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An Initial Step Towards Improving the Accuracy of the Oscillometric Blood Pressure Measurement

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Abstract—**We conceived a set of physical model-based techniques to compute blood pressure (BP) from a standard oscillometric cuff pressure measurement. First, the cuff pressure measurement is represented with a physical model. Then, the model unknowns are determined from the measurement via a parametric, non-parametric or hybrid technique. Finally, the entire BP waveform is computed from the determined model. For an initial assessment, we compared the accuracy of these techniques with the conventional fixed-ratio technique for computing systolic and diastolic pressures using cuff pressure measurements simulated with an established oscillometric model over a wide parameter range. The new techniques afforded much greater accuracy than the conventional technique. Our future efforts will focus on experimental validation.**

I. Introduction

Oscillometry is perhaps the most widely used principle for measuring blood pressure (BP). This non-invasive and automatic approach determines BP using an occlusive brachial artery cuff, which acts as both an external pressure actuator and an arterial volume sensor. More specifically, as shown in Fig. 1, the cuff is inflated to a supra-systolic pressure level and then slowly deflated to a sub-diastolic pressure level. So, during the cuff deflation period, the brachial artery experiences trans-mural pressures ranging from negative to positive values. Since brachial artery compliance changes considerably near zero trans-mural pressure [1], the pulsatile amplitude of the brachial artery volume oscillation varies greatly. This variation accordingly alters the amplitude of the resulting pressure oscillation that is sensed inside the cuff, as illustrated in Fig. 1. Because the compliance of the arterial vessel becomes maximal at zero trans-mural pressure [1], mean arterial pressure (MAP) is computed as the cuff pressure at which the maximum amplitude oscillation occurs. Systolic and diastolic pressures (SP and DP) are then computed as the cuff pressures at which the amplitude of cuff pressure oscillation is some ratio of its maximum value [1]. The ratios are fixed to empirically selected values (e.g., 0.55 before the maximum amplitude oscillation occurs for SP and 0.85 after the maximum amplitude oscillation occurs for DP as shown in Fig. 1) rather than being specific to the patient at the time of measurement. As a result, this "fixed-ratio" technique is heuristic and can be very inaccurate [2-5].

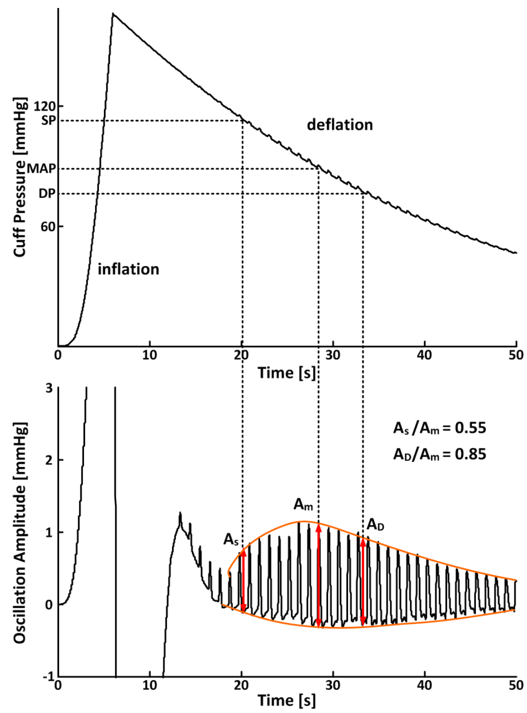

Fig. 1: Oscillometric fixed-ratio blood pressure measurement.

Our goal is to improve the accuracy of oscillometry. We conceived a set of techniques based on physical modeling of the phenomenon. More specifically, first, the standard cuff pressure measurement shown in Fig. 1 is obtained from a subject and represented with a physical model. Then, the model unknowns are determined in various ways from the measurement. Finally, SP, DP, and the entire BP waveform are computed from the determined model. In this way, the BP computation is specific to the subject at the time of measurement. For an initial assessment, we compared the accuracy of the new techniques with the conventional fixed-ratio technique using cuff pressure measurements simulated with an established model over a wide parameter range.

