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SeismoWatch: Wearable Cuffless Blood Pressure Monitoring Using Pulse Transit Time

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Abstract

The current norm for measuring blood pressure (BP) at home is using an automated BP cuff based on oscillometry. Despite providing a viable and familiar method of tracking BP at home, oscillometric devices can be both cumbersome and inaccurate with the inconvenience of the hardware typically limiting measurements to once or twice per day. To address these limitations, a wrist-watch BP monitor was developed to measure BP through a simple maneuver: holding the watch against the sternum to detect micro-vibrations of the chest wall associated with the heartbeat. As a pulse wave propagates from the heart to the wrist, an accelerometer and optical sensor on the watch measure the travel time – pulse transit time (PTT) – to estimate BP. In this paper, we conducted a study to test the accuracy and repeatability of our device. After calibration, the diastolic pressure estimations reached a root-mean-square error of 2.9 mmHg. The watch-based system significantly outperformed ($p < 0.05$) conventional pulse arrival time (PAT) based wearable blood pressure estimations – the most commonly used method for wearable BP sensing in the existing literature and commercial devices. Our device can be a convenient means for wearable BP monitoring outside of clinical settings in both health-conscious and hypertensive populations.¹

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Keywords

Blood pressure monitoring; Pulse transit time; Wearable sensing

CCS Concepts

Applied computing~Consumer health; *Human-centered computing~Mobile devices*

1 INTRODUCTION

Serial blood pressure (BP) measurements at home improve the control of hypertension (defined as BP > 140/90) and the effectiveness of treatment compared to infrequent in-clinic measurements alone [1]. As hypertension is a major cardiovascular risk factor associated with an increase in complications and mortality rates regardless of age group [31], and nearly one in three individuals in the United States have hypertension [3], improving the effectiveness of treatment is a major health care priority. Current methods for BP monitoring at home rely on the use of oscillometry-based BP cuffs. While such automated cuffs enable out-of-clinic monitoring without the need for a specialist, they are cumbersome, inconvenient for the patient, and known to be inaccurate [32, 37]. Additionally, cuff-based measurements can typically be obtained only once per day, and thus do not capture BP changes that can occur throughout normal activities of daily living, nor in response to exercise or other stressors.

Novel technologies for wearable BP monitoring could potentially enable convenient and unobtrusive BP monitoring at home, and frequently throughout the day. Such technologies would also potentially improve hypertension awareness, as a larger segment of the general population could potentially use consumer devices to measure BP. To successfully supplant, or complement, existing BP cuffs, such wearable approaches must be (1) more convenient than the cuff in their use, and (2) at least as accurate as the cuff.

1.1 Blood Pressure Monitoring Watch

The work in this paper explores a novel method of convenient, non-invasive BP measurement without the need for a cuff. The method comprises a watch-based system, similar to that of fitness monitoring wearable devices currently in market (Fitbit, Apple Watch, etc.). Our method utilizes the theory of pulse transit time (PTT) to extract BP. PTT is simply the time interval for a pulse wave to travel from a proximal (close to the heart) to a distal (further away from the heart, such as at the wrist or leg) location along the arterial tree and is inversely correlated to BP [23]. To obtain a BP reading with this system, the user simply places the face of the watch onto his sternum for a short period of time (< 15 seconds) as seen in Fig. 1. An accelerometer inside the watch measures thoracic vibrations associated with the heartbeat, or the seismocardiogram (SCG) signal, to obtain the proximal timing indicating the ejection of blood from the left ventricle into the aorta [12]. Optical sensors on the watch (Fig. 2) measure the blood volume pulse, or photoplethysmogram (PPG) signal, to obtain the distal timing associated with the arrival of the pulse wave at the radial artery (on the wrist). The time delay calculated between fiducial points on these two

measurements then provides the PTT, and is then used with subject-specific parameters (obtained using an initial calibration step) to calculate an estimate for BP. Though the required user input is not conducive for continuous measurement, the device can provide episodic BP estimation as an unobtrusive alternative to conventional BP cuffs.

