Published in final edited form as:

IEEE Trans Biomed Circuits Syst. 2017 August ; 11(4): 765–772. doi:10.1109/TBCAS.2017.2683801.

Robust Sensing of Distal Pulse Waveforms on a Modified Weighing Scale for Ubiquitous Pulse Transit Time Measurement

Andrew M. Carek and

School of Electrical and Computer Engineering, Georgia Institute of Technology, Atlanta, GA 30332 USA

Orner T. Inan [Senior Member, IEEE]

School of Electrical and Computer Engineering and, as an Adjunct, the Wallace H. Coulter Department of Biomedical Engineering, Georgia Institute of Technology, Atlanta, GA 30332 USA, phone: 404-385-1724

Abstract

The measurement of aortic pulse transit time (PTT), the time for the arterial pulse wave to travel from the carotid to the femoral artery, can provide valuable insight into cardiovascular health, specifically regarding arterial stiffness and blood pressure, BP. To measure aortic PTT, both proximal and distal arterial pulse timings are required. Recently, our group has demonstrated that the ballistocardiogram (BCG) signal measured on a modified weighing scale can provide an unobtrusive, yet accurate, means of obtaining a proximal timing reference; however, there are no convenient, reliable methods to extract the distal timing from a subject standing on the modified weighing scale. It is common to use a photoplethysmograph (PPG) attached to a toe to measure this distal pulse, but we discovered that this signal is greatly deteriorated as the subject stands on the scale. In this paper, we propose a novel method to measure the distal pulse using a custom reflective PPG array attached to the dorsum side of the foot (D-PPG). Twelve subjects of varying skin tones were recruited to assess the preliminary validation of this approach. Pulse measurements using the D-PPG were taken from seated and standing subjects, and the commercially available PPG were measured for facilitating comparison of timing measurements. We show that the D-PPG was the only sensor to retain the high detection rate of feasible timing values. To further test and optimize the system, various factors such as applied pressure, measurement location, and LED/photodiode configuration were tested.

Index Terms

Cardiovascular health; pulse transit time; home monitoring; photoplethysmogram

I. Introduction

AORTIC pulse wave velocity (PWV)—the velocity with which the aortic pressure wave travels along the arterial wall—has been demonstrated in large clinical studies to be

indicative of overall arterial stiffening [1, 2], and a prognostic cardiovascular risk factor independent of, and in some cases beyond, all other traditional risk factors [3, 4]. In 2015, the American Heart Association provided recommendations for standardizing vascular research on arterial stiffness, stating that it is “reasonable to measure arterial stiffness to provide incremental information beyond standard CVD risk factors in the prediction of future CVD events,” where CVD stands for cardiovascular disease, and providing guidance on measurement tools and technologies for PWV [5].

Currently, the measurement of aortic PWV is limited to clinical settings only. The most common approach for its measurement involves the use of an arterial tonometer, which is placed by a medical professional first at the carotid artery, then the femoral artery of the patient; this technique, referred to as carotid-femoral PWV is considered to be the “gold” standard measurement [6]. The time elapsed for the aortic pressure wave to travel from the carotid to the femoral artery—pulse transit time (PTT)—is measured from these two sites, and the aortic PWV is calculated as the ratio of the distance between the sites for the patient, d , to the computed PTT value: $PWV = d/PTT$. Thus, the measurement of PWV hinges on the reliable measurement of PTT.

In addition to providing the essence of PWV calculations, aortic PTT measurements can also be used to estimate a patient’s blood pressure (BP) without the need for an arm cuff, by leveraging the Bramwell-Hill and Moens-Korteweg equations relating arterial wall elasticity to PWV [7, 8]. Indeed, the relationship between PTT and BP has been documented in numerous studies [9, 10]. Recently, there have been many attempts to build wearable [11–13] and home monitoring [14] systems based on these principles to allow ubiquitous measurement of BP to improve hypertension detection [15] and control [16].

In theory, the correlation between PTT and BP should be least confounded by vasomotor tone, and thus require least frequent calibration, when PTT is measured through the aorta rather than a smaller artery (e.g., the brachial artery) [7]. However, since such measurements through the aorta require a true proximal timing reference (i.e., the timing at which the aortic valve opens) and a convenient distal timing reference, only a limited number of systems have successfully achieved convenient *aortic* PTT measurement. To this end, we demonstrated for the first time that proximal timing information can be accurately obtained from a distal location (the foot) using weighing-scale based measurements of the ballistocardiogram (BCG) [14], a signal representing the reaction forces of the body to cardiac ejection of blood [17].

While the BCG can be used to obtain proximal timing for PTT, the electrocardiogram (ECG) can be used as a proximal reference to obtain pulse arrival time (PAT). PAT refers to the timing between the ECG R wave and a distal location and is composed of the pre-ejection period (PEP) and PTT. PEP refers to the timing of an isometric contraction of the heart. Since this depends on the electrical activity of the heart and can move in the opposite direction of PTT, PEP has been shown to have a negative impact when using PAT to estimate BP [18]. Though this timing reference has been shown to have lower correlation to BP than PTT [19], the SNR is greater in ECG signals compared to BCG [20], and thus PAT has been shown to provide a strong correlation to systolic BP [21].

