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Wearable PWV technologies to measure Blood Pressure: eliminating brachial cuffs

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Abstract— **The clinical demand for technologies to monitor Blood Pressure (BP) in ambulatory scenarios with minimal use of inflation cuffs is strong: new generation of BP monitors are expected to be not only accurate, but also non-occlusive. In this paper we review recent advances on the use of the so-called Pulse Wave Velocity (PWV) technologies to estimate BP in a beat-by-beat basis. After introducing the working principle and underlying methodological limitations, two implementation examples are provided. Pilot studies have demonstrated that novel PWV-based BP monitors depict accuracy scores falling within the limits of the British Hypertensive Society (BHS) Grade A standard. The reported techniques pave the way towards ambulatory-compliant, continuous and non-occlusive BP monitoring devices, where the use of inflation cuffs is drastically reduced.**

I. Non-occlusive blood pressure measurement

For more than one century, the non-invasive measurement of Blood Pressure (BP) has relied on the inflation of pneumatic cuffs around a limb, typically the upper arm [1]. In addition to being occlusive, and thus cumbersome, cuff-based methods provide only intermittent BP readings, i.e. every twenty minutes, hence impeding the suitable monitoring of short-term BP regulation mechanisms. Moreover, cuff-based methods may not yield representative BP during sleep as repeated inflations induce awekenings, leading to non-representative overestimated BP values. Therefore, the development of novel technologies that reduce the recurrent use of pneumatic cuffs is clearly justified.

II. The pulse wave velocity principle

At each cardiac cycle, the opening of the aortic valve generates a pressure pulse that propagates along the walls of the entire arterial tree. The velocity at which this wall-distending wave propagates is referred to as Pulse Wave Velocity (PWV), and in humans typically ranges from 4 m/s in large elastic arteries, to 30 m/s in small muscular arteries [2]. Already in 1905 it was observed that PWV depended on BP, and half a century later a mathematical model was proposed in order to describe such a relationship. With the so-called Moens-Korteweg equation [3], this model predicts that in an elastic tube (say, an artery) PWVdepends on the diameter d, thickness h and stiffness E of its wall, as well as on the density of the filling fluid ρ, i.e.:

$$\mathrm{PWV} = \sqrt{\frac{hE}{d\rho}}$$

In a living cardiovascular system, increasing BP will distend the walls of its elastic arteries, augmenting their stiffness, and thus increasing PWV. According to this simple model, a unique BP-PWV relationship is to be expected in an elastic artery (as illustrated in Figure 1).

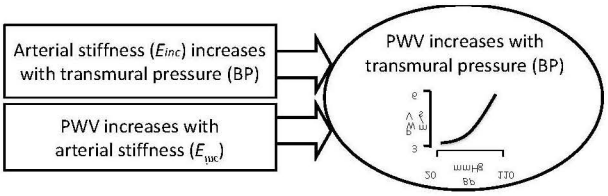

Figure 1. In elastic arteries, a unique relationship exists between measured PWV values and mean BP.

Unfortunately, and as depicted by the Moens-Korteweg model, the BP-PWV relationship is affected by several additional contributing factors, and a simple vasoconstrictor maneuver such as smoking a cigarette might modify it [4]. Said dynamic effect is particularly accentuated in those segments of the arterial tree depicting vasomotion, for which the BP-PWV relationship is not unique and might be time variant (as illustrated in Figure 2).

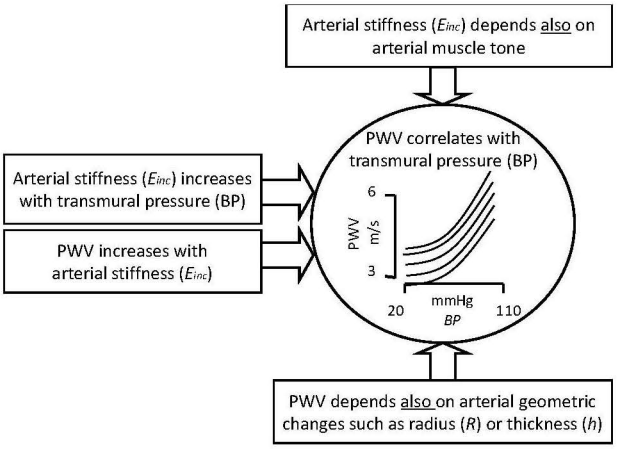

Figure 2. In muscular arteries, vasomotion continuously modifies the BP-PWV relationship. The use of PWV-based techniques to estimate BP is thus compromised.

