Coefficient-Free Blood Pressure Estimation Based on Arterial Lumen Area Oscillations in Oscillometric Methods

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***Abstract*—In this paper we present a novel algorithm destined to estimate systolic and diastolic blood pressures from lumen area oscillations of vessel underneath the cuff by using automated oscillometric method. This algorithm is composed of three procedures which process only the diastolic region information of oscillometric waveform (OMW) from 80 mmHg to 20 mmHg, a domain which is considered a low cuff pressure region. The standard oscillometric methods need inflating the cuff to Supra Systolic Pressure (SSBP) and, as such, require a deflation time proportionally longer than our method. Because of the relative low inflation pressure, our method represents a viable option for patients who need to be monitored continuously, since, in such circumstances, the cuff is not allowed to be inflated to higher pressures for longer periods of time.**

We developed a unified algorithm composed of three integrated procedures which are all based on the arterial lumen area oscillation model at diastolic region to estimate blood pressure. The first procedure estimates the compliance c of the blood vessel. The second procedure uses c and estimates the maximum lumen area (Am), lumen area at mean arterial pressure - MAP ($A0$) and the systolic arterial pressure (SBP) from the peaks of the OMW pulses. The third procedure uses $c, Am, A0$ found above to estimate diastolic arterial pressure (DBP) from the troughs of the OMW pulses. The OMW is obtained by filtering the cuff deflation curve (CDC) with a 2nd order Butterworth band pass filter and which has cut-off frequencies of 0.5 to 20 Hz. The proposed method avoids using empirical systolic and diastolic ratios for estimation of blood pressure (like the popular maximum amplitude algorithm - MAA), but rather employs the least square method to optimize the lumen area oscillations model for targeted parameters.

We applied this method on 150 oscillometric traces recorded from 10 healthy subjects composed of males and females from 25 to 63 years old, and validated the results with values measured by an Omron device that served as reference for each recording. Results are encouraging as mean absolute errors of estimated values from the Omron references over 150 recordings are 5.13 mmHg in terms of SBP and 3.18 mmHg for DBP with the standard deviation of errors of 3.60 mmHg and 2.58 mmHg respectively.

Keywords*—Arterial blood pressure; systolic blood pressure; diastolic blood pressure; blood vessel compliance; oscillometric waveform; artery lumen area; diastolic region; optimization

I. INTRODUCTION

Blood pressure (BP) is an important measurand recorded from a subject to diagnose and manage cardiovascular problems such as hypertension, hypotension or any kind of heart diseases. [1-2]. Nowadays non-invasive measurement of blood pressure has become more popular in many clinical observations. [3-4]. Automated non-invasive BP (NIBP) devices record cuff pressure oscillations during cuff deflation time [5-6]. There are many methods available in the literature that can estimate systolic BP (SBP), and diastolic BP (DBP) from the cuff pressure oscillations at deflation time.

Accuracy of the BP measurement is the most important topic, because sometimes even a small quantity of error in measurements can end up to wrong diagnoses and eventually put the patients in dangerous conditions like stroke and other critical heart diseases [7-9, 19]. Although there are significant improvements in NIBP monitoring devices addressed in the literature, accuracy of the measurements is still a big challenge and researches are working on different models to increase reliability of automated NIBP measuring devices. Also, noise is an important factor affecting the accuracy of measured BP, especially in chronic patients with critical heart diseases like atrial fibrillation (AF). Any kind of noise during BP measurement such as motion artifacts or breathing influence will mislead the physician for proper diagnosis [10-12].

There are a number of commercially available blood pressure monitors that meet the current standards for automated oscillometric devices [13-18]. The most common discussed algorithm is the conventional maximum amplitude algorithm (MAA) [20]. Oscillometric waveform envelope (OMWE) is constructed from peaks minus troughs of each oscillometric pulse and cuff pressure at maximum amplitude exhibits the mean arterial pressure (MAP) [25]. Systolic and diastolic blood pressures (SBP, DBP) are obtained from MAP by using two empirical coefficients determined a priori. MAA is simple and robust, but the problem is the empirical coefficients that limits the accuracy of the measurement because “no one fits all” patients and conditions.