A commonly used signal for distal pulse timing detection is the photoplethysmogram (PPG), a signal representing the blood volume pulse [22]. For PPG measurements, a light source and a photo-detector are placed on the skin: while part of the emitted light is absorbed into the tissue, a portion of the light reaches the photo-detector [23]. Depending on the placement, the photo-detector either detects the reflected light off the tissue or the transmitted light through the tissue [24]. The DC portion of the signal is correlated to stagnant components such as tissue and bone. The AC signal—the signal of interest—represents the PPG waveform and provides a measure of changes in blood volume [25].

With BCG or ECG representing the proximal timing measurement, the PPG could be used to extract the remaining key to PPT/PAT based BP measurements, the distal timing. A simple fingertip PPG sensor could potentially be retrofitted to adapt to the toes (Figure 2b). However, we have discovered that such an approach does not yield a reliable PPG signal that can be used for PTT calculations in subjects that are standing. Motion artifacts and varying contact pressure due to postural sway distort the signal and reduce the accuracy by which a distal timing reference can be obtained (Figure 2c).

A previous study by Paliakaite, et al. utilizing toe PPG also noted that the quality of the measured signal was highly dependent on the subject's motion. These researchers used median filtering over many cardiac cycles to smooth out the motion artifacts corrupting PPG feature detection for PAT, thus suggesting a need for an improved sensor design to facilitate recordings over shorter time periods and/or beat-by-beat PAT extraction [26]. Another study conducted by Kato, et al. demonstrated an interesting approach based on a PPG array without the need for direct contact to the skin (i.e., through the sock worn by the user); however, these authors also noted that motion artifacts can degrade the signal quality of the measured PPG waveform in the standing position, and results were unfortunately only provided for one subject [27].

Thus, to advance the quality of PPG sensing from the foot toward scale-based PTT measurement, we developed a custom reflective PPG array to measure the distal pulse wave on the dorsum of the foot (D-PPG), as shown in Figure 2a. The objective of this study is to assess the reliability and validity by rigorously testing the device and comparing it to multiple existing commercial PPG sensors. We aim to demonstrate that this method of extracting the pulse can serve as an accurate and repeatable reference tool for distal timing. With this validation, a convenient weighing-scale based system for measurement of aortic PTT and PAT could thus comprise the proximal timing measurement using the BCG and handlebar ECG, together with a distal pulse measurement at the foot using D-PPG (see Figure 1).

Impedance plethysmography (IPG) is a common measurement technique for extracting distal blood volume pulse signals by measuring the fluctuation of impedance associated with the pulsation of arterial vessels in a given area of interest [28]. Integrating the IPG to detect heartbeats from the feet on a weighing scale has already been implemented and verified [28, 29]. However, these methods utilize the peak of the IPG waveform associated with systole, where wave reflections are more prominent than at the diastolic “foot” of the waveform [8], and thus PTT may not provide an accurate estimate of BP [7]. Further studies attempting to

use IPG as a distal timing reference have seen difficulties in determining the true point of measurement [26]. Finally, IPG can show poor waveform quality in obese individuals, heart failure patients, patients with aortic/mitral valve problems, or other patients with weak circulation [30–32]. However, in future studies, the IPG can be compared against the D-PPG to assess whether accurate distal timing information can be obtained with a more convenient interface for the user.

II. Materials and Methods

A. Hardware Design and Data Acquisition

At a high level, the hardware includes a D-PPG sensor to be used to obtain a pulse measurement on the foot. The toe sensors comprised four commercially available LED/Photodiode combinations. These sensors were connected to either a custom-designed analog front end or a commercially available transducer amplifier.

To obtain a timing reference for cardiac cycles, Lead I ECG signals were acquired using a wireless amplifier (BN-EL50, Biopac Systems, Goleta, CA, USA) and gel electrodes. The D-PPG sensor comprised a combination of red light emitting diodes (LEDs) and photodiodes (L1915-02 and S2386-18k, respectively, Hamamatsu Photonics, Shizuoka, Japan). Red LEDs were used to match the color emitted from most of the commercial PPG sensors, and thereby ensure a fair comparison. The anodes of the photodiode on the D-PPG sensors were directly connected to a single node to assist in removing zero-mean uncorrelated noise.

The commercial transmissive PPG cables used were the 8000AA and 8000SL (Nonin), and the DS100A-1 (Medtronic, Minneapolis, MN). The TSD200C (Biopac) was included in the test to represent a commercial reflective PPG that is designed to attach to the distal phalanx of the finger or toe. The commercial sensors can be seen in Figure 2b.

