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1 Estimating absolute aortic pressure using MRI and a 2 one-dimensional model

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9 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 article, 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.

10 1. Introduction

11 To date, blood pressure (BP) is one of the most useful clinical indicator of
12 cardiovascular disease. Hypertension, more specifically, is a leading predictor
13 of death in atherosclerosis diseases worldwide Cohn et al. [13], Collaboration
14 [14]. Therefore, measuring BP is of great interest for diagnosis and risk pre-
15 vention of cardiovascular events. An elevated pressure gives information about
16 the aortic state, the presence of atherosclerotic plaques, stenosis, calcification

17 or aneurisms. In a clinical routine, a sphygmomanometer is used to measure
 18 systolic and diastolic brachial pressure. However, due to reflexion in the distal
 19 arteries, the aortic pressure waveform is altered while traveling through the vas-
 20 cular system. Thus distortion of the wave shape as well as systolic amplification
 21 occur on the systolic pressure measured in the brachial artery O’rourke et al.
 22 [32], Park and Guntheroth [34], Salvi [37]. Although models and transfer func-
 23 tions to link brachial BP to aortic pressure exist Chen et al. [12, 11], Liang [26],
 24 wave reflection in the arterial system makes it difficult to reproduce the wave
 25 contour with great fidelity from such methods. Until now, the gold-standard of
 26 aortic pressure measurement is catheterization, which is invasive and not repeat-
 27 able in a routine procedure Murgu et al. [30], Skinner and Adams [40]. In recent
 28 years, imaging techniques have been used to assess pressure gradients from ve-
 29 locity or acceleration maps and its combination with fluid mechanics equations
 30 has been exploited in order to measure the BP non-invasively. Doppler Ultra-
 31 sound (US) used to measure blood velocity in the arteries was associated to the
 32 standard simplified Bernoulli equation in order to determine pressure differences
 33 between two measurement sites. This latter technique is reported to be user-
 34 dependent and error-prone due to the wave angle of incidence Zananiri et al.
 35 [51], Muhler et al. [29], hence its accuracy in determining the maximum velocity
 36 in the artery is debatable. Also, extending US to other situations is limited
 37 by the inapplicability of Bernoulli equation to unsteady flows Yang et al. [50].
 38 Phase-Contrast (PC) MRI allows accurate encoding of the blood velocity in the
 39 arteries in the three directions; hence, it has been largely used for non-invasive
 40 pressure estimation. Some authors computed pressure differences using Poisson
 41 equation Yang et al. [50], others integrated the Navier-Stokes (NS) equations
 42 using MRI velocity maps Tyszka et al. [45], Thompson and McVeigh [44], Bock
 43 et al. [8], Ebberts et al. [17] or acceleration maps to avoid computational errors

arising from velocity derivation Buyens et al. [10]. 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 absolute pressure in a straight artery from
 MR velocimetry using a biomechanical one-dimensional (1D) model as proposed
 by Formaggia *et al.* Formaggia et al. [18]. 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 larger elastic vessels
 Hughes and Lubliner [22]. 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.
 [7], in a distributed arterial model Olufsen et al. [31], Alastruey et al. [2] and *in*
vivo Raymond et al. [36], Alastruey et al. [3]. 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 deter-
 mined 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 mea-
 surements recorded simultaneously with the MRI acquisition. The model was
 also tested on a real-sized compliant aortic phantom Then, the model is used to

71 estimate BP on healthy volunteers.

72 2. Methods

73 The 1D-model, studied in Formaggia et al. [18], is a reduced model describ-
74 ing blood flow in arteries and its interaction with wall motion. The artery is
75 considered as a cylindrical compliant tube of length L and radius R ($R \ll L$).
76 The model derivation approach consists of integrating the NS equations on a
77 generic section S . Some simplifying assumptions are made:

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

82 For large arteries such as the aorta, it is a safe assumption to consider a flat
83 velocity profile for the boundary layer is very thin compared to the vessel radius
84 Olufsen et al. [31].

