Air cuff transducer design for occlusion-based hemodynamic measurements - An experimental and simulation study

Air Cuff Transducer Design for Occlusion-Based Hemodynamic Measurements—An Experimental and Simulation Study

Laura Bogatu, Simona Turco, *Member, IEEE*, Massimo Mischi, *Senior Member, IEEE*, Lars Schmitt, Pierre Woerlee, R. Arthur Bouwman, Erik H. H. M. Korsten, and Jens Muehlsteff

***Abstract*— In standard clinical practice, cuff devices are widely used for noninvasive blood pressure (NIBP) measurements. However, cuff-based NIBP is prone to large errors, especially in cases of hypotension and hypertension. In addition to this, the cuff measurement principle allows for estimation of a number of other hemodynamic parameters (e.g., cardiac output, arterial stiffness, and augmentation index) by means of analysis of the pulse waveform and/or pulse amplitude recorded in the cuff pressure. However, in standard practice, the cuff is still only used for the measurement of blood pressure (BP). A key reason for the observed measurement errors and the limited set of extracted parameters lies in our poor understanding of the cuff device as a transducer; the cuff pressure response to arm volume pulsations (the cuff transfer function $\mathbf{TF_{cuff}}$) depends on a large number of factors. It is not yet clear to what extent the cuff contributes to the NIBP error or if the cuff in its current format is reliable for measurement of arterial pulse waveform/amplitude. In this study, we investigate the isolated cuff in order to gain a better understanding of the main sources of measurement errors. By using an experimental setup that measures the cuff response to mechanically simulated arm pulsations, we find that the cuff can explain part of the NIBP errors in cases of hypotension. Also, we find that pulse waveforms can be estimated using a standard cuff; however, the measurement of the pulse volume amplitude is unfeasible. We demonstrate that the observed inaccuracies can be eliminated by usage of a calibrator device, which obtains real-time information on $\mathbf{TF_{cuff}}$. These findings will enable further investigation of the effects related to the pulse travel along the compressed limb and the interaction between cuff, arm tissue, distal arm hemodynamics, and arterial walls for the development of improved occlusion-based measurement strategies.**

***Index Terms*— Arterial stiffness, arterial volume, artery oscillation measurement, blood pressure (BP) cuff, cuff hardware, cuff transducer, cuff-based measurement, oscillometry.**

Manuscript received 15 April 2022; revised 9 June 2022; accepted 6 July 2022. Date of publication 27 July 2022; date of current version 29 August 2022. The Associate Editor coordinating the review process was Ada Fort. This work was supported by the Eindhoven MedTech Innovation Center e/MTIC. *(Corresponding author: Laura Bogatu.)*

Laura Bogatu is with Philips Research, 5656 AE Eindhoven, The Netherlands, and also with the Biomedical Diagnostics Laboratory, Eindhoven University of Technology, 5612 AZ Eindhoven, The Netherlands (e-mail: laura.bogatu@philips.com).

Simona Turco, Massimo Mischi, and Pierre Woerlee are with the Biomedical Diagnostics Laboratory, Eindhoven University of Technology, 5612 AZ Eindhoven, The Netherlands (e-mail: s.turco@tue.nl; m.mischi@tue.nl; p.h.woerlee@tue.nl).

Lars Schmitt and Jens Muehlsteff are with Philips Research, 5656 AE Eindhoven, The Netherlands (e-mail: jens.muehlsteff@philips.com; lars.schmitt@philips.com).

R. Arthur Bouwman and Erik H. H. M. Korsten are with Catharina Ziekenhuis, 5623 EJ Eindhoven, The Netherlands (e-mail: arthur.bouwman@catharinaziekenhuis.nl; erik.korsten@catharinaziekenhuis.nl).

Digital Object Identifier 10.1109/TIM.2022.3193980

I. Introduction

THE cuff-based noninvasive blood pressure (NIBP) measurement is key to hemodynamic monitoring and to assessment of cardiovascular health. The measurement principle (oscillometry) consists of inflating a cuff usually placed at the brachial site in order to alter the transmural pressure (P_{tm}) across the arterial wall. Blood pressure (BP) oscillations inside the artery cause arterial volume pulsations of amplitudes that depend on P_{tm}. The arterial volume oscillations propagate through arm tissue, generating arm volume pulsations, which in turn generate cuff pressure (P_{cuff}) pulsations in the attached cuff. The resulting cuff pressure oscillations are processed via empirical methods to derive systolic and diastolic BP values. Such empirical methods have been developed based on observations that cuff pressures at certain ratios of oscillation amplitudes represent reference clinical BP measurements (invasive and auscultatory) [1].

While empirical methods are generally accurate in normotensive patients, large errors in BP values have been reported in hypotensive and hypertensive patients [2]. There is an obvious need for more improved noninvasive BP measurement technologies, especially in critical care, where patients are more likely to suffer from hemodynamic instability and severe variations in BP.

In addition to this, it is hypothesized that the cuff is underutilized in clinical practice and that other measurement strategies can be developed via occlusion-based modulation of arterial P_{tm} to extract additional physiological parameters. In principle, cardiac output [3], arterial stiffness [5], augmentation index [4], arterial viscosity, and heart and lung function [6] might be estimated by analysis of the pulse waveform and/or pulse amplitude recorded in the cuff pressure. In standard practice, however, the cuff is still only used for measurement of BP.

It is not yet clear if the standard air cuff in its current format is suited for improved NIBP measurement and measurement of arterial pulse waveform/amplitude or if hardware improvements are needed. In addition, it is not known what proportion of the measurement inaccuracies can be attributed to the cuff itself or to other factors such as effects related to the pulse travel along the compressed arm tissue [1], [7]. As stated by Drzewiecki *et al*. [8]:

"The occlusive arm-encircling cuff is probably one of the most widely used instruments in the measurement of BP. Yet, its function as a mechanical device has not been explored. It is a device that has been developed more out of practicality than of engineering design [8]."

