

REVIEW ARTICLE

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Estimation of Blood Pressure by Image-Free, Wearable Ultrasound

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Abstract

1.28 billion adults worldwide have high pressure, and only 21% of people have it under control. If high blood pressure (BP) is not diagnosed and treated properly, then there is a high risk for cardiovascular diseases, the main cause of mortality. Ultrasound has emerged as a potential medical imaging system to monitor cardiovascular health. It is comfortable, noninvasive, safe, and a very well-established and well-known technology. However, there are currently no commercial single ultrasound systems to directly quantify vascular parameters, without the need for complex imaging algorithms, additional software-based solutions, and high energy demands that limit portable and prolonged measurements. In this paper, we present the steps to design an image-free novel ultrasound device to continuously extract vital parameters, with a focus on BP. Moreover, as the BP waveforms are amplified toward the periphery (i.e., from the aorta to the radial artery), we applied mathematical models and algorithms, specific to the site of measurement, to accurately extract BP. We validated these algorithms *in silico*, *in vitro*, and *ex vivo*, where tissue and artery phantoms help emulate human physiology. In young subjects, an *in silico* pulse pressure (PP) correlation of 0.978 and a mean difference of (-2.85 ± 2.57) mmHg at the radial artery were achieved. The *ex vivo* PP correlation was 0.986, with a mean difference of (1.72 ± 3.29) mmHg. Soon thereafter, *in vivo* measurements of BP and stiffness, and their correlation analysis will be performed to corroborate the accuracy of the developed proof-of-concept device.

Keywords Vascular ultrasound, Blood pressure, Stiffness, Patient-specific measurements, Mock circulation systems, Device validation

1 Introduction

Hypertension is a major cause of premature death worldwide and 46% of adults with hypertension are unaware of their condition [1]. By diagnosing and treating hypertension, many cardiovascular diseases can be prevented. Up till now, the “gold standard” of validated techniques for noninvasive reading of blood pressure (BP) requires blood flow to be blocked completely for the measurement. This is done through an arm cuff, placed around

the patient’s upper arm that is inflated with air. After blocking the artery, air is released, and values of systolic and diastolic BP are measured with a stethoscope and a manometer [2]. However, these devices often cause discomfort for those who need frequent BP monitoring, such as hypertensive patients or postoperative organ transplant patients. In such cases, BP must be measured every 30 min for 24 h, even when the patient is sleeping [3]. There is a significant need for a noninvasive, unobtrusive, and continuous BP monitoring approach, with a simplified remote healthcare data integration process.

Ultrasound is a real-time imaging modality that measures the wall locations of human vessels with high precision. The acoustic wave emitted by an ultrasound device produces an echo when an acoustic impedance difference

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between two materials occurs. In an artery, the acoustic wave propagates at different speeds between the muscular tissue (1580 m/s), arterial walls (intima layer 1590 m/s, inner and outer media 1626 m/s and 1699 m/s [4], respectively), and the blood (1575 m/s [5]) that transports the nutrients through the body.

Due to this effect, ultrasound transducers receive acoustic signals delayed in time (in the form of echoes), which recorded in time give us information about the mechanic changes of vessels in time. Ultrasound is also able to see the scattering of the blood cell particles, which is used to monitor blood hemodynamics. Using models and calibration values, those recordings are converted into arterial distensibility, stiffness index, compliance, and BP. Thus, key parameters to predict cardiovascular risks can be noninvasively and continuously measured via ultrasound [6–8].

On top of that, advances in new transducer technologies and microsystems make possible the fabrication of image-free ultrasound patches that reduce operator dependency, are wearable, comfortable, and allow for continued cardiovascular monitoring even while doing exercise [7]. Ultrasound therefore presents a promising technology for monitoring individual biomarkers as indicators of aging and cardiovascular disease risk.

However, ultrasound systems still face many challenges, including power-efficiency driving of transducers, low-noise recording of ultrasound echoes, high-attenuation medium (human tissues), and low quality of the images [9]. To better assess cardiovascular health and vascular age, we developed a low-cost and low-energy ultrasound

device and implemented patient-specific models to better derive BP from distension waveforms. To guarantee the accurate monitoring of vital parameters, we pay special attention to three main aspects (see Fig. 1): (01) Specification for device design and development with a compromise between resolution, patient safety, and energy saving; (02) Determination of the best artery model/algorithm to extract cardiovascular parameters specifically for each patient and arterial site; (03) Device validation in the laboratory (in vitro, ex vivo) and in a clinical environment (in vivo, in progress) attending to the specific medical device regulations and reference device used for validation.

