Eleventh International Conference on CFD in the Minerals and Process Industries
CSIRO, Melbourne, Australia
7-9 December 2015

CFD SENSITIVITY STUDY FOR NEWTONIAN VISCOSITY MODEL IN CEREBRAL ANEURYSMS

Hashem M. ALARGHA[1], Mohammad O. HAMDAN[1], Amr ELSHAWARBY[2], Waseem H. AZIZ[2]

[1]United Arab Emirates University, Mechanical Engineering Department, United Arab Emirates

[2]Tawam Hospital, Neuro Endovascular Centre, United Arab Emirates, Al Ain

*Corresponding author, E-mail address: 201470434@uaeu.ac.ae

ABSTRACT

Computational fluid dynamics has become a popular clinical tool for studying cerebral aneurysms and predicting the cause of initiation, the growth rate and the rupture risk of aneurysms. The CFD simulations in cerebral aneurysm adopt different assumptions to simplify the simulation. One of these assumptions is the Newtonian blood viscosity model. In this work, a numerical study is constructed to inspect whether the Newtonian blood viscosity model assumption is appropriate or not and how such model is compared to a realistic non-Newtonian viscosity model. A comparison is done between the Newtonian model and the Carreau model for three different patients for maximum wall shear stress, area average of wall shear and velocity profiles. Results represented a relatively similar wall shear stress and velocity profile values for high velocity regions such as the parent artery. As for the aneurysm dome, there is a noticeable difference between the two viscosity models due to low velocity in this region. It is concluded that the Newtonian assumption is valid for high velocity regions due to the equal coefficient of viscosity for both Newtonian and non-Newtonian models. The Newtonian fluid assumption presents error of around 45% in WSS when compared to Carreau non-Newtonian model in regions of stasis or slowly recirculating secondary vortices mainly inside the aneurysm dome. Overall, the Newtonian assumption is deemed an invalid assumption when considering the pulsatile velocity of blood.

INTRODUCTION

In recent years, computational fluid dynamics (CFD) has become a major player in the clinical field and has dramatically grown due to the developments of high-resolution angiography techniques (Robertson and Watton 2012). The engineering knowledge and medical technologies have combined to support medical physicians in diagnosing different medical cases. For instance, CFD has been used by researchers to develop a patient specific model that simulates the pumping flow in a left ventricle. This is considered a great breakthrough (Schenkel et al., 2009). Also, one of the recent application where CFD is found useful is in the aneurysm field (Robertson & Watton, 2012). The cerebral aneurysm field is the study of a localized, blood-filled balloon in the wall of a blood vessel inside the brain (Dorland, 1980). The importance of studying aneurysms is that it affects around 2% of adults and in case of rupture that can lead to death if medical intervention did not take place (Rinkel, Djibuti, Algra, & Van Gijn, 1998). Another reason that pushes for more CFD implementation in aneurysm studies is the limited medical capabilities and the difficulty to construct in-vivo experimental studies especially in small and tortuous intracranial arteries (Milner, Moore, Rutt, & Steinman, 1998; Papathanasopoulou et al., 2003). Mainly when studying aneurysms, researches are aiming to predict the causes of aneurysm initiation, the growth rate, the rupture risk and the flow alteration as a result of interventional treatment (coiling and stenting). Still more research is needed to establish the relation between aneurysm initiation, the growth rate and, the rupture.

The CFD results are valuable data that should be capable to describe different complex flow behaviour. However to get such reliable data, one needs to properly model the flow behaviour using correct boundary condition and geometry. It is reported that the variation of inflow waveform boundary condition affects the CFD results (Karmonik et al., 2010; Karmonik et. Al., 2009). Also, it is reported that geometrical properties of the parent artery in cerebral aneurysms strongly impact the CFD results (Hoi et al., 2004; Meng et al., 2006).

