In vivo, in silico, in vitro patient-specific analysis of the haemodynamics of a Type-B Aortic Dissection

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1. Introduction

Aortic dissection (AD) is a serious condition that occurs when a tear in the aortic wall allows blood to flow within the layers of the vessel. The optimal treatment of 'uncomplicated' acute/subacute Type-B aortic dissections (uABADs) continues to be debated. uABADs are commonly managed medically, but up to 50% of the cases will develop complications requiring invasive intervention [1].

AD is a highly patient-specific pathology in which morphological features have high impact on the haemodynamics. However, there is still a limited understanding of the fluid mechanics phenomena influencing AD clinical outcomes. Flow patterns, pressure, velocities and shear stresses are at the same time difficult to measure and extremely important features for this pathology.

Personalised computational fluid dynamics models (CFD) are being investigated as a tool to improve clinical outcome [2]. However, such models need to be rigorously validated in order to be translated to the clinic, and such validation procedures are currently lacking for AD. This scarcity of data may be supplemented using *in silico* and *in vitro* models, in which these parameters can be studied and compared for validation purposes.

In this work, a unique *in vitro* and *in silico* framework to perform personalised analyses of Type-B AD, informed by *in vivo* data, is presented.

Experimental flow rate and pressure waveforms, as well as detailed haemodynamics acquired via Particle Image Velocimetry (PIV), are compared at different locations against computational CFD results.

2. Materials and Methods

2.1. Clinical dataset and vessel segmentation

A rich clinical dataset was acquired from a 77-year-old man with a chronic Stanford Type-B AD, which includes PC-MRI, 2D cine-MRI and CT scans (ethics: HRA ref. 12/YH/0551, Leeds Hospital ref. 788/RADRES/16).

The dissection originated approximately 40 mm distal to the left subclavian artery, extended throughout the length of the descending aorta and terminated about 10 mm distal to the coeliac trunk. From CT scans, one entry tear was located approximately 10 mm distal to the proximal end of the dissection; no other communication between the TL and FL was evident from CT data, confirmed by a reduced flow in the FL observed in PC-MRI data.

In order to perform *in vitro* flow measurements and *in silico* CFD simulations, the geometry of the dissected aorta was extracted using a semi-automatic segmentation based on thresholding, implemented in ScanIP (Synopsys Inc., USA). Smoothing operations were used on the resulting mask to reduce pixellation artefacts. The IF separating the FL from the TL was identified based on greyscale-value differences in the two lumina. The geometry was cropped right after the end of the dissection in the descending aorta. The resulting geometry includes the ascending aorta, aortic arch and three upper branches (i.e. Brachiocephalic Trunk, BT; Left Subclavian Artery, LSA; Left Common Carotid, LCC), and descending aorta (DA), proximal to the renal branches. The geometry was imported in a CAD software and regular-geometry outlet were created to facilitate the connection to the experimental flow rig.

2.2 Experimental setup

A 3D phantom of the patient-specific dissected aorta was manufactured by 3D printing technology (Materialise, Belgium) to obtain a rigid, transparent model. Surface polishing was performed to increase transparency and reduce inner layer roughness of the material.

The model was connected to a bespoke pulsatile flow circuit described in [3]. The latter comprises a computer controlled pulsatile pump and left ventricle simulator, aortic dissection phantom, a tunable 3-element Windkessel (*WK3*) impedance physical model for each outlet of the aorta, and a preload atrial reservoir (Figure 1a). The blood mimicking fluid was a Potassium thiocyanate water solution (70% by weight). The mixture was formulated in order to match the refractive index of the phantom (RI = 1.51).

Pressure and flowrate waveforms were continuously measured by several pressure transducers (Omega Engineering, UK) and an ultrasound flow meter (Sonotec, Germany), respectively. The signals were digitised at 200 Hz sampling frequency, recorded with an AD converter, and handled by a purpose designed LabView virtual instrument.

To perform PIV measurements, the flow was seeded with silver coated hollow microparticles with a mean diameter of 10 μm, injected into the flow upstream of the phantom and allowed to disperse uniformly within the aortic model.