This review summarizes my Career Development presentation at Artery 2023 (Bonn, Germany). It discusses the methods that we applied to develop a low-consumption, safe, operator-independent, and noninvasive proof-of-concept ultrasound device for accurate and patient-specific vital parameter monitoring. In Section II, the fundamentals of device development to measure arterial diameter are explained. In Section III, vascular models and algorithms to derive BP from the arterial diameter are introduced. In Section IV, the creation of mock circulatory loops (MCL) for in vitro and ex vivo validation is described. The review is finalized with an outlook of our results and the highlight of future research possibilities.

2 Ultrasound Device Development

The impact on the market and exponential growth of wearables, which is expected to reach USD 931.31 billion by 2030 [10], requires further research on the integration

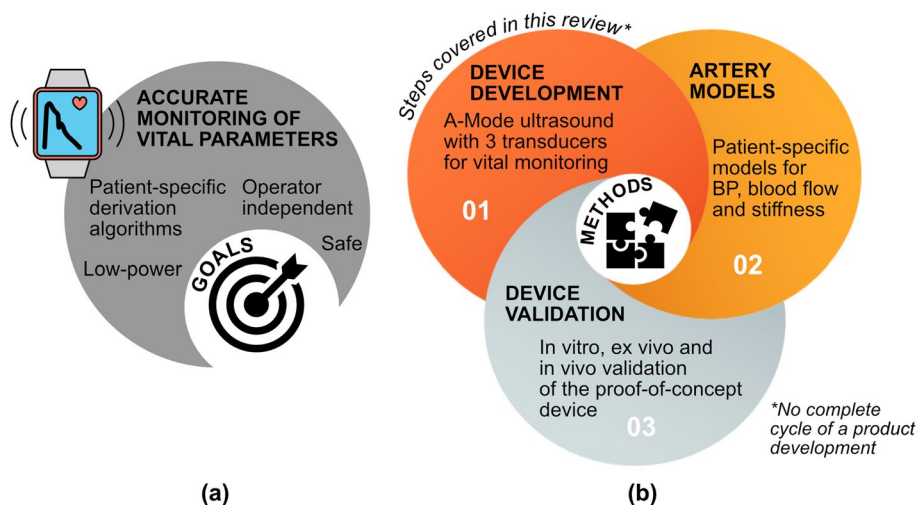


Fig. 1 **a** Goals needed to achieve accurate monitoring of vital parameters by employing portable devices. **b** Accuracy levels, patient comfort, and risks are affected by the methods used during device development (01), and the algorithms or models implemented to derive the cardiovascular parameters (02). Once the proof-of-concept device is available it is validated (03), first in a laboratory environment (in vitro and ex vivo), and later it will be done in animals or humans (in vivo) to assess its performance and validity as a medical device

of new microtechnologies and sensors into compact arm-bands and conformable patches for continuous monitoring of cardiovascular health. Specifically, ultrasound has been introduced in the wearable market through the creation of two new generations for the microfabrication of transducers: capacitive and piezoelectric micromachined transducers (CMUTs and PMUTs, respectively [11]). CMUTs are used for high bandwidth and resolution, while PMUTs offer their main advantage in terms of low-power requirements. Both technologies are easily integrated with microchips and show ongoing research and medical potential as stretchable patches and easy-to-wear wristbands [6, 7, 12–14]. Most of their applications are found in intravascular ultrasonography and endoscopy [11, 15].

In contrast to CMUTs and PMUTs, conventional piezoelectric-based transducers are the most widely chosen option for medical device production due to their mature technology, reliability, low cost, and broad knowledge of the fabrication process. In the market, a wearable piezoelectric-based ultrasound device for blood flow measurement is available [16]. Moreover, we demonstrated the functionality of a new bulk 5 MHz piezoelectric ultrasound probe [17]. The developed probe is composed of three sensors placed geometrically in a trapezoidal configuration. The sensor in the middle is used in amplitude mode (A-Mode) to detect the luminal diameter of the radial artery, whereas the other two transducers on the side are used for flow measurements based on the frequency shift produced by the movement of erythrocytes (Doppler mode).