II. Oscillometric Model

In this study, we used the physical model of oscillometry developed by Drzewieckiet al. [1]. As we described in [6], the model is illustrated in Fig. 2 and accounts for the pressure-dependent brachial artery compliance (Arterial P-A Relationship), the compressibility of air within the cuff (Inflation/Deflation), and the nonlinear elasticity of the cuff bladder (Cuff Bladder).The inputs to the model are the brachial artery BP waveform $P_a(t)$ and the volume of air

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pumped into and out of the cuff $V_p(t)$. The output is the cuff pressure $P_c(t)$, which also acts as feedback tothe blood vessel and the cuff. The cuff volume $V_c(t)$ is defined as the difference between the external sheath volume $V_e(t)$ and the inside volume contacting the arm $V_i(t)$. Details follow.

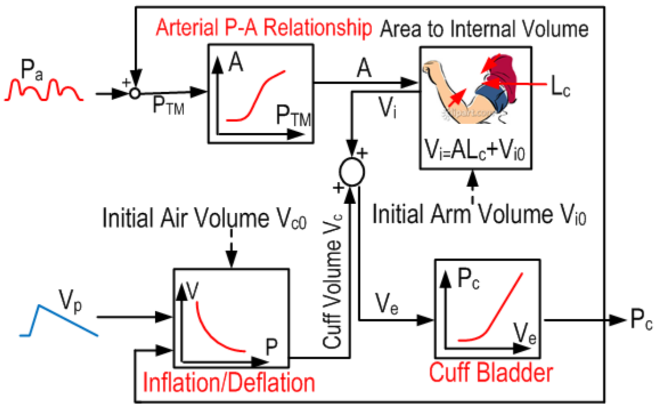

Fig. 2: Established physical model of oscillometry [1].

Arterial P-A Relationship: The cross-sectional area of the brachial artery $A(t)$ is determined via its trans-mural pressure $P_{TM}(t) = P_a(t) - P_c(t)$ as follows:

$$A(t) = d\frac{ln[aP_{TM}(t) + b]}{1 + exp[-cP_{TM}(t)]} \quad (1)$$

where a, b, c, and d are subject and time specific parameters.

Area to Internal Volume: The brachial artery area $A(t)$ is linked to the cuff via the arm volume $V_i(t)$ as follows:

$$V_i(t) = A(t)L_c + V_{i0} \quad (2)$$

where L_c is the length of the arm cuff, and V_{i0} is the initial arm volume for a collapsed brachial artery.

Cuff Bladder: The cuff pressure $P_c(t)$ is determined by the external cuff volume $V_e(t) = V_c(t) + V_i(t)$ as follows:

$$P_c(t) = E_c \cdot \{[V_e(t)/V_{eo}]^{1/n} - 1\}^n, \quad (3)$$

where E_c is the maximum cuff elastance, V_{e0} is the zero stretch volume of the bladder, and n is a nonlinearity term.

Inflation/Deflation: The cuff volume $V_c(t)$ is determined by $P_c(t)$ and $V_p(t)$ according to Boyle's law as follows:

$$P_A[V_p(t) + V_{c0}] = [P_A + P_c(t)]V_c(t) \quad (4)$$

where P_A is the atmospheric pressure, and V_{c0} is the initial air volume in the cuff.

$V_p(t)$ and $P_a(t)$: The two model inputs are defined as follows:

$$V_p(t) = \begin{cases} 81 \cdot t & 0 \le t \le 3 \\ 245 - 45 \cdot (t-3)/19 & t > 3 \end{cases} \quad (5)$$

and

$$P_a(t) = \overline{P_a} + A_0 \sin\left(\frac{2\pi f_{HR}}{60}t + \phi_1\right) + A_1 \sin\left(\frac{4\pi f_{HR}}{60}t + \phi_2\right) \quad (6)$$

where $\overline{P_a}$ is MAP, f_{HR} is heart rate in Hz, and A_0, A_1, ϕ_1, and ϕ_2 are parameters that define pulse pressure (PP = SP - DP).

For a given $V_p(t)$ and $P_a(t)$ and model parameters, $P_c(t)$ is computed by solving Eqs. (1)-(6) for each time instant using a root-finding algorithm. Drzewiecki et al. showed that the simulated cuff pressure is representative of experimental measurements despite the implicit assumption of incompressible arm tissue [1].

