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ANALYSIS OF BLOOD FLOW IN THE CAROTID ARTERY USING COMPUTATIONAL FLUID DYNAMICS

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Abstract. *Computational Fluid Dynamics (CFD) is an important tool in technological development in different areas. Its application in bioengineering is known as Computational Hemodynamics, a field that studies the blood flow in the circulatory system. Computational Hemodynamics helps bioengineering to understand the cardiovascular diseases and improve treatments. The present work studied the pulsatile blood flow in a three-dimensional model of the carotid artery. It was developed using open-source softwares and tools, providing an open and accessible methodology. The open-source fluid-flow software OpenFOAM, was used to solve the hemodynamics of the carotid flow. The blood flow regime is pulsatile, where the diastole phase is when the heart muscle relaxes and fills with blood, and the systole phase is when the heart cavity contracts and blood is ejected into the arteries. The hemodynamics in the carotid artery was solved for an incompressible, laminar and Newtonian fluid flow. Three heart frequencies was considered: 65, 80 and 100 bpm. A three-dimensional model of the carotid artery bifurcation was considered and the FreeCAD was used. The computational meshes was created using the software cfMesh. Post-processing tools such as ParaView and Gnuplot were used to visualize and analyze the results. The streamlines and velocity profiles show back flow, or recirculation, in the carotid artery due to the sinus geometry. Therefore, the carotid bifurcation is more prone to the deposition of plaques in the blood. These preliminary simulations show that blood behavior is strongly dependent on its geometric characteristics and that the carotid artery has regions of back flow, which may favor the appearance of atherosclerosis.*

Keywords: OpenFOAM®, CFD, Carotid artery, Pulsatile flow, hemodynamics

1. INTRODUCTION

Computational Fluid Dynamics (CFD) is an important tool in technological development in different areas. Its application in bioengineering is known as Computational Hemodynamics, a field that studies the blood flow in the circulatory system. Computational Hemodynamics helps bioengineering to understand the cardiovascular diseases and improve treatments. Cardiovascular diseases represent the main cause of death in the world, mostly caused by atherosclerosis (Benjamin *et al.*, 2018). It is characterized by the hardening of the arterial walls due to the deposition of plaques. According to research, the artery geometry and the dynamics of blood flow impacts on the atheroma (Auricchio *et al.*, 2010; Carvalho, 1998). Due to the relevance of this matter, the present work studied the pulsatile blood flow in a three-dimensional model of the carotid artery.

Recently, many tools are used to investigate experimentally the carotid blood flow, such as Doppler Ultrasound and Magnetic Resonance Imaging (MRI). Doppler Ultrasound is a non-invasive and real-time technique that captures data of image, 2D structure and blood flow. Magnetic resonance, on the other hand, is a 3D technique but with low resolution for turbulence flow, which limits its use in diseased arteries (Gharahi *et al.*, 2016; Marshall, 2010; Marshall *et al.*, 2004). Another approach uses a computer simulation of flow and stress in vessel models. The high level of computer technology, accurate models and the development of efficient numerical methods are important tools to reproduce and study the hemodynamic characteristics.

In general, accurate CFD calculation requires a realistic geometry, an appropriate choice of modeling equations coupled with adequate inflow and outflow boundary conditions. Numerical results using the pulsatile flow condition at the inlet velocity and zero relative pressure at the outlet at the bifurcation of the carotid artery have been published. In these previous studies, recirculation zones were observed in the geometry with greater intensity in the acceleration phase, systole. In addition, it was noted that the bifurcation angle is related to hemodynamic factors. This suggests that hemodynamic parameters can be estimated based on the geometry of the carotid artery (Ozden *et al.*, 2020; Ramiar *et al.*, 2017; Saho and Onishi, 2016). In their project, Sharifi and Niazmand (2015) used three types of carotid siphons to examine variations in hemodynamic parameters in different regions of the arteries. In addition, in order to have a better estimate of the risk sites, the accumulation of low density lipoprotein (LDL) particles within the lumen, which scientists believe to be

the main cause of atherosclerosis, was considered. From the results, it can be inferred that the risk sites depend strongly on the form of the internal carotid artery (ICA) (Sharifi and Niazmand, 2015).

