Ratio-Independent Blood Pressure Estimation by Modeling the Oscillometric Waveform Envelope

Mohamad Forouzanfar, Hilmi R. Dajani, Voicu Z. Groza, Miodrag Bolic, Sreeraman Rajan, and Izmail Batkin

Abstract—Oscillometry is the most common measurement method used in electronic blood pressure (BP) monitors. However, most of the existing oscillometric algorithms employ empirical ratios on the oscillometric waveform envelope (OMWE) to determine the systolic pressure (SP) and diastolic pressure (DP). As these algorithms do not consider the cardiovascular system parameters that may vary due to health conditions or age, the pressure estimates are not always reliable. In this paper, we develop a new mathematical model for the OMWE by incorporating an existing model of the cuff–arm–artery system. The unique feature of our developed model is that it explicitly represents the relationship between the OMWE and the SP and DP. Based on our developed model, we propose a new ratio-independent oscillometric BP estimation method. The proposed method is based on minimizing the sum of the squared errors between our model and the OMWE using the trust-region-reflective algorithm. Our proposed method is validated in a pilot study against Omron HEM-790IT and BpTRU BMP-100 BP monitors. It is found that the mean absolute error of the proposed method in estimating SP and DP is 4.60 and 4.53 mmHg, respectively, relative to the Omron monitor, and 3.66 and 2.84 mmHg, respectively, relative to the BpTRU monitor. The proposed model thus shows promise toward developing robust BP estimation methods.

Index Terms—Blood pressure (BP), estimation, mathematical model, oscillometry.

I. INTRODUCTION

Most electronic blood pressure (BP) monitors are based on the oscillometric approach. An occlusive cuff coupled to a pressure sensor is placed around the subject’s arm and gradually deflated from a suprasystolic to a subdiastolic pressure. Over the duration of the deflation, the cuff pressure oscillations form a signal known as the oscillometric waveform. The oscillometric waveform envelope (OMWE) is usually further processed to estimate the mean arterial pressure (MAP), the systolic pressure (SP), and the diastolic pressure (DP). However, most of the oscillometric algorithms are without physiological and/or theoretical foundation and rely on empirical relationships between the SP and DP and the OMWE [1].

In the literature, several models for oscillometry have been developed [2]–[7]. In [2] and [3], the cuff–arm–artery system was modeled by considering the mechanics of the occlusive cuff and of the brachial artery, the effect of arterial and cuff pressures, and the pressure transmission from the cuff to the brachial artery through the arm soft tissue. The developed models were used to study and test the validity of fixed empirically derived ratios for estimation of SP and DP from the OMWE. It was found that these empirical ratios should be changed as the parameters of the cardiovascular system vary between different health conditions, age groups, and so on. In [4] and [5], compliance models were developed to describe the volume changes of the artery at different cuff pressures. The compliance information was incorporated to estimate the subject’s BP. However, this method relied on preestimates of the SP and DP obtained by an independent method to estimate the vessel compliance. In [6], the relationship between the SP and DP values and the OMWE was implicitly modeled using neural networks. In [7], a mathematical model for the pulse transit time (PTT) was proposed and it was shown that BP can be determined directly from maxima of PTT signal.

While there have been extensive studies on the theory of oscillometry, to the best of the authors’ knowledge, no explicit mathematical model of the OMWE as a function of SP and DP currently exists. A precise model of the OMWE as a function of SP and DP eliminates the reliance of the oscillometric algorithms on empirical ratios and can play an important role in further development of accurate BP estimation algorithms. In this paper, 1) we derive a new physiologically based mathematical model for the OMWE as a function of SP and DP by incorporating an existing model of the cuff–arm–artery system and 2) based on the developed model, we propose a novel ratio-independent BP estimation method. Unlike the conventional oscillometric algorithms that employ empirically derived fixed ratios to estimate SP and DP from the OMWE, our proposed method estimates the SP and DP for each individual measurement by minimizing the sum of the squared errors between the developed model and the OMWE.

II. METHODOLOGY

A. Measurement Setup

In this paper, the oscillometric waveforms were acquired using an oscillometric device prototype designed in our research laboratory. Our measurement system consists of an arm cuff, a mini dc automatic air pump, an analog pressure transducer, the National Instruments C Series 9239 data acquisition module, and the National Instruments LabVIEW system design software. A detailed description of our measurement system can be found in [8], [9].

B. Model and Method for Estimation of BP

In this section, the necessary components of the oscillometry model proposed in [2] will first be introduced in (1)–(3). Equations (4)–(7) correspond to our OMWE model and BP estimation method that are proposed in this paper.

