Annals of Biomedical Engineering, Vol. 41, No. 3, March 2013 (© *2012*) pp. 587–597
DOI: 10.1007/s10439-012-0700-7

Error Mechanisms of the Oscillometric Fixed-Ratio Blood Pressure Measurement Method

JIANKUN LIU,[1] JIN-OH HAHN,[2] and RAMAKRISHNA MUKKAMALA[1]

[1]Department of Electrical and Computer Engineering, Michigan State University, 428 S. Shaw Lane, Rm. 2120 Engineering Building, East Lansing, MI 48824-1226, USA; and [2]Department of Mechanical Engineering, University of Maryland, College Park, MD, USA

(*Received 23 July 2012; accepted 8 November 2012; published online 21 November 2012*)

Associate Editor Joan Greve oversaw the review of this article.

Abstract—The oscillometric fixed-ratio method is widely employed for non-invasive measurement of systolic and diastolic pressures (SP and DP) but is heuristic and prone to error. We investigated the accuracy of this method using an established mathematical model of oscillometry. First, to determine which factors materially affect the errors of the method, we applied a thorough parametric sensitivity analysis to the model. Then, to assess the impact of the significant parameters, we examined the errors over a physiologically relevant range of those parameters. The main findings of this model-based error analysis of the fixed-ratio method are that: (1) SP and DP errors drastically increase as the brachial artery stiffens over the zero trans-mural pressure regime; (2) SP and DP become overestimated and underestimated, respectively, as pulse pressure (PP) declines; (3) the impact of PP on SP and DP errors is more obvious as the brachial artery stiffens over the zero trans-mural pressure regime; and (4) SP and DP errors can be as large as 58 mmHg. Our final and main contribution is a comprehensive explanation of the mechanisms for these errors. This study may have important implications when using the fixed-ratio method, particularly in subjects with arterial disease.

Keywords—Arm cuff, Mathematical model, Non-invasive blood pressure monitoring, Oscillometry, Sensitivity analysis.

INTRODUCTION

Oscillometry is a popular method for non-invasive monitoring of blood pressure. This method determines systolic, diastolic, and mean arterial pressures using an occlusive brachial artery cuff, which acts as both an external pressure applicator and an arterial volume sensor[4,5,8,9,15] More specifically, as shown in Fig. 1a, the cuff is inflated to a supra-systolic pressure level (e.g., 180 mmHg) and then slowly deflated to a sub-diastolic pressure level (e.g., 50 mmHg). So, during the deflation period, the brachial artery experiences trans-mural pressures ranging from negative to positive values. Since brachial artery compliance changes considerably around zero trans-mural pressure,[4] the amplitude of the brachial artery volume oscillation (due to the heart beat) varies greatly. This variation accordingly alters the amplitude of the resulting pressure oscillation that is sensed inside the cuff, as illustrated in Fig. 1b. Because the compliance of the arterial vessel becomes maximal when unloaded (i.e., at zero trans-mural pressure),[4] mean arterial pressure (MAP) is determined as the cuff pressure at which the maximum amplitude oscillation occurs. Systolic and diastolic pressures (SP and DP) are then determined as the cuff pressures at which the amplitude of cuff pressure oscillation is some ratio of its maximum value.[5] Due to the absence of a systematic method, the ratios are fixed to empirically selected values. As a result, the oscillometric fixed-ratio blood pressure measurement method is heuristic and susceptible to nontrivial errors arising from a number of factors.

A few investigators have used mathematical models of oscillometry to test factors that could affect the accuracy of the fixed-ratios method.[4,10,13] Drzewiecki *et al.*[4] varied SP and DP in their model over a wide range and showed that the fixed-ratio method largely maintained its accuracy. In a more comprehensive study, Ursino and Cristalli[13] varied several model parameters and found that alterations in arterial properties and pulse pressure (PP) affected the accuracy, leading to SP and DP errors as great as 15–20%.

Address correspondence to Ramakrishna Mukkamala, Department of Electrical and Computer Engineering, Michigan State University, 428 S. Shaw Lane, Rm. 2120 Engineering Building, East Lansing, MI 48824-1226, USA. Electronic mail: rama@egr.msu.edu

Jiankun Liu and Jin-Oh Hahn are equally contributing first authors.

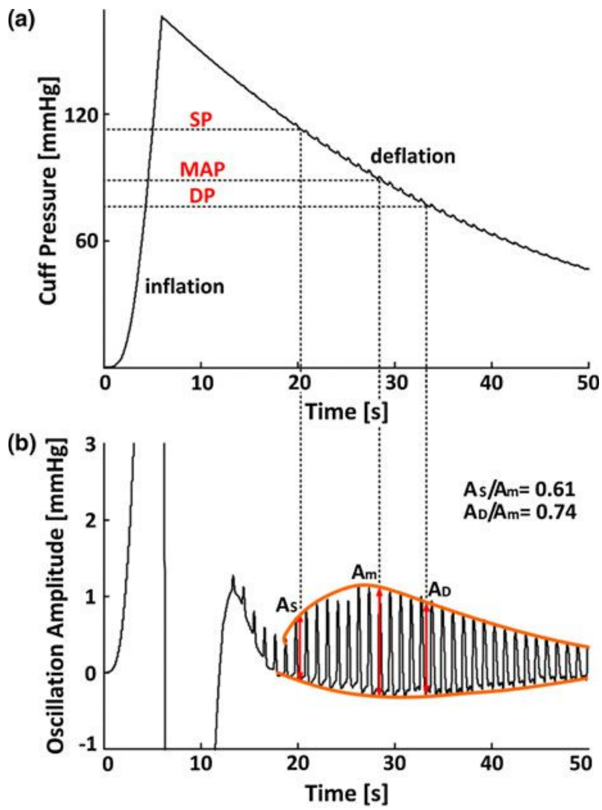

FIGURE 1. Oscillometric method for non-invasive blood pressure measurement. (a) Cuff pressure during cuff inflation and deflation. (b) Estimation of SP, DP, and MAP from cuff pressure oscillations *via* fixed-ratio method and maximum oscillation amplitude.

