Characteristic Ratio-Independent Arterial Stiffness-Based Blood Pressure Estimation

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***Abstract*—Noninvasive blood pressure (BP) measurement is an important tool for managing hypertension and cardiovascular disease. However, automated noninvasive BP measurement devices, which are usually based on the oscillometric method, do not always provide accurate estimation of BP. It has been found that change in arterial stiffness (AS) is an underlying mechanism of disagreement between an oscillometric BP monitor and a sphygmomanometer. This problem is addressed by incorporating parameters related to AS in the algorithm for BP measurement. Pulse transit time (PTT) is first used to estimate AS parameters, which are fixed into a model of the oscillometric envelope. This model can then be used to perform curve fitting to the measured signal using only four parameters: systolic BP, diastolic BP, mean BP, and lumen area at zero transmural pressure. The proposed technique is independent of the experimentally determined characteristic ratios that are commonly used in existing oscillometric methods. The accuracy of the proposed technique was evaluated by comparing with the same model without incorporation of AS, and with reference BP device measurements. The new method achieved standard deviation of error less than 8 mmHg and mean error less than 5 mmHg. The results show consistency with ANSI/AAMI SP-10 standard for noninvasive BP measurement techniques.**

***Index Terms*—Arterial stiffness (AS), blood pressure (BP), mathematical modeling, non-invasive blood pressure measurement, oscillometry.**

I. Introduction

BLOOD pressure (BP) is recognized as an important vital sign for managing hypertension and cardiovascular disease [1]. Despite the paramount importance of BP for making the right diagnostic and therapeutic decisions, it remains an imprecise measurement that is being performed daily in clinical environments [2].

The most popular technique employed in automated noninvasive BP measurement devices is the oscillometric method [3]–[9]. This technique relies on analysis of the oscillometric waveform (OMW) which represents the pressure oscillations detected during cuff deflation. Experimentally-determined characteristic ratios are used to determine systolic blood pressure (SBP) and diastolic blood pressure (DBP) based on the height of the oscillometric waveform envelope (OMWE) at different cuff pressures (CPs) [10]. These ratios are experimentally determined based on large population studies, and are likely vary among subjects with different disease conditions and vessel properties. Moreover, the amplitude of the oscillations is sensitive to the arterial stiffness (AS), and this sensitivity is a significant source of error in measurements [11], [12]. In particular, it is found that as the arteries stiffen, the characteristic ratios for SBP are affected resulting in an error of 20-25% in measurement of SBP, while DBP is influenced less [11]. Several studies [13]–[16] have attempted to estimate BP by modeling the OMW as a function of arterial BP and AS parameters. However, these studies lack sufficient accuracy as they estimate all the BP and AS parameters using the same model of oscillometry which consists of a large number of unknown parameters. For a review of different oscillometric BP estimation algorithms, the reader is referred to [17].

AS generally describes the extent of hardening of the inner wall of the artery, and so it is an indicator of vascular function and structure [1], [18]. AS is influenced by factors like age, BP, and atherosclerosis [19], and is quantified through a variety of indices as a function of pressure, diameter, volume, and pulse wave velocity (PWV) h is inversely related to the pulse transit time (PTT) [20], [21]. The PTT is defined as the time it takes for the pressure wave to travel between two sites on the arterial tree [22]–[28].

The oscillometric method overestimates SBP and underestimates DBP compared to the Korotkoff method in a normal healthy person while it overestimates both SBP and DBP in individuals with stiffened arteries, which also supports the dependency of oscillometric BP estimation on AS [12]. Baker *et al.* [4] formulated the measurement error in oscillometry and its dependency on AS parameters and pulse pressure and showed that multiple parameters including change in AS may affect the accuracy of BP estimates. However, estimation of AS

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often requires estimating the distance between two points on the arterial tree [29], which can be a significant source of error. Therefore, in order to improve measurement accuracy, there is a need to develop an automated BP estimation algorithm that takes AS into account and is independent of any characteristic ratios [10]–[12].

In this paper, we address the problem of AS dependency of the oscillometric BP readings by incorporating the AS-related parameters. The AS-related parameters are estimated from the PTT and are then incorporated into a model of the OMWE which is used for estimation of BP. To the best of our knowledge, this is the first work that employs parameters related to AS using a combined model of PTT and OMWE. This type of individualization is strongly recommended by American Heart Association Council on High Blood Pressure Research [10]. It should be noted that in [30], a model of PTT measured in oscillometry was developed. However, the focus was on studying the dependency of PTT on CP for BP estimation, and the AS parameters were not estimated nor incorporated for individualized BP estimation.