Arterial mechanics at different transmural pressures are used in modeling oscillometry [2]–[5]. Transmural pressure is defined as the difference in pressure between the two sides of the arterial wall, as follows:

\[ p_t(t) = p_a(t) - p_c(t) \]  \hspace{1cm} (1)

where \( p_t(t) \), \( p_a(t) \), and \( p_c(t) \) represent the transmural, arterial, and cuff pressures, respectively. The arterial pressure \( p_a(t) \) is modeled using a Fourier series representation [2]

\[ p_a(t) = MAP + \alpha_1 \cos(2\pi f_c t) + \alpha_2 \cos(4\pi f_c t + \phi) \]  \hspace{1cm} (2)

where two harmonics of the cardiac signal are considered to be adequate as they carry most of the signal power [2]. In this model,
$f_c$ is the cardiac rate, $a_1$ and $a_2$ represent the amplitude of the first and second harmonics of the cardiac signal, respectively, and $\phi$ is the phase difference between the two harmonics.

The combination of nonlinear geometric collapse and elastic distention of an artery is modeled as follows [2]:

$$A(t) = d \ln(a) SP - p_c(t)) + b)/(1 + e^{-c(p(t))}) \tag{3}$$

where $A$ is the arterial lumen area and $a$, $b$, $c$, and $d$ are subject-dependent constants that account for different cuff–arm–artery system characteristics. Fig. 1 shows an example of the arterial lumen area obtained by (1)–(3). Typical values of the cardiovascular parameters in human artery were used in this simulation [2]; where $a = 0.03 \text{ mmHg}^{-1}$, $b = 3.3$, $c = 0.1 \text{ mmHg}^{-1}$, $d = 0.08 \text{ cm}^2$, $f_c = 80 \text{ beats/min}$, $a_1 = 10 \text{ mmHg}$, $a_2 = 9 \text{ mmHg}$, $\phi = -1.2 \text{ radians}$, $\text{MAP} = 95 \text{ mmHg}$, $\text{SP} = 114 \text{ mmHg}$, and $\text{DP} = 82 \text{ mmHg}$.

The amplitudes of the oscillometric pulses form the OMWE and may be viewed mathematically as a function of cuff pressure. It has been shown that the amplitude of the oscillometric pulses is proportional to the amplitude of the arterial lumen area oscillations during cuff deflation [2], [3]. Therefore, a model of the amplitude of the arterial lumen area oscillations as function of cuff pressure can equivalently represent the OMWE. The amplitude of 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 oscillations, as shown in Fig. 1. According to (1)–(3), the peaks of the arterial lumen area oscillations occur when the arterial pressure $p_a(t)$ is maximum, i.e., at $p_a(t) = \text{SP}$. Therefore, the arterial lumen area upper envelope $A_{ue}$ can be modeled as

$$A_{ue}(t) = d \ln(a) \left(\text{SP} - p_c(t)\right) + b)/(1 + e^{-c(p(t))}) \tag{4}$$

Likewise, the troughs of the arterial lumen area oscillations occur when the arterial pressure $p_a(t)$ is minimum, i.e., at $p_a(t) = \text{DP}$. Therefore, according to (1)–(3), the arterial lumen area lower envelope $A_{le}$ can be modeled as

$$A_{le}(t) = d \ln(a) \left(\text{DP} - p_c(t)\right) + b)/(1 + e^{-c(p(t))}) \tag{5}$$

Now, the OMWE can be modeled as the difference between the upper and lower envelopes of the arterial lumen area oscillations times a scaling factor $\eta$, as follows:

$$\text{OMWE}(t) = \frac{d' \ln(a) \left(\text{SP} - p_c(t)\right) + b)}{1 + e^{-c(p(t))}} - \frac{d' \ln(a) \left(\text{DP} - p_c(t)\right) + b)}{1 + e^{-c(p(t))}} \tag{6}$$

where $d' = \eta \times d$. Note that the OMWE model in (6) consists of six unknown parameters, including SP and DP.

To estimate the SP and DP values, the proposed model in (6) is fitted to the OMWE extracted from the recorded BP signal. The optimum parameters of the model are found using the least-squares fitting approach, as follows:

$$\left[\text{SP}, \text{DP}, \hat{a}, \hat{b}, \hat{c}, \hat{d}'\right] = \arg \min_{\text{SP,DP}, a,b,c,d'} \int (\text{OMWE}(t) - \hat{\text{OMWE}}(t))^2 dt \tag{7}$$

where $\text{OMWE}(t)$ represents the extracted OMWE. The estimated parameters are shown on the left-hand side of the equation. The aforementioned optimization problem is solved using the trust-region-reflective algorithm [10].

