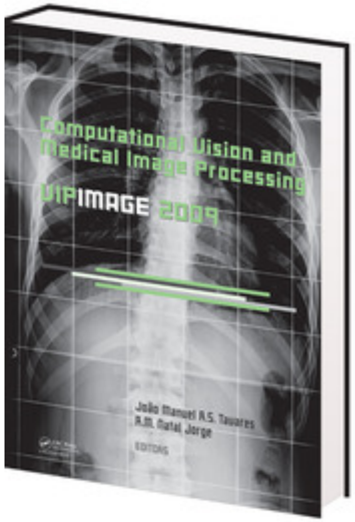

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Motions of particles and red blood cells in a bifurcation: comparison between experiments and numerical simulations

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The blood flow dynamics in microcirculation depends strongly on the microvascular networks composed with short irregular vessel segments which are linked by numerous bifurcations. This paper presents the application of a confocal micro-PTV system to track RBCs through a rectangular polydimethysiloxane (PDMS) microchannel with a bifurcation. By using a confocal micro-PTV system, we have measured the effect of bifurcation on the flow behaviour of both fluorescent particles diluted in pure water and RBCs in concentrated suspensions. After performing simulations with the commercial finite element software package POLYFLOW®, some experimental results were compared with the numerical results. Our preliminary qualitative results suggest that the *in vitro* blood flowing (Re = 0.007) through a bifurcation seems to have a tendency to behave more closely as a Power law model than as a Carreau or Newtonian model.

1 INTRODUCTION

Several studies have revealed that the information obtained on the rheological properties of blood in straight glass microtubes differs from the *in vivo* situation (Pries et al. 1994, Suzuki et al. 1996). The main potential causes for these discrepancies may be due to the endothelial surface layer and microvascular networks composed with short irregular vessel segments which are linked by numerous bifurcations (Maeda 1996, Pries and Secomb 2003).

The main purpose of this paper is to analyse the non-Newtonian flow characteristics of blood flowing in microvascular network models. To accomplish it experimental flow studies performed with a micro-PIV/PTV system will be complemented by hemodynamics computational models (commercial finite element software package POLYFLOW®). By using this combination we expect to gain understanding about several important parameters that affect the blood flow through a diverging microvessel bifurcation.

2 MATERIALS AND METHODS

2.1. *Working fluids and microchannel*

Two working fluids were used in this study: pure water and dextran 40 (Dx40) containing about 14% (14Hct) of human red blood cells (RBCs). The blood was collected from a healthy adult volunteer, where ethylenediaminetetraacetic acid (EDTA) was added to prevent coagulation. The RBCs were separated from the bulk blood by centrifugation and aspiration and then washed twice with physiological saline (PS). The washed RBCs were labeled with a fluorescent cell tracker (CM-Dil, C-7000, Molecular Probes) and then diluted with Dx40 to make up the required RBCs concentration by volume. All blood samples were stored hermetical at 4ºC until the experiment was performed at controlled temperature of about 37ºC (Lima et al. 2009).

The microchannel used in this study was a symmetric PDMS bifurcation (150 μm wide, 50 μm deep for parent vessel; 75 μm wide, 50 μm deep for daughter vessel) fabricated by a soft lithography technique (Lima et al. 2008).

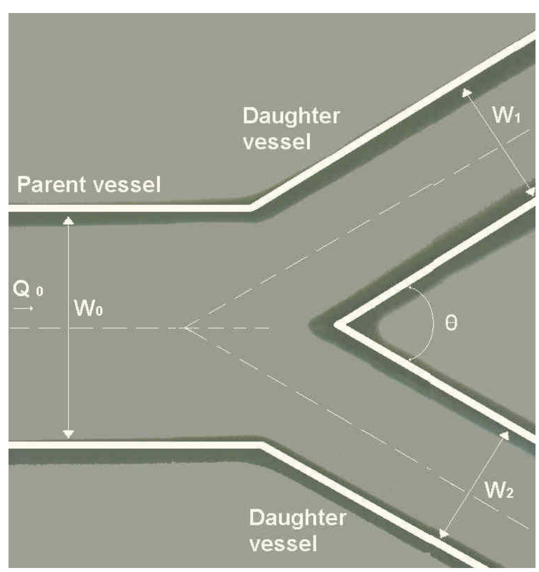


Fig. 1 Symmetrical bifurcation geometry used in this study: Q_0 = 0.18μl/min, W_0 = 150 μm, W_1 = 75 μm, W_2 = 75 μm, θ = 60°, depth = 50 μm.

