



HAL
open science

Effects of Channel Surface Finish on Blood Flow in Microfluidic Devices

D. Allen, D. Prentner, S. Marson, L. Larcombe, K. Jenkins, M. Saumer

► **To cite this version:**

D. Allen, D. Prentner, S. Marson, L. Larcombe, K. Jenkins, et al.. Effects of Channel Surface Finish on Blood Flow in Microfluidic Devices. EDA Publishing, pp.4, 2009. hal-00400589

HAL Id: hal-00400589

<https://hal.science/hal-00400589>

Submitted on 1 Jul 2009

HAL is a multi-disciplinary open access archive for the deposit and dissemination of scientific research documents, whether they are published or not. The documents may come from teaching and research institutions in France or abroad, or from public or private research centers.

L'archive ouverte pluridisciplinaire **HAL**, est destinée au dépôt et à la diffusion de documents scientifiques de niveau recherche, publiés ou non, émanant des établissements d'enseignement et de recherche français ou étrangers, des laboratoires publics ou privés.

Effects of Channel Surface Finish on Blood Flow in Microfluidic Devices

S. Prentner^{1,4}, D.M. Allen¹, L. Larcombe², S. Marson¹, K. Jenkins³ and M. Saumer⁴

¹ Precision Engineering Centre, Cranfield University, Bedford MK43 0AL, UK

² School of Health, Cranfield University, Bedford MK43 0AL, UK

³ School of Engineering, Cranfield University, Bedford MK43, UK

⁴ Fachhochschule Kaiserslautern, 66482 Zweibrücken, Germany

Abstract- The behaviour of blood flow in relation to microchannel surface roughness has been investigated. Special attention was focused on the techniques used to fabricate the microchannels and on the apparent viscosity of the blood as it flowed through these microchannels.

For the experimental comparison of smooth and rough surface channels, each channel was designed to be 10mm long and rectangular in cross-section with aspect ratios of $\geq 100:1$ for channel heights of 50 and 100 μm .

Polycarbonate was used as the material for the device construction. The shims, which created the heights of the channels, were machined from poly(ethylene terephthalate). Surface roughnesses of the channels were varied from Rz of 60nm to 1.8 μm .

Whole horse blood and filtered water were used as the test fluids and differential pressures ranged from 200 to 5000 Pa. The defibrinated horse blood was further treated to prevent coagulation.

The results indicate that a roughness above an unknown value lowers the apparent viscosity of blood dramatically due to boundary effects.

Furthermore, the roughness seemed to influence both water and whole blood almost equally. A set of design rules for channel fabrication is also presented in accordance with the experiments performed.

INTRODUCTION

It is well known that blood is a non-Newtonian liquid and in tubes and vessels a cell-free layer exists near the walls when blood is running through them [1]. This layer becomes more important with decreasing diameter especially between 10 μm and 300 μm . Also known from a number of previous studies, is that the thickness and the viscosity of this layer of plasma with a very low erythrocyte concentration, is dependent on the particle size, the channel diameter and the haematocrit value (Fig. 1).

For Newtonian liquids, Wang and Wang [3] reported that the higher the wall roughness the greater the velocity gradient at the channel wall. At a Reynolds number above 100 and a relative roughness (ϵ) larger than 10%, flow separation happens and regular perturbation methods, which are typically used to predict fluid flow in a microchannel, cannot be used any more. The effect of high roughness (R_z) walls [4] or hydrophobic surfaces [5] is better described with the Navier slip boundary model.

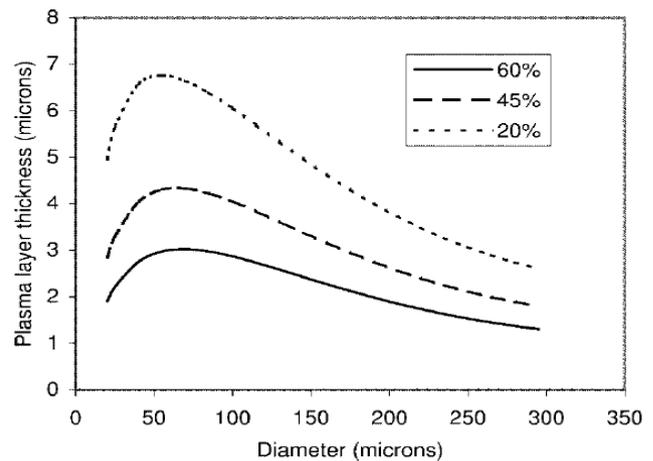


Fig. 1: Variation of plasma layer thickness over channel diameter and for discharge haematocrits HD=60, 45 and 20%. [2]

Thus, the velocity of the bulk liquid flowing in a rough channel rises in comparison to that in a smooth channel. However, it is still unclear what influence a rough wall has on non-Newtonian liquids such as whole blood. This paper focuses on quantifying the influence.

