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COMPARISON OF DIFFERENT RHEOLOGICAL MODELS FOR COMPUTATIONAL SIMULATION OF BLOOD FLOW IN CENTRAL VENOUS ACCESS

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Abstract. *Numerical models based on computational fluid dynamics allow the investigation of hemodynamic factors that lead to thrombus formation and progression. When simulating blood flow, physical modeling of its viscosity is a point of uncertainty. Although blood tends to behave as a non-Newtonian fluid in the venous circulation, it is possible that the insertion of a catheter into the jugular veins, during hemodialysis procedure, leads to higher strain rates, making blood behaves as a Newtonian fluid. This study proposes the development of a computational model to simulate blood flow in central venous access and to compare different rheological models commonly used in numerical simulations to represent blood viscosity. The software ANSYS-Fluent® 19.2 was used to perform simulations in transient regime comprising a total time of one cardiac cycle. Different rheological models commonly used for blood viscosity, as Carreau-Yasuda and Power-Law models, were compared with the Newtonian model, reveling no significant differences. It was found that, in many parts of the physical domain, blood behave as a Newtonian fluid along almost all the duration of a cardiac cycle. Only at the contraction of the right atrium, at t = 0.3 s, blood seems to behave, predominantly, as a non-Newtonian fluid along almost all the physical domain.*

Keywords: *central venous catheter, computational fluid dynamics, rheological models, blood rheology, blood flow, thrombus formation, hemodialysis*

1. INTRODUCTION

Non-maturation of the fistula in patients undergoing hemodialysis leads to prolonged use of the central venous catheter (CVC) and may compromise the catheter patency due to thrombus development. Non-tunneled catheters are generally designed with side holes facing different directions on the catheter wall. These side holes may affect the flow pattern, shear stress distribution, flow separation and recirculation, thus providing a critical hemodynamic environment (Lucas et al., 2014a).

Considering blood clot formation, initiation of platelet activation has been reported to occur in regions of high shear and stagnation, but also in recirculation zones where the time during which the platelets are exposed to shear becomes

an important factor. Another flow characteristic identified as a potential platelet activator is vortical flow structures that allows blood clots to develop (Berg et al., 2018)

Due to the correlation between the physical properties of the velocity field and the thrombus formation, it is essential to understand the flow characteristics close to the artificial devices. However, the exact influence of the physical mechanisms on thrombus formation still being only partially understood in literature (Fogelson and Neves, 2014b)

Numerical models based on computational fluid dynamics (CFD) allow the investigation of hemodynamic factors that lead to thrombus formation and progression in a relatively inexpensive and non-invasive way (Lucas et al., 2014a). However, studies are necessary to make those models more accurate and to reduce the amount of simplifications required.

When simulating blood flow, physical modeling of its viscosity is a point of uncertainty. The rheological behavior of blood depends on the shear stress, diameter of the vessel and hematocrit. Investigations have shown that at higher shear rates, greater than 100 s^{-1}, blood exhibits a constant apparent viscosity. At lower shear rates, however, blood viscosity significantly increases as the shear rate decreases (Silveira, 2017b)

The Newtonian blood rheology is one of the most popular assumptions in CFD models, yet it has been subject to extensive debate and criticism in the literature. It is well known that blood exhibits shear-thinning behaviour where its viscosity increases in the low shear rate regime (Fung, 2013). Shear-thinning generalized Newtonian models (commonly referred to as non-Newtonian models in the hemodynamics literature) have been proposed to model the non-Newtonian behaviour of blood (Yilmaz, 2008). At shear rates higher than 50 s, these models asymptote to the Newtonian limit therefore justifying the popular Newtonian assumption in large arteries and the non-Newtonian assumption in veins (Arzani, 2018).

Although blood tends to behave as a non-Newtonian fluid in the venous circulation, it is possible that the insertion of a catheter into the jugular veins, during hemodialysis procedure, leads to higher strain rates. Inside the catheter, as well as in its lateral orifices, blood flows with higher velocities than inside the vein. One of the questions that arises is whether, in those regions, the strain rates exceed, or not, the 100 s^{-1}.