Other studies have estimated the effects of stenosis on common carotid artery (CCA) based on the concept of energy loss minimisation at the carotid bifurcation. The geometry of the carotid artery varies from individual to individual, hence Sia *et al.* (2016) analyzed the flow rate and energy loss for eight internal carotid arteries diagnosed with stenosis, a healthy internal carotid artery and an idealized internal carotid artery model. It was observed that the minimization of energy loss can be used as an indicator to estimate the stenosis of the internal carotid artery because the flow distribution at the bifurcation has a minimum total energy loss (Sia *et al.*, 2016). When Zhang *et al.* (2013) compared the simulation results with a constant boundary condition at the outlet pressure, the results showed that CFD simulation based on the proposed boundary conditions gave an accurate prediction of the critical stenosis ratio of carotid arteries at around 65%.

The present work was developed using only open-source softwares and tools, providing an open and accessible methodology. The open-source fluid-flow software OpenFOAM[®], was used to solve the hemodynamics of the carotid flow. Even though previous authors have shown results of pulsatile blood flow, reproducing and replicating such CFD simulations is a challenging task, specially using open-source tools (Ozden *et al.*, 2020; Ramiar *et al.*, 2017; Saho and Onishi, 2016; Sia *et al.*, 2016). This work aims to tackle such matter by providing detailed description of the methodology used to solve the pulsatile blood flow in the carotid artery.

2. METHOD

The carotids are a pair of arteries located on both sides of the human neck, being the main responsible for the blood flow to face and brain. The carotid artery has a main branch, called the common carotid artery (CCA), that bifurcates into two other branches: the external carotid artery (ECA) and the internal carotid artery (ICA). The carotid has an enlargement called sinus (or bulb) located at the bifurcation of the CCA and the beginning of the ICA. The ICA supplies blood flow to the brain and the ECA supplies the neck and face.

Blood can be defined as a suspension of elements (erythrocytes, leukocytes and platelets) in a fluid called plasma. In percentage terms, red blood cells make up 40-45% of the entire blood volume, while leukocytes and platelets together make up less than 1% (Tu *et al.*, 2015). The rest of the composition corresponds to the plasma.

The blood flow regime is pulsatile, where the diastole phase is when the heart muscle relaxes and fills with blood, and the systole phase is when the heart cavity contracts and blood is ejected into the arteries. The arteries are flexible and dilate when blood is pumped. According to Houssay (1984) the length-diameter relationship of the elastic tube has an influence on the flow behavior and the carotid artery can be considered as a rigid tube. In this case, when the liquid rhythmically enters a rigid tube, the liquid that exits from the other side has the same pulsatile characteristics.

The present work studies the blood flow in a healthy carotid artery considering as rigid walls and the fluid can be assumed Newtonian with constant properties (Tu *et al.*, 2015). Three heart frequencies were considered: 65, 80 and 100 bpm. Human blood has density just above the water (1060 Kg/m^3) and a viscosity of $3.2 \cdot 10^{-3} \text{ Kg/ms}$ (Tu *et al.*, 2015). The Reynolds number range is from 400 for the diastolic phase and 1200 for the peak of systole, assuming a human common carotid artery radius of $R = 4.0 \text{ mm}$ (Ku and Giddens, 1983).

Open source CFD software OpenFOAM[®] was used to perform the simulations. The hemodynamics in the carotid artery was solved for an incompressible, laminar and Newtonian fluid flow. The appropriate governing equations were solved using the pimpleFoam solver. OpenFOAM[®] uses the Finite Volume Method and a second-order central discretization in space and second-order implicit discretization in time was adopted. The time step of the simulations were controlled with a criteria based on maximum Courant of 1 with an initial time step of 10^{-5} . It was used 4 iterations on the pimple control loop and convergence of 10^{-5} for each time step was guaranteed.

A velocity profile shown in Fig. 1 for a fully developed pulsatile flow was prescribed at the inlet using groovyBC-FixedValue from swak4Foam library and the following equations

$$\mathbf{U}(y, t) = 0.1088[1 + 1.0001 \sin(\omega t) + 1.0001|\sin(\omega t)|]\mathbf{P}_b(\mathbf{y}) \quad (1)$$

$$\mathbf{P}_b = -\frac{(Y_{max} - y)(y - Y_{min})}{0,25(Y_{max} - Y_{min})^2} \mathbf{n} \quad (2)$$

where ω is the frequency, \mathbf{P}_b is a parabolic function used to describe the laminar flow at the inlet parameterized as a function of the y coordinate and \mathbf{n} is the inlet surface normal vector. The values of the constants in the previous equations were determined based on the values of U_{max} and U_{min} from the Reynolds numbers. The values of ω were 6.8 s^{-1} , 8.4 s^{-1} e 10.5 s^{-1} respectively for 65, 80 e 100 bpm. No slip boundary condition was used at the wall boundaries. Boundary conditions for the outlet pressure were also defined using swak4Foam library, set as reference pressure to zero using the groovyTotalPressure.

