PDMS microparticle generation in a T-junction for hemodynamic studies

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Introduction

Motivation

Cardiovascular diseases (CVDs) are a class of diseases that involve the heart or blood vessels, and are the leading cause of death globally. Therefore, it is essential to understand how the heart distributes or pumps blood throughout the body. The study of hemodynamics is vitally important since the body needs oxygen to function. The aim of such studies is to assess this relationship between the cardiovascular system and the oxygen needs of the body's tissues. Such assessments are designed to allow medical professionals to make proper decisions for their patients and help diagnose and prevent CVDs. To perform such studies and in vitro experiments of blood flow, blood analogue fluids are generally used to avoid the ethical and practical considerations involved with using blood. Blood is a complex fluid composed of plasma, and formed elements such as platelets, white blood cells (WBCs) and red blood cells (RBCs). To match the rheological characteristics of blood, the blood analogue fluid should be a suspension of transparent particles with similar properties to red blood cells. Previously, rigid Polystyrene micro-particles have been used to mimic the RBCs in these suspensions. However, RBCs can adjust their shape and penetrate into narrow capillary vessels, which gives them an adaptability and consequently a complex blood rheology in different scales.

To account for the flexibility of the RBCs, we propose to use PDMS particles in the suspension. Transparency, non-toxicity, chemical inertness, thermal stability, biocompatibility are the characteristics which make PDMS an ideal choice. Here we present a study to produce PDMS micro-particles, to be used in biomimetic fluids, by droplet microfluidics using a T-junction. Conventional fabrication methods of micro-particles include mechanical agitation, resulting in wide size distribution and uneven components. Alternatively, droplet microfluidics could lead to formation of microscale droplets with exquisite control over size, shape, structure and components. As the advantages of this technology, it miniaturises the volume of reagents consumed as well as give highly monodispersed particles.

Objective

In the present study, we aim to develop a microfluidic device to produce PDMS micro-particles, which will be used in biomimetic fluids. The main objectives are as follows:

a) Theoretical simulation and understanding of micro-droplet generation
b) Fabrication of T-junction for droplet generation
c) Experimental studies for generation of deformable polymer droplets comparable to RBC size

Results and Discussion

In the T-junction device, where the continuous and dispersed phases flowing orthogonally meet at a junction producing droplets. Several parameters have been shown to influence the droplet size in T-junction devices. These include dimensionless parameters such as flow rate ratio ($Q=Q_D/Q_C$, where Q_D and Q_C are volumetric flow rates of dispersed and continuous phases, respectively), capillary number ($Ca=\mu_C U/\gamma$, where μ_C and U are the viscosity and velocity of the continuous phase, and γ is the interfacial tension), capillary number of the dispersed phase ($Ca_D=\mu_D U_D/\gamma$, where μ_D is the viscosity of the dispersed phase and U_D is the inlet velocity of the dispersed phase), Reynolds number ($Re=\rho_C U w_C/\mu_C$ where ρ_C is the density of the continuous phase and wC is the width of main channel), viscosity ratio ($\lambda=\mu_D/\mu_C$), and density ratio ($\rho=\rho_D/\rho_C$, where ρ_D is the density of the dispersed phase). In addition, geometrical parameters such as width ratio ($W=w_D/w_C$, where w_D is the width of side channel) and height ratio ($H=h/w_C$, where h is the height of the channel) can also influence the droplet size.

A two-dimensional simulation of droplet formation in T-junction geometries was performed using the volume-of-fluid (VOF) method. VOF is an Eulerian method of multiphase flow simulations where fluid properties such as viscosity and density are smoothed and the surface tension force is distributed over a thin layer near the interface as a body force. In VOF, a phase fraction parameter,

α, is used to indicate the presence of each phase at every location of the simulation domain. In our simulation, $\alpha = 1$ for phase 1 (i.e., continuous phase), $\alpha = 0$ for phase 2 (i.e., dispersed phase), and $0 < \alpha < 1$ in the interface region

