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Design and Analysis of Microchannel for Plasma Separation from Blood

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Abstract: Microchannel device is important and having great use in field of biological research. The micro channels having some advantages like reduction in the sample processing time, consumption of costly reagents and sample volumes. The current work for microchannel is based on the Zwiefach- fung bifurcation law which used to study for separation process. The microchannel based plasma separation is far better and rapid than conventional plasma separation techniques like centrifugation, filtration.

Microchannel flow simulation is carried out using CFD software. The analysis is done for different type channel with combination of various sets of different shape and size of obstacles placed along the channel walls. For this analysis we used computational fluid dynamic tool known as COMSOL MULTIPHYSICS 5.0, to study different configurations on pressure and velocity drop. Also, the effect of geometry, cross sectional area are studied. While analysis different obstacles placed along the channel like rectangular, semicircle and triangular are used. The analysis is done for each shape obstacle and observation are recorded while analysis it is observed of that pressure is decreases with increase in number of obstacles.

Keywords: Micro-channel, pressure drop, rectangular obstacles, velocity distribution, pressure distribution, flow rate

I. INTRODUCTION

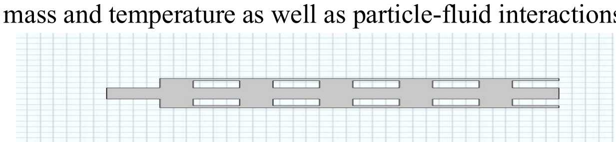
Micro-fluidics is a branch of science that includes the study of fluid flow in channels whose dimensions are in micrometers. It has a wide range of research as microfluidic tools it provides applications in pharmaceuticals, biomedical, lab-on-chip(LOC), drug delivery and DNA sequencing, etc. This technique has the advantage over the standard bench-top method as it requires a low volume of reagents and cost Micro-mixer is one of the very essential tools in microfluidics. It is a device that mixes two or more fluids at the micro-scale. The main purpose of microfluidic mixing is to achieve fast, complete, and through-out mixing of multiple fluids. In many devices mixing is achieved by enhancing the diffusion effect. There are two types of micromixers as active and passive type. In the passive type, micromixer diffusion Is achieved by external force or disturbance. In the passive type, diffusion is achieved by improving area contact, structure configuration, or adding obstacles and baffles in the flow path. In simple T and Y shape mixing take place by slow molecular diffusion. This mixing does not require any external source for mixing. Various research data showed that Low Reynolds number (Re<20) causes that mixing is only by diffusion so mixing efficiency is low. Also by increasing Reynolds number mixing index is also increased for proposed designs. The final result showed that the YRCSAR mixing efficiency is 99% at Re=100(Imran Shah et.al,2018). Therefore to improve mixing performance this research is concentrated on micromixers with different obstacles. The high values of wall shear rate, in the crossing zone, confirm the near-wall disturbance created by the impingement of flow and the appearance of vortices which allowed to improve the fluid mixing. The results of the study showed that the ┼ shaped system can improve the mixing process in comparison with the micromixers having ┬ geometry (Nassim Ait Mouheb et. Al, 2009). Also, the mixing can be improved by suitably choosing geometric parameters of the groove, such as the depth, the width, and the oblique angle (Yeng-Yung Tsui et. al,2008). The early initiation of vortices in the inlet streams of a T-shaped micromixer by either symmetric or antisymmetric VIOs in the inlet channels can lead to improvements in mixing efficiency(Chih-Yang Wu et. al,2020). The micromixer based on the SAR process is more effective at low Reynolds numbers than devices based only on molecular diffusion(Mohammad Nimafar et. al,2012) therefore the research design includes models based on the SAR technique. Micromixers with ellipse-like micropillars have a well-mixing status when their mixing efficiency is higher than 80% as $\leq$ 1. This micromixer has shown the improved performance of the previous micromixer (T-mixer) at a low Reynolds number(Tran-Minh et. al,2014). The flow regime of symmetrical 1:1 mixing in T-shaped micromixers ranged from straight laminar flow for Re numbers lower than 10 over vortex creation to symmetry break up of engulfment flow. Also for Re numbers larger than 240, the flow showed temporal fluctuations at the entrance of the mixing channel, starting with periodic pulsations.

