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Estimation of distributed arterial mechanical properties using a wave propagation model in a reverse way

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ABSTRACT

To estimate arterial stiffness, different methods based either on distensibility, pulse wave velocity or a pressure-velocity loop, have been proposed. These methods can be employed to determine the arterial mechanical properties either locally or globally, e.g. averaged over an entire arterial segment. The aim of this study was to investigate the feasibility of a new method that estimates distributed arterial mechanical properties non-invasively. This new method is based on a wave propagation model and several independent ultrasound and pressure measurements. Model parameters (including arterial mechanical properties) are obtained from a reverse method in which differences between modeling results and measurements are minimized using a fitting procedure based on local sensitivity indices. This study evaluates the differences between in vivo measured and simulated blood pressure and volume flow waveforms at the brachial, radial and ulnar arteries of 6 volunteers. The estimated arterial Young's modulus range from 1.0 to 6.0 MPa with an average of (3.8 ± 1.7) MPa at the brachial artery and from 1.2 to 7.8 MPa with an average of (4.8 ± 2.2) MPa at the radial artery. A good match between measured and simulated waveforms and the realistic stiffness parameters indicate a good in vivo suitability.

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1. Introduction

Arterial stiffness, S, is an independent predictor of cardiovascular risk at an early stage [1] and is increasingly assessed in clinical practice [2–6].

To study the relations between arterial properties such as arterial stiffness and parameters such as blood pressure and blood volume flow (BVF), lumped parameter [7–11] or one-dimensional wave propagation models [12–16] can be used. The wave propagation models are based on governing equations (conservation of mass and momentum) and constitutive equations that describe the mechanical properties of the arterial wall. Wave propagation models are adapted to simulate wave propagation phenomena while demanding only few minutes of calculation time. Experimental and clinical validation models have been performed and demonstrated the ability to qualitatively describe blood pressure and blood volume flow waveforms [15,17,18]. Furthermore, wave propagation models were successfully employed to simulate the effects of (surgical) interventions [19] or predict the effect of pathophysiological conditions.

Wave propagation models cannot be used to estimate arterial stiffness directly, since arterial diameters, wall thicknesses and arterial wall mechanical properties like the Young's modulus are required as input. However, the model output, expressed in the BVF and blood pressure waveforms, can be employed to estimate arterial mechanical properties in a reverse process comparing model output with the corresponding measurements. Input parameters (including the arterial mechanical properties) are optimized until the best fit between measured and simulated BVF and blood pressure waveforms is obtained.

For the measurements required for such reverse method, ultrasound is the favorable imaging tool. Both perpendicular echo M-mode and oblique Doppler measurements can be performed with a high temporal resolution, to determine the vessel wall distension waveform and centerline blood velocity, respectively [20]. Then, scaling of the measured distension waveform with the systolic and diastolic blood pressure can be performed to approximate the blood pressure waveform. Furthermore, BVF can be obtained from the centerline velocity using Womersley profiles [20].

An initial estimate of the arterial stiffness is also required. Several methods have been proposed to estimate arterial stiffness non-invasively. Most commonly, the arterial stiffness is assessed locally from the time average vessel radius, $\bar{a}$, and the linearized vessel wall distensibility, D_0. The distensibility D_0 is defined as the ratio between the linearized compliance C_0 and the time average

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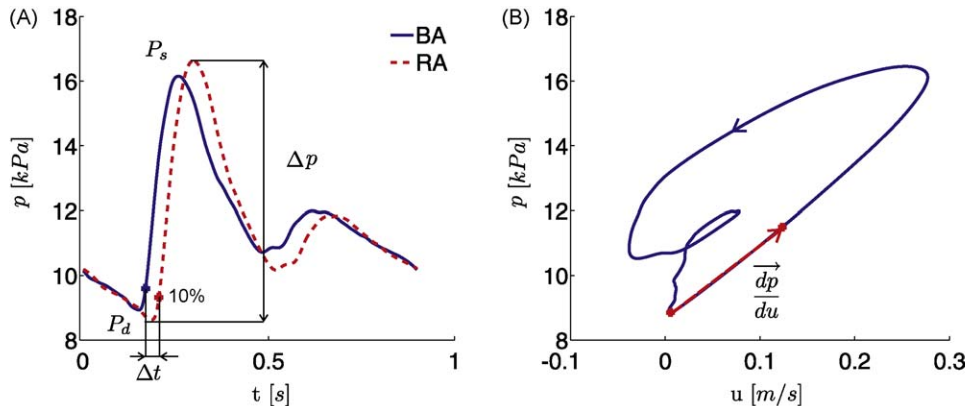

Fig. 1. (A) PWV method: the 10% foot (time when the pressure reaches 10% of Δp) during the systole is depicted. (B) The PU-loop method: the derivative of the pressure with respect to the average blood velocity is shown during the linear part of the systolic phase.

cross-sectional area $\bar{A}$:

$$D_0 = \frac{1}{\bar{A}}\frac{\Delta A}{\Delta p} = \frac{C_0}{A}, \quad \text{with} \quad C_0 = \frac{\Delta A}{\Delta p}, \tag{1}$$

ΔA being the maximum area difference and Δp the corresponding pressure difference within a heart cycle.