1.2 Contributions

This work presents the following contributions in ubiquitous health monitoring:

- The design of a watch-based prototype to record both SCG and PPG signals to determine the PTT between a proximal and distal location and enable a BP estimation. A key innovation in this work is the use of pressing the wrist against the sternum to measure a *true* proximal reference using the SCG signal as compared to an electrical *surrogate* proximal reference, such as from an electrocardiogram (ECG) signal, as used in conventional devices.
- Evaluation of the watch-based prototype to demonstrate accurate BP measurements. A controlled study and analysis of the prototype to demonstrate acceptable accuracy was performed. After calibration, the system could predict BP within 5 mmHg of ground truth (commercially available reference standard BP measurement system) in more than 95% of cases.
- Demonstration of day-to-day repeatability in the calibration coefficients linking watch-based PTT measurements to BP for a given subject. A different day follow-up study to test the repeatability of the device during both rest and physiological stress (i.e., cold pressor test) was conducted. The PTT method was compared to a pulse arrival time (PAT) based method commonly used in the existing literature, resulting in an average improvement of 12.3 mmHg during rest for mean arterial pressure (MAP) estimation and 13.9 mmHg during physiological stress (cold pressor) for MAP estimation.
- Setting the groundwork for creating a method that is capable of conveniently and accurately measuring BP throughout the day. This method could potentially result in a stand-alone device or be incorporated into a current smartwatch system.

2 RELATED WORK

2.1 BP Monitors Used in Clinical Settings

The gold standard for BP measurements in the hospital is intra-atrial catheterization; however, this method is invasive, requiring a strain gauge to be in fluid contact with blood at an arterial site [23]. The standard noninvasive method used in the clinic is the auscultation-based BP cuff [18, 23]: the brachial artery is occluded with a cuff placed around the upper arm inflated above the systolic BP, and Korotkoff sounds are then detected using a stethoscope and manometer during cuff deflation [18]. Both of these approaches are used in hospital/clinical settings only, requiring medical personnel to administer the test, and are not suitable for monitoring at home.

2.2 BP Monitors Used At Home

The most common approach to measure BP without clinical supervision is the automated BP cuff employing oscillometry. A cuff is wrapped around the upper arm and inflated above systolic pressure, then deflated to a pressure below diastolic BP. A pressure sensor inserted in the cuff records the arterial pulsations during the cuff deflation, and the amplitudes of these pulsations are used to calculate systolic and diastolic BP. This approach has many limitations, the primary one being that it relies on a set of empirical coefficients to map the pulse amplitudes to systolic and diastolic BP. Most of these coefficients are specific to the device and can vary across devices. Further, since it relies solely on pulse amplitudes, it fails to provide accurate readings in patients with atherosclerosis or obese patients whose pulse amplitudes can be weak [8]. Cuff-based approaches also have the pitfall that by occluding the vessel, BP is inherently altered, thus possibly corrupting repeated measurements.

Upper arm BP monitoring also has the disadvantage of requiring a measurement site that can be difficult to reach for elderly and handicapped. Wrist BP cuffs, such as the Omron R3 (Advance AG, Switzerland), have increased user-friendliness, allowing for a smaller device to be placed easily on the wrist. While the ease of measurement improved, these devices only monitor the pressure at the radial artery and factors such as hydrostatic pressure can affect the readings [41]. In clinical evaluation studies, these devices have significantly lower measurement accuracy when compared with upper BP monitoring devices [15].

Tonometry is a less-widely used non-invasive approach to blood pressure detection [23, 39]. The tonometer has a pressure sensor at the tip of a rod which is pressed perpendicularly on an artery to sense the deflections of the arterial wall with each heartbeat. The quality of the measured deflections depends heavily on the operator using a constant level of force and minimizing vibrations. In practice, since the artery is not perfectly applanated, the acquired waveform is uncalibrated, and whenever a major change in BP is anticipated the waveform needs to be calibrated with cuff BP [10]. To facilitate continuous monitoring via tonometry, Vardoulis et al. proposed a wrist-based wearable device that acquires pressure signals from the radial artery [39]. However, the device currently has only been shown to give a measure of arterial stiffness and not BP. Further, this device suffers from the same limitations of tonometry in that the waveforms will likely require calibration and that exerted pressure on the arteries can influence the recorded BP readings.

2.3 Wearable BP Measurement: Pulse Arrival Time Based Methods

Given the limitations of methods requiring external pressure to be applied to the arteries to determine BP, techniques based on passive timing interval measurements have become popular. The most commonly applied approach is based on the measurement of pulse arrival time (PAT) – the time delay from the ECG R-wave to the pulse arrival at a distal location such as the wrist/finger – and the subsequent correlation of PAT to BP [4, 5]. PAT can be measured conveniently, such as with a watch that has an external electrode the user can touch with the opposite hand to measure an ECG and reflectance PPG waveform simultaneously. Thus, the approach has been pursued widely. However, while PAT is convenient, it is not an accurate means of obtaining BP estimates for an important

underlying physiological factor which is often overlooked: the ECG provides only a *surrogate* proximal timing reference, not a *true* one.