Currently, different assumptions in CFD modeling are used by researchers when undertaking aneurysm research. These assumptions are meant to make the simulation easier and to consume less computational time. Some researchers studied the effects of truncating the parent artery of the aneurysm and replacing it with a cylinder as mentioned in the sensitivity study conducted by Castro et al. (Castro, Putman, & Cebral, 2006). Other researchers use mean velocities instead of using a real pulsatile blood cycle (Xiang et al., 2011). Different efforts have been done to validate the CFD results using three-dimensional X-ray angiography (Ford et al., 2005) or using variations of contrast agent distribution (Sun, Groth, & Aach, 2012). In two experimental canine aneurysms, researchers (Jiang et al., 2011) reported an acceptable agreement between experimental results collected by an accelerated 4D PC-MRA and CFD results mainly for predicting the intra-aneurismal velocity fields. However, other researchers suggest that 2D pcMRi presents a much higher in-plane resolution than the 4D pcMRI (Karmonik et al., 2008).

The concentration in this study is on the appropriateness of the Newtonian viscosity model assumptions. Blood is shear thinning non-Newtonian fluid and some researchers are modeling it as a Newtonian fluid to simplify the computational simulation (Castro et al., 2006; Galdi, Rannacher et al., 2007; Karmonik et al., 2014; Valen.Sendstad & Steinman, 2014).

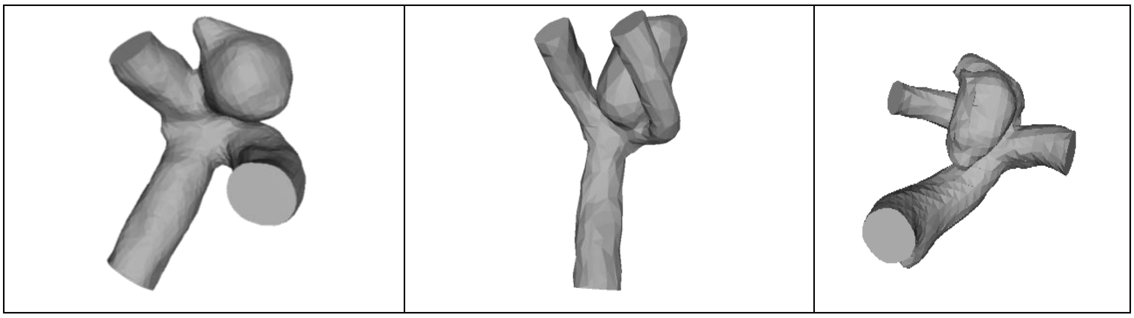


Figure 1: patient specific aneurysm models (from left patients 1, 2 and 3)

It is believed that this assumption causes a discrepancy when compared to real conditions especially for the data obtained at low velocity regions such as the aneurysm's dome. The objective of this paper is to examine the sensitivity of blood flow field, shear rate and wall shear stress (WSS) predictions using the Newtonian and non-Newtonian blood viscosity models. Wall shear stress (WSS) is one of the main pathogenic factors in the development of saccular cerebral aneurysms which plays an important role in the growth and the rupture of cerebral aneurysms. The aim is to conclude whether it is appropriate to conduct cerebral aneurysm simulations using the Newtonian viscosity assumption or not and under what conditions is this assumption valid.

PROBLEM FORMULATION

Three cases taken from Tawam hospital patient list were selected for this study, two anterior communicating artery aneurysms and one carotid bifurcation artery aneurysm (identified as patient 1, patient 2, and patient 3). Patient 1 is a small ruptured A-COM aneurysm in a 49-year-old woman; Patient 2 is also a ruptured A-COM aneurysm in a 48-year-old man; and Patient 3 is a ruptured left carotid bifurcation aneurysm in a 42-year-old woman. Acquisition of three-dimensional angiography (3DRA) images of the patients' aneurysms was done using Philips Clarity 3D work station. Figure 1 shows the three aneurysms used in this study.

