Numerical analysis of a brain artery in ANSYS

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Abstract

In the last decade the brain arteries were a source of lot of research, many of them try to study from an engineering perspective very difficult diseases such as aneurysm. They usually choose either the blood or the artery to study, but few study both without some bigger considerations like considering blood Newtonain or the structure elastic. This study tries to give a different light to those by trying to study the nature of a brain artery. For that the behavior of the brain arteries were treated as a Fluid-Structure Interaction problem and for that was used a 2-way coupling method between a non-Newtonian fluid and an isotropic hyperelastic structure. With this approach some conclusion about the nature of brain arteries, and arteries in general maybe, could be made. Those conclusions could lead to some new reflections about failure mode of arteries and the function of each layer of an artery.

Keywords*: FSI, 2-ways coupling, hyperelasticity, Non-Newtonian Fluid*

1. Introduction

The cerebral arteries are a very specific kind of artery, an artery can be divided into 3 tunicas, an intima (an inner one), a media and an adventitia (an outer one). Those three tunicas are usually very well divided by an elastin lamina. The cerebral arteries vary in that aspect, they don't have an elastin lamina dividing the adventitia and the media tunica, (Humphrey, 2014) therefore those two are virtually the same tunica (Holzapfel & Gasser, 2000). But it should be notice that there are studies (Wagner,2014) (Wicker,2007) presenting those two separated and the reason for that is the difference in the collagen fiber distribution.

The main structural component at an artery is the collagen fiber, those fibers give arteries their mechanical behavior as an anisotropic hyperelastic material (Hayashi, 2014). As anisotropic material arteries have a preferential tension direction, but to study the reason behind those direction one should look at the forces that an artery is undergoing and for that a study of Fluid-Structure Interaction (FSI) is a must.

To start the research, it was decided to look at the forces under FSI in a healthy individual, for that some assumptions were made, and it was decided to make a simulation using all the references researched. This simulation was between a non-Newtonian fluid and an isotropic hyperelastic material. By this choice it was expected to look at all the forces that a non-Newtonian fluid such as the blood could make at an hyperelastic structure, and with that study the forces that an artery undergoes on a healthy condition.

2. Methodology

2.1. Geometry

The geometry was provided by Damin, Samuel. He used MRI images to generate a 3D CAD image. Those show the real geometry of the blood domain at the circular system, but for FSI simulation it was needed to rebuild this geometry

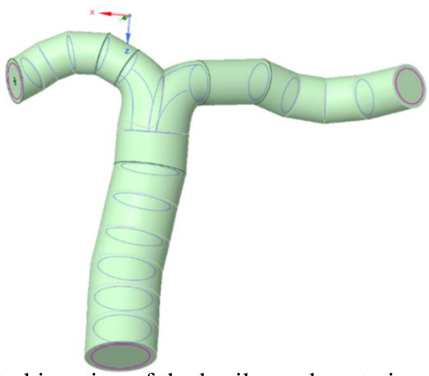


Fig. 1 Reconstructed junction of the basilar and posterior cerebral artery

because the geometry shown is the final geometry that the FSI simulation intends to reach, that is this geometry is the result of the FSI between blood and artery and therefore rebuilding the geometry was a necessary step for the simulation. The resulting geometry is shown on figure 1.

2.2. Fluid

The FSI study was done considering the blood as a non-Newtonian fluid. As a matter of effect, the blood is indeed a non-Newtonian fluid (Barnes *et al*, 1989), but depending on the regime it is put into it can behavior like a Newtonian fluid, especially when the shear rate is up to 100 s-1 (Oliveira *et al*, 2021). Since the goal for this research is to deeper the study about brain arteries it was consider treating blood as a non-Newtonian fluid necessary since this consideration was made thinking about the comparison with future works which could need to consider it Non Newtonian.

It was decided to use the Carreu-Yatsuda model since this model is very close to the power law model and cause it was usually used this model or the pure power law model to simulate the blood,(Barnes *et al*, 1989) (Wang X. e Li X,2011) (Karsheva *et al*, 2009) (Bernsdolf e Wang, 2009) (Shamloo *et al*, 2017) (Ahmed *et al*, 2007) (Bertelli *et al*, 2008) and just few articles that decided to consider the blood non-Newtonian treat the blood differently (Sodré *et al*, 2018).