In [17], we achieve precise measurement of arterial diameter in A-Mode by determining the difference in time of flight between echo signals from the anterior and posterior vessel walls. Thanks to the inclusion of a peak-detector receiver and high sample rates (125 Mega-Samples Per Second) during digitalization, we increase the vertical and the horizontal resolutions for accurate monitoring of diameter variations during the pulsation of the artery. Finally, we identified biomarkers of arterial hemodynamics by the application of signal processing, arterial models, and corresponding calibration parameters to the acquired raw data.

Important parameters obtained in the literature with image-free ultrasound devices are lumen diameter and arterial wall motion [18, 19], local and regional pulse wave velocity (PWV) [17, 20], intima-media thickness [17, 21] BP and compliance [6–8, 22], and blood flow [16]. Concretely, in [7], even continuous measurements of BP during exercise were possible thanks to the design of a conformal stretchable ultrasonic patch. In [6], BP measurements derived from in vivo ultrasonic acquisitions of arterial diameter distension achieved high

precision (within 2 mmHg) and high accuracy (grade A). Moreover, they also monitored the PWV using electrodes and the developed ultrasound device.

3 Derivation of Blood Pressure

The heart pumps blood through the arteries in our body in a pulsatile manner. Changes in blood flow velocity U contribute to continuous diameter D changes in our arteries. As seen in Fig. 2a, continuous acquisitions of U and D via ultrasound are converted into BP waveform $P(t)$ by applying specific arterial models. Those models require initial BP calibration values, obtained with measurements at the left brachial artery by the gold standard cuff sphygmomanometer (see Fig. 2b).

Multiple models to derive BP can be found in the literature [23–27]. All of those models assume that the artery is rotationally symmetrical so that the area waveform in time $A(t)$ is equivalent to:

$$A(t) = \frac{\pi \cdot D(t)^2}{4}. \quad (1)$$

Due to the pulsatile characteristics of the artery, $A(t)$ changes in each heart cycle and acquires minimum and maximum values, named the diastolic and systolic luminal areas (A_d , A_s), respectively.

In our research in [28], we compared the state-of-the-art transfer functions to derive BP. Most of those transfer functions are based on the A-Mode ultrasound signals (diameter distension), while others also include pulse wave Doppler signals (blood velocity) as input (see Fig. 2a). We showed that BP can be accurately derived from the diameter distension acquired by ultrasound by applying three best-performing models: linear model (M1), exponential model (M2), and Bramwell–Hill model (M3) [28]. The simplified equations of these three models are included in Fig. 3a. Concretely, in M3, the local PWV in the Bramwell–Hill equation can be extracted by the diameter-blood flow velocity (In DU-loop) method and the flow-area (QA-loop) method as explained in [29] and [30]. A non-invasive method to measure local PWV is using ultrasound-based pulse wave imaging [31].

Calibration also plays an important role in accurate BP measurements. In Fig. 3a, the calibration factors for models M1, M2, and M3 are indicated as k , α , and m , respectively. In M1 and M3, the conversion factors are calibrated to the MAP and DBP and use a linear calibration [28]. However, in M2, the factor α (stiffness coefficient) is determined by the cuff measurements of DBP and SBP (P_s and P_d), at the brachial artery. Because P_d and MAP are relatively constant throughout most of the arterial tree [29], but P_s changes significantly, a correction factor of α (α_i in Fig. 3), is applied for a certain number of iterations i until the mean pressure $\overline{P_i(t)}$ of