More recently, Raamat *et al.*[10] studied the effect of the arterial pressure waveform and the artery-cuff pressure-volume relationship in their model to conclude that alterations in these factors induced similar errors. These model-based studies have allowed the fixed-ratios method to be assessed in a manner that could not be achieved experimentally and have thus significantly contributed to the understanding of its potential pitfalls. However, the reported model-based errors appear smaller than those observed in practice (e.g., 40 mmHg for SP and 25 mmHg for DP)[1,3,7,12,14] perhaps due to the limited number of model parameters varied or the narrow parameter range studied. Further, to the best of our knowledge, neither these studies nor others have revealed the mechanisms for the errors of the fixed-ratios method.

In this study, we investigated the errors of the oscillometric fixed-ratio blood pressure measurement method based on a mathematical model. First, we determined the factors that significantly affect its accuracy *via* a thorough parametric sensitivity analysis. Then, we showed that the errors could be much higher than those reported by previous model-based studies through a realistic range of parameter values. Lastly and most significantly, we unveiled the mechanisms for these errors.

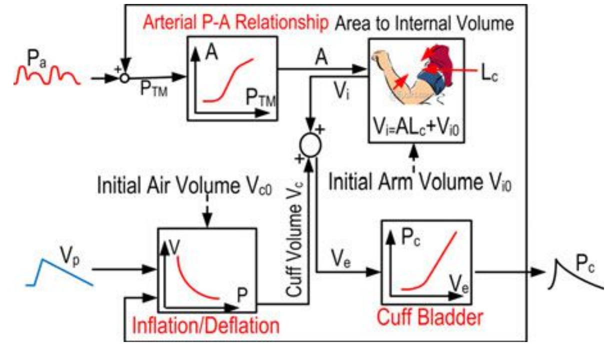

FIGURE 2. Functional block diagram of the mathematical model of oscillometry[4] used herein to investigate the accuracy of the non-invasive blood pressure measurement method. *A*: arterial lumen area; P_a: brachial arterial pressure; P_{TM}: arterial transmural pressure; L_c: length of cuff; P_c: cuff pressure; V_c: cuff volume; V_{c0}: initial cuff volume; V_e: cuff outer volume; V_i: cuff inner volume; V_{i0}: initial cuff inner volume; V_p volume of air pumped into cuff.

MATERIALS AND METHODS

Oscillometric Model

To reproduce the oscillometric measurement and study the root cause of its errors, we used the established mathematical model of Drzewiecki *et al.*[4] 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 as dictated by Boyle's law (Inflation/Deflation), and the deformation and stretch of the cuff bladder *via* a nonlinear pressure-volume relationship (Cuff Bladder). The inputs to the model are the brachial artery pressure waveform (P_a) and the volume of air pumped into and out of the cuff (V_p). The output is the cuff pressure (P_c), which also acts as feedback to both the blood vessel and the cuff. In this model, the cuff volume (V_c) is defined as the difference between the external sheath volume (V_e) and the inside volume contacting the arm (V_i). Details follow.

Arterial P-A Relationship: The cross-sectional area of the brachial artery (A) is determined *via* its transmural pressure (i.e., the difference between arterial pressure and cuff pressure ($P_{TM} = P_a - P_c$) according to the following nonlinear relationship:

$$A = d\frac{\ln(aP_{TM} + b)}{1 + \exp(-cP_{TM})} \tag{1}$$

where a, b, c and d are subject-dependent parameters (see the end of this section for a description of the effect of each of these parameters on the brachial artery compliance curve).

Area to Internal Volume: The brachial artery area A is linked to the cuff through the volume of the arm V_i as follows:

$$V_i = AL_c + V_{i0} \tag{2}$$

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 is determined by the external cuff volume, which is the sum of the cuff volume and arm volume (i.e., $V_e = V_c + V_i$), according to the following nonlinear relationship:

$$P_c = E_c \cdot \left[(V_e/V_{eo})^{1/n} - 1\right]^n \tag{3}$$

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

Inflation/Deflation: The cuff volume V_c is determined by the cuff pressure P_c and the pumped volume into and out of the cuff V_p according to Boyle's law as follows:

$$P_A(V_p + V_{c0}) = (P_A + P_c)V_c \tag{4}$$

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

V_p and P_a: The two model inputs are defined in terms of the following equations:

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

and

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

where $\overline{P_a}$ is MAP, f_{HR} is heart rate (HR) in Hz, and A_0, A_1, and ϕ_1 are parameters defining PP. For a given V_p and P_a, the cuff pressure P_c is computed by simultaneously solving the above equations for each time instant using a root-finding algorithm (FZERO routine in MATLAB). See Drzewiecki *et al.*[4] for additional model details including all parameter values.

The effect of the parameters a, b, c, and d on the brachial artery compliance curve ($C_a(P_{TM}) = dA/dP_{TM}$) is graphically shown in Fig. 3. The parameter a slightly shifts the location of the peak of the compliance curve, but its effect is not significant. By contrast, the parameters b and c play a crucial role in contributing to the compliance curve. On one hand, decreasing b and c results in a decrease in the amplitude of the curve in the neighborhood of zero trans-mural pressure. On the other hand, their effects are distinct outside of this range. In particular, decreasing b has different impact on the compliance curve in the negative and positive trans-mural pressure regimes. In the negative trans-mural pressure regime, decreasing b mostly shifts the compliance curve to the right without changing the width of the curve itself. In the positive trans-mural pressure regime, decreasing b mostly yields an increase in the width of the compliance curve (see the 30 mmHg trans-mural pressure range in Fig. 3b, where the slopes of the curves become reversed), with a slight shift of the curve to the right. In contrast to b, decreasing c results in a widening of the compliance curve over the entire pressure range. Finally, the parameter d acts simply as a scale factor. By defining arterial stiffness as the change in pressure divided by the change in area in the neighborhood of zero trans-mural pressure, arterial stiffness increases as b, c, and d decrease. It is important to emphasize that decreasing b and c increases arterial stiffness as defined here by both reducing the amplitude of the curve and expanding its width, whereas decreasing d enhances the arterial stiffness only by reducing the amplitude of the curve.

Error Analysis

To reveal the major factors affecting the accuracy of the oscillometric fixed-ratio method, we carried out a parametric sensitivity analysis. We specifically investigated the following seven parameters as candidate factors: PP, MAP, HR, and the brachial artery compliance parameters a, b, c, and d. We did not include the parameters defining the occlusive cuff in the analysis, as they can be regarded as fixed once a particular cuff is chosen. (Note that while the results are presented here for only the bladder cuff in Drzewiecki *et al.*,[4] the results were highly comparable for both the bladder and the Critikon Duracuf cuffs in Drzewiecki *et al.*[4].)