Excluding the PPG100C, all anodes of the photodiodes of the PPGs were connected to a transimpedance amplifier configured to act as a high pass filter ($f_c = 1\text{Hz}$, $G = 100\text{dB}$) followed by a gain stage ($G = 10\text{ dB}$). The relatively high cutoff may affect the amplitude and phase information for the PPG signal; however, motion artifacts that could distort the signal will be minimized. The signal was low pass filtered using a second-order Butterworth Sallen-Key filter ($f_c = 25\text{Hz}$, $G = 20\text{dB}$). Comparable transmit power was used for each of the commercial sensors to power the LEDs ($\sim 10\text{mW}$) with the D-PPG sensor having equal to or lower power.

The TSD200C interfaces to the PPG100C amplifier ($G = 100\text{dB}$, $\text{BW} = 0.01\text{Hz} - 10\text{Hz}$) The TSD200C consisted of a matched photodiode and an infrared (IR) sensor. Though PPG sensors of various waveforms have been shown to produce different SNR values [33], an IR sensor was deemed acceptable due to the study focusing on the *relative* changes in PAT detection rate value for standing versus sitting.

The contact pressure of the PPG sensor has been shown to affect the SNR value of the PPG signal [24]. To provide an estimate of the pressure, a force sensor was used to measure the

orthogonal force vector applied to the surface of the skin by the reflective PPG sensor (FS20-0000-1500-G, Measurement Specialties, Hampton, Virginia). The sensor's output had a 1V to 3V range corresponding to a linear range between 0 and 7.36N. The output pin was connected to a third-order Butterworth Sallen-Key low-pass filter ($f_c = 10\text{Hz}$) with unity gain.

The PPG, ECG, and force sensors were sampled at 2kHz and acquired by MPI50 DAQ and its corresponding software, Acqknowledge (Biopac), and stored on a desktop computer for further signal processing.

B. Subject Demographics and Testing Protocols

The study was approved by the Georgia Institute of Technology Institutional Review Board. Twelve young and healthy subjects (age: 25 ± 4 years; gender: seven male and five female; weight: $70 \pm 10\text{kg}$; height: $175 \pm 15\text{cm}$) were enrolled. Due to the levels of melanin influencing the PPG signal, the subjects were of varying race and ethnicity (seven Caucasian and five non-Caucasian) [34].

Three aspects of the D-PPG sensor were characterized for purposes of optimizing the design: (1) location on the dorsal surface, (2) applied pressure to the dorsum, and (3) LED/photodiode configuration. The last test (4) compared the D-PPG sensor to the commercially available sensors for seated and standing postures. For each of these tests, the subject was asked to sit in a chair and to remain seated until the cardiovascular system reached homeostasis. The following tests were then conducted.

1) Effects of Sensor Location—The dorsum of the foot was partitioned into seven regions chosen to map out the space of convenient surfaces that could be instrumented (Figure 5a), and the sensor was sequentially positioned on each of these seven surfaces. To ensure that contact pressure was held constant at each measurement region, the force sensor was placed on top of the D-PPG sensor. A strap was then placed around the foot to hold both sensors in place and to minimize the amount of external light that could reach the photodiodes. The strap was tightened to a constant pressure with the force sensor providing feedback. The subject remained seated for twenty seconds as the measurements were taken. At the end of each measurement, the PPG sensor was removed and placed in a new region, and this was repeated for the seven regions.

2) Effects of Contact Pressure—The reflective PPG sensor was placed on the dorsal side of the foot, near the most distal end of the first metatarsal, and the pressure sensor was placed on top of the PPG sensors. A strap was wrapped around the foot in a similar manner to the previous test. The strap was adjusted in order to vary the pressure through the entire range of the force sensor (0 mmHg to 195 mmHg) in intervals equal to one-eighth the max pressure. Note that, while this range may seem at first to be high – since typical aortic SBP values are less than 140 mmHg for healthy subjects, and contact pressures higher than this value would occlude the artery – BP at the foot in a standing subject is in fact much higher than brachial artery pressure as a result of hydrostatic pressure due to gravity. Specifically, an additive pressure of ρgh must be added, where ρ is the density of blood, g is acceleration due to gravity, and h is the height differential between the foot and the heart. Thus, SBP at

the foot can regularly exceed 200 mmHg in a healthy subject, and even the largest pressure applied of 195 mmHg was below the level required to occlude the arteries. Twenty-second measurements were taken of the PPG and ECG for each pressure interval.

3) Various LED/Photodiode configurations—Five different layouts were constructed with different combinations of LEDs and photodiodes (Figure 5c). Each configuration was placed on the subject at a constant pressure and recorded for twenty seconds.

4) Performance Comparison to Commercial PPG Sensors—Each commercial sensor was placed on either the first or second toe after visual inspection of the signal amplitude (to determine the best possible location). The D-PPG sensor in configuration I as shown in Figure 5c was placed near the larger toe on the dorsum and tightened to a constant pressure of 98 mmHg (which was found to be a comfortable level of tightness for all subjects). The subject was asked to remain seated for twenty seconds while the measurements were taken. Without moving either foot, the subject was asked to change position from seated to standing and to stand as naturally as possible. A brief rest period was taken to allow the subject to reach a state of cardiovascular homeostasis. A twenty second measurement was taken as the subject remained stationary. This process was repeated for each of the commercial sensors.