III. Model-Based BP Computation Techniques

As just indicated, the original purpose of the oscillometric model of Fig. 2 was to compute cuff pressure from the BP waveform and other model quantities. We conceived a set of techniques based on this model to compute the entire BP waveform from cuff pressure in an attempt to improve the accuracy of the oscillometric BP measurement. We describe each of these "inverse" techniques below.

A. Parametric Technique

The idea of this technique is to determine the unknown parameters of the physical model by fitting it to an oscillometric cuff pressure measurement. Then, the BP waveform is computed using the determined parameters.

More specifically, the following equation arises after combining Eqs. (1)-(4) and (6) and re-arranging terms:

$$\frac{1}{L_c}\left\{V_{e0}\left[\left(\frac{P_c(t)}{E_c}\right)^{1/n} + 1\right]^n - \frac{P_A}{P_A + P_c}[V_p(t) + V_{c0}] - V_{i0}\right\} = \frac{d \cdot ln\left[a\left\{\overline{P_a} + A_0 \sin\left(\frac{2\pi f_{HR}}{60}t + \phi_1\right) + A_1 \sin\left(\frac{4\pi f_{HR}}{60}t + \phi_2\right) - P_c\right\} + b\right]}{1 + \exp\left[-c\left\{\overline{P_a} + A_0 \sin\left(\frac{2\pi f_{HR}}{60}t + \phi_1\right) + A_1 \sin\left(\frac{4\pi f_{HR}}{60}t + \phi_2\right) - P_c\right\}\right]} \quad (7)$$

Here, L_c, V_{e0}, E_c, n, and V_{c0} are determined from *a priori* measurements on the employed cuff; $V_p(t)$ is the volume of air pumped into and out of the cuff; P_A is atmospheric pressure; $P_c(t)$ is measured via a sensor inside the cuff; f_{HR} is measured from the oscillations in $P_c(t)$; and $\overline{P_a}$ is measured as the value of $P_c(t)$ at which the maximum amplitude oscillation occurs. Hence, all these quantities are known. However, the parameters $, b, c, d$, A_0, A_1, ϕ_1, and ϕ_2 are subject and time specific and thus unknown. Note that $\overline{P_a}$ may also be regarded as an unknown parameter.

The parameters are determined by matching both sides of Eq. (7) to each other during the deflation period using a least squares search over a physiologic parameter range. Finally, $P_a(t)$ is determined via Eq. (6), and SP and DP are then given as the maximum and minimum of $P_a(t)$.

B. Non-Parametric Technique

The idea of this technique is to reduce the assumptions underlying the physical model by employing non-parametric models of the Arterial P-A Relationship and the BP waveform. Then, these non-parametric models are determined from an oscillometric cuff pressure measurement. A trade-off in using non-parametric models instead of parametric ones is reduced robustness to noise.

More specifically, the following equation arises by substituting the left-hand side of Eq. (1) into Eq. (7):

$$A(t) = \frac{1}{L}\left(V_{e0}\left\{\left[\frac{P_c(t)}{E_c}\right]^{1/n} + 1\right\}^n - \frac{P_A}{P_A + P_c}[V_p(t) + V_{c0}] - V_{i0}\right) \quad (8)$$

Based on this equation, $A(t)$ is solved from the known cuff parameters and known $P_c(t)$ and $V_p(t)$. Then, as shown in Fig. 3a, the Arterial P-A Relationship is obtained to within a horizontal offset by identifying the upper and lower envelopes of the plot relating $A(t)$ to $-P_c(t)$ and then averaging the two envelopes. Next, as indicated in Fig. 3b, the envelope is exactly determined by horizontally shifting it so that the peak derivative is located at zero trans-mural pressure. Thereafter, the resulting Arterial P-A Relationship is applied to compute $P_{TM}(t)$ from $A(t)$. Finally, $P_c(t)$ is added to $P_{TM}(t)$ to yield $P_a(t)$. SP and DP are then determined as the maximum and minimum of $P_a(t)$. In this way, the Arterial P-A Relationship and BP waveform are obtained without assuming any model (i.e., Eqs. (1) and (6)).