We studied the effects of the seven model parameters by quantifying the changes in the oscillometric blood pressure measurement errors caused by ±50% variations in each parameter. In particular, we first used the nominal model parameter values provided in Drzewiecki *et al.*,[4] which define the brachial artery pressure waveform, the brachial artery compliance curve, and the occlusive cuff, to simulate the cuff pressure output. We then applied the fixed-ratio method and the maximum oscillation amplitude to the simulated output to estimate SP, DP, and MAP. Finally, we computed the errors between these estimated blood pressure values and the corresponding blood pressure values that were used to simulate the cuff pressure output. We repeated this process with +50% and -50% variations to each of the seven parameters. We specifically implemented changes in MAP and HR by directly altering $\overline{P_a}$ and f_{HR} and in PP by multiplying a factor of 1.5 (for 50% increase)

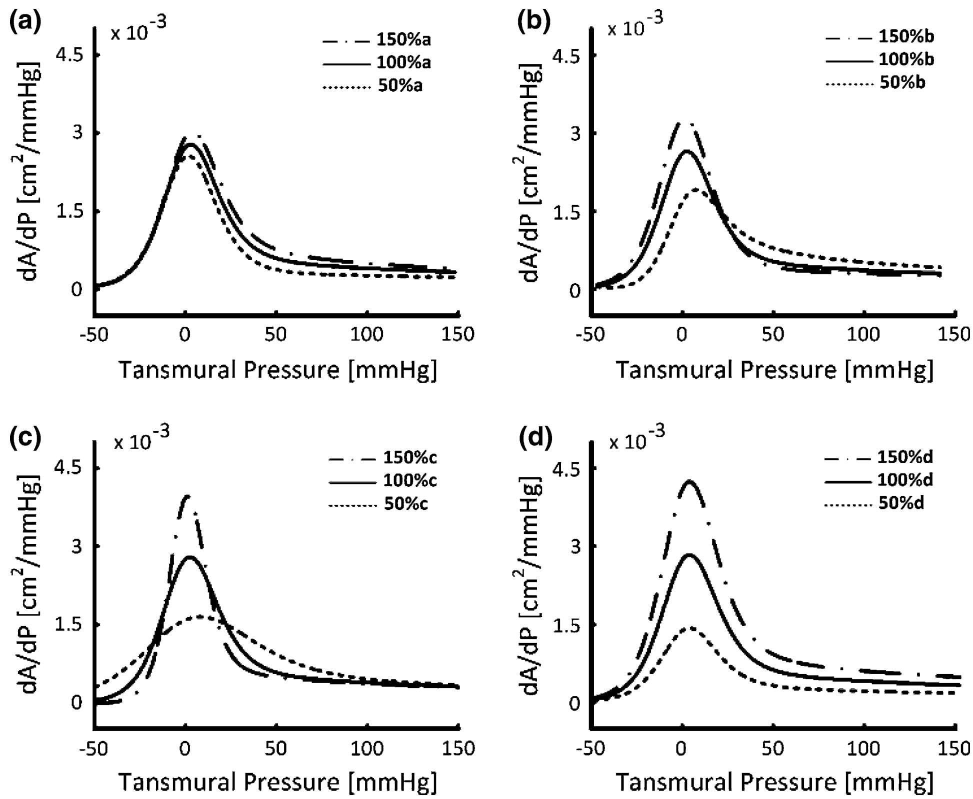

FIGURE 3. Model brachial artery compliance curve under different parameter values (see Eq. (1)).

and 0.5 (for 50% decrease) to A_0 and A_1. We quantified the impact of each parameter on the blood pressure errors in terms of the percent change in error caused by the parametric variation. Throughout the analysis, we used fixed-ratios of 0.61 for estimating SP and 0.74 for estimating DP, as these values yielded small SP and DP errors for the nominal model parameter values.

Finally, to scrutinize the impact of the significant parameters, we examined the blood pressure errors over physiologically relevant ranges of those parameters.

RESULTS

Parametric Sensitivity Analysis

Table 1 summarizes the results of the parametric sensitivity analysis. There were appreciable changes in SP and DP errors but not MAP errors.

Recall that MAP is determined as the cuff pressure at which the maximum amplitude oscillation occurs on the basis that the compliance of the brachial artery is maximal at zero trans-mural pressure. Hence, since alterations in *a*, *b*, *c*, and *d* hardly changed the location of the peak of the arterial compliance curve (see above), MAP accuracy was maintained despite the major parametric variations.

The model parameters, PP, *b*, and *c*, had great impact on SP and DP errors. As PP decreased, the fixed-ratio method overestimated SP and underestimated DP. This finding is consistent with other studies.[1] As *b* and *c* declined, which increased the width and decreased the amplitude of the compliance curve, significant errors were incurred. In particular, both SP and DP were underestimated as *b* decreased, while SP was overestimated and DP was underestimated as *c* decreased. In contrast, the compliance parameter *d* (as well as the remaining model parameters) hardly impacted the errors. Hence, increasing arterial stiffness (see definition above) specifically *via* an increase in the width of the compliance curve was a crucial factor.

Error Magnitudes

Figure 5 illustrates the SP and DP errors over a physiologic range of PP and *c*. A physiologic range for the latter parameter was determined based on the human aorta data at autopsy shown in Fig. 4. The SP and DP errors gradually decreased as PP increased, whereas these errors dramatically increased as arterial stiffness increased (*via* a decrease in *c*). The errors were as high as 58 mmHg in the case of severe arterial stiffening. Such errors are consistent with the experimental investigations of Coleman *et al.*[1] who reported errors up