For these reasons, many efforts have been focused toward characterization of the cuff and, more specifically, the cuff "transfer function," which is the relationship between the input and output signals, i.e., the arm volume oscillations and the recorded cuff pressure oscillations, this being an essential step toward better interpretation of the oscillometric measurement principle. This led to the development of mechanical models of different levels of complexity describing the cuff behavior [1], [10]–[14]. Despite the amount of work, cuff behavior is still not understood in sufficient detail. The existing studies are mostly qualitative, assume idealized nonrealistic, purely elastic material properties, and ignore the frequency dependence of material stresses in the cuff wall. Experimental characterization is often missing or incomplete. In our previous study, we attempted to provide a more detailed mechanical characterization of the cuff properties [15]. A dedicated experimental setup was used to study the cuff transfer function, including air compliance and viscoelastic cuff material effects. While a number of insights were acquired, physical characterization of the cuff was found to require much more complex modeling than expected one. The cuff behavior was found to depend on factors, such as cuff material time constants, pulse frequency content, and patient-specific arm compressibility. In addition to this, our MRI results [7] show that the cuff changes in shape and folds in unpredictable ways during inflation, i.e., it cannot be approximated by a cylinder most of the time, further revealing the unfeasibility of mechanical modeling without detailed knowledge of the cuff design and patient-specific tissue properties.

Therefore, from the practical perspective of improving BP estimation accuracy and introducing new parameters in clinical practice, two questions remain.

1) To what extent does the standard air cuff design and materials impact the BP measurement accuracy and the measurement of arterial pulse waveform/amplitude?
2) Can the standard air cuff be modified to solve the current inaccuracies?

In this study, we tackle the two questions as follows.

1) Via an experimental setup, we measure the response of the cuff to mechanically simulated arm volume pulsations under a range of factors occurring under typical clinical circumstances, i.e., varying cuff inflation speeds, different pulse frequency content, and arm compressibility. In essence, we measure the dependence of the cuff transfer function $\mathrm{TF}_{\mathrm{cuff}}$ on these factors. An analysis is performed to quantify the impact of the $\mathrm{TF}_{\mathrm{cuff}}$ variability on the accuracy of the estimated BP, pulse waveform, and pulse amplitude under different physiological conditions.
2) We overcome the current inaccuracies by means of the "cuff calibration" principle [9]. A hardware feature is added to the cuff in order to enable automatic real-time $\mathrm{TF}_{\mathrm{cuff}}$ estimation. In this way, cuff response to arm volume pulsations can be estimated regardless of the absolute cuff pressure, amount of air volume within the cuff, cuff material time dependence, arm compressibility, and cuff folding during inflation. By using the experimental setup described above, we validate the performance of this novel hardware feature.

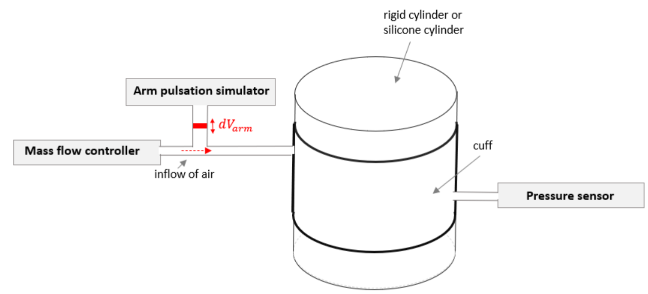

Fig. 1. Illustration of the experimental setup for measurement of the cuff transfer function.

The presented calibration approach is beneficial and practical because it solves the problem of cuff-related uncertainties by circumventing the need for complex, assumption-prone physical characterization of the cuff, and patient arm characteristics. Moreover, the method enables for development of cuff features specifically aimed at overcoming the observed measurement inaccuracies. In this way, only minimal changes to standard clinical equipment are needed, favoring the clinical uptake of the method by employment of cuff devices that stay compatible with existing hospital equipment and clinical procedures [16].

II. Methods

A. Measurement Accuracy of the Standard Cuff Design

1) Cuff Transfer Function Measurement: The realized experimental setup is shown in Fig. 1. A two-input standard air cuff is inflated via a mass-flow controller (Brooks SLA5800). During the inflations, sinusoidal arm volume variations of amplitude dV_{arm} are artificially generated via an electronically controlled piston pump connected to the cuff input via a 25-cm-length, 4-mm-diameter air tube. dV_{arm} amplitudes and frequencies are chosen close to the expected values for arm volume changes: 1-mL volume at frequencies from 1 to 10 Hz. The cuff is inflated over 20 or 40 s. A 40-cm air tube connects the second cuff input to a gauge pressure sensor (a sampling rate of 125 Hz). The cuff is wrapped around either a rigid phantom arm or around a compressible silicone phantom arm. Silicone is a material commonly used to simulate tissue behavior in phantom arms [14]. The rigid and silicone phantoms are meant to mimic two arms of different compressibility.

$\mathrm{TF}_{\mathrm{cuff}}$ is defined via the following equation:

$$\mathrm{TF}_{\mathrm{cuff}} = \frac{dV_{\mathrm{arm}}}{dP_{\mathrm{cuff}}} \tag{1}$$

where dP_{cuff} is the cuff oscillation as response to dV_{arm}. $\mathrm{TF}_{\mathrm{cuff}}$ varies depending on dV_{arm} frequency, absolute cuff pressure, cuff inflation speed, and arm compressibility. Note that in previous studies, this equation has been defined under

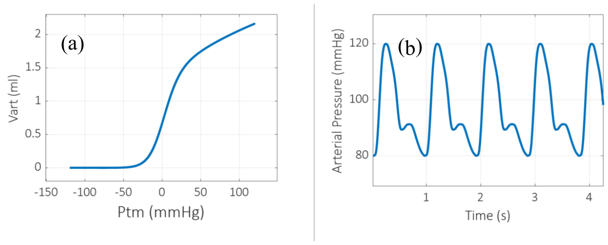

Fig. 2. (a) Illustration of arterial volume with respect to transmural pressure across the arterial wall. (b) Example of simulated arterial pressure.

