# Design and Parametric Analysis of a Wearable Dual-Photoplethysmograph Based System for Pulse Wave Velocity Detection

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***Abstract*—Ambulatory blood pressure monitoring (ABPM) is a term used to describe measurement of blood pressure (BP) at regular intervals and can be an important diagnostic tool, especially for hypertensive patients. Since traditional cuff-based measurement of blood pressure is not convenient for a continuous, ABPM measurement, one approach that is being explored is pulse wave velocity (PWV), which is known to be correlated with blood pressure. However, an easily wearable and reliable system has not been realized to date. In this work, we examine the feasibility and design requirements for measuring PWV using two photoplethysmography sensors placed 4 cm apart on the arm. Measurements of PWV *in vivo* were made with our system and we showed that PWV measurements changed in accord with induced variations in blood pressure of the subject. Furthermore, we examined the minimum requirements for the sampling rate and bit resolution for the analog-to-digital converter (ADC) in our system with error in PWV measurement as the criterion. Our results show the feasibility of measuring PWV at small distances and outlined the design requirements for an ABPM device.**

***Keywords—Blood pressure; pulse wave velocity; photoplethysmography; wearable sensors***

## I. Introduction

According to the Centers for Disease Control and Prevention (CDC), there are 70 million American adults who have hypertension [1]. A patient can be diagnosed with hypertension with a systolic blood pressure (SBP) of >140 mmHg and a diastolic BP (DBP) of >90 mmHg, and pre-hypertension with a SBP of 120-139 mmHg, and a DBP of 80-89 mmHg [2]. Without proper lifestyle changes or medical intervention, hypertensive patients are at an increased risk for developing cardiovascular disease (CVD). One method used to diagnose hypertension, and related CVD risk, is through ABPM. It was shown in previous studies that ABPM could be used to determine the risk factor of CVD in patients who had never been treated for hypertension or had been untreated for weeks [3]. ABPM could also aid in measuring the effectiveness of antihypertension drugs to personalize treatment for each hypertension patient. One method for ABPM utilizes an occluding cuff and methods such as oscillometry to measure BP at defined intervals throughout a 24-hour period. Two limitations of these types of ABPM devices are their cumbersome size because of the required occluding cuff, and the low frequency of measurements that can be used because of the risk of pressure ischemia from the occluding cuff.

Another attempt at continuous or ambulatory BP monitoring is the approximation of BP through the measurement of pulse wave velocity or the use of pulse transit time (PTT). PTT is the time interval between the peak of the QRS complex of an electrocardiogram (ECG) and a characteristic point (e.g. point of max slope or peak value) on a photoplethysmograph (PPG) wave [4]. PWV is a measure of two quantities, the time an arterial pulse takes to travel through an arterial segment generated by the contraction of the left ventricle, and the distance between the two measuring sites [5]. Through approximation of path length for blood flow and measurement of PTT, the PWV can be calculated. PWV is known to be correlated with BP can potentially be used to provide cuffless ABPM capabilities [4],[6]-[8].

Devices that use PTT and PWV to estimate BP do so to create wearable systems that are less cumbersome and more user friendly than traditional ABPM devices. The size limitation of the occluding cuff in current ABPM systems is removed in these devices because they rely on PPG which is traditionally measured on the wrist or finger, and ECG which can be obtained by at least three electrodes placed across the heart. The low sampling frequency limitation of existing ABPM systems is addressed in two ways: (1) the lack of an occluding cuff removes the risk of pressure ischemia, and (2) the devices measure the PTT, PWV, and calculate BP for each pulse continuously which produces more BP measurements in a 24 hour period. The backbone of these sensing devices is their analog front end (AFE) circuitry that enables transduction of physiological signals.