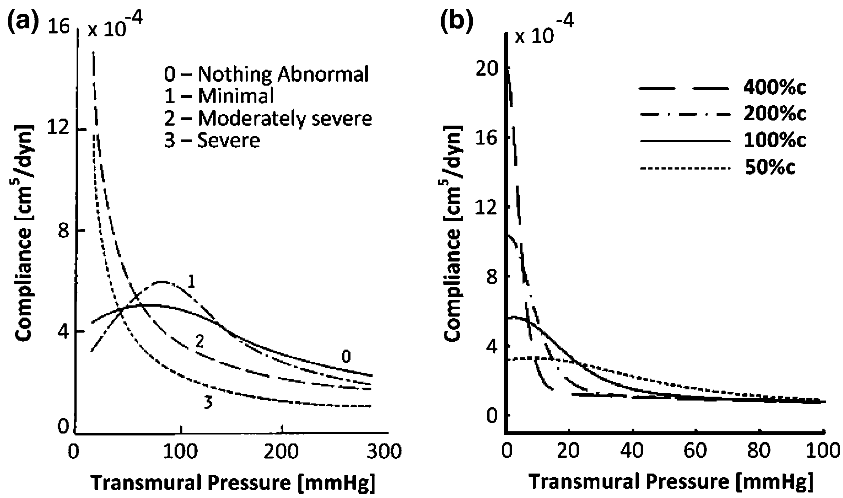

FIGURE 4. (a) Experimental brachial artery compliance curves determined from humans with varying degrees of atherosclerosis at autopsy[11] and (b) corresponding model brachial artery compliance curves determined by varying *c*.

TABLE 1. Results of the parametric sensitivity analysis.

Oscillometric error	PP +50%	PP −50%	MAP +50%	MAP −50%	HR +50%	HR −50%	*a* +50%	*a* −50%	*b* +50%	*b* −50%	*c* +50%	*c* −50%	*d* +50%	*d* −50%
SP	−9.37	+14.1	−3.02	+3.33	+0.33	−0.62	+3.60	+3.60	+3.93	−16.8	−4.72	+21.9	+0.40	−0.40
DP	+6.99	−9.93	−1.75	+5.73	−1.34	+3.90	+3.51	+3.51	+3.72	−14.1	+4.77	−19.5	−0.02	−0.10
MAP	+0.53	−0.92	−0.14	+0.44	−0.01	−0.13	+0.15	+0.38	+0.41	−1.59	+0.13	−0.55	+0.03	−0.00

Values represent the difference between the oscillometric error upon indicated parametric perturbation and the oscillometric error under the nominal parameters divided by the oscillometric error under the nominal parameters.

to 35 mmHg for the OMRON MX3 Plus device and Greeff *et al.*[3] who reported errors up to 40-45 mmHg for the OMRON MIT and OMRON M7 devices.

DISCUSSION

The oscillometric fixed-ratio method is widely employed for non-invasive blood pressure measurement. Its accuracy under nominal physiologic conditions is supported by several studies.[7,14] However, our model-based analysis shows that the method becomes susceptible to major SP and DP errors when arterial stiffness (defined as the change in brachial artery pressure divided by the change in brachial artery area in the neighborhood of zero trans-mural pressure) and PP deviate significantly from nominal levels. While these findings are consistent with the previous model-based analysis of Ursino and Cristalli[13] (despite the use of different models), the error magnitudes shown here are markedly higher due to our exploration of a wider, realistic range of model parameter values. Further, we describe the mechanisms for these errors below.

Error Mechanisms for Arterial Stiffness Variations

Variations in arterial stiffness *via* the model parameters b and c, which alter the shape of the brachial artery compliance curve (width in particular), significantly impact the accuracy of the fixed-ratio method. As exemplified for changes to c in Fig. 6, both parameters dictate the shape of the envelope of the cuff pressure oscillations (normalized to unity), which is the key factor in determining the accuracy of the method. So, even though the actual SP and DP are invariant in the figure, the fixed-ratio predictions (indicated with solid lines) deviate from these pressures with perturbations to c from its nominal value. By contrast, variations in arterial stiffness *via* the model parameter d, which alters the amplitude of the brachial artery compliance curve, have little impact on the accuracy, because the shape of the envelope of the cuff pressure oscillations hardly depends on the curve amplitude (not shown). So, the width of the compliance curve normalized by its maximum amplitude in the neighborhood of zero trans-mural pressure specifically represents the important factor affecting the accuracy of the fixed-ratio method.