2.2. *Confocal micro-PTV experimental set-up*

The confocal micro-PTV system used in our experiment consists of an inverted microscope (IX71, Olympus, Japan) combined with a confocal scanning unit (CSU22, Yokogawa) and a diode-pumped solid state (DPSS) laser (Laser Quantum Ltd) with an excitation wavelength of 532 nm. Moreover, a high-speed camera (Phantom v7.1) was connected into the outlet port of the CSU22. The microchannel was placed on the stage of the inverted microscope where the flow rate of the working fluids was kept constant (Re = 0.007) by means of a syringe pump (KD Scientific Inc.). The Reynolds number (Re) and associated experimental parameters are summarized in Table 1. A thermo plate controller (Tokai Hit) was set to 37ºC. All the confocal images were captured in the middle of the microchannels with a resolution of 640×480 pixels, at a rate of 100 frames/s with an exposure time of 9.4 ms. The recorded images were transferred to the computer and then evaluated in the Image J (NIH) (Abramoff et al. 2004) by using the manual tracking MtrackJ (Meijering et al. 2006) plugin. As a result it was possible to track single RBCs through the middle plane of the microchannel.

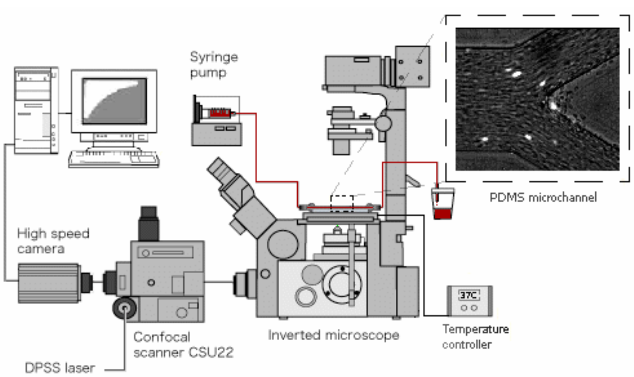


Fig. 2 Experimental set-up.

Table 1 - Experimental parameters used to calculate the Re.

Density (kg/m^3)	1046
Mean velocity (m/s)	3.8×10^{-4}
Hydraulic diameter (m)	7.5×10^{-5}
Viscosity of Dx-40 (Ns/m^2)	4.5×10^{-3}
Re	**0.007**

2.3. Simulation method

The numerical calculations for the laminar isothermal flow of pure water were performed using the finite-element computational fluid dynamics (CFD) program POLYFLOW®. The simulations were carried out in a 3D geometry representing the microchannel. The mesh used in the simulations was mainly constituted by quadrilateral elements, the discretization of the walls of the channel. The size of the elements was fixed after a grid independence test. The grids were successively refined and the velocity obtained with the different meshes were compared, the results being considered independent of the mesh when a difference bellow 1 % was achieved (Fernandes et al. 2007, 2008).

The equations solved were the conservation of mass and momentum equations for laminar incompressible flow of water. The problem is a non-linear problem, so it was necessary to use an iterative method to solve the referred equations. In order to evaluate the convergence of this process, a test based on the relative error in the velocity field was performed and the convergence test value was set to 10^{-4} (Fernandes et al. 2007, 2008).

The boundary conditions were established in order to reproduce the experimental conditions. The flow rate at the inlet of the microchannel was 0.18µl/min and slip at the walls of the channel was assumed to be non-existent.

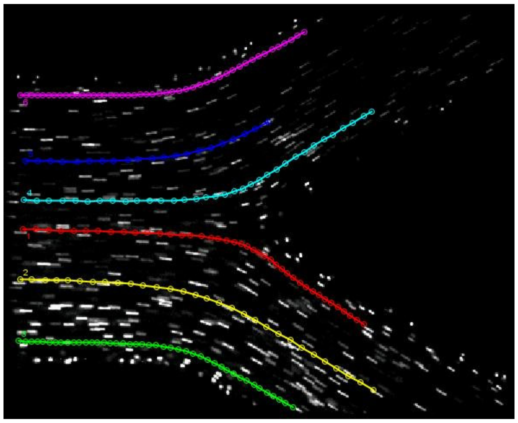
3 RESULTS AND DISCUSSION

The confocal micro-PIV system was first evaluated by comparing the experimental results not only with a well established analytical solution for steady flow in a rectangular microchannel (Lima et al. 2006) but also with a reliable numerical method that was used in past investigations to study the flow behaviour of Newtonian or non-Newtonian fluids at low Reynolds numbers (Fernandes et al. 2007, 2008).

