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A study on the computational fluid dynamics and fluid-structure interaction models for the hemodynamic of ascending thoracic aorta aneurysm

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Abstract: The ascending aorta is the first arterial segment, following the left heart. It is mechanically stressed by the blood flow that initiates the systemic circulation. Over time and due to changes in the microstructure of the vessel wall, the aorta may lose elasticity and dilate. If it increases by 50% over its original diameter in the ascending aorta, it is called an aneurysm. According to the European Society of Cardiology (ESC), the follow-up of this disease should be assisted by measuring the maximum diameter of the aneurysm. However, in practice, some patients have been observed to develop rupture or aortic dissection even before the criterion of critical diameter has been reached. In this sense, it is generally accepted that the geometric diameter criterion must be supported by other approaches that analyze multiple aspects to support the clinical medical diagnosis. Computational mechanics may be a suitable strategy. To start with, Computational Fluid Dynamics (CFD) allows a real-time evaluation of the hemodynamic behaviour and is able to estimate future life-threatening situations, even after surgery, by geometry modification. To improve the reliability of the results, however, the CFD approach can be enhanced by considering the possibility to couple Fluid-Structure Interaction (FSI), thus incorporating the elastic properties of the structure and their effects on the hemodynamics to be evaluated in the numerical simulations. The present work aims to develop a methodology to perform a numerical study of CFD with FSI for the hemodynamic of Ascending Thoracic Aorta Aneurysm (ATAA). The geometric model and boundary conditions were extracted from Computed Tomography scan (CT) scan and Magnetic Ressonance Imaging (MRI) imaging of the pilot patient. The open-source SimVascular software was used as a 3D segmentation tool, mesh generation, and numerical solver of Navier-Stokes equations and FSI. This methodology will improve the process of generating FSI numerical models to support the clinical outcome. In the future, this process will be automated to ensure consistent results.

1. Introduction

The aorta is the largest artery in the human body. It is split into segments based on its location and shape. The ascending aorta is the first section, located at the exit of the left ventricle. It is subjected to mechanical stress caused by the outflow of blood. With increasing age or due to multiple pathologies there may be a localized loss of elasticity which may yield to ATAA [1, 2]. According to the ESC, aneurysm status is characterized by an increase of at least 50% of the healthy diameter [3]. If this exceeds 55 mm then surgical intervention is recommended. However, there are documented cases of ruptures or dissections before reaching this limit, demonstrating that this methodology to assess the progression of this disease is not fully reliable [4, 5].

To overcome these limitations there are clinical and computational approaches that aim to analyze the state of the aneurysm by assessing the mechanical stress involved. Finite Element Method (FEM) has been gaining some headway in this area since, with the advancement of computational power, there is a possibility to model and analyze ATAA with some accuracy. These analyses have been focusing on stress distributions using Finite Element Analysis (FEA) and hemodynamics by means of CFD [6, 7].

FEA are used to analyze the dissipation and the effect of stresses and the pressure they have on the arterial wall in a macro and micro way [2, 8]. On the other hand, CFD aims to analyze the effect that hemodynamics has on the aortic wall by analyzing the surface tension, the pressure variation and possible sites at risk along with the domain [9, 10]. There is also the possibility of combining both methods through numerical models with FSI [6, 11, 12]. This process can analyze the hemodynamics and its relation with the aorta structure since it computes both in the same time step. However, it brings some disadvantages related to increased computational time, in patients where there is some sensitivity in the time.

There is a selection of directional open-source software for the hemodynamic and sequentially, the FSI. The software chosen is SimVascular, because of its ability to recreate ATAA geometries through DICOM images extracted from CT scan and to incorporate hyperelastic models and complex boundary conditions, such as the three-element Windkessel model [11, 13].

This paper aims to develop a method to develop a patient-specific ATAA numerical model aimed at solving FSI using SimVascular. Furthermore, it is intended to compare the variation of the detachment with the patient's MRI to understand and validate the expected behaviour. Initially, the process used to generate the solid dome will be discussed, followed by the analysis of the results. Finally, a discussion of the results will be carried out.

2. Methods

2.1 Geometric model in SimVascular

A numerical model with FSI requires a differentiation of the geometric model into Annulus Domain (AD) or Solid Domain and Lumen Domain (LD) also referred to as Fluid domain associated with the aortic structure and its lumen, respectively.

SimVascular was used to generate geometry associated with the LD from a three-dimensional segmentation tool with the SV MITKSegmentation, created by VMTK. This process extracts elements by defining a threshold ranging from 341 to 1706 on a patient's CT. However, this process generates some less relevant elements that need to be removed with mesh editing software, such as blender, to obtain the final LD geometric model.