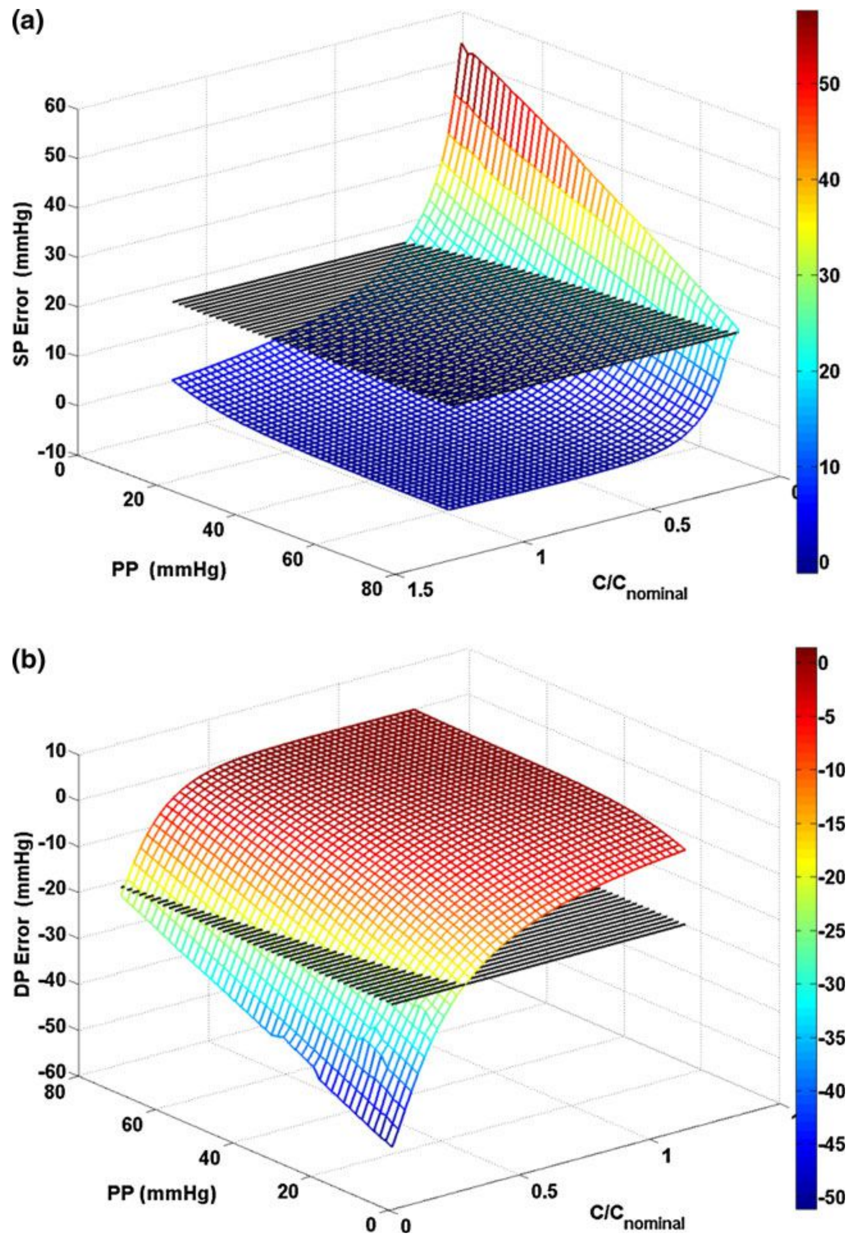

FIGURE 5. (a) SP and (b) DP errors of the fixed-ratio method as a function of *c* and PP. Black plane indicates error of ±20 mmHg.

Figure 7 illustrates the error mechanism when changes to c occur. From Fig. 6, the width of the cuff pressure oscillation envelope increases with decreasing c, which corresponds to arterial stiffening. For reduced c, the fixed-ratios predict higher SP and lower DP (see Fig. 7b), thereby overestimating SP and underestimating DP. On the other hand, for increased c, the fixed-ratios predict lower SP and higher DP (see Fig. 7c), resulting in underestimated SP and overestimated DP. Figure 7 also indicates that the cuff pressure oscillation envelope becomes flatter as c decreases. Hence, even small perturbations to the fixed-ratio values can yield large errors in SP and DP. In contrast, the envelope becomes steeper as c increases. So, the blood pressure errors are less sensitive to changes in the fixed-ratio values. In short, a reduction in c, which corresponds to arterial stiffening, is particularly problematic and results in overestimation of SP and underestimation of DP.

The error mechanism when changes to b occur is distinct due to the differing effect of b and c on the brachial artery compliance curve (see Fig. 3). First, recall that decreasing b in the negative trans-mural pressure regime, which corresponds to the region where SP is determined, shifts the compliance curve to the right without changing its width. Thus, the fixed-ratio

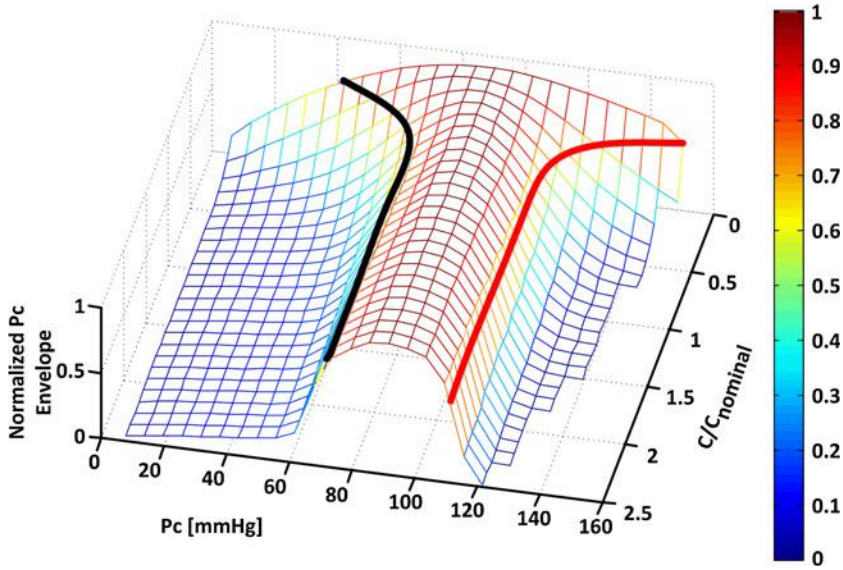

FIGURE 6. Envelope of the model cuff pressure (P_c) oscillations (normalized to unity) as a function of *c*. Solid lines indicate SP (red) and DP (black) estimated by the fixed-ratio method.

predicts a lower SP, thereby underestimating this pressure. Second, recall that decreasing b in the positive trans-mural pressure regime, which corresponds to the region in which DP is determined, increases the width of the compliance curve while also slightly shifting the curve to the right. Hence, the fixed-ratio predicts a lower DP, resulting in underestimated DP. Further, for similar reasons as c, decreasing b, which corresponds to arterial stiffening, is far more problematic than increasing this parameter.

Error Mechanisms for PP Changes

In addition to arterial stiffening, changes to PP significantly impact the accuracy of the fixed-ratio method. Figure 8 shows the envelope of the cuff pressure oscillations (again normalized to unity) for a range of PP values and three different values of c representing compliant (Fig. 8a), normal (Fig. 8b), and stiff (Fig. 8c) arteries. As PP increases, the actual SP and DP move away from MAP. The fixed-ratio predictions (indicated with solid lines) correctly show a tendency to likewise deviate from MAP. However, this tendency decreases markedly with increasing arterial stiffness (e.g., compare Figs. 8a–8c). Further, the fixed-ratio predictions appear more erroneous over the low PP range than the high PP range (see Fig. 8 where the red and black solid lines appear closer than they should be for the lower range of PP). Hence, smaller PP, particularly in combination with stiffer arteries, compromises the accuracy of the fixed-ratio method. As a final point, Fig. 8 also indicates that the value of PP in addition to the degree of arterial stiffening dictate the shape of the envelope of the cuff pressure oscillations.