C. Signal Processing

The following model is proposed to quantify the quality of the timing information extracted from the individual beats of the PPG waveform. The approach is based on computing – on a beat-by-beat basis – the arrival time of the pulse, and determining if that timing is physiological possible and therefore reliable (as described below). Using the full recording, a detection ratio (DR), defined as follows, is calculated to give an approximation of the reliability of the sensor for the given subject:

$$DR = \frac{1}{M} \sum_{i=1}^M 1_{\{|PAT_{\mu} - PAT_i| \leq 13ms\}} \quad (1)$$

Where PAT_{μ} is the ensemble average PAT for all beats and M is the total number of beats.

The first step was to partition the signal into individual beats using the ECG. This signal was digitally band-pass filtered (Finite impulse response, FIR, $f_{pass} = 0.8 - 25Hz$, Kaiser window) to extract the QRS complex. An automatic beat detection algorithm was then used to detect the timing of the R waves, which were then confirmed manually.

A standard algorithm based on the existing literature was used to locate the foot of the PPG. Each test (i.e., recording) was partitioned into a matrix X with size $N \times M$. The test signals were separated into M beats with length N samples. The timing of the R-wave of the ECG noted the beginning of the cardiac cycle and as the timing of the first sample of each beat. N was set to be the minimum R-to-R interval to minimize any overlap between beats. Each beat was processed separately to extract the timing reference. The timing of the foot represented the arrival of the pulse. As shown in Figure 2a, the foot was defined as the

intersection of a horizontal projection passing through the minimum and a line tangent to the maximum local gradient [35]. This method has been shown to be the preferred choice due to its robustness to morphological artifacts and its effectiveness in extracting PTT [14, 19].

When measuring beat-by-beat fluctuations in PAT for a person at rest in short-term recording, the only change that would be expected would be associated with respiration. During inspiration, systolic blood pressure decreases, thus increasing PTT [36], and the isovolumetric contraction time [37]; therefore, there is an overall increase in PAT with inspiration. The converse is true for expiration. It has been shown that the variation in PAT due to respiration should be no more than ± 13 ms [38–41]. Thus, by computing the ensemble average PPG beats and determining the PAT for that beat (PAT_{μ}), a range of PAT values deemed physiologically possible can be established. Beats that are within 13 ms of the PAT_{μ} are deemed acceptable while those outside the range are assumed to be detected incorrectly and corrupted with motion noise. The number of acceptable beats was divided by M to form the detection ratio (thus, a DR of 1 is ideal and 0 would mean no acceptable beats).

While a direct method of assessing the robustness of PAT timings detected based on the D-PPG signal would be preferable theoretically to the DR metric, there is no method short of intra-arterial catheterization that could yield a “gold standard” distal pulse timing reference for a standing subject [13]. Accordingly, we have decided to use the DR metric as a means of gauging the overall acceptability of various sensing methods for beat-by-beat PAT detection from the scale.

III. Results

A. Comparing the DR between the Custom and Commercial Sensors

The average DR of each sensor as the subject was either seated or standing can be seen in Figure 4 with a statistical summary shown in Table I. The average DR for each sensor in the seated position was sufficiently high for all sensors to facilitate heartbeat detection, though the D-PPG sensor did provide approximately 15% improvement in DR. Additionally, the inter-subject variability in waveform quality was substantially lower for the D-PPG sensor, as apparent from the lower standard deviation in DR across subjects.

The differences between the D-PPG sensor and the commercial sensors were more apparent in the standing posture. The average DR for the D-PPG sensor was 92.8% ($\pm 14.0\%$) while the commercial sensors averaged 42.9% ($\pm 39.1\%$), 41.0% ($\pm 32.7\%$), 40.1% ($\pm 40.0\%$), and 46.3% ($\pm 35.7\%$). For the D-PPG sensor, the waveform quality was sufficient for readily extracting signal peaks and the waveform foot (for pulse arrival detection); for the commercial sensors, the motion artifacts and changes in the waveform associated with contact pressure variations from the standing subject corrupted the signal such that the peak and foot timings were not discernable.

B. D-PPG Optimization Results

Figure 5 summarizes the results from changing the location, applied pressure, and sensor layout. Figure 5a shows a box plot of the DR values at different pressures. The larger

pressures generally correlated with the largest DR. Figure 5b presents a box plot of the DR values at each position. The area of the largest mean DR was position 3 followed by position 1 and 2. These three positions have a cumulative mean of $82.7\% \pm 26.5\%$ compared to a cumulative mean of $60\% \pm 31.5\%$ for the other positions. Figure 5c shows a box plot for each of the five LED and photodiode configuration tested. The best sensor was the two-LED, 3 photodiode combination (98.2%).