In the literature there are several models validated for non-Newtonian properties of blood and selected as the most suitable for blood flow velocity and shear stress predictions, for example: Carreau-Yasuda, Casson, Power-low and Cross. However, not many studies exist regarding the rheological behavior of blood in-vivo circulation (Ionescu., 2017a).

Aiming to contribute with a better elucidation of these questions, this study proposes the development of a computational model to simulate blood flow in central venous access and to compare different rheological models commonly used in numerical simulations to represent blood viscosity. The simulations were performed in transient state comprising a total time of one complete cardiac cycle.

2. MATERIALS AND METHODS

2.1 Three dimensional geometry

SolidWorks (SolidWorks, Inc., Concord, MA, USA) was used to design a CVC model similar to the clinical device MedCOMP/HEMO-CATH (Harleysville, PA, USA). It was designed with a dual-lumen system, arranged in a single shaft 2 mm in diameter and 20 cm in length.

To develop a three-dimensional model of the veins it was used the procedure described by Machado et al. (2018). The geometry model was designed using InVesalius3® (CTIRA, Campinas, SP, Brazil) to process images obtained from a computed tomography of a healthy 74-year-old male patient, as illustrated in Fig. 1
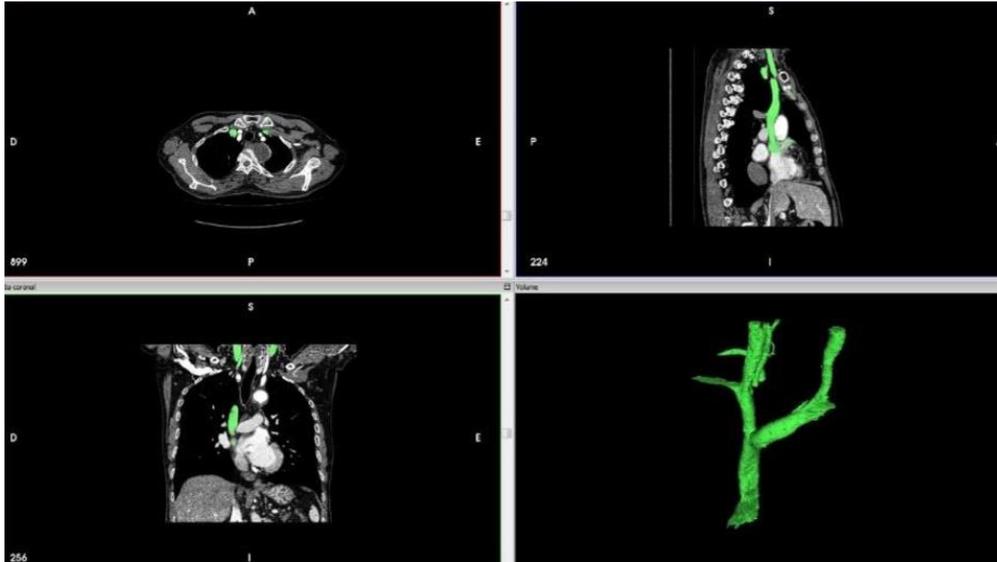
Figure 1. Development three-dimensional model of the central veins using InVesalius3® by (Machado et al., 2018).

Figure 2 illustrates the three-dimensional model of the central veins, with the catheter inserted, used in the simulations.
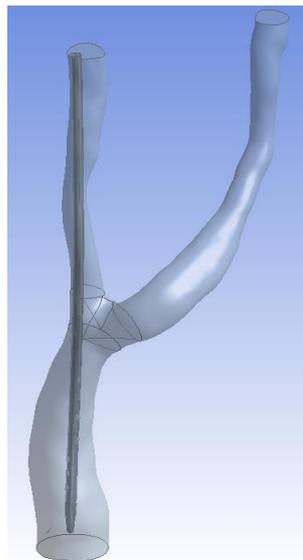

Figure 2. Three-dimensional model used in the simulations.

2.2 Computer mesh

The mesh used in the simulations consisted of 866930 nodes and 3174083 elements, being 2364026 tetrahedral, 1704 pyramidal and 808353 wedges elements. It was refined near the catheter, vein walls, and in the regions with the highest gradients of pressure, strain rates, and velocity.

