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Blood Pressure Models for Wearable Sensors

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Abstract—As cardiovascular diseases are one of the most prominent illnesses, a continuous, non-invasive, and comfortable monitoring of blood pressure (BP) is indispensable. This paper investigates the best method for obtaining highly accurate BP values in non-invasive measurements when using an ultrasound (US) sensor projected for a wrist-worn device. State-of-the-art BP models were analyzed and qualitatively compared. Relevant arterial parameters such as luminal area, flow velocity and pulse wave velocity, of 729 subjects were extracted from a computer simulated database and served as input parameters for the wearable US. A linear *in-silico* model calibrated to each arterial-site revealed to be most accurate model. The linear model was used for the extraction of BP by using the US sensor and validated with a commercial pressure sensor in an *ex-vivo* experimental setup. The results showed an *in-silico* pulse pressure correlation of 0.978 and mean difference of (-2.134 ± 2.477) mmHg at the radial artery and *ex-vivo* pressure correlation of 0.994 and mean difference of (0.554 ± 2.315) mmHg. Thus, with the linear model, the US measurement complies with the Association for the Advancement of Medical Instrumentation standard with deviations lower than 5 mmHg.

Index Terms—arterial pressure waveform, continuous and non-invasive BP measurement, hypertension monitoring, pulse pressure, ultrasound sensor, wearable devices.

I. INTRODUCTION

HYPERTENSION is the most common cause of cardiovascular diseases (CVD) worldwide. The higher the blood pressure (BP), the higher the risk of damage to the heart and blood vessels in the major organs. Hypertension is known as the "silent killer", because most people might even not show any symptoms at all, causing around half of all deaths from CVD [1]. Therefore, the need for continuous monitoring of BP in the daily life is evident. In a clinical context, hypertension can be diagnosed when multiple readings of systolic blood pressure (*SBP*) and diastolic blood pressure (*DBP*) are above 140 mmHg and 90 mmHg, respectively [2]. Moreover, the BP pulse wave (PW_{BP}) presents a wealth of information in the cardiovascular system, providing remarkable insights for CVDs diagnosis and prognosis at an early stage.

The gold standard for BP measurement is sphygmomanometry at the brachial artery, which besides being uncomfortable when used for long periods of time due to the periodic inflation

and deflation of the cuff, it only provides discrete values of *SBP* and *DBP*. Contrarily, invasive BP measurement is continuous and accurate, but attains a high-risk as it comprises in the insertion of a catheter in the artery of the patient and should only be used in an intensive care unit. The ideal BP sensor for continuous measurement should allow long-term monitoring, be non-invasive, and easy to integrate with a remote healthcare system. There are several sensor-based techniques that try to answer the requirements listed above, such as the tonometer method, vascular unloading method, plethysmography (PPG), pulse transit time (PTT), and ultrasound-based method [3]–[6].

When using non-invasive methods of pressure measurement, various models arise that relate a hemodynamic parameter to BP. The correlation of blood flow and/or artery distension with BP and different mathematical models that relate these arterial variables to the arterial PW_{BP} have been demonstrated [7]–[11].

The ultimate goal is to develop a wrist-worn device, that examines hemodynamic changes in the radial artery and meets with the requirements presented previously, so that its user can continue to lead a normal life. Up to now, non-invasive standard BP measurements are performed in the brachial artery with a cuff. Although *DBP* and mean arterial pressure (*MAP*) do not change significantly through the arterial tree, and the calibration of these values taken from a different artery (i.e. brachial artery) than the artery under study (i.e. radial artery) is generally accepted, this leads to inaccurate readings and wrong clinical prognoses. Thus, due to effects of increased viscoelasticity, it is demonstrated that pulse pressure (*PP*), defined as $PP = SBP - DBP$, changes through the arterial tree. Intra-radial BP can present differences higher

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than 15 mmHg in comparison to the intra-brachial BP [12]. It is necessary to deduce an algorithm to continuously extract highly accurate SBP and DBP values from measurements in the radial artery.