IV. Discussion

A. Comparing the DR between the Custom and Commercial Sensors

The average DR for each sensor in the seated posture was greater than 70%, and visually the peak and waveform foot locations were readily discernable. The DR decreased substantially, however, for the commercial sensors when the subject transitioned to a standing posture, decreasing by an average of 34.6% ($p < 0.01$ for all sensors). The only sensor to exhibit a high-quality PPG waveform during a standing position was the D-PPG. Though the change of posture did result in a decrease in DR, the mean standing DR remained at an acceptable level and was significantly better than all other sensors ($p < 0.005$).

For most of the subjects in this study, it appears that the change of posture from seated to standing caused variations in the contact pressure placed on the commercial sensors (as the subject had normal postural sway while standing), resulting in a decrease in DR. The strong effects of contact pressure variations on PPG signal amplitude are also known for finger-based PPG sensors [25, 42]. In fact, as the pressure increases above systolic blood pressure, the artery stiffens and the blood volume pulse for a given arterial blood pressure becomes smaller. During the standing posture, the pressure difference placed on the D-PPG sensor was minimized and the signal was able to remain intact. It should be noted that, based on our findings, motion artifacts had a greater effect on the commercial sensor than the D-PPG sensor, and the natural swaying of a person will have a greater affect in the commercial sensor and lead to a deteriorated DR value.

An exciting result from this study is that the D-PPG sensor packaging provided robust and repeatable results regardless of external factors such as body weight, skin tone, and posture. Two of the subjects with high levels of melanin exhibited less than 20% DR regardless of posture for the commercially available sensors. For the D-PPG sensor, eleven of the twelve subjects achieved a DR above 90% while seated or standing. Using an array of sensors, combined with the fact that the sensors were positioned on a well-perfused region of the foot (the dorsum) rather than the toe, were the main contributors to enabling this improvement in performance over the state of the art.

To compare the performance of a commercial sensor to that of the D-PPG sensor, the Biopac's reflective PPG sensor was affixed to the dorsum of the foot in each of the seven locations in Figure 5b and compared to the D-PPG sensor in the same positions. The D-PPG showed a significant increase in DR with a mean increase of 46% ($p < 0.05$).

This work clearly shows an improvement in the PPG signal with the D-PPG sensor when compared to the commercial sensors during standing. This is important due to the increase in

BCG signal quality seen as the subject is standing upright as opposed to seating [43], which is required for reliable PTT measurements.

B. D-PPG Optimization

The remaining tests focused on optimizing the reflective sensor. By increasing the pressure placed upon the sensor, the DR values showed an increasing trend. This result matches that seen when a reflective sensor was placed on the radial artery [24]. The most significant changes in terms of DR improvement occurred between 24 mmHg and 98 mmHg with a mean increase of 38.6% ($p < 0.05$). It is commonly known that the amplitude of a reflective PPG sensor increases as the pressure approaches zero transmural pressure [25, 42]. The increase in pressure towards zero transmural pressure increases arterial compliance [44]. Since PPG is a measure the change in blood volume, the PPG signal amplitude increases for a given arterial blood pressure amplitude. Venous pulsation may have also declined with the increase pressure, reducing the overall noise and increase the DR [45]. Contact pressures exceeding arterial pressure can lead to collapsing arterial walls [42], as well as reduced PPG amplitude. In the current study, this was seen at pressures greater than 146 mmHg where there were no significant differences between the DR values ($p > 0.05$) with six of the twelve subjects experiencing a decrease in PPG amplitude between 171 mmHg and 195 mmHg. In addition, some subjects reported different levels of discomfort when the applied pressure was greater than 146 mmHg. Accordingly, the sensor should be placed ideally at a contact pressure (normal to the skin) of approximately 25mmHg – 120 mmHg, which corresponds to a feeling that the strap is snug on the foot, but not so tight that it is uncomfortable.

Placing the sensor in different locations only showed general trends of the optimal placement: positions 1–3 showed the greatest DR values when compared to positions 4–7. For these positions (1–3), the sensors were placed directly above the deep plantar branch of dorsalis pedis artery and thus correspond to areas of high perfusion for the foot. Along this artery (i.e., among positions 1–3), there was no obvious ideal location for PPG measurements, as the results were similar; however, when the foot was partitioned into two regions (positions 1–3 versus 4–7), the DR was higher by an average of 22.8% ($p < 0.01$) for positions 1–3.

We did not find any significant differences in the signal quality obtained using the different array geometries. However, it should be noted that while sensor configurations I and II have similar DR values ($p > 0.05$), sensor I has an additional LED. Further testing will be needed to determine the optimal LED and photodiode configuration.