Tracers were illuminated by a pulsed Nd:YAG laser (Litron Lasers, Bernoulli, UK) emitting 532 nm wavelength light. Particle image pairs were acquired with a CCD camera (Imperx, USA) at a sapling rate of 25 Hz with a resolution of 4000 x 3000 pixels with a time interval of 200-1000 ms, depending on the velocity magnitude. Image pre-processing was performed to improve the image quality and minimize the error. The recordings were analysed using a cross-correlation method starting with an interrogation area of 64 x 64 pixels and ending with an area of 32 x 32 pixels, overlapping by 50%. For each time step, 200 pair images were used to obtain the vector field. Lastly, post-processing was performed using Tecplot (Tecplot, Inc., USA).

2.3 Haemodynamic Simulation

The extracted AD geometry was discretised with ICEM- CFD (ANSYS Inc., USA), adopting a tetrahedral mesh in the core region and seven prism layers at the walls. The resulting mesh consisted of about 506000 elements. The Navier–Stokes (NS) and continuity equations for 3D time- dependent flows were solved with finite-volume-based CFD solver ANSYS-CFX 19.0 (ANSYS Inc., USA). The vessel wall was modelled as rigid. The fluid was treated as Newtonian incompressible fluid with a dynamic viscosity and density matching the experimental ones. WK3 models were coupled to the outlets as boundary conditions. Simulations were run until reaching a periodic steady state. This was achieved within three cardiac cycles after appropriate initialization; the last cycle was used for the analysis of results. Post-processing was performed using both CFD-Post (ANSYS Inc., USA) and Matlab (Mathworks, USA).

2.4 Personalisation approach and model inputs

Available *in vivo* data was first used as a target for the *in vitro* experiment (Figure 1b). The inlet flow rate wave $Q_{in}(t)$, systolic and diastolic pressures (P_{sys} and P_{dia}) and cardiac output (CO) distribution were used as inputs of a previously developed personalisation procedure [2] to calibrate the parameters of the pump system and *WK3* models. In order to facilitate direct comparison between the two models, the *in vitro* flow rate curves at the inlet and upper branches, and pressures $P_{in}(t)$, were used to inform the boundary conditions for the *in silico* simulation (Figure 1c). Flow rate and pressure waveforms at different locations, as well as the velocity field in the vessel, were then qualitatively and quantitatively compared amongst the models.

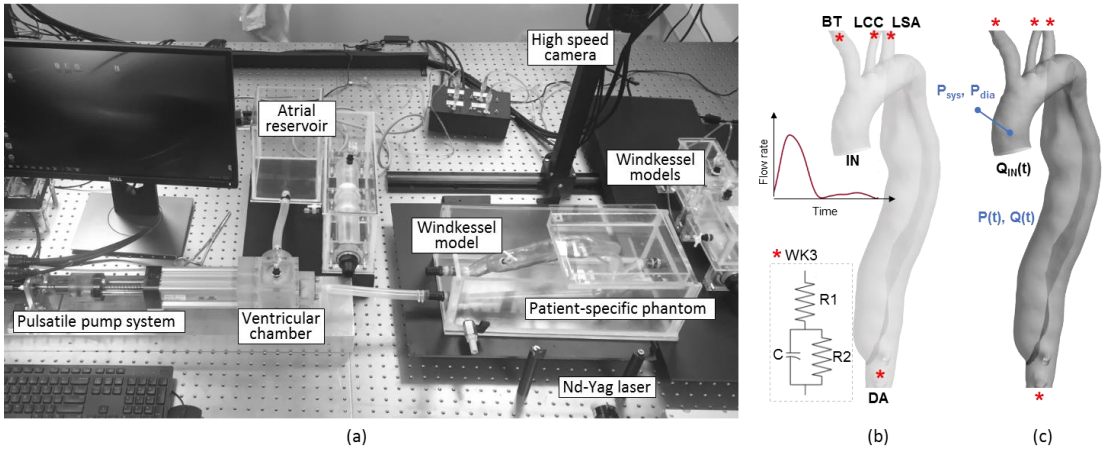

Figure 1. *(a) Picture of the in vitro apparatus; schematics of the boundary conditions for the (b) in vitro model and (c) in silico simulation.*