This paper is organized as follows: In Section II, the relation between AS and PTT is explained, a method of estimating AS parameters from PTT is introduced, and our proposed characteristic ratio-independent AS-based BP estimation method is formulated. Data collection and the study population are also given in this section. In Section III, the proposed method is validated on a dataset of 150 recordings acquired from 10 healthy subjects, and the overall results along with advantages and limitations of this work are discussed. In Section IV, the paper is concluded.

II. Methodology

PWV analysis is considered among the most established techniques for assessment of AS [20], [21]. PWV, which is inversely related to PTT, is defined as the propagation speed at which the pressure wave moves in the artery. According to the Bramwell and Hill equation, PWV is inversely proportional to the square root of vessel distensiblity, a marker of AS, which is defined as the relative change in arterial volume for a given pressure change [31], [32].

The ability of the arterial wall for expansion and compression considerably affects the PTT [21]. Extensive research has confirmed the relation between PTT and AS; PTT increases considerably as an individual gets older [33], [34]. It has been shown by experimental and theoretical analysis that change in AS strongly affects the lumen area pressure dependent variation. As the artery stiffens, higher pressure is needed to make the same small changes in lumen area. The slope of the arterial lumen area-CP exponential curve represents AS [4], [35]. Studies on isolated arteries have also confirmed the dependency of the lumen area-pressure curve on AS [36], [37].

Employing a cuff provides the opportunity to investigate lumen area variation at different levels of applied CP, and a model of PTT as a function of lumen area variation gives us the possibility of estimating AS parameters.

In this section, two physiologically-based mathematical models for PTT-CP and OMWE-CP, formulated as a function of the lumen area variation, are described. PTT is determined from the simultaneous ECG and oscillometric measurements and the PTT-CP model is used to estimate the AS parameters. The AS parameters are then incorporated in a model of OMWE-CP to estimate BP independent of any characteristic ratios or AS parameters.

TABLE I
Definition of Symbols

Symbol	Definition
ρ	Blood density
a	AS parameter at negative transmural pressure
b	AS parameter at positive transmural pressure
A_0	Arterial lumen area at zero transmural pressure
A_m	Arterial lumen area when the vessel is entirely distended
A_{cst}	Average unoccluded lumen area while the area at the center of cuff is zero
A_0'	A_{cst}/A_0
A_0''	$\rho \times A_0$
p_a	Arterial pressure
p_c	CP
p_t	Transmural pressure ($p_a - p_c$)
T	Total PTT
τ_0	PTT from the heart to brachial artery
τ_c	PTT underneath the cuff
τ_{c1}	PTT underneath the cuff for negative transmural pressure
τ_{c2}	PTT underneath the cuff for positive transmural pressure
L_0	Length of the artery from the heart to the brachial artery
L_c	Length of the artery underneath the cuff (cuff width)

A. AS Estimation From PTT

We adopt the PTT-CP model developed in [30], [38] to estimate the AS parameters. The main features of the model will be introduced and the focus would be on the estimation of AS parameters using the adopted model.

The lumen area is modeled using two exponential equations for the positive and negative transmural pressures as follows [30], [39]:

$$A(t) = \begin{cases} A_{\mathrm{cst}} + A_0 e^{a p_t(t)} & p_t(t) \le 0 \\ (A_{\mathrm{cst}} + A_m) + (A_0 - A_m) e^{-b\, p_t(t)} & p_t(t) \ge 0 \end{cases} \tag{1}$$

where

$$b = \frac{a}{\frac{A_m}{A_0} - 1}. \tag{2}$$

$A(t)$ represents the average lumen area under the cuff, is lumen area when transmural pressure is zero, A_m is lumen area when the vessel is entirely distended. A_{cst} is the average unoccluded area, while the area at the center of cuff is zero. Transmural pressure, $p_t(t)$, is defined as the difference between intra-arterial pressure $p_a(t)$ and the external applied pressure, $p_c(t)$, *i.e.* $p_a(t) - p_c(t)$, a and b are AS parameters. These two parameters are expressed in mmHg^{-1} and the units of $A(t)$, A_0 and A_m are m^2. The definition of symbols used in this paper is provided in Table I.

According to the Bramwell and Hill equation [31], PWV is directly related to the arterial volume and inversely related to the lumen area pressure-dependent changes. The PTT in oscillometry is composed of two main components [39]: (1) PTT from the heart to brachial artery, $\tau_0(t)$, which is not affected by CP, and so can be considered as a constant, and (2) PTT underneath the cuff, $\tau_c(t)$, which is locally affected by external pressure.