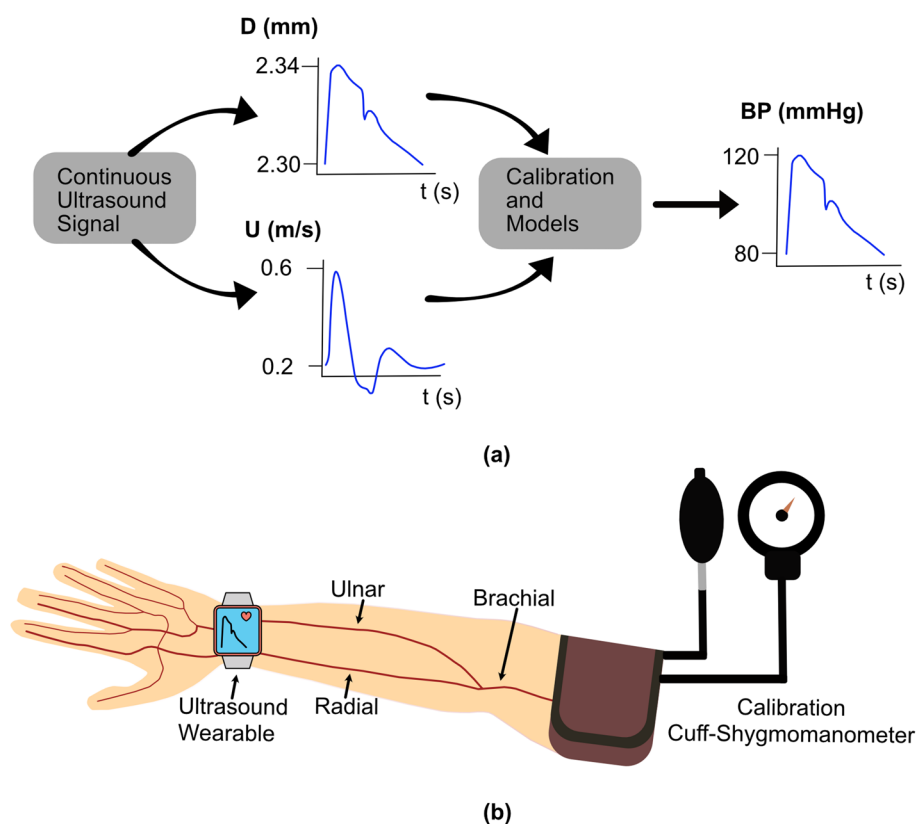


Fig. 2 Illustration of the concept of the derivation of BP from the distension and blood flow velocity waveform using ultrasound. **a** The acquired diameter and/or velocity waveform from the sensor is converted into the pressure waveform by applying validated models and initial mean arterial pressure (MAP) and diastolic BP (DBP) values. **b** Schematic of the wearable ultrasound that acquires the physiological signals of either the ulnar or radial artery. Simultaneously, the MAP and DBP values are measured at the brachial artery using the gold standard cuff sphygmomanometer for calibration

the projected waveform converges to the brachial MAP. For more information about this correction, please refer to [23].

In [28], we compared in silico the three best-performing models (M1, M2, and M3) for young (25- and 35-year-old) subjects ($n=1458$) at carotid, brachial, and radial arteries. The in silico data was obtained from an open-source database and was validated with in vivo measurements [32]. Results are shown in Fig. 3b [28]. This comparison is based on PP and SBP differences from the novel device to the reference device (ground truth) extracted from Bland–Altman plots. The mean values of the difference are shown in the graphs on the left, while the standard deviations (SD) of the SBP and PP differences are included in the graphs on the right. From the obtained results, we concluded that when measuring the carotid artery in a healthy young subject, the model that better approaches the ground truth is M3, with SBP and PP differences of 2.64 ± 1.79 mmHg and 0.64 ± 1.73 mmHg, respectively. However, when measuring at the brachial or radial artery, the linear model

(M1) is more accurate, with SBP and PP differences of: -0.76 ± 2.21 mmHg and -2.85 ± 2.57 mmHg (radial); and -0.57 ± 1.98 mmHg and -0.57 ± 1.99 mmHg (brachial). As shown in Fig. 3b, those values are within the recommended values from the Association for the Advancement of Medical Instrumentation (AAMI).

The AAMI recommends for BP cuff devices a maximum mean difference of ± 5 mmHg and SD of ± 8 mmHg between the reference and the proof-of-concept device [33]. As seen in Fig. 3b, BP measurements of young subjects at the carotid or radial artery derived from the M2 model (exponential relationship) provide higher errors in SBP and PP than the maximum allowed by the AAMI. Surprisingly, the exponential model is still the most extended in the literature and is mostly used for BP determination when using ultrasonic devices. This may be due to its better accuracy for aged (middle and late adulthood) subjects and to the tube law employed in the 1D model, which has a profound effect on the pressure estimation. The application of other 1D models may lead to other conclusions. Thus, the validity of the proposed