A three-dimensional model of the carotid artery bifurcation was considered. Its geometric parameters were based on data from Bharadvaj *et al.* (1982) and were obtained from 57 biplanar carotid angiograms of patients ranging from 34 to 77 year of age, with the average radius of the common carotid artery being 4 mm. The model used was constructed by

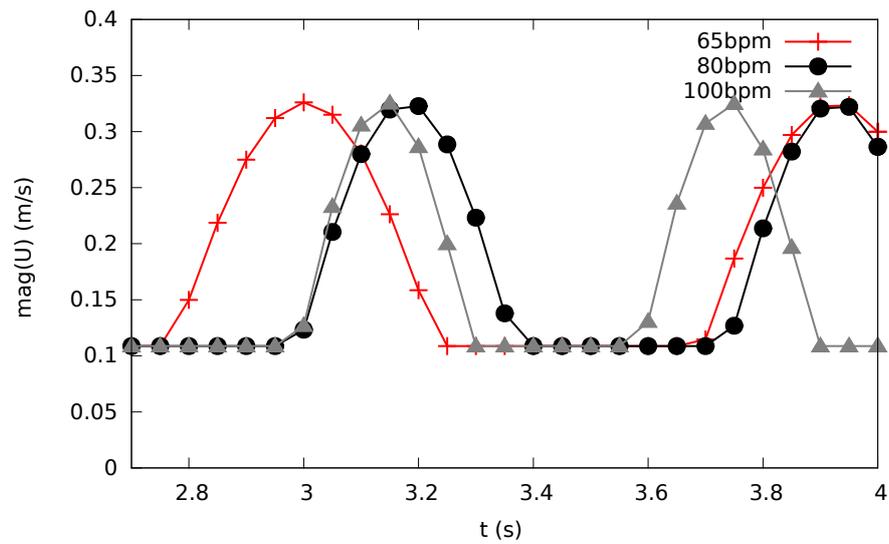


Figure 1. Magnitude velocity at center-line at the inlet (CCA)

Souza *et al.* (2017) in the software FreeCAD[®]. The computational meshes shown in Fig. 2 was created using the software cfMesh. Post-processing tools such as ParaView[®] and Gnuplot[®] were used to visualize and analyze the results.

Three different computational meshes were constructed for the carotid bifurcation. The unstructured meshes created had 46,000, 250,000 and 1,500,000 cells. For the analysis of mesh convergence, the velocity profiles at the sinus region was considered, as shown in Fig. 3. The meshes present the same qualitative behavior, with the 250,000 cells mesh having a 5.7% error in respect to the more refined mesh. From these results, it was decided to carry on the analysis for the different frequencies with the mesh of 250,000 cells.

