A Non Newtonian Model for Blood Flow behind a Flow Diverting Stent

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Abstract: Usually blood is modeled as a Newtonian fluid, neglecting its shear-thinning behavior, when calculating the blood flow in cerebral arteries and intracranial aneurysms. Treatment with flow diverting devices leads to much slower and more constant flows inside the aneurysm sack, thus the accuracy of the Newtonian model had to be reviewed. We measured the viscosity at different shear rates and used the results for a Carreau Yasuda model. We then used Comsol Multiphysics to model a bend cerebral parent vessel with a side wall aneurysm and calculate the velocity of the blood stream using both viscosity models. Against the first intuition the Newtonian model overestimates the effect of the flow diverting stent. Using the Carreau Yasuda model the velocities behind the device are about 4% to 6% higher.

Keywords: CFD simulation, blood flow, flow diverting stent, Non Newtonian, Carreau Yasuda

1. Introduction

Intracranial (also cerebral) aneurysms (see figure 1) are pathological extensions of, in most cases, arterial blood vessels within the head or brain. Most sidewall and bifurcation aneurysms are balloon-like outward bulges in the vessel wall. Due to the permanent pulsating pressure these lesions grow in size resulting in constant thinning of the vessel wall. With increasing size and many other partly unknown factors there is a significant risk of rupture followed by a bleeding (hemorrhagic stroke) [1].

Beside the surgical treatment of cerebral aneurysms, clipping, there are two endovascular methods of treatment:

1. coiling [2] and
2. using flow diverting devices [3].

Both endovascular treatment methods aim to slow down the blood flow inside the aneurysm sack to start clotting and finally lead to cicatrization of the pathological structures. Thereby the morbid vessel wall can be protected and the risk of rupture and following hemorrhage can be minimized.

When calculating the blood flow in cerebral arteries and intracranial aneurysms usually the blood is modeled as a Newtonian fluid. This simplification is based on the assumption, that the shear-thinning behavior of blood can be neglected due to the very high velocities and shear rates [5].

The flow past such endovascular devices is much slower and more constant leading to much lesser shear rates. Thus

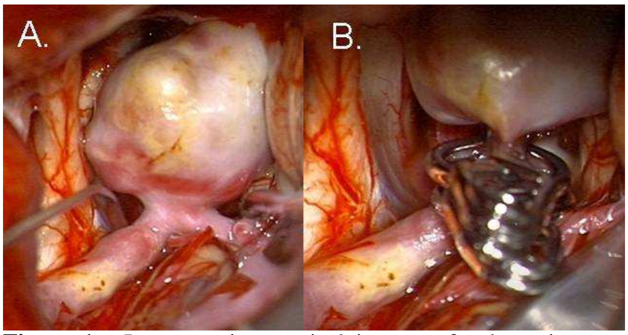

Figure 1. Intraoperative surgical images of a large intracranial aneurysm before treatment (A) and after successful clipping (B) [4].

the assumption of a nearly constant dynamic viscosity (μ) and consequential the modeling of blood as Newtonian fluid should be reviewed.

2. Methods

To test the influence of the viscosity model on the velocity profile of the blood stream we first measured the viscosity of human blood at different shear rates to get parameters for different viscosity models. Thereafter we modeled a bend intracranial blood vessel with a side wall aneurysm and a flow diverting stent. We used our blood viscosity parameters to calculate the blood flow within the stented structure using CFD simulations and compared the results of both, the Newtonian and the non Newtonian models.

2.1 Blood Viscosity Model

The first step was to get values of the dynamic viscosity of human blood in dependency of the shear rate. Therefor blood samples of multiple human probands were taken and the viscosity was measured at a constant temperature of 37 °C and various shear rates from $0.1\,\mathrm{s}^{-1}$ to $1000\,\mathrm{s}^{-1}$ using a double gap cylinder viscosimeter.

The viscosity of blood is amongst others dependent to the shear rate, the hematocrit (HCT) and the temperature. Physiological values of HCT, the volume percentage of red blood cells, are from about 37% to 50%, depending on the sex, the age, the state of health and many other factors. Since natural values of hematocrit vary from proband to proband, the viscosity values had to be normalized. The standard physiological HCT in literature is most often set to 40%, so we used this value too to enable comparability. The normalized results of our measurements can be seen in figure 2.

