Two-Way Aorta Blood-Artery Interaction using Computational Fluid Dynamics (CFD) Simulation

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ABSTRACT

The CFD method is a promising approach for developing a new non-invasive diagnostic tool in biomedical research. For example, analysis of the components of the blood flow through arteries can be investigated using CFD modelling. This is based on solving fluid dynamic equations such as the continuity and conservation of energy equations. CFD can also be used to develop and optimise devices and system design, to improve existing equipment and reduce costs. Therefore, biomedical engineers and scientists are increasingly employing CFD models to examine the circulatory system on both human and animal subjects to obtain valuable information that is difficult to determine clinically. CFD methods can also accurately simulate the stresses affecting the artery wall and the dynamic behaviour of the blood flow in its pulsatile form. Therefore, the outputs from CFD analysis can be used to reduce the risk of disease complication and enable a better understanding of the effects of hemodynamic stresses. The aim of this study is to investigate the behavior of the pressure waveforms in a realistic 3D Aorta geometry obtained from 2D-images. This investigations are done using Two-Way Fluid Structure Interaction (FSI) methods in ANSYS 2020 R2.

Keywords: Aorta; CFD; TwoWay FSI; Fluent; Transient Structural; Pressure Waveforms

Introduction

In this paper we investigate the aorta’s geometry which is classified as a complex geometry [1-8]. The aorta has three main branches: Brachiocephalic (BC); Left Common Carotid (LCC); and Left Subclavian (LS), which are all connected to the aortic arch [9-16] as shown in Figure 1. In constructing and modelling this geometry, it is necessary to consider the thickness of the artery wall and incorporate the bifurcation and branches. This simulation of the bifurcation and branches takes a long time; however, this can be reduced with a faster processor. Previous studies [9-16] modelled the aorta with simplified assumptions which give rise to minor divergences between the model and clinical trials. However, these differences are minimal and the results can still be validated against clinical trials. In this paper we are presenting a realistic model as shown in Figure 1. This figure shows the stages for converting 2D-images to a 3D STL files with fluid body (blood) and solid body (Aorta wall).

The modelling of the arterial pulsatile structure involves the analysis of its variables, namely, the deformation of the arteries, the pulsatile blood flow, and the interaction between them. This can be related to complex processes that could be the origin of arterial
diseases such as atherosclerosis [1]. Computational modelling analyses this disease using the Fluid Structure Interaction (FSI) scheme. The most common ill-effect of cardiovascular disease is to cause arterial narrowing called stenosis, which results in a reduction of blood flow to the tissues [2-3]. Previous studies incorporating FSI models with different degrees of stenosis showed that the wall deformation resulting from stenosis leads to a turbulent blood flow and higher Wall Shear Stress (WSS) [4-6].

Material and Methods

Artery Wall Properties. The healthy artery wall is modelled assuming a Hyperelastic Neo-Hookean model with Density of 1150 kg/m$^3$, Young’s modulus of E=0.5 MPa, Initial Shear Modulus of 1.74 MPa, Initial Bulk Modulus of 34.7 MPa and Poisson’s ratio of v=0.45 [8]. The boundary conditions are also set for the solid-fluid interface and the fixed ends of the inlet and outlets of the aorta geometry. These conditions combined with the CFD-Fluent analysis provide greater understanding of arterial wall displacement and deformation resulting through the pulsatile blood flow as shown in Figure 2. The inlet boundary condition is a waveform of blood flow velocity [9-10]. Three of the outlets’ boundary conditions are brachial artery pressure waveform, while the fourth outlet is an iliac pressure waveform as shown in Figure 2. All of these waveforms were invasively measured in a healthy person and were collected from patients undergoing left heart catheterization at Green Lane Hospital under Ethics approval number (NTX/09/11/109). The FSI model is used to investigate the effect of pulsatile blood flow on the aorta wall deformation, compliance and displacement for healthy conditions. Blood Density 1060 kg/m$^3$. The material properties of blood as a non-Newtonian fluid. According to literature, a common shear-thinning model used to replicate blood is the Carreau-Yasuda model [8], which takes the form of:

$$\mu = \mu_\infty + \frac{(\mu_s - \mu_\infty)}{\left(1 + \left(\frac{\dot{\gamma}}{\lambda}\right)^{\frac{1}{n}}\right)^{\frac{1}{n}}},$$

where $\mu_s$ is the zero-shear viscosity, $\mu_\infty$ is the infinite-shear viscosity, and $\dot{\gamma}$ is the shear strain rate, time constant $\lambda=3.313$ (s) and Power-Law Index n=0.3568 [8].