85 The main variables of the problem are (Figure 1):

- 86 • axial section area A

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

- 87 • mean flow Q

$$Q(t, z) = \int_S u_z d\sigma \quad (2)$$

- 88 • blood pressure $P(t, z)$,

89 where $d\sigma$ denotes the area element. Their evolution is described by the mo-
90 mentum conservation and the mass conservation equations, while considering a

91 constant viscosity:

$$\frac{\partial A}{\partial t} + \frac{\partial Q}{\partial z} = 0 \quad (3)$$

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

92 K_r is the friction coefficient; for a flat profile in blood flow problems, $K_r = 22\pi\nu$,
 93 where ν is the vessel wall kinematic viscosity Formaggia and Veneziani [19]. ρ is
 94 the blood density.

95 *Pressure law*

96 To close the system, a relation between the section area A and the pressure
 97 P is defined. This pressure law depends on section area $A_0(z) = \pi R_0^2(z)$ at time
 98 $t = 0$ and on a set of parameters $\beta = (\beta_0, \beta_1, \dots, \beta_n)$ related to the vessel wall
 99 physical and mechanical properties. P_{ext} is the external pressure exerted by the
 100 environment of the vessel whereas P is the intravascular pressure. The pressure
 101 law should meet these conditions:

102 • $\frac{\partial P}{\partial A} > 0$

103 • when $A = A_0$, $P = P_{ext}$.

104 Some pressure laws are proposed in the literature to link the pressure behavior
 105 to the section area. A frequently used relation between A and P is stated as
 106 follows Quarteroni and Formaggia [35], Alastruey [1]:

$$P - P_{ext} = \beta_0 \frac{\sqrt{A} - \sqrt{A_0}}{A_0} \quad (5)$$

107 $\beta_0 = \frac{\sqrt{\pi} h_0 E}{1 - \xi^2}$ using Young's modulus E and the vessel thickness h_0 and Poisson's
 108 ratio $\xi = 0.5$ for an incompressible material deformed elastically at small strains.

109 A more general law proposed in Hayashi et al. [21], Smith and Hunter [41]
 110 is written as:

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

111 The parameters β_0 and β_1 values can be obtained either by fitting experi-
 112 mental pressure versus section measurements Smith [42], or by solving an inverse
 113 problem with a 3D-model solution Martin et al. [27], Dumas [16]. Hence, these
 114 laws cannot be determined non-invasively, and need knowledge about the vessel
 115 properties. Additionally, they seem too complex to determine during a clinical
 116 application. Finding a simple non-invasive way to determine a pressure law is of
 117 great interest, consequently, we turned to the compliance. Indeed, in physiolog-
 118 ical conditions, the aorta section deformation is commonly assumed to be linked
 119 to the intravascular pressure by the aortic compliance Langewouters et al. [24].

120 Aortic compliance establishes a linear relation between the pressure and
 121 section area; it represents the arterial wall ability to deform in response to a
 122 pressure variation Conrad [15]. Also, it is clinically used and can be estimated
 123 non-invasively by measuring the pulse wave velocity (PWV) in MRI. Hence, it
 124 provides a simple and non-invasive pressure law that can be applied *in vivo* and,
 125 as it includes compliance changes, is patient-specific.

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

$$\mathcal{C} = \frac{dA}{dP_t} \quad (7)$$

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

132 We write equation 7 as $\mathcal{C} = \frac{A-A_0}{P-P_{ext}}$, thus deriving the pressure law:

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

133 In the limit of small displacements, this pressure law is equivalent to the com-
134 monly used linear law ((5)) for

$$\beta_0 = \frac{2A_0^{3/2}}{\mathcal{C}}, \quad (9)$$

135 In fact, by linearizing ((5)) (see also Alastruey et al. [5]), we have:

$$\beta_0 \frac{\sqrt{A} - \sqrt{A_0}}{A_0} = \frac{\beta_0}{2A_0^{3/2}} (A - A_0) + O(A - A_0)^2$$

136 *Numerical approximation*

137 System (3)-(4) is written in the conservative form :

$$\frac{\partial \mathbf{U}}{\partial t} + \frac{\partial \mathbf{F}(\mathbf{U})}{\partial z} = \mathbf{S}(\mathbf{U}) \quad (10)$$