Figure 9 illustrates the error mechanism when changes to PP occur. To describe this mechanism, it is important to first reiterate that the amplitude of the cuff pressure oscillations is determined by the amplitude of the brachial artery area oscillations. In fact, the two variables are proportional to each other in the mathematical model herein. The amplitude of the brachial artery area oscillations (δA) arises according to the following equation:

$$\delta A = \int_{\mathrm{DP}-P_c}^{\mathrm{SP}-P_c} dA = \int_{\mathrm{DP}-P_c}^{\mathrm{SP}-P_c} \frac{dA}{dP} dP = \int_{\mathrm{DP}-P_c}^{\mathrm{SP}-P_c} C_a(P_{\mathrm{TM}}) dP \qquad (7)$$

Hence, PP determines the width of the brachial artery compliance curve $C_a(P_{\mathrm{TM}})$ that is integrated to establish the amplitude of the brachial artery area oscillations and thus the amplitude of the cuff pressure oscillations. Computing this area at each cuff pressure level (see green rectangles in left panels of Fig. 9) and plotting the resulting areas against the cuff pressures yield the cuff pressure envelope (see right panels of Fig. 9). In this way, PP impacts the shape of the cuff pressure envelope and thus the accuracy of the fixed-ratio method.

For normal or compliant brachial arteries (see Fig. 9a), the area difference across the cuff pressure levels is large for small PP (see upper, left panel) and small for large PP (see lower, left panel). The reason is that large PP will include area under the central part of

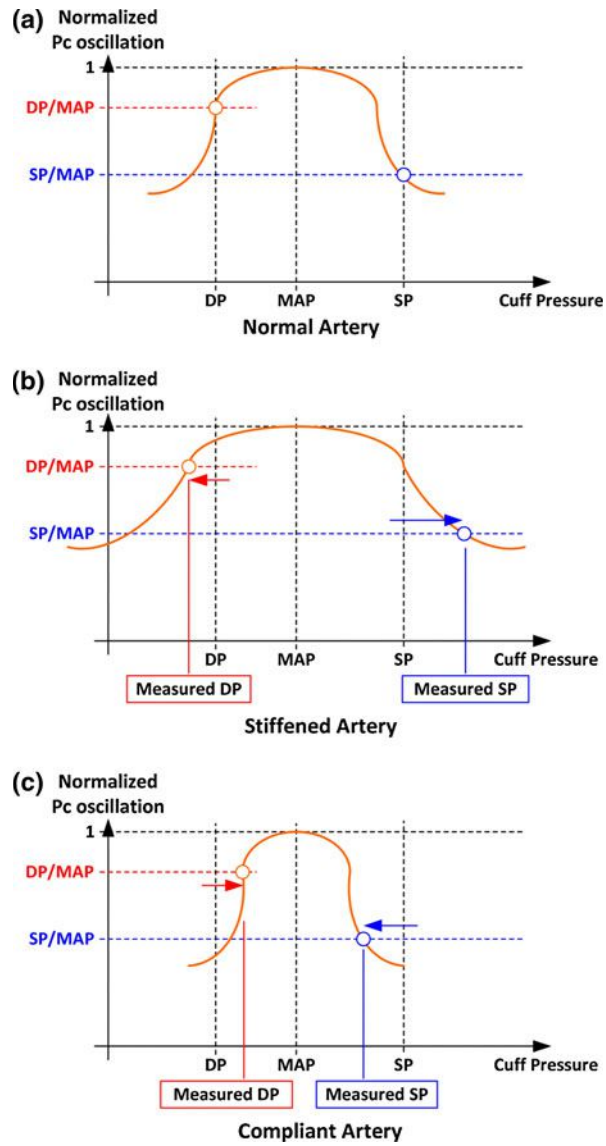

FIGURE 7. Error mechanism of the fixed-ratio method for changes in arterial stiffness at zero trans-mural pressure.

the brachial artery compliance curve, which is responsible for the majority of its total area, for many cuff pressure levels, whereas small PP will not. Hence, as indicated in Fig. 9a as well as in Figs. 8a and 8b, the cuff pressure oscillation envelope becomes flatter as PP increases and steeper as PP decreases. On the other hand, for stiff brachial arteries (see Fig. 9b), the area difference across the cuff pressure levels is relatively small, because the brachial artery compliance curve is inherently flat (compare the left panels in Figs. 9a and 9b). As a result, as indicated in Fig. 9b as well as in Fig. 8c, the impact of PP on the cuff pressure oscillation envelope is relatively small. For these reasons, as indicated in Fig. 5, the worst-case scenario for the fixed-ratio method is when PP is low and the brachial artery is stiff. In this scenario, SP will be markedly overestimated and DP will be significantly underestimated (see upper, right panel of Fig. 9b). However, when the brachial artery is normal or compliant, the fixed-ratio method may be more effective against PP changes, yet far from impervious, due to greater responsiveness of the cuff pressure oscillation envelope to PP (see right panels of Fig. 9a).

Study Limitations

The main limitation of this study is that the mathematical model that we employed ignored the effect of arm tissue mechanics. More precisely, the model assumed incompressible arm tissue. We settled upon this model, despite this assumption, for the following reasons. First, the model was shown to be able to generate realistic oscillometric cuff pressure waveforms.[4] Second, the model offered a simple foundation upon which error mechanisms could be elucidated. Third, the model turned out to afford similar parametric sensitivity results as the model of Ursino and Cristalli,[13] which did account for arm tissue mechanics. That said, Ursino and Cristalli[13] did not include variations to the arm tissue parameters as part of their sensitivity analysis, so these parameters could conceivably be more important than arterial stiffness and PP in impacting the accuracy of the fixed-ratio method. However, in another modeling effort, these same investigators[2] did assess how changes to arm tissue parameters affect the pressure transmission from the artery to the arm outer surface. They found that changes to the Young's modulus of the arm tissue hardly changed the pressure transmission and that changes to the Poisson ratio of the arm tissue changed the pressure transmission by up to only about 10 mmHg. These and our results indicate that arterial stiffness and PP are indeed the dominant factors influencing the accuracy of the fixed-ratio method. However, further studies are needed to conclusively determine the relative impact of arm tissue mechanics on the accuracy.

Concluding Remarks

In summary, we employed a mathematical model to investigate the blood pressure measurement accuracy of the popular oscillometric fixed-ratio method. We found that its accuracy was highly sensitive to the levels of PP and brachial artery stiffness (at zero transmural pressure). In particular, over a physiologic range of these parameters, the SP and DP errors of the method increased as the brachial artery stiffened and reached 58 mmHg. We also explained, perhaps for the first time, the mechanisms for these errors.