This work performs an analytical analysis and comparative study between different models based on their physical principles, with their assumptions and simplifications explained. A computer-simulated arterial pulse wave (PW) database developed by Charlton et al. [13] was used to demonstrate how the pressure values vary between different arteries and to compare the response to the models at different measuring sites. In the *in-silico* study, accuracy of different mathematical models applied to the radial artery was investigated through parameters derived from the database. The best performing model was validated with a porcine artery and a customized ultrasound sensor. Data were post-processed in a custom analysis program written in the MATLAB environment. Finally, the accuracy of the *ex-vivo* measurements was studied through its comparison with a commercial sensor. The results showed promising advances and new insights for the extraction of vital parameters when using ultrasound sensors.

II. NON-INVASIVE CONTINUOUS BP MEASUREMENT

A. Tonometer Method

The tonometer method is based on the measurement of back-force at contact when the device is placed over a superficial vessel. Applanation of the vessel wall requires a stiff or bony structure as abutment. The change in pressure of the flow is measured while the (static) pressure is applied. Controlled force has to be applied to avoid occlusion of the blood vessel. Thus, accurate positioning over the measuring site is required. Applanation tonometry cannot be applied to obese subjects, as it needs lean skin to avoid cushioning of the pressure values. This method is highly sensitive to motion, proving impractical for daily activities [3], [4].

B. Vascular Unloading Method

Also known as volume clamp method, vascular unloading method relies on a finger PPG sensor that adjusts pressure in a small finger cuff and keeps the optically measured blood flow constant. The varying adjusted cuff pressure can then be correlated to the intra-arterial BP. Continuous measurement of BP is possible with this method, but the cuff is uncomfortable for long periods of time and there is a risk of venous congestion for the patient [3]–[5].

C. PTT Method

PTT is the time it takes for a PW to travel between two points of the cardiovascular system. The configuration that gives the most accurate SBP and DBP values is the PTT obtained between an eletrocardiogram (ECG) and a finger PPG. The recorded signals can be translated to BP through a calibration procedure. Compared to the previously presented methods, PTT does not need to partly occlude a vessel, but comes with an increase in the system's complexity, as it adds a second PW measuring site [3], [14].

D. Ultrasound-based Method

Assessment of the BP waveform via ultrasound (US) technique utilizes a high-speed acoustic probe to capture the pulsation of arteries. On the one hand, echo signals and Doppler shift principle have been used in this technique to detect the arterial wall displacement and blood flow, which with a mathematical model and signal conditioning reveal good accuracy in arterial measurements and is, therefore, convenient to use. On the other hand, it is necessary to maintain a stable contact between the probe and the measuring site in order to achieve a reliable acoustic coupling interface [4], [8]–[11]. Taking the indicated advantages and disadvantages into account, the US-based method seems to be the most promising method, when stable acoustic coupling is achieved.

III. PRESSURE WAVEFORM MODELS

For the application of a wrist wearable device to measure BP through an US-sensor, models that rely on the measurement of both hemodynamics (blood flow and arterial distension) of the arterial segment were analyzed and compared.

A. Model 1 (M1): Linear Relationship

Through a direct conversion, the diameter waveform, $D(t)$, can be projected to a pressure waveform [8]. The end-diastolic and mean arterial diameter, D_d and $\bar{D}(t)$, are calibrated to the brachial DBP and MAP, deriving the following conversion factor k :

$$k = \frac{MAP - DBP}{\bar{D}(t) - D_d}. \quad (1)$$

The diameter waveform is then converted to the pressure waveform by the factor k (1) and the slope intercept b calculated at the diastolic pressure,

$$P(t) = D(t) \cdot k + b. \quad (2)$$

Although a study revealed that this method of assessing BP may on average only underestimate invasive PP by 1.6 mmHg [15], this is not expected if the measurement is taken in the periphery. In normotensive subjects, the diameter of elastic arteries (e.g. carotid artery) change linearly with pressure, while peripheral arteries (e.g. radial artery) exhibit saturation at near-systolic pressure [16]. *In-vitro* and *in-vivo* studies have demonstrated that the diameter-pressure relationship exhibits an exponential characteristic [8], [16], [17].

