**Technical note**

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


**Mirko Bonfanti[a,*], Stavroula Balabani[a], Mona Alimohammadi[a], Obiekezie Agu[b], Shervanthi Homer-Vanniasinkam[a,c,d], Vanessa Díaz-Zuccarini[a,*]**

[a] Department of Mechanical Engineering, University College London, WC1E 7JE, UK

[b] University College London Hospital, NW1 2BU, UK

[c] Leeds Teaching Hospitals NHS Trust, LS1 3EX, UK

[d] University of Warwick Medical School & University Hospitals Coventry and Warwickshire NHS Trust, CV4 7AL, UK

* Correspondingi Londi

## Abstract

Aortic dissection (AD) is a complex and highly patient-specific

# 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) 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 [12,14,15]. Chen et al. [16] 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

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variables was carried out, as described in the paper by Alimohammadi et al. [12]. The grid consisted of about 483,000 elements.

### 2.3.1. Geometry and mesh

The 3D flow domain was generated from the geometry used for the FSI model, representing an acute type-B AD of a 54-year-old female patient. The original geometry was extracted from a computed tomography (CT) scan of the entire aorta performed with a 64-slice Siemens scanner (Siemens AG, Germany; in-plane resolution=0.7 mm, inter-slice distance = 0.7 mm; for details on the image segmentation the reader is referred to our previous work [12]). The geometry did not include the abdominal aortic branches and renal arteries because the CT scan resolution did not allow an accurate segmentation of these small arteries. The surface of the IF was discretised in roughly 200 patches (average surface area = 22 ± 12 $mm^2$) with the aid of the +NURBS module of ScanIP image-processing software (Synopsys Inc., CA, USA).

The fluid volume was discretised using ICEM-CFD (ANSYS Inc.) with an unstructured tetrahedral mesh in the core region and 7 prism layers at both IF sides and vessel wall, with dimensionless height of the near wall cells ($y^+$) < 1. The mesh was created using the same parameters adopted for the fine mesh used in the FSI model, for which a mesh sensitivity analysis checking for Time-Averaged Wall Shear Stress (TAWSS), Oscillatory Shear Index (OSI) and velocity

### 2.3.2. Boundary conditions

In order to perform a comparison between the MBM and FSI models, the same CFD boundary conditions (BCs) and fluid properties employed in our previous work [12] were applied to the MBM model. Blood was modelled as an incompressible fluid with a density of 1056 kg/$m^3$ and a non-Newtonian viscosity described by the Carreau-Yasuda model, with parameters from [25]. The shear-stress transport turbulence model was used with 1% turbulence intensity at the inlet, as in our previous work [12]. The fluid flow and mesh motion BCs are detailed in Table 1. WK3 models were coupled at the outlet branches with parameters shown in Table 2. The mean Reynolds and Womersley numbers, based on the inlet diameter of the aorta, were equal to 831 and 25, respectively. The peak Reynolds number was 4252, which is close to the critical Reynolds number for transition to turbulence uewWLLHEJ3W,WLLHEJ3W,63K7,r*zLl

**The parameter $K_i$ is related to the wall distensibility □ via Eq. 2. □ can be estimated in different aortic regions *k* using Eq. 6 [27]. In this case, we selected the following regions: ascending aorta, upper aortic branches and descending dissected aorta, where different vessel diameter or wall-structure alteration (due to the dissection) would be expected to affect the value of □.**

$$\square_k = \frac{\Delta A_k}{A_k^0 \, \Delta p_k} \qquad \textbf{(Eq. 6)}$$

**where**

significantly lower OSI in these regions (about -0.15 in absolute values [12]). In fact, the motion of the wall (both in the FSI and MBM models) highly affects the flow in the closed-end parts of the FL, where the alternate expansion and contraction of the vessel due to pressure fluctuations enhances the oscillatory nature of the flow. The correct assessmeKíEZ,I*zWHEKíLHJJ3ZE,I*zWHEJKJL,a*zLHEJ3W,*zEHUK33U, *LKHzLHLLW

aJ,e*HíJKJL,n*ZHL3KZ3W,t*z3WH3LW3,7KU4zZLHíL3JJ,e*H*zEHJJ3ZE,o*gnificantly IceiozLHJJ3K7, *z3WH3LW3,O*E

L3KZE, *zZZHWW777Woeussoo o InciczLHEJJ77,o*HL3KZE, *zczLHEJJ77,o*ZHL3KZE,e*íH33JJ77, *LKHJEZU,m*zUHHJJJ77,o*ZHL cross-section of the dissected aorta largely deviates from the physiological shape. Due to the high complexity and heterogeneity of ADs, careful consideration needs to be given to each case, and it is up to the modeller to assess if these approximations are appropriate for the specific case under analysis.

The MBM implements a linear, elastic relatidLiat

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## Ethical approval

The study was ethically approved (NHS Health Research Authority, ref: 13/EM/0143). The patient signed the appropriate consent form.

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