138 where

$$\begin{aligned} 139 \quad & \bullet \mathbf{U} = \begin{bmatrix} A \\ Q \end{bmatrix} \\ 140 \quad & \bullet \mathbf{F} = \begin{bmatrix} Q \\ \frac{Q^2}{A} + C_1 \end{bmatrix}, \quad C_1(A) = \int_0^A c_1^2(\tau) d\tau \text{ and } c_1 = c_1(A) = \sqrt{\frac{A}{\rho} \frac{\partial P}{\partial A}}, \\ 141 \quad & \bullet \mathbf{S} = \begin{bmatrix} 0 \\ -K_r \frac{Q}{A} \end{bmatrix} \end{aligned}$$

142 Since $A > 0$, the matrix $\mathbf{DF}$ possesses two real distinct eigenvalues $\lambda_{\pm} = \frac{Q}{A} \pm c_1$
143 and system (10) is a hyperbolic system of partial differential equations For-
144 maggia and Veneziani [19]. The blood flow is assumed to be sub-critical, thus
145 $\lambda_- < 0$, and $\lambda_{\pm}$ have opposite signs.

146 Following Formaggia and Veneziani [19], system (10) is written in diago-
 147 nal form, which is more suitable to understand the solution behavior and for
 148 numerical purpose:

$$\frac{\partial \mathbf{W}}{\partial \mathbf{t}} + \mathbf{\Lambda} \frac{\partial \mathbf{W}}{\partial \mathbf{z}} = \mathbf{L} \cdot \mathbf{S} \quad (11)$$

149 where $W = (W_1, W_2)$ are the characteristic variables, $\mathbf{\Lambda} = \begin{bmatrix} \lambda_+ & 0 \\ 0 & \lambda_- \end{bmatrix}$ and

150 $\mathbf{L} = \begin{bmatrix} \ell_+^T \\ \ell_-^T \end{bmatrix}$ is a matrix of left eigenvalues of $\mathbf{D}\mathbf{F}$ such that $\mathbf{L} \cdot \mathbf{D}\mathbf{F} = \mathbf{\Lambda} \cdot \mathbf{L}$. In

151 the case of the law (8), we have

$$c_1 = \sqrt{\frac{A}{\rho \mathcal{C}}},$$

$$W_1 = \frac{Q}{A} + 2\sqrt{\frac{A}{\rho \mathcal{C}}}, \quad W_2 = \frac{Q}{A} - 2\sqrt{\frac{A}{\rho \mathcal{C}}}. \quad (12)$$

152 Inversely, the primitive set of variables (A, Q) is written in terms of the
 153 diagonal variables (W_1, W_2) as

$$A = \left(\sqrt{\rho \mathcal{C}} \frac{W_1 - W_2}{4} \right)^2, \quad Q = A \frac{W_1 + W_2}{2}. \quad (13)$$

154 For system (10) approximation, a finite volume scheme is defined by a grid over
 155 $[0, L] \times [0, T]$, with uniform mesh-spacing Δz and time step Δt^n . The cells are
 156 given by $[z_i, z_{i+1}] \times [t^n, t^{n+1}]$, where $z_i = i\Delta z$, $i = 0, \dots, M$, with $M\Delta z = L$,
 157 and $t^n = n\Delta t^n$, $n = 0, \dots, N$, with $N\Delta t^N = T$. We seek U_i^n approximating
 158 $\frac{1}{\Delta z} \int_{z_i}^{z_{i+1}} U(x, t^n) dz$.

159 The numerical scheme is a classical finite volume scheme of Rusanov type
 160 for hyperbolic problems:

$$\frac{1}{\Delta t^n} (U_i^{n+1} - U_i^n) + \frac{1}{\Delta z} (F_{i+1/2}^n - F_{i-1/2}^n) - S_i^n = 0, \quad (14)$$

161

$$F_{i+\frac{1}{2}} = \frac{1}{2} \left[F(U_i^n) + F(U_{i+1}^n) - \lambda_{i+\frac{1}{2}}(U_i^{n+1} - U_i^n) \right]$$

162 with $\lambda_{i+\frac{1}{2}} = \max\{\lambda_-(U_i^n), \lambda_+(U_{i+1}^n)\}$, and where $S_i^n = \begin{bmatrix} 0 \\ -K_r \frac{Q_i^n}{A_i^n} \end{bmatrix}$. Time-
 163 space Δt^n must verify the well-known stability condition $2 \max_i |\lambda_{i+\frac{1}{2}}| \Delta t^n \leq \Delta z$.