B. Model 2 (M2): Exponential Relationship

Many researchers have studied the exponential relationship between arterial diameter distension and BP [7]–[9], [18]. After obtaining the arterial vessel cross-section area, $A(t)$, from $D(t)$, the BP waveform can be calculated as follows:

$$P(t) = P_0 \cdot e^{\alpha \left(\frac{A(t)}{A_d} - 1 \right)}, \quad (3)$$

with

$$\alpha = \frac{A_d \cdot \ln(P_s/P_d)}{A_s - A_d}, \quad (4)$$

where α is the vessel rigidity coefficient, P_0 and P_d is the diastolic pressure, P_s is the systolic pressure, A_d is the diastolic luminal area and A_s the systolic luminal area.

As (3) is valid over a large pressure range [8], α is pressure-independent and constant. To accurately determine α , it is worth noting that P_s and P_d must be obtained at the same position as $D(t)$. If the pressures are unknown, calibration to brachial sphygmomanometer could be done. Wang et al. [9] designed an ultrasonic device that uses the brachial calibration method with a measurement precision higher than 2 mmHg. However, it is known that P_s changes drastically through the arterial tree due to arterial stiffening and wave reflection phenomenon. Moreover, brachial P_s should not be used for calibration in other measuring sites, especially if the changes between systolic pressures are known to be very different as is the case between the brachial and radial artery [12].

In addition, an iterative correction of α was proposed by Meinders et al. [8]. For the first iteration, the rigidity coefficient, α_i with $i = 1$, is calculated as indicated in (4), taking use of brachial SBP as P_s . After determining $P_i(t)$, the measured mean pressure $\bar{P}_i(t)$ is compared to the MAP determined at the brachial artery. If $|\bar{P}_i(t) - MAP| > 0.01$ mmHg, a new α_{i+1} is calculated until the mean pressure of the projected waveform converges to brachial MAP . The iterative α_{i+1} is calculated as follows:

$$\alpha_{i+1} = \alpha_i \cdot \frac{MAP}{\bar{P}_i(t)}. \quad (5)$$

C. Model 3 (M3): Laplace's Law + Moens-Korteweg

An US-based method for the assessment of the PW_{BP} based on the integration of pressure was introduced by Vappou et al. [10]. The method relies on the measurement of local distension waveforms using US signals, together with the measurement of the local Pulse Wave Velocity (PWV) that leads to an estimation of the local stiffness. The theoretical principle combines Laplace's Law and the Moens-Korteweg (MK) equation.

Laplace's law,

$$dP = \frac{E \cdot h \cdot dR}{R^2}, \quad (6)$$

relates an infinitesimal variation of the lumen radius dR to the variation of internal fluid pressure dP , where E is the Young's elastic modulus of the arterial wall, h is the wall thickness and R is the artery's luminal radius.

The MK equation relates the PWV to the elasticity of the arterial wall,

$$PWV = \sqrt{\frac{E \cdot h}{2\rho \cdot R}}, \quad (7)$$

where ρ is the blood density.

By replacing the elasticity factor in (6) with E calculated from (7), and integrating it over the cardiac cycle, the PP waveform is given by

$$\Delta P = 2\rho \cdot PWV^2 \cdot \log\left(\frac{R(t)}{R_0}\right), \quad (8)$$

with $\Delta P = P(t) - P_0$, and P_0 and R_0 corresponding to diastolic values of pressure and artery radius, respectively. As

in most indirect BP measurement models, the methodology of Vappou et al. [10] only determines the ΔP , maintaining P_0 unknown. A conversion to absolute BP measurement is done by adding the DBP measured at the brachial artery to the ΔP calculated in (8).

In this study, a linear waveform calibration to brachial DBP and MAP was proposed. The calibration procedure takes the model's projected mean pressure $\bar{P}(t)$ (through integration over a cardiac cycle) and end-diastolic pressure P_{end} to calculate the calibrating factor m as

$$m = \frac{MAP - DBP}{\bar{P}(t) - P_{end}}. \quad (9)$$

The pressure waveform is then multiplied by the calibration factor and the slope intercept b is added (calculated for $P(t) = MAP$).

In the study with 11 subjects [10], good correlation ($0.94 < r < 0.98$) was found between the PP measured through radial tonometry using a commercially available system and the PP obtained through the method described above. The Bland-Altman study [10] revealed a positive bias of 4.7 mmHg and standard deviation of 4.45 mmHg.