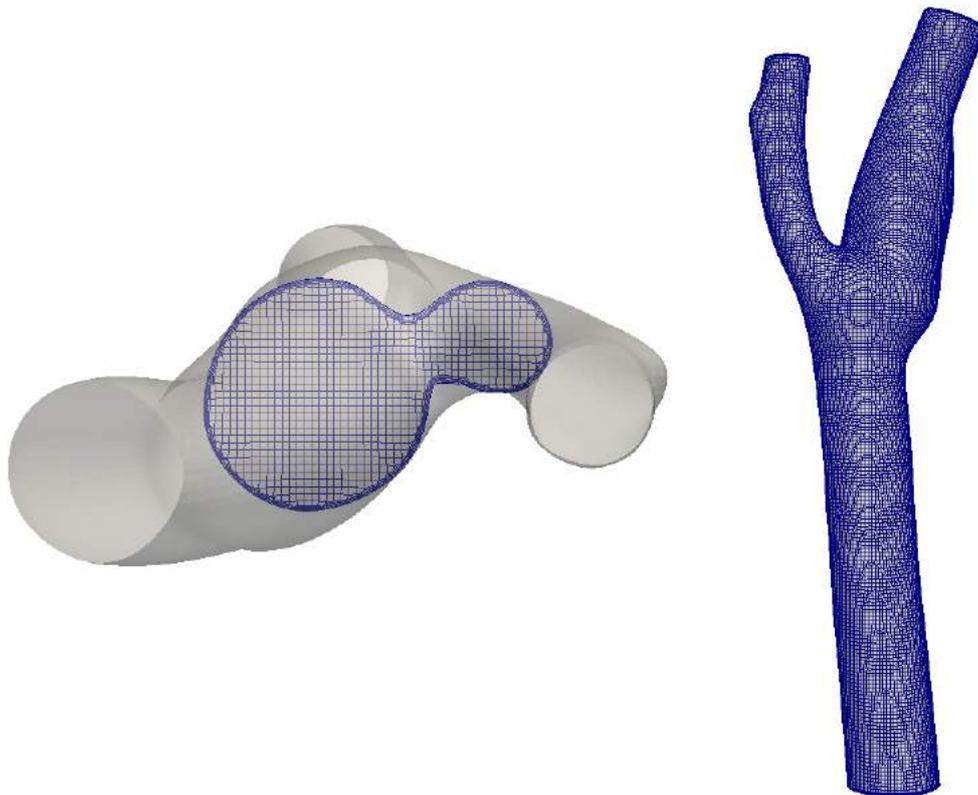


Figure 2. Geometry and mesh characteristics with 250,000 cells.

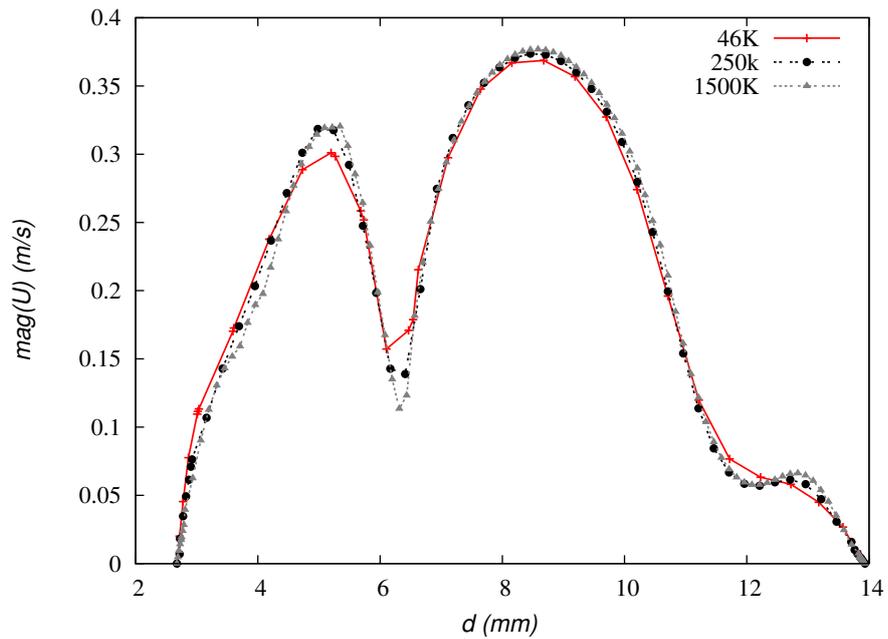


Figure 3. Magnitude of velocity in sinus region for all meshes

3. RESULTS

The carotid bifurcation geometry impacts on the flow characteristics of the region, such as: the curvature at the junctions of the common carotid artery - internal carotid artery and common carotid artery - external carotid artery; the widening of the diameter at the entrance of the internal carotid artery, known as the sinus, which abruptly narrows at the exit; and the asymmetry of the internal and external arteries. In this study, it is shown the results for the velocity field for pulse frequencies of 65, 80 and 100 bpm. The normalized velocity profile and streamlines for the carotid artery is shown in Fig. 4, Fig. 5, Fig. 6 and Fig. 7 for the systolic and diastolic phases at each heart rate (65, 80 and 100 beats per minute), after a time-periodic state was guaranteed. The velocity results presented were normalized in respect to the maximum velocity obtained between the three frequencies, being $U_{max} = 0.435 \text{ m/s}$.

