Identification of the dynamic component of the finger arterial pressure-volume relationship

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Abstract

Similar to most biological tissues, the arterial wall has both elastic and viscous properties. This is the reason why recordings of the pressure-volume (*P-V*) characteristics show a shape of hysteresis loops and are different for static and dynamic measurements. The exact model of the viscoelastic *P-V* relationship would be complicated, containing several sections with potentially nonlinear properties. The volume changes, in the simplest arrangement, can be viewed as a sum of fast and slower components.

We propose a method to identify the fast (dynamic) component of the nonlinear *P-V* relationship. Continuous finger blood pressure is measured by applying the Finapres monitor, while volume pulses at a controlled cuff pressure are recorded by the transmittance mode photoplethysmograph. The nonlinear relationship is modelled by five parameters. The identification is based on the fitting of parameters of the *P-V* relationship to obtain maximum similarity between the filtered waveforms of the modelled and measured volume signals. The method was tested on 9 normal subjects. The results show that the applied algorithm can be used even in case of remarkable viscous creep effects.
Keywords: modelling, system identification, pressure-volume relationship, arterial compliance, blood pressure, photoplethysmography.

1 Introduction

Peripheral circulation is strongly influenced by the peripheral resistance and compliance. According to the definition, peripheral compliance is equal to the slope of the peripheral pressure-volume (*P-V*) relationship and can be expressed as $C=dV/dP$. In general, this relationship is nonlinear, and this nonlinear nature of the arterial mechanical relationship results in appearance of a maximum energy transfer from the artery to the occluding cuff and is often used in indirect blood pressure measuring devices, e.g. oscillometric instruments. Knowledge of the arterial *P-V* relationship is of great importance for the correct interpretation of results in indirect blood pressure measurement.

On the other hand, data about patients' arterial pressure-volume relationship have an independent prognostic value in the assessment of the patients' vascular condition. It has been revealed that an elevated level of arterial blood pressure is usually originated by an increase in the arterial wall stiffness and peripheral vascular resistance.

As a rule, the local compliance of an artery is assessed from the simultaneous recordings of pressure and volumetric signals. The pulse pressure wave is travelling along the vascular tree with finite velocity and its shape is formed as a sum of the forward and reflected waves. Because of that it would be ideal, if the pressure and volume signals, used in compliance estimation, are recorded on the same measuring site. In the case of real indirect measurements this condition is hard to achieve. The pressure and volume signals should be recorded at least on similar sites, not disturbing each other. Two adjacent fingers can be regarded as appropriate sites for such a measurement. With the Finapres it is now possible to record the whole arterial pressure waveform non-invasively from the finger artery. The volume changes can independently be recorded on the other finger. Such a combination, mostly by the use of finger photoplethysmography (PPG), has been applied by several authors. Allen and Murray [1] used PPG, secured with moderate tension not restricting blood flow. Wesseling *et al* [2], Peñaz *et al* [3] and Guerrisi *et al* [4] used PPG at controlled cuff pressures to record PPG signals over a wide transmural pressure range.

It is well known that the arterial wall has both elastic and viscous properties. This is the reason why recordings of the *P-V* characteristics show a shape of a hysteresis loops and are different for static and dynamic measurements.

Wesseling *et al* [2] recorded the pseudo-static relationship by applying Finapres and a special variable setpoint device. Peñaz *et al* [3] used vibration plethysmography to record dynamic vascular compliance and found that it gives information about the instantaneous visco-elastic state of small peripheral arteries. Guerrisi *et al* [4] constructed the static relationship by using end-diastolic points of several cardiac cycles.

Allen and Murray [1] modelled the relationship between blood pressure and blood volume pulses by using linear and neural network system identification techniques. Attention was paid to the extraction of information on time constant and gain of the model. The character of nonlinearity was not given.

In mathematical studies the arterial wall *P-V* curve is usually modelled by using exponents [5] or the arctangent function [6]. The exact model of the viscoelastic *P-V* relationship would be complicated, containing several sections with potentially nonlinear properties.

The aim of present study was to model the fast (dynamic) component of the *P-V* relationship on the basis of PPG signals, recorded at controlled cuff pressures.

