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# THEORETICAL STUDY OF DISCREPANCIES IN ESTIMATION OF MEAN BLOOD PRESSURE BY MAREY'S CRITERION

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**Abstract.** A simulation study was performed to evaluate how the arterial pressure–volume (*P–V*) relationship and blood pressure pulse height and shape affect the accuracy of non-invasive blood pressure measurement in fingers by applying Marey's criterion. The obtained results demonstrate that the pressure, estimated by this criterion, does not exactly agree with the true mean blood pressure defined as pressure corresponding to the maximum arterial compliance. The error may exceed 20 mmHg depending on the form of the arterial *P–V* diagram, arterial pressure pulse height and shape.

**Key words:** arterial pressure, arterial compliance, arterial pressure–volume relationship, oscillometric blood pressure, maximum oscillation criterion, Marey's criterion, non-invasive measurement.

## 1. INTRODUCTION

The method most widely used by physicians for non-invasive blood pressure measurement has been the auscultatory method, based on the detection of Korotkoff sounds. In recent years, however, increasing attention has been paid to the oscillometric technique, since it can be easily automated. Oscillometric method was first introduced by the French physiologist Etienne-Jules Marey [1]. Marey placed the arm into a compression chamber and observed that the chamber pressure fluctuated with the pulse and that the amount of pulsation varied with the pressure. He concluded that the arterial wall becomes tension free at the counterpressure corresponding to maximum pulsations.

This principle of measuring internal blood pressure by applying external pressure to obtain an unloaded (tension free) condition of the arterial wall was afterwards named Marey's principle and the maximum oscillation criterion (MOC) is known as Marey's criterion. Subsequent experimental *in vivo* and *in vitro* studies, concerning the relationship between arterial pressure and counter-pressure for maximum oscillations, have confirmed that maximum oscillations occur when the mean cuff pressure is approximately equal to the mean arterial pressure [2–9]. Yamakoshi et al. [6] reported that the cuff pressure, revealing the maximum pulsation amplitude, showed a close agreement with the corresponding mean arterial pressure, recorded simultaneously by catheterization in the brachial artery. Mauck et al. [7] analysed theoretically and experimentally the relationship between the cuff pressure for maximum oscillation and the true mean arterial pressure and found that the two pressures coincide, provided the cuff–air volume is kept sufficiently low. It has been commonly accepted that mostly the peak of the oscillometric curve (OC) corresponds to the mean intraarterial blood pressure ($P_{\text{mean}}$).

Some recent studies [10,11], however, have shown that non-invasive estimates of the blood pressure may be influenced by mechanical properties of the arterial wall, pulse amplitude of the blood pressure ($P_{\text{pulse}}$), and pulse shape. Some clinical investigators have also reported differences between the results of the individual invasive measurement of the $P_{\text{mean}}$ and the corresponding values estimated by MOC [12].

This study was initiated by our results obtained by simultaneous non-invasive beat-to-beat recording of $P_{\text{mean}}$ in fingers by two different methods, the volume clamp [13,14] and differential oscillometric method [15,16]. The differential oscillometric device estimates $P_{\text{mean}}$ by the maximum value of oscillations in a finger cuff (Marey's criterion), while the volume clamp monitor Finapres calculates $P_{\text{mean}}$ by integrating the pressure wave over the cardiac cycle. The setpoint of Finapres (transmural pressure $P_{\text{transm}} = 0$) is adjusted according to the shape of plethysmographic oscillations [17], without considering MOC. In most cases (rest, tilting test, deep breathing test) the results obtained by the two different devices demonstrated a good agreement [18], while some experiments (handgrip test, rapid changes of posture, cold stress test) have revealed an overestimation of $P_{\text{mean}}$ by the device based on MOC.

The aim of this study is to analyse theoretically the effect of the arterial pressure–volume ratio, pressure pulse amplitude, and pulse shape index on the accuracy of MOC estimation of the arterial mean blood pressure.