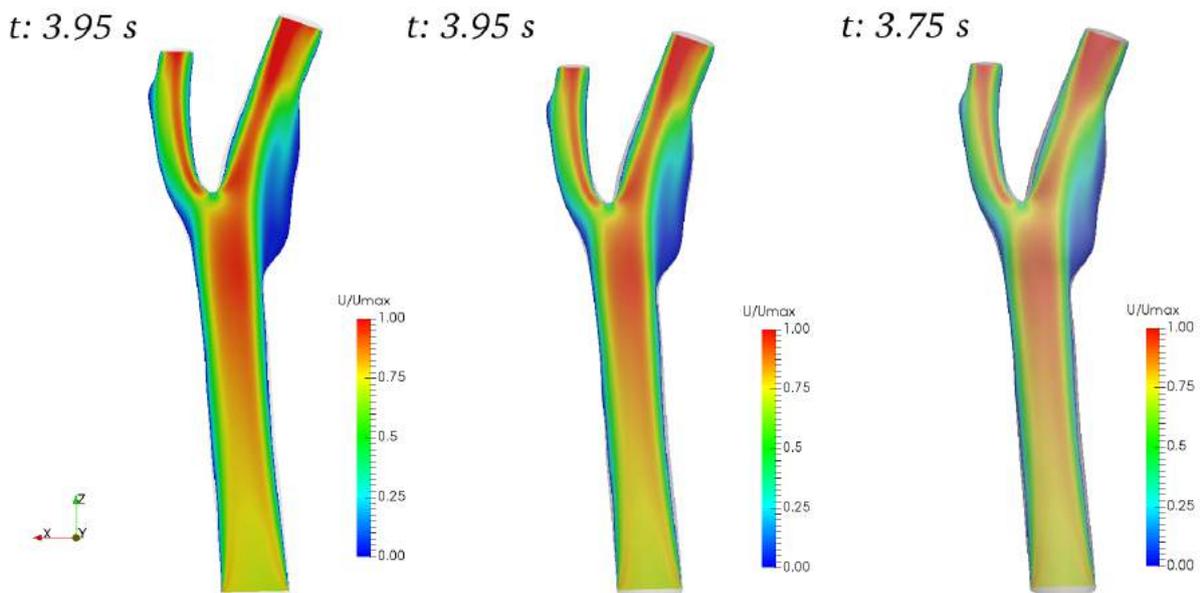


Figure 4. Magnitude of normalized velocity field in the systolic phase: (a) 65 bpm, (b) 80 bpm e (c) 100 bpm.

Figure 4 and Figure 5 shows magnitude of normalized velocity field in the systolic and diastolic phase for all frequencies. The authors call the attention to the fact that the simulations results were saved at every 0.05 s. As a result,

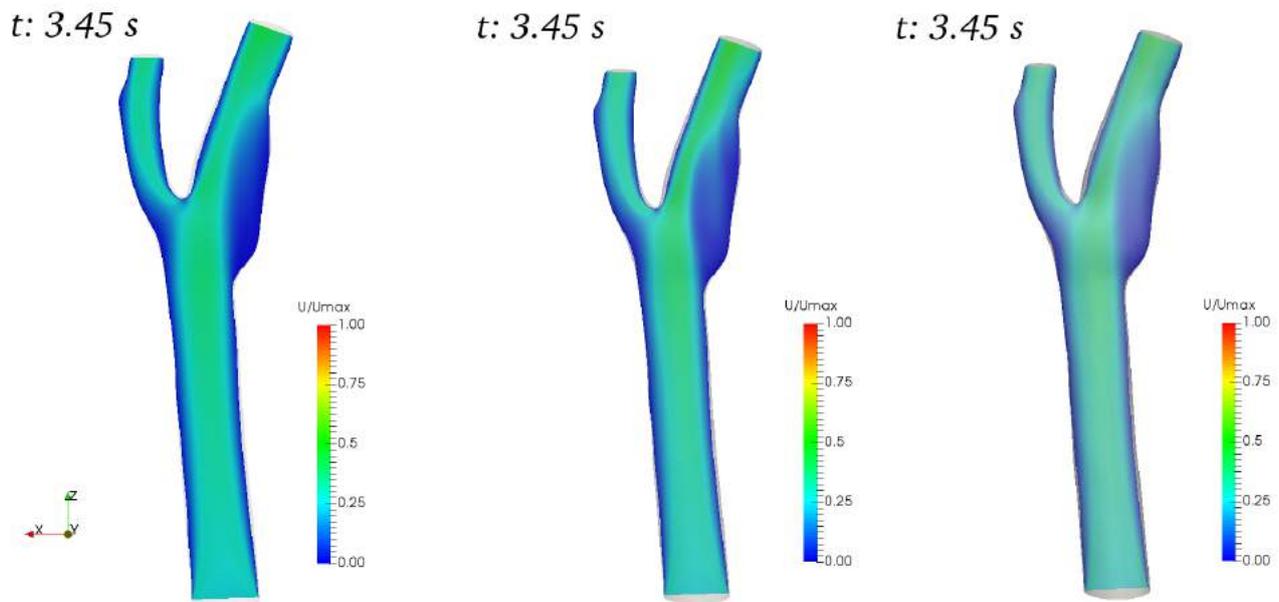


Figure 5. Magnitude of normalized velocity field in the diastolic phase: (a) 65 bpm, (b) 80 bpm e (c) 100 bpm.

the temporal resolution for the frequencies of 65 and 80 bpm was the same, and results are shown at 3.95 s, where for 100 bpm the systolic phase is shown at 3.75 s. Nevertheless, these results are in agreement with the flow characteristics expected in the carotid artery for all cases. The velocity gradient and stagnation point in the sinus region were observed. It can also be noted that the results are similar for all frequencies.