A realistic inlet velocity model to use is the transient profile suggested by Sinnott et al. (Sinnott, Cleary, & Prakash, 2006) with a sine wave having a maximum blood velocity of 0.5 m/s, the period of each cycle is 0.5 s and the simulation is done for one cycle. No slip boundary conditions applied at the artery walls and the walls are assumed as rigid surfaces. Figure 2 shows the average inlet velocity boundary condition as it varies with time during a period of 0.5 s (i.e. 120 pulse/min).

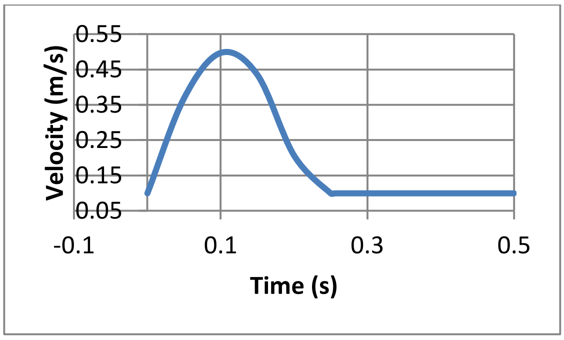


Figure 2: pulsatile velocity vs. time

NUMERICAL SIMULATION SETUP

The 3D images which are collected as described in the earlier section are refined and truncated using a commercial CAD software package. A quadric edge collapse simplification is applied to the 3D images to reduce the number of geometric faces. Later on, the images are exported as a solid 3D part to ANSYS design modeller. After importing the CAD file by ANSYS-FLUENT, the 3D CAD image is meshed using a 3D pyramid cell. A mesh independence study is implemented to make sure that the output values are independent of mesh size. Figure 3 shows the maximum WSS at Aneurysm dome vs. Number of grid elements.

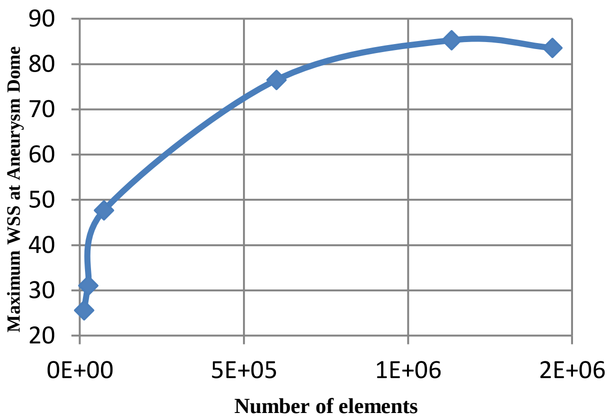


Figure 3: Maximum WSS vs. Mesh elements

The model for each patient is divided into three sections of interest, the aneurysm dome, the parent artery and the outlet arteries which is referred to as "rest". The final mesh used in this study contains around 10^6 elements for patients 1, 2 and 3 with minimum mesh size of 0.08 mm. The hemodynamic behaviour of blood is determined by numerically solving Navier-Stokes equations using a finite volume discretization and laminar flow model (Reynolds number < 700). The QUICK scheme is used to approximate the momentum equation while pressure-velocity coupling is realized using the SIMPLEC method with the standard under-relaxation parameters. The outlet back flow is specified using normal velocity at the outlet. The maximum convergence residuals criteria are set to 10^{-5} for all variables. Two viscosity models are considered in this study, the Newtonian model with viscosity of 0.0035 kg/m/s and the non-Newtonian Carreau model with constants proposed by Siebert et al, where the time constant was set to 3.313 s, the Power-Law Index was set to 0.3568, the zero shear viscosity was set to 0.056 kg/m/s and the infinite shear viscosity was set to 0.0035 kg/m/s.