There are coefficient-free algorithms that rely on specific features of OMWE. In Maximum/Minimum Slope Algorithm

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(MMSA) SBP is estimated from maximum slope of OMWE in systolic region, and DBP from minimum slope of OMWE in diastolic region [20]. Temporal positions of slopes are detected from first derivative of OMWE where the absolute value of amplitudes of the derivative is maximum in either systolic or diastolic regions. This algorithm is very sensitive to noise, and, if the recorded signal is a bit noisy, the estimated blood pressures are affected and we need to remove noise from the recorded trace, which will increase the complexity of the algorithm.

Another approach is to estimate blood pressure from pulse transit times (PTT) [22]. Pulse transit times are delays from onset of R_peak of each ECG signal to corresponding peaks (PTT_pk), zero_crossings (PTT_zc), or troughs (PTT_tr) of the arterial pressure pulses. Theoretically, there should be only one maximum for each of the three mentioned vectors that will indicate temporal positions of systolic, diastolic, and mean arterial pressures: SBP is cuff pressure when PTT_pk becomes maximum, and, similarly, MAP, and DBP are cuff pressures when PTT_zc, and PTT_tr are maximum respectively [22]. In practice, it rarely happens, because this method is very sensitive to noise. There are different kinds of noises present at recording time that affect measured pulse transit times over cuff deflation period. Breathing is the most important factor that affects blood pressure signal during measurement and changes amplitude and frequency of OMW at peaks, zero_crossings, and troughs, so we cannot expect to get correct pulse transit times unless we remove the noise. Removing noise will again increase the complexity of the algorithm which is not desired.

II. Methods

In this paper we propose a reliable and accurate coefficient-free method that estimates blood pressure from lumen area oscillations of vessel underneath the cuff without using ECG. Lumen area oscillates under the influence of two pressure components: the first one is the arterial pressure (Pa) that is exerted by the blood onto the internal arterial wall, and the other one is the cuff pressure (Pc) at opposite direction, onto the exterior of the arterial wall. The arterial wall underneath the cuff is at transmural pressure $Pt(t)$, defined as the difference between arterial and cuff pressure [21].

$$Pt(t) = Pa(t) - Pc(t) \tag{1}$$

Arterial lumen area $A(t)$ properties are different at systolic and diastolic regions. Systolic region is where arterial pressure is lesser than cuff pressure, and conversely diastolic region is where arterial pressure is greater than or equal to cuff pressure.

$$\begin{cases} A(t) = Ac + A0e^{aPt(t)} & for\ Pt(t) < 0 \\ A(t) = (Ac + Am) - (Am - A0)e^{-cPt(t)} & for\ Pt(t) \geq 0 \end{cases} \tag{2}$$

In equation (2), Ac is the lumen area when the cuff is completely inflated and vessel is collapsed. Since the vessel at all portions underneath the cuff is not completely closed at supra systolic blood pressure (SSBP), Ac represents the average area of collapsed vessel, and its value is close to zero. Am is the maximum lumen area at fully expanded vessel, and $A0$ is the lumen area at MAP where arterial and cuff pressures are equal. The two remaining parameters a and c reflects the compliance of the vessel at systolic and diastolic regions respectively [21].

The lumen area oscillation $A(t)$ is composed of two components: a slow varying component $A1(t)$ due to the deflating cuff pressure, and an oscillating component $A2(t)$ because of the blood pressure pulses.

$$A(t) = A1(t) + A2(t) \tag{3}$$

Since our method needs only the oscillating component $A2(t)$ to be processed, the slow varying component $A1(t)$ has to be removed by subtracting it from the composed lumen area $A(t)$.

$$\begin{cases} A1(t) = Ac + A0e^{a(MAP-Pc(t))} & for (MAP - Pc(t)) \leq 0 \\ A1(t) = (Ac + Am) - (Am - A0)e^{-c(MAP-Pc(t))} & \\ & for\ (MAP - Pc(t)) \geq 0 \end{cases} \tag{4}$$

In equation (4), the slow varying component $A1(t)$ is derived when the arterial pressure is equal to MAP [22]. In our model MAP is set to 80 mmHg which gives a good approximation of MAP pressure for all traces.