## 2. PRINCIPLES AND METHODS

### 2.1. Oscillometric method

We determined the pressure pulse amplitude $P_{\mathrm{pulse}}$ as the intraarterial pressure wave height (i.e., as the difference between systolic ($P_{\mathrm{syst}}$) and diastolic ($P_{\mathrm{diast}}$) pressure)

$$P_{\mathrm{pulse}} = P_{\mathrm{syst}} - P_{\mathrm{diast}}, \tag{1}$$

and the pulse shape index $k$ as

$$k = (P_{\mathrm{mean}} - P_{\mathrm{diast}})/P_{\mathrm{pulse}}. \tag{2}$$

Transmural pressure is the difference between intraarterial ($P_{\mathrm{a}}$) and cuff ($P_{\mathrm{cuff}}$) pressure

$$P_{\mathrm{transm}} = P_{\mathrm{a}} - P_{\mathrm{cuff}}. \tag{3}$$

During cuff deflation manoeuvre the pressure oscillation amplitude in the occluding cuff exhibits a specific pattern. This pattern is characterized by an increase in the cuff oscillation amplitude up to a maximum level and then by a slower decrease. Values of $P_{\mathrm{syst}}$, $P_{\mathrm{mean}}$, and $P_{\mathrm{diast}}$ are estimated according to the amplitude of oscillations in the cuff. It is generally accepted that to estimate $P_{\mathrm{mean}}$ the peak of OC can be used. Values of $P_{\mathrm{syst}}$ and $P_{\mathrm{diast}}$ can be identified as pressures at a fixed percentage of maximum oscillations.

Figure 1 shows an example of the simultaneous recording of the finger $P_{\mathrm{cuff}}$ and the pulsatile component of the pneumoplethysmographic (PPG) signal obtained from the index finger of a normotensive subject. The $P_{\mathrm{cuff}}$ and pulsatile PPG signal were simultaneously recorded during the decrease in $P_{\mathrm{cuff}}$ at the constant rate of 2–3 mmHg/s. The value of $P_{\mathrm{mean}}$ was determined from the PPG signal according to MOC: $P_{\mathrm{cuff}}$ corresponding to the point of maximum amplitude of the pulsating signal was estimated as $P_{\mathrm{mean}}$. As shown in Fig. 1a, the maximum amplitude of the pulsatile component did not exceed 2 mmHg (by $P_{\mathrm{pulse}} \cong 50$ mmHg) and therefore the used pneumatic system can be considered elastic enough to provide an undamped recording of arterial volume pulses. Pressure transducer model MPX5050 (Motorola) and an A-D converter (12 bit, 120 Hz) were applied to record the pneumatic signals.

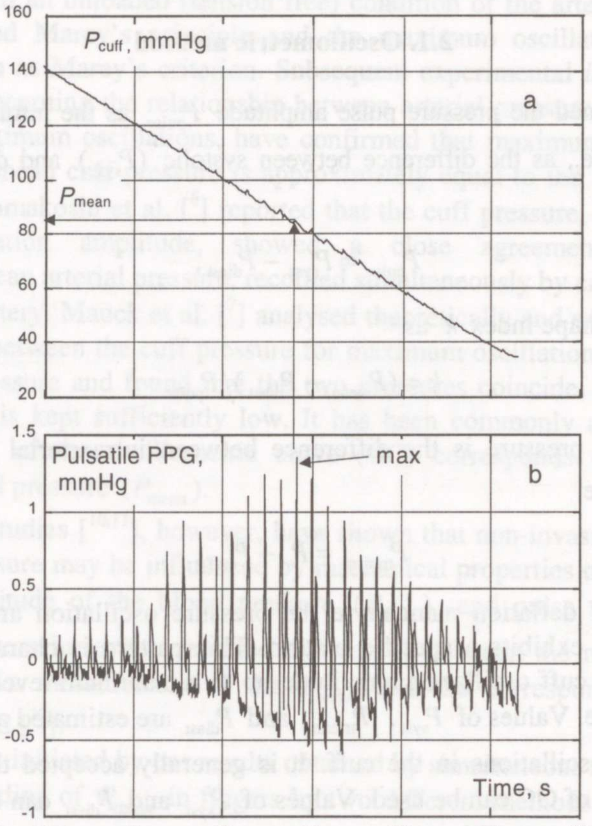


**Fig. 1.** An example of simultaneous recording of finger cuff pressure $P_{cuff}$ (a) and the pulsatile component of the PPG signal (b), obtained from the index finger of a normotensive subject; the arrow indicates the point of the maximum amplitude of the pulsatile PPG signal.