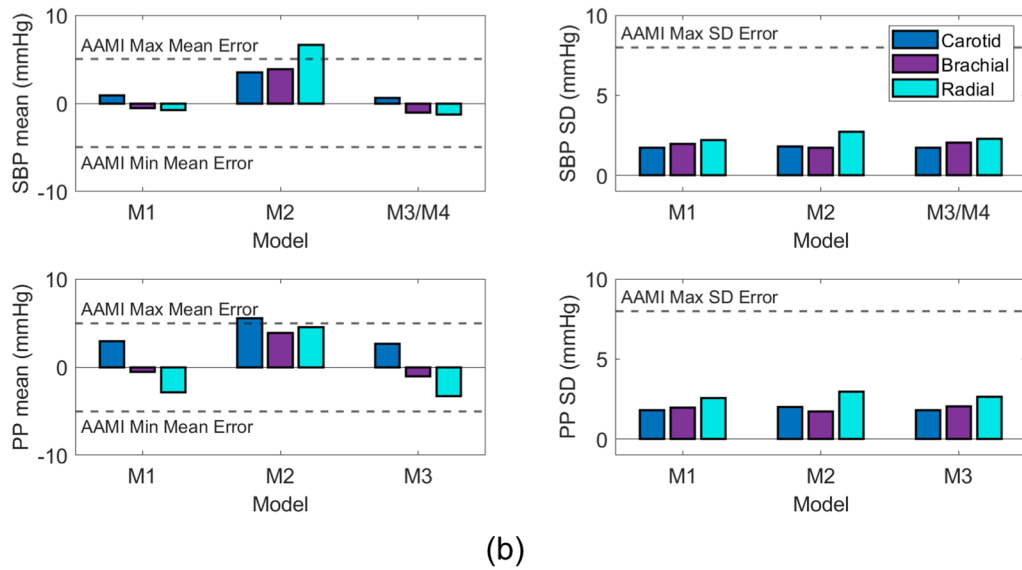
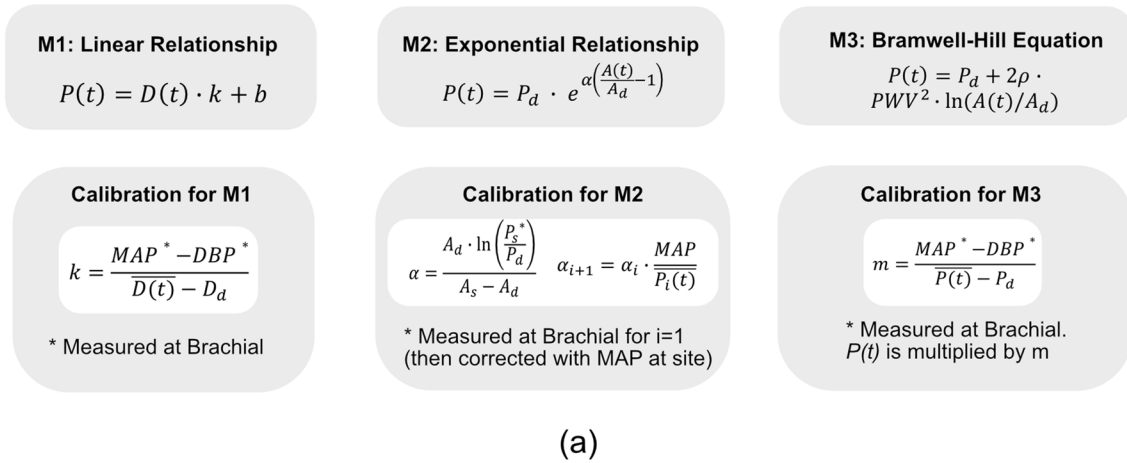


Fig. 3 Comparison of the three best-performing models (M1, M2, and M3) for BP acquisition with ultrasonic devices from our previous publication [30]. **a** Transfer BP and calibration functions for the three best-performing models. The exponential model is calibrated through a previous determination of the stiffness coefficient. PWV in M3 is the local pulse wave velocity, and ρ is the blood density. **b** Mean pulse pressure (PP) and systolic BP (SBP) difference between the three models at three main peripheral arteries (carotid, brachial, and radial) for a young population with a size of $n = 1458$

models M1-M3 still requires in vivo data for a better and more accurate physiological validation.

4 Proof-of-Concept Validation

The emergence of new cutting-edge technologies to continuously measure vital parameters and prevent cardiovascular risks requires the validation of proof-of-concept devices at different stages of development [34]. In vivo experiments provide the most valuable information about the performance of the device. However, they are very expensive for iterative optimizations

or modifications required in the development of a new medical device [35]. Thus, before performing in vivo measurements in animals or humans, the performance of the new device needs to be demonstrated in a laboratory environment (technology readiness level 4, TRL4 [36]). Hemodynamic evaluations of flow and distension at peripheral arteries and central arteries (i.e., aorta) can be modeled in vitro by implementing mechanical MCL [37]. MCL evaluates the human cardiovascular system and offers the advantage that complications in the circulation and potential risk of cardiovascular diseases, such as atherosclerosis, stenosis, and changes in

the arterial structure, can be mechanically modeled and controlled with high repeatability.