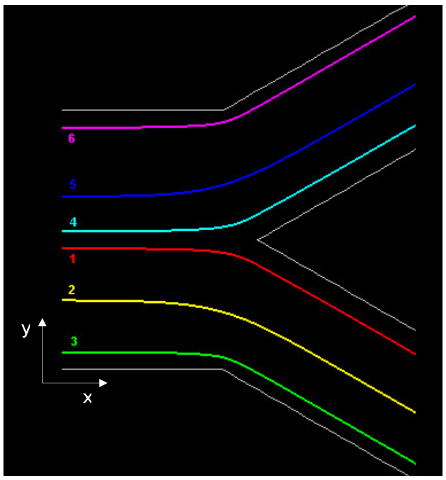
The numerical, experimental and analytical results of the present work were obtained for the middle plane (25 μm height) of the rectangular microchannel. The averaged velocity data obtained from the confocal micro-PTV measurements, analytical solution and numerical simulation were in close agreement. A more detail description of these results can be found elsewhere (Oliveira et al. 2009).

By using a confocal micro-PTV system it was possible to obtain series of successive images at the middle of the bifurcation. Figures 3a and 4 show images with both fluorescent particles and labeled RBCs (laser-emitted light) flowing through a symmetric bifurcation, together with the correspondent time position tracking of both particles and individual RBCs.

By comparing the experimental data from pure water (see Fig. 3a) with the results of the numerical simulation (see Fig. 3b), it is possible to observe that in both cases the trajectories do not exhibit any appreciable deviations in the transversal (*yy* axis) direction.



a)



b)

Fig. 3 a) Paths displacement of fluorescent particles flowing in pure water; b) Numerical trajectories using pure water.

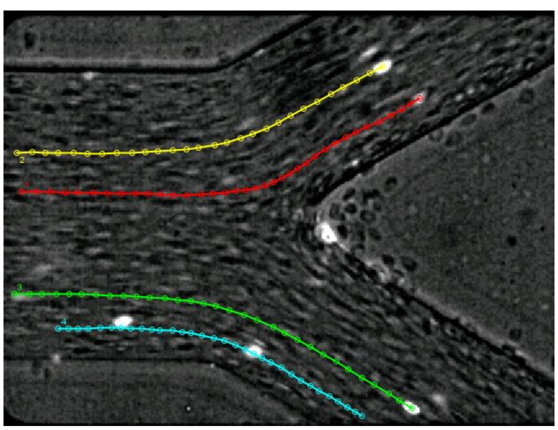


Fig. 4 Paths displacement of labeled RBCs (bright spots) flowing in physiological fluid with 14% Hct (32×).

In addition by comparing qualitatively the experimental data from both pure water and in vitro blood (14% Hct) it is possible to observe that some RBC paths seems to suffer small deviations from the streamlines of the plasma flow probably due to flow perturbations caused by cell interactions in the neighbourhood of the apex of bifurcation.

Moreover numerical simulations of non-Newtonian models were performed around the bifurcation and compared with the experimental results at the regions 1 to 7 (see Fig.5)

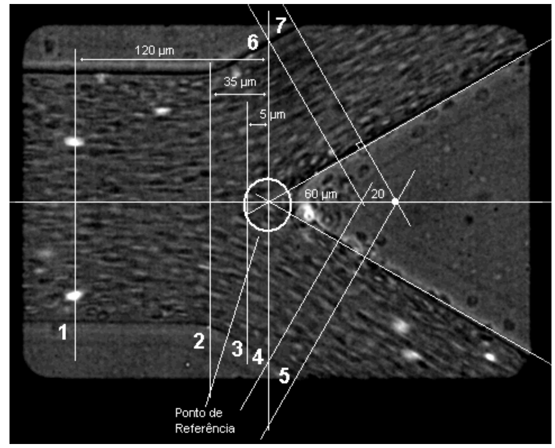
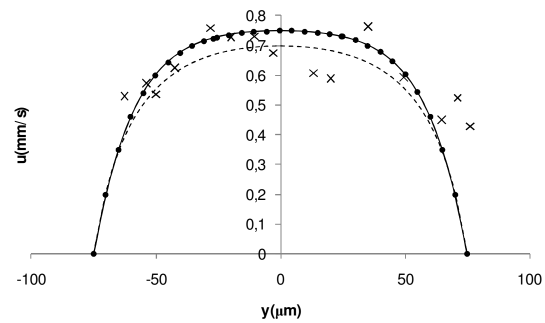
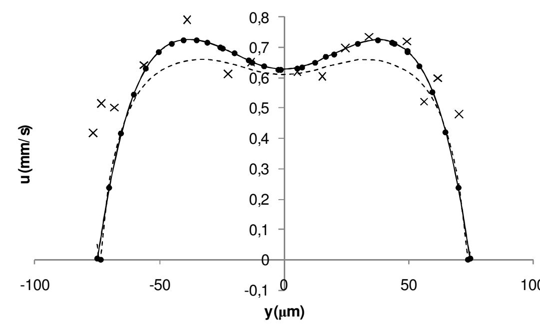


Fig. 5 Regions where the velocity profiles of the numerical and experimental results were compared.