The mesh test was done following ASME (2009) standards, in which the mesh was refined using adaptive criteria in relation to the velocity, pressure and strain rate gradient until the results become mesh independent.

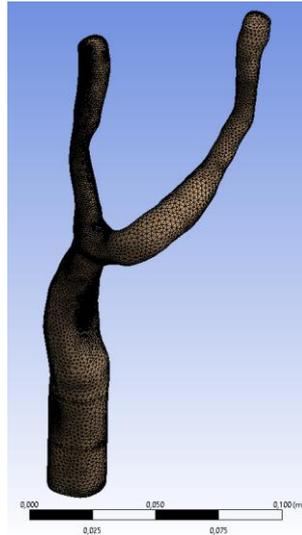Figure 3 shows the mesh used in the simulations.

Figure 3. Mesh used in the simulations.

2.3 Boundary conditions and numerical simulation

The software ANSYS-Fluent® 19.2 (ANSYS-Fluent Inc., Lebanon, NH, USA) was used to solve the equation of continuity and the time-dependent incompressible equation of momentum for each different rheological model considered in this study.

Table 1 contains blood properties considered for each rheological model. It was set a valor of 1060 kg/m^3 for the blood density.

Table 1. Properties and rheological models considered for blood.

Rheological Model	Maximum Viscosity [kg/m.s]	Minimum Viscosity [kg/m.s]	Power-Law index (n)	Time constant (λ) [s]	Consistency index (k) [Pa.s^n]	Dynamic viscosity (μ) [kg/m.s]
Carreau-Yasuda	0.056	0.0035	0,3568	3.313s	----	----
Power - Law	0.0058	0.0035	0,7755	----	0,01467	----
Newtonian	----	----	----	----	----	0.0035

The simulations were performed using k-ω Shear Stress Transport (SST) turbulence model. The advantage of this model over the k-ε model is its improved performance for boundary layers under adverse pressure gradients. Therefore, this model is substantially more efficient to determine the wall shear stress in the regions closed to the wall of the catheter (Lucas et al., 2014a).

For spatial discretization of the momentum and continuity it was used the second order *Upwind* method. For the gradients it was used the *Least Squares Method Cell-Based Gradient*. The *Coupled Method* was adopted to solve the pressure-velocity coupling. Finally, the *PRESTO* scheme was used to the pressure equation discretization.

The duration of a cardiac cycle was assumed to be 0.8 s, yielding a heart rate of 75 beats per minute. The second-order backward Euler method was used for time integration. A constant time step was employed, $\Delta t = 0.002$ s, corresponding a total of 400 times steps per cardiac cycle. Whit this time steps was possible to obtain a courant number close to 1. In this study we have simulated just one cardiac cycle.

The waveforms of the jugular blood inlet velocity and atrium pressure outlet were approximated from the experimental curves found in the literature and intravenous color Doppler lines. The (A) wave corresponds to the right atrium contraction, which increases the pressure in the central veins and propel blood in a retrograde direction. When the tricuspid valve closes, the systole wave (S) occurs. The transitional (V) wave corresponds to atrial overfilling against a closed tricuspid valve, anticipating the opening of the valve in diastole (D). The inlet and outlet wave forms designed for this study showed a correlation between flow velocity in the internal jugular veins and the pressure in the right atrium, as shown in Fig. 4 (Lucas et al., 2014a).
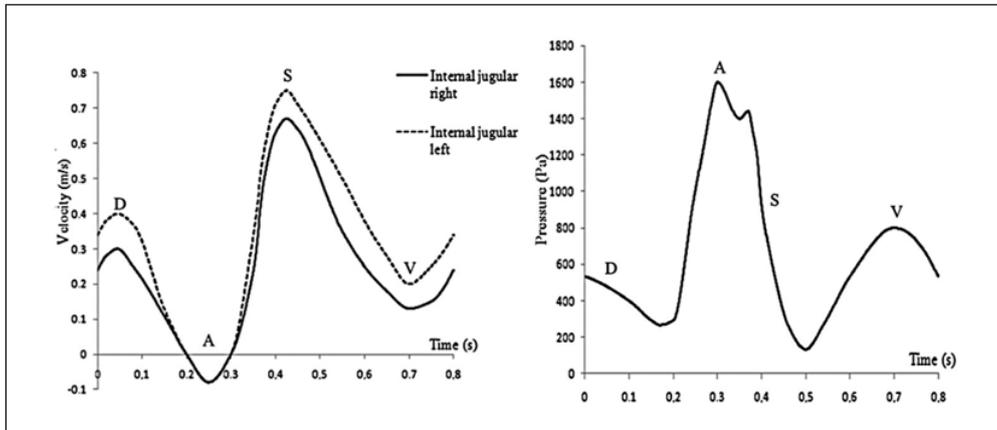
Figure 4. Boundary conditions by (Lucas et al., 2014a).