To generate the AD, SimVascular recommends reusing the concentric mesh generated in the LD directed at the boundary layer. This methodology allows the development of a perfect interface [1] interaction since the meshes are coincident. However, this process introduces some disadvantages, since the AD mesh is generated from the LD mesh, there is no independency in the element size on both domains. In addition, the LD mesh does not include the concentric refined mesh dedicated to resolving the boundary layer, forcing a global AD element size increase.

A new method is developed to generate the AD and solving both limitations. In this new approach, the AD is obtained by extruding the outer LD layer (interface) with a $1.5\ \mathrm{mm}$ homogeneous thickness in a mesh editing program, such as Blender [14]. With this process the LD remains with the concentric mesh, reducing the total number of elements. Furthermore, there is an almost independent process in the element size of both domains.

The meshes were developed in SimVascular (Figure 3), given its hability to generate tetrahedral meshes using the TetGen library.

2.2 Boundary and initials conditions

The boundary conditions are intended to adapt the numerical model to the ATAA operating state. For this purpose, an inlet flow variation similar to a cardiac pulse was defined in Figure 1.

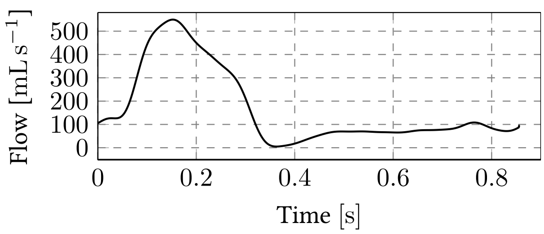


Figure 1. Flow variation as inlet boundary condition [9, 15].

To consider the effect on the pressure variation of the blood vessels downstream of the domain under analysis, a three-element Windkessel model, also called the RCR model, was defined at the output. This model predicts the pressure variation through the surface flow rate

[1] The interface is the inner surface of the ATAA that is shared by the meshes of the LD and the Aortic Dissection.

in a way that can be compared to an electrical arrangement with the resistor, R_p, in series with a combination of the resistor, R_d, and the capacitor, C, in parallel. Its arrangement can be considered as two resistances: Proximal resistance, R_p, which considers the blood vessels close to the domain and the distal resistance, R_d, which assumes the pressure of the capillary vessels. This model also includes a capacitor, C, which recreates compliance.

Figure 1 presents the values used to describe the behaviour of the three-element Windkessel model at each output. These values were tuned as the pressure in ATAA is between a normal arterial pressure [16].

Table 1. Values used to describe the three-element Windkessel model in CGS units.

	R_p	C	R_d
Thoracic aorta	39	4.82×10^{-4}	1016
Brachiocephalic aorta	139	8.74×10^{-5}	3637
Left common carotid artery	520	7.70×10^{-5}	13498
Left subclavian artery	420	9.34×10^{-5}	10969

FSI require considerable computational effort as each time instant increases. In addition, in this type of numerical model, the first cardiac cycles under analysis do not consider the impact of the previous cycle. In this sense, it would be interesting from an efficiency point of view to incorporate an initial condition that contains the impact of the previous cardiac cycle.

Prestress is a process to incorporate more complex initial conditions in aFSI model. The method chosen to develop this initial condition involves developing a CFD model in the LD to calculate the pressure variation and flow in these conditions. Furthermore, the calculated pressure is used to perform a FEM analysis to extract the stress and displacement variation in the AD [11].

2.3 Constitutive models

The aortic wall has a hyperelastic behaviour, meaning that the state of tension at each moment depends on the deformation rate over time. In this sense, the hyperelastic Neo-Hookean model was used to define the aorta's elastic behaviour since its largely used for this porpose [11].

The LD mathematical model is considered a Newtonian, incompressible, homogeneous flow with a laminar flow [11]. These simplifications of the non-Newtonian characteristics of blood are well accepted for the larger blood vessels since the high strain rate in the blood does not affect the viscosity [17, 18].

In this article, the Arbitrary Lagrangian-Eulerian (ALE) description will be used to describe the interaction between the mesh and the material. This description is largely used in FSI models since the mesh motion is calculated based on the material velocity.

There are some constants needed to define both solid and fluid behaviour. In the LD, density, ρ, of $1.060\ \mathrm{g} \cdot \mathrm{cm}^{-3}$ and a viscosity, μ, of $0.04\ \mathrm{P}$ were considered. On the other hand the AD was considered to have a density, ρ_s, of $1.120\ \mathrm{g} \cdot \mathrm{cm}^{-3}$, Young's modulus, E, of $10\ \mathrm{Mdyn} \cdot \mathrm{cm}^{-2}$ and Poisson ratio, ν, of 0.49 [11].