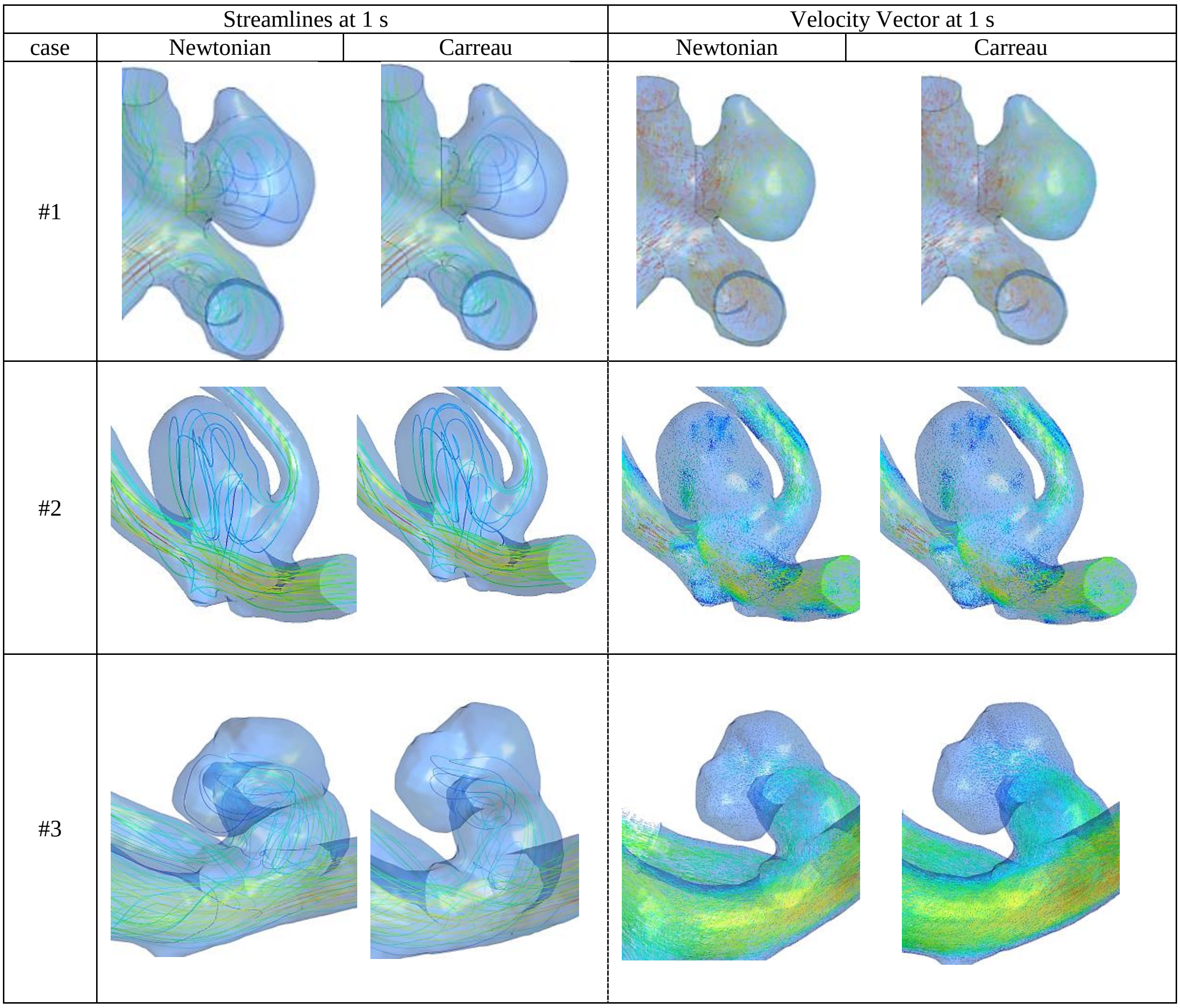


Figure 4: Velocity vectors and streamlines for 3 patients

(Siebert & Fodor, 2009). For the Newtonian model the regular form of Navier-Stokes equation is solved with constant viscosity value and for the non-Newtonian model the Carreau fluid form is considered for the viscosity term in Navier-Stokes equations. Blood is considered as an incompressible fluid, and the continuity equation is solved accordingly.

