I, MING LIU
hereby submit this work as part of the requirements for the degree of:
MASTER OF SCIENCE
in:
MECHANICAL ENGINEERING
It is entitled:
BLOOD FLOW IN MICRO-CHANNEL CAPILARY
USING NON-NEWTONIAN VISCOSITY:
A NUMERICAL STUDY

This work and its defense approved by:
Chair: Rupak. K. Banerjee
Yijun Liu
Jandro. L Abot
Blood Flow in Micro-channel Capillary Using non-Newtonian viscosity: A Numerical Study

A dissertation submitted to the
Graduate School
Of the University of Cincinnati
In partial fulfillment of the
requirements for the degree of

MASTER OF SCIENCE

In the Department of Mechanical, Industrial and Nuclear Engineering
of the College Engineering

By
Ming Liu
B.Tech, Mechanical Engineering
Shan Dong University, China, July 2003

Committee Chair: Dr. Rupak Banerjee
ABSTRACT

Blood flow in micro-channel capillary using non-newtonian viscosity: a numerical study

Ming Liu

Significant achievements have been made in the area of lab-on-a-chip devices for biomedical analysis. The recent development of microfluidics has further promoted the successful development of lab-on-a-chip devices by enabling researchers to use numerical methods to improve device performance. In this work, by using the Volume of Fluid (VOF) model, we studied the non-Newtonian shear-thinning viscosity of blood capillary flow in micro-channels with diameters under 100 µm. The blood penetration, withdrawal and final equilibrium are observed. The effects of the size of the microchannels and the concentration of red blood cells (RBCs) in blood on filling distance are also determined. It is found that the blood with 65% RBC concentration withdraws more than those with lower RBC concentrations in the same microchannel. For the blood with lower RBC concentration, increasing the microchannel diameter results in a significant increase in filling distance. This effect was less pronounced for blood with higher RBC concentration. The demonstrated results can be used to improve the design and efficiency of lab-on-a-chip devices that use sample transport by capillary forces.
ACKNOWLEDGEMENTS

I would like to express profound appreciation to my advisor, Dr. Rupak Banerjee, Associate Professor of the Mechanical Engineering and Biomedical Engineering Department at University of Cincinnati, for his guidance and support throughout the course of this study.

I thank Mr. Surendra Balaji Devarakonda for his help on this research. I thank Mr Michael J. Rust for his help for this project. I also wish to express thanks to Professors for their advice and participation in my qualification and/or defense committees. Many thanks to all my Professors and the research associates, staff and graduate students who helped me in one way or another.

I thank all my other friends here and abroad for helping me so much, enriching my life and helping me to pursue my dreams.

I wish to thank my parents for their financial and spiritual support. Without them I can not finish this work.

Finally I must say that it has been an honor for me to be a part of the University of Cincinnati community and always felt proud to come every morning and make my little contribution.
Table of Contents

Abstract iii
Copyright iv
Acknowledgements v
Table of Contents vi
List of Figures viii
List of Tables x
List of Symbols xi

1 Introduction 1

2 Methodology 6
  2.1 Geometry ................................................................. 6
  2.2 Formulation ............................................................. 7
  2.3 Boundary conditions .................................................. 8
  2.4 Materials ................................................................. 8
  2.5 Finite volume method .................................................. 11

3 Results and discussions 13
  3.1 Water validation ....................................................... 13
3.2 Results of blood

3.3 Discussions

4 Conclusions

4.1 Conclusions

4.2 Future work

References
List of Figures

Fig. 1. Geometry of micro-channels and computational grid ........................................ 7

Fig. 2. Shear-dependent viscosities for the Carreau model ........................................ 10

Fig. 3. Convergence of Max distance for blood with 25% RBC concentration entering microchannels with 25 µm, 50 µm, and 100 µm diameters with increasing aspect ratios ................................................................. 12

Fig. 4. Numerical and theoretical results of Distance vs. Time for water entering microchannels with 25 µm, 50 µm, and 100 µm diameters ............................................................... 14

Figure 5 (a) : Numerical results for Distance vs. Time for blood with 25% RBC concentration entering microchannels with 25 µm, 50 µm, and 100 µm diameters ......... 15

Figure 5 (b) : Numerical results for Distance vs. Time for blood with 45% RBC entering microchannels with 25 µm, 50 µm, and 100 µm diameters ........................................... 16

Figure 5 (c): Numerical results for Distance vs. Time for blood with 65% RBC entering microchannels with 25 µm, 50 µm, and 100 µm diameters ........................................... 16