Considering the relation between PWV and PTT, the following formulas are obtained for PTT between heart and cuff $(\tau_0(t))$ and PTT underneath the cuff $(\tau_{c1}(t))$ and $(\tau_{c2}(t))$ at different transmural pressure [30]:

$$\tau(t) = \tau_0(t) + \begin{cases} \tau_{c1}(t) & p_t(t) \le 0 \\ \tau_{c2}(t) & p_t(t) \ge 0 \end{cases} \tag{3}$$

where

$$\tau_0(t) = L_0\sqrt{\rho b\left(\frac{1}{1-\frac{A_m-A_0}{A_m}e^{-b\,p_a(t)}}-1\right)} \tag{4}$$

$$\tau_{c1}(t) = L_c\sqrt{\rho a\left(1-\frac{1}{1+\frac{A_0}{A_{\text{cst}}}e^{ap_t(t)}}\right)} \tag{5}$$

$$\tau_{c2}(t) = L_c\sqrt{\rho b\left(\frac{1}{1-\frac{A_m-A_0}{A_m+A_{\text{cst}}}e^{-bp_t(t)}}-1\right)} \tag{6}$$

where L_0 is length from heart to brachial artery and L_c is length of the artery beneath the cuff (cuff width). According to (4)-(6), $\tau_{c1}(t)$ and $\tau_{c2}(t)$ are functions of the transmural pressure, and $\tau_0(t)$ is independent of CP. More details on this model can be found in [30].

In this work, we employ the PTT-CP model (i.e. equations (3)-(6)) to estimate the AS parameters. The PTT is measured as the time delay between the onset of R-peaks of the ECG signal and zero-crossings of the oscillometric pulses [40], which are located between peaks and troughs where the waveform changes its sign [41]. R-peak is the peak of the R wave in the ECG QRS complex that corresponds to the beginning of the ventricular contraction [42]. It has been shown that the zero-crossings of the oscillometric pulses correspond to the time points at which the arterial pressure is equal to MAP [30]. Therefore, the model of PTT measured from the zero-crossings of the oscillometric pulses can be obtained by replacing the arterial pressure by the MAP in equations (3)-(6):

$$\tau_z(t) = \tau(t)\,|_{p_a(t)=\text{MAP}} \tag{7}$$

where $\tau_z(t)$ denotes the PTT measured from zero-crossing of the oscillometric pulses. According to the relation between intra-arterial pressure and transmural pressure $(p_t(t) = p_a(t) - p_c(t))$, we can conclude that $p_t(t) = \text{MAP} - p_c(t)$ at $p_a(t) = \text{MAP}$. Since $\tau_0(t)|_{p_a(t)=\text{MAP}}$ is not a function of CP it can be approximated as a constant over the cuff deflation period, and the only CP varying components of the PTT model would be $\tau_{c1}(t)|_{p_a(t)=\text{MAP}}$ and $\tau_{c2}(t)|_{p_a(t)=\text{MAP}}$.

Based on the two-segment lumen area-CP exponential model presented in equation (1), AS will change at zero transmural pressure (i.e. at $p_c(t) = \text{MAP}$) [4], with different slopes at positive and negative transmural pressures. Therefore, two different AS indices would characterize these two regions, the one below and the one above MAP. Based on the model of PTT expressed in equations (3)-(6), when we use zero-crossing points to measure the PTT, two separate models are provided for these two regions expressed by $\tau_{c1}(t)|_{p_a(t)=\text{MAP}}$ and $\tau_{c2}(t)|_{p_a(t)=\text{MAP}}$ [4], [30]

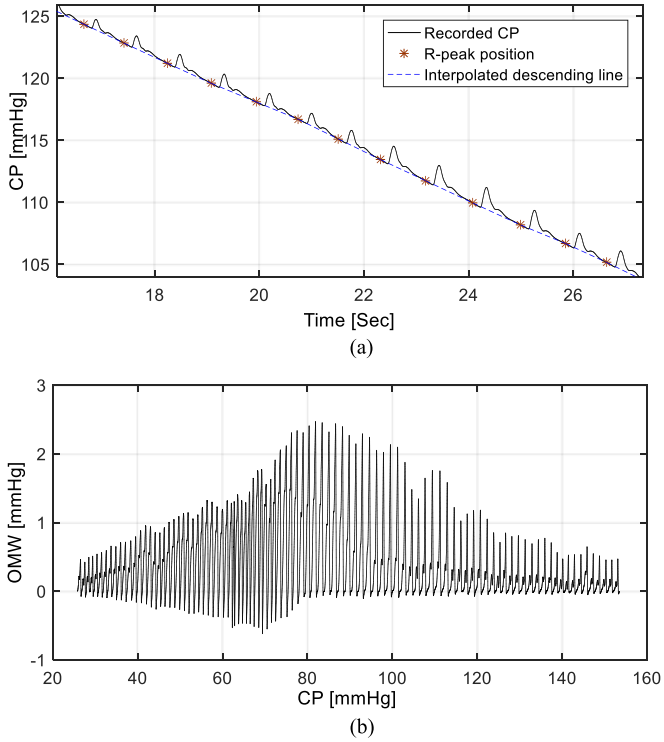

Fig. 1. (a) Recorded CP with R-peaks positions marked by stars. The interpolated descending trend line connecting the dots is also shown. (b) The obtained OMW after subtraction of the descending line from the recorded CP signal.

