Towards quantifying the impact of blood rheology model on shear stress estimates throughout the cardiac cycle

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1. Introduction

Blood is a shear-thinning fluid, which flows only when subject to stresses greater than a certain yield stress. This rheological behaviour arises from the presence of red blood cells (RBC) suspended in a medium known as blood plasma (mostly made of water). Despite this fact, a large body of literature concerning haemodynamics characterises blood as a Newtonian fluid under the assumption that in large arteries the shear rate is large enough for the viscosity to remain constant [1]. More recently, it has been suggested that, in intracranial aneurysms, this simplification overestimates wall shear stress (WSS) (see, *e.g.* [2, 3]). Therefore, applications requiring accurate shear stress (SS) estimates (*e.g.* those concerning vascular remodelling and biomechanics) will suffer from modelling inaccuracy unless the generalised Newtonian (GN) properties of blood are taken into account.

In this work, we analyse the impact of choice of rheology model on the estimates of SS of a parallel lattice-Boltzmann haemodynamics solver (namely HemeLB). So far, similar analyses have been carried out in idealised two- and three-dimensional geometries (see *e.g.* [1]) and to a lesser extent in complex vascular networks obtained from patient-specific data (see *e.g.* [3]). Furthermore, little evidence exists about how the variation in blood peak velocity throughout the cardiac cycle affects SS distribution.

We hypothesise that the difference in SS estimated by GN and Newtonian rheology models varies considerably during the cardiac cycle. More precisely, we hypothesise that this difference becomes particularly evident during diastole, when the velocity magnitude is lowest.

2. Materials and Methods

HemeLB

HemeLB [4] is a software platform for modelling and simulation of blood flow in sparse geometries. It is comprised of tools for geometrical model generation, simulation on massively parallel architectures, real-time visualisation and steering, and data post processing. To date, HemeLB has been successfully applied to the simulation of blood flow in healthy brain vasculature as well as in the presence of intracranial aneurysms. Particular attention has been paid to obtaining and presenting simulation results in a clinically meaningful way. HemeLB uses the lattice-Boltzmann method of fluid dynamics (see, *e.g.* [5]) since it allows efficient implementations in large-scale High Performance Computing infrastructures. For this work, we developed an extension of HemeLB's LBGK collision operator in order to accommodate both Newtonian and GN rheology models. The no-slip boundary condition at the walls was implemented with the bounce-back rule [5].

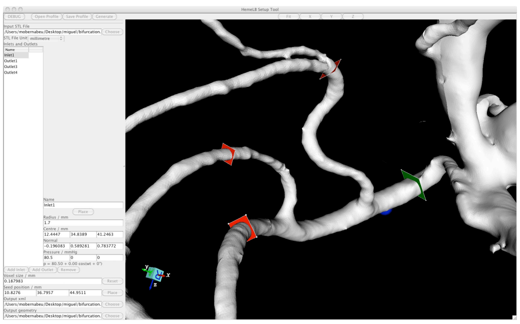

Figure 1: HemeLB's setup tool. Green/red planes define the simulation inlet/outlets, respectively.

Vascular network geometry generation

The geometry used in this work (see Fig. 1) is a subset of a computational model of intracranial vasculature reconstructed from rotational angiography scans of a patient presenting an intracranial aneurysm. It corresponds to a section of the middle cerebral artery (MCA) in the vicinity of the internal carotid artery. The main geometrical features of the model are: i) vessels of variable diameter, including narrowing, ii) two bifurcations, and iii) vessel bending.

Generalised Newtonian rheology

The Carreau-Yasuda (CY) model is broadly used to describe the shear-thinning behaviour of blood [6]. In this model, dynamic viscosity, η, is given as a function of shear-rate, $\dot{\gamma}$, by:

$$\eta(\dot{\gamma}) := \eta_\infty + (\eta_0 - \eta_\infty)\left(1 + (\lambda\dot{\gamma})^a\right)^{\frac{n-1}{a}}$$

where a, n, and λ are empirically determined to fit a curve between regions of constant η_∞ and η_0. This model defines three different regimes: a Newtonian region with η_∞ for low shear-rate, followed by a shear-thinning region where η decreases with the shear-rate, finally once η_0 is reached a third Newtonian region with constant viscosity η_0 is defined for high shear-rates. In this work, we will use the values given in [6]: $\eta_\infty = 0.16$ (Pa·s), $\eta_0 = 0.0035$ (Pa·s), $\lambda = 8.2$ s, $a = 0.64$, and $n = 0.2128$ (both dimensionless).

Fig. 2 presents viscosity as a function of shear-rate for the previous model and the Newtonian model considered in this work. Carreau-Yasuda displays a smooth transition between η_∞ and η_0. Significant differences in velocity profile between the two models are expected for simulations with $\dot{\gamma}$ in the range $[10^{-3}, 10^{3}]$.