2.4 Mesh sensitivity analysis

To improve the efficiency and reduce the computational time of the numerical model, it is wise to perform a mesh sensitivity analysis. The methodology developed for this step goes through two distinct phases associated with each domain. The first, associated with numerical models with the nomenclature “E”, consists in linearly increasing the number of elements in the AD domain while keeping the LD domain constant until a convergence in the results is obtained. The same procedure was performed for the LD, with the numerical model “F”, obtaining an ideal option for each size of elements of each domain.

Figure 2 shows the flow velicty “Point A” (3) for time instance 245 ms in relation of the variable mesh elements in the “E” and “F” analysis, respectively. Similarly, it shows the computational time variation for both numerical models.

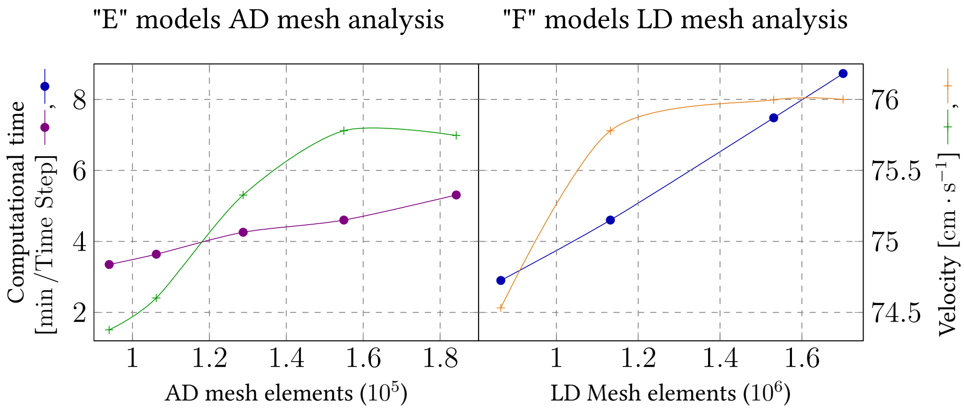


Figure 2. Velocity variation of each numerical model for the time frame 245 ms in parallel with the computational time required for both phases in the Mesh sensitivity analysis.

Both models “E” and “F” show an almost linear increase in computational time as the elements number increases. In numerical models “E” there is a convergence for the AD element number greater than 1.5×10^5 that is equivalent to mesh “E4”. In this same process, we verified that the numerical models “F” present a convergence for values bigger than 1.2×10^6. This figure facilitates the mesh selection and allows the conclusion that, for this model, the ideal mesh has an element size of 1.35 mm for the LD mesh and 1.1 mm for the AD mesh. Figure 3 demonstrates the final mesh obtained for both domains.

3. Results

The proposed FSI model implemented in SimVascular can describe the aortic hemodynamics with the impact of the aorta hyperelastic structure. Figure 4 demonstrates the variation of velocity modulus, displacement magnitude and Wall Shear Stress (WSS) in the ATAA for the 150 and 300 ms time instants. These time intervals were selected because they are extremes of the flow applied to the inlet (Figure 1).

To better understand the numerical model behaviour, a comparison was performed between patient MRI data and computational results. Figure 5(a) shows the comparison between the area

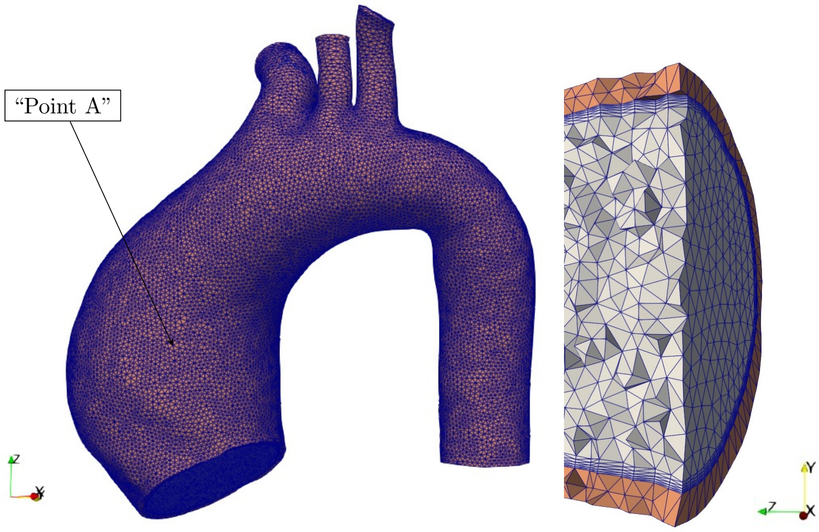


Figure 3. Final mesh with solid and lumen domain and "Point A" location.

variation obtained from the patient's MRI slice (Figure 5(b)) and the area variation obtained by computational methods on a similar slice. Furthermore, the pressure variation in the numerical model is shown in Figure 5(a) to verify the effectiveness of three-element Winkessel model boundary conditions. As desired, the pressure varies between 123.7 mmHg and 72.1 mmHg.