The principle behind using PAT for estimating BP is that BP impacts the stiffness of the arteries, and thus the time it takes the pressure pulse to travel along the arterial wall – pulse transit time (PTT). The stiffer the arteries, the faster the pulse will travel. This phenomenon has been known for nearly one hundred years, and is deeply-rooted in the theoretical and practical understanding of cardiovascular biomechanics [24]. To measure this propagation time of the pulse wave along the arterial wall (i.e., PTT), one would need to measure the timings of the propagation at two points along the wall – ideally one close to the heart (such as at the aortic output from the left ventricle) and another far from the heart (such as at the femoral artery or dorsal artery). The latter is readily achieved with current technologies, for example with reflectance PPG sensing at the finger, wrist, or foot of the subject. However, the former – the measurement of the proximal timing, or when the pulse first begins to travel along the aortic wall – is quite challenging. Researchers pursuing PAT-based approaches have chosen to use the ECG R-wave as a surrogate for this proximal timing, arguing that the ECG represents the electrical depolarization of the heart, and thus marks the start of the heart's contraction which later ensues in the ejection of blood into the aorta. Herein lies the important detail that is often overlooked yet is the main reason behind the inaccuracy in PAT-based BP estimation methods.

There is a non-zero, variable, and non-negligible time delay between the ECG R-wave and the ejection of blood from the heart into the aorta – the timing is actually defined as the pre-ejection period (PEP), and has been studied extensively in the physiology literature [19]. In some scenarios, PEP can increase with increased BP, while in others it can actually decrease with increased BP. Thus, since PAT is actually a sum of PEP and PTT, its relationship to BP will depend on the way that PEP changes with respect to BP - a factor that is highly variable and can change substantially due to stress, neuro-hormonal factors, muscle contractions, and vasomotor tone. Prior work has shown that PTT more strongly correlates to BP than PAT and that PAT-based BP measurement is unable to meet FDA BP error limits [44]. Prior works have developed watch-based PAT devices. The user touches the back of the watch with the opposing hand to obtain an ECG while LEDs are mounted to monitor wrist PPG [17, 38]. While this is a simple and convenient maneuver, the resulting BP measurement would be inaccurate in certain vasoactive interventions such as vasoconstriction [45].

3 PROPOSED SYSTEM

3.1 Pulse Transit Time Background

The proposed device will facilitate wearable measurement of PTT using a novel approach, and thus will result in improved BP estimation accuracy compared to conventional techniques. PTT has become a well-established physiological phenomenon for estimating BP. According to the Moens-Kortweg equation, as arterial stiffness increases, so does the speed with which a pulse wave travels along the arterial wall due to the mechanical properties of the wall. Furthermore, BP has a close relationship to arterial stiffness, allowing for an inverse correlation between PTT and BP to be given by the following equation and shown in Fig. 3.

$$BP = \frac{K_1}{PTT} + K_2 \quad (1)$$

where K_1 and K_2 are subject-specific parameters.

Conventionally, two arterial sites are directly measured to represent the proximal and distal locations for PTT measurement. Common measuring techniques for the proximal timing reference include ballistocardiogram (BCG) [16], impedance cardiogram (ICG) [29, 36], and PPG at the carotid [6]. BCG can be an unobtrusive measurement, but cannot readily be implemented in a wearable form factor such that the waves of the signal can be interpreted to locate aortic valve opening. ICG can be a true measurement of aortic valve opening, but is not convenient enough to be practical as a wearable device, and has low signal quality in subjects with low cardiac output and obese subjects [28, 34]. PPG has some potential in providing a proximal timing reference if placed on the carotid or ear, but needs direct skin contact at those sites and would likely require two separate wearable devices to be worn and synced in time (i.e., one at the carotid, and a second at the wrist or foot). For a distal timing measurement, PPG is commonly used [21, 25, 35], since it is convenient and provides high quality recordings from the fingers, wrist, or foot.

Our prototype system will focus on obtaining SCG and PPG measurements (Fig. 4) to detect both *true* proximal and distal timing references, respectively. While this theory has been explored in previous work [40], PTT via SCG and PPG was only performed in a clinic while the subject was supine. The proposed system will be able to obtain, for the first time, both references from the same convenient wearable device that can be used in an at-home setting. As previously mentioned, SCG represents the low frequency thoracic vibration induced by mechanical motions of the heart and blood. It should not be confused with the phonocardiogram (PCG), the measurement of higher frequency acoustic components associated with the *closing* of the mitral and the aortic valve. In the SCG, the waveform includes waves associated with the *opening* of the aortic valve, and thus the theoretically ideal point in time for a proximal timing reference for PTT. With regards to the PPG sensor, larger arteries will be targeted to minimize the effects of smooth muscle contractions (SMC) and obtain a reliable distal timing for the BP estimation. The proposed system will capture the opening of the aortic valve (t_{AO}) and the arrival of the distal pulse wave (t_{PPG}) to determine PTT using:

$$PTT = t_{PPG} - t_{AO} \quad (2)$$

3.2 Hardware Design For Watch Prototype

A prototype watch was created to resemble the size and form factor of commercially available smartwatches. The watch includes electronics already commonly integrated: an accelerometer, photodiodes, and light emitting diodes (LEDs). For the prototype, the ADXL354 accelerometer (Analog Devices, Norwood, MA) was chosen to measure the SCG, due to its very low output noise (20 $\mu\text{g}/\text{Hz}$). The wrist PPG consisted of three pairs of

infrared (IR) LEDs (VSMF2893GX01, Vishay Semiconductors, Malvern, PA) and photodiodes (S2386-18K, Hamamatsu Photonics, Shizuoka, Japan). Though most heart rate monitoring smartwatches use green LEDs for maximizing the signal-to-noise ratio, the IR LEDs allow for deeper penetration into the skin and a more accurate capture of the arterial pulse wave from larger arteries deeper within the tissue than the cutaneous circulation illuminated by green wavelengths. A custom watch face was designed to house both the SCG and PPG sensor. The photodiode and LEDs were placed in the inside face of the watch to allow direct contact with the wrist and to sense the pulsations of the radial artery. Three sets of photodiode and LED combinations were used to adequately cover the wrist and to capture the pulsation of either the radial or ulnar.

The cathode of the photodiode was biased at 5V, and the anode was connected to a transimpedance amplifier configured to act as a first-order lowpass filter ($f_c = 8\text{Hz}$, $G = 110\text{ dB}$) followed by a bandpass filter ($\text{BW} = 0.7\text{--}8\text{Hz}$, $G = 59\text{dB}$). The signal was then lowpass filtered using a second-order Butterworth filter in a Sallen-Key topology ($f_c = 20\text{Hz}$, $G = 20\text{dB}$). The LEDs' intensities were modulated by a potentiometer in series with the LEDs. The intensities were altered to account for differing melanin levels or depth within the skin of the arteries that could deteriorate the PPG signal.

3.3 Data Collection

Multiple studies were conducted to test the robustness and accuracy of the prototype. All studies were approved by the Georgia Institute of Technology Institutional Review Board. In total, thirteen young and healthy subjects with no history of cardiovascular conditions were recruited to partake in all the studies (age: 23 ± 3 ; gender: 8 males; weight: $68 \pm 16\text{ kg}$; height: $173 \pm 10\text{ cm}$). Written informed consent was obtained from each subject.

To acquire a timing reference for the start of a cardiac cycle, a wireless ECG module (BN-EL50, Biopac Systems, Goleta, CA, USA) in a 3-lead configuration was attached to the subject with Ag-AgCL gel electrodes. A finger-cuff BP sensor based on the volume-clamp method [27, 42] was placed on the hand contralateral to the watch, acquiring continuous beat-by-beat BP (ccNexfin, Edwards Lifesciences, Irvine, CA).

All the sensors were sampled at 2 kHz and interfaced to a computer by a MP150 Data Acquisition system and its corresponding software, Acqknowledge (Biopac Systems). The files were saved to a desktop computer for post-processing.

3.4 Study Protocols

Validation of SeismoWatch as a blood pressure monitor was divided into three studies. The first study was performed to vary the subject's BP to allow a calibration curve to be fit between PTT^{-1} and BP. Subjects returned a following day to test the repeatability of the device. During the final study, varying conditions were imposed to quantify the effects of possible errors in usage on performance. During all studies, subjects were asked to remove coats or other bulky clothing.

The first study was intended to establish proof-of-concept and a correlation between PTT^{-1} and BP and included all 13 subjects. The protocol was divided into three sections: one-

minute rest, one-minute exercise, and five minutes recovery. During the rest and recovery periods, the subject was positioned in a chair with the ccNexfin placed on the left index finger and the watch donned around the right wrist. During the exercise session, the watch was removed to avoid damage, and the subject performed a stair stepping exercise. The purpose of performing exercise is to increase the heart's cardiac output and BP, thus increasing MAP and varying PTT. Since exercise can raise a person's systolic BP by up to 80 mmHg [22], a substantial range will be available for accurate correlation between cuff and estimated BP. Each subject was asked to place the device against his sternum for 20 seconds to enable the measurement of SCG and PPG signals and, thereby, PTT for the baseline (pre-exercise) and recovery (post-exercise) periods.