RESULTS AND DISCUSSION

Velocity

In all three cases there is a relatively similar pattern when comparing between the Carreau and the Newtonian viscosity models for velocity. Velocity vectors are mainly dependent on the geometry of each aneurysm. Figure 4 Shows the velocity vectors and streamlines for 3 different patients of cerebral aneurysms. In these figures red represents the highest value for velocity and blue represents the lowest values of velocity. It is noticed that velocities are slower at the dome and low circulation occurs in some of the cases. As for the velocity vectors, for the three cases it is noticed that the flow patterns are alike. Patients 1 and 2 exhibit higher velocities at the aneurysm dome due to the neck size as well as geometry of the aneurysm that allows more flow to enter. On the other hand, patient 3 shows slower velocity patterns inside the aneurysm which allows blood to reach near zero velocities which leads to weaker low circulation.

Wall Shear Stress

Figure 5 shows the variation of WSS for the three cases for the Carreau and Newtonian viscosity models. It is noticed that corresponding to the velocity figure, areas with low velocity exhibit low WSS values and vice-versa. At areas with low velocities a considerably high difference in values is noticed in WSS between the Carreau and the Newtonian model. Major differences between the viscosity models are always noticed at the dome region. For patient 1, the WSS variation in WSS is noticed nearby the blip and it is also noticed at the neck of the aneurysm. For patient 2, a considerable amount of variation is noticed in the WSS at the dome while a smaller amount of variation is visible at the neck. This is also visible in figure 7 as it shows the normalized WSS (maximum, minimum and area averaged) for the two models at two different time instances. It is noticed that at the first instant (i.e. $t = 0.05$ s) at the dome great difference is evident between the two viscosity models. Looking at the second instant when $t = 0.1$ s for the same area we can see that the difference in readings is minor. This is evident in all readings for the first instant at the dome. The large difference between the two viscosity models always

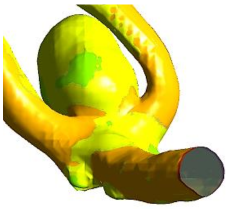
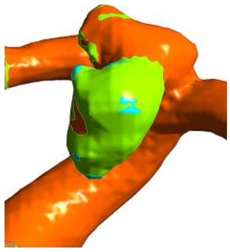
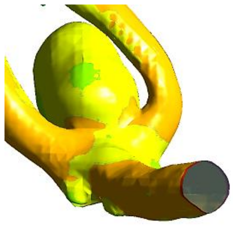
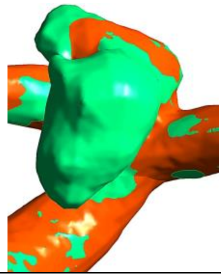
WSS at 1s	#1	#2	#3
Newtonian			
Non-Newtonian			

Figure 5: WSS for 3 patients

occurs when looking at the minimum WSS section because it happens at minimum speeds hence minimum strain rate and maximum viscosity difference. At the parent artery and at the rest of the arteries it is noticed that WSS for the two models is relatively the same due to high magnitude of velocities at these areas.

To further compare between the viscosity models and to consider the pulsatile velocity of blood figure 6 is plotted to show the Maximum WSS versus time step for both viscosity models. The difference between Newtonian and non-Newtonian models results is quantified using the following error calculation:

$$Difference\ \% = \left|\frac{WSS_{Newtonian} - WSS_{non-Newtonian}}{WSS_{non-Newtonian}}\right| \quad (1)$$

At 0.1 s of the simulation the maximum velocity occurs with a value of 0.5 m/s. While between 0.25 and 0.5 seconds at the diastole, the velocity is minimum and stays at 0.1 m/s. It is noticed that at 0.05 s the difference between the two viscosity models is around 55% which reduces along the systole until it reaches the peak time step of 0.1 s with velocity at 0.5 m/s and a minimum error of 5% between both tested viscous models. Newtonian model underestimates the WSS when velocity starts becoming constant at the diastole (between 0.25-0.5 s) with a difference of around 45%. As expected, the results presented a high variation between the two viscosity models especially at points where velocities are low. The overall difference between the viscosity readings is around 45% overall which is deemed unacceptable.