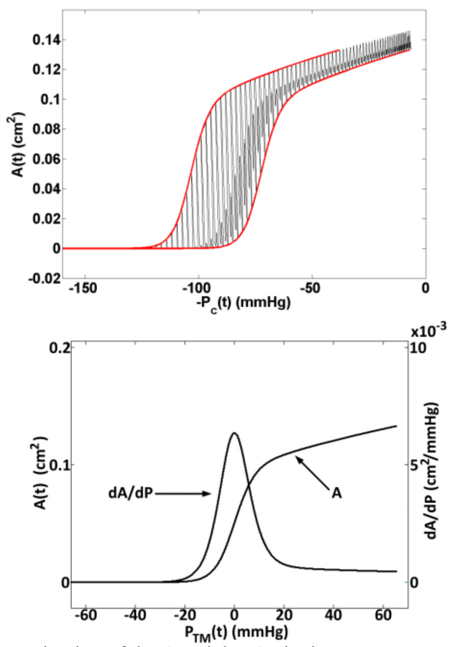

Fig. 3: Determination of the Arterial P-A via the non-parametric technique.

A key assumption is that the peak derivative of the Arterial P-A Relationship is located at zero trans-mural pressure. This assumption may not always be valid. For example, the assumption breaks down when the c parameter is very small, which corresponds to severe arterial stiffening in the neighborhood of zero trans-mural pressure. To mitigate this problem, the computed $P_a(t)$ is vertically shifted so that its mean value is equal to the $P_c(t)$ at which the maximum amplitude oscillation occurs (i.e., standard MAP determination).

C. Hybrid Technique

The idea of this technique is to combine components of the parametric and non-parametric techniques so as to eliminate the error introduced by the standard MAP determination in the non-parametric technique while preserving benefits of this technique. Then, the "hybrid" model is determined from an oscillometric cuff pressure measurement.

More specifically, $A(t)$ is solved from the known cuff parameters and known $P_c(t)$ and $V_p(t)$ according to Eq. (8). Then, the Arterial P-A Relationship is obtained to within a horizontal shift as described above and shown in Fig. 3a. Next, this relationship is represented with the parametric model described above as follows:

$$A(t) = d\,\frac{\ln\left[a\left(SP - P_c(t)\right) + b\right]}{1 + \exp\left[-c\left(SP - P_c(t)\right)\right]} \qquad \textit{for upper envelope} \tag{9}$$

$$A(t) = d\,\frac{\ln\left[a\left(DP - P_c(t)\right) + b\right]}{1 + \exp\left[-c\left(DP - P_c(t)\right)\right]} \qquad \textit{for lower envelope} \tag{10}$$

Here, $A(t)$ and $P_c(t)$ are known, while $a, b, c, d,$ and SP and DP are unknown. These unknown parameters are determined by matching both sides of Eq. (9) and (10) to each other during the deflation period using a least squares search over a physiologic parameter range. Finally, the two resulting Arterial P-A Relationships are averaged and then applied to compute $P_a(t)$ from $A(t)$ as described above.

IV. Technique Evaluation

For an initial evaluation, we applied the non-parametric and hybrid techniques to cuff pressure measurements simulated with the oscillometric model of Fig. 2. More specifically, we simulated $P_c(t)$ via Eqs. (1)-(6) using the nominal model parameter valuess provided in Drzewiecki et al. [1]. We then applied the techniques to $P_c(t)$ to compute the BP waveform. We next determined the SP and DP errors using the known values of SP and DP. We repeated this process for a physiologic range of values of PP and the c parameter, as we previously showed that variations in these two quantities had the greatest impact on the accuracy of the fixed-ratio technique [6].

V. Results

Fig. 4 shows the SP and DP errors of the non-parametric and fixed-ratio techniques as a function of PP and the c parameter. The conventional technique showed absolute errors as high as 57 mmHg, whereas the new technique revealed absolute errors within just 12 mmHg. Fig. 5 shows an example of the BP waveform computed by the non-parametric technique (red) and the actual BP waveform (blue). Note that the fixed-ratio technique is not able to yield a BP waveform. The error in the non-parametric technique was due to the error in the standard MAP determination. Indeed, the hybrid technique determined SP, DP, and the BP waveform without error over the entire PP and c parameter range (results not shown). We anticipate that the parametric technique would likewise be able to achieve perfect BP determination here.