3. RESULTS

It was obtained the strain rates distributions in time, along one cardiac cycle, for each one of the rheological models considered in this study, as shown in Fig. 5. It is perceptible that in regions closed to the catheter, vein walls and as well as in many regions inside the left jugular vein, strain rates exceed $100 \ s^{-1}$ along almost all the duration of the cardiac cycle, which suggests that blood behaves as a Newtonian fluid. Only for $t = 0.3$ s, when occurs the contraction of the right atrium, blood seems to behave, predominantly, as a non-Newtonian fluid in most parts of the physical domain. This observation contradicts the assumption that blood behave as a non-Newtonian fluid in the venous circulation. It may have occurred due to the disturbance caused by the catheter in the flow field. It emphasizes that, when in contact with artificial devices blood is submitted to different conditions of shear stress and strain rate then when it is in physiological flow conditions, and it may be analyzed to the safety and efficiency of the clinical procedure.
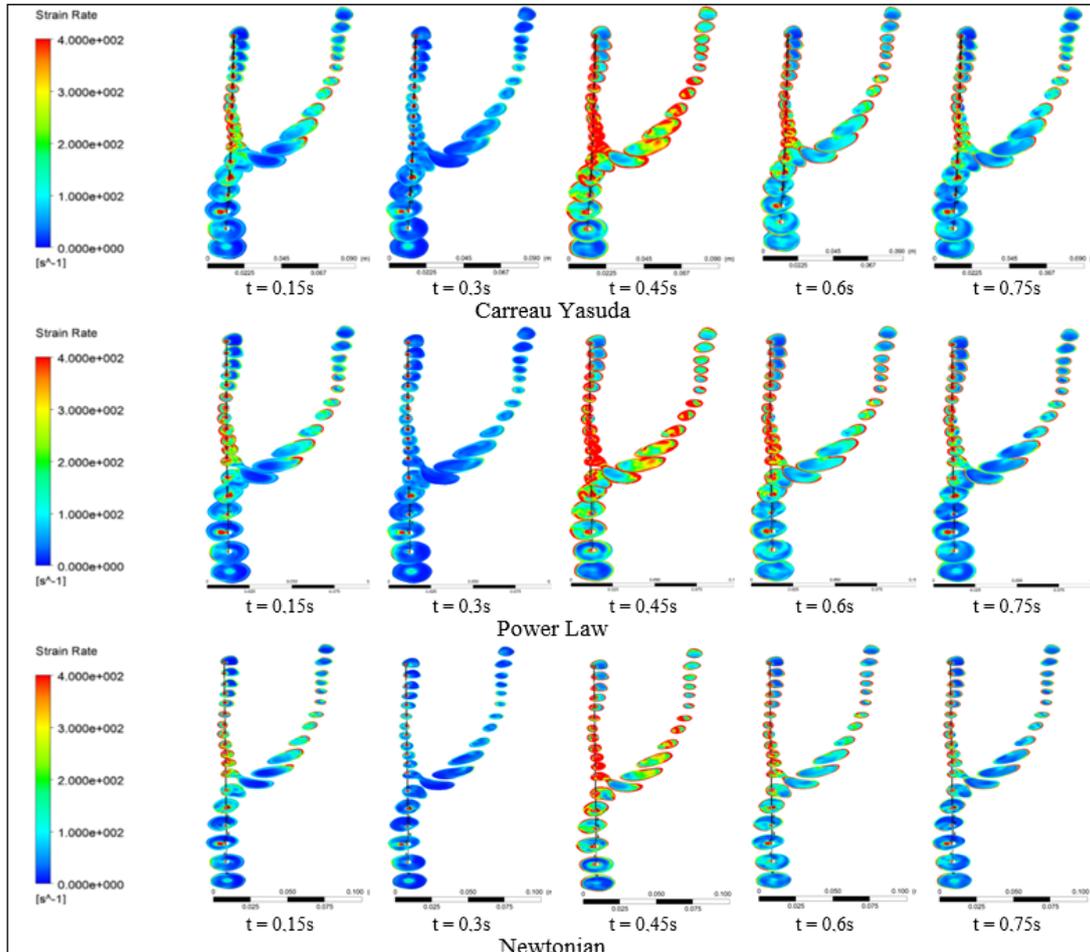
Figure 5. Strain rate distributions along the cardiac cycle.