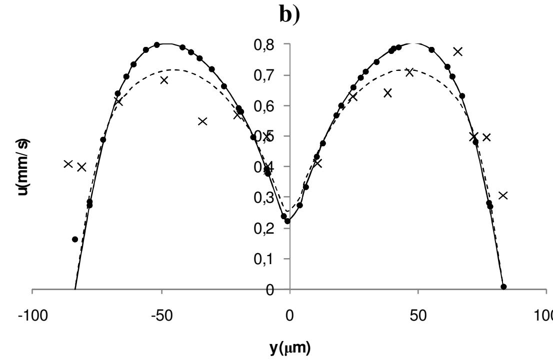
Figures 6 and 7 show the velocity profiles for both computational and experimental results before and after the bifurcation, respectively.



a)

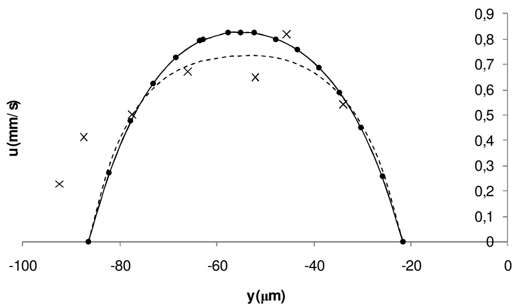


b)

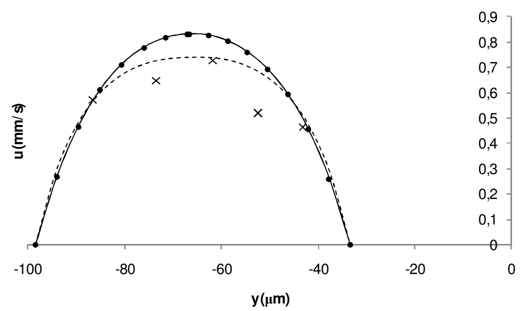


c)

Fig. 6 Velocity profiles for both computational and experimental results before the bifurcation: a) region 1; b) region 2; c) region 3. (—) Newtonian; (- - -) Power Law; (•) Carreau Model; (×) Confocal micro-PIV.



a)



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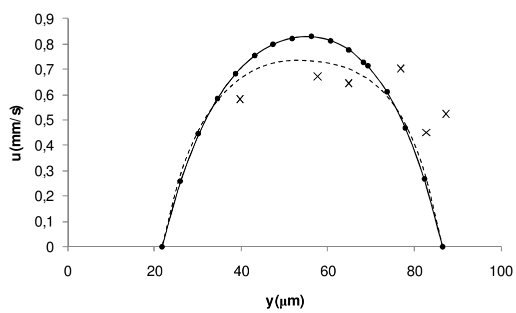
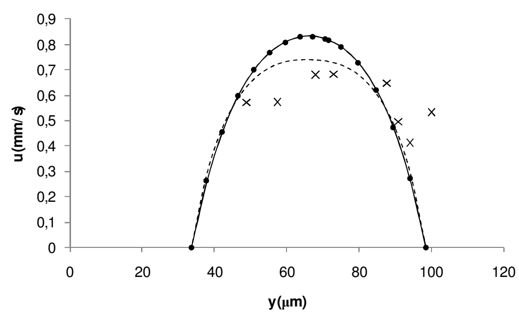


c)



d)

Fig. 7 Velocity profiles for both computational and experimental results after the bifurcation: a) region 4; b) region 5; c) region 6; d) region 7. (—) Newtonian; (- - -) Power Law; (•) Carreau Model; (×) Confocal micro-PIV.

A qualitative examination of Figures 6a) and 6b) does not show a clear correspondence of the experimental results with the numerical models used in this study. However Figure 6c) reveals a tendency of the *in vitro* blood to behave more closely as a Power law model. This is further confirmed by the results obtained in Figure 7, in particular Figures 7 a) and b). An on going study to obtain more detailed quantitative measurements of the blood flow behavior through a bifurcation is currently under way.

ACKNOWLEDGMENTS

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