Another objective of this work was to do all estimations at diastolic region from 80 to 20 mmHg, so it will be possible to estimate blood pressure in a range of smaller cuff pressure. To this end we use only the second equation in (2) where the arterial pressure is greater or equal to the cuff pressure. This equation is the base model used for this work, and we developed three integrated procedures based on this equation to estimate blood pressure. Each procedure has its own inputs, outputs, and objective function. All three procedures employ the least square method for optimization. Outputs are derived when all parameters inside the models are optimized with least square method. Objective function is the difference between experimental values provided by the recordings, and calculated values from our method. Objective functions are minimized to produce optimized unknown parameters as the outputs for each model. Experimental values from recordings are recorded OMW values at peak, and trough of oscillometric pulses in diastolic region. We obtained OMW from filtering the cuff deflation curve (CDC) which is the input recording signal. Butterworth band pass filter with the order of 2 and cut-off frequencies from 0.5 to 20 Hz is used to obtain OMW form CDC. OMW is proportional to the oscillating component of lumen area oscillations $A2(t)$ [23]. In (5), φ is a proportional factor which is a constant to convert $A2(t)$ to $OMW(t)$:

$$OMW(t) = \varphi \cdot [A(t) - A1(t)] \tag{5}$$

This conversion is used to obtain calculated OMW and compare it with experimental OMW for each peak or trough used in the method, and minimize the objective functions accordingly.

The first procedure estimates the compliance from peaks of OMW in diastolic region. The second procedure uses the estimated c as one of the inputs and estimates optimized values for $Am, A0, Pa_{pk}$ from peaks of OMW in diastolic region. The third procedure uses estimated $c, Am, A0$ as some of the inputs and estimates Pa_{tr} from troughs of OMW in diastolic region.

The lumen area oscillation $A(t)$ has two different exponential behaviors at systolic and at diastolic regions as shown by equations (2) which are illustrated by Fig. 1. Our method employs the diastolic region where arterial lumen area expands from $A0$ to Am, i.e., the second equation of (2).

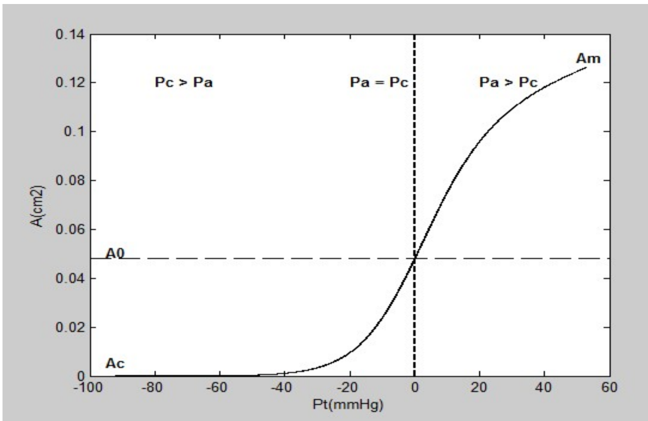

Fig. 1. Arterial lumen area versus transmural pressure for one recording

We developed an arterial blood pressure simulator from the literature [23] to show the discussed behaviors of lumen area oscillations. The simulated arterial pressure is represented by two sinusoidal harmonics with the specified default parameters [23]. This simulator is not a main onus of this paper, but the lumen area components and behaviors.