. In VOF, the governing equations including continuity (Eq. (1)), momentum balance (Eq. (2)), and phase fraction equations (Eq. (3)) are solved simultaneously,

$$\nabla \cdot \boldsymbol{U} = 0, \quad (1)$$

$$\partial \rho_b \boldsymbol{U} \partial \bar{t} + \nabla \cdot (\rho_b \boldsymbol{U}\boldsymbol{U}) = -\nabla p + \nabla \cdot \boldsymbol{T} + \rho_b \boldsymbol{f} + \boldsymbol{F}s, \quad (2)$$

$$\partial \alpha \partial \bar{t} + \nabla \cdot (\alpha \boldsymbol{U}) = 0. \quad (3)$$

Parameters μb and ρb are bulk viscosity and density which are based on the weighted average of the distribution of the phase fraction,

$$\mu_b = \alpha \mu_C + (1-\alpha)\mu_D, \quad (4)$$

$$\rho_b = \alpha \rho_C + (1-\alpha)\rho_D. \quad (5)$$

The continuous phase is a mixture of glycerol and water (1:1 wt/wt). The dispersed phase is a mixture of PDMS and t-Butyl Alcohol (1:3 wt/wt).

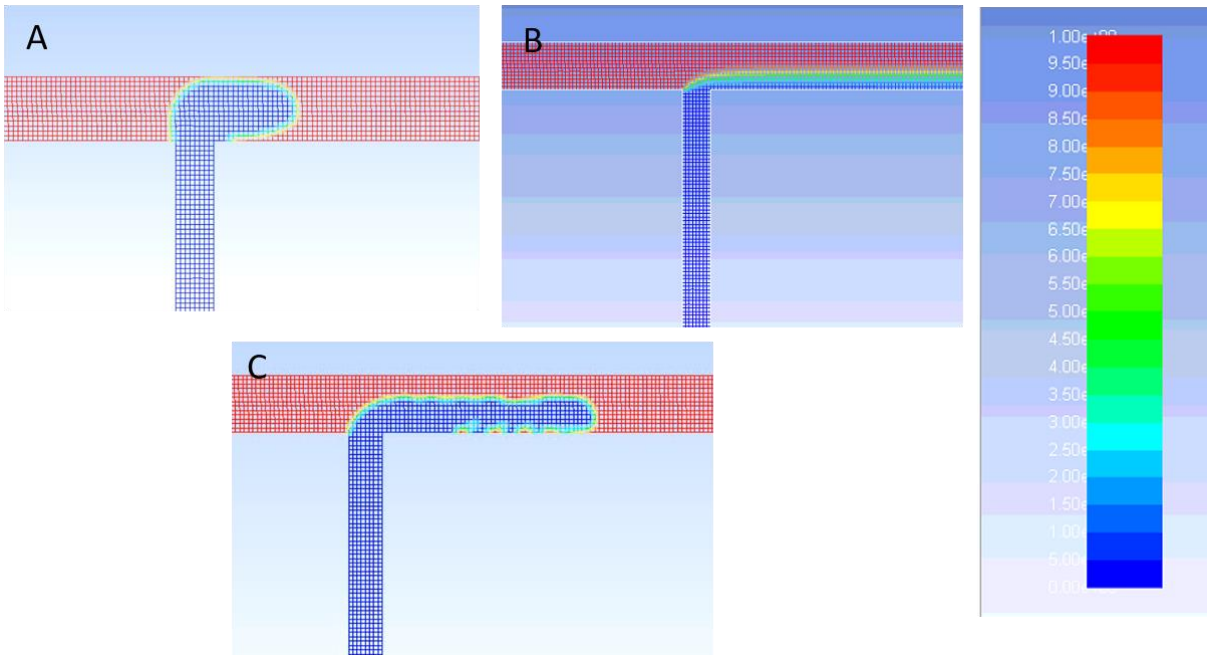

Figure 1. Computational results for three cases. A (Q_C/Q_D = 0.85) represents the channel being completely filled by dispersed phase, no droplet generation. B (Q_C/Q_D = 10) represents the dispersed phase forming a laminar flow in a separate layer, no droplet generation. C (Q_C/Q_D = 5) represents jetting regime, droplet generation.

Conclusions

Simulations as described in this work, enable us to predict the optimum flow ratio. This allows us to effectively perform the experiments.

Further Scope

The future steps would involve performing experimental studies for generation of deformable polymer droplets comparable to RBC size.

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