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Technical note

A simplified method to account for wall motion in patient-specific blood flow simulations of aortic dissection: Comparison with fluid-structure interaction

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ABSTRACT

Aortic dissection (AD) is a complex and highly patient-specific vascular condition difficult to treat. Computational fluid dynamics (CFD) can aid the medical management of this pathology, yet its modelling and simulation are challenging. One aspect usually disregarded when modelling AD is the motion of the vessel wall, which has been shown to significantly impact simulation results. Fluid-structure interaction (FSI) methods are difficult to implement and are subject to assumptions regarding the mechanical properties of the vessel wall, which cannot be retrieved non-invasively. This paper presents a simplified 'moving-boundary method' (MBM) to account for the motion of the vessel wall in type-B AD CFD simulations, which can be tuned with non-invasive clinical images (e.g. 2D cine-MRI). The method is firstly validated against the 1D solution of flow through an elastic straight tube; it is then applied to a type-B AD case study and the results are compared to a state-of-the-art, full FSI simulation. Results show that the proposed method can capture the main effects due to the wall motion on the flow field: the average relative difference between flow and pressure waves obtained with the FSI and MBM simulations was less than 1.8% and 1.3%, respectively and the wall shear stress indices were found to have a similar distribution. Moreover, compared to FSI, MBM has the advantage to be less computationally expensive (requiring half of the time of an FSI simulation) and easier to implement, which are important requirements for clinical translation.

1. Introduction

Aortic Dissection (AD) is a life-threatening vascular condition initiated by a tear in the intima layer that allows the blood to flow within the aortic wall and leads to the formation of two distinct flow channels, the true lumen (TL) and the false lumen (FL), separated by the so-called intimal flap (IF) [1].

The clinical decision-making process around Stanford type-B dissections (i.e. ADs involving only the descending aorta) is complex and patient-specific [2]. Surgical intervention is the preferred choice in the presence of complications, whereas 'uncomplicated' ADs (referring to ADs without complications, such as organ malperfusion, rupture, refractory pain or hypertension, at presentation) [3] are usually managed by controlling the blood pressure [4]. Long-term prognosis of medically treated ADs remains poor, with aortic dilation and late-term complications reported in 25–50% of the cases within 5 years [5].

Patient-specific computational fluid dynamics (CFD) can inform the decision-making process around the disease and aid the identification of patients prone to adverse outcomes by providing detailed information about haemodynamic factors [6–9]. Moreover, numerical models may support clinicians by virtually simulating different interventional strategies [10,11].

The use of three-dimensional (3D) rigid models that neglect the effects that vessel wall motion exerts on the fluid dynamics has been shown to impact simulation results considerably [12]. Vascular compliance, IF motion and the critical role of haemodynamics on AD prognosis (e.g. tear propagation and rupture) necessitate the use of more advanced fluid-structure interaction (FSI)

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approaches to simulate the flow in this complex aortic condition. FSI couples CFD simulations with finite element modelling (FE) of the aortic wall; however, this method is subject to significant and additional modelling assumptions regarding the mechanical properties of the vessel, which are patient-specific and not known for the case of AD [13]. In addition, FSI models are difficult to setup and demand significant computational effort to be resolved. ADs are arguably one of the most challenging aortic pathologies to simulate and hence it is not surprising that there are only a handful of studies on AD accounting for wall motion in the literature [12,14,34]. Chen et al. [15] recently presented an FSI model of an idealised dissected porcine aorta without re-entry tear, assuming a homogeneous linear-elastic material model. The study presents a first attempt to validate AD FSI simulations against bench experiments.

Two key objectives of patient-specific modelling and simulation for clinical support are (a) to gather as much information as possible from the patient, if possible, via non-invasive techniques and (b) to perform detailed computations in clinically-meaningful timescales. Neither is currently achievable with FSI due to the lack of patient-specific arterial wall properties and associated high computational costs mentioned above.

However, imaging can help in this respect by providing significant patient-specific detail on the motion of the wall and the IF. With this in mind and in view of the aforementioned limitations of FSI, this paper presents a simplified and computationally efficient method to account for the motion of the IF and vessel compliance in type-B AD CFD simulations, circumventing the need to use full-FSI techniques. The 'moving-boundary method' (MBM) presented here can be tuned with non-invasive patient-specific measurements (e.g. two-dimensional cine magnetic resonance imaging, 2D cine-MRI). It aims at capturing the main fluid dynamic effects due to the fluid-solid interaction by representing the motion of the vessel wall and IF in a simplified, and yet meaningful way. It adopts a physiologically-supported calculation based on pressure differences and fluid forces calculated in the computational domain along with wall stiffness estimated in different regions of the vessel.