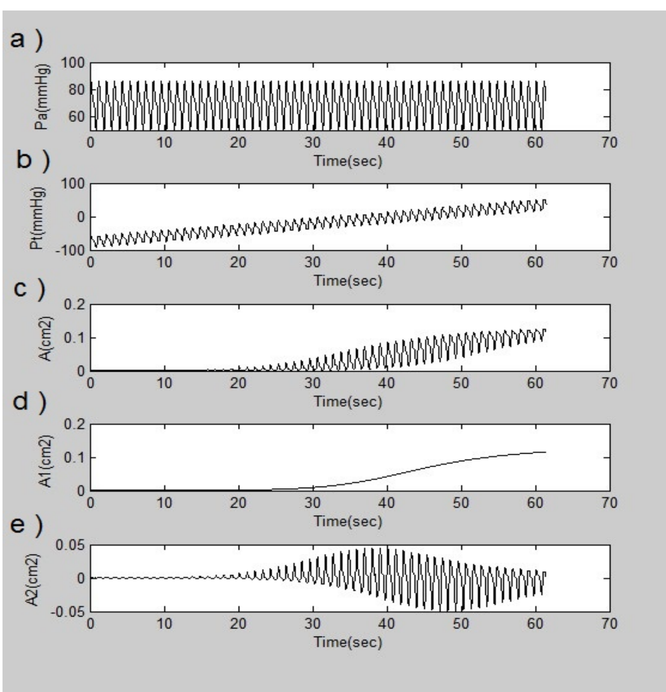

Fig. 2. Oscillometry model. (a) Simulated arterial pressure. (b) Simulated transmural pressure. (c) Simulated arterial lumen area oscillations. (d) Simulated slow varying component. (e) Simulated oscillating component

As shown in Fig. 2, transmural pressure is increasing in time as cuff pressure declines. Also, Fig. 2 shows how arterial lumen area $A(t)$ is influenced by the slow varying component $A1(t)$, and the oscillating component $A2(t)$. We obtain the oscillating component by subtracting slow varying component from the composed oscillations.

A. Dataset

The dataset employed in the experimental part comprises 150 oscillometric recordings that were obtained from 10 healthy subjects (15 recordings per subject) using an NIBP monitoring prototype that we developed, along with their corresponding NIBP estimates of *SBP* and *DBP* determined with an Omron monitor having the golden standard. The NIBP estimates made by the Omron monitor serve as reference measurements for the oscillometric recordings. Out of the 10 subjects, 6 were males while 4 were females, and their age range was 24 to 63 years. More details about this dataset can be found in our earlier published work [24].

B. Compliance estimation (procedure 1)

The artery compliance is estimated by tuning the model to best fitting the values of the peak ratios of the two successive OMW pulses in diastolic region, as presented in Fig. 3. The following equations refer to the i^{th} OMW pressure pulse:

$$OMW_i = \varphi\{\left[(Ac + Am) - (Am - A0)e^{-c(Pa - Pc_i)}\right] - \\ -[(Ac + Am) - (Am - A0)e^{-c(MAP - Pc_i)}]\}$$

where Pc_i is the cuff pressure (mmHg) at the i^{th} peak of the recorded OMW, and Pa is the respective arterial pressure (mmHg).

with $a = \frac{Ac+Am}{Am-A0}$:

$$\frac{OMW_i}{Am - A0} = \varphi\{\left[a - e^{-c(Pa - Pc_i)}\right] - \left[a - e^{-c(MAP - Pc_i)}\right]\} = \\ = \varphi\{e^{-c(MAP - Pc_i)} - e^{-c(Pa - Pc_i)}\}$$

$$\frac{OMW_{i+1}}{Am - A0} = \varphi\{e^{-c(MAP - Pc_{i+1})} - e^{-c(Pa - Pc_{i+1})}\}$$

$$OMW1_{ratio\_i} = \frac{OMW_{i+1}}{OMW_i} = \frac{e^{-c(MAP - Pc_{i+1})} - e^{-c(Pa - Pc_{i+1})}}{e^{-c(MAP - Pc_i)} - e^{-c(Pa - Pc_i)}} \quad (6)$$

