BLOOD VISCOELASTICITY: EXPERIMENTAL CHARACTERIZATION AND 2D-NUMERICAL SIMULATION OF THE BLOOD FLOW

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ABSTRACT

Circulatory system diseases are the main cause of death in Europe and United States of America. This fact has promoted increasing research related to human body biomechanics. In this context, huge increase of computational capacity has become in a valuable tool for cardiovascular disease studies. Blood viscoelastic flow through healthy carotid artery was here studied. Both experimental and computational techniques were carried out in order to simulate a realistic human situation. On the one hand, regarding the experimental study, relaxation tests were performed on blood by using a rheometer. After that, the Generalized Maxwell model was used to fit experimental data time dependence of relaxation modulus. On the other hand, several numerical simulations were carried out by using finite volume methods implemented in open source software (Foam Extend 3.1). The effect of blood viscoelastic behaviour in a common human carotid artery was tested. Results from Viscoelastic and Newtonian models for blood were compared. The main conclusion of the work is that assuming blood as a Newtonian fluid can give rise to wrong predictions, especially at the near wall region.

NOMENCLATURE

G	[Pa]	Relaxation modulus
G'	[Pa]	Storage modulus
G''	[Pa]	Loss modulus
f	[Hz]	Frequency

Special characters

γ	[-]	Amplitude
δ	[rad]	Phase angle
λ	[s]	Relaxation time
τ	[Pa]	Stress
ω	[rad/s]	Oscillation frequency

INTRODUCTION

First studies on blood rheology have been referred to viscosity shear dependence. Since 1972 and thanks to growing rheometer technology development, blood viscoelastic behaviour has been properly studied [1]. In that sense, different authors [2,3] have studied the viscoelasticity effect on blood coagulation process as well as the pulsatile flow inside a pipe. Besides that, a commercial software based on finite volume methods has allowed to carry out both 2D [4] and 3D [5] blood flow simulations in realistic situations thanks to the increasing of the computational capability in modern computers.

In this work, we characterized blood viscoelastic behaviour by means of rheological tests. The experimental rheological data were fitted with the Generalized Maxwell model to obtain blood relaxation time. The results were used for 2D numerical simulations of blood flow. Numerical results on blood flow assuming viscoelastic behaviour was compared to Newtonian fluid flow.

METHODOLOGY

Regarding rheological tests and before getting reliable experimental data, a pre-shear test should be always applied. Pre-shear consists in applying continuous shear to the sample until a steady state is reached. Thus, any trace of previous external influence should be eliminated.

It is well known [6] that materials exhibit solid-like or liquid-like behaviour depending on the relation between its characteristic relaxation time (λ) and the characteristic time of the experiment. Blood relaxation time can be estimated with stress relaxation test. To that end, constant deformation is applied to the sample and shear stress is monitored. The deformation (γ_0) must belong to the viscoelastic region where linear behaviour is observed, i.e. relaxation modulus must not vary when deformation changes. In order to determine the linear viscoelastic behaviour region of human blood, an amplitude sweep oscillatory test was performed on samples supplied by a healthy male donor. Amplitude (γ) sweep test consists in applying increasing range of amplitude deformations (γ_o) while maintaining constant the oscillation frequency (ω), i.e. the amplitude sweep test is described by

$$\gamma(t) = \gamma_o sin\omega t. \tag{1}$$

This means that the stress (τ) in the sample is also sinusoidal but with certain phase shift (δ),

$$\tau(t) = \tau_o sin(\omega t + \delta). \tag{2}$$

The development of trigonometric function in Equation (2) leads to define the storage (G') and the loss (G'') modulus as,

$$G' = \frac{\tau_o}{\gamma_o} cos\delta, \quad (3)$$

$$G'' = \frac{\tau_o}{\gamma_o} sin\delta. \quad (4)$$

Therefore, monitoring G' and G'', the linear viscoelastic region is defined by G' and G'' having constant values.