To analyze and compare the average values of strain rate obtained with the use of each rheological models in different instants of time of the cardiac cycle, it was defined 8 planes inside the physical domain. The location of these planes is illustrated in Fig. 6. Table 2 presents the results of this comparison and the respective percentual difference.
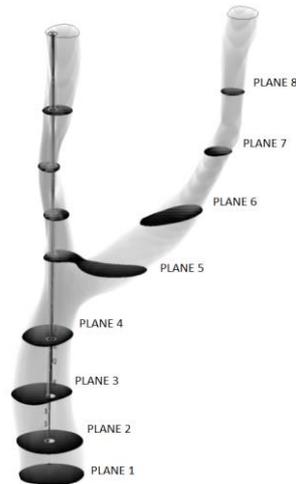

Figure 6. Planes location to analyze mean strain rate values.

Table 2. Comparison of mean strain rate values obtained with the use of each rheological model.

Plane	Carreau-Yasuda	Newtonian	Power Law	Error [%]
Strain Rate Values (s^{-1}) for t = 0.15 s				
1	50.25	51.90	49.86	3.94%
2	231.34	234.52	232.55	1.35%
3	454.69	457.49	453.55	0.86%
4	926.03	922.32	924.79	0.40%
5	871.14	896.95	882.93	2.88%
6	1270.75	1265.06	1262.57	0.64%
7	1096.83	1102.33	1092.24	0.92%
8	1152.93	1153.08	1149.58	0.30%
Strain Rate Values (s^{-1}) for t = 0.3 s				
1	35.41	37.85	35.94	6.44%
2	234.06	238.62	234.97	1.91%
3	490.83	494.53	489.21	1.07%
4	862.01	861.29	862.93	0.19%
5	644.09	640.27	643.52	0.59%
6	906.64	905.69	908.39	0.30%
7	1057.59	1054.45	1052.88	0.45%
8	1072.35	1077.34	1067.28	0.93%
Strain Rate Values (s^{-1}) for t = 0.45 s				
1	118.97	118.70	118.34	0.53%
2	399.68	401.92	401.73	0.56%
3	611.04	611.14	609.06	0.34%
4	1060.53	1067.08	1064.96	0.61%
5	1365.97	1368.02	1371.66	0.41%
6	2349.85	2318.83	2323.98	1.32%
7	1931.79	1957.46	1954.37	1.31%
8	1919.09	1916.13	1915.74	0.17%
Strain Rate Values (s^{-1}) for t = 0.6 s				
1	105.97	107.45	105.70	1.63%
2	319.47	316.84	310.13	2.92%
3	507.30	513.30	507.31	1.17%
4	932.85	938.48	927.50	1.17%
5	1044.24	1048.39	1046.83	0.40%
6	1649.40	1638.96	1640.77	0.63%
7	1284.24	1282.60	1276.89	0.57%
8	1444.35	1442.07	1441.05	0.23%
Strain Rate Values (s^{-1}) for t = 0.75 s				
1	80.03	85.75	81.01	6.67%
2	289.01	290.67	288.21	0.85%
3	467.80	467.99	465.14	0.61%
4	892.72	910.64	904.29	1.97%
5	776.27	780.67	774.46	0.79%
6	1315.62	1315.82	1311.82	0.30%
7	1202.14	1202.48	1198.04	0.37%
8	1332.40	1331.26	1329.40	0.23%