V. Conclusion

We demonstrated that the D-PPG sensor yields more reliable and consistent pulse waveforms than commercial toe-based PPG sensors, particularly in a standing posture. The results were consistently better than commercially available sensor packaging options in terms of detection ratio (DR). Future work will focus on further refinements to optimize the distal timing extraction, as well as providing an automatic indication to the user when the strap is placed with optimal contact pressure to enable high signal quality. To make this method more convenient for the user, the strap can be attached to the scale directly such that

the user would simply place his/her foot into the strap when stepping onto the scale. Overall, the weighing-scale based approach for PTT measurement with the BCG as a proximal timing reference and the D-PPG as the distal timing reference may provide a convenient and accurate means of extracting aortic PTT towards arterial stiffness and blood pressure assessment.

Acknowledgments

This material is based upon work supported by the National Institutes of Health, National Institute of Biomedical Imaging and Bioengineering, under Grant 1U01EB018818-01.

References

1. Lehmann ED. Clinical value of aortic pulse-wave velocity measurement. *The Lancet*. Aug.1999 354:528–529.
2. O'Rourke MF, Staessen JA, Vlachopoulos C, Duprez D, Plante G. Clinical applications of arterial stiffness; definitions and reference values. *American Journal of Hypertension*. May.2002 15:426–444. [PubMed: 12022246]
3. Hansen, T Willum, Staessen, JA., Torp-Pedersen, C., Rasmussen, S., Thijs, L., Ibsen, H., et al. Prognostic Value of Aortic Pulse Wave Velocity as Index of Arterial Stiffness in the General Population. *Circulation*. Feb.2006 113:664–670. [PubMed: 16461839]
4. Tomiyama H, Koji Y, Yambe M, Shiina K, Motobe K, Yamada J, et al. Brachial - Ankle Pulse Wave Velocity is a Simple and Independent Predictor of Prognosis in Patients With Acute Coronary Syndrome. *Circulation Journal*. 2005; 69:815–822. [PubMed: 15988107]
5. Townsend RR, Wilkinson IB, Schiffrin EL, Avolio AP, Chirinos JA, Cockcroft JR, et al. Recommendations for Improving and Standardizing Vascular Research on Arterial Stiffness: A Scientific Statement From the American Heart Association. *Hypertension*. Sep.2015 66:698–722. [PubMed: 26160955]
6. Van Bortel LM, Laurent S, Boutouyrie P, Chowienczyk P, Cruickshank JK, De Backer T, et al. Expert consensus document on the measurement of aortic stiffness in daily practice using carotid-femoral pulse wave velocity. *Journal of Hypertension*. 2012; 30:445–448. [PubMed: 22278144]
7. Mukkamala R, Jin-Oh H, Inan OT, Mestha LK, Chang-Sei K, Toreyin H, et al. Toward Ubiquitous Blood Pressure Monitoring via Pulse Transit Time: Theory and Practice. *Biomedical Engineering, IEEE Transactions on*. 2015; 62:1879–1901.
8. Nichols, WW., O'Rourke, MF., Vlachopoulos, C. McDonald's Blood Flow in Arteries Theoretical, Experimental, and Clinical Principles. London: Hodder Arnold; 2011.
9. Geddes L, Voelz M, Babbs C, Bourland J, Tacker W. Pulse transit time as an indicator of arterial blood pressure. *Psychophysiology*. 1981; 18:71–74. [PubMed: 7465731]
10. Ochiai R, Takeda J, Hosaka H, Sugo Y, Tanaka R, Soma T. The Relationship Between Modified Pulse Wave Transit Time and Cardiovascular Changes in Isoflurane Anesthetized Dogs. *Journal of Clinical Monitoring and Computing*. Dec.1999 15:493–501. [PubMed: 12578047]
11. Poon CCY, Zhang YT. Cuff-less and Noninvasive Measurements of Arterial Blood Pressure by Pulse Transit Time. *Proc IEEE 27th Annu Int Conf Eng Med Biol Soc*. 2005:5877–5880.
12. Sola J, Proenca M, Ferrario D, Porchet JA, Falhi A, Grossenbacher O, et al. Noninvasive and Nonocclusive Blood Pressure Estimation Via a Chest Sensor. *IEEE Transactions on Biomedical Engineering*. 2013; 60:3505–3513. [PubMed: 23864147]
13. Payne RA, Symeonides CN, Webb DJ, Maxwell SRJ. Pulse transit time measured from the ECG: an unreliable marker of beat-to-beat blood pressure. *Journal of Applied Physiology*. 2005; 100:136–141. [PubMed: 16141378]
14. Kim C-S, Carek AM, Mukkamala R, Inan OT, Hahn J-O. Ballistocardiogram as Proximal Timing Reference for Pulse Transit Time Measurement: Potential for Cuffless Blood Pressure Monitoring. *IEEE Transactions on Biomedical Engineering*. 2015; 62:2657–2664. [PubMed: 26054058]

