Comparison of Newtonian and Non-Newtonian Blood Flow in an Ascending Aortic Aneurysm

Aleksandra Petuchova[1], Algirdas Maknickas[1]

[1]Department of Biomechanical Engineering, Faculty of Mechanics, Vilnius Gediminas Technical University, Vilnius, Lithuania

Abstract

This work aimed to perform a numerical study of aortic hemodynamics and evaluate both Newtonian and non-Newtonian blood flow parameters in an ascending aortic aneurysm model.

An aortic model was reconstructed from a medical computed tomography (CT) image, and finite element method laminar blood flow modelling was performed using different blood parameters. The inflow boundary conditions were defined as a flow profile, and the outlet boundary conditions were defined as the pressure at each outlet. The first simulation was calculated by considering blood as a Newtonian fluid, while in the second simulation, using the Carreau model, blood was assumed to be a non-Newtonian fluid.

The results showed that average systolic and diastolic velocities were 2% and 9% higher, respectively, for the non-Newtonian fluid. In addition, the wall shear stress (WSS) values on the surface of the aneurysm were 30% higher during systole in the non-Newtonian simulation, while the average WSS on the artery surface in diastole was 20% higher for the Newtonian fluid.

1. Introduction

Numerical modelling of the cardiovascular system enables investigation of the properties of blood flow, simulation of the mechanical workings of the vascular system, and observation of hemodynamic models, pressure gradients, stress distributions and deformation of blood vessel walls under certain conditions [1,2]. Patient-specific computational fluid dynamic models are used to address aortic aneurysms, dissection, atherosclerosis and heart valve pathologies [3–6].

Computational vascular modelling consists of several steps: data acquisition or reconstruction, identification and description of boundary conditions, choice of computational model and equipment, and computation, processing and display of the results [7].

Newtonian fluids exhibit constant viscosity and zero shear coefficient at zero shear stress. This means that the shear coefficient is directly proportional to shear stress, and the ratio of shear stress to shear rate is constant throughout the fluid. Non-Newtonian liquids can change viscosity when a force is applied, becoming more fluid or more solid [8]. Human blood has non-Newtonian properties, becoming less viscous at high shear rates, for example, with increased flow, such as during exercise or at peak systole. Blood viscosity increases when shear rates decrease with increasing vascular diameter or with low flow, such as downstream of an obstacle or during diastole [9].

One of the main challenges of describing blood conditions is to model blood as either a Newtonian or non-Newtonian fluid, as the differing fluid dynamics affect the distribution of stresses on the aortic wall. Studies to evaluate the effect of these types of distributions on the risk of rupture of an aortic aneurysm are required. The current investigation aimed to evaluate and compare the two types of fluid using the same boundary conditions in an ascending aorta model.

2. Materials and Methods

An ascending aorta model was constructed from a medical CT image using SimVascular software [10]. Finite element method (FEM) laminar blood flow modelling of six cardiac cycles was performed using different blood parameters.

Navier-Stokes equations for incompressible fluids were used for blood flow simulations generated by the COMSOL 5.5 CFD tool [11]. The model was discretized using FEM to perform the calculations. The mesh parameters chosen for the simulation were designed for fluid dynamics problems with a boundary wall layer.

Inflow boundary conditions were a fluid flow waveform that simulated the flow of the aortic valve. As the blood flow in the human body is pulsatile, a constant velocity at the outlet does not simulate the actual flow; in this case, flow was indicated as a periodic change profile (Figure 1). Assuming a heart rate of 60 beats per minute, the duration

 ISSN: 2325-887X DOI: 10.22489/CinC.2022.085

of each period was 1 s, yielding a blood flow of Q = 5.5 l/min under these conditions. The first cardiac cycle started at 0.5 s, and outlet pressure varied with time according to a given function; pressure was 125.4 mmHg]·f(t) in the branches of the aortic arch and 125.1[mmHg]·f(t) in the descending aorta.

