Modelling the hemodynamics in a realistic cerebral aneurysm with clot through transient FSI simulations

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Abstract. Modelling the hemodynamics of a patient-specific cerebral aneurysm with a clot in the most realistic conditions possible is still a challenge. The present work focus on the validation of the numerical procedure, implemented in ANSYS® software, for further accurate results. The 3D geometry used for simulations is a real patient cerebral artery with an aneurysm. The aneurysm is an Internal Carotid Artery (ICA) and is a saccular type. A Womersley velocity profile at the inlet boundary condition was used and blood was considered as a non-Newtonian fluid. These conditions were implemented in User-Defined Functions (UDFs) of ANSYS®. Moreover, an isotropic linear elastic model was used for the arterial walls and a hyperelastic Ogden model for the clot. The Fluid-Structure Interaction (FSI) model was also implemented in ANSYS® in order to really mimic the deformability of the artery during the pulsatile blood flow. In this patient case, the maximum WSS was 38 Pa and the TAWSS on the sac surface was 0.1 Pa. These values are within the expected since the results are similar to previous research. Thus, this numerical procedure can be considered valid for further hemodynamic studies of the aneurysm clot migration process in a realistic cerebral aneurysm.

Keywords: hemodynamics, cerebral aneurysm, FSI simulations, UDF implementation.

1 Introduction

An aneurysm is an outward ballooning in a blood vessel and occur predominantly in arteries at several locations in the body [1]. The aneurysms have different classifications according to the size, shape and location and due to the absence of symptoms, the rupture of a cerebral aneurysms has a high mortality rate [2]. Endovascular coiling has being one of the most common treatment for cerebral aneurysms [3], in this procedure a metal wire is inserted into the aneurysm sac enhancing the blood clot formation, and improving the occlusion of the aneurysm. The clot formation mechanisms depends on the location of the aneurism, having a complex interaction between blood flow and coils. However, during endovascular embolization of intracranial aneurysms, the coil migration is a serious complication. The majority of coil migration occurs during the procedure [4].

CFD and FSI researches have been widely used for research on aneurysm. CFD modeling has great potential for studying the endovascular treatment with coils, different previous researches have demonstrated the CFD capacity to simulate with good accuracy the hemodynamics in aneurysms. Nevertheless, CFD methods are often limited due to the inability to reproduce changes in the vascular walls, specifically important in aneurysm hemodynamic simulation, these problems might be partially solved including FSI models.

Despite of a high number of studies in this field, modelling the hemodynamics of a patient-specific aneurysm with a clot in the most realistic conditions possible is still a challenge [5]. As far as we know, there are no FSI hemodynamic analysis of forces suffered by clots inside an aneurysm. The present work focus on the validation of the numerical procedure, implemented in ANSYS® software, to obtain further accurate results.

2 Methodology

2.1 3D Geometric Model

The 3D geometric model used for the numerical simulations is a real patient cerebral artery with an aneurysm, selected from the open-source Aneurisk online database [6]. Representation can be observed in Fig. 1. The aneurysm is an Internal Carotid Artery (ICA) and is a saccular type, which did not have rupture problems. The patient-specific case is a 42 years-old woman. The geometry of the clot is assumed to be spherical. The reconstruction of the 3D model was carried out using ANSYS SpaceClaim 2020 R1® software, a design tool oriented to repair, simplify and treat geometries imported in STL format.

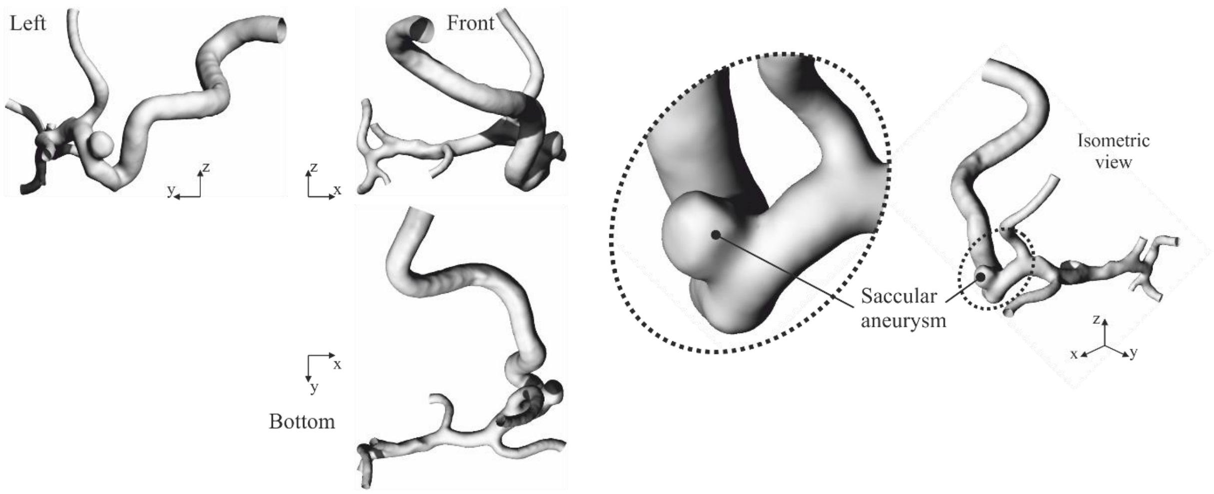

Figure 1. 3D Geometry of a patient cerebral artery with aneurysm [6]