It was assumed a laminar flow, because it also seemed to be well established since all the references searched assumed a laminar flow.

Therefore, for a laminar flow it follows the continuity equation and the moments equations:

$$\nabla \bar{u} = 0 \quad (1)$$

$$\rho\left(\frac{\partial u_i}{\partial t} + u_j \frac{\partial u_i}{\partial x_i}\right) = \frac{\partial p}{\partial x_i} + \frac{\partial \tau_{ij}}{\partial x_i} + f_c \quad (2)$$

Were

$$\tau = \mu(S)S \quad (3)$$

S is the deformation rate tensor, this deformation rate tensor also defines the shear rate:

$$\gamma = \sqrt{0{,}5 * S{:}S} \quad (4)$$

The viscosity was defined as a function of the shear rate γ using the Carreu-Yatsude model, so for the viscosity μ:

$$\frac{\mu - \mu_\infty}{\mu_0 - \mu} = \frac{1}{(1+(K_1\dot{\gamma})^2)^{m_1/2}} \quad (5)$$

The constants μ_∞, μ_0, K_1 and m_1 are material constants of the blood and were used the following values from Barnes (1989).

For the Boundaries conditions it was defined an outlet of 100mmHg which is the average pressure of the blood circulation, and it was defined an inlet condition with 0 Pa of gauge pressure and a flow rate defined by seventeen constants of a Fourier transform found by Linninger, Andreas A. *et al* (2009). Those constants resulted the flow rate shown in fig 2.

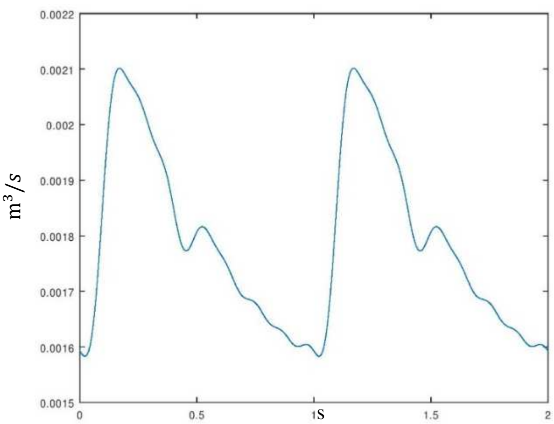


Fig. 2 – Flow rate

2.3. Structure

An artery is an anisotropic and incompressible material due to the collagen fibers but since one of the goals of this study was the use of ANSYS workbench material manager, the version used (2021 R1) didn´t supported hyperelastic anisotropic materials, then it was decided to make the hypothesis of an isotropic material that would replicate to the other three directions the behavior of one of directions of the real artery. The chosen direction was the tangential one since it is the preferential direction of the collagen fiber from tunica media (Wicker, 2007), the thickest tunica (Wagner, 2014).

Then it was decided to use constants of Holzapfel model (Holzapfel,G.A *et al*, 2000) and replicate the data from the biaxial test used in Wagner (2014), there Wagner calculated the stress at both the tangential and axial directions and build a stress-strain curve between stress and strain at the tangential direction. This curve was recalculated and used as input to the isotropic model as shown at figure 3.

Then the isotropic model chosen was the 3rd order Odgen model (Odgen, R.W. 1972), this model was chosen because it defines the material using just the deformation at the principal direction and due to this it doesn't need shear test to calculate its constants, which was critical cause it is most common to find biaxial and uniaxial test data for arteries and not shear test.