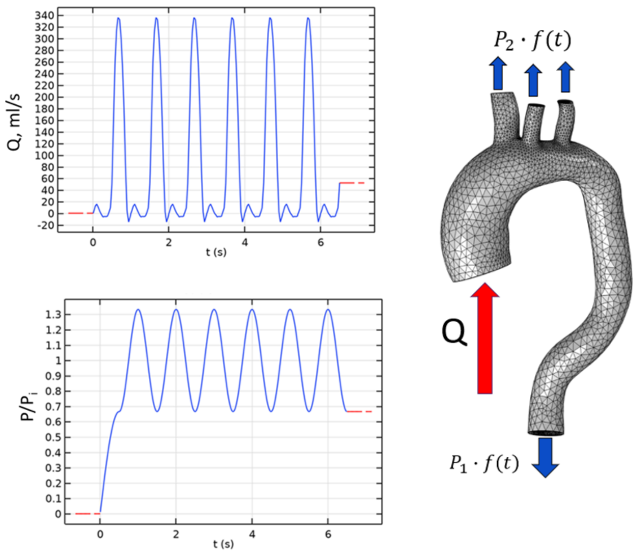

Figure 1. Boundary conditions at the inlet (Q) and outlet (P_i*f(t)) in the discretized ascending aorta model

Three-dimensional Navier-Stokes equations for an incompressible fluid were applied to simulate blood flow using the CFD tool COMSOL Multiphysics:

$$\rho \frac{\partial v}{\partial t} + \rho(v \cdot \nabla)v = \nabla\tau \quad in\, \Omega^t, \qquad \forall t \in I, \tag{1.1}$$

$$\nabla \cdot v = 0 \quad in\, \Omega^t, \qquad \forall t \in I, \tag{1.2}$$

where I is the time interval and ρ is the blood density. The unknown variables are blood velocity and pressure, which depend on Cauchy stress.

In the case of Newtonian fluids, the general stress tensor is simply defined:

$$\tau(v, p) = -PI + 2\eta\, D(v), \tag{1.3}$$

Here, μ is the dynamic viscosity of the blood, and D is the strain rate tensor given by:

$$D(v) = \frac{(\nabla v) + (\nabla v)^T}{2}. \tag{1.4}$$

A common stress tensor for a non-Newtonian fluid:

$$\tau(v, p) = -PI + 2\eta(\dot{\gamma})D(v). \tag{1.5}$$

Viscosity is variable and depends on the shear rate $\gamma = \sqrt{2tr(D^2)}$.

The second simulation treated the blood as a Newtonian fluid using the Carreau model:

$$\eta = \eta_\infty + \left(\eta_0 - \eta_\infty\right)[1 + (\lambda\dot{\gamma})^2]^{\frac{n-1}{2}}, \tag{1.6}$$

where the viscosity at a high shear rate (η∞) is equal to the value of the Newtonian model (0.0035 Pa s), and at zero shear η_0 = 0.056 Pa s; λ = 3.313 s and n = 0.3568 [12]; blood density, ρ = 1060 kg/m^3.

The simulation ran for 6 cardiac cycles (giving a total time of 6.5 s as the first cardiac cycle started at 0.5 s), while the number of steps was 65 and the step size was 0.1 s; results were recorded at each step.

3. Results

The use of different fluid parameters in blood flow models with the same boundary conditions enabled haemodynamic parameters relevant for the assessment of cardiovascular function to be investigated, namely blood flow velocity and wall shear stress (WSS).

