multiple phones, each with a different optimal driving signal requirement for maximum efficiency, and our device has the potential to adjust and find that point. To quantify the effect of our parameter adjustment, we define a metric for the error that accounts for both repeatability and accuracy in the HR and $SpO_2$ signals,

$$Error = \sqrt{(HR_{\%\mathrm{Err}}HR_{\sigma})^2 + (SpO_{2,\%\mathrm{Err}}SpO_{2,\sigma})^2} \quad (1)$$

where $HR_{\%Err}$ is the normalized average error between the set of values measured by a medical grade pulse oximeter (Masimo RAD 87) and the set of values simultaneously measured by our device, and $HR_{\sigma}$ is the normalized standard deviation of the measurements taken. $SpO_{2,\%Err}$ and $SpO_{2,\sigma}$ are similarly defined for $SpO_2$. Each point was sampled >30 times to ensure that the measurement was statistically relevant.

There are two variables that the phone can adjust that affect the phones power and SNR: 1) the duty cycle and 2) the amplitude of the driving signal, $V_{LED}$. By reducing the duty cycle, one can reduce the LED "on" time, which is the most power consumptive process. However, reducing the duty cycle too much degrades the signal quality by integrating

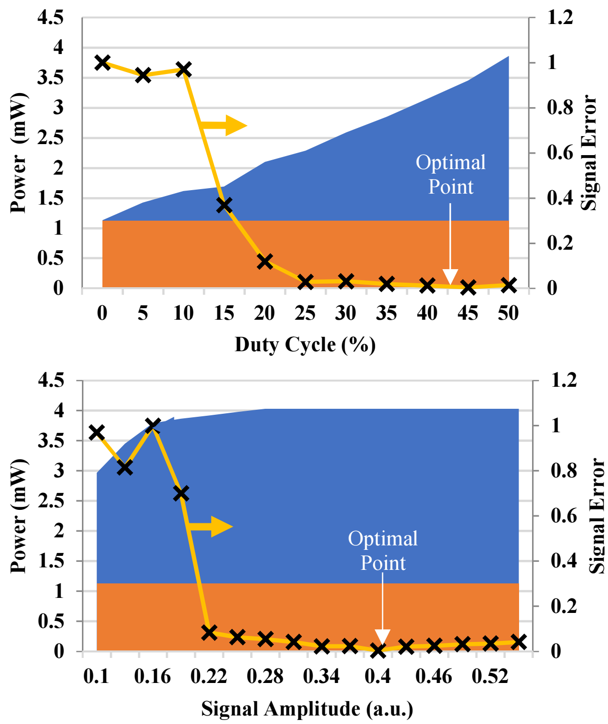


Fig. 3. Plot of measured signal error (yellow line) and the total power consumed with quiescent power shown in orange and dynamic power shown in blue vs. a) the duty cycle and b) the amplitude of the LED driving waveform.

more noise due to a wider noise bandwidth. Figure 3(a) shows the effect of the duty cycle on the signal error. As one can see, there exists an optimal duty cycle that balances the signal quality and power consumption. By adjusting $V_{LED}$, the light intensity of the LEDs can be modulated, thereby modulating the SNR accordingly (Fig. 3b). At low $V_{LED}$, the LEDs are not being driven strongly enough to return a measurable result, as is often the case where there is a high amount of ambient light. A large amplitude, on the other hand, may saturate the LED driver, and overload the microphone ADC while consuming unnecessary power. Similarly, there is a tipping point after which the signal error drastically reduces. With our device's ability to dynamically adapt, the phone will be able to find the minimum power setting that provides adequate SNR required for an accurate measurement given the current environmental conditions.

## IV. Signal Processing

One of the main advantages in creating a mobile phone-based device compared to a standalone equivalent is that the phone provides a powerful CPU with abundant memory to perform intensive calculations at no extra cost. This processing power combined with our device's adaptability provides the opportunity for a wide range of advanced signal processing techniques that standard pulse oximeters would find prohibitive to implement due to their computational complexity [5], [6].