It has been found that PWV, which is inversely related to the PTT, is better correlated with AS in SBP pressure range (i.e. CPs above MAP) than in DBP range (i.e. CPs lower than the MAP) [43]. Thus, PTT in SBP range τ_{c1} is used to estimate AS parameter a. Parameter b, the AS for the DBP range, is estimated using the relation expressed in equation (2) which will be explained in the next subsection. Based on the experimental results, finding b (for pressures lower than MAP) using this equation, not independently using PTT as is done for parameter a, was found to give more accurate BP estimates.

The OMW is extracted from the recorded CP signal during deflation by: (1) locating pressure pulses corresponding to R-peaks in the CP signal, and forming an interpolated descending line, and (2) subtracting this constructed line from the CP signal [44]. An example of identified pressure pulses corresponding to R-peaks, the interpolated descending line, and the resulting OMW is shown in Fig. 1.

Zero-crossings of the OMW are found using a 2nd order high-pass Butterworth filter with cutoff frequency 0.2 Hz to completely suppress the descending trend in the CP. Then, the minima of the absolute value of all data points within each cycle, from trough to peak, are determined and assigned as zero-crossing points (Fig. 2).

When the zero-crossings are employed to estimate PTT, and hence the AS parameters, the MAP should be estimated and fixed into the PTT-CP model as intra-arterial pressure, $p_a(t)$. The MAP is obtained by finding the CP that corresponds to the maximum amplitude oscillometric pulse [30]. The estimation of

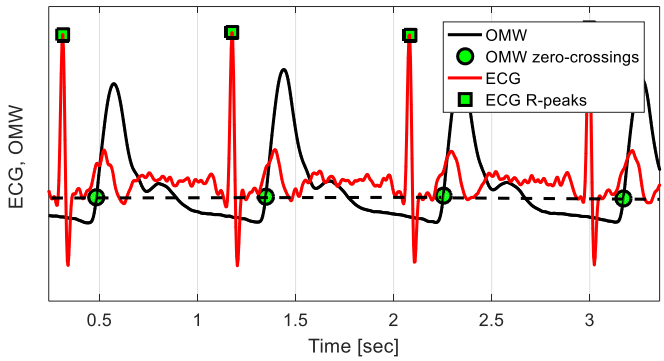

Fig. 2. Zero-crossings (circles) shown superimposed on the OMW. PTT is obtained as the difference between R-peaks (squares) and zero-crossings.

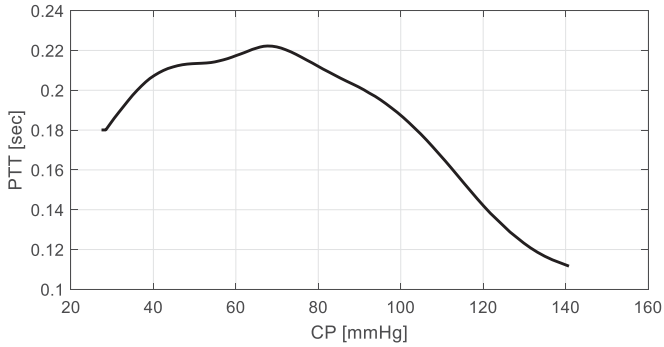

Fig. 3. Smoothed PTT obtained as the time difference between ECG R-peaks and OMW zero-crossings. The obtained PTT is smoothed with a 5-point moving-average filter.

PTT based on zero-crossings in the OMW is illustrated in Fig. 3, after smoothing with a 5 point moving average filter.

Nonlinear least-squares curve fitting based on the trust region reflective method was used to fit the PTT-CP model to the recorded PTT [45]:

$$\underbrace{\min}_{\tau_0, a, A_0'} \left\| \left(\tau_0 + \tau_{c1}\left(t\right)|_{p_a(t)=\text{MAP}}\right) - \tilde{\tau}\left(t\right) \right\|_2^2 \tag{8}$$

where $A_0' = \frac{A_{\text{cst}}}{A_0}$ and $\tilde{\tau}(t)$ is the measured PTT. L_c was set to 0.12 m, and ρ to 1060 kg/m^3. As mentioned earlier τ_0 was assumed to be constant over the cuff deflation period. Initial values for τ_0, A_0', and a were chosen as 0.1 s, 5, and 0.09 mmHg^{-1}, respectively, based on the values reported in [46]. Ranges for these variables over which the optimization was performed were set to $[0-500]$, $[0-\infty]$, and $[0-\infty]$, respectively.