With increasing Re number, the regularity of the pulsations is disturbed by higher frequencies leading to chaotic flow behavior at Re = 1000(Norbert Kockmann et. al,2007). A bend in the microchannel induced stirring and it also enhanced mixing performance. In case of square wave microchannel number of bend is higher than the zig-zag microchannel therefore it has higher mixing efficiency (Sharhawat Hassain et. al,2009). The mixing analysis of different configurations of channel under various operating conditions were carried out using CFD software (Shinde et al). The CFD based results showed that the mixer efficiently stirs the fluids, generates secondary vortex flows in the microchannels which results in higher mixing index. The results also revealed that baffles can break the fluid streams, produce fluid convection and increase the contact area of the fluid by folding and deflecting fluid streams which in turn helps to improve the mixing index (Ranjitsinha R. Gidde et. al,2019). Also because of the reversed area, the mixing performance of TPZ has an highlighted advantage over the zigzag micromixer's and mixing index are all beyond 93% for a wide range of Re ($Re \geq 5$ or $Re \leq 0.5$). The reversed area enhanced the intensity of molecular diffusion. Also the pressure drop in TPZ is bigger than the zigzag micromixer at each Re inevitably (Xueye Chen et. al,2016). The mixing analysis of different configurations of channel under various operating conditions were carried out using CFD software (Shinde et al). With the respect of pressure drop, teardrop obstacle-based micromixer showed less pressure drop. Both mixing index and pressure drop increase with increase in aspect ratio of the obstacle (Ranjitsinha R. Gidde,2019). Various results also showed that , the mixing angle, or equivalently the tilt angle, in arrow mixers also has a strong impact on the mixing performance at relatively high values of Re in unsteady periodic regimes, and it appeared that a small downward tilt angle leads to an improvement in the mixing performance compared with classical T mixers that is robust toward variations in the operating conditions up to the transition to chaotic dynamics(Simone Camarri et. al,2020). It is necessary to improve mixing performance and study the effect of various parameters because of its applications in the chemical industry, fuel cells, micro-reactors, and biomedical. Therefore, in the present work, we studied different types of micromixers consisting of rectangular, elliptical, and circular obstacles for comparing the mixing performance of each micromixer using COMSOL Multiphysics 5.0. Also, various parameters such as pressure drop, velocity, and pressure distribution were studied in this paper.

II. CFD MODELLING and FORMULATION

A. Rectangular Microchannel

For our model we use 2d geometry then we select two physics first one is creeping flow and second is particle tracing for fluid flow. The creeping flow is used for the creeping Flow interface is used for simulating fluid flows at very low Reynolds numbers for which the inertial term in the Naiver-Stokes equations can be neglected. This single-phase flow type is also referred to as Stokes flow and occurs in systems with high viscosity or small geometrical length scales (for example, in microfluidics and MEMS devices). The fluid can be compressible or incompressible, and Newtonian or non-Newtonian. The equations solved by the Creeping Flow interface are the Stokes equations for conservation of momentum and the continuity equation for conservation of mass. The Creeping Flow interface can be used for stationary and time-dependent analysis. The particle tracing for fluid flow is applied for the Particle Tracing for Fluid Flow interface is used to compute the motion of particles in a background fluid. Particle motion can be driven by drag, gravity, and electric, magnetic, and acoustophoretic forces. User-defined forces can be added. It is also possible to compute the particle mass and temperature as well as particle-fluid interactions.



B. Parameters

In model we create 2D geometry having length 2000micron and height 400 micron. The model has inlet and 3 outlet out of them 2 outlet for plasma separation and one for cell collection. It has 5 set obstacle inside. The ratio of inertia force and viscous force, formulated as

$Re = \rho\ v\ d\boldsymbol{h}/\ \mu$

dh= 4∗Area of microchannel/ Perimeter of microchannel

v= Total Flow rate/ area of the microchannel

dh is the hydraulic diameter

v is the average velocity

ρ = density of blood

μ = viscosity of blood.