The signal processing steps implemented, shown in Fig. 4, begin with digitally low-pass filtering the signal recorded at the microphone at 700 Hz, then down sampling the waveform

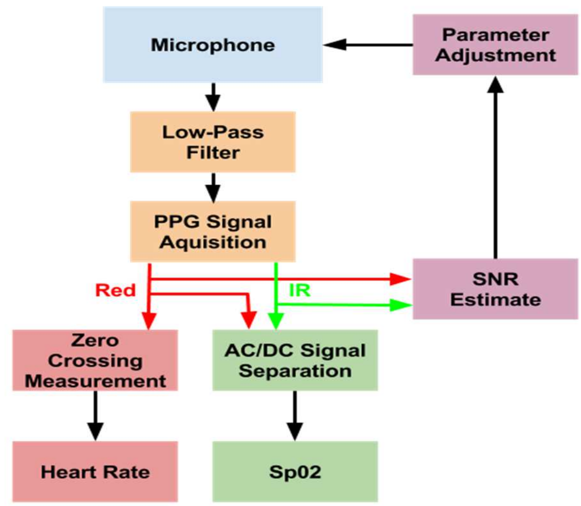


Fig. 4. Signal processing block diagram for the phone application to acquire heart rate and $SpO_2$ from the proposed pulse oximeter.

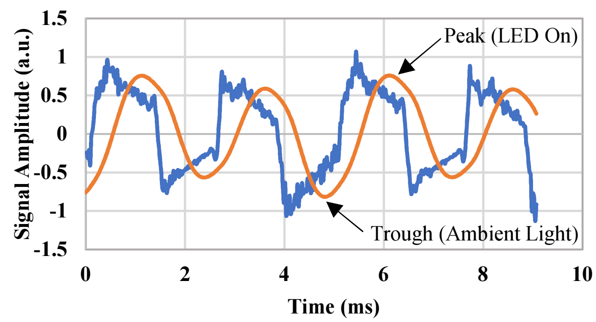


Fig. 5. Microphone waveform before low-pass filter (blue) and after (orange).

from the sampling rate of the audio codec (44.1 kHz) to 200 Hz to construct a PPG waveform (Fig. 5). This is done by separating the peaks and troughs from the filtered microphone output and building the red and IR PPG's from this data. Since the trough represents the signal level without the LEDs illuminated, by using the difference between the peak and trough to form a single PPG point the effect of ambient light is eliminated.

Due to the embedded HPF at the microphone input, the received signal is heavily distorted (blue curve in Fig. 5) and cannot be restored without knowing the precise transfer function of this filter, which differs between phones. Instead of equalizing the distorted signal, we propose to extract only the fundamental tone (400 Hz) amplitude, which lies higher than the HPF corner frequency. To enable this, a 3$^{rd}$ order Butterworth FIR filter was implemented to aggressively low-pass filter the signal to reduce the square wave to its fundamental frequency and first few harmonics (Fig. 5). The sample from each segment becomes the maximum value of each peak and the minimum of each trough. This method was able to eliminate the problems created by each phone and still preserve the shifts in the peak and trough magnitudes due to changing absorption rates. This allowed us to acquire the PPG signals used in standard pulse oximetry. From there the PPG was windowed into blocks of 500 samples each and the DC and AC components were extracted to calculate the $SpO_2$

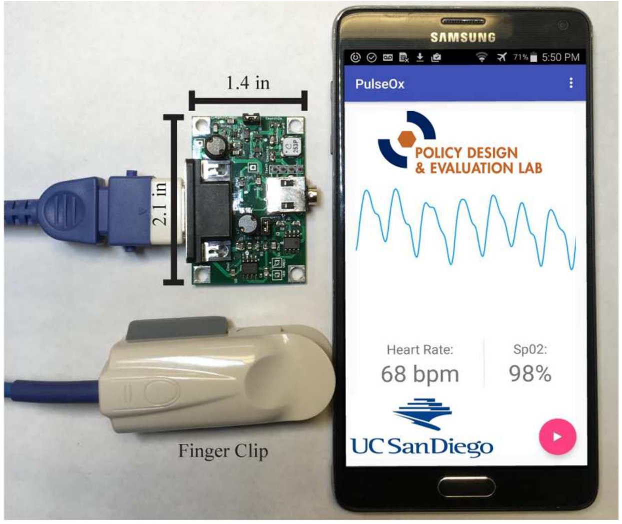


Fig. 6. a) Photograph of PCB, finger clip, and a smartphone. b) Screenshot of app while measurement is in progress.