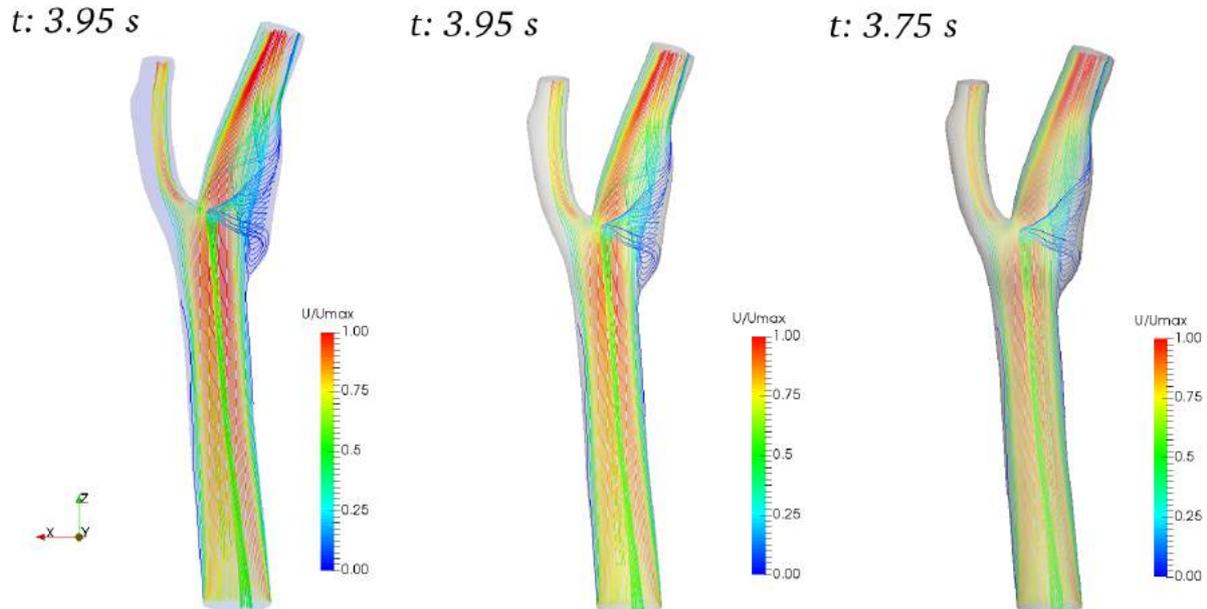


Figure 6. Streamlines of the magnitude of normalized velocity field in the diastolic phase: (a) 65 bpm, (b) 80 bpm e (c) 100 bpm.

The analysis of streamlines of the magnitude of normalized velocity field, Figs. 6 and 7, show the area where the flow presents critical behavior due to the complex geometry of the sinus. The results show back flow, or recirculation, as expected in the carotid artery and in agreement with the literature. The sinus and the non-dividing wall at the junction of the common carotid and the external carotid arteries are low-speed regions. Because of this condition and the blood being a fluid composed of solid particles suspended in the plasma, this region is more prone to the deposition of these particles, that is, even under normal health conditions, the carotid bifurcation tends to accumulate particles in this region.

Figure 8 compares magnitude of velocity field for all frequencies at the sinus region and systolic phase. The results show the same profile for all frequencies where a maximum difference of 4% was observed. However, the authors call