2 Methods

2.1 Experimental setup

Nine healthy volunteers, 3 males and 6 females were studied at rest at room temperature 22-23°C. The subject was seated comfortably with the left arm resting at heart level.

Finger arterial pressure was recorded by Finapres (Ohmeda, USA). Photoplethysmograms were recorded by applying one channel of the UT9201 physiograph (University of Tartu, Estonia) containing a transmittance mode photoplethysmograph. The finger cuffs were wrapped around the middle phalanx of the middle finger (Finapres) and ring finger (UT9201) of the subject's left hand. After a 10-minute equilibrium period, an automatic setpoint adjustment (Physiocal) was switched off. The pressure in the UT9201 cuff was lowered to zero and then increased with the rate of ~3 mmHg/s to suprasystolic level until PPG oscillations disappeared. After that the pressure was slowly decreased back to zero.

2.2 Mechanical model of the finger vasculature

The finger vasculature consists of vessels of different sizes and functions, and their filling and emptying may take place with very different time constants (at very different rates). The volume changes, in the simplest arrangement, can be viewed as a sum of relatively fast and slower components. The corresponding mechanical model to describe the response of finger vascular volume to pressure changes is shown in fig. 1. The stress represents arterial transmural pressure, while the strain represents volumetric changes in PPG.

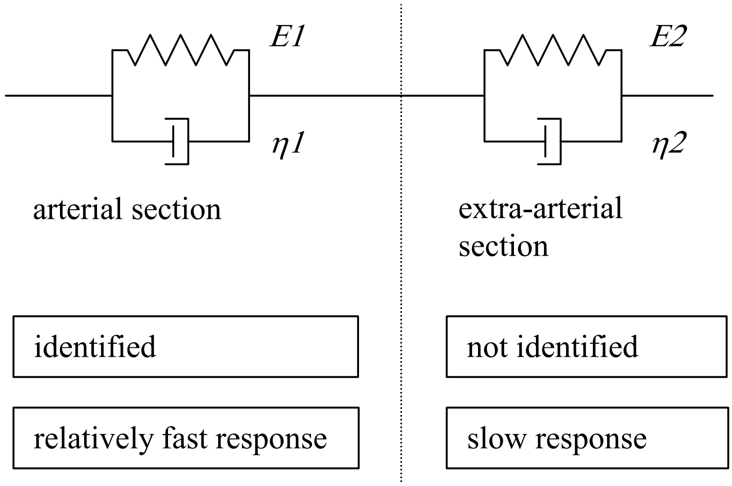

Figure 1: Simplified mechanical model of the finger vasculature.

Slower creeps in the vascular volume are considered to relate mainly to the extra-arterial phenomena (the right part of the figure) and are not described in this paper. The left part of the figure represents an arterial section with a relatively fast response. We suppose, there exists a nonlinear arterial *P-V* relationship, but the volume cannot follow the pressure changes immediately, and volume changes occur with the time constant of creep, modelled by the left Voigt cell in fig. 1.

2.3 Identification technique

Identification of the *P-V* relationship is performed according to the functional diagram shown in fig. 2. Transmural pressure is calculated as the difference between two pressures, i.e. blood pressure, measured by the Finapres (P_a), and the UT9201 cuff pressure (P_c)

$$P_{trF} = P_a - P_c \tag{1}$$

Since Finapres is known to follow changes in the intraarterial pressure well, but the bias can be remarkable [7], we added a member P_{bias} to eqn. (1). The biased transmural pressure is expressed as

$$P_{trB} = P_a - P_c + P_{bias} \tag{2}$$

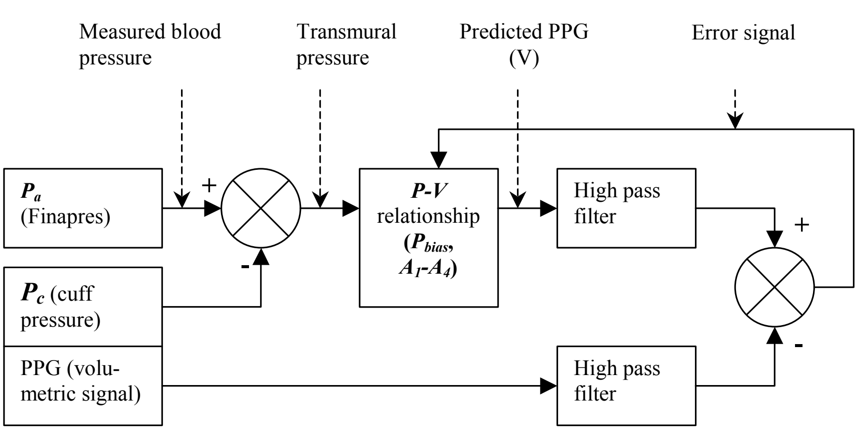

Figure 2: Principles of the identification of the *P-V* relationship. The error signal is used to adjust the *P-V* model parameters for minimising the modelling error.