MATERIAL AND METHODS

The viscosity of blood is variable and it is not possible to measure it within a channel directly. Thus, the viscosity has to be measured by calculating the volume flow and the differential pressure. To be able to measure these values, they must be kept constant over a certain time at a constant temperature. Therefore, in the experiments to be described, the volume flow was controlled with a pump (Minipuls3, Model M312 from Gilson Inc.), and the differential pressure was measured with pressure sensors (HCXPM005D6V, HCXM100D6V and HCX005D6V from Sensortech). The readout of the sensors was carried out via a LabView program connected to an A/D converter (LabJack U3 High Voltage A/D). The volume flow was measured by monitoring the increase in outflow blood weight (Mettler balance AT460) over a period of 10 minutes.

All experiments were performed at room temperature and every measurement for water (used as a reference) and whole blood was repeated two or three times to check the

repeatability.

Whole horse blood was obtained from TCS Biosciences Ltd. It was delivered already defibrinated (HB035) but, in addition, 6 g/l of trisodium citrate was added to prevent its coagulation. With the aim of magnifying the effects of the inner walls on the flow, the wall surface was made as large as possible, keeping the flow conditions constant everywhere within the device. Therefore, two large and long rectangular conduits were constructed [6] with a constant width and length of 10mm and with channel heights of 50µm and 100µm respectively (Fig. 2). For comparison, at least two different channel heights with two different roughnesses were measured at three differential pressures (200, 300 and 400Pa for the 100µm channel and 1000, 3000, 5000 Pa for the 50µm channel). The appropriate pressure range was selected to maintain comparable flow velocities for both the 50µm and the 100µm channels.

illustrated in Fig. 4.

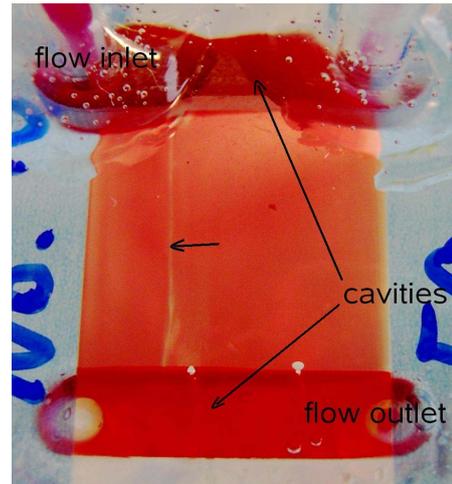


Fig. 4: Blood flow through a channel; the arrow highlights a streamline within the channel

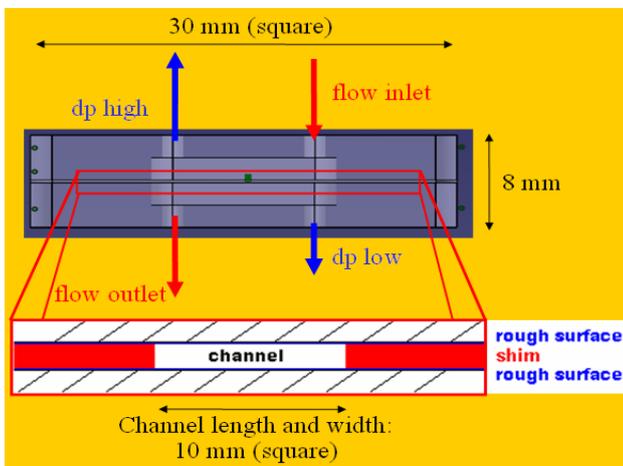


Fig. 2: Geometry of the channel within the device

For additional comparisons, a model of the device was drawn in a CAD program and the velocities in the microchannels were simulated with ANSYS 11(Fig. 3). With this method it was proven that the laminar flow velocities are equal across the whole width and are parallel to the whole length of the conduit.

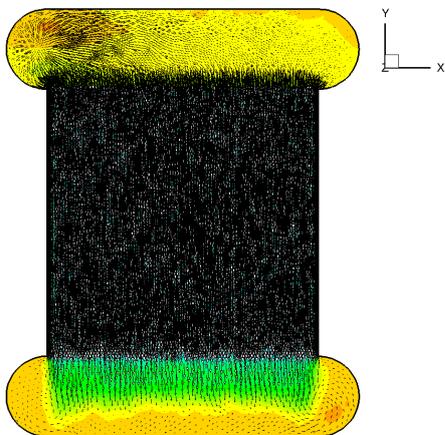


Fig. 3: Simulation of channel velocity in y-direction

Such streamlines have been shown to exist in practice as

A polycarbonate (PC) polymer sheet (Makrolon® GP clear 099) was selected for the test device. The height of the channel was created by inserting a poly(ethylene terephthalate) (PET) sheet (Polymex®) of the desired thickness between two PC sheets and fixed in position by super glue after clamping.