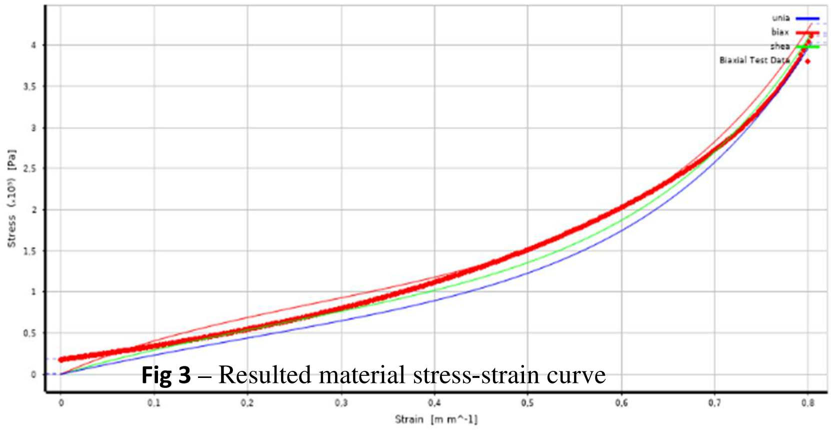


Fig 3 – Resulted material stress-strain curve

After the stress-strain curve was used as input for the hyperelastic curve fit of ANSYS workbench 2021R1 for the 3rd order Odgen model some alterations had to be done in the constants μ_1 and α_1 to make the material useable. Then the resulted material had the stress-strain curve shown in Fig. 3.

The red bold line is the Stress-Strain curve calculated using the constants calculated by Wagner (2014) and the other three lines, blue, red, and green are from the customed material at the uniaxial, biaxial and shear theorical tests. The values of constants used to define this material were at table 2.

Table 2 – Material Constants from the 3rd order Odgen model

μ_1	90000
$\boldsymbol{\alpha_1}$	1,8
μ_2	264,8
$\boldsymbol{\alpha_2}$	11,673
μ_3	264,857
$\boldsymbol{\alpha_3}$	11,864
$\boldsymbol{d_1}$	1,2E-08
$\boldsymbol{d_2}$	1,2E-08
$\boldsymbol{d_3}$	1,2E-08

The only downside of this model for this application is the fact that it isn´t an isochoric model, but it is divided between an isochoric and a volumetric part. This downside was contorted by finding a big value for the bulk modulus, enough to make the model nearly isochoric. For that we based our choice of value on the first bulk modulus which is:

$$K = \frac{2}{d_1} \tag{6}$$

To find a good value of bulk modulus it was used as a comparation base an article that studied porcine Linea alba which is also an incompressible biological material and Cooney *et al* (2015), there it concluded that for their material which is mix of lots different kinds of biological material the bulk modulus should be higher than 20 Mpa (Cooney *et al*, 2015), therefore we assumed that for a human artery an error for a bigger value shouldn´t affect too much, but just made the material more incompressible. Therefore, we used the same values for Young and Poisson modulus than the one used by Ahmed *et al* (2007), which gives a bulk modulus of 166,67 MPa way higher than the 20 Mpa. Then with the calculated value of the first bulk modulus the constant d_1 was obtained, it was considered that the other two volume constants had the same value. It should be noted that this was just done because we were looking for a reasonable value to make the volumetric change very low.

The constrain conditions for the structure were defined to make the lateral area of the basilar and of both posterior cerebral arteries and its contours fixed.

3. Results

3.1. Fluid

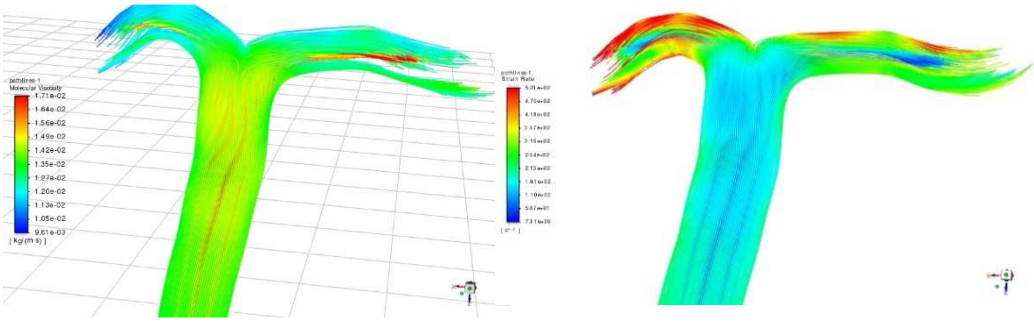


Fig. 4– Viscosity (left) and Strain rate (right)