C. Validation Setup

Our proposed method was initially validated with a pilot investigation on two sets of recordings. The first set consists of 150 oscillometric BP recordings acquired from 10 healthy subjects aged 24–63 years. Recordings from each subject were obtained on three separate days with five sets of recordings on each day. Each set of recordings started with the reference Food and Drug Administration-approved Omron HEM-790IT monitor measurement on the right arm. As soon as the Omron measurement had ended, the measurement on the left arm using our device prototype started. For more details regarding the data collection protocol, the reader is referred to [8]. An additional data collection on two subjects with six paired measurements with an accurate device designed for the office and clinic settings, the BpTRU BPM-100 monitor, was also done. The BpTRU’s and our device’s cuffs were attached on different arms and paired measurements were done one after the other with 3 min pause between the paired measurements. These studies were approved by the University of Ottawa Research Ethics Board and written informed consent was obtained from all subjects. The performance of the proposed method was evaluated in terms of mean error (ME), mean absolute error (MAE), and standard deviation of error (SDE).

III. Experimental Results

The oscillometric waveform was extracted from the recorded cuff pressure signal using a second-order bandpass digital Butterworth filter with lower cutoff frequency of 0.5 Hz and upper cutoff frequency of 20 Hz. The OMWE was formed by subtracting the consecutive peaks and troughs of the oscillometric waveform. The optimum model parameters, including SP and DP, were obtained as the solution to the optimization problem in (7), which was implemented in discrete time. Typical values of the cardiovascular parameters in human artery mentioned in Section II were used as initial values for the trust-region-reflective algorithm. Examples of the OMWE and the fitted model are shown in the solid gray and dashed black curves, respectively, in Fig. 2.

It was found that the ME, MAE, and SDE in the estimation of SP are 0.04, 4.60, and 5.84 mmHg, and in the estimation of DP are −1.75, 4.53, and 5.97 mmHg, respectively, relative to the Omron monitor. It was also found that the ME, MAE, and SDE in the estimation of SP are −1.33, 3.66, and 4.33 mmHg, and in

\begin{figure}[h]
\centering
\includegraphics[width=0.5\textwidth]{fig1.png}
\caption{Example of the arterial lumen area (solid gray curve). The upper and lower envelopes are shown in dashed and dashdot curves, respectively.}
\end{figure}

\begin{figure}[h]
\centering
\includegraphics[width=0.5\textwidth]{fig2.png}
\caption{Examples of the OMWE and the fitted model proposed in (6).}
\end{figure}
the estimation of DP are $-2.50$, $3.00$, and $2.84$ mmHg, respectively, relative to the BpTRU monitor.

To demonstrate the agreement between our proposed method and the Omron monitor, the Bland-Altman analysis was performed. Fig. 3 shows the Bland–Altman plot of the SP and DP estimates for our proposed method versus the Omron monitor. The $x$-axis of the plots shows the average of our proposed method and the Omron monitor, while the $y$-axis shows the difference between the two devices. The bias (ME) and the limits of agreement (ME $\pm 1.96 \times$ SDE) are shown in dashed and dotted lines, respectively. It is observed that our method slightly overestimates the SP and DP at pressures approximately under 90 and 60 mmHg, respectively, compared with the Omron monitor. However, in total, the errors are almost evenly distributed over the measured pressure range. That is, the BP estimates made by our proposed method are in close agreement with those made by the Omron device. Although not shown in Fig. 3, our proposed method also provided estimates that were in close agreement with BpTRU BP monitor.

IV. CONCLUSION

The main contributions of this paper are the development of a new mathematical model for the OMWE as a function of the SP and DP, and the formulation of a novel ratio-independent method for estimating BP. Unlike existing models of oscillometry, our proposed model provides an explicit relationship between the OMWE and the SP and DP. Therefore, our proposed model can be used to develop accurate BP estimation algorithms and study the effect of different cuff–arm–artery system parameters on the accuracy of BP estimates. Our proposed method estimates the SP and DP for each individual measurement as the values that minimize the sum of the squared errors between the OMWE and our developed model. Unlike the conventional oscillometric algorithms, the proposed method does not use empirical ratios and all the parameters used in the method for the estimation of SP and DP are derived from the subject’s oscillometric waveform itself. Therefore, our developed method shows promise toward providing individualized robust BP estimation in a variety of patients, age groups, and monitoring situations.

The successful cross validation with empirical data supports the accuracy of the proposed theoretical model and BP estimation method. Future work will involve undertaking clinical testing on a dataset of 150 measurements collected from a minimum of 15 subjects, where the method will be compared against invasive intraarterial measurements.

REFERENCES