D. Model 4 (M4): Bramwell-Hill Equation

Bramwell et al. [19] improved the MK equation such that the formula may be independent of Young's modulus E and thickness h of the vessel wall, directly relating PWV to the compliance dA/dP as follows:

$$PWV = \sqrt{\frac{A(t) \cdot dP}{\rho \cdot dA}}. \quad (10)$$

By rearranging (10), and assuming that PWV remains relatively constant during a cardiac cycle, a pressure waveform can be expressed in terms of PWV and change in cross-sectional area as [20]:

$$P(t) - P_0 = \rho \cdot PWV^2 \cdot \ln\left(\frac{A(t)}{A_0}\right). \quad (11)$$

It should be noted that the PWV is dependent on the compliance and area of an artery and it should be determined at the local site of diameter distension measurement [20].

The same calibration procedure as in M3 was applied to this model.

E. Model 5 (M5): Joukowsky's Equation + PWV

Soleimani et al. [11] developed a system where the internal diameter and blood velocity waveforms were extracted from consecutive sonograms. The authors assume that blood is an incompressible fluid traveling through a flexible tube, and pressure can be determined through the fundamental water hammer phenomenon. Thus, water hammer principle can be projected to the arterial vascular system, where the opening and closing of cardiac valves correspond to mechanical valves in a hydraulic piping system. The water hammer equation, known as Joukowsky's equation, measures the change in

pressure of a fluid resulting from a change in the fluid's velocity, taking into account its density and PWV,

$$\Delta P = \rho \cdot PWV \cdot \Delta v, \quad (12)$$

in which ΔP and Δv are changes in pressure and velocity, respectively, relative to the initial value.

Solving (12) with (10), and applying the increment of pressure and area instead of its derivative, results in

$$\Delta P = \rho \cdot (\Delta v)^2 \cdot \frac{A(t)}{\Delta A}, \quad (13)$$

where, once again, relative ΔP is obtained instead of the absolute BP. The brachial DBP is then added to the measured pressure and the absolute PW_{BP} is determined. Soleimani et al. [11] developed a subject-specific calibration procedure based on the correlation between $\bar{P}(t)$ calculated from the proposed model and $\bar{P}(t)$ calculated from tonometry measurements at the radial artery. The calibrated model was validated on 20 male subjects via a standard sphygmomanometer, revealing a PP correlation of 0.91 and mean absolute difference (through Bland-Altman analysis) of 1.333 ± 6.548 mmHg [11].

F. Comparison between models

It is worth noting the similarities between models M3 and M4. Both (8) and (11) are derived from the Bramwell-Hill equation that relates the MK equation to the Laplace's Law. However, when integrating over a cardiac cycle, Vappou et al. [10] proposed a model with the decimal logarithm, whilst the direct integration of the Bramwell-Hill equation applies the natural logarithm. Although it is expected that both models produce similar responses, the raw, non-calibrated BP values derived from (11) are higher than the ones obtained from (8). This changes after applying the corresponding MAP and DBP calibration, where both calibrated models exhibit the same curve. From here on, M3/4 refers to both calibrated models M3 and M4.

Table I summarizes the different model equations and the implemented calibration methods. All models are calibrated with DBP and MAP . In M2, the first iteration is calibrated with SBP . Calibrating values DBP , SBP and MAP can be taken from a brachial sphygmomanometer measurement.

IV. MATERIALS AND METHODS

A. Database

For the benchmarking of the different PW_{BP} models, the *open-source* simulated database of arterial PWs developed by Charlton et al. was used [13]. The computational model consists of three main components, based on the work of Alastruey et al. [21]. First, the arterial tree is decomposed into arterial segments modeled as thin viscoelastic tubes of constant length and linearly tapered diameter. Then, a periodic inflow waveform is introduced at the aortic root. Third, terminal three-element Windkessel boundary conditions are imposed at the outlets of peripheral arterial segments, modeling vascular beds. The database was validated by the comparison of PWs and hemodynamic characteristics with *in-vivo* data.