It should be noted that, conventionally, in order to calculate PWV, length should be manually measured or estimated based on height and weight. Length measurements are less accurate for older subjects because their aortas become sinuous due to atherosclerosis. Moreover, length measurement requires involvement of an expert. Errors in estimating of propagation distance lead to imprecise estimation of PWV [29]. The PTT-based physiological model incorporated in this study has the potential to quantify AS using the PTT-CP curve, without need for distance measurement.

B. Ratio-Independent AS-Based BP Estimation

The OMW is modeled by considering lumen area pressure-dependent variation [47], [48]. The arterial lumen area oscillations are transferred to the cuff forming the OMW [11], [49]. The arterial lumen area oscillations during the cuff deflation can be modeled as the difference between the upper and lower envelopes of the arterial lumen area. The mathematical model of the OMWE is therefore defined as follows [13]–[16]:

$$\text{OMWE}\left(t\right) = \varphi(t) \times \left(A\left(t\right)|_{p_a(t)=\text{SBP}} - A\left(t\right)|_{p_a(t)=\text{DBP}}\right) \tag{9}$$

where the vertical bar signifies "at" and $\varphi(t)$ is a proportionality factor between the amplitude of the arterial lumen area oscillations and the amplitude of the oscillometric pulses (OMWE). The rest of variables are as defined previously. The proportionality factor $\varphi(t)$ can be approximated as a constant φ if a linear CP-volume relationship is assumed [13], [16]. On the other hand, some studies have suggested a non-linear CP-volume relationship [3], [5], [11], [14]. In this paper, we considered both a constant and a CP dependent proportionality factor. The CP dependent proportionality factor is given by $\varphi(t) = \varphi \times (p_c(t) + 760)$ [14]. The best estimation results were obtained using a constant proportionality factor and are reported in Section III.

Equations (1) and (9) present the relation between lumen area and OMWE. By substituting equation (1) into equation (9) and simplification, the OMWE model is expressed as follows:

$$\text{OMWE}(t) = \varphi \times \begin{cases} A_0(e^{a(\text{SBP}-p_c(t))} - e^{a(\text{DBP}-p_c(t))}) \\ \text{for SBP} \leq p_c\left(t\right) \\ A_m + (A_0 - A_m)e^{-b(\text{SBP}-p_c(t))} \\ \quad -A_0(e^{a(\text{DBP}-p_c(t))}) \\ \text{for DBP} \leq p_c\left(t\right) \leq \text{SBP} \\ (A_0 - A_m)(e^{-b(\text{SBP}-p_c(t))} \\ \quad -e^{-b(\text{DBP}-p_c(t))}) \\ \text{for } p_c(t) \leq \text{DBP} \end{cases} . \tag{10}$$

Since the arterial lumen area has the maximum oscillation at zero transmural pressure [4], the second term of the equation should have a maximum when CP equals the MAP. Thus taking the derivative of this equation and setting it equal to zero gives the following equation:

$$A_m = A_0 \left(\frac{\text{SBP} - \text{DBP}}{\text{MAP} - \text{DBP}}\right). \tag{11}$$

The OMWE model derived through equations (10)-(11) is only a function of six parameters including SBP, DBP, MAP, *a*, *b*, and $A_0'' = \varphi \times A_0$. In our proposed method, a is estimated according to equation (8) and b is obtained by replacing A_m in equation (2) by equation (11) as follows:

$$b = a \left(\frac{\text{MAP} - \text{DBP}}{\text{SBP} - \text{MAP}}\right). \tag{12}$$

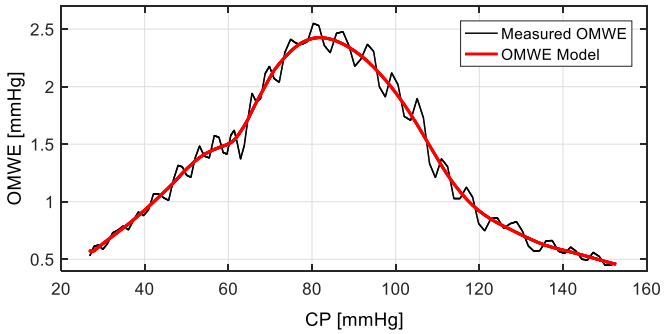

Fig. 4. An illustration of curve fitting the OMWE model to OMWE obtained from real data to estimate SBP and DBP.