The AFE proposed in this paper requires no ECG sensor, and instead utilizes two PPG sensors to detect PWV. The advantage of this approach is, although ECG can be easily measured, the need to have electrodes placed on either side of the heart creates device requirements that can limit the usability of a wearable device. The AFE proposed also uses a high sampling frequency which enables the PPG sensors to be placed in closer proximity with lower separation distance and still maintain adequate sensitivity of PWV detection. These two advantages will allow for this AFE to be utilized in a compact, wearable BP monitoring device. The contributions of this paper are:

- Development of a custom AFE for measuring PWV using two PPG sensors placed close to each other on the arm to emulate a wearable system.
- Statistical verification of measurements using *in-vivo* experimental data collection
- Mathematical modeling and discussion of ADC design requirements for robust and accurate measurement of PWV in a wearable system.

## II. Related Work

A previous device developed to estimate BP through PWV is the BioWatch [6]. The BioWatch is a wrist based ECG/PPG coupled device which places three dry ECG electrodes on the underside and top of the device. The PPG sensor is a flat reflectance-type PPG placed on the underside of the device. PWV was calculated by the BioWatch by approximating the arterial path length as half of the user's height and measuring PTT using the peak in the QRS complex of the ECG signal and point of max inclination on the pulse wave. The limitation with this device is that it requires the use of both hands to measure ECG and hence PWV. The proposed device will only require sensing on a single arterial segment on the underside of the wrist, the other hand of the user is not required to measure PWV.

Another device developed was an adhesive chest sensor that detected a so-called pulse wave transit time (PWTT), which is synonymous with PTT [7]. The differential electrodes for ECG are placed 40 mm apart on the underside of the device along with a reflectance-type PPG sensor centered between the two electrodes. The device was meant to be placed on the user's chest, specifically the sternum, to reduce myoelectric interference from muscle tissue. Placement on the sternum also reduces the risk of detachment because there is less displacement of the sensors due to motion. Fuke et al. used the onset, or foot, of the pulse wave and the previously determined QRS peak to calculate their PWTT. For our system, we will test the viability of measuring PTT using the time difference between the onsets of the pulse waves.

Both of the previous devices utilized ECG and PPG signals to calculate PWV, and apart from being relatively inconvenient, using the ECG in conjunction with PPG can result in errors due to variable ejection fraction times across patients. The possibility of measuring local PWV over the carotid artery was tested in another work with an AFE circuit that generated two PPG signals with a single source [8]. The prototype AFE circuit consisted of two photodiodes placed equidistant from a single IR LED. The system utilized a National Instruments® DAQ that acquired samples simultaneously at 100kSPS. Our system will use an ADC chip to collect, store, and visualize data in order to give us the opportunity to reduce the size of the intended device instead of relying on a larger DAQ system.

## III. Materials And Methods

### A. Design of Analog Front End Circuit

The AFE circuit consists of transmit (TX) and receive (RX) blocks. The TX block controls the LED current amplitude, and the RX block includes the transduction, conditioning, and digitization circuits. There are two reflective type PPG sensors, and each sensor has one photodiode (Fairchild Semiconductor – QSB34CGR) centered between two green LEDs (Kingbright – APL3014MGC-F01) which have a peak wavelength of 574 nm. Saturation does not occur despite the absence of an optical barrier because of the arrangement ensuring the viewing angles of the LEDs do not intersect with the reception angle of the photodiode. Current control to the LEDs is carried out using a constant +5V voltage and a potentiometer. The current output from the photodiodes are sensitive to environmental noise so ground shielded cables (General Cable – C1352A. 18. 10) were used to connect the PPG sensors to the trans-impedance amplifiers (TIA) (Texas Instruments – OPA381AIDGKT). After the I-V conversion, the signal passes through a conditioning block, composed of a passive RC anti-aliasing filter with a cutoff frequency of 50 kHz and a fully differential amplifier (FDA) stage which buffers the differential inputs to the ADC (Texas Instruments – THS4521DGK). The digitization is carried out using a delta-sigma ADC (Texas Instruments – ADS1274IPAPR). The FDA and ADC are used as part of the ADS1274 performance design kit (Texas Instruments - ADS1274EVM-PDK) to allow us to easily modify platform settings to develop the AFE. The circuit prior to the digitization circuits is shown, with functional blocks marked, in Fig. 1.