The interest of constructing BP monitors upon the PWV principle is justified by the fact that there exist technological solutions to measure PWV in a beat-by-beat and non-occlusive manner [5]. Therefore, after establishing a subject-dependent BP-PWV relationship, beat-by-beat and non-

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occlusive BP estimates can be obtained from PWV measurements. A good illustrative development of the state of the art is that proposed in [6]. Assuming that the factors that interfere with the BP-PWV relationship have slow time dynamics, *i.e.* slower than actual blood pressure changes, Chen *et al* investigated the use of a hybrid BP monitor: while a brachial oscillometric cuff regularly inflates to perform reference BP measurements, continuous PWV measurements are performed through a non-occlusive fingertip sensor. Chen then proposed to use the measured PWV series to interpolate BP intermittent readings in a beat-by-beat basis. Such a setup still demanded the recurent use of brachial cuffs, but was a first step towards the non-occlusive beat-to-beat BP monitoring.

III. Pulse Wave Velocity Glossary

Since Chen's pioneering works, a large body of literature on PWV-based BP measurements has been released. For the sake of clarity, a list of common terms is here provided:

- Pulse Wave Velocity (**PWV**): propagation velocity of a pressure pulse along a segment of the arterial tree (m/s).
- Local Pulse Wave Velocity: propagation velocity of a pressure pulse at a given particular location of the arterial tree (m/s).
- Segmental Pulse Wave Velocity: propagation velocity of a pressure pulse along a given concatenation of heterogeneous segments of the arterial tree (m/s).
- Pulse Arrival Time (**PAT**): arrival time of a pressure pulse at a given location of the arterial tree (s). Given a cardiac cycle, PAT is typically defined as the delay between the R-Wave of an Electro-Cardiogram (ECG) and the associated pressure pulse arrival event at that particular location.
- Pre-Ejection Period (**PEP**): time delay between the onset of the Q-Wave of the Electro-Cardiogram (ECG) and the associated opening of the aortic valve (s). Because of implementation issues, the R-Wave is typically used instead of the Q-Wave.
- Pulse Transit Time (**PTT**): time required by a pressure pulse to propagate along a given segment of the arterial tree (s). Given a proximal location p and a distal location d, one defines $PTT = PAT_d - PAT_p$. In some setups, given a single location in the arterial tree, one defines $PTT = PAT - PEP$.
- Propagation Distance (**D**): length of the arterial segment involved in a PWV measurement (m). Therefore, $PWV = D/PTT$.
- Calibration function (**CAL**): subject-dependent BP-PWV relationship. Depending on the segment of the arterial tree involved in the measurement, CAL might be time-variant or time-invariant.

IV. Peripheral versus Central Measurements

As illustrated in Section II, time-invariant BP-PWV relationships exist in a living cardiovascular system, provided that only arterial segments depicting no vasomotion are involved in the measurements. In other words, in order to minimize the frequency of recalibration maneuvers required to obtain accurate PWV-based BP estimates, one must rely on PWV values measured on central, *i.e.* elastic, arteries.

Historically, the first generation of PWV-based BP sensors relayed on peripheral PTT measurements. For this generation, PAT values were typically assessed in superficial arteries in anatomical locations such as the wrist [7, 8], ear [9], or forehead [10]. A brachial cuff provided then reference BP measurements to intermittently estimate a CAL function. The main advantage of said first generation was that peripheral PAT values could be easily assessed via simple technologies such as Photo-Plethysmography (PPG), applanation tonometry or Impedance-Plethysmograpy (IPG) [11]. The main limitation of this approach was obviously that the associated CAL functions were time-variant due to vasomotion and hydrostatic effects [8], leading to a need of continuous CAL function re-estimation. Examples of first generation developments are: [6, 12-15]. Note that for some of these examples, PTT is simply approximated by PAT, thus underestimating the influence of PEP variability [16].

A second generation of PWV-based BP sensors has been recently proposed relying on central PTT measurements [11]. While said generation requires more complex measuring technologies such as Electrical Impedance Tomography (EIT) or opto-electrical hybrid sensors, it presents the main advantage of reducing the need of recurrent CAL re-estimations because of the lack of vasomotion perturbations. Figure 3 illustrates the benefits of second generation sensors.