## 2.2. Pressure–volume relationship and compliance

The envelope curve of oscillations in the cuff (Fig. 1b) has a specific pattern caused by both the elastic and viscoelastic properties of arteries. The main character of the transformation of blood pressure pulses from the artery to an occlusive cuff can be explained by a nonlinear *P–V* relationship of the arteries. The point of the maximum slope of the *P–V* curve approximately (but not exactly, as will be shown later) corresponds to the point of maximum oscillations in the cuff. The slope of the *P–V* relation is known as vascular compliance $C$:

$$C = \mathrm{d}V/\mathrm{d}P. \tag{4}$$

Several categories of arterial compliance have been used by investigators to describe mechanical parameters of arteries: arterial static compliance ($C_{stat}$) [11,19,20], arterial dynamic compliance ($C_{dyn}$) [19,21], arterial beat-to-beat compliance ($C_{btb}$) [20,22], and arterial instantaneous compliance ($C_{inst}$) [19] ($C_{btb}$ and $C_{inst}$ are variations of $C_{dyn}$). Although a larger number of analyses have been performed by using the static *P–V* relationship, we decided to use the beat-to-beat *P–V* diagram to obtain more realistic pulse transformation parameters for normalized conditions. As pointed out by Peñaz et al. [19], the arterial dynamic compliance in fingers may have a form similar to the static one except the amplitude which is only a fraction of the value that could be expected on the basis of the static relationship.

To obtain information on how the general character of the *P–V* relations affects the accuracy of MOC estimates, we have chosen two typical shapes of this relationship (Fig. 2a, b) for mathematical modelling. The diagram in Fig. 2a was derived from 5 original recordings of the pulsatile component of $P_{cuff}$ in the index finger of a normotensive subject by PPG technique during cuff deflation.

The second *P–V* diagram (Fig. 2b) was selected to reflect the situation in the finger with partly contracted small arteries. On the basis of Wesseling's et al. findings [17], it can be claimed that in narrow thick-wall arteries the negative transmural pressure achieves 25–50 mmHg before full collapse occurs. Such arteries may be present in patients with hypertension or in condition when the smooth muscle in the artery wall is partially contracted. A partially contracted artery shows a more gradual collapse and has a smaller diameter at positive transmural pressure.

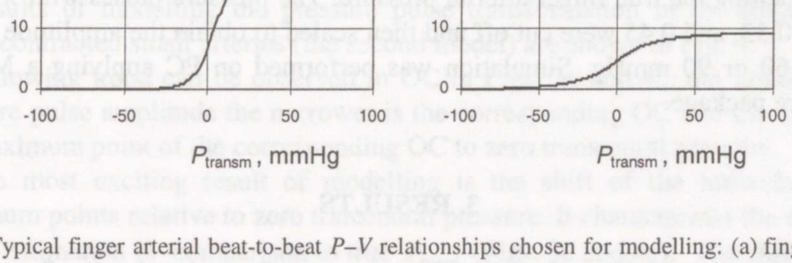


**Fig. 2.** Typical finger arterial beat-to-beat *P–V* relationships chosen for modelling: (a) finger with relaxed arteries; (b) finger with partly contracted small arteries.

### 2.3. Mathematical modelling

Mathematical modelling allows us, considering the nonlinear relation between the transmural pressure and arterial volume changes, to simulate a variety of arterial pressure pulses of different shape and height. Then, gradually altering the occlusive cuff pressure, we achieve a series of simulated arterial volume pulses (oscillations). Thereafter OC can be drawn and shifts of the maximum point determined. The error of every MOC estimation can be found as the ratio of the measured shift to zero transmural pressure. The latter is defined as the pressure corresponding to maximum slope of the arterial *P–V* relationship.