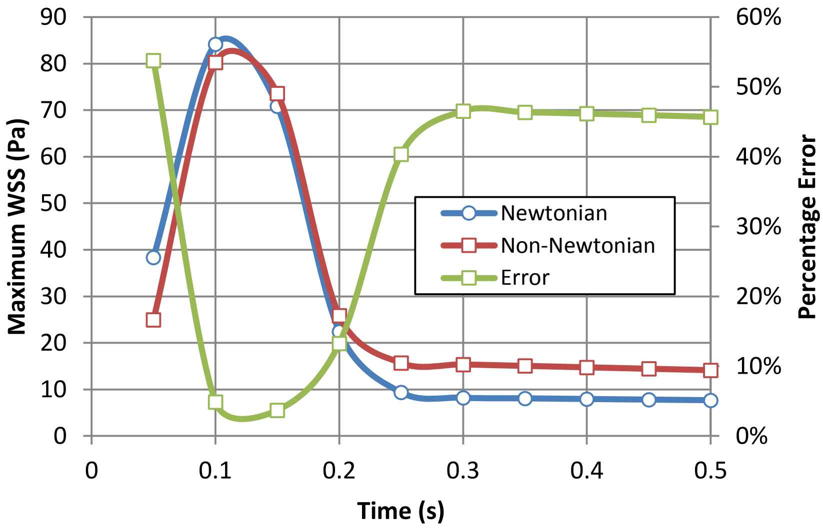


Figure 6: Maximum WSS for viscosity models and differences

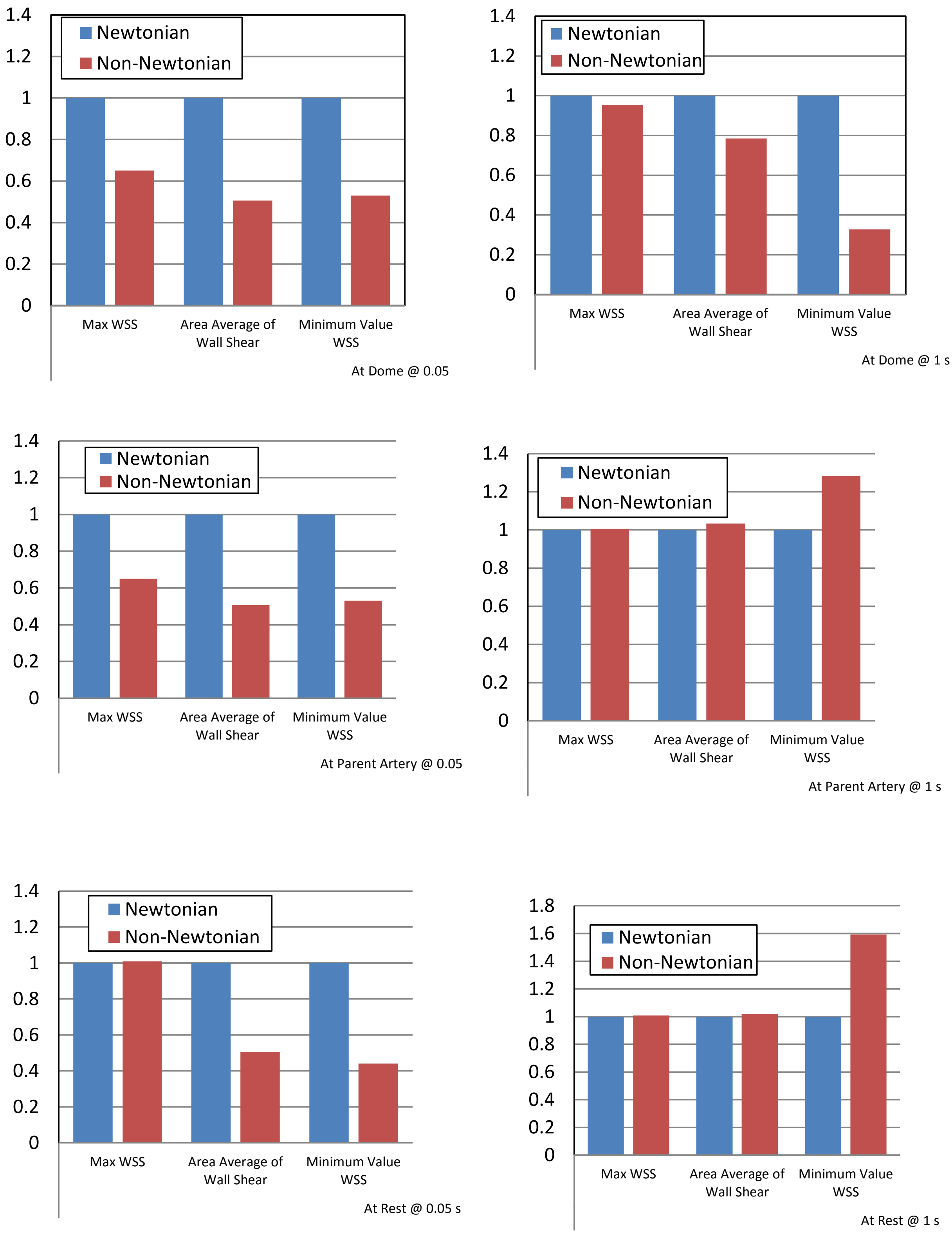


Figure 7: Normalized WSS at different locations and instances

CONCLUSION

The Newtonian model assumption is not appropriate to be used when dealing with blood flow in a low circulation zone such as the aneurysm dome. The Newtonian results are over estimating WSS at some regions and under estimating WSS in other regions. Mainly when considering the pulsatile velocity of blood the Newtonian model presents a difference between models up to 55% when compared to the actual case. It is recommended to use realistic viscosity models for blood such as the Carreau model or other power non-Newtonian models.

REFERENCES

CASTRO, M., PUTMAN, C., & CEBRAL, J. (2006). Computational fluid dynamics modeling of intracranial aneurysms: effects of parent artery segmentation on intra-aneurysmal hemodynamics. *American Journal of Neuroradiology, 27*(8), 1703-1709.

DORLAND, W. A. N. (1980). *Dorland's medical dictionary*: Saunders Press.