Figure 6 (a): Numerical results for Mass vs. Time for blood with 25%, 45% and 65% RBC concentration entering the microchannel with 100 µm diameter .......... 20
Figure 6(b): Numerical results for Mass vs. Time for blood with 25%, 45% and 65% RBC concentrations entering the microchannel with 50 µm diameter

Figure 6 (c): Numerical results for Mass vs. Time for blood with 25%, 45% and 65% RBC concentrations entering the microchannel with 25 µm diameter

Figure 7 (a): Effect of size of micro-channels and blood RBC concentration on blood filled distance

Figure 7 (b): Effect of size of micro-channels and blood RBC concentration on blood filled distance
List of Tables

1. Properties of water and blood ................................................................. 9
2. Coefficient values in Carreau model for blood with different RBC concentrations .......... 10
3. Variation of Max distance for blood with 25% RBC concentration in microchannels with 25 µm, 50 µm, and 100 µm diameters with increasing aspect ratios ........................................ 12
4. Maximum distance, final distance and withdrawal percentage in blood with 25% RBC concentration ............................................................................................................. 19
5. Maximum distance, final distance and withdrawal percentage in blood with 45% RBC concentration ............................................................................................................. 19
6. Maximum distance, final distance and withdrawal percentage in blood with 65% RBC concentration ............................................................................................................. 20
7. Maximum, final mass and withdrawal percentage of blood with different RBC in 100 µm diameter micro-channel case ................................................................. 24
8. Maximum, final mass and withdrawal percentage of blood with different RBC concentrations in 50 µm diameter microchannel ................................................................. 24
9. Maximum, final mass and withdrawal percentage of blood with different RBC concentrations in 25 µm diameter microchannel ................................................................. 25
List of Symbols

\( \dot{\gamma} \) shear rate

\( \varepsilon_f \) volume fraction

\( \lambda \) time constant

\( \mu \) viscosity

\( \mu_0 \) zero shear viscosity

\( \mu_\infty \) infinite shear viscosity

\( \rho \) density

\( F_\sigma \) surface tension force

\( n \) power law index

\( p \) pressure

\( \frac{dP}{dX} \) pressure gradient

\( u \) velocity
Chapter 1

Introduction

The last decade has seen the emergence of lab-on-a-chip devices in the area of biochemical analysis [Ahn et al., 2003; Srinivasan et al., 2004]. Lab-on-a-chips have been developed for a variety of applications, including DNA analysis [Burns et al., 1998], clinical analysis and diagnostics [Vespoorte, 2002; Choi et al., 2004]. Most of the advantages of lab-on-a-chip devices for clinical diagnostics applications (such as reduced reagents and sample waste) rely on the small volume of blood sample tested in lab-on-a-chip systems [Figeys et al., 2000]. One of the primary reasons these systems have been successful has been the simultaneous development of the field of microfluidics [Karniadakis et al., 2002], which studies the fluid flow through channels that have dimensions on the order of 500 $\mu$m or less. Microfluidic flows are typically driven by either applied pressure or electrokinetic techniques [Nguyen, Wereley, 2002], both of which require external equipment and added design complexity. An alternative approach is to use capillary forces to introduce samples to lab-on-a-chip devices and provide sample transport on the chip. To effectively use
capillary forces, a fundamental understanding of capillary transport in microchannels needs to be obtained, especially when the sample is of non-Newtonian nature such as blood.

When a droplet of fluid comes in contact with a micro-capillary channel, surface tension pulls it into the channel and generates fluid motion [Chakraborty et al. 2005]. The effect of surface tension becomes significantly more important in this micro-capillary than in macro-capillaries because the surface area to volume ratio of the micro-capillary increases with the corresponding decrease in characteristic length scales. Pioneering attempts to understand capillary-driven fluid flows for practical applications have been made by Lucas et al., 1918 and Washburn, 1921. Recently, Lukas and Soukupova, [1999] have carried out a data analysis to test the validity of the above-mentioned approach for some fibrous materials, and have obtained a solution for the Lucas–Washburn equation that includes a gravity term. Recently, such non-homogeneous flows have been studied extensively using an alternative approach of stochastic simulation. Additionally, Lukkarinen et al. [1995] have studied the mechanisms of fluid droplet spreading on flat solid objects.