We have chosen an exponential model of the arterial beat-to-beat *P–V* relationship. For the first type of relationship (Fig. 2a, the finger with relaxed arteries) the model parameters were identified by the experiment described above:

$$V = \begin{cases} V_1 e^{\frac{a}{V_1} P_{\text{transm}}}, & \text{for} \quad P_{\text{transm}} \le 0, \\ V_1 + V_2 (1 - e^{-\frac{a}{V_2} P_{\text{transm}}}), & \text{for} \quad P_{\text{transm}} \ge 0, \end{cases} \tag{5}$$

where $V$ is the blood volume at the specified $P_{\text{transm}}$, $V_1$ is the blood volume when the artery is collapsed, $V_1 + V_2$ is the blood volume when the artery is fully expanded, and $a$ is a coefficient.

Following values of the parameters were obtained: $V_1 = 6.60$ µl, $V_2 = 30.37$ µl, and $a = 0.87$ µl/mmHg.

For the second type of the *P–V* relationship (Fig. 2b, the finger with partly contracted small arteries) we chose the following values of parameters: $V_1 = 4.62$ µl, $V_2 = 4.62$ µl, and $a = 0.22$ µl/mmHg.

In both models the maximum slope (compliance) occurs at the point $P_{\text{transm}} = 0$. For both models three values of $P_{\text{pulse}}$ were applied: 30, 60, and 90 mmHg. Every pulse height was simulated at three different values of the pulse shape index 0.2, 0.33, and 0.45, respectively.

In our archive of original Finapres recordings we found waveforms suitable for modelling the true finger arterial pressure. The pressure pulses with $k$ equal to 0.2, 0.33, and 0.45 were cut off and then scaled to obtain the amplitude values of 30, 60 or 90 mmHg. Simulation was performed on PC applying a MatLab software package.

## 3. RESULTS

Results of modelling the pressure pulse transformation in the finger with relaxed arteries of a normotensive person (the first model) are shown in Fig. 3.

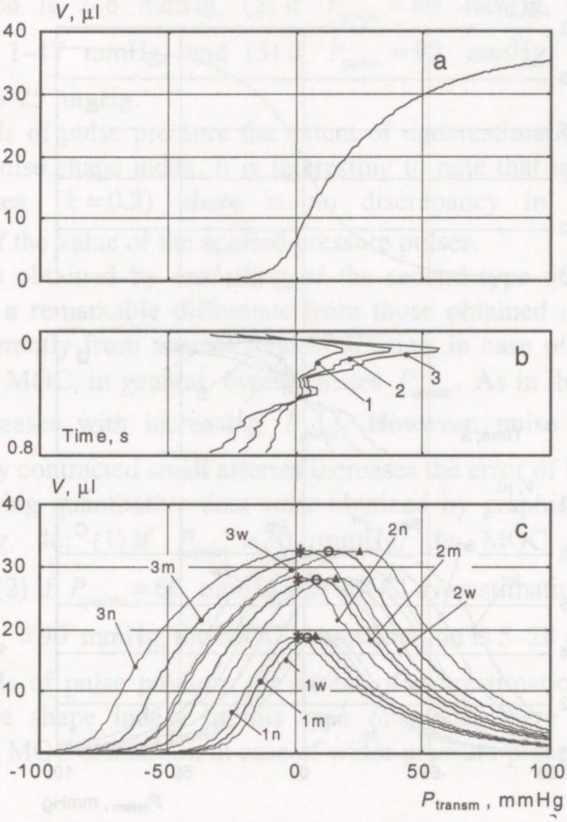


**Fig. 3.** Results of modelling the pressure pulse transformation in the finger with relaxed arteries of a normotensive person: (a) *P–V* relationship; (b) evolution of three arterial pressure pulses of constant shape ($k = 0.33$) and different height (1, 30 mmHg; 2, 60 mmHg; 3, 90 mmHg); (c) three sets of calculated volume oscillation curves (indices 1, 2, and 3 denote pulse heights 30, 60, and 90 mmHg; indices n, m, and w denote pulse shape indices 0.2, 0.33, and 0.45, respectively).

In Fig. 3c the sets 1, 2, and 3 can easily be distinguished from one another as the higher pressure pulses have higher volume response as well. In every set three curves are shown which correspond to different shape indices.

Results of modelling the pressure pulse transformation in the finger with partly contracted small arteries (the second model) are shown in Fig. 4.

Following trend can be observed in OC in Figs. 3c and 4c: the lower is the pressure pulse amplitude the narrower is the corresponding OC and the closer is the maximum point of the corresponding OC to zero transmural pressure.