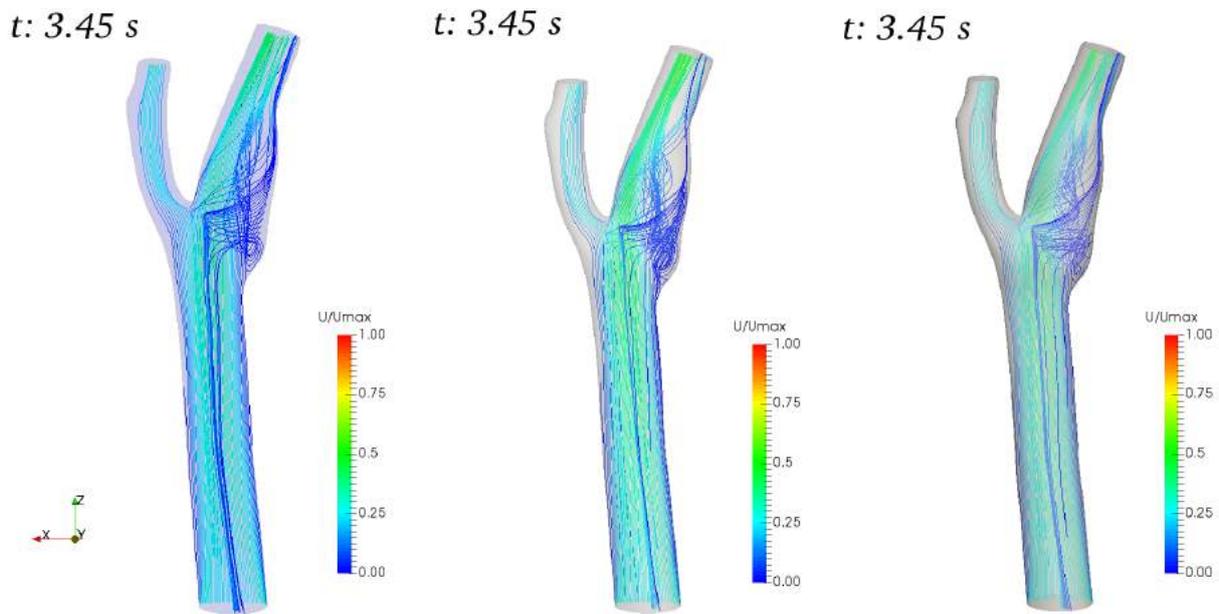


Figure 7. Streamlines of the magnitude of normalized velocity field in the diastolic phase: (a) 65 bpm, (b) 80 bpm e (c) 100 bpm.

the attention to the fact that the mesh error was up to 5.7%. Therefore, for this mesh accuracy, all frequencies showed the same results.

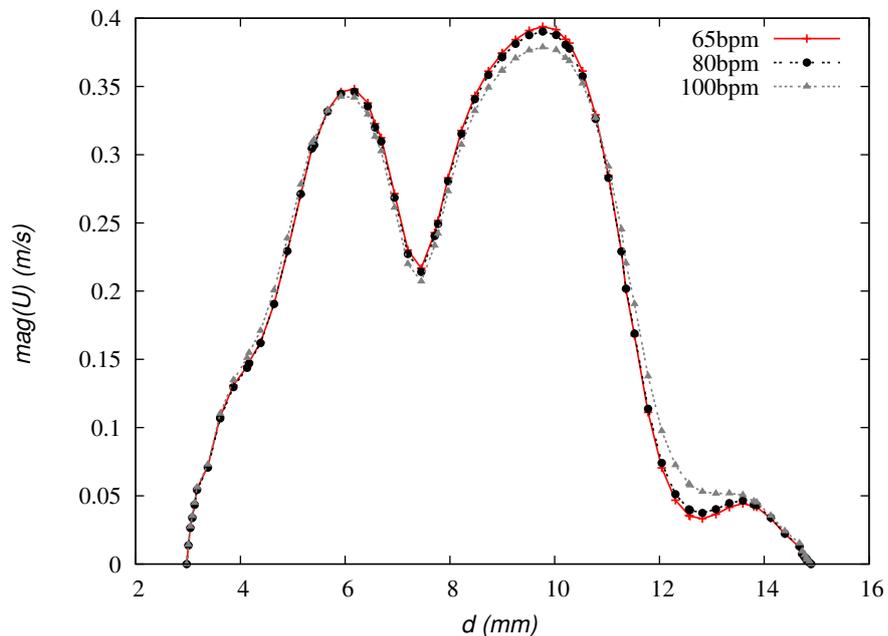


Figure 8. Magnitude of velocity field for all frequencies at the sinus region and systolic phase.

4. CONCLUSIONS

The present work shows that it was possible to develop a set of simulation parameters capable of reproducing the flow in a carotid in accordance with the results previously published in the literature. These preliminary simulations shows that blood behavior is strongly dependent on its geometric characteristics and that the carotid artery has regions of reverse flow, which may favor the appearance of atherosclerosis. The importance of simulations in this artery lies in the fact that if the flow pattern is known and the influence of geometry is understood, it is possible to verify the presence of disturbances in the velocity and pressure profiles, and thus contribute to a better understanding of the conditions that lead

to atherosclerosis.