Once the experimental tests were carried out, the Generalized Maxwell model was used to fit relaxation stress rheological data. This model relates current shear stress tensor $(\tilde{\tau})$ with shear rate tensor $(\dot{\tilde{\gamma}})$ evaluated for past time (t') that precedes current time (t),

$$\tilde{\tau}(t) = \int_{-\infty}^{t} \sum_{k=1}^{N} G_k e^{-\frac{t-t'}{\lambda_k}} \dot{\tilde{\gamma}}(t')dt'. \quad (5)$$

In Equation (5),

$$G(t) = \sum_{k=1}^{N} G_k e^{-\frac{t}{\lambda_k}}. \quad (6)$$

This is the memory function, which coincides with the time dependent relaxation modulus. G_k are weighting constants and λ_k are relaxation times corresponding to each structural level formed into the material. Each one of these structural levels is mechanically represented by a dashpot and a spring connected in series.

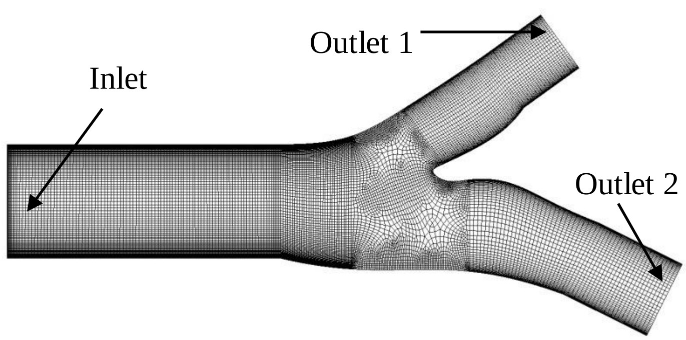

Figure 1 Healthy carotid geometry and mesh.

In order to recreate the geometry on which the blood flows, as well as its meshing, ANSYS designModeler and ANSYS meshing were used. The two-dimensional geometry and mesh can be seen in Figure 1. It represented a typical bifurcation of an uniform healthy carotid.
The numerical simulations were performed using Foam-extended 3.1. To solve transient viscoelastic problems, viscoelasticFluidFoam library were used, together with the PISO algorithm with 2 correctors. A BiCGStab solver was also used to solve the velocity, with a relative tolerance of 10^{-6}. Regarding the inlet boundary condition, a transient profile was used in order to describe the time dependent carotid inlet blood velocity, as shown in Figure 2. This profile was obtained from the experimental measurements shown in [7].

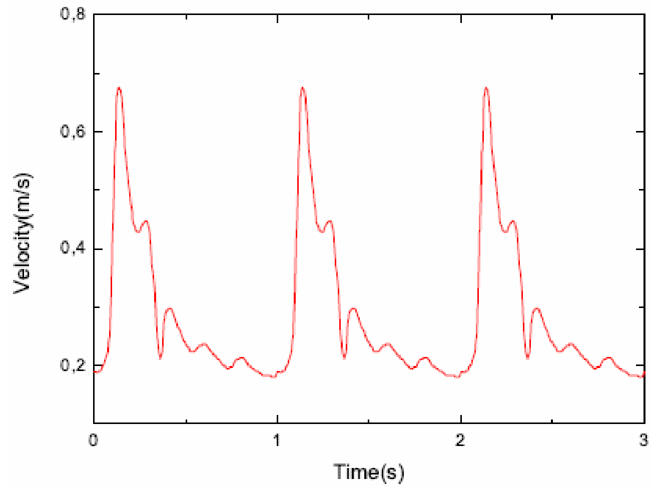

Figure 2 Velocity inlet boundary condition

RESULTS

Amplitude sweep test of a blood sample is shown in Figure 3. Linear viscoelastic (LVE) region extends to γ_0=4% deformation. Only material functions can be defined in LVE region; therefore, relaxation stress test were carried out by applying a constant deformation lower than 4%. Only one term of Equation (6) was used to fit experimental relaxation modulus values (Figure 4),

$$G(t) = (1.6 \pm 0.1) \cdot 10^7 e^{-t/(7.56 \pm 0.01) \cdot 10^{-4}}. \quad (7)$$

Then, the characteristic relaxation time of blood was obtained, i.e. $\lambda = (7.56 \pm 0.01) \cdot 10^{-4} s$.