![Figure 2:](image)

For the aorta geometry several attempts are made to identify the size of the meshes to be used. Starting with simple geometry and gradually increasing the number of elements, we are able to identify the most suitable and accurate mesh for the current investigation. This leads to a mesh of 1736473 nodes and 8252952 elements for the blood body, and 13831 nodes and 13791 elements for the aorta wall, as shown in Figure 3. The Aorta (blood and Artery) are meshed using the tetrahedron path independent method, which has the advantage of producing uniformly sized meshes, especially for small arteries. Furthermore, there are several methods for assessing mesh element quality (mesh metrics) such as the element skewness, which is available within ANSYS® 20R2. In the present work the Skewness is between a minimum of 0.46 and a maximum of 0.75 which is within the acceptable range. The above mesh has demonstrated a good statistical measure for moving meshing nodes, and convergences are assessed as being in the acceptable range by comparing them with the literature [11,12]. In this computational model, a maximum number of iterations are performed, based
on the level of residual reduction or absolute residual levels as a convergence criterion; that is iterations continue until an appropriate level of convergence is reached.

**Results and Discussion**

The model presented in this paper is compared with the literature and validated against clinical trials on artery geometry. Firstly, the healthy model is validated against literature data and the clinical trial results. These validations are also performed by investigating the pressure pulse wave at different locations such as the ascending, descending and the renal arteries. The advantage of using the FSI approach is the ability to investigate the effect of pressure waveforms in healthy aorta on the interface between blood flow and the artery wall. This method allows us to investigate the stresses and strains which are affecting the artery wall and blood flow. Consequently, Fluent results have a big impact on the mechanical arterial wall function for example deformation, which we is being investigated in this paper as shown in Figure 4. This Figure shows the WSS and Pressure contours for the Aorta at time 0.6 (s). The results is based on the FSI method which is based on solving the Arbitrary Lagrangian Eulerian (ALE) approach which allows the grid to track the material by adjusting grid generation and measuring the flux of the material during the adjustment. The ALE method is suitable for arteries that have large deformations resulting from the pulsatile flow because it allows the grid to track the movement of the material by adjusting the grid size and measures the flux of the material when the grid deforms excessively. Figure 5 shows the blood velocity contours at (0.4s) and (0.9s) and Figure 6 shows the Strain Rate for the Fluid-Structure Interaction (FSI) at (0.4s) and (0.9s).

Clinical observations indicate that the increase in the abdominal aorta systolic pressure in older patients with hypertension is the result of faster waves travelling down the artery and merging with stronger reflected waves [2-4]. This together with the increase in the ratio of lumen diameter, results in an increase in systolic pressure in the aorta. This obviously causes an increase in the wall stresses. In this study, the variation of WSS (representing the shear stress attacking the inner wall) and strain rate (representing the stretching of the artery wall due to the pulsatile blood flow) waveforms in terms of propagation time delay are presented in Figure 6 for the aorta.
Conclusion

This paper investigates the P, Q, WSS and Strain Rate contours for a healthy 3D aorta using the two-way, fluid-structure interaction method. These contours were achieved using ANSYS-Fluent 2020 R20, which allowed transient analysis performance to understand the interaction between the Aorta wall and the blood flow. The 3D-aorta geometry was constructed based on converting 2D-images to the artery wall as a solid and the blood body as a fluid. The resultant realistic aorta geometry is investigated using clinical data on pressure and blood flow waveforms to capture accurate physical parameters and behaviour. The simulated model shows promising results for demonstrating the behavior of the WSS, Strain Rate, P and Q profiles, so they can be investigated and deformed to simulate the case of cardiovascular disease.

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Conflict of Interest

The authors report no declarations of interest.
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