Fig. 1. Compliance curves (P/A) for (a) carotid, (b) brachial and (c) radial artery, extracted from the Charlton database [13] (for the baseline subject). The increase in hysteresis and PP towards the periphery due to an increase in stiffness is noticeable.

The primary parameters (flow velocity and luminal area) that can be directly extracted from ultrasound measurements were taken from the database, and then simulated in MATLAB for a comparative realistic analysis of the different five models. One of the most important simulations performed consisted of the compliance curves (arterial pressure versus luminal area), as it provided a valuable information about the viscoelasticity properties of the arteries. Fig. 1 shows the compliance curves derived from the baseline of a 25 year old virtual subject (one simulated subject created with the age-specific mean values for all variables, e.g. heart rate, stroke volume and arterial diameters). The viscoelasticity of the arterial walls leads to a hysteresis seen in the compliance curve. Furthermore, the composition of the arterial wall, in particular the elastin and collagen content, changes from central to peripheral arteries. Starting in the proximal aorta, elastin is the dominant component, but by the periphery collagen becomes the governing element [22]. As seen in Fig. 1, it is clear that there is a change in behavior and compliance through the arterial tree, so that the carotid artery presents an approximately linear relationship, whilst towards the periphery, in the radial artery, the hysteresis is much more accentuated due to the collagen content. Thus, when performing arterial parameters readings with ultrasound in the arterial tree (i.e. blood pressure measurements), changes through the different arteries are expected. Therefore, specific models or concrete stiffness coefficients should be applied to each local peripheral artery in the cardiovascular tree.

B. Models simulation

To evaluate and analyze the changes in the hemodynamic parameters in the arterial tree, the five BP models were studied at three main arteries: carotid, brachial and radial. For each measuring sites, luminal area PW , in combination with (i) PWV for M3 and M4, and (ii) flow velocity PW for M5, were extracted from the database and applied to each model. As DBP and MAP do not change significantly through the arterial tree, calibration was made to brachial DBP and MAP (additionally to SBP in the case of M2). The aim of calibrating the curves to the brachial values is to approximate the simulation to a practical situation where calibration of the measured hemodynamic PWs and values would be done using a brachial sphygmomanometer. Furthermore, in the sense of maintaining real-life application plausibility, brachial MAP was not calibrated to its gold standard - integration of BP curve

Fig. 3. BP models performance for the baseline subject at the (a) carotid, (b) brachial and (c) radial artery. All models are calibrated as described in section III. Notice that M1 and M3/M4 curves overlap. Curve *Db* refers to the database pressure waveform (*ground truth*).

performing model at the radial artery, revealing a correlation of 0.978 and mean difference of (-2.134 ± 2.477) mmHg.

After rejecting M5 because of its high error and low correlation with the *PP* extracted from the database, further error studies between the best performing models were done. *DBP* and *MAP* display the same error profile between the remaining models, since the models are calibrated from the same values (brachial *DBP* and *MAP*). The results of the error studies for 729 subjects are depicted in the boxplots in Fig. 4, where mean *PW* error, its *SD* and *SBP* error are included. Even though M2 always exhibits a lower mean *PW* error (carotid artery (c.a.)= -0.313 mmHg, brachial artery (b.a.)= 0.013 mmHg, radial artery (r.a.)= 1.160 mmHg), the *SD* (i.e. r.a.= 2.864 mmHg) and *SBP* error (i.e. r.a.= 7.301 mmHg) are higher. The mean *PW* error is the same in M1 and M3/4 at all sites (c.a.= -0.318 mmHg, b.a.= 0.023 mmHg, r.a.= 1.170 mmHg). In the carotid artery, M3/4 has a smaller *SD* and *SBP* error (1.002 mmHg and 1.252 mmHg, respectively) than M1 (1.008 mmHg and 1.505 mmHg, respectively). However, at the brachial and radial artery M1 has a lower *SD* (b.a.= 0.658 mmHg, r.a.= 2.013 mmHg) and *SBP* error (b.a.= 0.135 mmHg, r.a.= -0.151 mmHg) than M3/4 (i.e. *SBP* error: b.a.= -0.345 mmHg, r.a.= -0.657 mmHg). Thus, it is corroborated that in a 729 subject-study, M1 is the most accurate model for the radial artery.