164 Numerical boundary conditions are prescribed, thus $U_{in}^n = U_{-1}^n$ and $U_{out}^n =$
 165 U_{M+1}^n are given to compute U_0^{n+1} and U_M^{n+1} at $z = 0$ and $z = L$ respectively.
 166 Since flows involved in this model are sub-critical, the numerical scheme (14)
 167 requires one condition at each boundary. An admissible condition imposes the
 168 incoming characteristic and allows the wave corresponding to the outgoing char-
 169 acteristic to leave the domain. As W_2 remains constant along the characteristic
 170 curve defined by $\frac{dz}{dt} = \lambda_-(U(z, t))$, if we approximate $\lambda_-(U(z, t))$ by its numer-
 171 ical value $\lambda_-(U(z, t^n))$ in $[t^n, t^{n+1}]$, we consider that

$$W_2(z, t^{n+1}) \simeq W_2(z - \{\lambda_-\}_0^n \Delta t^n, t^n),$$

172 thus

$$W_{2,in}^{n+1} = \frac{\Delta z + \lambda \Delta t}{\Delta z} W_{2,-1}^n - \frac{\lambda \Delta t}{\Delta z} W_{2,0}^n$$

173 where $\lambda = \lambda_{-0}^n$ and $\Delta t = \Delta t^n$. Similarly, we impose

$$W_{1,out}^{n+1} = \frac{\Delta z - \lambda \Delta t}{\Delta z} W_{1,M+1}^n + \frac{\lambda \Delta t}{\Delta z} W_{1,M}^n$$

174 where $\lambda = \{\lambda_+\}_M^n$.

175 *Boundary conditions*

176 We used MRI measured boundary conditions on the inlet of the tube phan-
 177 tom or the arterial segment. Flow measured on the entrance was imposed as
 178 a condition on the first mesh. We extrapolated, as explained above, the inlet
 179 value of W_2 , and A was computed by using relations (12) and (13). The outlet

180 boundary condition is obtained by coupling the 1D model with a 0D model that
 181 consists of a system of differential equations linking pressure to flow at the 1D
 182 model exit. In fact, as shown in Alastruey et al. [4], the behavior happening
 183 beyond the 1D-modeled arteries can be represented by a 0D lumped-parameters
 184 model, also known as the Windkessel model, that simulates hemodynamics by
 185 electrical circuit analogy. Pressure gradient is represented by a potential dif-
 186 ference, blood flow by the electrical current and hydraulic impedance by an
 187 electrical impedance. Hydraulic impedance combines friction loss, arterial wall
 188 elasticity and blood flow inertia, which are modeled by a resistance, a capacitor
 189 and an inductance respectively. A three-element (RCR) Windkessel model is
 190 a good compromise accounting for wall compliance and resistance in the pe-
 191 ripheral network as well as proximal aortic impedance Wetterer [48], Westerhof
 192 et al. [47]. It is a combination of an R-model and an RC-model that improves
 193 considerably the behavior of the original two-element model proposed by Otto
 194 Frank Frank [20] by removing pressure and flow oscillations Stergiopoulos et al.
 195 [43], Alastruey [1]. The first resistance Z_c is the proximal aorta characteristic
 196 impedance, the compliance C in parallel to the resistance R_2 simulate respec-
 197 tively the volume compliance and the resistance of the vascular network found
 198 downstream the abdominal aorta (Figure 2).

199 The inlet flow of the R model, which is the 1D-model outlet flow ($Q_{in-0D} =$
 200 Q_{out-1D}), is:

$$Q_{in-0D} = \frac{P_{in-0D} - P_C}{Z_c} \quad (15)$$

201 As for the RC-model:

$$C \frac{dP_C}{dt} - \frac{P_{out-0D} - P_C}{R_2} - Q_{in-0D} = 0 \quad (16)$$

202 P_{out-0D} is the pressure at the 0D-model exit. We compute P_C at each time
 203 step by explicitly discretizing equation (16):

$$P_C^{N+1} = f(P_C^N, Q_{in-0D}^N, P_{out-0D})$$

204 then use the relation at the 1D-model exit:

$$Q_{in-0D}^{N+1} = \frac{P_{in-0D}^{N+1} - P_C^{N+1}}{Z_c}$$

205 *MRI experiments*

206 A straight flexible silicone tube was used as a phantom to validate the
 207 1D-model. The MRI setup (Figure 3), described in a previous work Khalife
 208 et al. [23], consisted of the phantom connected to a non-magnetic CardioFlow
 209 5000MR pump (Shelley Automation, Toronto, Canada) that can generate a pro-
 210 grammable flow pattern. We used a blood mimicking fluid (BMF, 62% water
 211 and 38% glycerol) with similar magnetic and physical properties as blood. It
 212 has a Newtonian behavior which is a safe assumption for blood circulating in
 213 large arteries Ottesen et al. [33]. Two MRI-compatible optical pressure sensors
 214 (Opsens, QuÃbec, Canada) were introduced into the phantom. These sensors
 215 allow simultaneous pressure measurements used as validation for the simulation.

216 To test the 1D-model application on a curved segment and estimate the
 217 error, the tube phantom was replaced with an aortic phantom (Elastrat, Geneva,
 218 Switzerland) comprising an ascending aorta branch (AAo) and a descending
 219 aorta branch (DAo). Pressure sensors were introduced at the AAo entrance and
 220 at the DAo exit.

221 A sinusoidal waveform of amplitude 150mL.s^{-1} was applied in the tube phan-
 222 tom setup. In the aorta phantom setup, a flow wave similar to the one measured

223 at the left ventricle exit was programmed on the pump control system with an
224 amplitude of 250mL.s^{-1} and a period of 0.8s (75bpm).

225 Experiments were performed on a 1.5T MRI system (Achieva; Philips, Best,
226 the Netherlands) using SENSE Flex-L coils. Phase-Contrast (PC) flow-encoded
227 gradient-echo multiphase sequences were performed to assess through-plane ve-
228 locity on two slices, one situated at the phantom entrance with the first pressure
229 sensor and the second at the phantom exit, with the second pressure transducer.
230 Scan parameters: spatial resolution: $1 \times 1 \times 8\text{mm}^3$, echo time/repetition time
231 $\text{TE/TR} = 5.75/9.59\text{ms}$, velocity encoding $V_{enc} = 80\text{cm.s}^{-1}$ for a flowrate of
232 150mL.s^{-1} and $V_{enc} = 100\text{cm.s}^{-1}$ for a flowrate of 250mL.s^{-1} , 40 time frames
233 per cardiac cycle with ECG-triggering on the pump signal.

234 For the *in vivo* feasibility of the 1D-model, 3 healthy subjects (male, ages 27,
235 30 and 33) were included in our study after having filled out a written consent
236 form. All measurements were performed on a 1.5T MRI system with SENSE
237 cardiac coil. PC flow-encoded sequence covering the thoracic aorta was acquired
238 during breath-hold and ECG-gated gradient echo sequence (spatial resolution:
239 $1.1 \times 1.1 \times 8\text{mm}^3$, $\text{TE/TR} = 3.0/5.1\text{ ms}$, $V_{enc} = 180\text{cm.s}^{-1}$, 40 time frames per
240 cardiac cycle). Data were acquired in an axial plane to measure through-plane
241 velocity in the descending and the abdominal aorta (AbAo). A coronal plane of
242 the aorta was acquired to measure the distance between velocity encoding sites.

243 Arm pressure was measured before and after MRI acquisitions and used as
244 a reference for the 1D-model pressure simulations.

245 *Compliance measurements*

246 MRI images were processed using an in-house code. Aorta section was seg-
247 mented semi-automatically and visually assessed by the operator on the magni-
248 tude image then flow waveforms were computed from velocity data provided by
249 the corresponding phase images. The inlet flow is thus used as the model input

condition.

MRI non-invasive measurements were used to determine phantoms and aortic compliance through PWV. The method, described in a previous work Khalife et al. [23], is based on the transit time (TT) method commonly used for the PWV estimation Murgo et al. [30], Avolio et al. [6] (Figure 4).

For each experiment, the linear compliance is determined from PWV based on the Bramwell-Hill theory: $\mathcal{C} = \frac{A}{\rho(PWV)^2}$ Bramwell and Hill [9], Vullimoz et al. [46], where A is the vessel lumen area and ρ the fluid density.