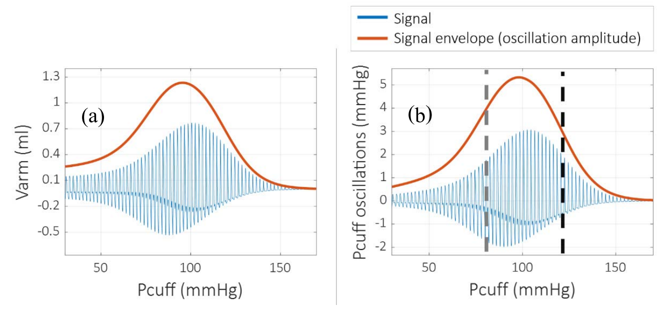

Fig. 3. (a) Example of simulated arm volume pulsations with respect to absolute cuff pressure. (b) Example of corresponding simulated cuff pressure pulsations with respect to absolute cuff pressure under slow cuff inflation and rigid arm conditions. Dashed lines represent diastolic and systolic values inferred via empirical processing of the signal envelopes. This example simulates normotension with $P_{\text{dia}} = 80$ mmHg and $P_{\text{sys}} = 120$ mmHg. Empirical ratios are typically optimized such that BP is estimated correctly under normotension conditions. In this example, $P_{\text{diaPenv}} = 80$ mmHg and $P_{\text{sysPenv}} = 121$ mmHg.

the term "cuff compliance," $C = \Delta V/\Delta P$ [1]. In our study, we make use of the term "cuff transfer function" to account for frequency dependence.

TF_{cuff} is measured under the different conditions: volume oscillations of different frequencies, over a range of cuff pressures under the two different inflation speeds on both the rigid and the silicone phantom arms.

2) Impact of Cuff Transfer Function on BP Estimation: The measured TF_{cuff} is then used as part of proprietary simulation software, which estimates the error in empirical BP estimation due to cuff behavior variability. The software framework enables analysis over a wide range of BP values, pulse pressures, and control of arterial volume waveform according to nonlinear arterial collapse principles (see Fig. 2).

In the simulation, arterial volume V_{art} is modeled via the following equation:

$$V_{\text{art}}(P_{\text{tm}}) = L_{\text{cuff}}\, d \frac{\ln(a P_{\text{tm}} + 3.3)}{\left(1 + e^{-c P_{\text{tm}}}\right)} \tag{2}$$

where a, c, and d are parameters describing arterial collapse [11, 5]. L_{cuff} is the cuff length of 14 cm. P_{tm}, the transmural pressure across the arterial wall, depends on the arterial pressure P_{art} and cuff pressure P_{cuff}

$$P_{\text{tm}} = P_{\text{art}} - P_{\text{cuff}}. \tag{3}$$

P_{art} is simulated to follow a typical arterial waveform, oscillating between systolic value, P_{sys}, and diastolic value, P_{dia}, at either 60 or 200 beats per minute (BPM). As the cuff pressure increases, the V_{art} oscillations adjust in amplitude according to the resulting P_{tm}.

It is assumed that dV_{art} equals dV_{arm}; transmission of volume pulse through arm tissue is outside the scope of this article (although preliminary evidence is showing that this could be of importance at lower cuff pressures).

The amplitude of the V_{arm} oscillations with respect to the absolute cuff pressure is defined as the volume envelope Venv (P_{cuff}); an example illustration of Venv (P_{cuff}) can be seen in Fig. 3(a).

The cuff pressure envelope is then simulated. Each V_{arm} frequency component corresponds to a TF_{cuff} value; the V_{arm} waveforms are processed in the frequency domain to obtain the resulting P_{cuff} waveforms [example in Fig. 3(b)]. Cuff pressure envelopes are simulated under hypotension, normotension, and hypertension conditions (diastolic BP ranging from 25 to 125 mmHg and systolic BP ranging from 50 to 200 mmHg [2]), different inflation speeds (the cuff is inflated over 20 or 40 s), and different heart rates (1 and 3.3 Hz).

For each simulated P_{cuff} envelope, P_{sysPenv} and P_{diaPenv} are computed and compared to reference P_{sys} and P_{dia}, respectively. To achieve this, the P_{cuff} envelope is normalized. According to the common empirical method [1], the cuff pressure on the falling phase of the envelope at which the pulsation amplitude is about 50% of the maximum amplitude is the systolic BP P_{sysPenv}. Cuff pressure on the rising phase of the oscillation signal at which the pulsation amplitude is about 70% of the maximum amplitude is the diastolic BP P_{diaPenv} [1].

However, the ratios are empirical, and their exact values depend on the specific implementation of algorithms in commercially available devices. It is important to understand whether TF_{cuff} is relevant in selection of such ratio values.

To achieve this, for each simulated P_{cuff} envelope, the ideal ratios that would lead to correct BP inference are computed.

In addition, we also give an illustration of the potential error that could result from the empirical estimation of BP by fixed ratios that do not account for TF_{cuff} variability. For this, P_{sysPenv} and P_{diaPenv} are computed for each P_{cuff} envelope based on two often mentioned sets of fixed ratios, which are likely to be encountered in practical algorithm implementations:

1) 50% and 70%, respectively, for P_{sys} and P_{dia} estimation.
2) 55% and 75%, respectively, for P_{sys} and P_{dia} estimation.

A second empirical method commonly used for BP inference is based on derivatives [18]. The cuff pressure at which the P_{cuff} envelope has maximum slope is the diastolic BP (P_{diaPenv}) and the cuff pressure at which the P_{cuff} envelope has minimum slope is the systolic BP (P_{sysPenv}). We provide an illustration of the potential error that could result from the derivative-based estimation of BP in the presence of TF_{cuff} variability.

3) Impact of Cuff Transfer Function on Quantified Volume Pulse Amplitude Measurement: The variability of TF_{cuff} across different conditions (cuff inflation speeds, heart rate (HR) values, absolute cuff pressure, and arm compressibility) is assessed. The degree of variability can reveal if the translation

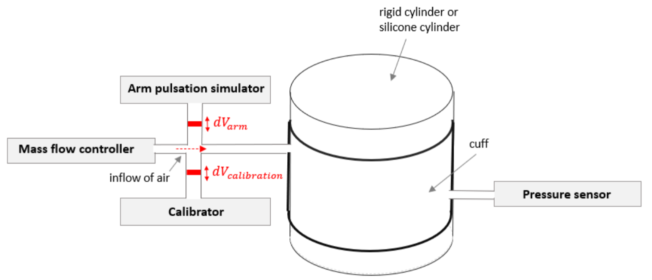

Fig. 4. Illustration of the experimental setup for validation of calibrator device.