The most exciting result of modelling is the shift of the individual OC maximum points relative to zero transmural pressure. It characterizes the error of MOC estimation in comparison to true $P_{mean}$ (Figs. 3c and 4c). The shift of the maximum point to the right (towards the positive values of $P_{transm}$) means

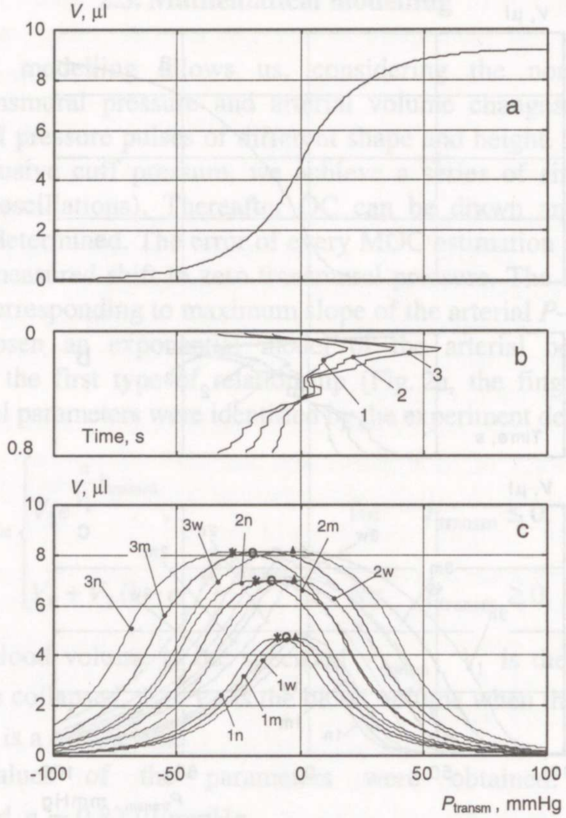


**Fig. 4.** Results of modelling the pressure pulse transformation in the finger with partly contracted small arteries: (a) $P$–$V$ relationship; (b) evolution of three arterial pressure pulses of constant shape ($k = 0.33$) and different height (1, 30 mmHg; 2, 60 mmHg; 3, 90 mmHg); (c) three sets of calculated volume oscillation curves. Notations are the same as in Fig. 3.

underestimation of intraarterial pressure, and, on the contrary, the shift of the maximum point to the left (towards the negative values of $P_{transm}$) means overestimation of intraarterial pressure. If the maximum points remain located close to the zero transmural pressure, there is no difference between $P_{mean}$ and MOC estimates.

The results obtained by applying the first type of model (Fig. 3c) demonstrate that in case of normal relaxed finger arteries, the MOC usually underestimates $P_{mean}$. The underestimation increases with increasing $P_{pulse}$. On the other hand, narrowing of the pulse decreases the error of the MOC estimate.

For simulated pressure pulse amplitudes and pulse shape indices the following quantitative data were obtained: (1) if $P_{pulse} = 30$ mmHg, the MOC underestimation is 2–8 mmHg, (2) if $P_{pulse} = 60$ mmHg, the MOC underestimation is 1–17 mmHg, and (3) if $P_{pulse} = 90$ mmHg, the MOC underestimation is 2–25 mmHg.

At all levels of pulse pressure the extent of underestimation depends on the value of the pulse shape index. It is interesting to note that in case of narrower pressure pulses ($k = 0.2$) there is no discrepancy in MOC estimation independent of the value of the applied pressure pulses.

The results obtained by modelling of the second type of the $P$–$V$ relation (Fig. 4) show a remarkable difference from those obtained on the first model (Fig. 3). Differently from normal relaxed arteries, in case of partly contracted finger arteries MOC, in general, overestimates $P_{mean}$. As in the previous model, the error increases with increasing $P_{pulse}$. However, pulse narrowing in the model of partly contracted small arteries increases the error of MOC estimation.

The following quantitative data were obtained by graphical analysis of the curves in Fig. 4c: (1) if $P_{pulse} = 30$ mmHg, the MOC overestimation is 2–10 mmHg, (2) if $P_{pulse} = 60$ mmHg, the MOC overestimation is 3–18 mmHg, and (3) if $P_{pulse} = 90$ mmHg, the MOC overestimation is 5–28 mmHg.