Simulations

Simulations were performed on an in-house program using Matlab (the Mathworks, Natick, USA) by setting the domain geometry, fluid properties and MRI measured compliance for the pressure law. Flow Q , section area A and pressure P are computed on the whole domain as functions of time and space, then compared to measured data on the corresponding nodes.

The values of the input parameters were the following:

- Tube phantom : $L = 40\text{cm}$, $\mathcal{C} = 0.0012 \text{ cm}^2.\text{mmHg}^{-1}$. The RCR-model components were estimated from measurements: $Z_c = 1.24 \cdot 10^8 \text{ Pa.s.m}^{-3}$ was computed using the approach proposed in Mitchell et al. [28], $R_2 = 1.63 \cdot 10^8 \text{ Pa.s.m}^{-3}$, $C_{per} = 0.288 \text{ cm}^3.\text{mmHg}^{-1}$ and $P_{out-0D} = 0$.
- Aortic phantom: $L = 21.5\text{cm}$, $\mathcal{C} = 0.051\text{cm}^2.\text{mmHg}^{-1}$, $Z_c = 2.63 \cdot 10^6\text{Pa.s.m}^{-3}$, $R_2 = 1.07 \cdot 10^8\text{Pa.s.m}^{-3}$, $C_{per} = 0.137 \text{ cm}^3.\text{mmHg}^{-1}$.
- *In vivo* : the RCR parameters at the outlet were taken from Alastruey et al. [4] because of the absence of invasive measurements. The characteristic impedance is $Z_0 = \rho \frac{c_0}{A_0}$. The peripheral resistance is $R_2 = R_T - Z_c$ where $R_T = 1.89 \cdot 10^8\text{Pa.s.m}^{-3}$ is the complete vascular system resistance.

275 The simulations were performed on a 100 meshgrid on a scope of 10 periods,
 276 with an average simulation time of 328 s. A and Q are computed at each time
 277 step, $\Delta t = 10^{-4}$ s. To evaluate results, the root mean square error in % (RMS)
 278 was calculated between computed and measured pressure waves.

279 Furthermore, to evaluate the equivalence between the pressure law (5) and
 280 the compliance pressure law (8) proposed in this work, two simulations were
 281 carried out with the same flow conditions, but with different pressure laws. The
 282 flow waveform and the model parameters used were those of a physiological
 283 scenario as proposed in Alastruey et al. [4]: $A_0 = \pi \cdot 10^{-4} \text{m}^2$, $\beta_0 = \frac{4}{3} \sqrt{\pi} h E$ for
 284 the β_0 pressure law, with $E = 0.4 \cdot 10^6 \text{Pa}$, $h = 1.5 \cdot 10^{-3} \text{m}$, $C = 7.85 \cdot 10^{-9} \text{m}^3/\text{Pa}$,
 285 and compliance of $\frac{2A_0^{3/2}}{\beta_0}$ for the compliance pressure law.

286 3. Results

287 The results of the equivalence test between the pressure law (5) and the
 288 compliance pressure law (8) are depicted in figure 5. Based on these curves, we
 289 can say that the two pressure laws provide nearly identical results which means
 290 that the β_0 pressure law agree well with the linear pressure law with $C = \frac{2A_0^{3/2}}{\beta_0}$,
 291 as in equation (9). The RMS between both pressure curves is 6.2% . Hence,
 292 even if the pressure to section area relation is not perfectly linear in the range of
 293 physiological pressure, the error committed by considering it linear is negligible.

294 In phantom experiments, simulated pressure was compared to pressure sen-
 295 sor measurements.

296 The inlet flow and the simulated pressure within the tube phantom are
 297 shown in figure 6 for the sinusoidal flow waveform of amplitude 150mL.s^{-1} .
 298 Computed pressure curves show good agreement with the measured pressure in
 299 the tube phantom. RMS error computed between both is 8.15% and confirms
 300 the agreement of the model results with measurements.

301 Furthermore, the simulated pressure at the DAo outlet in the aortic phantom

was compared to pressure sensor measurements acquired at the same location (Figure 7b). RMS between both pressure curves is 4.97%. Velocity measurement noise caused oscillations on the MR measured flow which were reported on the simulated pressure. This is due to the relatively low velocity measured in the descending branch. Thus, the flow was filtered before simulations (Figure 7a).