The device cavities, inlets and outlets were machined on the Kern Evo Micromachining Centre using an endmill of 1.5mm radius at 10,000 rpm and a feed rate of 750 mm/min. The same cutting strategy was used to produce the rough channel surface on the PC sheets.

The height of each device was measured directly by a micrometer with an accuracy of about 1 µm.

The roughness measurements were carried out with a Taylor Hobson Form Talysurf 120L and afterwards calculated by a Scilab program. Three roughness measurements on three different regions were made on the rough surface and one on the smooth surface of PC for comparison.

Design of experiments

The aim of the work is to measure a change of volume flow (or viscosity) by changing the inner surface roughness of the channel while keeping all the other parameters constant. The Hagen-Poiseuille equation (1) describes the change of volume flow (I_V), related to the differential pressure (ΔP), viscosity (η), hydraulic diameter (d_h), cross-sectional area of channel (A), and length (l) [7].

$$I_V = \frac{V}{t} = \frac{A \cdot d_h^2 \cdot \Delta P}{32 \cdot \eta \cdot l} \quad (1)$$

For a rectangular channel with the height (h) \ll width (w) of the channel and with $l = w$, equation (1) can be simplified to equation (2) as follows:

$$\eta \approx \frac{h^3 \Delta P}{8 I_V} \quad (2)$$

RESULTS

The experiments carried out on water showed that the calculated volume flow in the rough channel was significantly higher than for the smooth one. In particular, the apparent viscosity of water in the rough channels dropped to 58.1±7.8 % and to 79.3±7.9 % (where 100% represented the viscosity of water in the smooth channel) in the 50µm and 100µm high channels respectively. This suggests a decrease in the flow resistance and a decrease in the apparent viscosity that is in distinct contrast to the predictions made by previous simulations [8].

For horse blood with a haematocrit of 44 % (blood sample 1) the apparent blood viscosity was reduced in the rough channels to 66.4±2.2 % in the 50µm high channel and to 79.7±1.9 % in the device with a 100µm high channel. At a lower haematocrit of 42 % (blood sample 2) a drop in the apparent viscosity in the channel to 51.6±2.7 % was noticed compared to the smooth one (Table I).

For all measurements the Reynolds number was increased 1.7 times higher in the rough channels compared to the smooth channels.

The results indicate that a change of surface roughness, above an unknown value (possibly with Rz between 100 and 1,000 nm) lowers the apparent viscosity of the blood dramatically, probably due to the presence of boundary effects. Furthermore the roughness seems to influence both water and whole blood almost equally.

TABLE I
CALCULATED APPARENT VISCOSITY OF WHOLE HORSE BLOOD FOR SMOOTH (Rz = 60nm) AND ROUGH (Rz = 1.8µm) CHANNELS AND DIFFERENT HAEMATOCRIT VALUES

(Apparent viscosity of blood in smooth channel: assigned as 100 %)	Blood sample 1 with 44% haematocrit	Blood sample 2 with 42% haematocrit
50 µm channel	66.4 ±2.2% (n=9)	51.6 ±2.7% (n=10)
100 µm channel	79.7 ±1.9% (n=12)	N/A

DISCUSSION

The results of the experiments showed a significant change of the apparent channel viscosity of both water and whole horse blood when comparing rough to smooth surfaces. These effects have already been observed [4, 9] but never been compared directly.

However, a direct comparison was carried out in this work and as a result it can be stated that water and blood flows are very similar with respect to the change in apparent viscosity within microchannels. This can be explained by the surface roughness influencing only the Newtonian plasma layer which surrounds the central stream of the blood flow containing the majority of the erythrocytes. This is confirmed by the observation that a decrease in haematocrit, which will increase the plasma layer thickness, seems to increase the influence of the wall roughness too. Furthermore the influence of the roughness seems to decline with higher channels and larger channel diameters. Here the plasma layer is smaller (Fig. 1) and, in the whole channel, the area cross-section for volume flow becomes bigger. In contrast, an increase in the volume

flow by changing the differential pressure, seems to have no influence on the apparent viscosity and also has no effect on the plasma layer.

The reason for this effect may be a decrease in the no-slip boundary effect, enabling the plasma at the wall to flow with a velocity significantly higher than zero. This is dependent on the peak-to-valley roughness and the rise of the Reynolds number to a value of 70 % would confirm the prediction made by Cohen and Feaster [9].

Thus there has to be a limit where the roughness will increase flow resistance and above which it will decrease as modeled in the simulations of Wang and Wang [3].