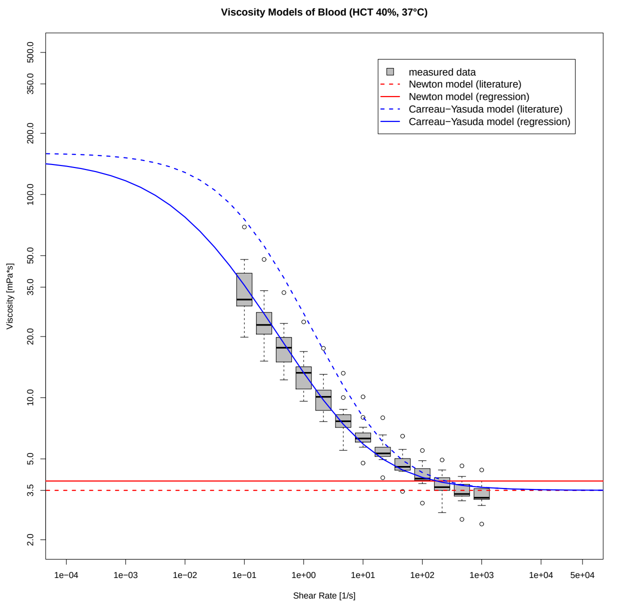

Figure 2. Viscosity of blood. The measured data is shown in grey; the Newtonian model (red) and the Carreau Yasuda model (blue) are depicted with values from literature (dashed lines) as well as with the parameters found from regression (solid lines).

The most common non Newtonian models of human blood viscosity in literature [6] are:

- Power Law,
- Carreau and Carreau Yasuda,
- Casson and
- Walburn-Schneck.

Power Law and the Carreau model are implemented in Comsol Multiphysics. Since we suspect blood to have a zero viscosity μ_0 (a plateau at very low shear rates), the Power Law, the Casson and the Walburn-Schneck model dropped out. We used the Carreau Yasuda model, which is basically a more generalized Carreau model. The dependency of the dynamic viscosity of the shear rate is modeled by equation 1.

$$\mu = \mu_\infty + (\mu_\infty + \mu_0)\left(1 + (\lambda + \dot{\gamma})^a\right)^{\frac{n-1}{a}} \tag{1}$$

The Casson model is a Casson Yasuda model with the factor $a = 2$ but we found better fitting results using other values for a. The comparatively easy Newtonian model is described by $\mu = \mu_\infty$.
The parameters a, λ, μ_0 and n for the Casson Yasuda model as well as the parameter μ_∞ for both models were found using weighted non linear least square regression on the normalized measurements. The weights represented the number of cells with the respective shear rates using a simulation of the later described stented aneurysm model with Newtonian blood flow. The results of the regressions are depicted in table 1 as well as in figure 2.

Table 1. Values for the parameters of the viscosity models derived from regression.

	Newtonian	Carreau Yasuda
$a[1]$	—	0.500
$\lambda\,[\mathrm{s}^{-1}]$	—	46.530
$\mu_0\,[\mathrm{mPa\,s}]$	—	150.000
$\mu_\infty\,[\mathrm{mPa\,s}]$	3.892	3.500
$n[1]$	—	0.342

2.2 Aneurysm Model

We used the CSG capabilities of Comsol Multiphysics to model a bend cerebral parent vessel with an inner diameter $d_{\text{vessel}} = 2.14\,\text{mm}$ as well as a side wall aneurysm with a neck length $l_{\text{neck}} = 4.41\,\text{mm}$ and a height of the dome $h_{\text{dome}} = 4.09\,\text{mm}$ (see figure 4). Also a complex flow diverting device was created. The stent consists of 16 wires with a diameter $d_{\text{wire}} = 70\,\mu\text{m}$ which are knitted to a mesh. The permeability of the device lies at round 55%, which corresponds to industrial samples. Real stents consist of even more wires with smaller cross section. Figure 3 illustrates the weaving pattern.