B. Ex-vivo validation

For the *ex-vivo* validation, the linear model (M1) was selected as the best performing model and applied to the setup. Fig. 5 shows the results of the commercial pressure sensor measurements and the curve when applying M1 to the distension measurements acquired by the custom fabricated ultrasound sensor. The Pearson's linear correlation between

Fig. 4. BP models error ($n=729$) at the carotid, brachial and radial artery for the best models (M1, M2, M3/4). First and second row illustrates mean pulse wave error and its standard deviation. *SBP* error is shown in third row.

Fig. 5. Pressure waveforms obtained by the commercial pressure sensor and linear model (M1) during 2.5 seconds. Pump flow was set to 5 l/min. Pulse repetition frequency at 2 kHz.

the obtained PW_{BP} when applying M1 to the ultrasound measurement and the commercial pressure sensor was of $r = 0.994$ and mean difference of (0.544 ± 2.315) mmHg. The *SBP* error was of -2.131 mmHg.

VI. DISCUSSION

US-based methods are feasible in most vascular sites and cost effective, being promising as indirect, non-invasively, continuous, and comfortable means of obtaining arterial pressure waveforms, and providing anatomical and functional information about arterial health. Several models have been proposed that relate hemodynamic variables indirectly to BP. In the present work, five models that ultimately take the distending luminal area were studied *in-silico* through an arterial *PW* simulated database. To the best of our knowledge, this was the first time a comparison study of BP models was done in a large ($n=729$) and variable database.

All the studied models rely on the physical assumptions that the luminal area is a circle, neglecting the possibility of irregular and asymmetric area and, therefore, excluding the cases of highly torturous arteries. Additionally, the models assume linear elasticity, meaning that pressure and luminal area *PWs* have the same temporal profile. However, due to the natural viscoelasticity of arteries, this is known not to be true and it is corroborated in this study with the compliance curves and the time delay between the *ground truth* pressure and the calculated PW_{BP} from the different models. The time delay

TABLE II

PEARSON'S LINEAR CORRELATION AND MEAN DIFFERENCE BETWEEN PP MODELS. *Ground truth* PP (N=729) AT THE CAROTID, BRACHIAL AND RADIAL ARTERY.

	Carotid artery		Brachial artery		Radial artery	
	<i>r</i> -Pearson	Mean difference (mmHg)	<i>r</i> -Pearson	Mean difference (mmHg)	<i>r</i> -Pearson	Mean difference (mmHg)
M1	0.979	3.459 ± 1.560	0.989	0.080 ± 1.719	0.978	-2.134 ± 2.477
M2	0.970	6.317 ± 1.887	0.985	4.665 ± 1.511	0.969	5.661 ± 2.837
M3'	0.571	-17.758 ± 6.233	0.485	-23.578 ± 7.545	0.818	-30.665 ± 7.315
M3	0.980	3.167 ± 1.558	0.989	-0.428 ± 1.806	0.978	-2.655 ± 2.572
M4'	0.571	-1.475 ± 6.238	0.485	-1.907 ± 8.053	0.818	-2.032 ± 5.810
M4	0.980	3.167 ± 1.558	0.989	-0.428 ± 1.806	0.978	-2.655 ± 2.572
M5'	0.917	-12.179 ± 3.457	0.812	-14.525 ± 5.061	0.490	-16.318 ± 8.985
M5	0.917	7.791 ± 3.459	0.812	5.529 ± 5.059	0.485	3.853 ± 9.030

MX' identifies uncalibrated model.

is larger at the radial artery, due to the increase in collagen content and, consequently, viscoelasticity.