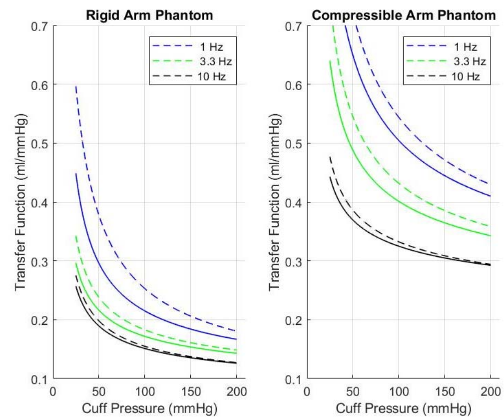

Fig. 5. TF_{cuff} measured via the experimental setup (shown in Fig. 1). The cuff pressure response to mechanically generated arm volume pulsations of 1, 3.3, and 10 Hz is plotted. TF_{cuff} is measured on rigid (left) and compressible (right) arm phantom. The dashed lines represent TF_{cuff} measured under fast inflation and the solid lines represent TF_{cuff} measured under slow inflation (the cuff is inflated over 20 or 40 s).

of dV_{arm} absolute value by means of analysis of the corresponding dP_{cuff} oscillation is feasible.

4) Impact of Cuff Transfer Function on Pulse Waveform Measurement: The measured TF_{cuff} is used as part of a software framework to estimate the error in pulse waveform estimation due to variability in the cuff behavior. By the same method described in Section II-B, V_{arm} and the corresponding P_{cuff} oscillations occurring during cuff inflation are simulated under different conditions. Each pulse in the V_{arm} signal is normalized and is compared to the corresponding normalized P_{cuff} pulse. The difference in waveform between the two pulses is quantified via the root-mean-square error (RMSE) metric.

B. Cuff Calibrator Device

A second piston pump generating controlled sinusoidal volume oscillations $dV_{calibration}$ is designed for real-time measurement of TF_{cuff} (see Fig. 4). dV_{arm} is programmed to generate a waveform similar to the pulsations recorded in typical NIBP measurements performed on real arms. dV_{arm} can be set to 60 or 200 BPM.

The controlled volume oscillations $dV_{calibration}$ and dV_{arm} both generate cuff pressure oscillations $dP_{calibration}$ and dP_{arm}. The frequency of $dV_{calibration}$ is designed such that it differs from the frequency components of dV_{arm}. Therefore, simple filtering can be applied to the P_{cuff} signal to measure the amplitude of $dP_{calibration}$. The amplitude of $dV_{calibration}$ is also well-defined; therefore, information on TF_{cuff} can be obtained in real time.

For validation of the measurement strategy, dV_{arm} and $dV_{calibration}$ oscillations are generated. The resulting P_{cuff} signal is filtered to obtain the $dP_{calibration}$ amplitude, which is divided by the $dV_{calibration}$ amplitude to obtain TF_{cuff}. The amplitude of dV_{arm} is then computed based on dP_{arm} and TF_{cuff} information. Since the exact amplitude of dV_{arm} is known, the percentage error in the dV_{arm} amplitude estimation can be computed.

The validation of the measurement is conducted at 40- and 90-mmHg absolute cuff pressures, for 60 and 200 BPM dV_{arm} heart-rate values and for rigid and compressible phantom arms. For each of the absolute cuff pressure value, heart rate, and arm compressibility conditions, the dV_{arm} estimation process is repeated 20 times. For each repetition, the percentage error between the estimated amplitude of dV_{arm} and the true value of dV_{arm} is calculated. The results are summarized by computing the percentage error mean and standard deviation.

Across all measurement conditions, $dV_{calibration}$ is set to a predefined frequency of 8 Hz, which differs sufficiently from the frequency components of the dV_{arm} signal. Note that future developments will enable real-time adjustment of $dV_{calibration}$ frequency depending, e.g., on the recorded patient-specific dP_{cuff} waveform frequency components, or changes in heart rate. Also, $dV_{calibration}$ composed of multiple frequency components is ideal for a more complete sampling and subsequent identification of the cuff response. However, this is outside the scope of this article, which aims at conducting a first investigation of the TF_{cuff} measurement principle, along with the applicability and use of a calibrator device.

III. Results

A. Measurement Accuracy of the Standard Cuff Design

1) Cuff Transfer Function Measurement: Fig. 5 reveals that the cuff response varies significantly depending on arm compressibility, inflation speed, and absolute cuff pressure. This finding suggests the need to quantify and possibly to correct the impact of TF_{cuff} on the accuracy of cuff-based hemodynamic measurements. TF_{cuff} cannot be predicted; in standard clinical practice, TF_{cuff} is expected to change with every wrapping, depending on exact location on the upper arm where cuff is positioned, subsequent compression of the tissue, wrapping tightness leading to unpredictable cuff folding, inflation procedure, and so on.

2) Impact of Cuff Transfer Function on BP Estimation: P_{cuff} envelopes are simulated under different conditions: BP values ranging across hypotension, normotension, and hypertension, different pulse pressures, for normal and high HR, fast and slow inflation, and compressible and rigid arms. For each of the conditions, V_{arm} oscillations are generated via the simulation framework and $\text{Venv}(P_{cuff})$ is computed. The V_{arm} oscillations are processed together with the corresponding TF_{cuff} (measured via the experimental setup) to generate Penv (P_{cuff}).

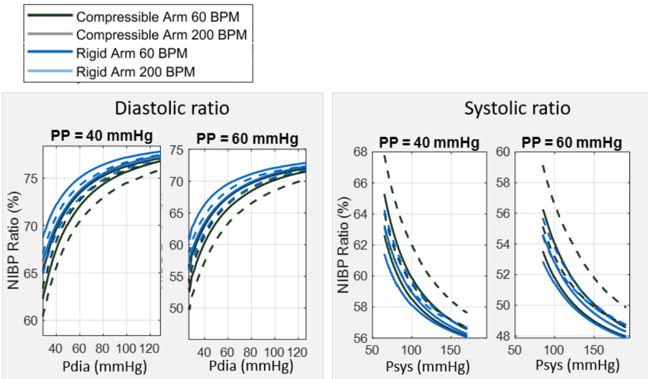

Fig. 6. Ideal ratios for deriving correct BP values based on the pressure envelope for rigid and compressible arms, 60 and 200 BPM, and 40- and 60-mmHg pulse pressure values. The dashed lines represent BP ratios corresponding to fast inflation. The solid lines represent BP ratios corresponding to slow inflation.