4. Discussion

The present numerical model uses SimVascular to develop the preprocessing and numerical calculations and demonstrates the ability to integrate the FSI into the numerical model. The new process for generating the geometric model can optimize the number of elements. Furthermore, it enables the incorporation of boundary layer-directed concentric refined mesh. These advances are a relevant step towards computational efficiency in SimVascular, as the geometric model is optimizable.

The prestress process adds to preprocessing complexity but it reduces the computational time for numerical stability. Stress and displacement inserted as initial condition demonstrated that the first cardiac cycle in this analysis already has a fully defined flow. By comparison, the same numerical model but without prestess may need at least three cardiac cycles to replace the same results [11].

The mesh sensitivity analysis concluded that there is an almost linear relationship between increasing the number of elements and the computational time involved. By the convergence of the velocity at "Point A", it is concluded that the optimal mesh for this numerical model has an element size of 1.35 mm for the LD mesh and 1.1 mm for the AD mesh.

From the comparison with the patient's MRI, is concluded that the amplitude area variation is higher than the obtained by the numerical model. Since the pressure variation is similar to the patient's arterial pressure, the displacement error can be explained by the simplifications made to the AD. The AD thickness homogeneity and the hyperelastic behaviour simplification

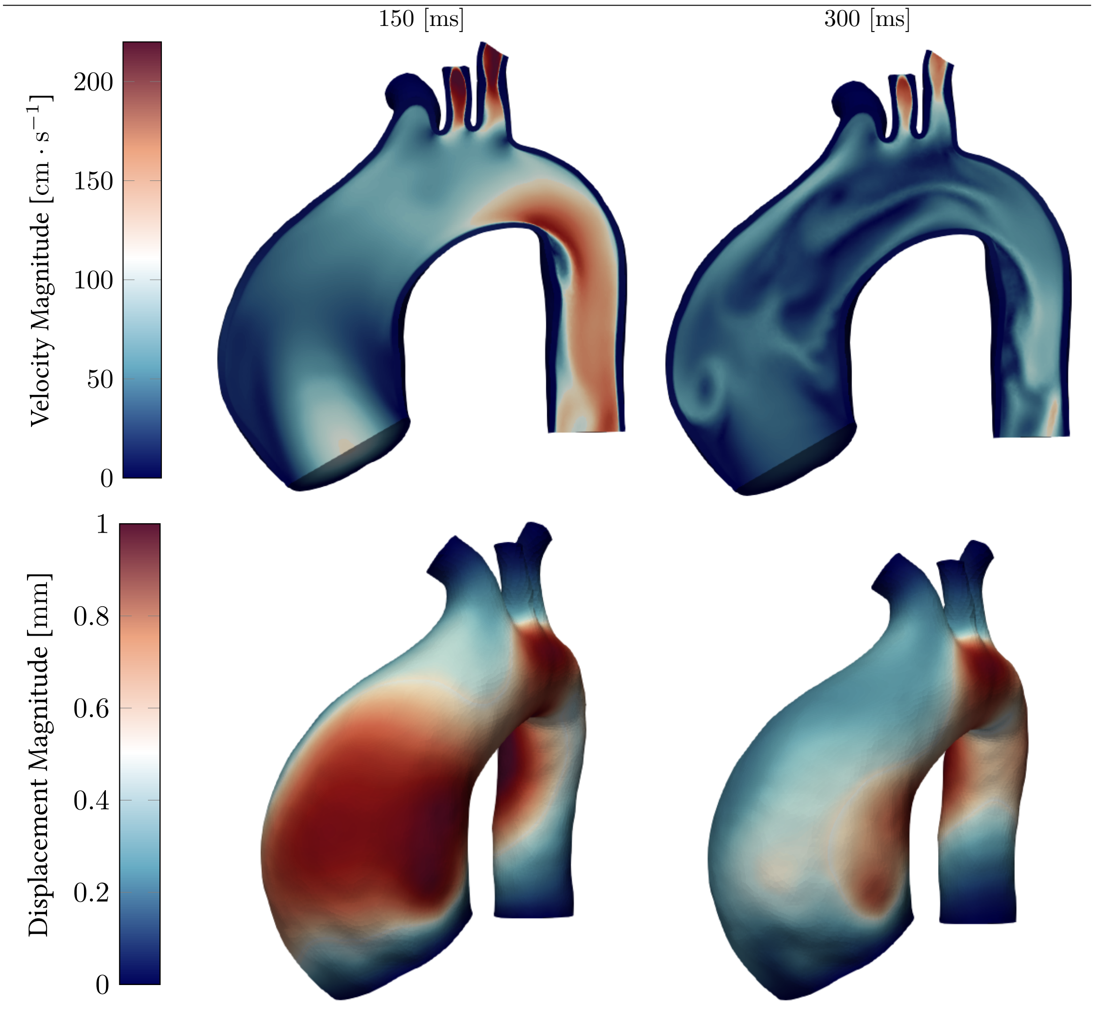
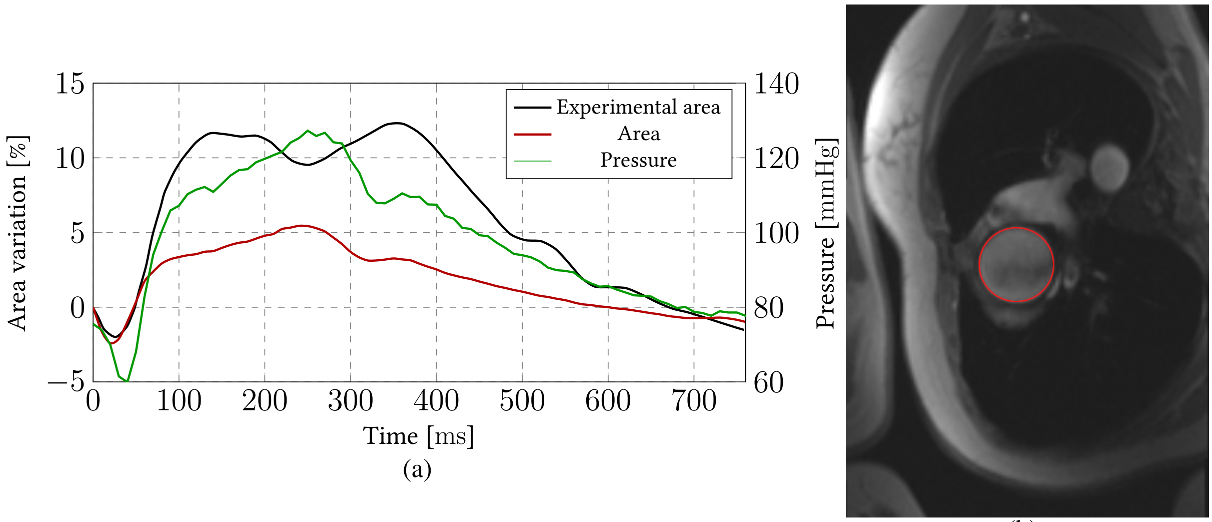


Figure 4. Velocity and displacement magnitude distribution in a vertical slice for prestress analysis at 150 and 300 ms.