Eight of the subjects returned for a follow-up study to test the multi-day repeatability of the PTT to BP correlation parameters. During this test, each subject was asked to remain seated with the same sensor configuration as for the previous study. Upon sitting down, the subject was asked to place the face of the watch to his sternum for 20 seconds. The procedure attempted to closely replicate the maneuver that would be performed at home with this device. The subject was then asked to place his feet into a bucket of cold water for 60 seconds to obtain a cold pressor reading. The cold water was chosen since it induces local vasoconstriction and thereby increases BP [46]. After the first 30 seconds of immersing the feet, the subject was asked to perform the same maneuver (placing the watch against his sternum) to obtain the SCG and PPG measurements.

Five of the thirteen subjects were selected to test the robustness of the SCG signal for ultimate use in an at-home, unsupervised, setting. Varying contact pressures of the device against the sternum and sensor locations were deemed as possible user errors and were chosen as the independent variables for the studies. To test the effects of contact pressure variations on the measured SCG, the subject was asked to vary the applied pressure three times, ranging from a light touch to a hard press. A pressure sensor with continuous feedback was placed between the watch and the subject's wrist ensuring a constant pressure was achieved during the study for each of the three conditions. To test the effects of placement error, the subject transitioned the watch to varying locations on the chest while maintaining a constant pressure. These included locations that are in contact with light clothing and in direct skin contact.

3.5 Signal Processing

The accelerometer, PPG and ECG signals were post-processed to calculate PTT. Note that the ECG signal was selected for its high signal quality and simply used in this study as a fiduciary point to validate the other sensor measurements and would not be required in the actual deployment of the device for home use.

One of the technical challenges in using the SCG signal is the variability in morphology between subjects and its susceptibility to motion artifacts and respiration-induced waveform distortion. A traditional SCG signal contains peaks representing the mitral-valve closure (MC), isovolumic moment (IM), and AO. Classical models of the SCG denote the AO as simply the highest peak immediately following the R-wave of the ECG [33]. However, by

this simple method, the MC and AO can sometimes be confused with each other even in healthy subjects [14].

Fig 5. depicts the method of locating AO in a SCG waveform that we employed. To determine the correct AO peak, the SCG and PCG signal – both measured from the accelerometer – were extracted. A band of 0.8Hz to 30 Hz was used for the SCG signal, while a higher frequency bandpass filter (30Hz – 150Hz) was applied to extract the PCG. Note that the PCG is identical to the signal one would obtain using a digital stethoscope, as the accelerometer on the chest simply acts as a contact microphone for high-quality measurement of heart sounds. Spline interpolation of the PCG peaks was applied to yield the amplitude envelope for the signal. Using the timing of the R-wave of the ECG, the SCG and enveloped PCG signals were partitioned, and thirty beats were averaged together into individual beats of size N , where N is the minimum R-to-R interval to minimize any overlap between beats. The first averaged beats on the SCG and PCG were used to find the AO point. Since the PCG is commonly used to identify the closing of the mitral valve, the MC peak was identified as the first maximum preceding the maximum PCG envelope point [14]. Due to the shortening of the PEP following exercise, the AO point can be assumed to occur directly following the MC. Using a simple peak detection algorithm, all the peaks were identified in this window. Peaks were removed if a window around the peak was deemed too noisy using a noise-to-signal ratio (NSR) detection algorithm described in [11], removing any false peaks that did not correspond to cardiac events. The AO peak and those occurring afterwards formed a template for future AO detections. The same NSR detection algorithm was applied to the future SCG beats to determine possible SCG peaks. The template was applied to the later peaks to determine the AO point.

The PPG signal was first filtered using a digital bandpass filter (0.8Hz – 15Hz). For each signal, ensemble averages of thirty beats were formed in the same manner as the SCG and enveloped PCG beats. These averaged waveforms were used to extract the distal timing from the PPG. Specifically, the timing of the “foot” represented the arrival of the pulse. To determine the foot timing, a horizontal projection passing through the signal’s minimum and a line tangent to the maximum local gradient was placed on the signal. The intersection of these two lines represented the distal timing reference as commonly used in previous work [9]. This position was seen to provide a better marker of distal timing toward accurate BP estimation than such timing references as the minimum, maximum, or maximum second derivative [16].

For each averaged cardiac cycle, the location of the foot of the PPG was subtracted from the AO timing to determine PTT. Using linear regression, a PTT^{-1} to BP calibration curve was formulated for the subject. This curve was used for the follow-up study when estimating BP. PAT was calculated by determining the time delay between the R-wave of the ECG and the foot of the PPG.