15. Pickering TG, Shimbo D, Haas D. Ambulatory Blood-Pressure Monitoring. *New England Journal of Medicine*. 2006; 354:2368–2374. [PubMed: 16738273]
16. Agarwal R, Bills JE, Hecht TJW, Light RP. Role of Home Blood Pressure Monitoring in Overcoming Therapeutic Inertia and Improving Hypertension Control: A Systematic Review and Meta- Analysis. *Hypertension*. Jan.2011 57:29–38. [PubMed: 21115879]
17. Inan OT, Migeotte PF, Kwang-Suk P, Etemadi M, Tavakolian K, Casanella R, et al. Ballistocardiography and Seismocardiography: A Review of Recent Advances. *IEEE Journal of Biomedical and Health Informatics*. 2015; 19:1414–1427. [PubMed: 25312966]
18. Martin SLO, Carek AM, Kim C-S, Ashouri H, Inan OT, Hahn J-O, et al. Weighing Scale-Based Pulse Transit Time is a Superior Marker of Blood Pressure than Conventional Pulse Arrival Time. *Scientific Reports*. Dec.2016 6
19. Zhang G, Gao M, Xu D, Olivier NB, Mukkamala R. Pulse arrival time is not an adequate surrogate for pulse transit time as a marker of blood pressure. *Journal of Applied Physiology*. Dec.2011 111:1681–1686. [PubMed: 21960657]
20. Inan OT, Park D, Giovangrandi L, Kovacs GTA. Noninvasive Measurement of Physiological Signals on a Modified Home Bathroom Scale. *IEEE Transactions on Biomedical Engineering*. 2012; 59:2137–2143. [PubMed: 22318479]
21. Cattivelli FS, Garudadri H. Noninvasive Cuffless Estimation of Blood Pressure from Pulse Arrival Time and Heart Rate with Adaptive Calibration. 2009 Sixth International Workshop on Wearable and Implantable Body Sensor Networks. 2009:114–119.
22. Allen J. Photoplethysmography and its application in clinical physiological measurement. *Physiological Measurement*. 2007; 28:R1.
23. Nijboer JA, Dorlas JC, Mahieu HF. Photoelectric plethysmography-some fundamental aspects of the reflection and transmission methods. *Clinical Physics and Physiological Measurement*. 1981; 2:205. [PubMed: 7338024]
24. Rafolt D, Gallasch E. Influence of Contact Forces on Wrist Photoplethysmography. *Biomedical Engineering*. 2004; 49:22–26. [PubMed: 15032494]
25. Teng XF, Zhang YT. The effect of contacting force on photoplethysmographic signals. *Physiological Measurement*. 2004; 25:1323. [PubMed: 15535195]
26. Paliakaite B, Daukantas S, Sakalauskas A, Marozas V. Estimation of pulse arrival time using impedance plethysmogram from body composition scales. *Sensors Applications Symposium (SAS)*, 2015 IEEE. 2015:1–4.
27. Kato Y, Nambu M, Imura M, Kuroda Y, Oshiro O. Smart sensing of cardiovascular physiological information from soles without direct skin contact. *Proc IEEE 34th Annu Int Conf Eng Med Biol Soc*. 2012:5806–5809.
28. Landaeta RG, Casas O, Pallas-Areny R. Heart Rate Detection from Plantar Bioimpedance Measurements. *Proc IEEE 28th Annu Int Conf Eng Med Biol Soc*. 2006:5113–5116.
29. Park D, Inan OT, Giovangrandi L. A combined heartbeat detector based on individual BCG and IPG heartbeat detectors. *Proc IEEE 34th Annu Int Conf Eng Med Biol Soc*. 2012; 2012:3428–31.
30. Patterson R. Fundamentals of impedance cardiography. *Engineering in Medicine and Biology Magazine, IEEE*. 1989; 8:35–38.
31. Sherwood A, Allen MT, Fahrenberg J, Kelsey RM, Livallo WR, Doomen LJPV. Methodological Guidelines for Impedance Cardiography. *Psychophysiology*. 1990; 27:1–23. [PubMed: 2187214]
32. Critchley LA. Impedance cardiography. The impact of new technology. *Anaesthesia*. Jul.1998 53:677–684. [PubMed: 9771176]
33. Vizbara V. Comparison of Green, Blue and Infrared Light in Wrist and Forehead Photoplethysmography. *Biomedical Engineering*. 2013; 17
34. Fallow BA, Tarumi T, Tanaka H. Influence of Skin Type and Wavelength on Light Wave Reflectance. *Journal of Clinical Monitoring and Computing* vol. Jun.2013 27:313–317.
35. Gaddum NR, Alastruey J, Beerbaum P, Chowienczyk P, Schaeffter T. A Technical Assessment of Pulse Wave Velocity Algorithms Applied to Non-invasive Arterial Waveforms. *Annals of Biomedical Engineering*. 2013; 41:2617–2629. [PubMed: 23817766]