$OMW2_{ratio\_i}$ = ratio of amplitude of two consecutive pulses from experiment (i^{th} and $i+1^{th}$)

$OMW1_{ratio\_i}$ values are calculated from equation (6), while their experimental equivalent $OMW2_{ratio\_i}$ are estimated from experiments for each peak of the oscillometric pulse in diastolic region. By comparing these two sets of values at each peak position we can estimate objective function OF_c (7) for each recording.

$$OF_c = \sum_{i=1}^{number\ of\ peaks-1} (OMW1_{ratio\_i} - OMW2_{ratio\_i})^2 \quad (7)$$

The least square method was used to minimize objective function OF_c to get optimized values for the compliance c and the corresponding peak arterial pressure Pa. Compliance is the main output of this procedure and is used as one of the inputs of the next two procedures.

Inputs:
a. MAP which is approximated with 80 mmHg
b. Pc_i is the cuff pressure (mmHg) at the i^{th} peak of the recorded OMW
c. Pc_{i+1} is the cuff pressure (mmHg) at i+1th peak of the recorded OMW
d. $OMW2_{ratio\_i}$ is the ratio of peaks of i^{th} and i+1th consecutive OMW pulses from the experiment
Outputs:
a. Optimized Pa which is arterial pressure (mmHg) at the peak of pulse i
b. Optimized c (mmHg^{-1}) which is the compliance of the vessel
Procedure:
1- Estimate $OMW1_{ratio\_i}$ from inputs for all peaks of the recorded OMW in diastolic region
2- Estimate $OMW2_{ratio\_i}$ from the experiment for all peaks of the recorded OMW in diastolic region
3- $OF_c = \sum_{i=1}^{number\ of\ peaks-1}(OMW1_{ratio\_i} - OMW2_{ratio\_i})^2$
4- Minimize OF_c using the least square minimization method
5- Store optimized compliance parameter c (mmHg^{-1})

Fig. 3. Compliance estimation procedure

C. SBP estimation (procedure 2)

The systolic blood pressure SBP is the maximum of the arterial blood pressure pulses and it can be approximated by the amplitude of the best fit of the peaks of the lumen area model referred above. To this extend, as shown in Fig. 4, the lumen area values $A2_{calc\_i}$ and $A2_{ex\_i}$ are calculated for each peak i in the diastolic region using the compliance parameter c found in the previous step, and the objective function (8) is minimized with least square optimization method for a given trace. Unknown parameters of the model ($Am, A0, Pa_{pk}$) are optimized and exported to the outputs. The resulted optimum peak arterial pressure Pa_{pk} is considered the best estimation of SBP of the trace.

$$OF_{pk} = \sum_{i=1}^{number\ of\ peaks} (A2_{calc\_i} - A2_{ex\_i})^2 \qquad (8)$$

Inputs:
a. MAP which is approximated with 80 mmHg
b. Pc_i which is cuff pressure (mmHg) at i^{th} peak of the recorded OMW
c. Compliance c (mmHg^{-1})
d. $Ac = 0$ which is lumen area (cm^2) when vessel is completely collapsed
Outputs:
a. Optimized Am which is maximum lumen area (cm^2)
b. Optimized $A0$ which is lumen area (cm^2) at $Pt(t) = 0$
c. Optimized Pa_{pk} at peaks equal to SBP (mmHg)
Procedure:
1- Estimate $A2_{calc\_i} = A_i - A1_i$ from inputs for all peaks of the recorded OMW in diastolic region
2- Estimate $A2_{ex\_i} = \frac{OMW_{pk\_i}}{\varphi}$ from the experiment for all peaks of the recorded OMW in diastolic region
3- $OF_{pk} = \sum_{i=1}^{number\ of\ peaks}(A2_{calc\_i} - A2_{ex\_i})^2$
4- Minimize OF_{pk} using the least square minimization method
5- Store optimized parameters $Am, A0, Pa_{pk}$
6- $SBP = Pa_{pk}$ (mmHg)