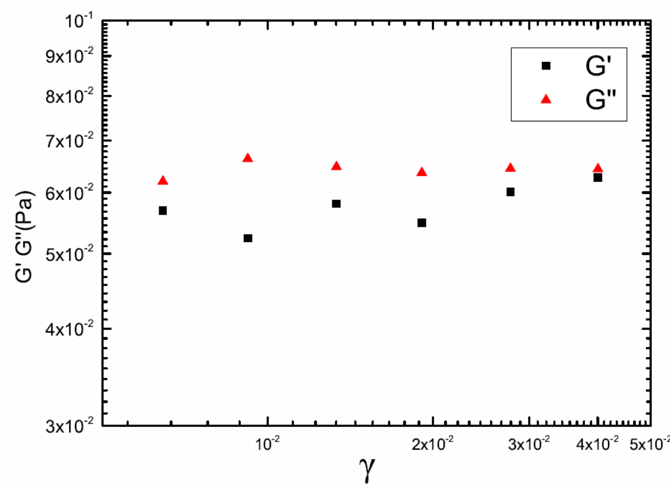

Figure 3 Amplitude sweep test of blood at 36.5°C. $\omega = 2\pi\, rad/s$.

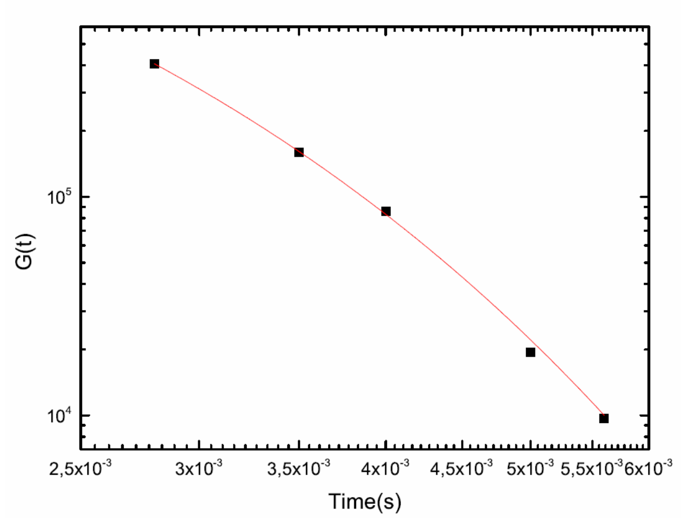

Figure 4 Relaxation test experimental measures

Therefore, the constitutive equation of blood was, finally,

$$\tilde{\tau} = \int_{-\infty}^{t} (1.6 \pm 0.1) \cdot 10^7 e^{-t/(7.56 \pm 0.01)\cdot 10^{-4}} \tilde{\dot{\gamma}}(t')dt'. \quad (8)$$

Comparison between blood flows in 2D geometry described before (Figure 1) when blood is assumed Newtonian or Viscoelastic fluid (Equation (8)), was made. This study helped us to establish the real effect of using viscoelastic model on the blood flow circulation instead has considered blood as Newtonian fluid. To that end, after the simulations were carried out with both Newtonian and viscoelastic model, the velocity profiles at the geometry outlets were compared. The results of the comparisons are shown in Figures 5 and 6 for different times. Symbols represent velocity profiles obtained with the viscoelastic model while error bars quantify the deviation of the Newtonian solution from the viscoelastic one. As can be seen in Figure 5, close to the solid walls the differences between both solutions (indicated by error bars) were higher than at any other outlet1 position. The same result is observed at outlet2 (Figure 6). This can be explained as due to the fact that near the walls flow velocities are enough small to enhance blood viscoelastic behaviour and, consequently, differences must be higher when velocity profile is obtained assuming blood Newtonian behaviour.