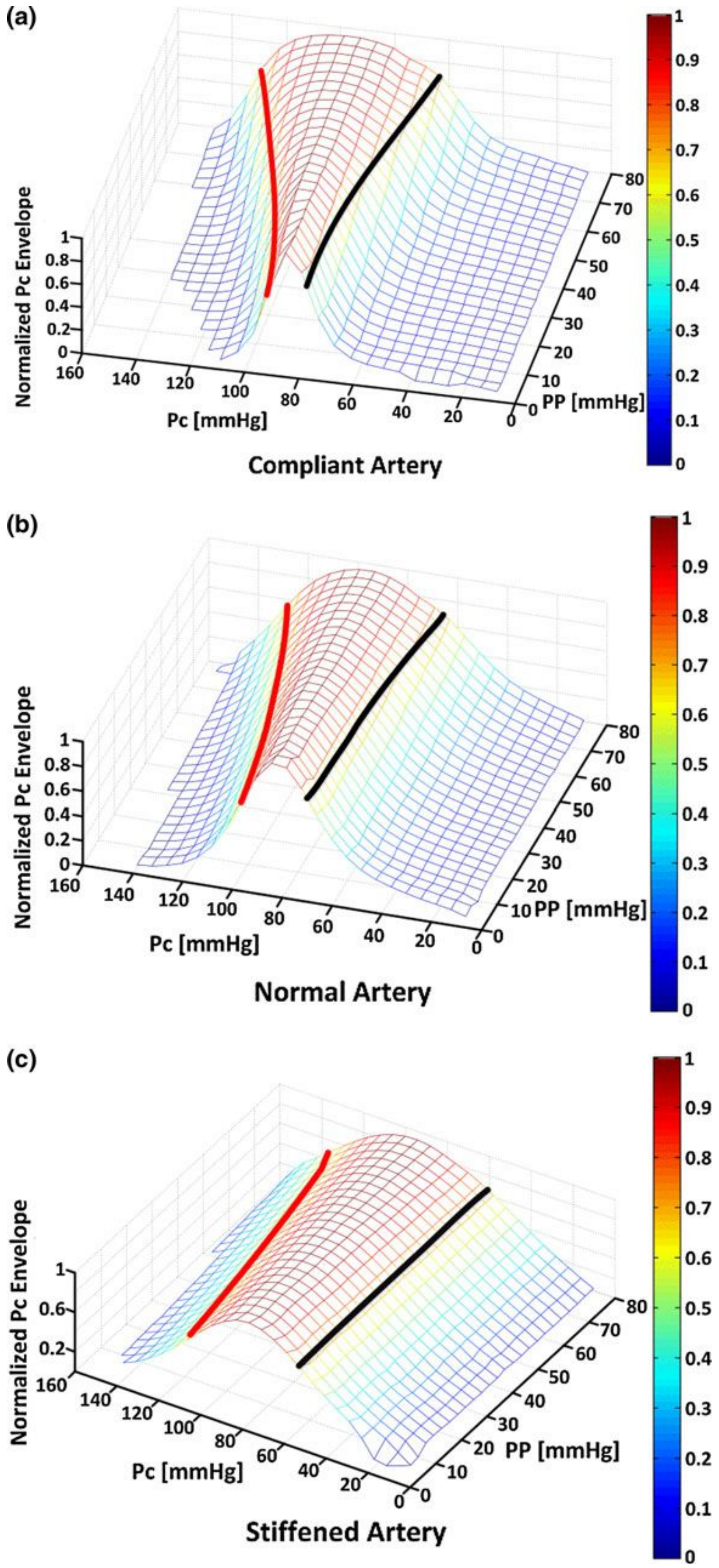

FIGURE 8. Envelope of the model P_c oscillations (normalized to unity) as a function of PP for three different values of *c* representing (a) compliant, (b) normal, and (c) stiff arteries. Solid lines indicate SP (red) and DP (black) estimated by the fixed-ratio method.