A description of the proposed method is presented in the following section, including its validation against the one-dimensional (1D) solution of flow through an elastic straight tube. The MBM is then applied to a type-B AD case, which was previously simulated with full-FSI [12]. In order to directly compare both methods, displacement data available from the FSI simulation was taken as the benchmark upon which the MBM was tuned. Results for both cases are presented and discussed in Section 3, including the comparison of haemodynamic results obtained with the MBM and FSI simulations of the AD case.

2. Methods

2.1. Description of the method

The proposed moving-boundary method (MBM) allows the motion of the 3D model boundaries in a CFD framework by means of a deformable mesh, avoiding the detailed structural analysis of the arterial wall. It is assumed that the vessel wall and IF are in static equilibrium with the fluid forces at each time-step, and that dynamic and viscoelastic effects are negligible. The displacements of the aortic external wall and IF follow the local surface-normal direction, and are linearly related to the fluid forces acting on them. The displacement δ_i [m] of each mesh node i on the external vessel wall is prescribed by Eq. (1):

$$\delta_i = \frac{p_i - p_{ext}}{K_i}\vec{\mathbf{n}}_i \tag{1}$$

where p_i [Pa] is the pressure at node i, p_{ext} [Pa] is the external pressure set as equal to the diastolic pressure, $\vec{\mathbf{n}}_i$ is the local unit normal vector in the outward direction, and K_i [N/m^3] is a measure of the wall stiffness at node i. Under the hypothesis of a circular cross section, K_i can be related to the vessel wall distensibility $\mathcal{D}$ [Pa^{-1}] as follows:

$$K_i = \frac{2}{\mathcal{D}}\sqrt{\frac{\pi}{A_i^0}} \tag{2}$$

where A_i^0 [m^2] is the diastolic cross-sectional area at the location of node i.

The IF is modelled as a zero-thickness membrane and is discretised into a number of patches (i.e. surface regions); the displacement δ_j [m] of each mesh node on patch j is prescribed by Eq. (3):

$$\delta_j = \frac{\vec{\mathbf{F}}_j^{tm}}{K_j\ A_j} \tag{3}$$

where $\vec{\mathbf{F}}_j^{tm}$ [N] is the surface normal-transmural force (TMF) on patch j, considering the viscous and pressure forces acting on both TL and FL sides of the IF patch; A_j [m^2] is the surface area of patch j; K_j [N/m^3] is the stiffness coefficient assigned to patch j. For each patch, K_j can assume two values, namely K_j^{FL} and K_j^{TL} depending on whether $\vec{\mathbf{F}}_j^{tm}$ points in the direction of the FL or the TL, respectively. Thus, it is possible to account for the different mechanical behaviour that the IF can exhibit in case of extension (i.e. TL expansion) or contraction (i.e. TL compression), as highlighted by Karmonik et al. [16]. The tuning of K_i and K_j is based on patient-specific displacement data obtainable, for instance, from cine 2D MR images.

In this study, the MBM was implemented in ANSYS-CFX 17.0 (ANSYS Inc., PA, USA). The mesh motion was obtained by specifying the displacements of the boundaries following Eqs. (1) and (3), defined in CFX via the CFX Expression Language (CEL) [17]. The mesh displacement equations were solved so as to obtain an implicit two-way coupling between mesh motion and fluid dynamics.

2.2. Flow in an elastic straight tube

The flow through an elastic straight tube was studied to provide a preliminary validation of the proposed method, and to prove its ability to capture wave transmission phenomena. The solution obtained with a 3D model implementing the MBM was compared to the solution of a 1D elastic tube, which is a common approach in the study of wave propagation in the arteries and has been thoroughly validated in the literature [18–20].

The parameters of the tube resemble those of a healthy human aorta, and were taken from Alastruey et al. [19]. The tube has a length $l = 40$ cm, an initial lumen radius $r = 1$ cm, a wall thickness $h = 1.5$ mm and a Young's modulus $E = 0.4$ MPa. The blood was modelled as a Newtonian fluid (density $\rho = 1056$ kg/m^3, dynamic viscosity $\mu = 3.5$ cP). At the inlet, a periodic flow waveform was applied, with the systolic phase modelled as a half-sinusoidal waveform, and the diastolic phase as zero-flow [19] (mean flow $Q = 3.8$ l/min, cycle period $T = 0.8$ s, systolic phase $T_{sys} = T/3$). A three-element Windkessel (WK3) model was coupled at the outlet. WK3s are electrical analogues of the downstream vasculature and consist of a proximal resistance, R_1, connected to a compliance, C, and a distal resistance, R_2 (Fig. 1a). WK3 parameters were taken from Alastruey et al. [19] ($R_1 + R_2 = 1.418$ mmHg s ml^{-1}, $C = 0.840$ ml mmHg^{-1}). R_1 was set equal to the characteristic impedance of the elastic tube ($R_1 = 0.155$ mmHg s ml^{-1}), and an outflow pressure $P_{out} = 9.98$ mmHg was considered.

A 3D model of the tube employing the MBM was implemented in ANSYS-CFX 17.0. The parameter K_i was derived from the

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