5. ACKNOWLEDGEMENTS

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6. REFERENCES

- Auricchio, F., M., B., G., S. and B., V., 2010. “Carotid artery stenting simulation: From patient-specific images to finite element analysis”. *Elsevier*, , No. 9, p. 1.
- Benjamin, E.J., Virani, S.S., Callaway, C.W., Chamberlain, A.M., Chang, A.R., Cheng, S., Chiuve, S.E., Cushman, M., Dellinger, F.N., Deo, R., de Ferranti, S.D., Ferguson, J.F., Fornage, M., Gillespie, C., Isasi, C.R., Jiménez, M.C., Jordan, L.C., Judd, S.E., Lackland, D., Lichtman, J.H., Lisabeth, L., Liu, S., Longenecker, C.T., Lutsey, P.L., Mackey, J.S., Matchar, D.B., Matsushita, K., Mussolino, M.E., Nasir, K., O’Flaherty, M., Palaniappan, L.P., Pandey, A., Pandey, D.K., Reeves, M.J., Ritchey, M.D., Rodriguez, C.J., Roth, G.A., Rosamond, W.D., Sampson, U.K., Satou, G.M., Shah, S.H., Spartano, N.L., Tirschwell, D.L., Tsao, C.W., Voeks, J.H., Willey, J.Z., Wilkins, J.T., Wu, J.H., Alger, H.M., Wong, S.S. and Muntner, P., 2018. “Heart disease and stroke statistics—2018 update: A report from the American Heart Association”. Technical report, American Heart Association.
- Bharadvaj, B.K., Mabon, R.F. and Giddens, B.P., 1982. “Steady flow in a model of the human carotid. part i: Flow visualization”. *Journal Biomechanics*, Vol. 15, pp. 349–362.
- Carvalho, S., 1998. *Uma Simulação Numérica do Escoamento na Artéria Carótida*. Mestrado em matemática aplicada, Instituto de Matemática, Universidade Federal do Rio Grande do Sul, Porto Alegre, Brasil.
- Gharahi, H., Zambrano, B.A., Zhu, D.C., DeMarco, J.K. and Baek, S., 2016. “Computational fluid dynamic simulation of human carotid artery bifurcation based on anatomy and volumetric blood flow rate measured with magnetic resonance imaging”. *International Journal of Advances in Engineering Sciences and Applied Mathematics*.
- Houssay, B.A., 1984. *Fisiologia Humana*. Guanabara Koogan, Rio de Janeiro, Brasil, 5th edition.
- Ku, D.N. and Giddens, D.P., 1983. “Pulsatile flow in a model carotid bifurcation”. *Journal of the American Heart Association*, , No. 9, pp. 31–39.
- Marshall, I., 2010. “Computational simulations and experimental studies of 3d phase-contrast imaging of fluid flow in carotid bifurcation geometries”. *Journal of Magnetic Resonance Imaging*.
- Marshall, I., Zhao, S., Papathanasopoulou, P., Hoskins, P. and Xu, X.Y., 2004. “Mri and cfd studies of pulsatile flow in healthy and stenosed carotid bifurcation models”. *Journal of Biomechanics*.
- Ozden, K., Sert, C. and Yazicioglu, Y., 2020. “Effect of stenosis shape on the sound emitted from a constricted blood vessel”. *Medical Biological Engineering Computing*.
- Ramiar, A., Larimi, M.M. and Ranjbar, A.A., 2017. “Investigation of blood flow rheology using second-grade viscoelastic model (phan-thien-tanner) within carotid artery”. *Acta of Bioengineering and Biomechanics*.
- Saho, T. and Onishi, H., 2016. “Evaluation of the impact of carotid artery bifurcation angle on hemodynamics by use of computational fluid dynamics: a simulation and volunteer study”. *Radiological Physics and Technology*.
- Sharifi, A. and Niazmand, H., 2015. “Analysis of flow and ldl concentration polarization in siphon of internal carotid artery: Non-newtonian effects”. *Computers in Biology and Medicine*.
- Sia, S.F., Zhao, X., Li, R., Zhang, Y., Chong, W., He, L. and Chen, Y., 2016. “Evaluation of the carotid artery stenosis based on minimization of mechanical energy loss of the blood flow”. *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*.
- Souza, T.P., H., B.E. and G., J.F., 2017. “Simulação do escoamento sanguíneo na bifurcação da artéria carótida”. In *Encontro Nacional de Modelagem Computacional*. p. 10.
- Tu, J., Inthavong, K. and Wong, K.K.L., 2015. *Computational Hemodynamics*. Springer Netherlands, New York City, USA, 1st edition.
- Zhang, Y., Furusawa, T., Sia, S.F., Umezu, M. and Qian, Y., 2013. “Proposition of an outflow boundary approach for carotid artery stenosis cfd simulation”. *Computer Methods in Biomechanics and Biomedical Engineering*.

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