A simple MCL to measure heart rate, BP, compliance and resistance of the arteries can be composed of the following mechanical elements: pressure motor or peristaltic pump to mimic the heart function, bubble trap to remove the bubbles produced by the pump, variable valves to reduce or increase the peripheral resistance, tubes to emulate arteries, compliance chamber to simulate arterial compliance by changing the air pressure column on top, and various reservoirs. The main challenge of MCLs is the design of patient-specific mimicked arteries and tissues that possess the same characteristics as humans or animals. For this purpose, phantoms are widely used. The phantom design for validation of a new device needs to resemble the human tissue as it is seen by the wave that is going to be propagated through it. This means that if the device to be evaluated is an ultrasonic device, the phantom needs to have the same acoustic impedance, speed, and attenuation as the human tissue.

For the validation of our ultrasonic device, we tested multiple phantoms [38]. Agar, gelatin, gel-wax, and polyvinyl (PVA)-based phantoms are the most commonly used [39]. Gel-wax, agar, and gelatin phantoms are low-cost, easy, and fast to develop. However, they are not long-lasting, and their properties change very fast in time, so they can be used only for some days. As we demonstrated in [38], PVA-cryogel is the most preferred alternative for ultrasound applications due to its stability and very optimal acoustic attenuation. In addition, its mechanical and speed properties can be easily tunable by freeze-drying processes, changing concentration, or adding other chemicals. PVA can be made transparent by adding dimethyl sulfoxide, which is of high advantage when visualization of the flow in the phantom is needed. The limitation is that the viscoelastic properties of the arterial walls cannot be modeled with those phantoms. Still, the elastic modulus can be tuned with the freeze-drying process [40].

Multimodal phantoms that can be used for more than only one signal modality provide the advantage that they can be applied in MCL for the validation of various devices, or for one single device that integrates two or more sensor technologies [41].

We also used extracted arteries from animals provided by slaughterhouses instead of phantoms for a better evaluation of the ultrasonic device. In this case, we are talking about ex vivo cardiovascular setups. The advantage is that they provide a more realistic mimicking of the non-linear mechanics of the tissue.

In [17, 28], we validated in silico, in vitro, and ex vivo our newly developed US device. In young subjects, an

in silico pulse pressure (PP) correlation of 0.978 and a mean difference of (-2.85 ± 2.57) mmHg at the radial artery were achieved. At the same artery, the ex vivo PP correlation was 0.986, with a mean difference of (1.72 ± 3.29) mmHg. The next steps are being done now for the in vivo US validation.

5 Outlook

The development of new noninvasive, portable, and compact cutting-edge devices offers the possibility to continuously monitor BP comfortably and safely. Ultrasound is a technology that is widely used in clinics, is very well known, and can potentially be used to non-invasively measure vital parameters for in-home monitoring. However, limitations in power consumption, reduced accuracy affected by the arterial-site measurement, placement of the device, and the challenge of creating patient-specific algorithms to derive multiple vital parameters still require further research. New technologies can be further expanded to obtain a better penetration depth and amount of signal transmitted to the tissue. Furthermore, specific algorithms applied according to the arterial age and sex of the patient significantly improve the accuracy of BP derivation. On top of that, MCL setups using mechanical in vitro and ex vivo approaches are simplified solutions for iterative modifications and improvement of medical devices, and help emulate arterial hemodynamics and cardiovascular risk conditions before proceeding with in vivo trials.

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Author Contributions

Ana Belen Amado Rey is the full contributor of this review paper.

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Availability of Data and Materials

The datasets generated and/or analyzed during this review (Fig. 3) are available in Table II of this published article: A. C. Gonçalves Seabra, A. F. d. Silva, T. Stieglitz and A. B. Amado-Rey, "In Silico Blood Pressure Models Comparison,"

in *IEEE Sensors Journal*, vol. 22, no. 23, pp. 23,486–23,493, 1 Dec.1, 2022, <https://doi.org/10.1109/JSEN.2022.3215597>.

Declarations

Conflict of Interest

The authors declare that they have no competing interests.

Ethics Approval and Consent to Participate

Not applicable.

Consent for Publication

Not applicable.

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