2.2 3D Mesh

In order to generate the mesh in this research, the ANSYS® meshing pre-processing module was used (Fig. 2a). For the mesh convergence, 12 different sizes of the surface mesh were evaluated and 13 numbers of prismatic cell layers in the near-wall region were considered. The mesh convergence criterion was 10^{-4} in the wall shear residual and the optimal meshing parameters were achieved (Fig. 2b). Moreover, the maximum Skewness value is lower than 0.79, which highlights the quality of the mesh. The total number of computational cells considering fluid and solid domains is around 2.5 million.

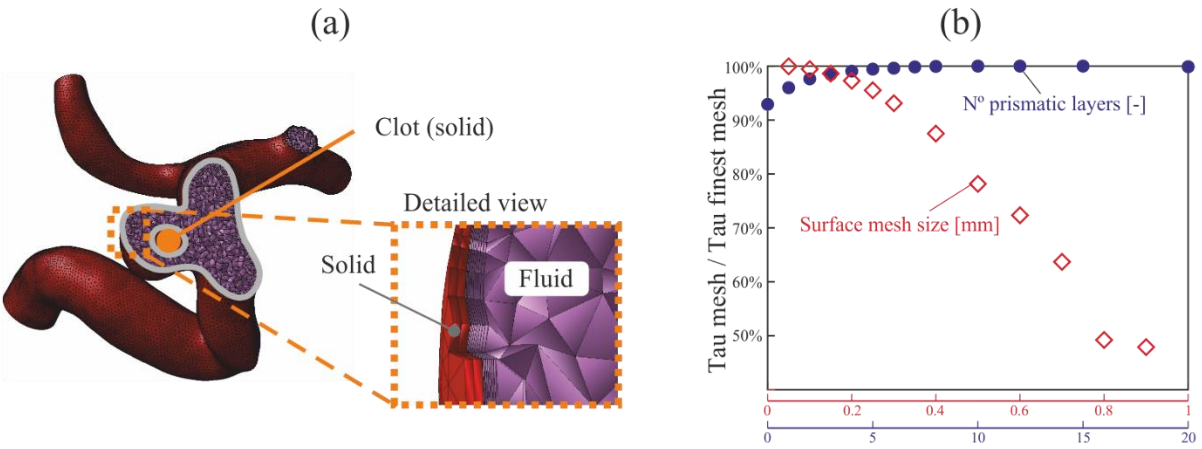

Figure 2. (a) Mesh details, (b) Mesh convergence study

2.3 Properties of blood, vessel walls and clot

There are two phases in the present model: a fluid region corresponding to the blood and a solid region, the artery walls and the clot, which have different properties.

Blood was assumed to be a homogeneous and incompressible fluid with constant density equal to 1050 kg/m3. In the present work, blood was modelled following the non-Newtonian Carreau model [7]. This model was implemented through a User-Defined Function (UDF) of ANSYS® software.

$$\mu = \mu_\infty + (\mu_0 - \mu_\infty)[1 + (\lambda\dot{\gamma})^2]^{\frac{n-1}{2}} \tag{1}$$

The viscosity, μ, varies with the shear rate, $\dot{\gamma}$, where the time constant λ is equal to 3.313, *n* equal to 0.3568, zero shear viscosity, μ_0 , equal to 0.056 Pa and infinite shear viscosity, μ_∞, equal to 0.00345 Pa.

The model used for the arterial wall is an isotropic linear elastic model governed by Hooke's Law, with 1.8 MPa for the Young's modulus and 0.45 for the elastic isotropic Poisson's ratio [8]. For the clot, the Ogden model replicate the stress-strain behavior of clots with acceptable accuracy [9]. The strain energy density function for Ogden model (symbol) is given by the following equation:

$$\psi = \sum_{i=1}^{N} \frac{2\mu_i}{\alpha_i^{\,2}}\left(\bar{\lambda}_1^{\,\alpha_i} + \bar{\lambda}_2^{\,\alpha_i} + \bar{\lambda}_3^{\,\alpha_i} - 3\right) + \sum_{i=1}^{N} \frac{1}{D_i}(J-1))^{2i} \tag{2}$$

where μ_i, α_i, D_i are material parameters, $\bar{\lambda_i}$ are the isochoric principal stretches, $J = \lambda_1\lambda_2\lambda_3$ is the Jacobian, and all the parameters are defined in Table 1:

Table 1. Parameters of the Ogden model (N=3) [9]

μ_1 (kPa)	α_1	μ_2 (kPa)	α_2	μ_3 (kPa)	α_3	D_1 (1/kPa)	D_2 (1/kPa)	D_3 (1/kPa)
0.00812	6.41	0.00248	-5.4	0.00009	5.16	207.43	193.37	473.03

The arterial wall and clot densities were considered the same, equal to 1050 kg/m^3.