3. Results

The personalisation approach and tuning procedure, as well as the methodology to exchange information amongst the models, allowed to obtain the target clinical flow rate distribution and systolic and diastolic pressures with a very good agreement. A comparison between the *in vivo* target pressures, *in silico* and *in vitro* inlet pressure curves is shown in figure 2a. The bar plot in figure 2b shows the *in silico – in vitro* comparison of the mean flow rate at all the outlet branches, with a maximum error of 7% for DA. The 2D velocity field acquired in different sections of the geometrical domain is compared to the corresponding velocity predictions of the computational model (figure 2c).

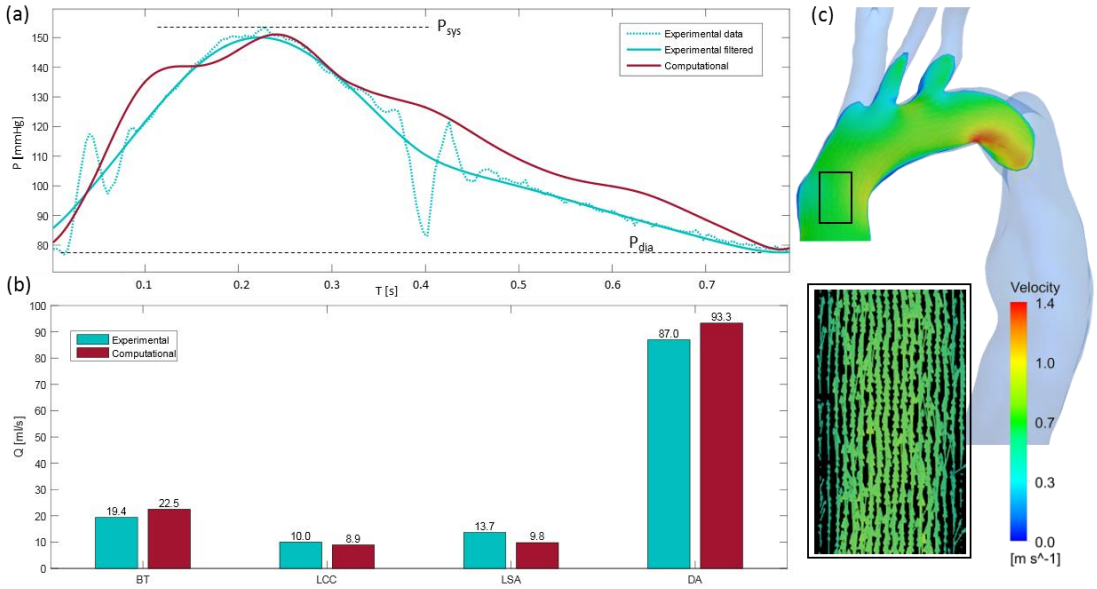

Figure 2. *(a) Comparison between the target systolic and diastolic in vivo pressures, and experimental and computational inlet pressure waveforms; (b) mean flowrate comparison between the two models and (c) example of velocity vector field obtained experimentally with PIV and in silico study.*

4. Discussion and Conclusions

Validation of numerical fluid dynamic predictions is essential to translate *in silico* tools for complex vascular pathologies to the clinic. This study represents the first attempt in literature to perform a comparison and validation of a patient-specific case of AD via computational, experimental and clinical data. The comparison of haemodynamic parameters reveals a good qualitative agreement. The results of this study demonstrate the accuracy and reliability of numerical and experimental methods to reproduce *in vivo* patient-specific conditions.

This work provides confidence that these models can be further utilised to study the haemodynamics in this patient-specific case of AD and potentially provide valuable clinical markers for disease progression and intervention, and test of surgical strategies.

5. References

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