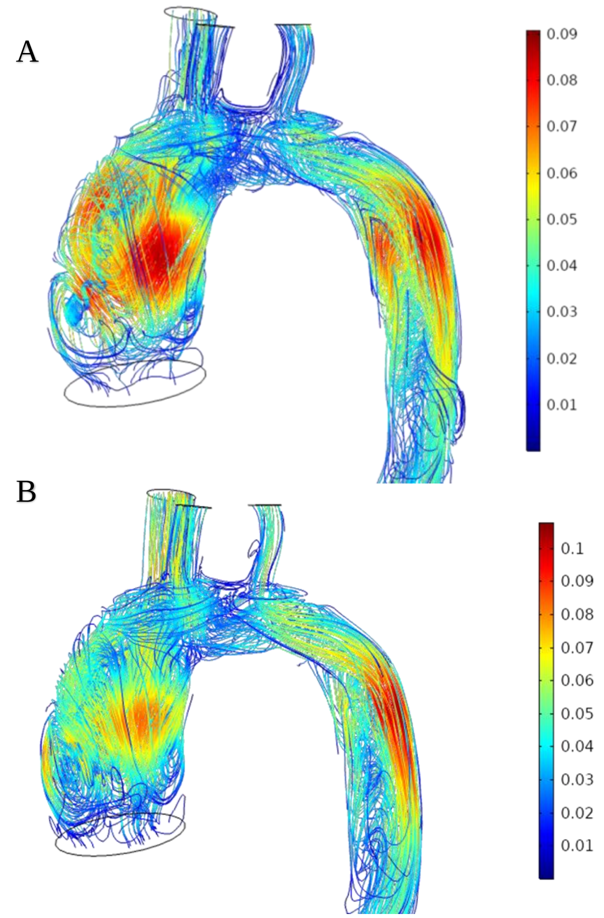

Figure 2. Distribution of blood flow velocities in an aneurysm in diastole phase: (A) non-Newtonian fluid, (B) Newtonian fluid

In the non-Newtonian model, peak blood velocity during systole (0.82 m/s) was 2% higher than in the Newtonian model (0.806 m/s). In diastole, the non-Newtonian fluid, blood velocity varied from 0 to 0.09 m/s, while for the Newtonian fluid the range was 0 to 0.105 m/s. (Figure 2 A,B) Maximum flow velocity values occurred in the aneurysm zone and the descending aorta. Cross-sections of the aorta at the site of the aneurysm show the differences in the trajectories of the flow lines (Figure 3 A,B). To compare blood velocities within the areas delineated by the sections, another section was made on each section (a straight line), and a graph of the distribution of values on the straight line was plotted (Figure 3 C).

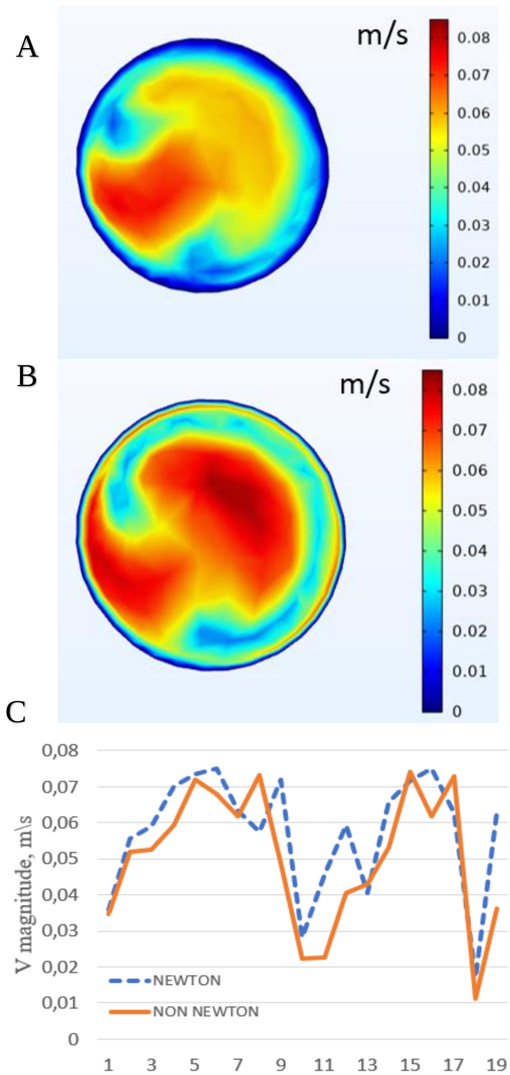

Figure 3. Distribution of blood flow velocities in an aneurysm in diastole phase in section plot: (A) non-Newtonian fluid, (B) Newtonian fluid, (C) section comparison plot.

Comparison of the distribution of WSS on the surface of the aneurysm also revealed some differences: in the non-Newtonian simulation, systolic WSS was 0.3 Pa, while for the Newtonian simulation, WSS was 0.2 Pa. The highest time-average one-cycle wall shear stress (TA WSS) value was the same for both non-Newtonian and Newtonian simulations (2 Pa) (Figure 4).