If we assume a linear elastic and isotropic behaviour, arterial stiffness S can be defined as the product between the Young's modulus, E, and the time average vessel wall thickness, $\bar{h}$, and can be calculated from the following relation [11]:

$$S = E\bar{h} = \frac{2[(2\bar{a}^2(1-\mu^2))/\bar{h} + (1+\mu)(2\bar{a}+\bar{h})]}{D_0((2\bar{a}/\bar{h})+1)}, \tag{2}$$

Herein μ is the Poisson ratio. When the wall thickness, $\bar{h}$, is an order of magnitude smaller than the radius $\bar{a}$, the following thin-walled tube formulation is obtained:

$$S \approx \frac{2\bar{a}(1-\mu^2)}{D_0}. \tag{3}$$

This relation is usually applied to common carotid, femoral or brachial arteries to derive a local arterial stiffness [4,21,22]. For all these arteries, the local pulse pressure is assumed to equal the brachial pulse pressure Δp. The radius $\bar{a}$ and ΔA can be assessed using an ultrasound scanner.

For arteries that are not accessible by ultrasound, an alternative method based on the global pulse wave velocity, c_0, can be employed. In the so-called pulse wave velocity (PWV) method, pressure waveforms obtained with applanation tonometry, or wall distension waveforms obtained with ultrasound techniques, are measured at a proximal and a distal site. For both waveforms, the foot of the wave is defined as the time when the pressure reaches 10% of its pulse Δp (see Fig. 1A). Alternatively for the common carotid artery the dicrotic notch can be chosen as reference point [23]. The distance between the measurement sites, Δl, is divided by the foot-to-foot or dicrotic notch transit time, Δt, to calculate the pulse wave velocity c_0. The distensibility D_0, is then estimated with the Moens-Korteweg wave speed c_0 for thin-walled tube assuming that the vessel wall behaves linear elastically, isotropically and incompressibly:

$$D_0 \approx \frac{1}{\rho c_0^2}, \quad \text{with} \quad c_0 = \frac{\Delta l}{\Delta t}, \tag{4}$$

with ρ the blood density. Next, either (2) or (3) can be used to estimate an average S over the arterial segment. Most commonly, the PWV is estimated between the carotid and femoral arteries, or between the brachial and radial arteries.

An alternative method, developed by Parker and Jones [24], is based on the pressure-velocity-loop (PU-loop) and requires the simultaneous assessment of the mean blood flow velocity over the arterial cross-section, u, and blood pressure, p. It is assumed that no reflections are present during the early systole and that the relation between p and u is linear (see Fig. 1B). The PWV, c_0, is then estimated from the slope of the PU-loop, with:

$$c_0 = \frac{1}{\rho}\frac{dp}{du}. \tag{5}$$

Since it is not possible to assess the blood pressure waveform non-invasively, p is approximated by the vessel wall distension waveform scaled with the systolic and diastolic blood pressures assuming a linear pressure–area relationship [25]. The scaling will not affect the temporal relationships.

Arterial stiffness can be estimated globally with the PWV method or locally using either the distensibility or the PU-loop method. All these methods have some drawbacks. The distensibility method reflects the arterial mechanical properties at a single location. Furthermore, it requires local pulse pressure while it can only be measured directly in the brachial artery. The PWV method estimates an average stiffness over a long arterial trajectory whereas stiffness differs between arteries and increases towards the peripheries [26]. Finally, for both PWV and PU-loop methods, hemodynamic viscous forces and pressure wave reflections originating from transitions in arterial stiffness, the presence of bifurcations, arterial lumen tapering and the peripheral bed are neglected. These assumptions might lead to inaccuracies in the estimates of arterial stiffness.

The aim of the present study is to investigate the feasibility of a non-invasive method to estimate distributed arterial mechanical properties by means of a reverse approach of a 1D wave propagation model, as developed by Bessems et al. [13], in combination with several independent ultrasound and blood pressure measurements.

An estimate of the distributed arterial properties along each arterial segment is obtained whereas other techniques are providing either local or average estimates. Furthermore, several independent ultrasound measurements are combined, which might reduce the sensitivity of the method to measurement errors. Moreover, this approach is based on a model taking into account hemodynamic viscous forces, pressure wave propagation and reflection phenomena.

We focus on the estimation of the mechanical properties of the arteries of arm because these arteries are frequently subject of medical investigations [27–29]. Furthermore, the local systolic and diastolic blood pressure can be measured directly in the brachial artery and ultrasound measurements can be performed in the main arteries from the arm pit until the wrist, enabling the estimation of vessel wall distension and BVF at several sites.

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