Fig. 4. SBP estimation procedure

D. DBP estimation (procedure 3)

This procedure is very similar to the SBP estimation procedure, except $A2_{calc}$ and $A2_{ex}$ are estimated from the troughs of the oscillometric wave form. Also, the objective function is different, as shown in (9). Imported parameters are $c, Am, A0$ from previous steps that are shown in Fig. 5 among other inputs of the model. After several iterations of the procedure, the unknown parameter of the lumen area model Pa_{tr} is optimized and exported as DBP of the trace.

$$OF_{tr} = \sum_{i=1}^{number\ of\ troughs} (A2_{calc\_i} - A2_{ex\_i})^2 \qquad (9)$$

Inputs:
a. MAP which approximated with 80 mmHg
b. Pc_i which is cuff pressure (cm^2) at ith trough of the recorded OMW
c. Compliance c (mmHg^{-1})
d. Optimized Am which is maximum lumen area (cm^2)
e. Optimized $A0$ which is lumen area (cm^2) at $Pt(t) = 0$
f. $Ac = 0$ which is lumen area (cm^2) when vessel is completely collapsed
Outputs:
a. Optimized Pa_{tr} at troughs equal to DBP (mmHg)
Procedure:
1- Estimate $A2_{calc\_i} = A_i - A1_i$ from inputs for all troughs of OMW in diastolic region
2- Estimate $A2_{ex\_i} = \frac{OMW_{tr\_i}}{\varphi}$ from the experiment for all troughs of OMW in diastolic region
3- $OF_{tr} = \sum_{i=1}^{number\ of\ troughs}(A2_{calc\_i} - A2_{ex\_i})^2$
4- Minimize OF_{tr} using the least square minimization method
5- Store optimized parameter Pa_{tr}
6- $DBP = Pa_{tr}$ (mmHg)

Fig. 5. DBP estimation-procedure

III. RESULTS

In this paper we have used three procedures integrated to each other by exporting/importing the required parameters to finally estimate the blood pressure non-invasively from the oscillations of the arterial lumen area underneath the cuff. The whole work is done in the diastolic region from 80 to 20 mmHg. We used OMW values from experiments for each peak and through of recorded OMW as known parameters to minimize specific objective functions. Least square method is used to minimize objective functions and export optimized values for all unknown parameters of procedures as outputs. The first procedure is used to optimize compliance of the vessel and export it to systolic and diastolic estimation procedures. The lumen area model is optimized for compliance unconditionally from the OMW peaks with the initial guess of zero for its unknown parameters (c, Pa). Lower and upper bands for these parameters are 0.01 to 0.1 ($mmHg^{-1}$) for c, and 80 to 200 (mmHg) for Pa respectively. The second procedure uses known parameters shown is Fig. 2 as inputs along with the imported compliance value from the first phase. This procedure uses only the peaks information of oscillometric waveform pulses at diastolic region and optimizes the lumen area model for $Am, A0, Pa_{pk}$ parameters. Arterial pressure Pa_{pk} is estimated from OMW peaks and is considered as SBP. Optimization is conditioned to $Am \geq 1.5 \cdot A0$ with the initial guess of zero for all unknown parameters. The lower and upper bands are 0 to 0.1 (cm^2) for both Am, and $A0$ parameters, and 80 to 200 (mmHg) for Pa_{pk} respectively. The third procedure operates very similar to the second one, but Pa_{tr} is estimated from the troughs of the OMW pulses. The model receives compliance from the first procedure, and Am, Ao parameters from the second procedure and optimizes the unknown parameter Pa_{tr} as DBP of the trace. Initial guess for Pa_{tr} is zero with lower and upper bands of 20 to 80 (mmHg). Unlike the second model, there is no condition defined between unknown parameters as there is only one unknown parameter Pa_{tr} in this procedure. The three described procedures are integrated as a single unified method and were applied on all 150 recordings. As a sample, Table I. shows the optimized values of all unknown parameters $(c, Am, A0, Pa_{pk}, Pa_{tr})$ as well as the Omron references for one recording. Results are encouraging as there are small differences between estimated blood pressure provided by our method and respective Omron references. Also, we estimated mean and standard deviation of results from model and Omron references for all 150 recordings to be able to compare them in terms of mean and standard deviation of the measured values. As shown in Table II., the differences are very small and results from the proposed method are very close to the Omron references. Moreover, we validated our method with the Omron references to investigate accuracy of the method as well. MAE is the mean absolute error or the absolute difference between the estimated values provided by the proposed method from the Omron references for each of the systolic and diastolic blood pressures averaged over 150 recordings which is shown in Table III. Furthermore, standard deviation of the absolute errors is also shown as SDE for each of the systolic and diastolic pressures accordingly.