### B. Local Pulse Wave Velocity

The first step to measuring PWV is the calculation of the PTT over an arterial segment. The proposed AFE utilizes two dual-source PPG sensors to detect a pulse wave at two separate locations on the radial artery. The onset or foot of the pulse wave captured by the two sensors will be identified offline, and the time between every two successive feet will give us the PTT, as illustrated in Fig. 2. Local PWV of the radial artery will then be determined using (1).

$$Local\ PWV = \frac{PTT}{D} \quad (1)$$

Where $D$ is the distance between the two PPG sensors along the radial artery.

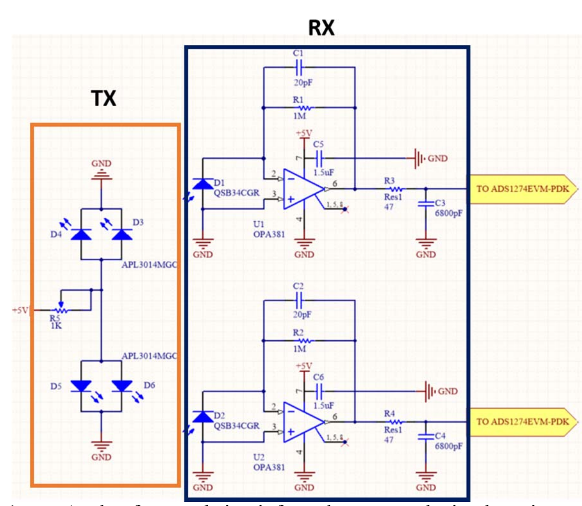


Fig. 1. Analog front end circuit for pulse wave velocity detection system prior to the digitization phase.

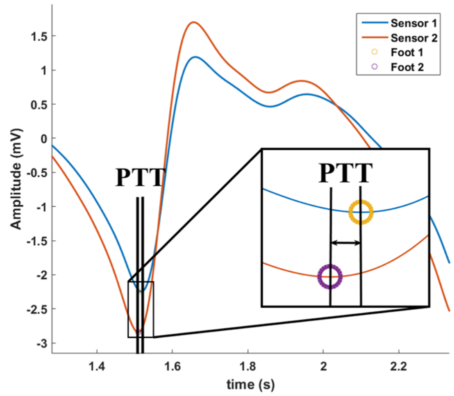


Fig. 2. Pulse wave measured at two separate sensors indicating pulse transit time between feet of the two signals. Sensor 1 is placed distal to Sensor 2 along the radial artery. Signals have been inverted to show typical shape of PPG signals.

### C. Importance of Sampling Frequency

The sampling frequency for the ADC used for this AFE is important because it determines the sensitivity for detecting PWV. Calculating the sensitivity starts with (2), which calculates the number of samples between the feet of the two measured pulse waves given an initial PWV value.

$$N_1 = nint(\frac{D}{PWV_i} f_s) \tag{2}$$

$N_1$ is the number of samples between feet, $D$ is the distance between the two PPG sensors in meters, $PWV_i$ is the initial PWV measurement, $f_s$ is the sampling frequency of the ADC, and *nint()* returns the nearest integer value because it is impossible to have a non-integer value for $N_1$. Adding or subtracting one sample from $N_1$ gives us the number of samples that represent the smallest change in PWV. Using a variation of (2), we can calculate the PWV from the single sample number difference as shown in (3).