Although surface tension driven liquid flows, such as those mentioned above, have been extensively studied for over a century, many fundamental aspects related to such fluid motions in micro-capillary channels are yet to be clearly understood. It is well known that at the points of intersection of the liquid–gas interface and solid
capillary wall, the surface tension forces originating from solid–liquid, liquid–gas and solid–gas interactions, respectively, keep the interface in local equilibrium, by establishing a contact angle [Huang et al., 2006]. Maintenance of this equilibrium state effectively induces fluid motion towards the contact surfaces. However, the surface tension forces governing this situation cannot be determined in a straightforward manner. One fundamental reason behind this is based on the fact that the velocity of the liquid meniscus itself can be related to its surface energy. For example, immediately after a layer of fluid molecules is adsorbed by the channel surface, the next set of fluid molecules can march more easily in the micro-channel, giving rise to a dynamic evolution of the contact angle. Although significant advancements have been made in microfluidics and lab-on-a-chip technology in recent years, rarely have attempts been devoted towards fundamental theoretical understanding of surface-tension driven non-Newtonian flow of liquids from a droplet into micro-capillary channels, under the realistic conditions of velocity-dependent contact angles. Yet interactions between a dynamically-evolving contact angle (with the associated transience in surface tension) and viscous effects offered by suspended red blood cells (RBCs) have not been well-demonstrated in the literature. However, such analysis can act as a key towards more fundamental understanding of the working principles of some of the newly synthesized bio-microfluidic devices.

Capillary rise experiments of different liquids in glass capillaries were carried out by Siebold et al. [2000]. A detailed review of theoretical and experimental
studies on capillary flow of Newtonian fluids through various conduits has been
presented by Turian and Kessler [Turian et al., 2000]. Early experimental work to
understand the effect of non-Newtonian viscosity of blood on the flow was carried out
by Forrester and Young [Forrester et al., 1970]. They reported the different flow
pattern between blood and water in vessels. Moravec and Liepsch found the
difference between the velocity distributions of Newtonian and non-Newtonian fluids
in a simplified 3-dimentional model [Moravec et al., 1983]. Nakamura conducted a
numerical study and the results indicated that the non-Newtonian effects weakened
the wall shear stress associated with the stenosis [Nakamura et al., 1988]. Various
models used by researchers were examined and the model constants were summarized
by Young et al. [1997].

The penetration rate of power-law non-Newtonian liquid in a circular tube
driven by capillary pressure was derived by Turian et al. [2005]. It was found that
the rate of penetration as well as withdrawal of liquid is dramatically retarded as the
liquid becomes more strongly shear-shinning, i.e, as n becomes small. Blood flow
through simple mesoscopic geometry was investigated using power law model by
Trebotich et al. [2001]. In this paper, numerical computations for Newtonian fluids
(water) have been conducted and the results have been compared with theoretical
analysis based on Washburn’s equation. Then non-Newtonian fluids, including
blood with 25% RBC concentration (similar to blood with anemia, particular in cases
of women), 45% RBC concentration (normal whole blood) and 65% RBC
concentration (blood typically from the patients with polycythemia), are studied using the Carreau model. The work is unique as computations for non-Newtonian fluids in micro-capillaries have been done for the first time using the Carreau model. The penetration and withdrawal of blood samples in microchannels were studied and the effects of RBC concentration and microchannel diameter investigated.
Chapter 2

Methodology

2.1 Geometry

Axis-symmetric micro-channels, as shown in Figure 1, were considered in this work. The diameters of circular microchannels studied in this research are 25, 50, and 100 μm and the microchannel length is fixed at 5 cm. A computational mesh was generated using GAMBIT software (Fluent). Water and blood with different red blood cell (RBC) concentrations were considered in the computation. The dark region of Figure 1 indicates the portion of the microchannel that is filled with fluids (water or blood) whereas the light region is filled with air.
2.2 Formulation

The Volume of Fluid model (VOF) was used to track the interface between the fluids and air. Conservation equations 1 and 2 are the governing equations for this multi-flow problem.

\[
\frac{\partial \varepsilon_f}{\partial t} + \nabla \cdot (\varepsilon_f \vec{u}) = 0 \tag{1}
\]

\[
\rho \left( \frac{\partial \varepsilon_f}{\partial t} + \nabla \cdot (\varepsilon_f \vec{u}) \right) = -\varepsilon_f \nabla p + \rho g + \left( \nabla \cdot \varepsilon_f \mu \left( \nabla \mu + \left( \nabla \mu \right)^T \right) \right) + F_s \tag{2}
\]

where, \( \varepsilon_f \) is the volume fraction. In a single cell, \( \varepsilon_f \) is the volume fraction of the fluid and \( 1-\varepsilon_f \) is that of air. \( \vec{u} \) is the velocity of liquids, \( p \) is the pressure, and \( \mu \) is the coefficient of viscosity. The source term \( F_s \), accounts for the surface tension force. Here, ‘\( \cdot \)’ on the top of \( \vec{u} \) and \( \vec{g} \) means that they are the vector quantities.