Beyond this, pressure curves simulated by the 1D-model on a straight segment formed by the DAo and the AbAo in healthy volunteers are plotted in figure 8. Their contour is in good agreement with pressure curves reported *in vivo* in the literature Schnabel et al. [38], Murgu et al. [30]. Also, according to the reflection principle described in Salvi [37], the systolic pressure increases along the arterial tree and consequently the brachial systolic pressure overestimates the aortic systolic pressure up to 20mmHg Levick [25]. Thus in figure 8, the simulated systolic pressure in the aorta is smaller than the systolic pressure measured on each volunteer. Mean arterial pressure (MAP) computed on the simulated and measured pressure is added in figure 8.

4. Discussion

In this work, we showed that by combining a 1D aorta model to non-invasive MRI velocity measurements, we were able to estimate aortic pressure. Flow MRI had commonly been used to derive pressure differences, so estimating a local pressure without having to set a null pressure point is a step forward in cardiovascular diagnosis.

The MR-measured flow was used as input for the 1D-model and a coupled 0D-model was used as an output condition. The pressure law, which relies on invasive measurements in formal studies, is replaced by a non-invasive pressure law based on the aortic compliance. This compliance was measured from MRI assessed PWV. The model simulations were successfully validated on an MRI setup with a straight phantom, an aortic arch phantom and *in vivo*.

329 At first, the simulations were applied on a simple straight tube to respect the
 330 model conditions and simplifications. Pressure values obtained with simulations
 331 were in good agreement with measurements and showed satisfactory results as
 332 in previous studies Alastruey et al. [2]. Then, simulations were applied on an
 333 aortic phantom and results showed a negligible error when compared to pressure
 334 measurements, meaning that, in the physiological range of blood velocity, the
 335 1D-model coupled with MRI flow measurements and subject-specific compliance
 336 pressure law would be enough to describe the aortic flow behavior. Results *in*
 337 *vivo* showed that it can estimate realistic pressure curves in the DAo, in agree-
 338 ment with the brachial pressure measured in the clinical routine. A bias was
 339 observed between the two brachial measurements carried out before and after
 340 the MRI exam due to the subject changing state going from agitated prior to
 341 the exam to rested state after the exam was completed. Another explanation
 342 could be the difference in the subjects position (seated vs. supine) during mea-
 343 surements. Consequently, the measured arm pressures might have been different
 344 than what the simulations would yield.

345 This 1D-model, which is frequently used to model the aorta, is based on
 346 strong assumptions, considering only radial and axisymmetric wall deforma-
 347 tions, constant pressure on a section along the tube axis, thus only depending
 348 on time and on the axial coordinate z , dominance of axial velocity, absence of
 349 turbulence and neglecting the vessel wall inhomogeneities. Nevertheless, the
 350 model serves its initial role of estimating an absolute pressure in a simple and
 351 fast non-invasive manner, thus it is widely used in large artery modeling for
 352 its fair reproduction of the physiological configuration. Even if the model is
 353 one-dimensional and must be applied to a straight segment, its application to a
 354 curved aortic phantom did not greatly affect the agreement of simulated pres-
 355 sure and measured pressure at the phantom outlet. However, the effect of an

356 inertance in the 0D-model represented by an inductor in the electrical circuit
357 could be considered. The inertance element as proposed in Segers et al. [39]
358 could absorb the oscillations which appear on the pressure curve.

359 The compliance based pressure law is simple, non-invasive and widely an-
360 chored in clinical practice and its association to the 1D-model may be more
361 advantageous than the previously used pressure laws reported here. The model
362 showed satisfactory results and good reproduction of invasive measurements
363 recorded on a tube phantom. Additionally, we have shown that this simpli-
364 fied linear pressure law is in good agreement with the $\sqrt{A}$ law proposed in the
365 literature.

366 Some improvements could be made to the 1D-model in the future. The model
367 could be made more patient-specific by adapting the 0D-model parameters to the
368 patient at the outlet boundary condition. Since these parameters are taken from
369 the literature and are not patient-specific, previous work Willemet et al. [49] has
370 proposed to measure peripheral resistance by introducing pressure catheters in
371 the femoral artery. However, this is a disadvantage to a method which offers to
372 be strictly non-invasive. Further studies are needed to propose a non-invasive
373 way to measure the RCR parameters. Also, the model supposes a constant
374 compliance along the artery which is a non-realistic condition when studying
375 aortic disease or stenosis. This problem will be addressed in future studies.