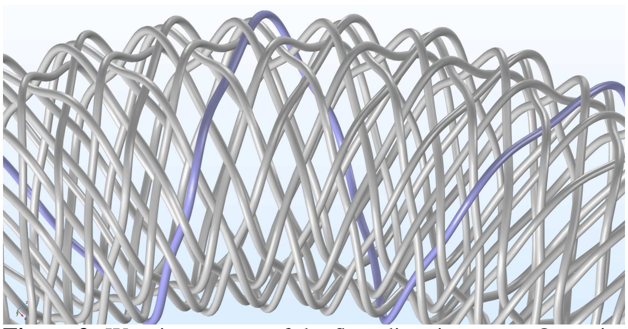

Figure 3. Weaving pattern of the flow diverting stent. One wire is highlighted in blue.

The stent was placed in the parent vessel at the position of the aneurysm and subtracted from the domain representing the blood filled vessel. The expansion of the vessel wall by the flow diverting device was also modeled as can be seen in figure 4.

2.3 CFD simulation

To model the blood flow, we used the CFD module solving the incompressible Navier Stokes equation (iNSE),

$$\overbrace{\rho\left(\frac{\partial \vec{u}}{\partial t} + \vec{u}\cdot\vec{\nabla}\vec{u}\right)}^{\text{Inertia}} = \overbrace{-\vec{\nabla}p + \mu\nabla^2\vec{u}}^{\text{Divergence of stresses}} + \underbrace{\vec{f}}_{\substack{\text{Other}\\ \text{body}\\ \text{forces}}}, \tag{2}$$

where $\vec{u}$ is the velocity, p is the pressure and $\vec{f}$ are other body forces such as gravity.

We first used a steady state solution of a single phase laminar flow problem with the Newtonian model to get suitable start values for the time dependent studies. Afterwards we performed the pulsatile flow simulations (also single phase laminar flow) using a material sweep with two blood models:

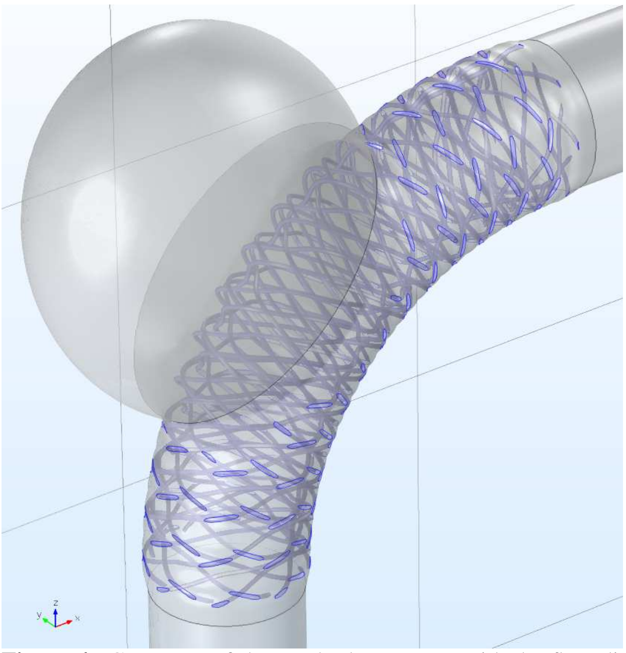

Figure 4. Geometry of the cerebral aneurysm with the flow diverting device in place. The aneurysm neck plane (slightly moved towards the dome to keep it outside the stent) as well as the expansion of the vessel due to the stent can be seen.

1. a Newtonian model with constant dynamic viscosity,
2. a Carreau Yasuda model like used in [7].