4 RESULTS

4.1 Assessing Correlations Between Wearable PTT and BP

Fig. 6 illustrates the correlation plot and Bland-Altman plots after a best-case calibration of BP based on PTT for MAP, diastolic pressure (DP), and systolic pressure (SP). The root mean square of the difference between the estimated BP and the measured BP was calculated to obtain the root-mean-square-error (RMSE) in all three BP values. The group RMSE was 3.4 mmHg, 2.9 mmHg, and 4.8 mmHg for MAP, DP, and SP respectively. Diastolic estimation had the best confidence interval (95%) at 5.8 mmHg. Systolic estimation proved to be the least accurate, with the highest RMSE and a confidence interval at 9.7 mmHg. This is consistent with physiological expectations since the foot of the PPG waveform representing distal pulse arrival occurs during diastole rather than systole.

Table 1 summarizes the individual results of the thirteen subjects when PTT was converted to DP with calibration curves. DP represented the lowest error and thus was chosen for further evaluations. μ_{DP} is the average diastolic pressure, RMSE is the root mean square error, R is the correlation coefficient, $e < 5$ mmHg is the ratio between estimations that are less than 5 mmHg from measured BP and the total number of estimations and $e < 10$ mmHg is the ratio between estimations that are less than 10 mmHg from measured BP and the total number of estimations. Individual RMSE was less than 5 mmHg for twelve of the thirteen subjects. BP estimation errors were below 5 mmHg in eight subjects. Only one subject had BP estimation error exceeding 10 mmHg.

4.2 Quantifying Day-To-Day Repeatability in Calibration Coefficients and BP Estimation

Fig. 7 depicts the follow-up study that tested the day-to-day repeatability of the watch with MAP, DP, and SP calibration for both PAT and PTT calibration. During rest, PTT estimation significantly improved both MAP and DP estimations when compared to PAT estimation ($p < 0.02$, $p < 0.005$). During rest, BP estimation improved by an average of 12.3 mmHg when using PTT based calibration over PAT based calibration. All eight of the subjects had estimated values what were within 5 mmHg of their measured BP levels. While the best-case BP using PAT produced less than 1.0 mmHg error, three of the eight estimations had errors of more than 15 mmHg, which renders the measurement not sufficiently accurate for clinical use. DP had a similar comparison. BP estimation improved by an average of 13.9 mmHg with only one subject with greater than 5 mmHg error ($e = 5.7$ mmHg). In contrast, SP estimation was poor. Though PTT calibration produced an error that was, on average, 12.3 mmHg lower than PAT estimation, there was no significant difference between the two estimations ($p > 0.05$).

During the cold pressor challenge, PTT estimation significantly improved both MAP and DP estimations when compared to PAT estimation ($p < 0.01$, $p < 0.005$). However, PTT based BP measurements worsened when compared to rest values, averaging 6.2 ± 4.1 mmHg error for MAP estimation and 5.0 ± 5.0 mmHg error for DP estimations. Though there was a decline in performance during this perturbation, BP estimation error was still substantially lower than estimations made by using the PAT method which resulted in an average error of 20.1 mmHg for MAP and 18.2 for DP. At the resting phase, the SP estimations were not

within an acceptable error. Estimations yielded an average of 13.3 mmHg for PTT based estimation and 27.4 mmHg for PAT based estimation.

4.3 Determining Variability in SCG Signal Quality For Unsupervised Settings

Varying the watch in different locations and positions tested the robustness and reliability of the watch in at-home scenarios when trained personnel are not available to supervise. The three contact pressures tested resulted in a similar SCG signal with the AO peak only deviating an average of 3.2 ms, well within normal variability of PEP. Qualitatively, the signals had similar morphology regardless of contact pressure, exhibiting comparable numbers, and relative amplitudes, of peaks and troughs.

However, varying locations on the chest seemed to influence the signal shape and timings. When the contact area was located at the bottom half of the sternum, between the sternal angle and the xiphoid process, the signal was consistent and reliable. Positions above this section distorted the signal, and the AO peak deviated by as much as 85%. Furthermore, locations away from the sternum, such as on the softer tissue of the pectoral muscles, resulted in a misclassification of the AO peak. To obtain a reliable SCG signal, the subject will need to place the watch face on the lower half of the sternum, which should be feasible even in unsupervised settings. The presence of light clothing seemed to have little effect on the quality of the signal. On the same location, direct skin contact and contact with a clothing layer showed similar SCG signal.