2.4 Numerical settings

The flow inside the artery has a pulsatile behavior, due to the cardiac cycle. Thus, transient state was considered for simulations.

At the inlet of the artery, a fully developed flow was applied using a Womersley velocity profile [10]:

$$w(r,t) = \frac{2B_0}{\pi R^2}\left[1 - \left(\frac{r}{R}\right)^2\right] + Re\left[\sum_{n=1}^{N} \frac{B_n}{\pi R^2}\left[\frac{1 - \frac{J_0\left(\alpha\sqrt{n}\frac{r}{R}i^{\frac{3}{2}}\right)}{J_0\left(\alpha\sqrt{n}i^{\frac{3}{2}}\right)}}{1 - \frac{2J_1\left(\alpha\sqrt{n}i^{\frac{3}{2}}\right)}{\alpha\sqrt{n}i^{\frac{3}{2}}J_0\left(\alpha\sqrt{n}i^{\frac{3}{2}}\right)}}\right] e^{i2\pi n\frac{t}{T}}\right] \tag{3}$$

Where w is the normal velocity, r is the radius, t is the instant time, B_n is the complex Fourier coefficients, R is the equivalent radius of the circumference with the same input area, J_0 is the Bessel function of the first type of zero order, J_1 is the Bessel function of the first type and first order, i is the imaginary number, , T is the cardiac cycle period, α is the Womersley number ($\alpha = R\sqrt{(2\pi/vT)}$ which indicates that for values between 0 and 1 the flow can approach Poiseulle's law, while for values greater than 10 the flow approaches to a flat profile, and v is the kinetic viscosity.

The Womersley velocity profile was implemented in a UDF of ANSYS® software. 20 Fourier coefficients were obtained using Matlab® software and sufficient to mimic the cardiac cycle curve achieved by Ford et al. [11].

The zero gauge-pressure was defined as the outlet boundary condition following Souza et el. [12] and the walls were defined as stationary with no slip-boundary condition.

FSI simulations were run in ANSYS® software. The interactions between fluid (blood) and solid (arterial wall and clot) domains were taken into account in the coupling mode (Fig. 3). For this, the displacement of the fluid and solid interfaces must be compatible and the transition at the interface must be equilibrium. A dynamic mesh was used on the arterial wall, defining a diffusion factor of 1.5. It allows the creation of a changing mesh without loss of quality depending on the results of the deformations obtained.

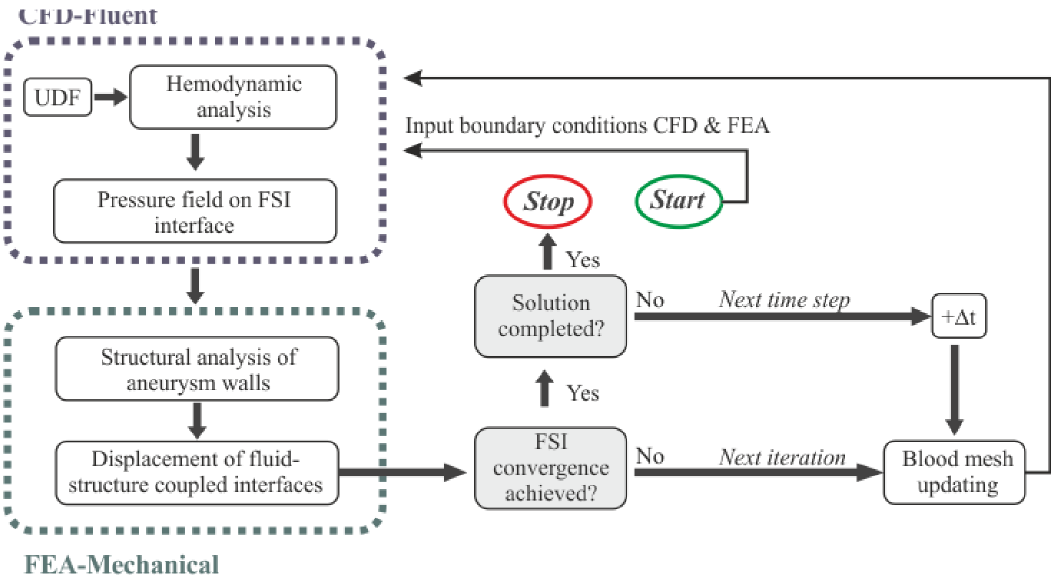

Figure 3. Coupling mode of the FSI process

The SIMPLEC algorithm was used with square cell based for the gradient, second order spatial discretization settings for pressure and second order upwind for momentum. The simulations were running according to the convergence criterion of ANSYS® Fluent software [13]. The scale residual of all variables must be below 10^{-4}.