As shown in Tab. 2, it is perceptible that no great difference was evidenced with the use of each rheological model considered in this study, which suggests that all of them are equally suitable for modeling blood viscosity. Only in plane 1, it was evidenced mean values of strain rate lower than 100 s^{-1}. That probably happened because this plane does not cross the catheter, and the mean strain rate values of blood inside the catheter are much higher than in order regions of the veins, approaching values of 5000 s^{-1}. This might have influenced the average values of strain rate in each plane. Figure 7 illustrate the strain rate distributions in each plane analyzed in Tab. 2, for the Carreau-Yasuda model. It is perceptible that the flow inside the catheter always presents high shear rates. The flow near the catheter presents shear rates lower than 100 s^{-1} only at t = 0.3 s, when occurs the contraction of the right atrium.

Thrombus formation and hemolysis are directly related to fluid shear stress, so that the higher the shear stress, the greater the thrombolytic and hemolytic potential. Thus it is noteworthy that near the catheter region there is a high probability of these events occurring, which leads to concern about the time of blood exposure to these devices.
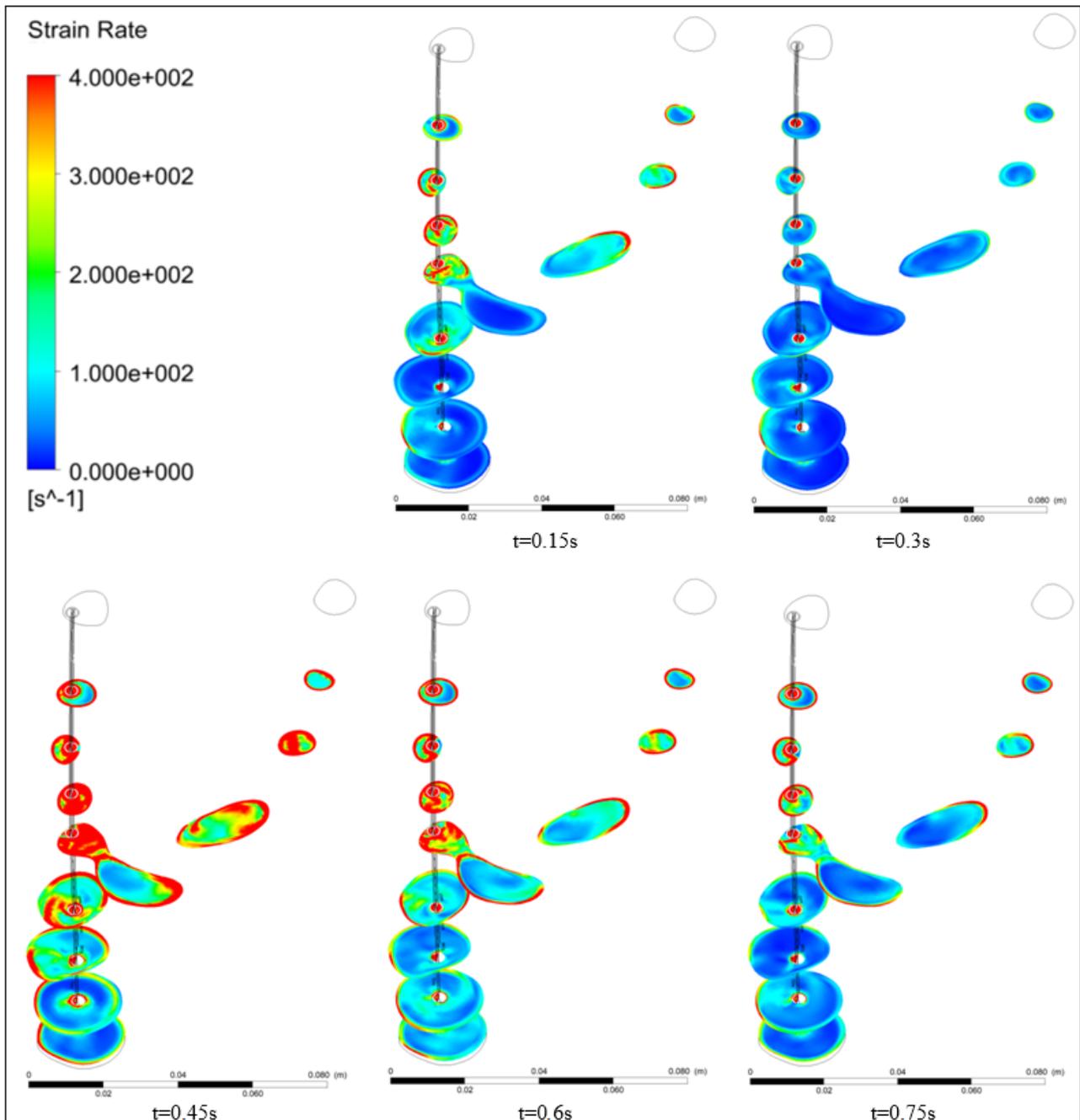