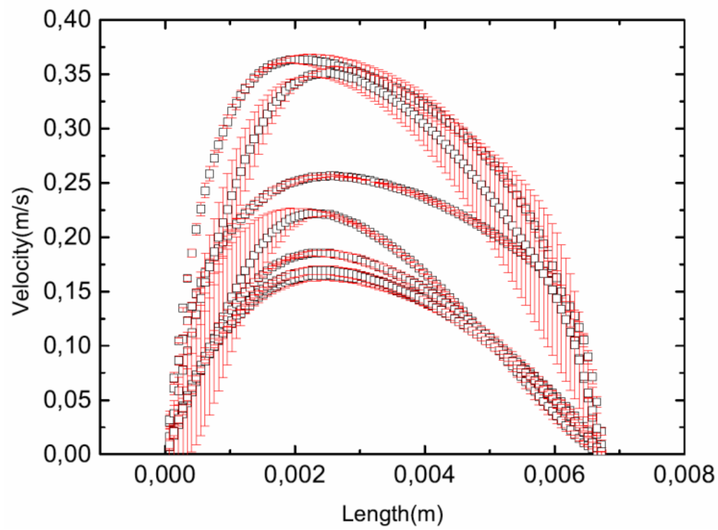

Figure 5 Comparison between viscoelastic and Newtonian velocity profiles at outlet1 and at different times.

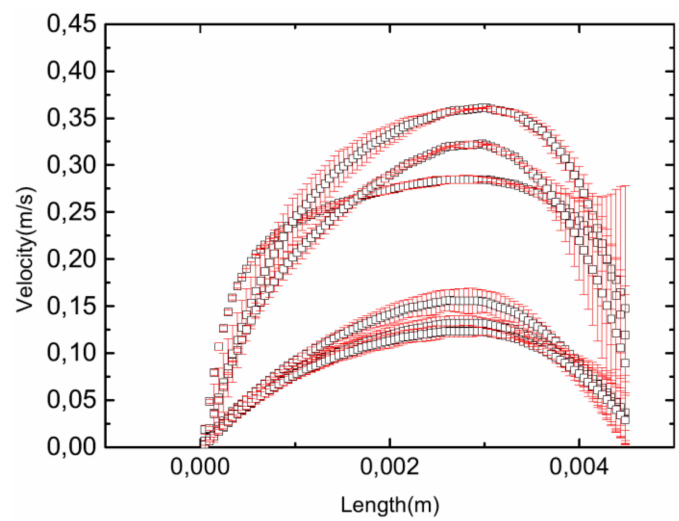

Figure 6 Comparison between viscoelastic and Newtonian velocity profiles at outlet2 and at different times.

CONCLUSIONS

Generalized Maxwell model for blood LVE behaviour was used to fit experimental data from relaxation stress tests. This procedure was used to obtain representative blood relaxation time. Due to the small value of this characteristic time ($\lambda = (7.56 \pm 0.01) \cdot 10^{-4} s$), blood viscoelastic behaviour could appear as not meaningful. Nevertheless, numerical simulations were performed in order to observe viscoelastic effects on blood flow through a simplified healthy carotid in comparison with blood Newtonian flow behaviour in the same geometry. Despite the small relaxation time value, it was obtained that blood viscoelasticity cannot be neglected. In particular, the maximum differences between Newtonian and viscoelastic results occurred near the walls. This result could be explained as due to the low velocities near the walls favours the dominance of the blood elastic component. It is concluded from this study that assuming blood as Newtonian fluid could give rise to wrong flow predictions, specifically if they refer to blood flow near walls, for instance in the common case friction forces exerted by the blood on artery/vein walls must be obtained.

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