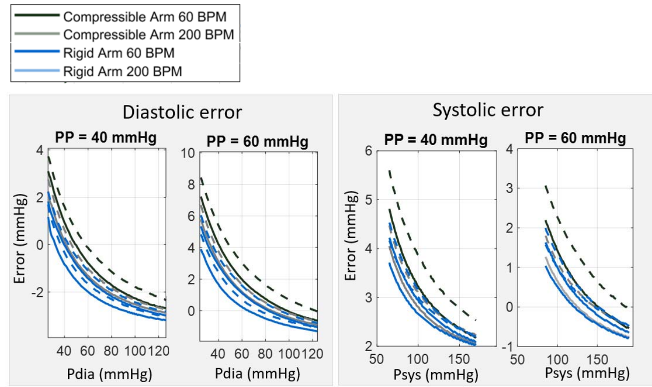

Fig. 7. $P_{sysPenv}P_{sys}$ and $P_{diaPenv}P_{dia}$ are plotted for rigid and compressible arms, 60 and 200 BPM, and 40- and 60-mmHg pulse pressure values, respectively. The dashed lines represent BP error corresponding to fast inflation. The solid lines represent BP error corresponding to slow inflation. The Penv(P_{cuff}) envelopes are generated based on the simulated V_{arm} oscillations and the measured TF_{cuff} (see Fig. 5). $P_{sysPenv}$ and $P_{diaPenv}$ are obtained based on the 50% and 70% empirical ratios, respectively.

For each simulated P_{cuff} envelope, the ideal ratio values that would lead to correct systolic and diastolic values are computed. Fig. 6 shows that the ideal ratio for systolic value inference varies from 48% to 68% and the ideal ratio for diastolic value inference varies from 50% to 78%. This result reveals that fixed ratios might not be optimal and that knowledge of TF_{cuff} can contribute to improved selection of such ratio values.

In [2, Fig. 1], large errors in NIBP are reported in hypotensive and hypertensive patients. We quantify the error that could result from the empirical estimation of BP by fixed ratios that do not account for TF_{cuff} variability.

Figs. 7 and 8 show the NIBP errors defined as $P_{sysPenv}$–P_{sys} and $P_{diaPenv}P_{dia}$, respectively, for BP values ranging across hypotension, normotension, and hypertension, for different pulse pressures, normal and high HR, fast and slow inflation, and compressible and rigid arms. $P_{sysPenv}$ and $P_{diaPenv}$ are computed based on the following:

1) the 50% and 70% ratios (see Fig. 7);
2) the 55% and 75% ratios (see Fig. 8).

Trends similar to the errors reported in [2, Fig. 1] are observed: the systolic and diastolic errors under hypotension are positive; as normotension is reached, the errors decrease. In [2, Fig. 1], BP inference in cases of hypotension is overestimated; BP inference in cases of hypertension is underestimated. Our results show that the cuff can contribute to BP overestimation during hypotension. In this simulation, the cuff transfer function contributes by a negligible amount to BP underestimation during hypertension.

In terms of error absolute values, the results show that the cuff mainly impacts the inference of diastolic values in cases of hypotension, where a maximum in the order of 10-mmHg error is found. Note that these absolute error values depend not only on the cuff behavior but also on the BP inference algorithm and chosen ratio (see Figs. 6 and 7).

Fig. 9 shows the NIBP errors defined as $P_{sysPenv}$–P_{sys} and $P_{diaPenv}P_{dia}$, where $P_{sysPenv}$ and $P_{diaPenv}$ are obtained based on the derivative algorithm.

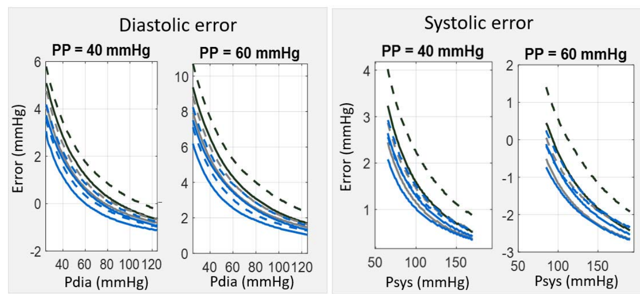

Fig. 8. $P_{sysPenv}P_{sys}$ and $P_{diaPenv}P_{dia}$ are plotted similar to Fig. 6, respectively. $P_{sysPenv}$ and $P_{diaPenv}$ are obtained based on the 55% and 75% empirical ratios, respectively.

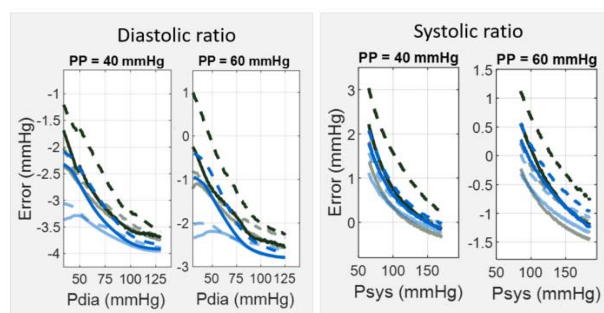

Fig. 9. $P_{sysPenv}P_{sys}$ and $P_{diaPenv}P_{dia}$ are plotted similar to Figs. 6 and 7, respectively. $P_{sysPenv}$ and $P_{diaPenv}$ are obtained by identifying the cuff pressure at which minimum and maximum slope of the $P_{sysPenv}$ occurs, respectively.

A maximum error in the order of 4 mmHg is found. This suggests that the accuracy of the derivative-based algorithm is less affected by effects stemming from the cuff when compared to the ratio-based algorithm. Note, however, that derivative-based algorithms are particularly sensitive to noise. A combination of ratio- and derivative-based methods is likely necessary for practical use.