$$PWV_f = \frac{D}{N_1 \pm 1} f_s \tag{3}$$

$PWV_f$ is the new PWV as a result of changing the distance between the feet of the two waves by the minimum amount: one sample. Considering the fact that the increase or decrease of one sample produces the same change in PWV, we will only consider a single sample decrease in the number of samples between the feet. Putting the equations together and simplifying gives

$$S_i = \frac{PWV_i}{N_i - 1} \tag{4}$$

Where the sensitivity $S_i$ is ($PWV_f - PWV_i$), the minimum detectable change in PWV with a change in PTT from a one sample number difference between pulse waves with a given $D$ and $fs$. Knowing the sensitivity, we can also model the maximum percent error in PWV due to quantization error:

$$Max\ \%\ Error = \frac{S_i}{PWV_i} \tag{5}$$

### D. Signal Processing

The ADS1274EVM-PDK was connected to a PC using a USB connection, and the data was saved and processed offline in MATLAB®. The data was filtered in the digital domain using a 2nd order Butterworth high pass filter with a cutoff frequency 0.8 Hz to remove baseline wander, and a 4$^{th}$ order Butterworth low pass filter with a cutoff frequency of 10 Hz to remove AC power line noise. PTT was calculated using the onset or foot of the wave, which was calculated using threshold based peak detection.

### E. Valsalva and Mueller Maneuvers

In order to simulate an elevated BP *in vivo,* we leveraged the Valsalva maneuver [9]. The Valsalva maneuver is a process wherein a subject will exhale against a closed airway to increase intrapleural pressure and subsequently increase arterial BP. The Mueller maneuver, which is a process wherein the subject attempts to inspire at residual capacity against a closed/restricted airway to create a negative intrathoracic pressure, was used to simulate a drop in BP [10]. Since a change in BP has been correlated with a change in PWV, the two maneuvers were incorporated into the *in vivo* experiments to assess the AFE's capability of detecting corresponding changes in PWV [6]-[10].

## IV. Results and Discussion

### A. In-Vivo Experiments

In order to verify the functionality of the system, data collection was carried out on the underside of the right wrist for a single subject. The informed consent and protocol were approved by the IRB at Texas A&M University. The PPG sensors were placed on the radial artery 4cm apart using adhesive wrap. Subject was instructed to remain in the supine position throughout the experiment, and asked to either continue breathing normally, perform the Valsalva maneuver, or perform the Mueller maneuver in successive trials. Three trials were conducted for each of the three breathing states, and each trial consisted of 8 seconds of data collection, with sufficient time in between trials to allow for recovery from the physiological effects of the breathing maneuvers. The average PWV as measured offline for the resting state (normal BP) trials, Valsalva maneuver (elevated BP) trials, and Mueller maneuver (reduced BP) trials were 7.11m/s, 9.02m/s and 4.35m/s respectively. The resting average PWV is within the expected human physiological range of about 6 – 14 m/s.

### B. ADC Parameter Analysis

We evaluated the minimum requirements and the effects of varying two parameters of the ADC: sampling rate and bit resolution. We saved all the data at a sampling rate of 128kHz and the full resolution of 24 bits. Subsequently, in MATLAB we progressively decimated and reduced the resolution of the data to simulate different ADC configurations. In Fig. 3, we plot the measured experimental percentage error in PWV as a function of the sampling frequency, and compare it to the modeled maximum percentage error from (5). Here experimental error is defined as the error with respect to the PWV calculated in the highest ADC configuration: 128kHz sampling and 24-bit resolution. We see that the measured error for the various breathing maneuvers largely follows the expected model described in Section III C. In Fig. 3(a) the experimental error is higher than the model in places, and this mismatch is likely because the PWV itself varied from beat to beat during the trial whereas the modeled error is for the average PWV over the whole trial.