The viscosity can vary a lot, it varies from 9,81e-3 kg/(ms) to 1,71e-2 kg/(ms), but it has an average close to 1,35e-2 kg/(ms). It´s higher values are close to the center of the transversal section and therefore very close to the pattern of a viscous laminar flow inside a pipe. The Strain rate mean was 358,6421 s^{-1} which is a higher than the value stipulated of 100 s^{-1} in which the blood could be simulated as Newtonian. To confirm the laminar regime, it was calculated the Reynolds Number for the blood flow, and it was found a value of 55,79 which was found by using the mean viscosity, the mean velocity magnitude at the inflow 0,187 m/s, the final basilar diameter 4,11mm as the characteristic linear dimension and a density of 1050 kg/m^3. Which is way less than the Reynolds number calculated by other authors cited previously, but still, provides the same conclusion that the flow regime is laminar.

The Wall Shear Stress had a maximum value of 20,47 Pa and had the distributing as shown on figure 8 and this distribution could be linked with the results found in the structure.

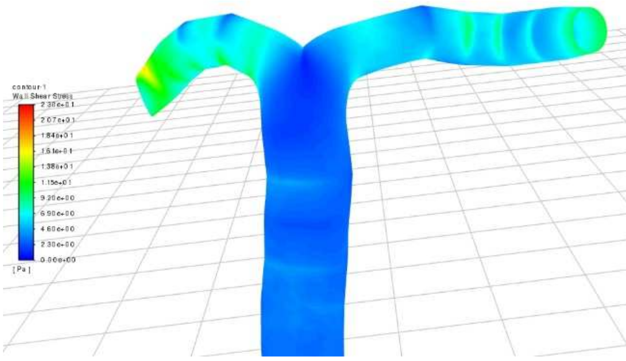


Fig. 7 – Wall Shear Stress

3.2. Structure

In Figure 8 (b) is shown the elastic strain intensity, it had an average of 38,075% of deformation and it should be said that the regions where we found the higher values are very close to the regions reported by Sheikh *et al* (2020) as regions with high wall shear stress and therefore subjectable to start aneurysm growth.

(a) (b)

Fig. 8 – Equivalent Stress and Elastic Strain Intensity

It also should be noticed that the points with higher elastic strain are all regions close to where the wall shear stress has a bigger gradient, which is at right posterior artery (at the figure 8 is the right branch and at the figure 7 the left one) close to the bifurcation of the basilar. This region is also the region with the maximum equivalent Stress shown in figure 8 (a) which is another indicator that this region is more prompt to occur an aneurysm. It also should be noticed that despite the results from fluid and structure point to the same region, the patient from which the geometry was based had an aneurysm on the other posterior cerebral artery, this occurrence couldn't be explained with this study.

At Figure 9 is shown the vector principal strain, there the maximum principal direction is at tangential direction, the middle at the axial and minimum at the normal directions. The minimum at the normal direction is expected since the artery is a thin-walled tube, but one should pay attention to the other two, the maximum principal is at the tangential direction which is the same direction of the tunica media collagen fibers and where there is a very concentrated collagen fiber distribution and is the larger tunica. The middle principal is the axial direction the same preferred direction of the collagen fiber of the tunica adventitia, and since it is the middle it is also the direction of the maximum shear stress as it is known the shear stress is higher the farther it is from the center of a thin-walled tube and the adventitia is not only the most external layer it is also the one with the largest collagen fiber distribution. Wagner (2014) found a mean angle of 39,5°for the theorical Holzapfel layer model and Wicker (2007) found a mean angle of layer model 43,65°, Holzapfel presents an angle of 40° for modeling his adventitia as well (Holzapfel, 2014), very close to the ideal 45° which would be the best position to withstand shear stress.