Equation 3 was used to calculate the local average density \( \rho \), linearly weighing the densities of the fluid \( (f) \) and air \( (a) \).
\[ \rho = \varepsilon_f \rho_f + (1 - \varepsilon_f) \rho_a \]  

(3)

2.3 Boundary Conditions

Boundary conditions applied to the problem are as follows:

1) No slip boundary condition is considered on the walls of the micro-channels.

\[ \bar{u} = 0 \]  

(4)

2) Pressure boundary condition is used for the inlet and outlet of the channels.

\[ P_{\text{inlet}} = P_{\text{outlet}} = P_{\text{atmosphere}} \]  

(5)

3) No pressure gradient between the inlet and outlet of the micro-channels.

\[ \frac{dP}{dX} = 0 \]  

(6)

2.4 Material Properties

Water and blood with different RBC concentration were considered in the computation. The density, surface tension, viscosity and contact angle of water are 998 kg/m³, 0.072 N/m, 1.0012 cp and 61 degrees, respectively. The surface tension and contact angle of blood are 0.056 N/m and 92 degrees, respectively. The density of blood varies according to different RBC concentrations in blood. The densities of blood with 25%, 45% and 65% RBC are 1040 kg/m³, 1060 kg/m³ and 1080 kg/m³, respectively.
Table 1: Properties of water and blood

<table>
<thead>
<tr>
<th>Material</th>
<th>Density (kg/m(^3))</th>
<th>Sur.Ten(N/m)</th>
<th>Viscosity(cp)</th>
<th>Contact Angles (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td>998</td>
<td>0.072</td>
<td>1.0012</td>
<td>61</td>
</tr>
<tr>
<td></td>
<td>1040 for 25% Hct</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>1060 for 45% Hct</td>
<td>0.056</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>1080 for 65% Hct</td>
<td></td>
<td>Carreau Model</td>
<td></td>
</tr>
<tr>
<td>Blood</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>1080 for 65% Hct</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

The non-Newtonian viscosity of the blood was modeled using the Carreau model. For a Newtonian fluid, the coefficient of viscosity \( \mu \) is constant. In the Carreau model, the viscosity of a non Newtonian fluid is dependent upon shear rate by the following equation:

\[
\mu_{\text{eff}}(\dot{\gamma}) = \mu_{\infty} + (\mu_0 - \mu_{\infty})(1 + (\dot{\gamma}/\lambda)^n)^{\frac{n-1}{2}}
\]  

(7)

where \( \mu_{\text{eff}}(\dot{\gamma}) \) is viscosity as a function of shear rate, \( \mu_{\infty} \) is infinite shear viscosity, \( \mu_0 \) is zero shear viscosity, \( \lambda \) is time constant, and \( n \) is power law index [Cho et al., 1991].

The values of all the coefficients for blood with different RBC concentrations are
cited in Table 2. The Carreau model for the shear thinning behavior of blood is shown in Figure 2.

![Figure 2: Shear-dependent viscosities for the Carreau model](image)

**Table 2: Coefficient values in Carreau model for blood with different RBC concentrations**

<table>
<thead>
<tr>
<th>Hct</th>
<th>Infinite shear viscosity (cP)</th>
<th>Zero shear viscosity (cP)</th>
<th>Time constant(s)</th>
<th>Power law index</th>
</tr>
</thead>
<tbody>
<tr>
<td>25%</td>
<td>2.57</td>
<td>17.8</td>
<td>12.448</td>
<td>0.33</td>
</tr>
<tr>
<td>45%</td>
<td>3.45</td>
<td>161</td>
<td>39.42</td>
<td>0.3568</td>
</tr>
<tr>
<td>65%</td>
<td>8.02</td>
<td>859.2</td>
<td>103.088</td>
<td>0.389</td>
</tr>
</tbody>
</table>
2.5 Finite Volume Method