The flow profile was derived from averaging various human artery Doppler sonograms and set to a frequency of 1 Hz. The result is depicted in equation 3 and figure 5.

$$v(t) = v_{min} + (v_{max} - v_{min}) \cdot \frac{1}{10} \cdot \Big(42 \cdot (\sin(2\pi t) + |\sin(2\pi t)|) \cdot e^{-20\cdot(t - \mathrm{rnd}(t-0.5))} + 3 \cdot \sin(2\pi t)^2 + |\sin(2\pi t)| \cdot \sin(2\pi t) + \left|\sin(\pi t)^2\right|\Big) \quad (3)$$

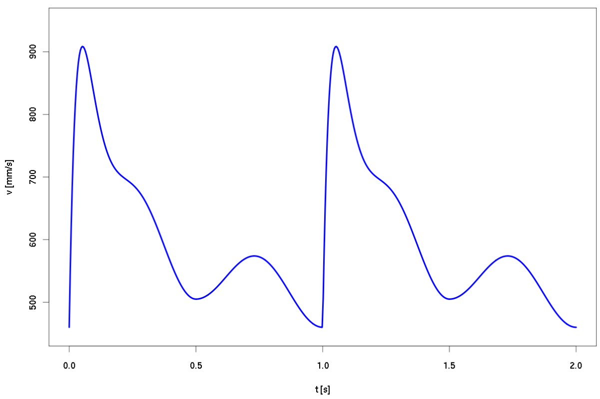

Figure 5. Flow profile derived from several Doppler sonograms.

At the end we used the "Join Dataset" functionality to compare both models to each other.

3. Results

All results are referring to the time of systole (highest value of the velocity profile) appearing at the inlet plane. The probes were taken within the aneurysm sack, which is defined to be the region above the aneurysm neck plane.

Using the Carreau Yasuda model resulted in a range of dynamic viscosity from 3.57 mPa s to 7.1 mPa s (see figure 6). Also the range of shear rates is slightly expanded and of course the velocity profiles differ (see figures 7 and 8).

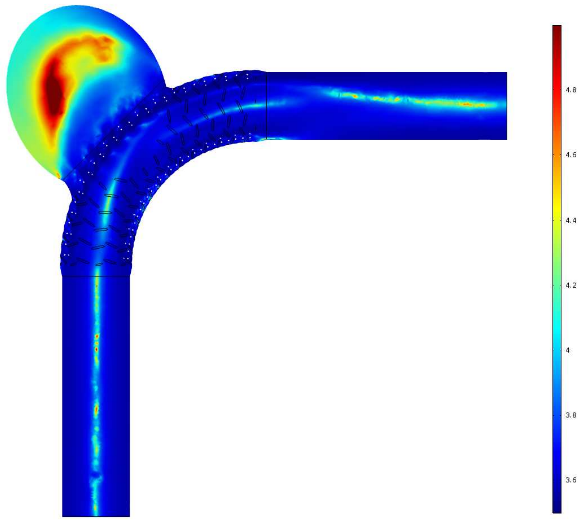

Figure 6. Viscosity profile of the Carreau Yasuda model. Values are given in mPa s.

The main differences between the two models are shown in table 2. The most interesting values are the flow rate into the aneurysm sack (of course identical to the outflow rate), which raised from 810 $mm^3\,s^{-1}$ to 843 $mm^3\,s^{-1}$, and the average velocity within the aneurysm sack, which increased from 4.7 $cm\,s^{-1}$ to 5.0 $cm\,s^{-1}$.

Table 2. Results of the two time dependent simulations. The values are taken at the time of systole at the inlet plane within the aneurysm sack. The aneurysm is referred to as the region above the neck plane.

	Newtonian	Carreau Yasuda
min. shear rate	6.81 s^{-1}	5.30 s^{-1}
max. shear rate	2174.9 s^{-1}	2178.2 s^{-1}
min. viscosity	3.89 mPa s	3.57 mPa s
max. viscosity	3.89 mPa s	7.10 mPa s
avg. velocity	4.71 $cm\,s^{-1}$	5.03 $cm\,s^{-1}$
max. velocity	42.94 $cm\,s^{-1}$	43.98 $cm\,s^{-1}$
tot. inflow	809.51 $mm^3\,s^{-1}$	842.92 $mm^3\,s^{-1}$

4. Conclusion and Outlook

A Carreau Yasuda model was introduced as a non Newtonian model for blood flow simulations and parameters were found using weighted non linear least square regression on normalized whole blood viscosity measurements. An intracranial sidewall aneurysm treated with a flow diverting stent was modeled and CFD simulations with a Newtonian and the non Newtonian model have been performed. The