Therefore, by incorporating the AS parameters, the number of unknown parameters of the OMWE model is reduced to four. This not only improves the optimization required to estimate the unknown parameters but also eliminates the dependency of BP estimates to AS.

The maximum and minimum amplitude of the recovered OMW (Fig. 1) between two successive R-peaks are found and assigned as peak and trough. The OMWE is then constructed by subtracting troughs from peaks in the OMW.

The OMWE-CP model, equation (10), is fit to the recorded OMWE by incorporating stiffness parameters obtained from the PTT-CP model. The SBP and DBP values are obtained through nonlinear least-squares curve fitting between the derived model in equations (10)-(12) and the measured OMWE:

$$\min_{\text{SBP, DBP, MAP, } A_0''} \left\| (\text{OMWE}(t) - \widetilde{\text{OMWE}}(t) \right\|_2^2 \qquad (13)$$

where $\widetilde{\text{OMWE}}(t)$ represents the measured OMWE. Initial values for SBP, DBP, MAP, and A_0'' were 120, 80, 100 mmHg, and 0.1. The ranges of SBP, DBP, MAP, and A_0'' over which the optimization was performed were set to $[70-150]$, $[40-100]$. $[50-100]$, and $[0-\infty]$, respectively [46]. Different sets of initial values and ranges were examined and the method was not sensitive to the examined initial values and ranges. An illustration of the fitting the OMWE model to real data is shown in Fig. 4.

The algorithm was repeated after smoothing the recorded OMWE with a seven-point moving average filter followed by fitting the curve with a spline function using a smoothing parameter of 0.1, and there was little change in the results, so the technique is insensitive to the smoothing function.

C. Data Collection

The performance of the proposed method was evaluated on a pilot dataset obtained from 10 healthy subjects aged from 24 to 63 years (six males and four females) without a history of cardiovascular disease or hypertension. All subjects provided their informed consent in accordance with the guidelines of the institutional Research Ethics Board.

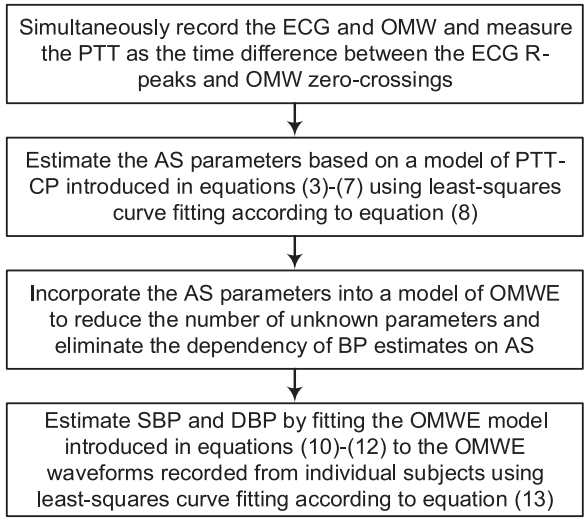

Fig. 5. Summary of our proposed method, which derives from AS parameters being one of the determinant factors of the oscillometric measurement and its governing equation.

A multi-parameter monitoring device (named InBeam) was developed in our group and is able to simultaneously register the ECG and CP during cuff deflation [38], [44]. ECG is recorded by employing two electrodes. The conductive fabric attached to the cuff provides one electrode. For the second electrode another conductive fabric is placed inside a wristband which is worn on the right wrist of the subject. The CP signal is acquired using the cuff, which is placed over the brachial artery of the left arm. Apart from the wristband, from the point of view of the user, the InBeam device operates like a regular automated BP monitor [50].

The BP was first recorded by the FDA-approved Omron HEM-790IT device to obtain a reference measurement with a cuff placed on the right arm. Subsequently, a measurement was obtained with the InBeam device and the cuff is placed on the left arm. This procedure was then repeated five times, with a gap of 3 minutes between each repetition. Although there may be differences in BP measured from right and left arms, studies have shown that the mean interarm difference is not significant in healthy subjects [51] such as the ones tested in this investigation.

The data was collected on three different days (not necessarily subsequent days) with five measurements per day, which resulted in a dataset of 150 recordings. More details on the measurement setup and protocol are found in [44].

III. Experimental Results and Discussion

A summary of the overall method proposed in this work is given in Fig. 5. In this section, our proposed method is evaluated on the dataset of 150 recordings and the results are discussed.

To assess the performance of our proposed method, the following error measures are used: mean error (ME), mean absolute error (MAE), and standard deviation of error (STDE), where error is defined as the difference between estimated BP and the reference value in mmHg.