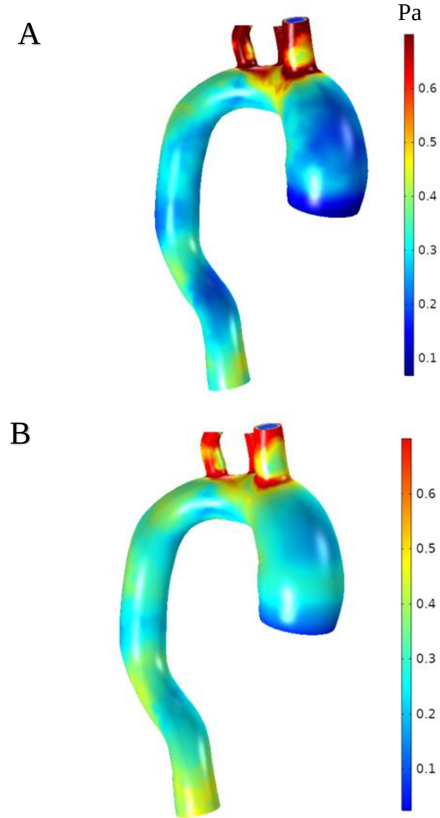

Figure 4. Aortic time-average wall shear stresses over the cardiac cycle: (A) non-Newtonian fluid, (B) Newtonian fluid.

3. Discussion

Comparison of Newtonian fluid simulation values using the same geometry model and inlet boundary conditions but different simulation software (SimVascular) in our previous study showed differing results for both systole and diastole. Previously obtained values were 0 to 1.18 m/s in systole and 0 to 0.24 m/s in diastole [13], whereas the maximum systolic velocities obtained using SimVascular were 1.46- and 2.6-fold higher for systole and diastole, respectively.

In addition, WSS values at systole obtained using the same geometry but with SimVascular software were the same as in the present study; however, the distribution of WSS values was different. In addition, TA WSS values were higher in the previous simulation. The aneurysm model with ascending aorta, aortic arch and descending aorta yielded WSS values ranging from 1.5 Pa to 3.3 Pa [13].

Other researchers have obtained WSS of 15.29 Pa in a non-Newtonian simulation and 16 Pa using a Newtonian blood model [14]. Another study obtained TA WSS values ranging from 0.128 to 12 Pa (the highest values were observed in the branches of the aortic arch) and 0.45 Pa in the ascending aorta [14]. A model of a non-Newtonian fluid in the ascending aorta resulted in maximum WSS values of 7.5 Pa during systole and 1.6 Pa during diastole [15]. Furthermore, using the non-Newtonian Carreau model with ANSYS Fluent software yielded a WSS of 32 Pa at systole (maximum blood velocity of 1.6 m/s) in the ascending aneurysm zone [16].

Simao et al., 2017 suggested that higher or lower WSS values can lead to dysfunction of the inner aortic layer, potentially contributing to the progression of atherosclerosis. Extremely high WSS values are associated with rearrangement of the vascular structure responsible for the initiation and progression of aneurysms, while regions of low WSS are correlated with atherosclerosis progression. WSS is related to blood flow in the artery and is also dependent on the size and geometry of the aorta. Areas with low and fluctuating WSS are vulnerable to the development of pathologies such as aneurysms and vascular dissections [6].

5. Limitations

The main limitation of the work was that the modelling used a rigid aortic wall. In addition, non-individualized patient parameters were used: all values were taken from the scientific literature.

6. Conclusions

Comparing blood flow models using Newtonian and non-Newtonian fluids, the following general conclusions were made:

1. Mean systolic and diastolic velocities were 2% and 9% higher, respectively, for the non-Newtonian fluid.
2. WSS on the surface of the aneurysm was 30% higher for the non-Newtonian fluid during systole.
3. Mean WSS on the arterial surface during diastole was 20% higher for the Newtonian fluid.
4. Non-Newtonian fluid modelling consumes more computational resources and takes 60% longer.