The governing equations were discretized and solved using a finite volume method (FLUENT). The mesh was generated using GAMBIT and was composed of hexahedral elements. The mesh contained approximately 7,200 cells in the 25 μm microchannel, 14,400 cells in the 50 μm microchannel and 28,800 cells in the 100 μm microchannel. Time step is 1x10\(^{-7}\). The non linear governing equations were discretized using an implicit linearization scheme. The PRESTO (Pressure Staggering Option) algorithm was used to solve for pressure and velocity from the equations. The second order upwind scheme was employed in the computation. This process continued iteratively until all the residuals of the mass continuity and momentum equations were less than 1x10\(^{-6}\). Mesh independence was assured by comparing the fluid filling distances under different aspect ratios of 1:5, 1:10 and 1:15. The results showed deviations of less than 5% by decreasing the aspect ratio. The aspect ratio of elements was kept at 1:5 for the experiments. Figure 3 shows the convergence of Max distance for blood with 25% RBC concentration into 25 μm, 50 μm, and 100 μm channels with increasing aspect ratios. The difference of Max distances got by using aspect ratio of 1:10 are 2.1%, 2% and 1.1% from those got by using aspect ratio of 1:5 in 25 μm, 50 μm, and 100 μm channels.
Figure 3: Convergence of Max distance for blood with 25% RBC concentration entering microchannels with 25 µm, 50 µm, and 100 µm diameters with increasing aspect ratios

Table 3: Variation of Max distance for blood with 25% RBC concentration in microchannels with 25 µm, 50 µm, and 100 µm diameters with increasing aspect ratios

<table>
<thead>
<tr>
<th>Aspect Ratio</th>
<th>Diameters</th>
<th>1:15</th>
<th>1:10</th>
<th>1:5</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>25µm</td>
<td>135µm</td>
<td>139µm</td>
<td>142µm</td>
</tr>
<tr>
<td></td>
<td>50µm</td>
<td>391µm</td>
<td>400µm</td>
<td>409µm</td>
</tr>
<tr>
<td></td>
<td>100µm</td>
<td>1140µm</td>
<td>1157µm</td>
<td>1170µm</td>
</tr>
</tbody>
</table>
Chapter 3

Results & Discussion

3.1 Water validation

The effects of microchannel diameter and RBC concentration of blood on filling characteristics of microchannels were studied and the results are discussed. Specifically, the filling distance and blood sample mass were calculated along with any withdrawal observed for the cases studied. Additionally, water was used in computation and the results were compared with the theoretical prediction.
Figure 4: Numerical and theoretical results of Distance vs. Time for water entering microchannels with 25 µm, 50 µm, and 100 µm diameters.

The computational results for water entering microchannels with 25 µm, 50 µm, and 100 µm diameters are shown in Figure 4. As expected, the liquid penetration speed increases when increasing the diameter of microchannels. After 2 seconds, the liquid has penetrated 40 mm into the micro-channel with 100 µm diameter, which is 42% farther than that in the microchannel with 50 µm diameter. Water reaches 29 mm in the microchannel with 50 µm diameter, which is 41% farther than that in the microchannel with 25 µm diameter. The computational results for water in microchannels were compared with the theoretical prediction based on Washburn’s equation.
\[ X = \sqrt{\frac{\gamma_{lv} R \cos \theta}{2 \eta}} \sqrt{t} \]  

(8)

The maximum difference between the theoretical prediction and the computational results are 4.2%, 1.5%, and 4.8% for 25, 50, and 100 µm diameters. Thus, the computational results match well with the theoretical prediction.

3.2 Results of blood

Figure 5 (a): Numerical results for Distance vs. Time for blood with 25% RBC concentration entering microchannels with 25 µm, 50 µm, and 100 µm diameters.
Figure 5 (b) : Numerical results for Distance vs. Time for blood with 45% RBC entering micro-channels with 25 µm, 50 µm, and 100 µm diameters.

Figure 5 (c): Numerical results for Distance vs. Time for blood with 65% RBC entering micro-channels with 25 µm, 50 µm, and 100 µm diameters.
Figure 5 (a) shows the results for the case of blood with 25% RBC concentration entering microchannels with 25 µm, 50 µm, and 100 µm diameters. The figure displays the movement of blood from the microchannel entrance to its stopping location for microchannels with different diameters. The results show that the blood enters the microchannels rapidly and stops at the maximum distance. No withdrawal of blood was observed in the three microchannels; thus the final distance is equal to maximum distance in this case. The maximum distances traveled by the 25% RBC concentration samples in the different microchannels were obtained and compared. Maximum distances of 1170, 409, and 142 µm were reached by blood in channels with diameters of 100, 50, and 25 µm, respectively. Arrow #1 in figure 5 (a) indicated that maximum distance filled by blood increased with increasing diameters of microchannels. The maximum and final distances and withdrawal percentages for each microchannel diameter are shown in Table 4. The withdrawal percentage is given by the following equation:

\[
\text{Withdrawal percentage} = \frac{\text{Maximum distance} - \text{Final distance}}{\text{Maximum distance}} \tag{9}
\]