Nonlinearity of the *P-V* relationship is modelled by two arctangent functions, one for positive and the other for negative transmural pressures:

$$V_{pv} = A_1 \cdot \arctan\left(\frac{A_3}{A_1} \cdot P_{trB}\right) \quad \text{for} \quad P_{trB} \geq 0$$

$$V_{pv} = A_2 \cdot \arctan\left(\frac{A_3}{A_2} \cdot P_{trB}\right) \quad \text{for} \quad P_{trB} < 0 \tag{3}$$

The slope of the relationship has its maximum value at zero biased transmural pressure.

The above described formulas are based on assumption that there exists a one-to-one relationship between the pressure and the PPG signal. In reality, there exists a phase shift between the pressure and volume changes, and a hysteresis effect is present. This effect is modelled as a unity-gain first order lag. Each value of the volume is computed as a combination of the volume at a past time point and the volume corresponding to transmural pressure at the current time point:

$$V\ (n) = A_4 * V_{pv}(n) + (1 - A_4) * V\ (n-1) \tag{4}$$

where the value of A_4 between 0 and 1 determines the rate of viscous creep.

The predicted and measured PPG waveforms are filtered by two identical high pass filters (0.7 Hz, 3rd order) and then used to derive an error signal. This signal was used to adjust model parameters for minimising the modelling error. The filters serve to suppress the disturbing influence of low frequency creeps of the measured PPG signal.

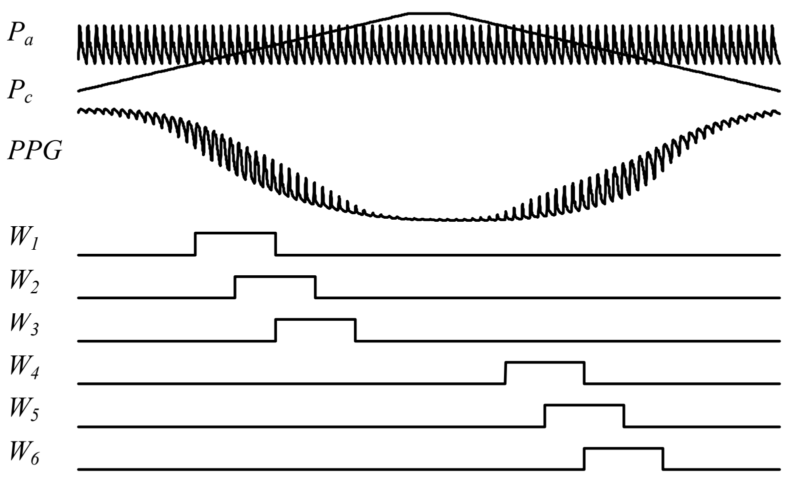

Figure 3: Data sampling diagram.

To examine the stability of the proposed method, from one experiment 6 datasets were created according to fig. 3, i.e. three sets during the cuff pressure increase and next three during the pressure decrease time.

The Levenberg-Marquardt method was used to identify model parameters P_{bias}, A_1, A_2, A_3 and A_4.