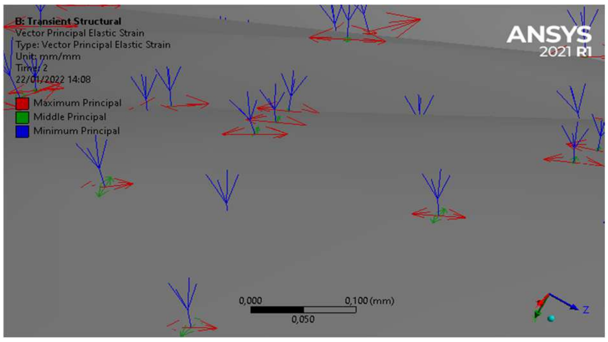
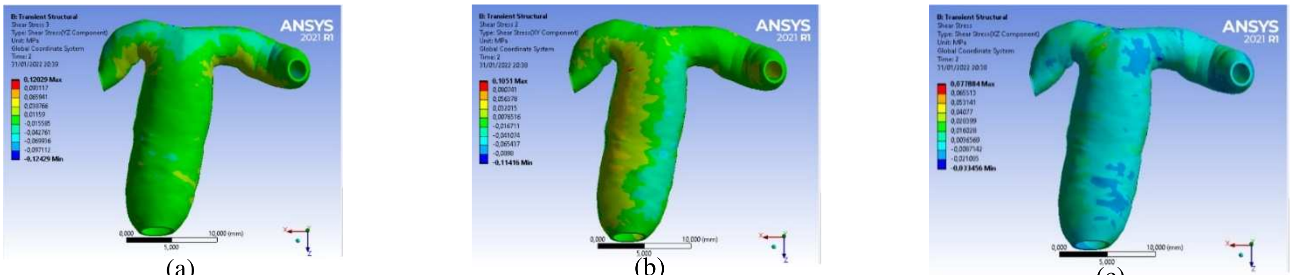


Fig. 9 – vector principal elastic strain

To show that the shear stress occurs with higher values at the middle principal elastic strain direction it was calculated the shear stress at x,y and z directions as shown at figure 10. At figure 10 (a) is shown the shear stress at z direction, at (b) the shear stress at x direction and at (c) the shear stress at y direction. As it can be seen at figure 10 the maximum shear stress is found at (b) which is the axial direction of the left and right posterior cerebral artery and (a) has the second lagger shear stress due to the basilar axial direction and both are way bigger than the values found in (c).The explanation for this shear stress is the geometry, due to the FSI our model had its shape changed from an Y like design to a T like design. This change made each of the arteries subject to forces at the axial and tangential direction, which generated the shear stress presented at figures 10.



(a) (b) (c)

Fig 10 – Shear Stress

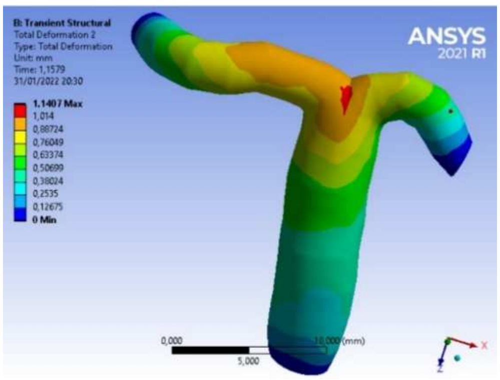


Fig 11 – Total deformation

To confirm this change of the geometry the total deformation is shown at Fig 11, and it should be noted that the maximum deformation occurs at the bifurcation and had a value of 1,1407 mm. This value is high enough to change the geometry and to realize that one should remember that the initial diameter of the basilar artery, the larger artery studied here, is 3,24 mm, this way one can see that the deformations are very large which is typical of a hyperelastic material.

4.Conclusion

The most important component in cerebral arteries is the collagen fiber and with this study it was able to understand one of the reasons why different tunica have different fiber distribution since each tunica have a preferential direction that matches the maximum or medium elastic strain direction and the one with the largest fiber distribution is also the one that respond to the biggest shear stress. It seems fair to say that each layer has a structure function such as to withstand shear or normal stress depending on the fiber direction. Namely the media seems to have the function to withstand the normal stress at tangential direction and the adventitia to withstand the shear stress. Since each layer has a structure function due to its fiber distribution, the failure to withstand one of those stress could bring the whole structure to fail since the other layer won't respond well to forces at a direction that don't match to its fiber distribution. About the fluid domain one could realize that the fluid at heathy conditions could be treated as a Newtonian fluid at a laminar regime.

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