In [2, Fig. 1], errors in the range of 25 mmHg are reported. This suggests that the TF_{cuff} variability and empirical processing of the P_{cuff} envelope can explain part of the

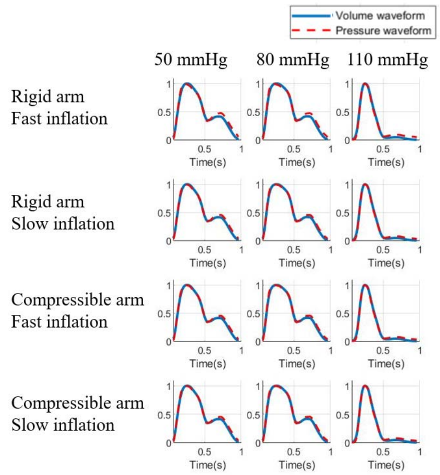

Fig. 10. Simulated and normalized arm volume waveforms and corresponding cuff pressure waveforms. The cuff pressure waveforms are generated based on the simulated arm volume waveforms and the measured TF_{cuff} (see Fig. 5). Example waveforms at 50-, 80-, and 110-mmHg absolute cuff pressures are shown.

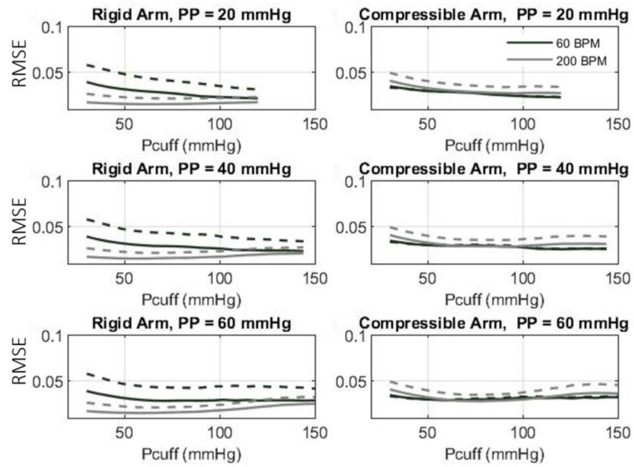

Fig. 11. RMSE representing the difference between normalized arm volume waveform and corresponding cuff pressure waveform is plotted for rigid and compressible arms, 60 and 200 BPM, and various pulse pressure values. The dashed lines represent RMSE corresponding to fast inflation. The solid lines represent RMSE corresponding to slow inflation.

reported errors in NIBP, while the rest of the error likely stems from arm tissue effects, pulse travel along the compressed arm, arm–cuff interaction, or issues of the accuracy of the invasively measured BP.

3) Impact of Cuff Transfer Function on Quantified Volume Pulse Amplitude Measurement: The TF_{cuff} variation (shown in Fig. 5) can lead to errors up to a factor $\sim$4 in the estimation of the arm volume pulsation amplitude based on a measured cuff pressure oscillation. Therefore, this is a clear indication that the amplitude of the volume pulse cannot be measured via the standard cuff dependent on the situation-specific measurement conditions (cuff/arm tissue compression).

4) Impact of Cuff Transfer Function on Pulse Waveform Measurement: According to our simulation results, the cuff impact on the pulse waveform measurement is minimal. To give a visual impression, Fig. 10 shows the examples of different simulated V_{arm} waveforms and corresponding P_{cuff} oscillations (computed based on Fig. 5 measurements). In Fig. 11, the difference between the two waveforms via the RMSE metric for each of the simulated conditions is quantified (different pulse pressures, normal and high HR, fast and slow inflation, and inflation on compressible and rigid arms). For each of the conditions, V_{arm} oscillation is generated via the simulation framework. The V_{arm} oscillation is processed together with the corresponding TF_{cuff} (measured via the experimental setup) to generate P_{cuff}.

B. Cuff Calibrator Device

As shown in Fig. 5, large errors in estimation of arm volume pulsation amplitude can occur if TF_{cuff} is not known (Section III-A3). We demonstrate the feasibility of the real-time TF_{cuff} measurement method via the experimental setup shown in Fig. 4.

Table I shows the mean and standard deviation of the dV_{arm} amplitude estimation percentage error for each of the measurement conditions.

TABLE I

dV_{arm} AMPLITUDE ESTIMATION ACCURACY BASED ON MEASURED dP_{cuff} AND CALIBRATOR DATA. RESULTS ACQUIRED VIA THE EXPERIMENTAL SETUP SHOWN IN FIG. 4

	HR / Cuff Pressure	**60 BPM, 40 mmHg**	**200 BPM, 40 mmHg**	**60 BPM, 90 mmHg**	**200 BPM, 90 mmHg**
Rigid arm	Mean	0.2%	0.3%	3.9%	1.6%
	Standard deviation	0.7%	0.2%	0.8%	0.6%
Silicone arm	Mean	4.2%	1.5%	2.0%	1.3%
	Standard deviation	1.8%	1.6%	1.5%	0.7%

The dV_{arm} estimation percentage error is minimal and ranges from 0.2% to 4.2%. This demonstrates that the calibrator enables for measurement of dV_{arm} amplitude and that measurement inaccuracies stemming from the cuff can be tackled via real-time measurement of TF_{cuff}.

IV. DISCUSSION

A. To What Extent Does the Standard Air Cuff Design Impact BP Measurement Accuracy and Measurement of Arterial Pulse Waveform/Amplitude?

First, we find that the cuff contributes partially to the NIBP measurement error. TF_{cuff} needs to be considered for optimized selection of BP inference ratios (see Fig. 6). When using ratio-based algorithms, not accounting for TF_{cuff} variability can lead to error contribution up to 10 mmHg in the estimation of BP values in some cases of hypotension. The performance of derivative-based algorithms appears to be less affected by TF_{cuff} variability, with an error contribution of up to -4 mmHg in the estimation of BP values in some cases of hypertension. These results might indicate that derivative-based algorithms could be useful in minimizing errors stemming

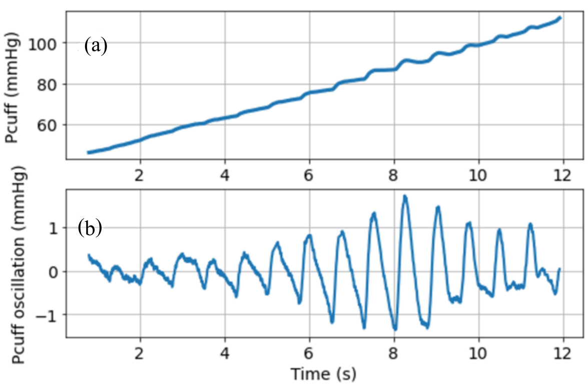

Fig. 12. (a) Example of typical cuff pressure signal obtained during cuff inflation performed on patient arm. (b) Waveforms acquired via high pass of the cuff signal.

from TF_{cuff} variability. However, it is known that derivative-based algorithms are particularly sensitive to noise, e.g., breathing, air pump, and motion artifacts, which might hamper performance in practice. A combination of ratio- and derivative-based methods is likely necessary for practical use.