5 DISCUSSION

5.1 Reliability of the Watch as a BP Monitor

The study presents a device that shows promise in being able to conveniently obtain BP measurements outside of clinical settings. The results show a strong correlation between BP and PTT obtained from the timing reference derived from an SCG and PPG for MAP and DP. SP did not follow as strong of a trend; however, this is consistent with previous studies [21, 26].

The watch showed significant increase in accuracy when compared with estimations using the PAT based method as in conventional approaches. The addition of an accelerometer to measure SCG as the proximal timing reference for PTT proved to be superior to using the ECG as the proximal reference for PAT during rest. While ECG has the advantage of a higher signal-to-noise ratio, the confounding effect of PEP negatively impacts MAP and DP estimation. In our study, exercise during the calibration phase had a large effect in shortening the PEP. During post-exercise recovery, BP decreased substantially while PEP only slightly increased. This had a negative impact on estimations during the follow-up study where PEP typically was much higher than during post-exercise while BP did not substantially change. Therefore, PAT values were lengthened, and underestimated BP. In conditions that vary BP but have little to no effect to PEP, BP estimation will be poor with PAT, and such conditions are common during normal daily living activities.

Discrepancies were found between the rest and cold pressor results. The difference can be attributed to the presence of smooth muscle in the brachial artery. External stimuli such as

cold pressor can modulate vasomotor tone independent of blood pressure, changing the arterial stiffness and affecting PTT. Though this may induce error, current standards for blood pressure monitoring recommend subjects to sit in a room temperature room and remain still for a period. Abiding by these recommendations can minimize the impact of the vasomotor tone based changes in PTT.

The average power consumption of the device is 360 mW. While this would be difficult to sustain for long periods of time using an average smart watch battery, only a few seconds of measurements are needed to obtain a reading. During the follow up study, only ten beats were required, averaging the recording time to 8.7 seconds. The device can be gated to minimize power consumption. The user can either by preform a certain gesture or by a manual input to start the measurements. If longer periods of time are required, power consumption could be substantially improved by lowering the output intensity of the LEDs with a varying current amplitude or duty cycle or by selecting only one of the LEDs and turning off the other two. These methods will insure that only the minimal amount of power would be used to measure the pulse signal.

5.2 Advantages Over Blood Pressure Cuffs

The form factor and measurement procedure were more convenient than a traditional BP cuff. During the study, there were only basic instructions provided on how to operate the watch. Even with these simple instructions, all of the subjects could easily and comfortably place the watch in the correct location and obtain a clean SCG signal. BP cuffs, on the other hand, are susceptible to positioning, cuff size, obesity and other user errors [30].

The subjects could perform the maneuver for five minutes while taking multiple readings (>10) without loss of quality in either the SCG or PPG. This would be difficult to achieve with the BP cuff since it is recommended to have one-minute in between successful measurements [30]. Thus, transient changes in BP in response to stressors such as exercise or mental stress can more readily be quantified with our approach.

One limitation of current BP measurement devices is the strict guidelines imposed to obtain a consistent reading. Typically, these devices, such as a wrist mounted BP cuff, require either a pressure differential sensor between the device and the heart level or the device to be place at a set distance from the heart level. The effect of hydrostatic pressure can alter BP as much as 2 mmHg per inch [30]. With the watch, the user is required, by definition, to place the device at or near the heart level, allowing for consistent hydrostatic pressures between readings.

6 Limitations and Future Design

The primary limitation of the proposed device is that a cuff-based BP measurement is required for calibration during the first use and updated periodically to account for slow changes (over the course of months or years) in arterial stiffness. However as seen in prior work, current commercially available BP monitoring devices require cuff-based calibration as well [47]. A potential solution would be to leverage posture-induced changes in hydrostatic pressure to calibrate the watch. The subjects would be required to adjust their

arms to a specific height above and below heart, varying the average BP and thus PTT to allow calibration. This method would first require an ECG and SCG to estimate PEP which would be assumed to remain constant during the remaining maneuvers. The ECG and PPG would measure the PAT at the different levels, in which the PEP would be removed. A BP cuff reading would still be necessary for the initial calibration.

An obvious limitation of this study is that only young and healthy subjects were studied. Nevertheless, younger subjects are an important sub-population for early detection of hypertension [21]. Another limitation is that the wrist based approach measures PTT through the brachial artery, which has a higher variability of elasticity independent of BP than larger arteries such as the aorta. While constraining the pulse through these larger arteries can improve tracking of BP, it would have been difficult to reliably measure both a distal and proximal pulse conveniently. Future work can investigate a solution for measurement of aortic PTT with a wearable system.