376 In the future, it would be interesting to compare the pressure estimation
377 found in our model with catheter measurements in an animal experiment. Also,
378 more *in vivo* experiments would allow assessing the reliability and the repeata-
379 bility of such estimation techniques in the aorta and to study pressure changes
380 in patients suffering from cardiovascular diseases or under various stress condi-
381 tions.

382 **Conflict of interest statement**

383 The authors declare that there are no conflicts of interest.

384 **References**

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Figure 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 in respect to the radial vector.

Figure 2: Coupling the 1D model outlet with the 0D model: a three-element Windkessel model (RCR) forms the outlet boundary condition of the 1D model. $(Q_{in-0D}, P_{in-0D}) = (Q_{out-1D}, P_{out-1D})$ and (Q_{out-0D}, P_{out-0D}) are flow and pressure at the inlet and the outlet of the RCR model respectively. To solve the RCR model, two sub-models, RC and R, are considered and then combined. P_C is the pressure across C at the exit of the RC model and at the entrance of the R model.

Figure 3: Experimental MRI setup with the MR-compatible pump and the aorta phantom. Optical pressure sensors are introduced in two locations on the phantom.

(a)

(b)

(c)

Figure 4: Transit time estimated on MR flow waves in subject A. (a) magnitude image of the acquired axial plane. (b) AAo and DAo are visible in hypersignal; through-plane velocity image in AAo and DAo : flow waveforms are extracted from the encoded velocity. (c) The TT method for PWV measurement. The TT, which is the delay Δt that a waveform needs to travel a distance L , is used to compute the wave velocity: $PWV = \frac{L}{\Delta t}$. Here, Δt was measured between the feet of flow waveforms extracted at two sections: tangent lines are obtained by the linear regression of the upslope segment on each waveform.

(a)

(b)

(c)

Figure 5: Results of the 1D model using pressure laws with β_0 in equation (5) (straight line) vs. with compliance $C = \frac{2A_0^{3/2}}{\beta_0}$ (dotted line). Simulations are performed with a pulsed ventricular flowrate and parameters given in Alastruey et al. [4]. Plotted results are such as: (a) flow waveform, (b) mean velocity and (c) pressure waveform simulated at three sites of the artery, at the inlet (0m), in the middle (0,2m) and at the exit (0,4m).

(a)

(b)

Figure 6: Pressure curves in the tube phantom with a sinusoidal flow waveform of amplitude 150 mL.s^{-1} . Boundary conditions used are the measured flow at the inlet (a) and given by the 1D-0D coupled model at the outlet. (b) Simulated pressure curves at the sensors locations are compared to pressure measurements.

(a)

(b)

Figure 7: (a) Flow measured at the entrance of the aorta phantom (solid line). Due to the motion of the phantom and the low velocity amplitude, oscillations are visible at the measurements curves and are reported in the 1D model simulation results. Hence, the fitted measured flow (dashed line) is used for pressure calculation. (b) Simulated vs. measured pressure at the exit of the aorta phantom with a ventricular flow of amplitude 250 mL.s^{-1} . Boundary conditions used are the flow at the entrance of the ascending aorta (AAo) and given by the 1D-0D coupled model at the outlet of the phantom.

(a) subject A, 27 years old

(b) subject B, 30 years old

(c) subject C, 33 years old

Figure 8: Healthy volunteers' aortic pressure simulated in a straight segment going from the descending aorta (DAo) to the abdominal aorta (AbAo). The horizontal dashed lines in orange show the averaged brachial diastolic and systolic pressure (P_{diast} and P_{syst}) measured for each subject. In green, the corresponding mean arterial pressure (MAP) lines are computed on the simulated pressure (solid line) and the arm measurements (dashed line). The inlet condition is the flow measured at the base of the descending aorta and the outlet condition is given by the 1D-0D coupled model. Pressure law based on compliance measurement (equation 8) is used. Mean arterial pressure is given by: $MAP = P_{diast} + \frac{1}{3}(P_{syst} - P_{diast})$.