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Estimating absolute aortic pressure using MRI and a one-dimensional model

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ABSTRACT

Aortic blood pressure is a strong indicator to cardiovascular diseases and morbidity. Clinically, pressure measurements are done by inserting a catheter in the aorta. However, imaging techniques have been used to avoid the invasive procedure of catheterization. In this paper, we combined MRI measurements to a one-dimensional model in order to simulate blood flow in an aortic segment. Absolute pressure was estimated in the aorta by using MRI measured flow as boundary conditions and MRI measured compliance as a pressure law for solving the model. Model computed pressure was compared to catheter measured pressure in an aortic phantom. Furthermore, aortic pressure was estimated *in vivo* in three healthy volunteers.

1. Introduction

To date, blood pressure (BP) is one of the most useful clinical indicator of cardiovascular disease. Hypertension, more specifically, is a leading predictor of death in atherosclerosis diseases worldwide (Cohn et al., 2004; P.S. Collaboration, 2002). Therefore, measuring BP is of great interest for diagnosis and risk prevention of cardiovascular events. An elevated pressure gives information about the aortic state, the presence of atherosclerotic plaques, stenosis, calcification or aneurisms. In a clinical routine, a sphygmomanometer is used to measure systolic and diastolic brachial pressure. However, due to reflexion in the distal arteries, the aortic pressure waveform is altered while traveling through the vascular system. Thus distortion of the wave shape as well as systolic amplification occurs on the systolic pressure measured in the brachial artery (O'rourke et al., 1968; Park and Guntheroth, 1970; Salvi, 2012). Although models and transfer functions to link brachial BP to aortic pressure exist (Chen et al., 1996, 1997; Liang, 2014), wave reflection in the arterial system makes it difficult to reproduce the wave contour with great fidelity from such methods. Until now, the gold-standard of aortic pressure measurement is catheterization, which is invasive and not repeatable in a routine procedure (Murgo et al., 1980; Skinner and Adams, 1996). In recent years, imaging techniques have been used to assess pressure gradients from velocity or acceleration maps and its combination with fluid mechanics equations has been exploited in order to measure the BP non-invasively. Doppler Ultrasound (US) used to measure blood velocity in the arteries was associated to the standard simplified Bernoulli equation in order to determine pressure differences between two measurement sites. This latter technique is reported to be user-dependent and error-prone due to the wave angle of incidence (Zananiri et al., 1993; Muhler et al., 1993), hence its accuracy in determining the maximum velocity in the artery is debatable. Also, extending US to other situations is limited by the inapplicability of Bernoulli equation to unsteady flows (Yang et al., 1996). Phase-Contrast (PC) MRI allows accurate encoding of the blood velocity in the arteries in the three directions; hence, it has been largely used for non-invasive pressure estimation. Some authors computed pressure differences using the Poisson equation (Yang et al., 1996), others integrated the Navier–Stokes (NS) equations using MRI velocity maps (Tyszka et al., 2000; Thompson and McVeigh, 2003; Bock et al., 2011; Ebbers et al., 2001)or acceleration maps to avoid computational errors arising from velocity derivation (Buyens et al., 2005). These methods compute a pressure gradient, and to estimate an absolute pressure, require a zero-pressure reference point which has to be measured with a catheter, or user-defined in a gross assumption. Consequently, these methods are not an alternative to catheter measurements, which remain more accurate. In this work, we propose a non-invasive technique to extract

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absolute pressure in a straight artery from MR velocimetry using a biomechanical one-dimensional (1D) model as proposed by Formaggia et al. (2003). Although a three-dimensional (3D) model gives a more complete and realistic reproduction of the aortic flow, 1D models are able to describe the non-linear flow behavior in large elastic vessels (Hughes and Lubliner, 1973). As these models are reasonably accurate, they are widely used for aortic flow simulations. Their accuracy has been assessed by comparison with experimental data acquired in a tube phantom (Bessems et al., 2008), in a distributed arterial model (Olufsen et al., 2000; Alastruey et al., 2011) and *in vivo* (Reymond et al., 2009; Alastruey et al., 2009). Furthermore, the 1D model relies on the establishment of a pressure law consisting of a relation between pressure and vessel section area. The pressure laws used in 1D model equations are determined experimentally with invasive measurements or estimated using complex algorithms. Additionally, these pressure laws are complicated and involve the determination of multiple parameters. Here, we propose a pressure law based on the aortic compliance which reflects arterial elasticity and can be determined non-invasively with MRI. Using this pressure law, we coupled the 1D model with realistic boundary conditions measured by MRI to estimate absolute pressure in the aortic segment. The derived model was tested on a straight compliant phantom and computed pressure was compared to experimental pressure measurements recorded simultaneously with the MRI acquisition. The model was also tested on a real-sized compliant aortic phantom. Then, the model is used to estimate BP on healthy volunteers.

2. Methods

The 1D-model, studied in Formaggia et al. (2003), is a reduced model describing blood flow in arteries and its interaction with wall motion. The artery is considered as a cylindrical compliant tube of length L and radius R ($R \ll L$). The model derivation approach consists of integrating the NS equations on a generic section S. Some simplifying assumptions are made:

- the model assumes axisymmetry
- the wall displacement is supposed to solely be in the radial direction
- pressure is assumed to be uniform in each section
- the axial velocity u_z is predominant.