Another common limitation between the models is that all non-invasively acquired arterial pressure waveform require calibration to a known BP, imposing potential sources of measurement errors. It is generally accepted that *DBP* and *MAP* do not change considerably through the arterial tree, however, there are small changes, as seen in this study, introducing over- and under-estimation in *DBP* at the carotid and radial artery, respectively, when calibrated to brachial *DBP*. Furthermore, it is known that every formula to calculate *MAP* that uses a form factor (*FF*) that quantifies the change in the shape of *PW_{BP}* and relates its mean pressure to *SBP* and *DBP* have flaws. As the shape of BP changes along the artery tree due to propagation and reflection phenomena, it is understandable that a common *MAP* equation is not possible. Whilst a *FF* of 0.3 might be closest to the invasive measured *MAP* at the radial artery, the same could not be said about the brachial artery. A study by Papaïonnou et al. [23] demonstrated that the estimation of *MAP* using the *FF* value 0.412 provide results that are more accurate to invasive brachial *MAP* calculation when compared with *MAP* determined by the traditional formula, which uses *FF* = 0.3. In the current work, *FF* = 0.412 was used to best approximate brachial *MAP* to its true value.

The M5 pressure shape shown in Fig. 3 converges to the one displayed by Soleimani et al. [11], which is characterized by a distinct and separate two-peak waveform. All models except for M5 were able to approximate the shape of their BP waveform to the *ground truth*. The differences between the shapes can be appointed to M5 taking the flow velocity profile into account, while for the rest of the models the only time dependent variable is the change in luminal area and its shape is similar to the pressure PW. Additionally, it is worth noting that the calibration procedure proposed by the authors, that correlates a subject *MAP* values from its proposed method and values measured from the radial artery by tonometry over a group of cardiac cycles, could not be applied in the current study, as the database only supplies one cardiac cycle per subject. This underlying impracticability may be the subject of an increased *PW_{BP}* error, as the subject-specific calibration was done with the values supplied by Soleimani et al. [11].

The simulation study showed a slight difference in the best applicable model (M1) at different measuring sites. Thus,

M1 attributes to each artery distinct composition and, consequently, specific physical behavior. However, while in theory the carotid artery exhibits a more linear compliance behavior than the radial artery, M3/4 (logarithmic model) proved to have a lower error value than M1 (linear model). The contrary was seen at the radial artery, where although there is a stronger viscoelasticity and arterial rigidity, M1 turned out to be the model with the lowest error. Nonetheless, the differences in error between M1 and M3/4 are quite small (< 0.6 mmHg) and may be neglected.

The Association for the Advancement of Medical Instrumentation (AAMI) recommends a maximum mean difference of ±5 mmHg and standard deviation of ±8 mmHg between the standard and the novel method of pressure measurement (AAMI/ISO 81060-2:2013) [25]. Taking the database's *PW_{BP}* as standard, M1 and M3/4 accomplished this requirement, whilst M2 and M5 introduced higher mean difference and variability ((5.661 ± 2.837) mmHg for M2 and (3.853 ± 9.030) mmHg for M5, when applied at the radial artery), not complying to the recommendations. As the exponential relationship is the most used model for extracting pressure waveform from ultrasound measurements of arterial wall distensibility in known literature review [7]–[9], it was expected that M2 would reveal a better accuracy than M1 (linear model) but that was not the case. Whilst the linear model (M1) follows AAMI recommendations and has the best results in the brachial and radial artery, the exponential model (M2) metrics are above the recommended values and is the second worst performing model in the present study.

The *ex-vivo* study revealed a better correlation (*r* = 0.994) than *in-silico* (*r* = 0.978). This may be due to the fact that the models calibration in *ex-vivo* was done by using the in site pressure sensor's *DBP* and *MAP* recording. Thus, the calibration was done specifically to the measured artery (radial) and not to the brachial artery, which increases the error for comparison with realistic and practical measurements.

VII. CONCLUSION

The present study provides a novel *in-silico* analytical comparison between different models that relate arterial pulse waves to the pressure waveform for a continuous, non-invasive monitoring of arterial health parameters. The best model for the radial artery has been validated with an *ex-vivo* experiment, in which a porcine artery and an ultrasound sensor were used.

The *ex-vivo* experimental measurements demonstrated the high accuracy of the linear model based on the MAP calibration with extremely low error values of (0.544 ± 2.315) mmHg. This research introduces a new insight into the extraction of arterial parameters based on ultrasound, that will help the researchers to get much more precise clinical non-invasive readings. The changes in the pressure waveform through the arterial tree makes it necessary a site-specific model calibration method for a high accurate non-invasive sensor measurement.

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