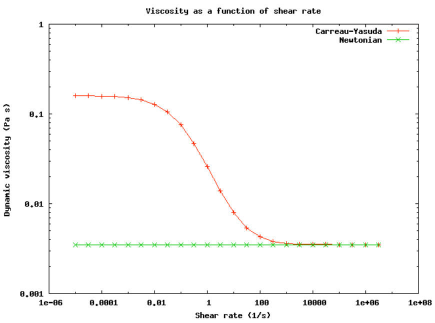

Figure 2: CY and Newtonian viscosities.

Simulation protocol

The vascular network in Fig. 1 was discretised as a regular lattice at a resolution of 0.075mm, with a total of 357,154 lattice sites. At its narrowest point the domain has, at least, 20 lattices sites along its diameter. A constant pressure difference was defined between the inlet and the outlets. The magnitude of this difference was tuned for both rheology models in order to reproduce peak velocity values corresponding to diastolic and average velocity values observed experimentally in the MCA (*i.e.* 33 cm/s and 62 cm/s [7]). The values obtained were 0.8 and 2 mmHg for Newtonian, and 0.9 and 2 mmHg for Carreau-Yasuda. The Newtonian viscosity was 0.0035 Pa·s and the simulation time step 1.71e-2 ms.

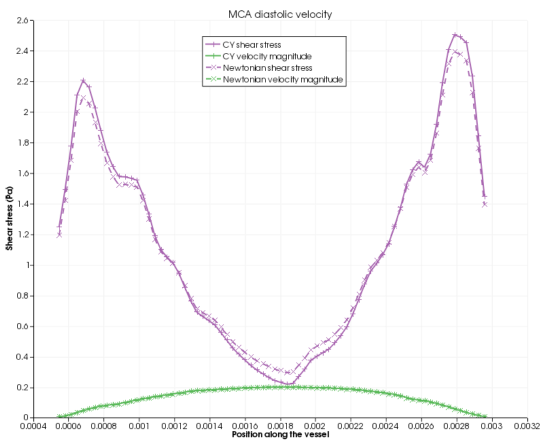

Figure 3: HemeLB's velocity and shear stress estimates during diastole.

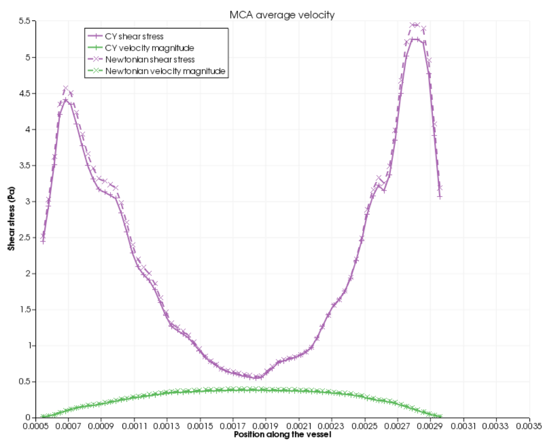

Figure 3: HemeLB's velocity and shear stress estimates for MCA average velocity.

3. Results and discussion

Simulations were run on 96 cores of HECToR, UK's national supercomputing resource. Simulations took 11-14 minutes each. Figures 3 and 4 show velocity and shear stress estimates for the two rheology models considered in this work at 33 cm/s and 62 cm/s (*i.e.* diastolic and average MCA blood velocity). The values were sampled along the diameter of a slice of the MCA between the two bifurcations in Fig. 1. It can be appreciated how differences in shear stress arise in both scenarios, although of different sign. For average MCA velocity (see Fig. 4), Newtonian SS is greater than CY (in agreement with [2, 3]). However, in diastole (see Fig. 3) the opposite is observed. This is due to the slowdown in blood velocity at the end of the cardiac cycle, which decreases shear-rate and as a consequence viscosity and shear stress increase (see Fig. 2).

4. Conclusions

In this work we presented simulations of blood flow in a cerebral vascular network reconstructed from MRI scans and corresponding to the MCA. We compared a Newtonian rheology model of blood flow with the Carreau-Yasuda GN rheology model. Since the CY model yields a large increase in viscosity for low shear-rates, we chose multiple velocity magnitudes in order to compare the models at different shear-rate regimes. The comparison was performed at diastolic and average MCA velocities observed experimentally. We hypothesised that this increase in viscosity would translate into noticeable differences in the SS estimates provided by both models, especially during diastole.

Our preliminary results show how the Carreau-Yasuda model predicts shear stress values higher than the Newtonian model during diastole, especially in the vicinity of the vessel wall. However, the opposite is observed when the average MCA blood velocity is considered. This is a direct consequence of the non-linear changes in blood viscosity throughout the cardiac cycle. These results are a first step towards the validation of the previous hypothesis. We also observed artefacts in HemeLB's shear stress calculator in the vicinity of the blood vessel (*i.e.* shear stress should be maximum in that area). We believe that this is due to the use of the bounce-back boundary condition, which is known to affect the accuracy of the WSS estimates of the LB method (see, *e.g.* [8]). In future work, we will investigate alternative no-slip boundary condition implementations in order to fully validate the previous results.

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