36. Johansson A, Ahlstrom C Fau-Lanne T, Lanne T Fau - Ask P, Ask P. Pulse wave transit time for monitoring respiration rate. *Medical and Biological Engineering and Computing*. Jun.2006 44:471–478. [PubMed: 16937198]
37. Smith RP, Veale D Fau - Pepin JL, Pepin J1 Fau - Levy PA, Levy PA. Obstructive sleep apnoea and the autonomic nervous system. *Sleep Medicine Reviews*. May.1998 2:69–92. [PubMed: 15310503]
38. Michael JD, John A, Alan M. Relation between heart rate and pulse transit time during paced respiration. *Physiological Measurement*. 2001; 22:425. [PubMed: 11556663]
39. Ma T, Zhang YT. A Correlation Study on the Variabilities in Pulse Transit Time, Blood Pressure, and Heart Rate Recorded Simultaneously from Healthy Subjects. 2005 IEEE Engineering in Medicine and Biology 27th Annual Conference. 2005:996–999.
40. Pitson DJ, Sandell A Fau - van den Hout R, R van den Hout R Fau - Stradling J, Stradling JR. Use of pulse transit time as a measure of inspiratory effort in patients with obstructive sleep apnoea. *European Respiratory Journal*. 1995; 8:1669–1674. [PubMed: 8586120]
41. Teng XF, Zhang YT. Theoretical Study on the Effect of Sensor Contact Force on Pulse Transit Time. *IEEE Transactions on Biomedical Engineering*. 2007; 54:1490–1498. [PubMed: 17694870]
42. Tamura T, Maeda Y, Sekine M, Yoshida M. Wearable Photoplethysmographic Sensors—Past and Present. *Electronics*. 2014; 3:282.
43. Javaid AQ, Wiens AD, Fesmire NF, Weitnauer MA, Inan OT. Quantifying and Reducing Posture-Dependent Distortion in Ballistocardiogram Measurements. *IEEE Journal of Biomedical and Health Informatics*. 2015; 19:1549–1556. [PubMed: 26058064]
44. Zheng D, Allen J, Murray A. Effect of external cuff pressure on arterial compliance. *Computers in Cardiology*. 2005:315–318.
45. Shelley KH, Tamai D, Jablonka D, Gesquiere M, Stout RG, Silverman DG. The effect of venous pulsation on the forehead pulse oximeter wave form as a possible source of error in SpO₂ calculation. *Anesth Analg*. Mar.2005 100:743–747. [PubMed: 15728063]

Fig. 1.

Block diagram of the proposed method of PTT/BP extraction. A representative pair of electrocardiogram (ECG), ballistocardiogram (BCG), and photoplethysmogram (PPG) heartbeats from one subject (heart rate = 56 bpm) are shown, and the time points for deriving pre-ejection period (PEP), pulse transit time (PTT), and pulse arrival time (PAT) are annotated.

Fig. 2.

a) Front view of the D-PPG sensor (sensor number (i)) placed under the foot strap, (b) commercially available PPG probes used, with (ii–iv) being transmissive sensors, and (v) a reflective sensor, and (c) typical normalized signals from each of the sensors while a subject is sitting (left pane) and standing (right pane). The commercially available PPG sensors all demonstrate significant artifacts in the standing measurements that reduce the quality of distal timing measurements from the foot. In contrast, the D-PPG sensor preserves the waveform quality and demonstrates distinct features for timing detection.

Fig.3.

(a) Filtered PPG waveforms from the D-PPG and a commercially available PPG while the subject was sitting and standing. A zoomed-in window shows the arrival time of the pulse. The intersection of the horizontal projection going through each beat's minimum and a projection tangent to the maximum derivative represents the foot of the PPG. (b) Extracted PAT for the complete recording. The horizontal black line represents the range of possible PAT values centered on the PAT for the ensemble averaged beat. The D-PPG PAT values were able to stay within the range for both seated and standing while the PAT for commercial sensor regularly showed values outside of the range during standing. To note, the commercial sensor's PAT was shifted to have equal means.

Fig. 4.

Comparison of sitting and standing DR value for each sensor shown in Figure 2. The custom reflective sensor designed in this work achieved the best DR for both the seated and standing postures. In particular, the improvement was substantial in the standing posture, with an average improvement of 52.9% over the existing methods.

Fig. 5.

Results from the three separate optimization tests along with error bars. (a) Applied pressure to the dorsum, (b) location on the dorsal surface, and (c) LED/photodiode configuration.

Author Manuscript

Author Manuscript

Author Manuscript

Author Manuscript

TABLE I

Comparison Between Custom and Commercial Sensors

	Sitting				Standing			
	μ	σ	max	min	μ	σ	max	min
i	0.95	0.15	1.00	0.47	0.93	0.14	1.00	0.53
ii	0.83	0.34	1.00	0.00	0.43	0.39	1.00	0.00
iii	0.82	0.30	1.00	0.00	0.41	0.33	1.00	0.00
iv	0.71	0.33	1.00	0.04	0.40	0.40	0.98	0.00
v	0.73	0.32	1.00	0.10	0.46	0.36	1.00	0.00