The results for the case of blood with 45% RBC concentration entering into microchannels with 25 µm, 50 µm, and 100 µm diameters are shown in Figure 5 (b). The penetration and withdrawal of blood can also be observed in all three microchannels. As shown in Table 5, The maximum and final distances traveled by blood with 45% RBC concentration was obtained and withdrawal percentage.
calculated for the microchannels with different diameters. A maximum distance of 558 µm was reached by blood in the 100 µm microchannel and after withdrawal it attained a final distance of 258 µm. The blood withdrawal distance is 300 µm; thus the withdrawal percentage is 54%. A maximum distance of 400 µm was reached by blood in the 50 µm channel before withdrawing and stopping at a final distance of 124 µm. The blood withdrawal distance is 276 µm and the withdrawal percentage is 69%. A maximum distance of 155 µm was reached by blood in the 25 µm channel before withdrawing and stopping at a final distance of 65 µm. The blood withdrawal distance is 90 µm, thus the withdrawal percentage is 58%. Arrow #1 and #2 in figure 5 (b) show the trends that both the maximum and final distances increase as the diameters of microchannels are increased.

Figure 5 (c) shows the results for blood with 65% RBC concentration in microchannels with 25 µm, 50 µm, and 100 µm diameters. The penetration and withdrawal of the blood sample can also be observed in all three conditions. The maximum and final distances and withdrawal percentage for blood with 65% RBC concentration in microchannels with three different diameters are shown in Table 6. The blood sample in 100 µm diameter microchannel traveled a maximum distance of 385 µm before withdrawing to a final distance of 94 µm (76% withdrawal). For the 50 µm diameter channel, the blood sample traveled a maximum distance of 250 µm and then withdrew to a final distance of 116 µm (54% withdrawal). The blood sample in the 25 µm diameter microchannel traveled a maximum distance of 198 µm
before withdrawing to a distance of 153 µm (23% withdrawal). As the diameters of microchannels are increased, the maximum distances increase as arrow #1 indicates in figure 5 (c), while the final distances decrease as arrow #2 indicates.

Table 4: Maximum distance, final distance and withdrawal percentage in blood with 25% RBC concentration

<table>
<thead>
<tr>
<th>Diameters</th>
<th>Maximum distance</th>
<th>Final distance</th>
<th>Withdrawal %</th>
</tr>
</thead>
<tbody>
<tr>
<td>25 µm</td>
<td>142 µm</td>
<td>142 µm</td>
<td>0%</td>
</tr>
<tr>
<td>50 µm</td>
<td>409 µm</td>
<td>409 µm</td>
<td>0%</td>
</tr>
<tr>
<td>100 µm</td>
<td>1170 µm</td>
<td>1170 µm</td>
<td>0%</td>
</tr>
</tbody>
</table>

Table 5: Maximum distance, final distance and withdrawal percentage in blood with 45% RBC concentration

<table>
<thead>
<tr>
<th>Diameters</th>
<th>Maximum distance</th>
<th>Final distance</th>
<th>Withdrawal %</th>
</tr>
</thead>
<tbody>
<tr>
<td>25 µm</td>
<td>155 µm</td>
<td>65 µm</td>
<td>58%</td>
</tr>
<tr>
<td>50 µm</td>
<td>400 µm</td>
<td>124 µm</td>
<td>69%</td>
</tr>
<tr>
<td>100 µm</td>
<td>558 µm</td>
<td>258 µm</td>
<td>54%</td>
</tr>
</tbody>
</table>
Table 6: Maximum distance, final distance and withdrawal percentage in blood with 65% RBC concentration

<table>
<thead>
<tr>
<th>Diameters</th>
<th>Maximum distance</th>
<th>Final distance</th>
<th>Withdrawal%</th>
</tr>
</thead>
<tbody>
<tr>
<td>25 µm</td>
<td>198 µm</td>
<td>153 µm</td>
<td>23%</td>
</tr>
<tr>
<td>50 µm</td>
<td>250 µm</td>
<td>116 µm</td>
<td>54%</td>
</tr>
<tr>
<td>100 µm</td>
<td>385 µm</td>
<td>94 µm</td>
<td>76%</td>
</tr>
</tbody>
</table>

Figure 6 (a): Numerical results for Mass vs. Time for blood with 25%, 45% and 65% RBC concentration entering the microchannel with 100 µm diameter
Figure 6 (b): Numerical results for Mass vs. Time for blood with 25%, 45% and 65% RBC concentrations entering the microchannel with 50 µm diameter.