FORD, M. D., STUHNE, G. R., NIKOLOV, H. N., HABETS, D. F., LOWNIE, S. P., HOLDSWORTH, D. W., & STEINMAN, D. A. (2005). Virtual angiography for visualization and validation of computational models of aneurysm hemodynamics. *Medical Imaging, IEEE Transactions on, 24*(12), 1586-1592.

GALDI, G. P., RANNACHER, R., ROBERTSON, A. M., & TUREK, S. (2007). *Hemodynamical Flows: Modeling, Analysis and Simulation (Oberwolfach Seminars)*: Birkh\\&\\#228;user Basel.

HOI, Y., et. Al. (2004). Effects of arterial geometry on aneurysm growth: three-dimensional computational fluid dynamics study. *Journal of neurosurgery, 101*(4), 676-681.

JIANG, J., JOHNSON, K., VALEN-SENDSTAD, K., MARDAL, K.-A., WIEBEN, O., & STROTHER, C. (2011). Flow characteristics in a canine aneurysm model: a comparison of 4D accelerated phase-contrast MR measurements and computational fluid dynamics simulations. *Medical physics, 38*(11), 6300-6312.

KALLMES, D. F. (2012). Point: CFD—computational fluid dynamics or confounding factor dissemination. *American Journal of Neuroradiology, 33*(3), 395-396.