TABLE I. MODEL RESULTS FOR A SAMPLE RECORDING

Name	Symbol	Value	Units
Omron SBP	omSBP	90	mmHg
Omron DBP	omDBP	58	mmHg
Compliance	c	0.0402	$mmHg^{-1}$
Maximum lumen area	Am	0.0368	cm^2
Lumen area at MAP	$A0$	0.0181	cm^2
Model SBP	Pa_{pk}	89.85	mmHg
Model DBP	Pa_{tr}	58.12	mmHg

TABLE II. COMPARING RESULTS BETWEEN OMRON AND MODEL

***Comparing* Results (150 Recordings)**	**Omron**		**Model**	
	SBP	***DBP***	***SBP***	***DBP***
Mean	106.65	69.85	106.42	68.96
Standard Deviation	13.21	7.37	11.33	7.42

TABLE III. ACCURACY RESULTS OF MODEL

***Accuracy* Results (150 Recordings)**	**Absolute differences from Omron**	
	SBP	***DBP***
MAE	5.13	3.18
SDE	3.60	2.58

IV. CONCLUSIONS

In this paper we have proposed a novel blood pressure estimation method consisting of three different, but integrated procedures. The proposed method is coefficient-free and is just based on the second equation of arterial lumen oscillation model in (2), because the whole process is done in diastolic region where $Pt(t) \geq 0$ or $Pa(t) \geq Pc(t)$. We followed two goals in this paper. Firstly, it is coefficient-free and not based on empirical coefficients that are used in conventional MAA algorithm. This goal increases the reliability and accuracy of the measurements accordingly, because we are not using any empirical numbers. Secondly, the proposed method is based on information extracted from peaks and troughs of the oscillometric waveform only between 80 to 20 mmHg of cuff pressure, so it can be used for continuous monitoring of blood pressures since subject is not feeling high cuff pressure around his/her arm during measurement, so subject can be monitored for a long time, even when sleeping. The whole monitoring device can be designed like a bracelet to be wrapped around the arm of the subjects. Moreover, the monitoring time is about half the standard time at the same cuff deflation rate, because cuff is inflated to about half the SSBP or even less. Accuracy of the model is another advantage as described in the results section. Not to mention that our proposed method is not using ECG at all. Table II. compares mean and standard deviation of the results of our method with the Omron references. Results from the method are very close to the Omron references in terms of mean and standard deviation of systolic and diastolic blood pressures over 150 recordings obtained from males and females between 24 to 63 years old.

Moreover, results are validated with respective Omron references, and mean absolute errors are 5.13 (mmHg) for SBP, and 3.18 (mmHg) for DBP. Also, standard deviations of absolute errors are 3.60 (mmHg) for SBP, and 2.58 (mmHg) for DBP accordingly which are very small for 150 recordings, and validates the accuracy and reliability of the proposed model.

Estimating the compliance model is the most time consuming procedure and future work will aim to finding another method that can estimate compliance from diastolic region more effectively. The ECG signal can be used to estimate compliance of the vessel based on pulse transit times from the peaks of the recorded OMW in diastolic region. This will increase robustness of the proposed method in turn. Also, future work should aim to expanding the scope of this method for sick subjects with cardiovascular diseases.

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