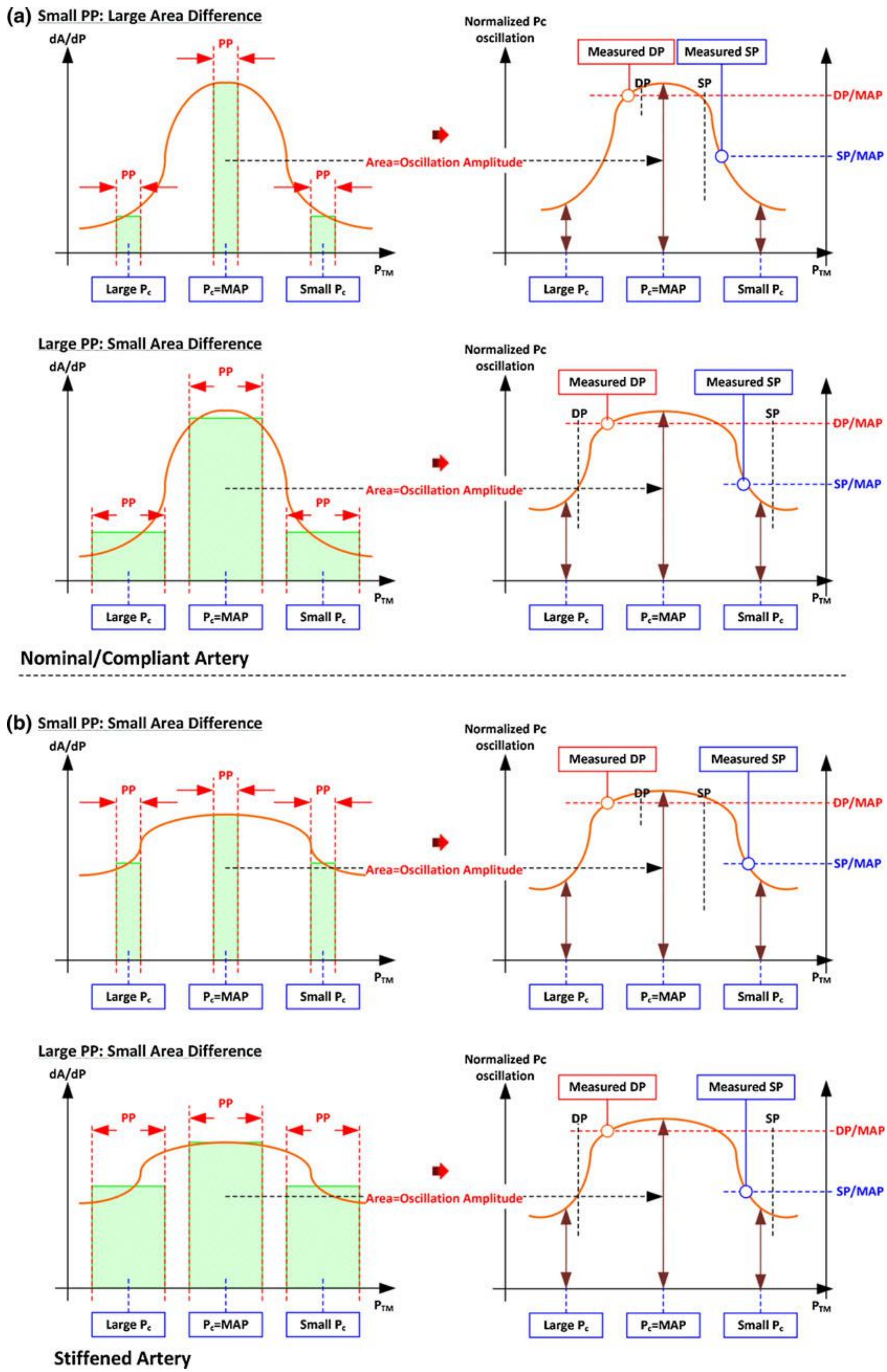

FIGURE 9. Error mechanism of the fixed-ratio method for changes in PP at different levels of arterial stiffness at zero trans-mural pressure.

The theoretical results herein have several important clinical implications. More specifically, as the brachial artery stiffens (*via* reduction in the model parameter c), the fixed-ratio method underestimates DP but overestimates SP. So, for example, isolated systolic hypertension could be wrongly diagnosed in elderly patients, while preeclampsia could be incorrectly concluded in pregnant women who may likewise have increased arterial stiffness.[6] Further, as PP decreases, the fixed-ratio method overestimates PP. So, for example, hemorrhage may not be readily apparent in trauma or surgery patients, especially those with stiff arteries. Accordingly, future efforts to improve the accuracy of the oscillometric BP measurement method are warranted.

ACKNOWLEDGMENTS

This work was supported by the Telemedicine and Advanced Technology Research Center (TATRC) at the U.S. Army Medical Research and Materiel Command (USAMRMC) through award W81XWH-10-2-0124, a US National Science Foundation CAREER Grant [0643477], and the Natural Sciences and Engineering Research Council of Canada (NSERC).

CONFLICTS OF INTEREST

None.

REFERENCES

[1]Coleman, A., P. Freeman, S. Steel, and A. Shennan. Validation of the Omron MX3 Plus oscillometric blood pressure monitoring device according to the European Society of Hypertension international protocol. *Blood Press Monit*. 10(3):165–168, 2005.

[2]Cristalli, C., and M. Ursino. Influence of arm soft tissue on non-invasive blood pressure measurements: an experimental and mathematical study. *Measurement* 14(3):229–240, 1995.

[3]de Greeff, A., Z. Beg, Z. Gangji, E. Dorney, and A. H. Shennan. Accuracy of inflationary versus deflationary oscillometry in pregnancy and preeclampsia: OMRON-MIT versus OMRON-M7. *Blood Press Monit*. 14(1):37–40, 2009.

[4]Drzewiecki, G., R. Hood, and H. Apple. Theory of the oscillometric maximum and the systolic and diastolic detection ratios. *Ann. Biomed. Eng*. 22(1):88–96, 1994.

[5]Geddes, L. A., M. Voelz, C. Combs, D. Reiner, and C. F. Babbs. Characterization of the oscillometric method for measuring indirect blood pressure. *Ann. Biomed. Eng*. 10(6):271–280, 1982.

[6]Kaihura, C., M. D. Savvidou, J. M. Anderson, C. M. McEniery, and K. H. Nicolaides. Maternal arterial stiffness in pregnancies affected by preeclampsia. *Am. J. Physiol. Heart Circ. Physiol*. 297(2):H759–H764, 2009.

[7]Longo, D., G. Toffanin, R. Garbelotto, V. Zaetta, L. Businaro, and P. Palatini. Performance of the UA-787 oscillometric blood pressure monitor according to the European Society of Hypertension protocol. *Blood Press Monit*. 8(2):91–95, 2003.

[8]Marey, E. J. Pression et vitesse dn sang. Masson, Paris: Physiologie Experimentale, vol.2, cb. VIII, 1876, pp. 307–343.

[9]Mauck, G. W., C. R. Smith, L. A. Geddes, and J. D. Bourland. The meaning of the point of maximum oscillations in cuff pressure in the indirect measurement of blood pressure—Part II. *J. Biomech. Eng*. 102:28–33, 1980.

[10]Raamat, R., J. Talts, K. Jagomägi, and J. Kivastik. Errors of oscillometric blood pressure measurement as predicted by simulation. *Blood Press Monit*. 16(5):238–245, 2011.

[11]Richter, H. A., and C. Mittermayer. Volume elasticity, modulus of elasticity and compliance of normal and arteriosclerotic human aorta. *Biorheology* 21(5):723–724, 1984.

[12]Stang, A., S. Moebus, S. Möhlenkamp, N. Dragano, A. Schmermund, E. M. Beck, J. Siegrist, R. Erbel, K. H. Jöckel, and Heinz Nixdorf Recall Study Investigative Group. Algorithms for converting random-zero to automated oscillometric blood pressure values, and vice versa. *Am. J. Epidemiol*. 165(7):848, 2007.

[13]Ursino, M., and C. Cristalli. A mathematical study of some biomechanical factors affecting the oscillometric blood pressure measurement. *IEEE Trans. Biomed. Eng*. 43(8): 761–778, 1996.

[14]Vera-Cala, L. M., M. Orostegui, L. I. Valencia-Angel, N. López, and L. E. Bautista. Accuracy of the Omron HEM-705 CP for blood pressure measurement in large epidemiologic studies. *Arq. Bras. Cardiol*. 96(5):393–398, 2011.

[15]Yelderman, M., and A. K. Ream. Indirect measurement of mean blood pressure in the anesthetized patient. *Anesthesiology* 50:253–256, 1979.