Figure 6 (c): Numerical results for Mass vs. Time for blood with 25%, 45% and 65% RBC concentrations entering the microchannel with 25 µm diameter.
The results for blood with different RBC concentrations (25%, 45% and 65%) entering a microchannel with 100 µm diameter are reported in Figure 6 (a). The figure records the mass of blood after it enters a microchannel of 100 µm diameter until it stops. The withdrawal of blood with 45% and 65% RBC concentrations can be observed, while the blood with 65% RBC concentration withdraws faster than that with 45% RBC. The comparison of maximum and final masses of blood with different RBC concentration entering the 100 µm diameter microchannel was evaluated and the results displayed in Table 7. The maximum and final masses of blood with 25% RBC concentration in the 100 µm diameter micro-channel are both 9552 ng, indicating no withdrawal. The maximum and final masses of blood with 45% RBC concentration in the 100µm diameter micro-channel are 4626 ng and 2147 ng, respectively, resulting in a withdrawal percentage of 54%. The maximum and final masses of blood with 65% RBC concentration in the 100µm diameter microchannel are 3264 ng and 797 ng, respectively, causing a withdrawal percentage of 76%. As shown by arrow #1 and #2 in figure 6 (a), both the maximum distances and final distances decrease as RBC concentration increases.

Figure 6 (b) shows the computational results for blood with different RBC concentrations (25%, 45% and 65%) entering a microchannel with 50 µm diameter. The withdrawal of blood with 45% and 65% RBC concentrations can be observed in this case as in 100 µm case, and the withdrawal of blood with 65% RBC is faster than
that of blood with 45% RBC. The comparison of maximum and final masses of blood with different RBC entering the 50 µm diameter microchannel was made and the results shown in Table 8. The maximum and final masses of blood with 25% RBC concentration entering the 50 µm diameter microchannel were both 833 ng, indicating no withdrawal. The maximum mass of blood with 45% RBC concentration entering the 50 µm diameter microchannel is 820 ng and the final mass is 258 ng after withdrawal (69%). Similarly, a maximum mass of blood with 65% RBC concentration entering the 50 µm diameter microchannel is 517 ng and the final mass is 246 ng after withdrawal (52%). Arrow #1 and #2 indicate that both the maximum distances and final distances decrease as RBC concentration increases.

The computational results for blood with different RBC concentrations (25%, 45% and 65%) entering a microchannel with 25 µm diameter are shown in Figure 6 (c). The withdrawal of blood with 45% and 65% RBC can be observed for this condition. The comparison of maximum and final masses of blood with different RBC concentrations entering a 25 µm diameter microchannel was evaluated and the results showed in Table 9. The maximum and final masses of blood with 25% RBC concentration in the 25 µm diameter microchannel are both 72 ng, indicating no withdrawal. The maximum and final masses of blood with 45% RBC concentration in the 25 µm diameter microchannel are 80 ng and 34 ng, respectively, yielding a withdrawal percentage of 58%. The maximum and final masses of blood with 65% RBC concentration in the 25 µm diameter microchannel are 105 ng and 81 ng,
respectively, yielding a withdrawal percentage of 23%. Unlike the arrows in figure 6 (a) and 6 (b), the arrow #1 in figure 6 (c) is upward, indicating that the maximum distances increase as RBC concentration increased in this case. The unexpected results in figure 6 (c) could be caused by the numerical artifact, which should be validated by the future experimental work.

Table 7: Maximum, final mass and withdrawal percentage of blood with different RBC in 100 µm diameter micro-channel case

<table>
<thead>
<tr>
<th>RBC</th>
<th>Maximum mass</th>
<th>Final mass</th>
<th>Withdrawal%</th>
</tr>
</thead>
<tbody>
<tr>
<td>25%</td>
<td>9552 ng</td>
<td>9552 ng</td>
<td>0%</td>
</tr>
<tr>
<td>45%</td>
<td>4626 ng</td>
<td>2147 ng</td>
<td>54%</td>
</tr>
<tr>
<td>65%</td>
<td>3264 ng</td>
<td>797 ng</td>
<td>76%</td>
</tr>
</tbody>
</table>

Table 8: Maximum, final mass and withdrawal percentage of blood with different RBC concentrations in 50 µm diameter microchannel

<table>
<thead>
<tr>
<th>RBC</th>
<th>Maximum mass</th>
<th>Final mass</th>
<th>Withdrawal%</th>
</tr>
</thead>
<tbody>
<tr>
<td>25%</td>
<td>833 ng</td>
<td>833 ng</td>
<td>0%</td>
</tr>
<tr>
<td>45%</td>
<td>820 ng</td>
<td>258 ng</td>
<td>54%</td>
</tr>
<tr>
<td>65%</td>
<td>517 ng</td>
<td>246 ng</td>
<td>69%</td>
</tr>
</tbody>
</table>
Table 9: Maximum, final mass and withdrawal percentage of blood with different RBC concentrations in 25 µm diameter microchannel