Another drawback is that the user is constrained to perform the maneuver while static. Direct contact to the sternum by the watch is required due to the SCG's low signal and the poor coupling to the rest of the body. With the mechanical nature of the signal, motion artifacts can corrupt the signal and lead to poor AO and PPG foot detection. These factors prevent the user from continuously monitoring PTT during sports or other high motion activities. However, static measurements are already necessary in current cuff-based BP monitors.

Future work on the proposed device will include embedding the electronics into a conventional smartwatch, making the technology less obtrusive and usable at home. Currently, the watch is externally tethered by a wire. The later version will transmit signals via Bluetooth to a smartphone which would perform the post-processing and record the measurements. The current prototype relies on the timing of an ECG to partition the cardiac cycles. This measurement will need to be removed for seamless integration into current smartwatches. Extracting the individual beats is a feasible engineering challenge that could be achieved either through automated detection of the aortic opening peaks using a supervised learning model [43] or through peak detection of the PPG signal. The current prototype requires manual selection of one of the three PPG sensors. Automatic detection of optimal signal quality is required for conventional use. Fortunately, much research has been done in evaluating signal quality for such waveforms [7, 13, 20].

Measurement reliability will also need to be addressed in future works. If not placed in a proper location on the sternum, signal quality will deteriorate, leading to a potential decline in accuracy. Potential solutions could include using machine learning algorithms to automatically assess whether SeismoWatch was positioned properly or not on the body and provide the appropriate user feedback for improved signal capture [2].

7 Conclusions

Due to its associations with cardiovascular diseases, management of hypertension can greatly decrease hospitalization costs and increase the standard of living for millions of

people. A reliable and unobtrusive device is needed to increase measurement frequency and quality compared to current monitors. This paper explores the feasibility of using a watch-based wearable device to estimate BP based on PTT. A study was conducted to first establish a strong correlation between watch PTT and BP. A follow-up study tested the consistency of the device and compared the PTT method to a PAT method that is commonly used. The results showed a significant improvement while using SeismoWatch when compared to the PAT method. This work established a method of extracting BP that is both convenient and robust enough for at-home usage. The increase in convenience that this device provides may allow for more ubiquitous and frequent BP measurements, ultimately improving the management of patients with cardiovascular diseases and assisting in prevention of hypertension.

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Fig. 1.

a) Subject position to obtain seismocardiogram (SCG) and reflectance photoplethysmogram (PPG) recordings from the watch. The watch is worn with the buckle on the top of the wrist and the accelerometer pressed to the sternum. While the current prototype is hardwired, future implementations can readily incorporate Bluetooth or other wireless protocols to send the data to a mobile device for storage/processing. b) Photo of the watch positioned on the wrist.

Fig. 2.

a) Isometric view of the watch. b) Close-up of the reverse side of the watch showing positions of the photodiodes and infrared light emitting diodes (LEDs). A high-resolution three-axis accelerometer enables the accurate capture of chest wall vibrations associated with the heartbeat – the SCG signal.

Fig. 3. Plot showing the linear correlations with MAP and PTT^{-1} taken from a participant in our study.

Fig. 4.

a) Subject position for blood pressure measurement with the arteries and pulse wave illustrated. b) Plot showing ECG, SCG, and PPG signals measured with our prototype. PTT is computed by identifying the time delay between the aortic valve opening point in the SCG and the arrival of the distal pulse wave in the PPG.

Fig 5.

Block diagram of signal processing techniques to extract the AO peak using both the SCG and the PCG signal.

Fig. 6.

Correlation and Bland-Altman plots for mean arterial pressure (MAP), diastolic pressure (DP), and systolic pressure (SP).

Fig. 7.

Comparison between PTT and PAT during both rest and cold pressor. * indicates $p < .05$, ** indicates $p < .01$

Table 1

Subject Data

Subject	μ DP [mmHg]	RMSE [mmHg]	R	e < 5 mmHg	e < 10 mmHg
1	80±10	3.06	0.95	0.85	1.00
2	81±4	2.29	0.77	1.00	1.00
3	76±4	2.45	0.74	0.94	1.00
4	73±5	2.73	0.80	0.94	1.00
5	80±4	1.42	0.93	1.00	1.00
6	61±3	1.56	0.69	1.00	1.00
7	60±5	2.32	0.88	1.00	1.00
8	76±3	1.41	0.89	1.00	1.00
9	85±10	6.12	0.76	0.72	0.94
10	78±5	1.71	0.92	1.00	1.00
11	73±6	1.67	0.95	1.00	1.00
12	77±6	3.37	0.81	0.83	1.00
13	88±5	2.64	0.85	1.00	1.00
μ	76	2.52	0.84	0.95	1.00
σ	10	1.25	0.09	0.09	0.04