Table I: $SpO_2$ Measurement Comparison

| | | **%Error Heart Rate** ($HR_{\%Err}$) | | | | **%Error $SpO_2$** ($SpO_{2,\%Err}$) | | | |
|---|---|---|---|---|---|---|---|---|---|
| | *Subject* | *1* | *2* | *3* | ***Mean*** | *1* | *2* | *3* | ***Mean*** |
| **This Work** | *Gal. S6* | -6 | 1 | 1.8 | **3.2** | 5.3 | 1.4 | 1.8 | **2.8** |
| | *Note 4* | 4.2 | 0 | -6 | **3.4** | 4.9 | 0 | -6 | **3.7** |
| | *Note 3* | 3.2 | 0 | -2 | **1.6** | -3 | 0 | -2 | **1.6** |
| | *LG V10* | 0 | 4 | -3 | **2.5** | -2 | 4.3 | -3 | **3.1** |
| **iOximeter** | *Gal. S6* | 5.3 | 4 | 0 | **3.1** | -1 | -1 | 0 | **0.7** |
| | *Note 4* | 4.9 | 0 | 3.4 | **2.8** | 1 | 1 | -2 | **1.3** |
| | *Note 3* | -3 | 6 | 3.5 | **4.1** | -2 | -2 | -2 | **2.0** |
| | *LG V10* | -2 | 3 | 0 | **1.6** | 2.1 | -5 | -3 | **3.4** |

where the AC component was defined as the standard deviation of the PPG waveform. The heart rate was acquired by counting zero crossings per sample span, and relating it to beats per minute (BPM).

## V. Experiments and Results

Figure 6 shows the constructed device as well as the phone application developed. The device was used with a Samsung Note 4 to conduct the following experiments to demonstrate the functionality and performance. First, we compared our device and a commercially available smartphone pulse oximeter (iOximeter) against a medical grade table top pulse oximeter (Masimo RAD 87). We took simultaneous measurements with all three devices and compared the measurements from both smartphone-based devices to the medical grade pulse oximeter for three different stationary subjects. This measurement process was then repeated with different phones to demonstrate the adaptability.

The results in Table I demonstrate that our pulse oximeter is capable of preforming well using several host devices. The maximum mean error in heart rate across three test subjects was 3.4% and a 3.7% error in $SpO_2$. These results compared well to the iOximeter, which had a maximum mean error of 4.1% and 3.4% in heart rate and $SpO_2$, respectively. We then used an artificial heart rate simulator (BE Biomedical PS-2110 Patient Simulator) to test our device over heart rates ranging from 30 to 180 BPM. As shown in Fig. 7, our device can accurately (error <1.84%) track over a wide range of

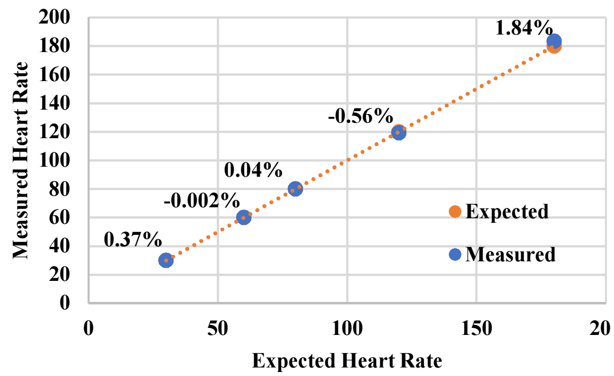


Fig. 7. Measured heart rate with error from the simulator heart rate annotated.

heart rates. It showed extremely accurate measurements for heart rates less than 120 BPM (<0.56%) and only increased to 1.84% at very high heart rates (180 BPM).

## VI. Conclusions

A portable pulse oximeter was presented that was designed to take advantage of the growing availability of powerful hand held devices. By interfacing the device with a smartphone, we create a low-cost device that has the hardware required to perform complex signal processing algorithms and excellent displays. The module demonstrates a high level of adaptably, allowing it to optimize its power consumption and SNR. The module requires only 2.5 mW, which is within the power constraint of most smartphones. We demonstrate the device's ability to run high complexity signal processing by preforming heart rate and $SpO_2$ acquisition in real time and displaying the results to the user. The performance compares well to bench top pulse oximeters, with a maximum 3.4% deviation in heart rate and 3.7% deviation in $SpO_2$ and matches well with similar commercially available smartphone pulse oximeters (4.1% in heart rate and 3.4% in $SpO_2$).

## Acknowledgment

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