<table>
<thead>
<tr>
<th>RBC</th>
<th>Maximum mass</th>
<th>Final mass</th>
<th>Withdrawal%</th>
</tr>
</thead>
<tbody>
<tr>
<td>25%</td>
<td>72ng</td>
<td>72ng</td>
<td>0%</td>
</tr>
<tr>
<td>45%</td>
<td>80ng</td>
<td>34ng</td>
<td>58%</td>
</tr>
<tr>
<td>65%</td>
<td>105ng</td>
<td>81ng</td>
<td>23%</td>
</tr>
</tbody>
</table>

3.3 Discussions
Figure 7 (a): Effect of size of micro-channels and blood RBC concentration on blood filled distance

Figure 7 (a) shows the maximum distance filled by blood with different RBC concentrations in microchannels with different diameters. Figure 7 (b) records the final distance. The two primary effects on filled distance are discussed here: the diameters of microchannels (25 µm, 45 µm and 100 µm) and blood RBC concentration (25%, 45% and 65%). As shown in figure 7 (a), for the blood with same RBC concentration, the maximum filled distance increases as the diameter of microchannel increases. The maximum distance filled by blood with 25% RBC was
increased about 7 times by increasing the diameter from 25 µm to 100 µm, which is more than the 2.6 times increase for blood with 45% RBC concentration and 94% increase for blood with 65% RBC concentration. Figure 7 (b) shows that the final filled distance also increases as the microchannel diameter increases for blood with 25% and 45% RBC concentrations. However, the final distance filled by blood with 65% RBC concentration decreased as diameters increase from 25 µm to 100 µm.

The effect of RBC concentrations of 25%, 45% and 65% was evaluated. In the microchannel with 100 µm diameter, the maximum and final distance filled by blood increases as the RBC concentration decreases. The maximum distance filled by blood with 25% RBC is 1.1 times more than that filled by blood with 45% RBC and 2 times more than that filled by blood with 65% RBC in the 100µm diameter micro-channel. For the 50 µm microchannel, the maximum distance filled by blood with 25% RBC is similar to that filled by blood with 45% RBC and only 1.6 times that filled by blood with 65% RBC. In the microchannel with 25 µm diameter, the maximum and final distance filled by blood increases as the RBC concentration increases.
Chapter 4

Conclusions

4.1 Conclusions

The blood flow in microchannels has been studied numerically using the Carreau model that incorporate shear thinning non-Newtonian viscosity of blood. The withdrawal of blood sample was observed for blood with higher RBC concentration (45% and 65%), which confirmed the conclusion of Turian [2005]. The effects of the size of microchannels and RBC concentrations of blood on filled distance into micro-channels were discussed.

For blood with 25% and 45% RBC concentration, the maximum and final filled distances increase with increasing microchannel diameter. For blood with 65% RBC concentration, the maximum distance as well as the withdrawal percentage increases with increasing microchannel diameter. The increase of the withdrawal distance is more than that of maximum distance, which causes the final filled distance
to decrease with increasing microchannel diameter in the case of blood with 65% RBC concentration. Therefore, more the blood is diluted, the linear correlation exists between filling distance and microchannel diameter improves.

For larger microchannel diameters (50 µm and 100 µm), the maximum and final filled distances increase with decreasing blood RBC concentration (diluted blood). For smaller microchannel diameters, diluting the blood has reduced impact on filling distance. In 25 µm diameter microchannel, the maximum and final distance filled by blood actually increases with increasing blood RBC concentration, although the effect is less pronounced.

4.2 Future work

These results can be used to design microfluidic lab-on-a-chip systems to efficiently handle small volumes of blood samples. Specifically, lab-on-a-chip systems for clinical diagnostics can be developed using capillary forces to transport blood samples from inlets to desired locations on the chip, such as sample preparation components and detection sensors, without the need for external power sources. Future work in this area can include investigations into the effect of shape and material of the microchannels on blood filling characteristics. The blood with 35% and 55% RBC concentrations will be studied to complement the effect of blood concentration on filled distance. We will also study the blood penetration and withdrawal in the microchannels made by different materials to observe the effect of contact angle on filled distance. Further, experiments of blood penetrating into microchannels should be done to validate the results from numerical study.
References