Overall, the results suggest that TF_{cuff}, even though very variable in absolute terms (see Fig. 5), does not cause significant distortion of the normalized Venv (P_{cuff}) signal. It is likely that other factors (arm tissue/pulse travels along the compressed arm and increases blood volume in distal arm during inflation, resistive drop over brachial artery) can explain the rest of the NIBP errors (as it was reported to be in the order of 25 mmHg in both hypotension and hypertension [2]). Also, it is relevant to note that the invasive BP measurement itself can sometimes be lead to errors [19]. This could be another potential reason for the reported discrepancy between invasive and noninvasive measurements.

Second, we find that the standard cuff does not allow for a quantified estimation of the dV_{arm} absolute value.

Third, our results indicate that the cuff behavior variability does not lead to significant errors in the measurement of the arm pulse waveform. Note that our results only reveal the impact of the cuff component on the arm volume waveform measurement. The arterial pulse waveform, however, might be distorted by the cuff–arm tissue–arterial wall interaction along the length of the cuff (arm tissue fat/muscle consistency, strain-dependent young modulus of tissue, and viscous properties). To illustrate this, Fig. 12 shows cuff pressure signals measured on a patient; this represents a typical example of cuff pressure waveforms acquired on humans. The waveforms appear damped when compared to the arterial waveforms (Romagnoli *et al.* [19] presented a comparison between optimally damped versus overdamped pressure waveforms). This study focused on measuring the arm volume pulsation. Next studies will focus on accessing arterial volume information [17].

B. Can the Standard Air Cuff Be Modified to Overcome the Observed Inaccuracies?

The measurement of dV_{arm} absolute value was found to be unfeasible due to variations in TF_{cuff}, this being the main limitation when investigating the cuff device, without considering cuff–tissue–artery interaction. In addition, a cuff-induced 10-mmHg error in the BP estimation can occur in some cases of hypotension, and this is dependent on the particular algorithms for oscillometry-based BP inference. Two potential solutions to these limitations exist.

1) Physics-based characterization of cuff behavior such that cuff pressure change with response to arm volume change is known.
2) Embedding of a hardware feature within the standard cuff as a method to obtain information on TF_{cuff} in real time.

In this study, we explored the latter solution and made use of a piston-pump calibrator device to demonstrate the measurement principle for acquiring information on TF_{cuff} in real time. We found that this solution can enable measurement of the arm pulsation amplitude. Therefore, it is suggested that cuff-induced NIBP error can also be improved by the same approach.

The presented solution is favorable from a clinical perspective. Embedding of a calibrator device within existing cuffs is feasible and it ensures that minimal changes to standard clinical equipment are made. Therefore, cuff devices stay compatible with existing hospital equipment and clinical procedures while enabling for absolute dV_{arm} measurement. Another benefit is in that the real-time measurement of TF_{cuff} is not prone to effects related to unpredictable cuff folding occurring during inflation or to unpredictable changes in cuff volume during inflation (caused by patient-specific arm tissue compression). Further investigations are needed to optimize the use of the calibrator in clinical practice; for example, tests are needed for identifying how often TF_{cuff} needs to be measured once the cuff has been placed on the arm.

This study provides an understanding of the isolated cuff. The obtained findings will aid further studies to investigate the effects related to pressure/volume pulse travel along the compressed limb and the interaction between cuff, arm tissue, and artery. Our study suggests that integration of a calibrator in the cuff is a valid and adequate solution for conducting occlusion-based measurements; implementation adjustments and additional cuff features might be needed as new insights of the interaction between cuff, arm tissue, and vasculature are acquired.

V. Conclusion

The oscillometric measurement principle is very complex; the P_{cuff} signal is the result of interference between multiple effects related to cuff properties, arm tissue, and arterial walls. Analysis of the isolated cuff is an essential step toward NIBP improvement and toward estimation of additional hemodynamic parameters from cuff pressure signals.

Our results show that the typically unknown cuff transfer function can explain part of the NIBP error and that absolute volume pulse amplitude cannot be measured via the standard air cuff. We find that a practical solution to overcome these limitations consists of introducing a calibrator device. Given the presented framework, we also find that the cuff transfer function itself does not lead to substantial waveform distortion of the localized mechanically simulated arm volume pulse. Further studies are required to investigate the cuff–arm tissue–arterial wall interaction along the length of the cuff and its potential impact on the accuracy of occlusion-based hemodynamic measurements. The presented research approach, experimental setup, and validation method can serve as a basis for such further studies.

ACKNOWLEDGMENT

The authors would like to thank Achim Hilgers, Edward Berben, Valentina Davidoiu, and Mathieu Oude Vrielink for their help with the experimental measurements.

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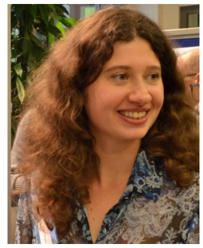

Laura Bogatu received the master's degree (Hons.) in electronic and software engineering from the University of Glasgow, Scotland, U.K., in 2017, and the Ph.D. degree from the Eindhoven University of Technology (TU/e), Eindhoven, The Netherlands, in 2022.

As part of her graduation project, she developed machine vision applications for congestive heart failure diagnosis in collaboration with Imec, Eindhoven. With an interest in interdisciplinary perspectives, she also worked on video compression applications in collaboration with Nallatech (BittWare), U.K., and developed remote control software tools for IBM, U.K., and has been part of projects conducted by Glasgow University research groups involving compressive sensing for single detector imaging and cloud computing. The research conducted at TU/e (in collaboration with Philips Research and Catharina Hospital, Eindhoven) was aimed at developing patient monitoring solutions with a focus on measuring novel hemodynamic parameters via noninvasive technologies. Since July 2022, she will continue this topic as a Scientist with the Department of Patient Care and Measurements, Philips Research.