Figure 7. Strain rates distributions along the cardiac cycle.

It was defined 4 points in the physical domain to analyze the strain rate distribution in time along a cardiac cycle for each rheological model considered. Figure 8 illustrate the location of those points. Figure 9 presents a comparison between the results obtained with the use of each rheological models. It seems that the comparison of the absolute values of strain rates results in greater difference than the comparison of its mean values. It probably happened because the greatest difference in strain rate values seems to occur in specifics instants of time of the cardiac cycle, as shown in Fig. 9. It suggests that each rheological model has a different response time. That deference seems to decrease to strain rates higher than 100 s^{-1}, which is expected, because that is the condition when the blood behaves as a Newtonian fluid, and it is predicted in the mathematical formulation of the non-Newtonian rheological models considered in this study.

This leads to the conclusion that the use of the Newtonian model demonstrates average behavior compatible with non-Newtonian models, however locally significant differences may occur.
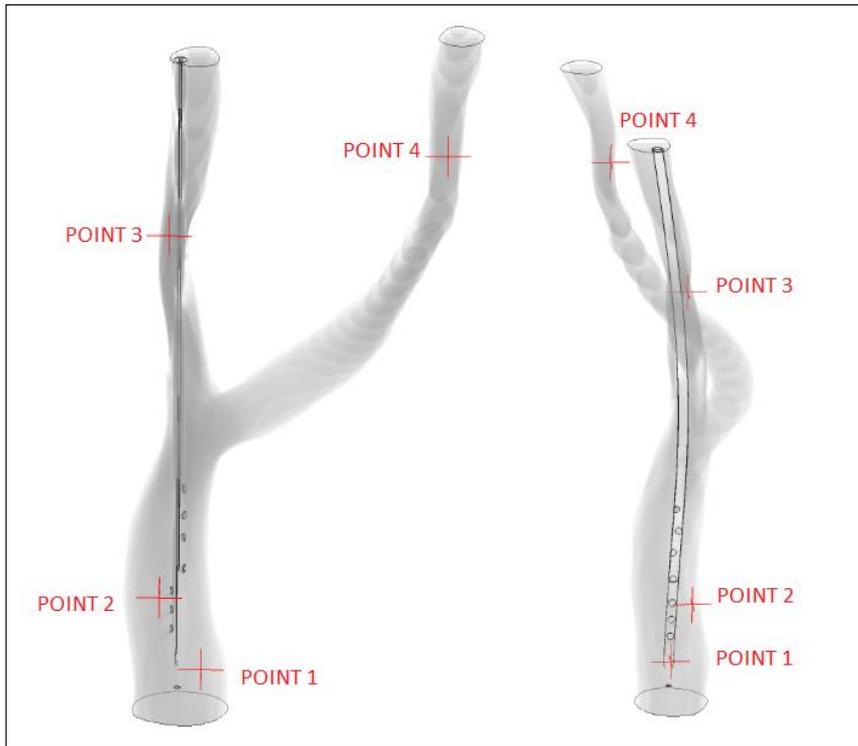
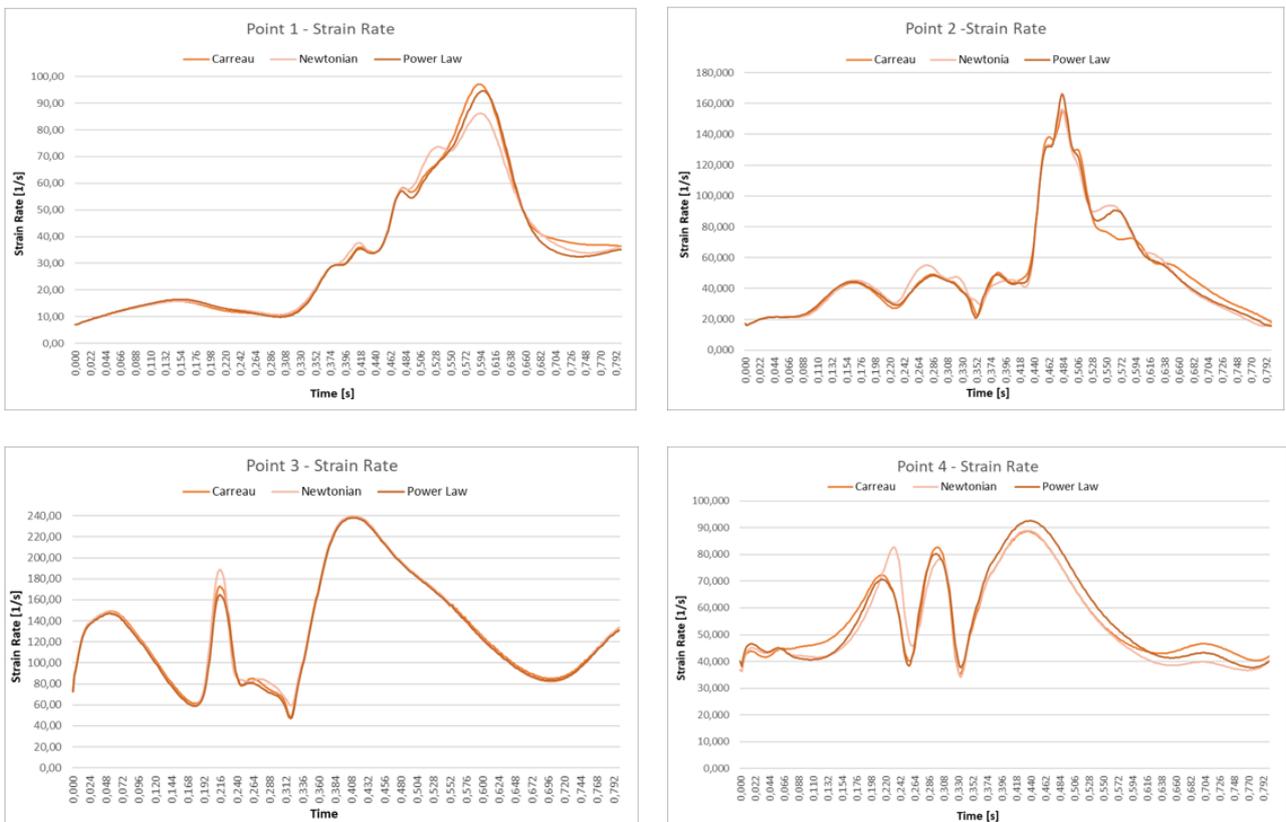Figure 8. Points defined to analyze the strain rate distribution in time for each rheological model.

Figure 9. Strain Rate distribution along the cardiac cycle in some points

4. CONCLUSION

No great difference in strain rate values was evidenced with the use of each rheological model considered in this study, which suggests that all of them are suitable for modeling blood viscosity.

In most parts of the geometrical domain blood exhibits a Newtonian behavior (shear rates over than 100 s^{-1}) during almost all the duration of the cardiac cycle. Only at the contraction of the right atrium, t = 0.3 s, blood seems to behave, predominately, as a non-Newtonian fluid in most parts of the physical domain.

The insertion of the catheter seems to modify the hemodynamic environment inside the central veins, thus providing higher shear rates, which may lead to higher shear stress and the formation of thrombus. To confirm this affirmation, simulations in the central veins without the catheter should also be performed.

5. ACKNOWLEDGEMENTS

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