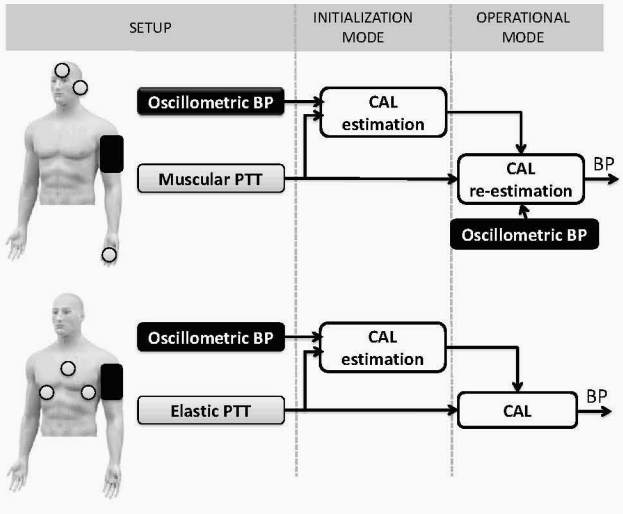

Figure 3. Upper panel illustrates first generation of PWV-based BP sensors relying on the measurement of peripheral (muscular) PTT values: re-estimation of CAL function is intermitently required. Lower panel illustrates **second generation** of PWV-based BP sensors relying on central (elastic) PTT values. CAL function is only estimated during the initialization mode, **eliminating the use of brachial cuffs in operational mode**.

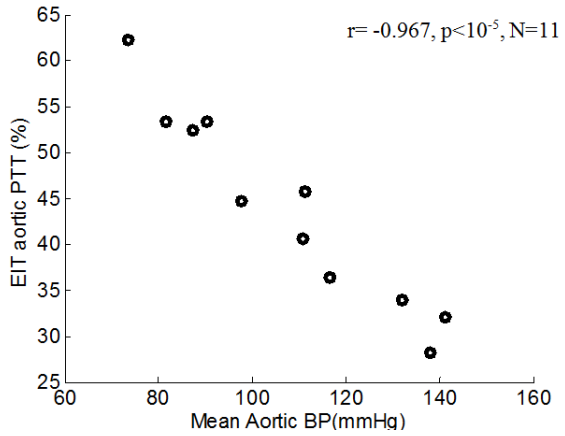

Figure 4. Correlation plot comparing EIT-derived aortic PTT values with invasive mean aortic BP values. While hypotensive conditions are associated to high aortic PTT values (decreased PWV), hypertensions conditions are associated to low PTT values (increased PWV).

V. Implementation Examples

In the following two particular examples of second-generation PWV-based BP sensors are provided, together with experimental data.

A. *Electrical impedance tomography chest sensor*

The first example of sensor to estimate BP via central PWV measurements is based on the so-called Electrical Impedance Tomography (EIT) technology [17]. The approach consists of placing an electrode belt around the thorax, the belt containing a set of electrical contact points with the skin (typically 16 or 32). Impedance measurements are then performed by injecting electrical currents between pairs of electrodes, and by reading the generated voltages between two different electrodes. From the measured impedance values, a tomographic image is generated: the tomographic image estimates the most likely distribution of impedance volumes within the thorax that generated such measurements [18]. By repeating this procedure at a high rate (up to 50 times per second) a sequence of EIT images is obtained.

From an electrical perspective, the thoracic cavity is composed of distributed impedance volumes. While the lungs (filled with air) form high impedance volumes, the heart and blood vessels (filled with blood) form low impedance volumes. During each cardiac cycle, vascularised structures within the thorax receive bursts of electrically conductive blood, which decrease the local impedance. Hence, when looking at a sequence of EIT images, local pulsations of the impedance signal (impedance pulse) are observed which are associated with the underlying pulsation of the blood (pressure pulse) [11].

Via the analysis of the arrival time of impedance pulses at different thoracic locations, pixel-wise PAT-based images are created: a PAT-based image assigns to each pixel a time value corresponding to the arrival time of a pressure pulse at the anatomical location represented by said pixel [19]. Via an unsupervised method described in [20], pixels depicting pulsatility information of the descending aorta are then identified. Impedance pulses from the aortic Region of Interest (ROI) are finally analysed as if a catheter was placed within the descending aorta, providing non-invasive estimates of aortic PAT.

In order to prove the feasibility of the novel approach, different hemodynamic conditions were induced in an anesthetized pig [21]. After inserting an arterial line into the ascending and descending aorta, noradrenalin and nitroglycerin were administrated in order to create BP modifications. Arterial-line time series provided both reference BP values, and PAT values at the ascending and descending aorta. In a retrospective analysis, aortic EIT-PTT values were determined as the delay between the opening of the aortic valve and the arrival of pressure pulses at the aortic ROI within the EIT plane.

For 11 experimental conditions, with mean aortic BP ranging from 73 to 141 mmHg, significant correlation ($r = -0.97$, $p<0.00001$) between aortic EIT-PTT and central BP was observed (Figure 4).

B. *Opto-electrical chest sensor*

The second example of sensor estimating BP via central PWV measurements is based on a novel opto-electrical sensing approach [22]. The approach consists of estimating two thoracic PAT values: the opening of the aortic valve (PAT_p) and the arrival of the pressure pulse at the sternum subcutaneous vasculature (PAT_d). While PAT_p is estimated based on the Impedance Cardiography (ICG) technology [23], PAT_d is estimated by the analysis of a multi-channel Photo-Plethysmograph (PPG) sensor [24].