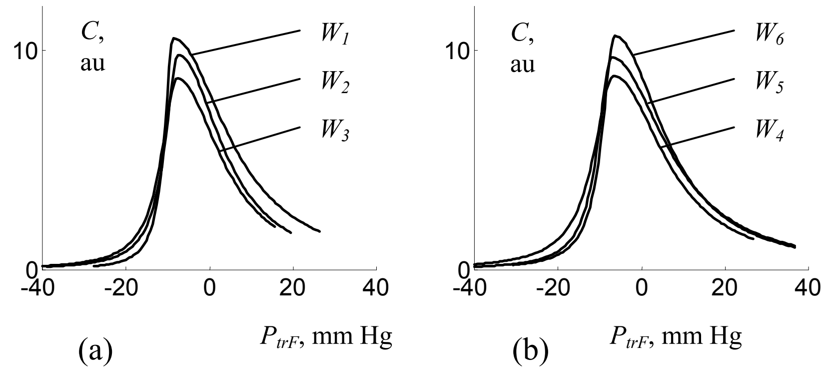

Figure 4: Example of the estimated compliance curves for one individual. (a): for P_c increasing episodes (W_1-W_3); (b): for P_c decreasing episodes (W_4-W_6). Compliance is in arbitrary units (au).

3 Results

An example of modelling results for one individual is shown in fig. 4.

Six curves, each for one data window W_1-W_6, show that the estimation of the character of the arterial compliance is relatively stable. However, some changes in compliance curves can be observed. The height of the compliance curve decreased when the cuff pressure was increased and began to increase again when the cuff pressure was lowered. A_4 was higher in case of a higher transmural pressure. Those phenomenona appeared in case of most patients.

An example of the measured and predicted PPG and their difference for one data window W_3 is shown in fig. 5.

4 Discussion

Typical shape of the compliance curve obtained in this study is similar to that found in [3]. The tendency that the estimated compliance curve changed its height during a cuff pressure ramp, also agrees with results in the above-mentioned study, where changes in visco-elastic properties of the arterial wall in response to the suprasystolic cuff pressure were reported.

The tendency that A_4 was higher in case of a higher transmural pressure, coincides with results in [1], where after hand raising the time constant decreased.

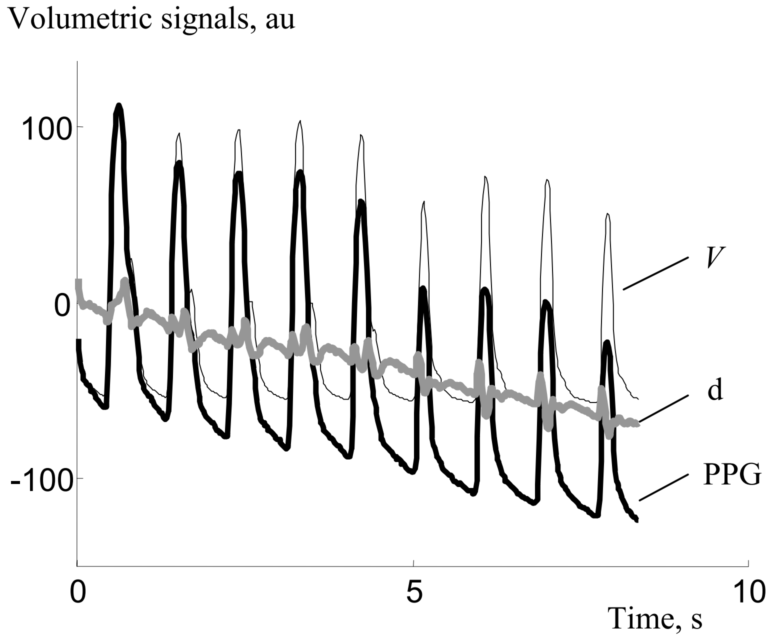

Figure 5: Measured (PPG) and modelled (V) waveforms and their difference (d).

In some cases the shapes of compliance curves found for P_c increasing and decreasing episodes were somewhat different, and it is our opinion that this is an estimation problem rather than a true physiological change.

Difference between the measured and predicted PPG signals (fig. 5) contains information about a fitting discrepancy of pulses, on one hand, and a slow change, related mainly to the extra-arterial section, on the other hand. As pointed out in [8], the simple Voigt model, although supposed to be a relatively crude representation of viscoelasticity, still represents the viscoelasticity in its minimal form. Similarly to all finger PPG measurements, the measured signal is not calibrated in absolute volume units.

In conclusion, the paper presents a method to identify the parameters of the dynamic finger pressure-volume relationship from synchronously recorded finger pressure and volume data. Results show the stability of the method when data obtained during the cuff pressure ramp are used repeatedly for several sliding time windows. The observed changes in the model properties are in agreement with results obtained by other authors.

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