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Sr.No	Name	Expression	Description
1	sigma_f	55[mS/m]	Fluid medium conductivity
2	rho_f	1000 kg/m^3	Fluid density
3	mu_f	1e-[Pa*s]	fluid dynamic viscosity
4	rho_p	1050[kg/m^3]	Particle density (RBC, plasma)
5	dp1	1.8[um]	particle diameter plasma
6	dp2	5[um]	Particle diameter RBC
7	N	2000	Number of particle per release

C. *Meshing of Geometry*

The fine meshing method is used for meshing the geometry in Comsol Multiphysics. The 2D geometry of the particles inside microchannel with extremely fine mesh. Material properties are defined in the material database in Comsol Multiphysics

III. PERFORMANCE PARAMETER

A. Pressure Drop

One of the most important design considerations for a micro-fluidic channel design is to minimize the pressure drop. The equation used to determine the pressure drop is as given below,

$$\Delta P = \left[\frac{F \times RE}{2}\right] \times \left[\frac{\mu \times V \times L}{D_h^2}\right]$$

Where p is the pressure difference within the channel, L is channel length in mm, μ is the dynamic viscosity of a fluid in Pa.s, V is the average velocity, (f×Re) is Poisseuille number. Also, pressure drop can be evaluated in simulation analysis as shown below,

$$\Delta P = P_i - P_o$$

Where P_i is inlet pressure and P_o is outlet pressure. The difference between inlet and outlet pressure will give the pressure drop.

B. Velocity of flow:

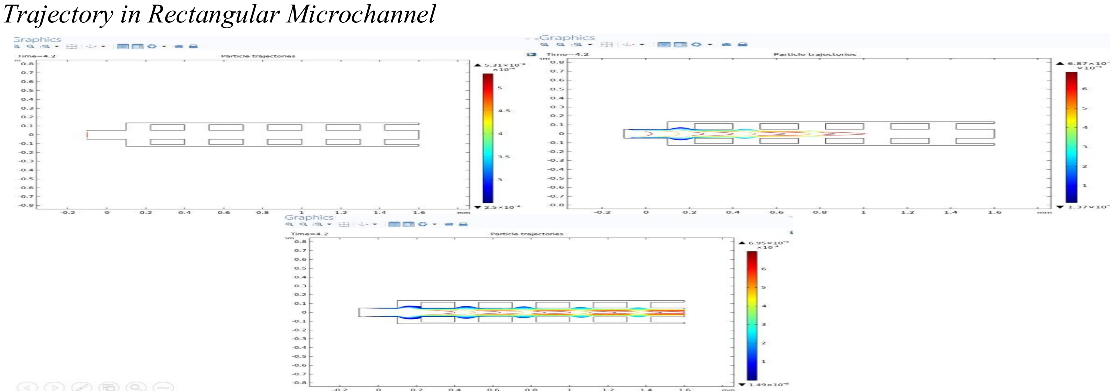
When a fluid is moving through a microfluidic channel, we can see that there is a velocity gradient across the cross-section of the channel. Since a fluid is defined as a substance that will deform continuously under the application of shear stress and know that the flow structure of fluid in a microfluidic channel is laminar. Newton's law of viscosity can be used to show the relationship between shear stress and deformation rate as below,

$$\tau = \mu \frac{du}{dy}$$

Where τ is the shear stress, μ is the viscosity of the fluid, u is the velocity of the fluid in the x-direction and du/dy is the velocity gradient in the y-direction in the above expression. This results in the highest velocity being in the portion of the fluid that is furthest from the wall. Also, there is a no-slip condition for the portion of the fluid that is adjacent to the wall of the channel where the shear force is the greatest.

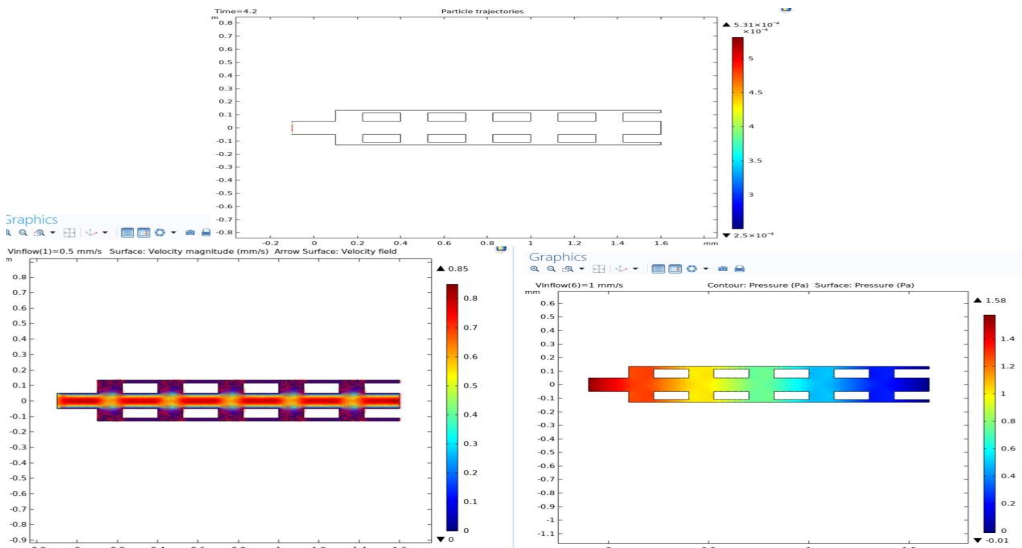
IV. SIMULATION AND RESULTS:

A. Particle Trajectory in Rectangular Microchannel



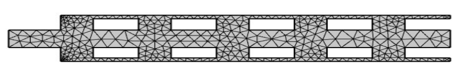
The above figures show the blood flow in microchannel having rectangular obstacle.

B. Pressure and velocity Drop across Micromixers



In the above figures velocity effect on microchannel is from 0.5mm/sec to 1mm/sec. As model obey Bifurcation law the different velocity regions created inside the microchannel. The axial region having high velocity and outer boundary having low velocity.. As blood flow inside channel it collide on obstacle. It results in creating vortex near the edges of obstacles the vortex create pressure difference inside the channel hence dense particles of blood settle down and move along high velocity channel which is in axial direction and low-density components move alongside the wall. The Pressure drop distribution with different inlet velocities (a) 0.5mm/sec (b) 0.6 mm/s (c) 0.7 mm/s (d) 0.8mm/s (e)0.9 mm/s (f)1mm/s. The red region inside microchannel describes the high pressure at the beginning of microchannel and it will get drop at outlet as the blue region in microchannel.

C. Meshing of Particles in Rectangular Microchannel



The above figures show meshing of blood particles in microchannel having rectangular obstacles. In this we used a control meshing which is fine meshing method. Finer the mesh more the accuracy.This mesh contains 12158 elements for getting accurate result in meshing we done corner refining and sizing of elements.

D. Comparison of velocity and preassure along wall and outlet of microchannel:

1) For Microchannel with Rectangular Obstacle

Flow rate mm/sec	Velocity Mm/sec		Pressure Pascal	
	axial	Along wall	beginning	outlet
0.5	0.65	0.15	0.65	0.15
0.6	0.72	0.1	0.7	0.09
0.7	0.8	0.08	0.9	0.1
0.8	0.9	0.09	1	0.16
0.9	0.9	0.2	1.1	0.2
1	1.3	0.3	0.3	0.15

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The table (I) shows various results such as pressure, velocity, flow rate, along the wall, at beginning and outlet for microchannel with a rectangular obstacle. When the flow rate is 0.5 mm/sec the velocity along axial is 0.65 mm/sec and along the wall it is 0.15 mm/sec corresponding pressure at beginning stage is 0.65 pascal and at outlet it is 0.15 Pascal. Similarly for 0.6 and 0.7 mm/sec flow rate the velocities are 0.72 and 0.8 and along wall they are 0.1 and 0.08 respectively the pressure at beginning are 0.7 Pascal and 0.9 Pascal and at the outlet they are 0.09 pascal and 0.1 Pascal respectively. for 0.8 and 0.9 mm/sec flow rate the velocities are 0.9 and 0.9 and along wall they are 0.09 and 0.2 respectively the pressure at beginning are 1 Pascal and 1.1 Pascal and at the outlet they are 0.2 pascal and 0.16 Pascal respectively. Hence and at last when the flow rate is 1 mm/sec the velocity along axial is 1.3 mm/sec and along the wall it is 0.3 mm/sec corresponding pressure at beginning stage is 0.3 pascal and at outlet it is 0.15 Pascal.

V. CONCLUSION

From various results, we have found that size, the geometry of mixing chambers and obstacles and pressure and velocity. As the floe rate increases the velocity at axial part of microchannel also increases but at the wall side i.e along the wall velocity decreases. Also the pressure goes on decreasing from inlet to outlet i.e there is pressure drop across the microchannel from beginning to end. Hence considering all the parameters the flow rate of 0.5 mm/sec is suitable in terms of velocity, pressure and efficiency of microchannel.

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