At all levels of pulse pressure the extent of overestimation depends on the value of pulse shape index. In this type of model there is practically no discrepancy in MOC estimation in case of wider pressure pulses ($k = 0.45$).

## 4. DISCUSSION

Investigating two different models of arterial pressure–volume relation we have found that Marey's criterion does not correctly estimate arterial mean blood pressure. The discrepancy between the true mean blood pressure (i.e. the pressure corresponding to the maximum of arterial compliance) and its MOC estimate may be more than ±20 mmHg depending on the form of arterial $P$–$V$ diagram, arterial pressure pulse height and shape.

The conception of "true mean blood pressure" is based on the generally accepted assumption that zero transmural pressure occurs when arterial $P$–$V$ relationship has maximum slope (maximum arterial compliance). The discrepancies between the mean blood pressure and its MOC estimation agree with the latest observations of other authors [10,11].

The simulation study has shown that in case of normal relaxed finger arteries the mean blood pressure tends to be underestimated by MOC, however, in case of partly contracted arteries the mean blood pressure tends to be overestimated by MOC. It was found that both models were characterized by the fact that decreasing of pressure pulse amplitude reduces the error of MOC estimation and *vice versa*. For relaxed finger arteries the steeper pressure pulse decreased, and for partly contracted arteries increased the error.

It has been established [23] that in finger arteries a specific pulse wave amplification may occur as a result of pulse wave reflection in the finger arterial system. It leads to the appearance of steeper pulse upstroke and higher pulse pressure amplitude. Considering results of our study, this phenomenon does not remarkably affect the accuracy of MOC estimates in case of normal relaxed arteries, but dramatically increases the error in case of partly contracted arteries. Concerning the numerical errors in MOC estimation obtained in this study, it should be mentioned that the exponential model was found not to be satisfactory by approximating the real arterial *P–V* relationship for normal relaxed finger arteries registered by us experimentally. The description based on two exponents was chosen because of its simplicity. For a more sophisticated model a better approximation should be found.

Considering the form of *P–V* diagrams we should like to note that the behaviour of the lower (convex to *x*-axis) part of the *P–V* diagram is strongly influenced by the technique used to record the relationship and does not adequately reflect the real properties of the arteries. It has been proved by observations that in case of applying photoplethysmographic or impedance plethysmographic methods [19,24,25] instead of pneumoplethysmography, the effect of generating oscillations in the proximal part of the cuff at a higher counterpressure is less apparent and, as a result, the convex component of the *P–V* diagram is less expressed.

## 5. SUMMARY

Assuming that zero transmural pressure occurs when arterial *P–V* relationship has maximum slope (this assumption being rational and commonly accepted), the value of the pressure estimated by the Marey's criterion does not exactly agree with the value of the true mean blood pressure. The difference between the true mean blood pressure (i.e., the pressure corresponding to the maximum of arterial compliance) and its MOC estimate may exceed ±20 mmHg depending on the form of the arterial *P–V* diagram and arterial pressure pulse height and shape. The simulation study has shown that in case of normal relaxed finger arteries the mean blood pressure tends to be underestimated and in case of partly contracted arteries overestimated by MOC.

## ACKNOWLEDGEMENT

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# MAREY KRITEERIUMI JÄRGI MÄÄRATUD KESKMISE VERERÕHU VIGADE TEOREETILINE UURIMINE

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Matemaatilise modelleerimise meetoditega on uuritud, kuidas mõjutavad arterite rõhu–mahu ($P$–$V$) diagrammi iseloom, pulsirõhu suurus ja pulsilaine kuju sõrme vererõhu mitteinvasiivse mõõtmise tulemusi Marey kriteeriumi rakendamisel. On näidatud, et rõhu väärtused, mis saadakse Marey kriteeriumi alusel, ei lange täpselt kokku tegeliku keskmise rõhu väärtustega. Viga võib ulatuda ±20 mmHg-ni, olles tingitud arterite $P$–$V$ sõltuvuse iseloomust, pulsirõhu suurusest ja pulsilaine kujust.