Simona Turco (Member, IEEE) received the M.Sc. degree (*summa cum laude*) in biomedical engineering from the University of Pisa, Pisa, Italy, in 2012, the P.D.Eng. degree in healthcare system design from the Stan Ackermans Institute, Eindhoven University of Technology (TU/e), Eindhoven, The Netherlands, in 2015, and the Ph.D. degree with thesis entitled "Pharmacokinetic modeling in cancer: From functional to molecular imaging of angiogenesis" from the TU/e in 2018.

She is currently an Assistant Professor with the Biomedical Diagnostics Research Laboratory, TU/e, focusing on quantitative model-driven analysis of biosignals. In this context, she received the Fellowship by the Hanarth Fonds to conduct research on artificial intelligence in oncology.

Massimo Mischi (Senior Member, IEEE) received the M.Sc. degree in electrical engineering from La Sapienza University, Rome, Italy, in 1999, and the Ph.D. degree from the Eindhoven University of Technology (TU/e), Eindhoven, The Netherlands, in 2004.

In 2007, he was an Assistant Professor with the Electrical Engineering Department, TU/e. In 2011, he joined TU/e as an Associate Professor and founded the Biomedical Diagnostics Research Laboratory, where he was involved in model-based quantitative analysis of biomedical signals with applications ranging from electrophysiology to diagnostic imaging. He is currently a Full Professor with the Electrical Engineering Department, TU/e. He has coauthored over 400 peer-reviewed publications, 13 patents, ten book chapters, and one book.

Prof. Mischi is a Board Member of the Imaging Section of the European Association of Urology, the Secretary of the Dutch Society of Medical Ultrasound, and the Chairperson of the IEEE EMBS Benelux Chapter. He was a recipient of the Stichting voor de Technische Wetenschappen (STW) VIDI Grant in 2009, the European Research Council (ERC) Starting Grant in 2011, and the ERC Proof of Concept in 2019 for his research on angiogenesis imaging. He serves as an Associate Editor for the IEEE TRANSACTIONS ON ULTRASONICS, FERROELECTRICS, AND FREQUENCY CONTROL (TUFFC), IEEE REVIEWS IN BIOMEDICAL ENGINEERING (RBME), *Computer Methods and Programs in Biomedicine* (CMPB), and *Innovation and Research in BioMedical engineering*.

Lars Schmitt received the Diploma degree in electrical engineering and information technology from Ruhr-Universität Bochum, Bochum, Germany, in 2001, and the Ph.D. degree in electrical engineering and information technology from RWTH Aachen University, Aachen, Germany, in 2008.

Since 2006, he has been with Philips Research, Eindhoven, The Netherlands, where he focused on innovations in the fields of medical device interoperability, biomedical sensors, vital sign measurements, and digital health solutions. He currently leads a competence team of scientists for patient care measurements and is a Product Owner of an Agile Team for hemodynamic measurements with Philips Research.

Pierre Woerlee received the Ph.D. degree in atomic physics from the University of Amsterdam, Amsterdam, The Netherlands, in 1979.

He joined Philips Research, Eindhoven, The Netherlands, in 1980. At Philips Research, he started a research project on electrical transport properties of nano-scale structures and quantum transport phenomena. In 1985, he switched to CMOS semiconductor technology. As a Research Fellow, he has been focusing on CMOS device scaling, device physics and technology, RF-CMOS, and solid-state storage. In 2001, he switched to DVD double layer recording technology, including the release of DVD + R DL disk technology. In 1998, he was the first non-U.S. Chairperson of the IEEE-IEDM conference. He was a part-time Professor with the Faculty of Electrical Engineering, University of Twente, Enschede, The Netherlands, from 1992 to 2001. From 2012 to 2016, he was leading a project on new patient monitor architectures. He retired from Philips on January 1, 2017. He joined the TUe-SPS Group on January 2017. Since 2006, he has been focusing on health care-related topics including low acuity patient monitoring, clinical measurements, resuscitation care, automated cardiopulmonary resuscitation (CPR) and monitoring devices for emergency care, and on clinical decision support for acute care during resuscitations. His current research interests include human cardio-vascular physiology, mechanical ventilation, patient monitoring, monitoring in low acuity care settings, new monitoring techniques and algorithms, and models for early warning for patient deterioration. He has authored or coauthored over 120 journal articles and more than 40 U.S. patents.

R. Arthur Bouwman is currently a Cardiac Anesthesiologist with Catharina Hospital, Eindhoven, The Netherlands. He combines his clinical work with a position as a Senior Research Associate with Technical University, Eindhoven; and a Clinical Advisor, Philips Research, Eindhoven. He has a strong background in cardiovascular patient monitoring and clinical decision support in acute settings. In his research, he focuses on new sensor technology, predictive decision support systems, clinical workflow integration, and value-based healthcare optimization. In his research group, he supervises several Ph.D. students. He has authored or co-authored several scientific publications.

Erik H. H. M. Korsten received the degree in medicine from Groningen University, Groningen, The Netherlands, in 1979, the degree with specialization in anesthesiology from the St. Antonius Hospital, Utrecht, The Netherlands, in 1982, and the Ph.D. degree from Leiden University, Leiden, The Netherlands, in 1984.

He joined the Catharina Hospital, Eindhoven, The Netherlands, in 1983, as an Anesthesiologist-Intensivist, and has worked there ever since. In 2001, he was appointed as a part-time Professor at the Eindhoven University of Technology, Eindhoven. In 2015, he joined Zhejiang University, Hangzhou, China, as a part-time Professor. He specializes in signal and data processing and decision support systems, such as his Gaston software that helps make sense of the data in the Dutch electronic patient files (EPD) and helps hospitals ensure that protocols and guidelines are followed. This contributes, for instance, to a reduced number of medication errors. He also focuses on signal processing in ultrasound and MRI images.

Jens Muehlsteff received the M.Sc. degree in physics from the University of Jena, Jena, Germany, in 1998, and the Ph.D. degree from the University of the Federal Armed Forces Germany, Munich, Germany, in 2002.

The main topic of his Ph.D. research was to develop control strategies for large-scale production processes using online infrared spectroscopy together with intelligent data interpretation techniques. The research was carried out at Siemens Corporate Technology, Munich. In 2002, he joined Philips Research, Eindhoven, The Netherlands, and has been focusing on biomedical sensors and measurements for monitoring solutions in clinical and in personal health care applications since that time. He is currently a Principal Scientist with the Patient Care and Measurement Group and an Owner of the Value Stream “Measurement innovations” with Philips Research.