For a given sampling rate, we see in Fig. 4 that reducing the ADC resolution does not have too much effect on the error until it goes below about 16 bits. In general, we note that to keep the error below 5%, the sampling rate must be higher than 1280Hz and the ADC resolution must be at least 15 bits when the separation distance is 4cm. The error due to sampling rate reduction is easily understood to be a result of deformed features due to missed samples. In order to illustrate the reason for resolution error, Fig. 5 shows the PPG signal sampled at 512 Hz for a single channel at full resolution, and that same channel after reduction to a 12 bit signal.  After reducing the resolution, the foot locations shift. We see that the reduction resulted in a quantization error on the actual foot of the PPG wave, with the peak detection identifying a neighboring location instead. The amount of shift due to this effect is not predictable. With a sensor separation of 4cm, a change in PTT of a few ms can cause the PWV to change dramatically, so these small changes observed in the two signals account for the large discrepancy in PWV calculations.

## V. CONCLUSION

In this work we have described an AFE to be used in the design of a wearable continuous blood pressure monitoring system. We showed the feasibility of measuring PWV with sensors at close distances on the arm with the use of PPG sensors which are more convenient for the user. We verified the results with *in-vivo* experiments and also provided insight into the minimum ADC requirements for measuring PWV at this distance. The focus of this work was on the validation of ADC requirements, but in the future we will look into optimizing the power consumption as well as validation of BP measurement.

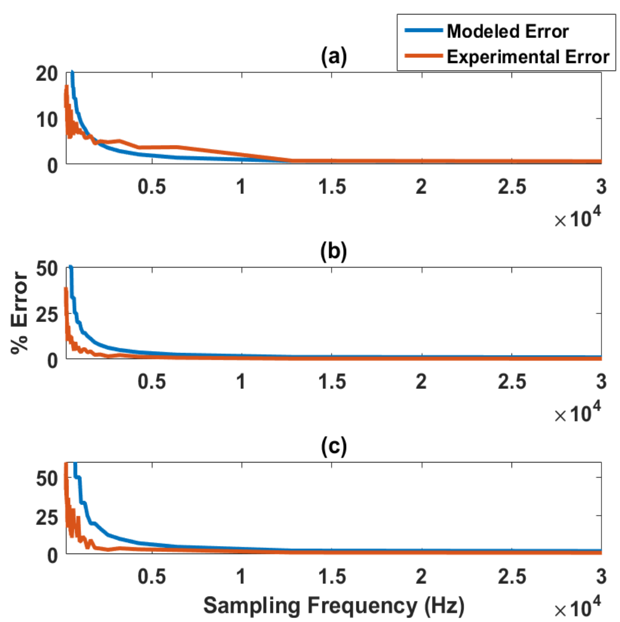


Fig. 3. Comparison of modeled and experimental error for mean PWV during (a) Mueller, (b) resting, and (c) Valsalva maneuver.

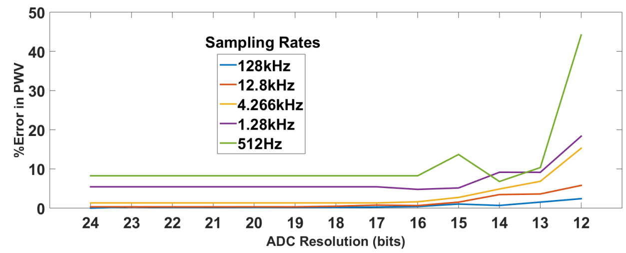


Fig. 4. Effect of ADC resolution on error of PWV measurement for various sampling rates

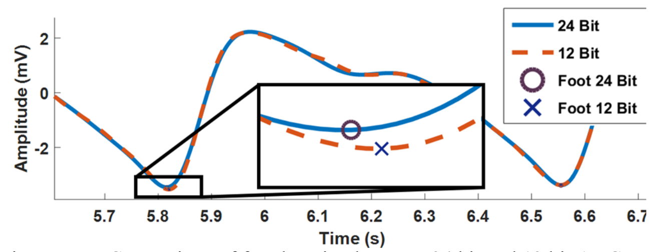


Fig. 5. Comparison of foot location between 24 bit and 12 bit ADC resolution and 512Hz sampling frequency.

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