For large arteries such as the aorta, it is a safe assumption to consider a flat velocity profile for the boundary layer which is very thin compared to the vessel radius (Olufsen et al., 2000).

The main variables of the problem are (Fig. 1)

- axial section area A

$$A(t,z) = \int_{S(t,z)} \mathrm{d}\sigma \tag{1}$$

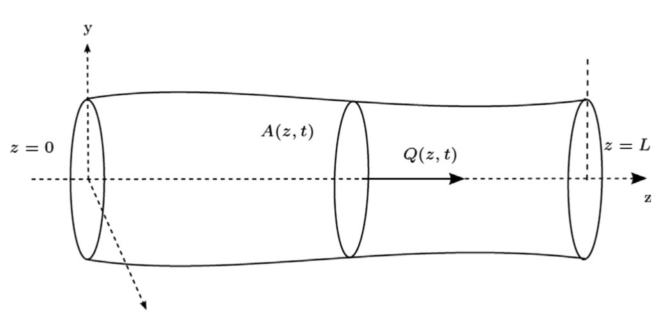

Fig. 1. The 1D model simplified geometry. It assumes that the artery is a straight cylinder of length L with a circular cross section $A(t,z)$ that deforms with respect to the radial vector.

- mean flow Q

$$Q(t,z) = \int_S u_z \, \mathrm{d}\sigma \tag{2}$$

- blood pressure $P(t,z)$,

where $\mathrm{d}\sigma$ denotes the area element. Their evolution is described by the momentum conservation and the mass conservation equations, while considering a constant viscosity:

$$\frac{\partial A}{\partial t} + \frac{\partial Q}{\partial z} = 0 \tag{3}$$

$$\frac{\partial Q}{\partial t} + \frac{\partial}{\partial z}\left(\frac{Q^2}{A}\right) + \frac{A \partial P}{\rho \partial z} + K_r\left(\frac{Q}{A}\right) = 0 \tag{4}$$

K_r is the friction coefficient; for a flat profile in blood flow problems, $K_r = 22\pi\nu$, where ν is the vessel wall kinematic viscosity (Formaggia and Veneziani, 2003), ρ is the blood density.

2.1. Pressure law

To close the system, a relation between the section area A and the pressure P is defined. This pressure law depends on section area $A_0(z) = \pi R_0^2(z)$ at time $t=0$ and on a set of parameters $\beta = (\beta_0, \beta_1, \ldots, \beta_n)$ related to the vessel wall's physical and mechanical properties. P_{ext} is the external pressure exerted by the environment of the vessel whereas P is the intravascular pressure. The pressure law should meet these conditions:

- $\frac{\partial P}{\partial A} > 0$
- when $A = A_0$, $P = P_{ext}$.

Some pressure laws are proposed in the literature to link the pressure behavior to the section area. A frequently used relation between A and P is stated as follows (Quarteroni and Formaggia, 2004; Alastruey, 2006):

$$P - P_{ext} = \beta_0 \frac{\sqrt{A} - \sqrt{A_0}}{A_0} \tag{5}$$

$\beta_0 = \sqrt{\pi} h_0 E/(1-\xi^2)$ using Young's modulus E and the vessel thickness h_0 and Poisson's ratio $\xi = 0.5$ for an incompressible material deformed elastically at small strains.

A more general law proposed in Hayashi et al. (1980) and Smith et al. (2000) is written as

$$P - P_{ext} = \beta_0 \left[\left(\frac{A}{A_0}\right)^{\beta_1} - 1\right]. \tag{6}$$

The parameters' β_0 and β_1 values can be obtained either by fitting experimental pressure vs. section measurements (Smith, 2004) or by solving an inverse problem with a 3D-model solution (Martin et al., 2005; Dumas, 2008). Hence, these laws cannot be determined non-invasively, and need knowledge about the vessel properties. Additionally, they seem too complex to determine during a clinical application. Finding a simple non-invasive way to determine a pressure law is of great interest, consequently, we turned to the compliance. Indeed, under physiological conditions, the aorta section deformation is commonly assumed to be linked to the intravascular pressure by the aortic compliance (Langewouters et al., 1984).

Aortic compliance establishes a linear relation between the pressure and the section area; it represents the arterial wall's ability to deform in response to a pressure variation (Conrad, 1969). Also, it is clinically used and can be estimated non-invasively by measuring the pulse wave velocity (PWV) in MRI. Hence, it provides a simple and non-invasive pressure law that can be applied *in vivo* and, as it includes compliance changes, is patient-specific.

In fact, compliance is given by the ratio of section variation to pressure variation:

$$\mathcal{C} = \frac{\mathrm{d}A}{\mathrm{d}P_t} \tag{7}$$

where $P_t = P - P_{ext}$ is the transmural pressure. The compliance is considered as a local constant on an arterial segment. By integrating Eq. (7) and knowing that when $P_t = 0$, i.e. $P = P_{ext}$, $A = A_0$, we write $A = \mathcal{C}P_t + A_0$ where A_0 is the section area at the equilibrium state.

We write Eq. (7) as $\mathcal{C} = (A - A_0)/(P - P_{ext})$, thus deriving the pressure law:

$$P - P_{ext} = \frac{A_0}{\mathcal{C}}\left(\frac{A}{A_0} - 1\right) \tag{8}$$