in the model can be correlated with the low amplitude. Furthermore, the inlet blood flow spatial vector is an approximation of the patient-specific *in-vivo* conditions and this may affect the ATAA displacement deviations, since it is known that hemodynamic will impact the stresses developed on the aortic wall [5, 19].

5. Conclusion

In this work, a FSI model of the ATAA was developed in SimVascular. A new methodology to generate the geometric domain and mesh was proposed. This method allowed including two improvements compared to standard procedures. The first was the possibility of including a refined concentric mesh in the LD to solve the boundary layer. The second improvement is the almost independent mesh generation on both fluid and solid domains. Both implementations resulted in the possibility of optimizing the geometric model with efficiency in mind. Furthermore, the numerical model demonstrated that the prestress yielded a numerical flow



(a)

(b)

Figure 5. Comparison between experimental and computation area and pressure variation. (**a**) Pressure variation and percentual comparason between the numerical area and experimental values measured on patient's MRI. (**b**) Patient-specific MRI horizontal cross-section to measure the area variation (red).

convergence within the first cardiac cycle, improving the efficiency. Otherwise, two to three cardiac cycles would be necessary for the same result [11, 20].

In the future, *ex-vivo* experimental analysis can be carried out to assist the definition of the aorta complex hyperelastic behaviour. Furthermore, both the heart movement and the outer pressure can be incorporated into ATAA numerical model to further improve its clinical relevance. To end with, a hemodynamic validation of the numerical model can be performed by a direct comparison using the patient 4D-MRI.

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