References

[1] Soudah E, Ng EYK, Loong TH, Bordone M, Pua U, Narayanan S. CFD modelling of abdominal aortic aneurysm on hemodynamic loads using a realistic geometry with CT. Comput Math Methods Med. 2013;2013.

[2] Caballero AD, Laín S. A Review on Computational Fluid Dynamics Modelling in Human Thoracic Aorta. Vol. 4, Cardiovascular Engineering and Technology. Springer Science and Business Media, LLC; 2013. p. 103–30.

[3] Morris PD, Narracott A, Von Tengg-Kobligk H, Soto DAS, Hsiao S, Lungu A, et al. Computational fluid dynamics modelling in cardiovascular medicine. Heart. 2016;102(1):18–28.

[4] Dwidmuthe PD, Mathpati CS, Joshi JB. CFD Simulation of Blood Flow inside the Human Artery: Aorta. 2018;(January 2018):1–5. Available from: https://www.researchgate.net/publication/330900048

[5] Polanczyk A, Piechota-Polanczyk A, Domenig C, Nanobachvili J, Huk I, Neumayer C. Computational fluid dynamic accuracy in mimicking changes in blood hemodynamics in patients with acute type IIIb aortic dissection treated with TEVAR. Appl Sci. 2018;8(8):1–14.

[6] Simao M, Ferreira JM, Tomas AC, Fragata J, Ramos HM. Aorta ascending aneurysm analysis using CFD models towards possible anomalies. Fluids. 2017;2(2):1–15.

[7] Gray RA, Pathmanathan P. Patient-specific cardiovascular computational modeling: Diversity of personalization and challenges. J Cardiovasc Transl Res. 2018;11(2):80–8.

[8] Gustavo V. Newtonian and Non-Newtonian Flow. Vol. II, Food Engineering. 2011. p. 1–10.

[9] Baskurt OK, Meiselman HJ. Blood Rheology and Hemodynamics. Semin Thromb Hemost [Internet]. 2003 Oct 21 [cited 2022 May 27];29(5):435–50. Available from: http://www.thieme-connect.de/products/ejournals/html/10.1055/s-2003-44551

[10] Updegrove A, Wilson NM, Merkow J, Lan H, Marsden AL, Shadden SC. SimVascular: An Open Source Pipeline for Cardiovascular Simulation. Ann Biomed Eng 2016 453 [Internet]. 2016 Dec 8 [cited 2022 May 23];45(3):525–41. Available from: https://link.springer.com/article/10.1007/s10439-016-1762-8

[11] William T V. CFD Module User 's Guide. CFD Modul User's Guid [Internet]. 2017;1–710. Available from: https://doc.comsol.com/5.3/doc/com.comsol.help.cfd/CFDModuleUsersGuide.pdf

[12] Siebert MW, Fodor PS. Newtonian and Non-Newtonian Blood Flow over a Backward- Facing Step – A Case Study. Excerpt from Proc COMSOL Conf 2009 Bost. 2009;5.

[13] Petuchova A, Maknickas A. Computational analysis of aortic haemodynamics in the presence of ascending aortic aneurysm. Technol Heal Care. 2021;30:1–14.

[14] Qiao Y, Fan J, Ding Y, Zhu T, Luo K. A primary computational fluid dynamics study of pre-and post-tevar with intentional left subclavian artery coverage in a type b aortic dissection. J Biomech Eng. 2019;141(11):1–8.

[15] Laín S, Caballero AD. Simulación transitoria de la dinámica del flujo sanguíneo en la aorta torácica. Ing e Investig. 2017;37(3):92–101.

[16] Campobasso R, Condemi F, Viallon M, Croisille P, Campisi S, Avril S. Evaluation of peak wall stress in an ascending thoracic aortic aneurysm using FSI simulations: Effects of aortic stiffness and peripheral resistance. arXiv. 2019;1–32.

Address for correspondence:

Aleksandra Petuchova.
Department of Biomechanical Engineering, Faculty of Mechanics, Vilnius Gediminas Technical University, Basanavičiaus str. 28, 06227 Vilnius, Lithuania.
aleksandra.petuchova@vilniustech.lt