3 Results and Discussion

The hemodynamics before coil procedure was analyzed in order to validate the numerical procedure with the literature. Therefore, the time average wall shear stress (TAWSS), defined as the wall shear stress magnitude normalized by temporal-spatial along the time of the cardiac cycle, and the total deformation and the equivalent stress in the walls of the artery and in the aneurysm were evaluated (Fig. 4).

The TAWSS distribution is significantly lower in the aneurysm than the rest of the artery (Fig.4a), causing a risk of blood stagnation and therefore clot formation. The maximum wall shear stress obtained was 38 Pa and the TAWSS on the sac surface was 0.1 Pa, taking into account the volume of the aneurysm. The distribution and values obtained are within the expected and in concordance with previous researches of Sejkorová et al. [14] and Aranda and Valencia [15].

The total deformation at the instant peak t=0.1s is very small, as can be observed in Fig. 4b. Thus, the generated forces in the arterial walls caused by the blood flow in the modelled conditions are relatively low, and the effects of the artery deformation are very limited in blood flow analysis. The contours of equivalent stress (Fig. 4c) are also at instant peak t=0.1s. It can be observed that the maximum stress is located in the region of the neck of the aneurysm, corresponding with the thinnest thickness of the arterial wall.

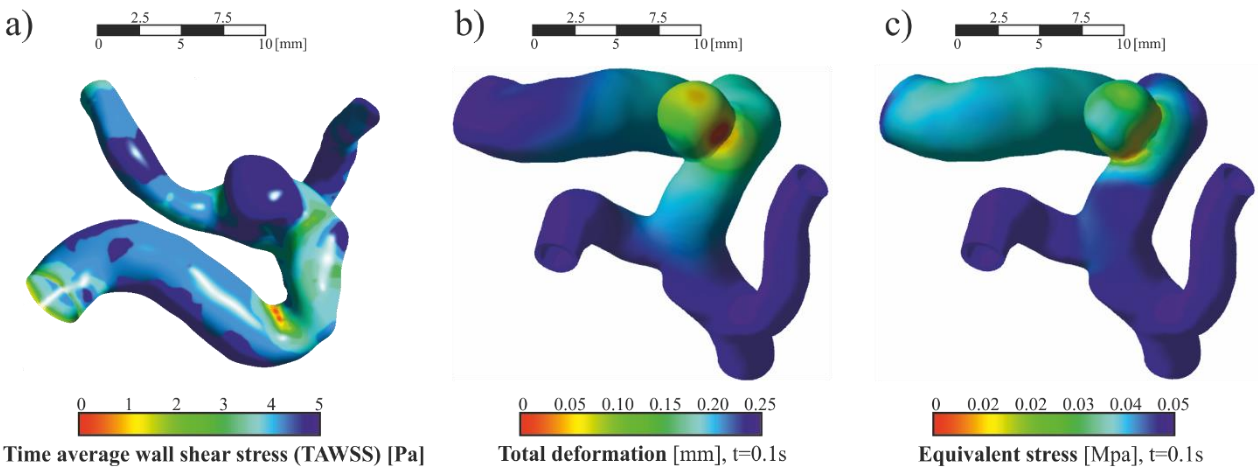

Figure 4. (a) Time Average Wall Shear Stress (TAWSS), (b) Total Deformation Contours at t=0.1s, (c) Equivalent Stress Contours at t=0.1s.

4 Conclusions

The focus of the present work is the validation of the numerical procedure to obtain the hemodynamics in a patient-specific cerebral aneurysm with a clot. Thus, the most real conditions were used and UDFs were implemented in ANSYS® software to mimic the deformability of this type of arteries during the pulsatile flow. A Womersley velocity profile was used as the inlet boundary condition, blood was considered as a non-Newtonian fluid and FSI was taken into account. After running the numerical simulations, the results obtained were promising for validation with the literature. For this patient, the maximum WSS was 38 Pa and the TAWSS on the sac surface was 0.1 Pa. These values are similar than previous researches of Sejkorová et al. [14] and Aranda and Valencia [15] and, thus, the numerical procedure is considered valid for further hemodynamic studies of the aneurysm clot migration process.

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