TABLE II
SBP AND DBP ESTIMATION ERRORS USING THE OMWE-CP MODEL ONLY

	MAE	STDE	ME
SBP	11.51	13.42	5.75
DBP	6.90	8.71	−1.59

Errors are reported in terms of MAE, STDE, and ME in units of mmHg.

TABLE III
SBP AND DBP ESTIMATION ERRORS WHEN AS VALUES ARE FIRST ESTIMATED FROM PTT-CP MODEL AND THEN FIXED INTO THE OMWE-CP MODEL

	MAE	STDE	ME
SBP	4.95	6.31	−1.23
DBP	5.40	7.38	2.60

Errors are reported in terms of MAE, STDE, and ME in units of mmHg.

The performance obtained with the dataset of 150 measurements by fitting the OMWE-CP model to the data without fixing AS values into the model is shown in Table II, while the performance obtained when the AS parameters are first estimated using the PTT-CP model and fixed in the OMWE-CP model are shown in Table III. As can be seen, except for the slight increase in magnitude of ME with DBP, all performance measures improved when estimates of AS are fixed in the OMWE-CP model, and particularly for SBP.

To further explore the performance of the proposed method compared to the reference method, Bland-Altman analysis was performed [52]. The difference between the results obtained with the two methods is plotted against the mean of the result obtained with the methods. The bias (the same as ME) is represented as the central horizontal line, and the limits of agreement (LoA) correspond to the bias ± 1.96 the standard deviation of differences between the two methods, and these are represented as the two horizontal lines above and below the bias.

Figs. 6 and 7 show the Bland-Altman plots for SBP and DBP obtained using the OMWE-CP model with and without fixing the AS parameters in the model.

The greater improvement obtained with estimation of SBP, indicates that SBP is more sensitive to AS than DBP. As stated previously, based on one study, PWV as an indicator of AS at SBP level is better correlated with vessel characteristics than PWV at DBP [43].

PWV, which is arguably the most common measure used for AS estimation [22], [24], [26]–[28], requires distance measurement and an appropriately trained device operator. However, distance can only be approximated and not precisely measured. Our method of arterial AS estimation is independent of any distance measurement and does not need a trained operator to perform the measurement. Therefore, our proposed approach could be readily incorporated into an automated BP measurement device, such as the device described earlier, which has the capability to simultaneously measure ECG with a dry electrode embedded in the cuff. However, it should be noted that the proposed method estimates AS in a peripheral vessel, whereas

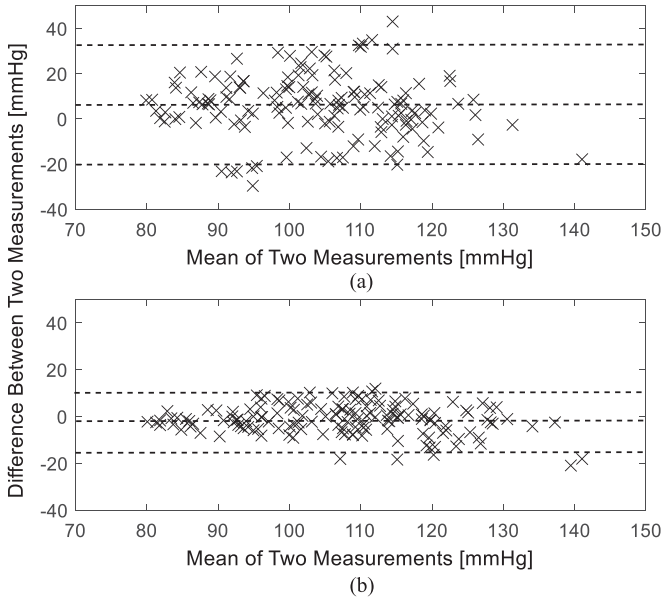

Fig. 6. (a) Bland-Altman plot which compares SBP estimation from OMWE-CP model without fixing the AS parameters and the reference measurements. The Bias is 5.75 mmHg and LoA are from –20.83 to 32.34 mmHg. (b) Bland-Altman plot which compares SBP estimation from OMWE-CP model with AS parameters first obtained from the PTT and the reference measurements. The Bias is –1.23 mmHg and LoA are from –13.75 to 11.28 mmHg.