KARMONIK, C., YEN, C., DIAZ, O., KLUCZNIK, R., GROSSMAN, R. G., & BENNDORF, G. (2010). Temporal variations of wall shear stress parameters in intracranial aneurysms—importance of patient-specific inflow waveforms for CFD calculations. *Acta neurochirurgica, 152*(8), 1391-1398.

KARMONIK, C., YEN, C., GROSSMAN, R. G., KLUCZNIK, R., & BENNDORF, G. (2009). Intra-aneurysmal flow patterns and wall shear stresses calculated with computational flow dynamics in an anterior communicating artery aneurysm depend on knowledge of patient-specific inflow rates. *Acta neurochirurgica, 151*(5), 479-485.

KARMONIK, C., ZHANG, Y. J., DIAZ, O., KLUCZNIK, R., PARTOVI, S., GROSSMAN, R. G., & BRITZ, G. W. (2014). Magnetic resonance imaging as a tool to assess reliability in simulating hemodynamics in cerebral aneurysms with a dedicated computational fluid dynamics prototype: preliminary results. *Cardiovascular diagnosis and therapy, 4*(2), 207.

MENG, H., WANG, Z., KIM, M., ECKER, R., & HOPKINS, L. (2006). Saccular aneurysms on straight and curved vessels are subject to different hemodynamics: implications of intravascular stenting. *American Journal of Neuroradiology, 27*(9), 1861-1865.

MILNER, J. S., MOORE, J. A., RUTT, B. K., & STEINMAN, D. A. (1998). Hemodynamics of human carotid artery bifurcations: computational studies with models reconstructed from magnetic resonance imaging of normal subjects. *Journal of vascular surgery, 28*(1), 143-156.

PAPATHANASOPOULOU, P., ZHAO, S., KÖHLER, U., ROBERTSON, M. B., LONG, Q., HOSKINS, P., . . . MARSHALL, I. (2003). MRI measurement of time-resolved wall shear stress vectors in a carotid bifurcation model, and comparison with CFD predictions. *Journal of Magnetic Resonance Imaging, 17*(2), 153-162.

RINKEL, G. J., DJIBUTI, M., ALGRA, A., & VAN GIJN, J. (1998). Prevalence and risk of rupture of intracranial aneurysms a systematic review. *Stroke, 29*(1), 251-256.

ROBERTSON, A., & WATTON, P. (2012). Computational fluid dynamics in aneurysm research: critical reflections, future directions. *American Journal of Neuroradiology, 33*(6), 992-995.

SCHENKEL, T., MALVE, M., REIK, M., MARKL, M., JUNG, B., & OERTEL, H. (2009). MRI-based CFD analysis of flow in a human left ventricle: methodology and application to a healthy heart. *Annals of biomedical engineering, 37*(3), 503-515.

SIEBERT, M. W., & FODOR, P. S. (2009). *Newtonian and non-newtonian blood flow over a backward-facing step–a case study.* Paper presented at the Proceedings of the COMSOL Conference, Boston.

SINNOTT, M., CLEARY, P. W., & PRAKASH, M. (2006). *An investigation of pulsatile blood flow in a bifurcation artery using a grid-free method.* Paper presented at the Proc. Fifth International Conference on CFD in the Process Industries.

SUN, Q., GROTH, A., & AACH, T. (2012). Comprehensive validation of computational fluid dynamics simulations of in-vivo blood flow in patient-specific cerebral aneurysms. *Medical physics, 39*(2), 742-754.

VALEN-SENDSTAD, K., & STEINMAN, D. (2014). Mind the gap: Impact of computational fluid dynamics solution strategy on prediction of intracranial aneurysm hemodynamics and rupture status indicators. *American Journal of Neuroradiology, 35*(3), 536-543.

XIANG, J., TREMMEL, M., KOLEGA, J., LEVY, E. I., NATARAJAN, S. K., & MENG, H. (2011). Newtonian viscosity model could overestimate wall shear stress in intracranial aneurysm domes and underestimate rupture risk. *Journal of neurointerventional surgery*, neurintsurg-2011-010089.