CONCLUSION

The aim of this work was to analyse the blood flow behaviour in microchannels with variation in wall roughness. Special attention was focused on micromilled surfaces. Therefore a device was designed with the ability to be constructed easily with different channel heights (50 and 100µm) and with variable channel wall roughness. The aspect ratio of the face was 200:1 and 100:1 respectively and the channel length was 10 mm.

As a smooth wall surface, the “natural” one of PC Makrolon® with Rz of about 60 nm was taken and the rough channels have been micromilled, producing Rz of about 1.8 µm. To improve the pump control and the sensor read-out, a closed-loop controlled system has been constructed and programmed with LabView. As a test fluid, whole horse blood was chosen.

The results of the experiments showed a significant drop in the apparent viscosity in the rough-surfaced microchannel compared to the smooth one. This effect seemed to increase with a smaller channel height and with lower erythrocyte concentration (or haematocrit value).

The thickness of the plasma layer, influenced by the channel surface roughness, is highly dependent on the erythrocyte size which is smaller for horses than for humans. Therefore, the results obtained in this work can only be used to predict qualitatively the behaviour of human blood.

As human erythrocytes are bigger than horse erythrocytes, it is expected that the plasma layer will be thicker [10, 11]. This could result in a more pronounced effect of surface roughness on human blood flow.

So, with all this information a set of design rules was created to provide indications on how to design a microfluidic device to get a low apparent viscosity and a thick plasma layer or the opposite (Table II).

TABLE II
DESIGN RULES FOR VARIABLE PARAMETERS THAT INFLUENCE THE APPARENT VISCOSITY AND PLASMA LAYER OF BLOOD

Variable parameters				Parameters influenced	
Haematocrit	Roughness (Rz)	Channel height	Pressure/velocity	Apparent viscosity	Plasma layer
42%	1.8 µm	50 µm	high	low	thick
44%	60 nm	100 µm	low	high	thin

ACKNOWLEDGMENT

D.M.A. wishes to thank EPSRC for Grand Challenge grant EP/C534212/1 to research “The design and manufacture of 3D-miniaturised integrated products” (“3D-Mintegration”). S.P. wishes to thank Kaiserslautern University of Applied Sciences for funding to carry out his research at Cranfield University.

REFERENCES

- [1] E.W. Merrill et al, “Rheology of human blood, near and at zero flow, Effects of temperature and hematocrit level”, *Biophysical Journal*, vol. 3, pp. 199-213, 1963.
- [2] M. Sharan and A.S. Popel, “A two-phase model for flow of blood in narrow tubes with increased effective viscosity near the wall”, *Biorheology*, vol. 38, pp. 415-428, 2001.
- [3] H. Wang and Y. Wang, “Flow in microchannels with rough walls: flow pattern and pressure drop”, *J. Micromech. Microeng.*, vol. 17, pp. 586-596, 2007.
- [4] M. Hodes et al, “Friction factors and Nusselt numbers in microchannels with superhydrophobic walls”, in: *Proceedings of the Fourth International Conference on Nanochannels, Microchannels and Minichannels*, 19-21 June 2006, Limerick, Ireland, ICNMM2006 -96134, pp. 1-11.
- [5] C-H. Choi et al, “To slip or not to slip – water flows in hydrophilic and hydrophobic walls” in: *Proceedings of the International Mechanical Engineering Conference and Exposition*, 13-16 November 2002, New Orleans, Louisiana, USA, IMECE2002-33707, pp. 1-8.
- [6] R. Lima, “In vitro blood flow in a rectangular PDMS microchannel: experimental observations using a confocal micro-PIV system”, *Biomed. Microdevices*, vol. 10, pp. 153-167, 2008.
- [7] P.A. Tipler and G. Mosca, *Physik: für Wissenschaftler und Ingenieure*, (Physics: for scientists and engineers) (2nd German ed), Elsevier GmbH / Spektrum Akademischer Verlag, Heidelberg, pp. 392-417, 2004.
- [8] A.S. Kulkarni, *Effects of surface roughness in microchannel flows*, MSc thesis, University of Florida, 2004.
- [9] D.S. Cohen and S.R. Feaster, “Rough channel microfluidic devices”, United States Patent Application Publication No. US 2007/0140913 A1, June 21, 2007.
- [10] V. Vand, “Viscosity of solutions and suspensions II”, *J. Physic. and Colloid Chem.*, vol.52, pp. 300-314, 1948.
- [11] O.K. Baskurt et al, “Erythrocyte aggregation tendency and cellular properties in horse, human and rat: a comparative study”, *American Journal of Physiology (Heart Circ. Physiol.)*, vol. 273(6), pp. H2604-H2612, 1997.