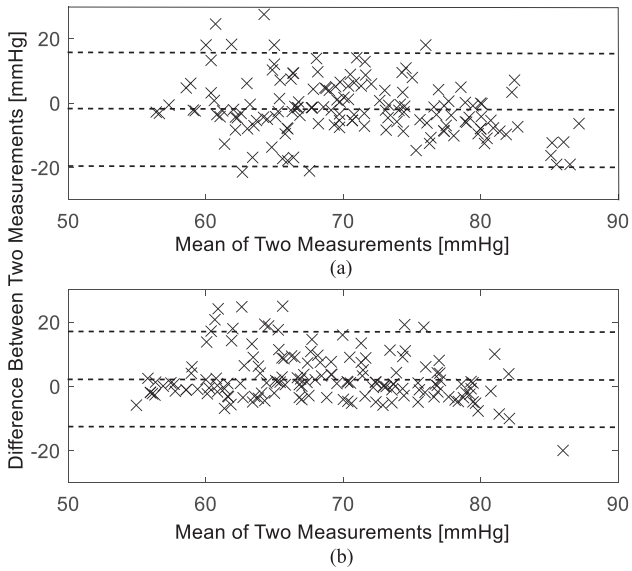

Fig. 7. (a) Bland-Altman plot which compares DBP estimation from OMWE-CP model without fixing the AS parameters and the reference measurements. The Bias is –1.59 mmHg and LoA are from –18.86 to 15.66 mmHg. (b) Bland-Altman plot which compares DBP estimation from OMWE-CP model with AS parameters first estimated from the PTT, and the reference measurements. The Bias is 2.60 mmHg and LoA are from –12.01 to 17.23 mmHg.

current PWV-based methods usually target stiffness of more central arteries.

Some other studies have also presented models of oscillometric CP waveform by taking into account AS parameters [13]–[16]. However, an important difference between these methods and our proposed approach is that they used the OMWE to es-

timate AS coefficients and BP values. In contrast, we used the PTT, which is obtained by simultaneously recording the ECG and the OMW, to estimate AS parameters from another channel of information which is separate from the OMWE. We expect that using these two sources of information in our method would add to the reliability of the BP estimates.

It should be noted that since the proposed method does not appear to be subject to consistent bias, as shown in the Bland-Altman plots, then using a less accurate reference such as the commercial device used in this study for validation would be expected to contribute additional random errors to the results. Therefore, it is possible that the errors that are associated with our proposed method are in reality smaller than we reported.

In this paper, we considered both linear and CP-dependent relationships between the amplitude of the oscillometric pulses (OMWE) and the amplitude of the arterial lumen area oscillations during cuff deflation (see equation (9)). The best estimation results were obtained using the linear model and are reported in Tables II and III and Figs. 6 and 7. The CP-dependent model was adopted from [14]. Using more accurate models for CP-volume relationship could lead to improved results.

The processing was done on a PC with an Intel Core i5-5200 U 2.2 GHz processor and 4.00 GB of RAM using MATLAB software. Compared to the OMWE curve fitting without considering the AS parameters that involves only one nonlinear least-squares optimization, our proposed method consists of two such optimizations; one to estimate the AS parameters using equation (8) and one to estimate the BP using equation (13). On average, the additional optimization process in equation (8) took approximately 0.8 sec for each measurement. However, after incorporating the AS parameters which reduced the number of unknown parameters, the optimization process in equation (13) took approximately 0.3 sec shorter. Overall, our proposed AS-based BP estimation algorithm was 0.5 sec slower than the OMWE curve fitting without considering the AS parameters. This additional delay is not significant for usual discontinuous measurements of BP using automated cuff-based devices.

IV. Conclusion

The aim of this work was to develop a new model-based method for estimation of BP which is optimized for each individual subject, is not sensitive to the AS variations, and is independent of any experimentally-derived characteristic ratios. A non-invasive method independent of any distance measurement was employed to estimate AS. This method was based on the measurement of PTT over the cuff deflation and estimating AS parameters using a model of PTT-CP. Afterwards, the estimated AS values were fixed into a model of the OMWE which reduced the number of unknown parameters in the model and eliminated the dependency of the oscillometric BP estimates on AS. Nonlinear least-squares curve fitting based on the trust region reflective algorithm was used to estimate the model parameters including BP values. Our physiologically-based mathematical models were formulated according to the vessel lumen area variation, the area through which blood passes in the artery. The validation results showed that the incorporation of AS values into the oscillometric model improves measurement accuracy compared to a purely oscillometric method without incorporation of AS.

Our proposed oscillometric BP estimation method is independent of any characteristic ratios that are experimentally determined from large scale group studies. Moreover, it is not sensitive to the AS variations. Therefore, our proposed method should be less sensitive to the variation between different individuals in different age groups and/or with different health conditions. We conclude that our approach can be used as a first step towards creating an individualized patient-specific oscillometric BP estimation method.

In a future study, comparing the results of the proposed approach with auscultatory or intra-arterial measurements would give a more definitive validation. A larger study population with different health conditions would also provide needed data for further validation.

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