The feasibility of the approach was tested in a cohort of 15 healthy male subjects, involving a total of 462 reference BP and chest PTT measurements. BP reference values corresponded to mean arterial pressure values as measured by a brachial cuff. Each subject was recorded at three different days: D, D+3 and D+14. Approval of University of Lausanne ethical committee was obtained for this study. In overall, the implemented protocol induced BP values to range from 80 ± 6 mmHg in baseline, to 107 ± 9 mmHg during isometric handgrip maneuvers (see Figure 6).

The cumulative percentage of BP values provided by the chest sensor falling within a range of ±5 mmHg compared to reference BP readings was of 70%, within ±10 mmHg of 91%, and within ±15mmHg of 98%. A mean BP reading error of 0.7 ± 5.1 mmHg was observed. These results point at the fact that the chest sensor complies with the British Hypertension Society (BHS) requirements of Grade A BP monitors, when applied to MAP readings (Figure 5).

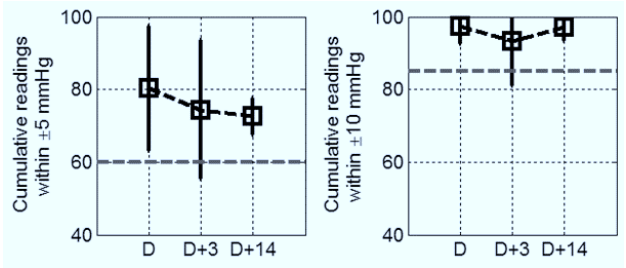

Figure 5. Day-dependent evaluation of the BHS criteria for the opto-electrical sensor. Error plots depict mean and standard cumulative percentage of readings for all subjects enrolled in the study. Gray dashed lines set BHS' minimum criteria for Grade A compliance. Reference BP values correspond to Mean BP as measured by a brachial cuff.

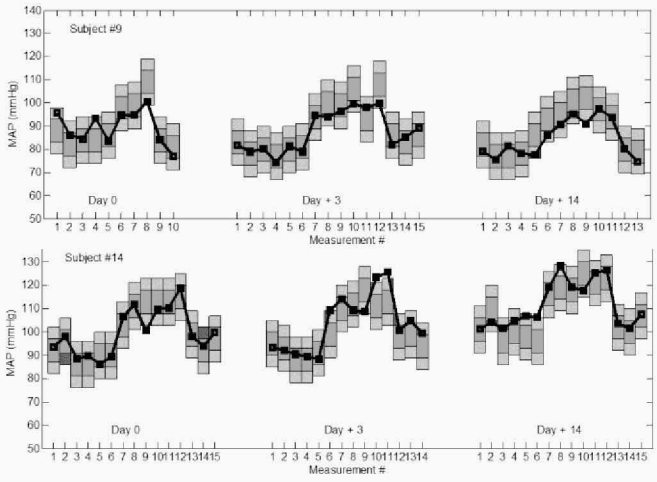

Figure 6. Two examples of the performance of the BP opto-electrical sensor. Black curve depicts Mean Arterial Pressure (MAP) as estimated by the chest sensor. Dark boxes comprise reference MAP by an oscillometric device ±5 mmHg, and light boxes reference MAP ± 10 mmHg.

VI. Conclusions

In order to respond to the demand of continuous, non-invasive, and non-occlusive BP measuring techniques, new approaches overcoming the limitations of brachial inflation cuffs are required. In this paper we reviewed the potential of a new generation of sensors based on the measurement of central PWV values. Because of its non-occlusiveness, this new generation of sensors is prone to be easily integrated in ambulatory and wearable measurement setups, as depicted in Figure 7.

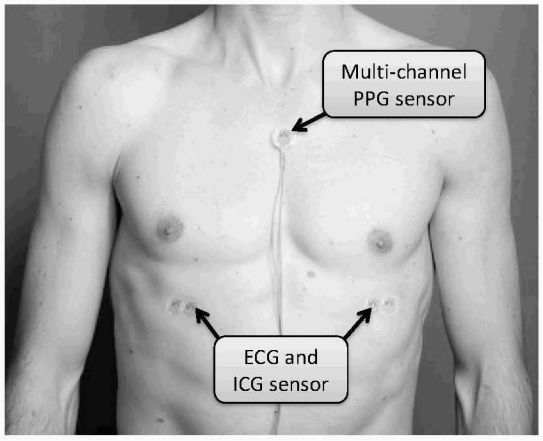

Figure 7. Example of implementation of opto-electrical sensor for the continuous non-occlusive estimation of blood pressure. All sensing technologies are implementable in ambulatory and wearable setups.

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