VI. Conclusion

The oscillometric fixed-ratio technique is widely used for measurement of SP and DP. However, this technique is heuristic and has thus proven to be error prone. We conceived a set of techniques based on physical modeling to improve the accuracy of SP and DP, while also providing the entire BP waveform, from a standard oscillometric cuff pressure measurement. As a first step, we tested these techniques using measurements simulated with a computer model without any systematic or measurement error. Our results showed that the techniques afforded much greater accuracy than the fixed-ratio technique.

Our future efforts will surely focus on experimental testing of the techniques. While the hybrid (and parametric) techniques performed best in this theoretical study, it is possible that the non-parametric technique will prove more efficacious experimentally due to its fewer underlying assumptions. If experimental testing proves successful, then BP may be more accurately measured by only altering the software component of a standard arm cuff system while maintaining its hardware component. Further, the added capability of providing the entire BP waveform may allow the convenient system to also track cardiac output via pulse contour analysis.

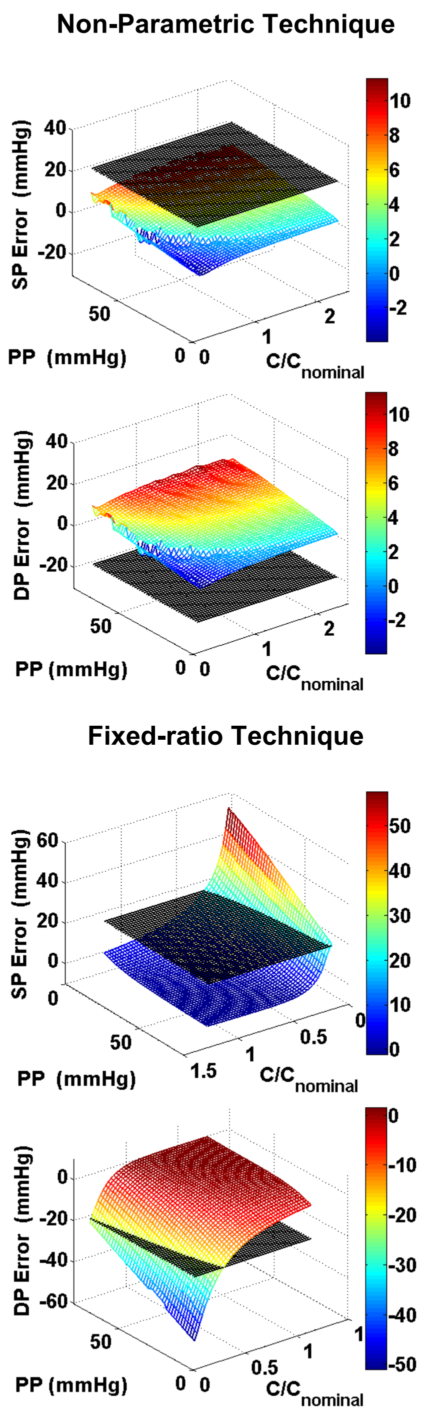

Fig. 4: Performance of the non-parametric and fixed-ratio techniques on simulated cuff pressure data. (The black planes correspond to + or – 20 mmHg error).

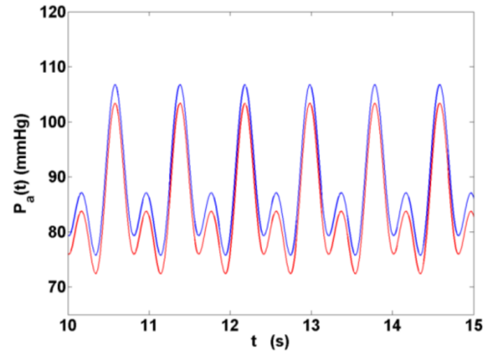

Fig. 5: Example of the actual (simulated blood pressure waveform (blue) and the waveform computed via the non-parametric technique (red).

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