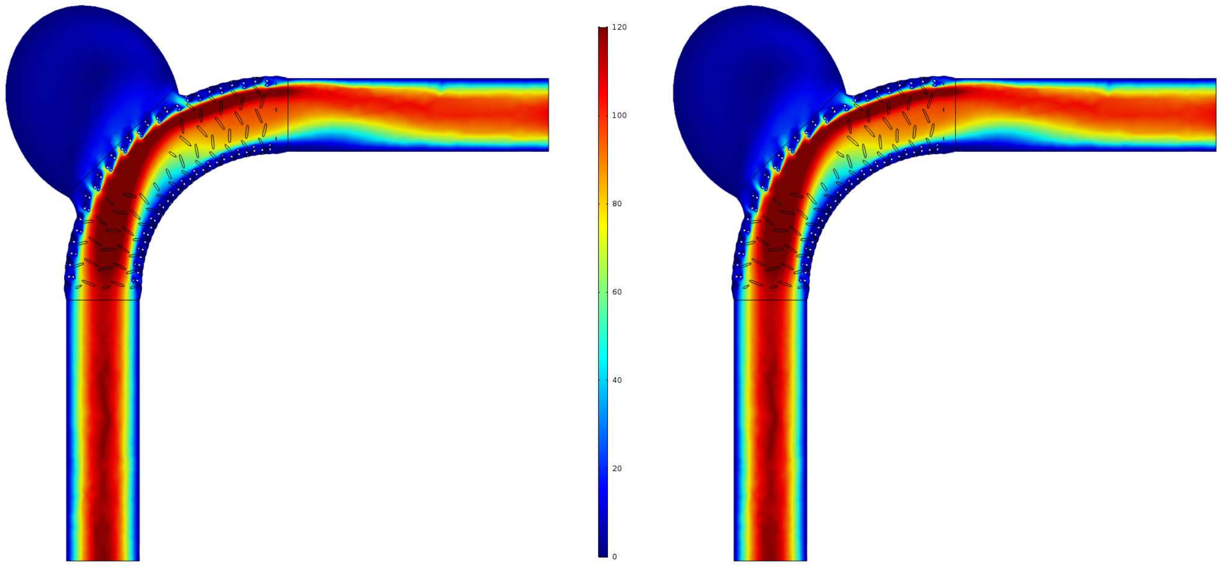

Figure 7. Velocity profiles of the Newtonian model (left) and the Carreau Yasuda model (right). Values are given in $\mathrm{cm\,s^{-1}}$.

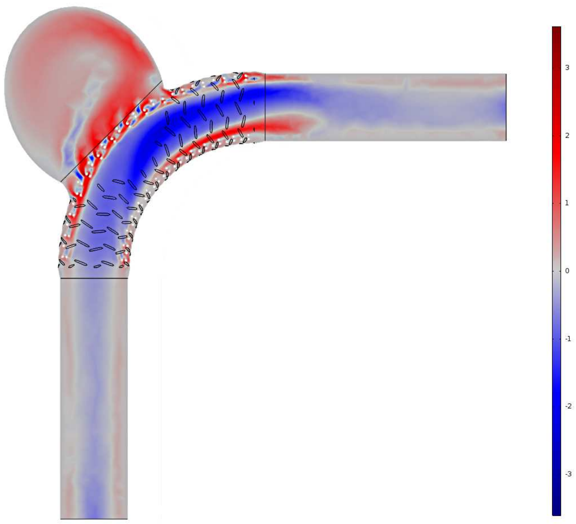

Figure 8. Difference of the velocity profiles (Carreau Yasuda model - Newtonian model). Values are given in $\mathrm{cm\,s^{-1}}$.

results of the comparison between the blood viscosity models have been presented.

Against the first intuition the Newtonian model overestimates the effect of the flow diverting stent. Even there cannot be seen relevant differences in the flow pattern, the error can be seen at the characteristic numbers. In our example this overestimation was about 4% to 6%. Thus in difference to the results of [8] the Newtonian Model seems not to be sufficient for flow calculations past endovascular devices.

The presented model is a solid foundation for future work. Anyhow further investigations with even better blood models and 3D images of real flow diverting stents in vivo have to be done. The resolution of tomographic images will be crucial in building sufficient models.

5. References

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