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DESIGN OF NOVEL RECIRCULATION SYSTEM FOR SLOW REACTING ASSAYS IN MICROFLUIDIC DOMAIN

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A simple design for a microfluidic flow system for use in mixing or reacting assays with limited sample availability has been proposed and analyzed using multiphysics simulation. The design is based on differential electroosmotic flow concept which has facilitated a number of interesting flow phenomena in micro-domains. For an average potential drop of about 86 kV/m in the setup, flow rates of about 0.7 nl/s were obtained. The component consists of a microchannel (~20 microns in width) designed to have different electroosmotic mobility in different sections to give a continuous, closed circuit flow. The system has been designed for continuous recirculation of the flow and hence appropriate for use as a micromixer or solid phase microreactor where there is a requirement for high residence time due to slow reaction/diffusion rates.

1. INTRODUCTION

Microfluidic devices are important components of chemical and biological sensors proposed for many applications. According to the operational functions, biochips can be classified into categories, such as gene chip, lab-on-a-chip, and protein chip. Among them, lab-on-a-chip is an application of miniaturized total analysis system (μTAS) first proposed by Manz et al [1] which integrate traditionally used equipments in biochemical industry such as micro-pumps, micro-actuators, micro-valves, microsensors and micromixers.

The mixing or reaction of liquid samples with possibly solid reagents in microfluidic assays with limited sample availability and high reaction time is a challenging problem since it requires continuous recirculation of the liquid sample. The paper demonstrates a design of the novel recirculation system that can be employed in solid phase microreactors as well as liquid-liquid micromixers. The device can function on a constant DC electric supply of about 12-20 Volts depending on the required speed of recirculation.

2. DESIGN & CONCEPTUALIZATION

2.1 Concept

Consider a circular closed loop microchannel as shown in fig. 1 with a small opening that is small enough to prevent excess outflow but large enough to prevent negative pressure drops in the loop. Two electrodes are placed at diametrically opposite ends of the circular loop. The walls of the loop on the left side of the two electrodes have a suppressed electroosmotic flow. This can be achieved either by coating of a viscous polymer coating [7] or the zeta potential is suppressed by other methods, like FEFC [6].In either case, electroosmotic flow is suppressed in the left side of the loop, which means that the wall does not form a sufficiently large electric double layer to actuate motion at the wall boundary. The method described in [6] realizes the scene by reducing the zeta potential whereas [7] describes ways of providing high viscosity coatings to counter the shear force created by EOF at the walls.

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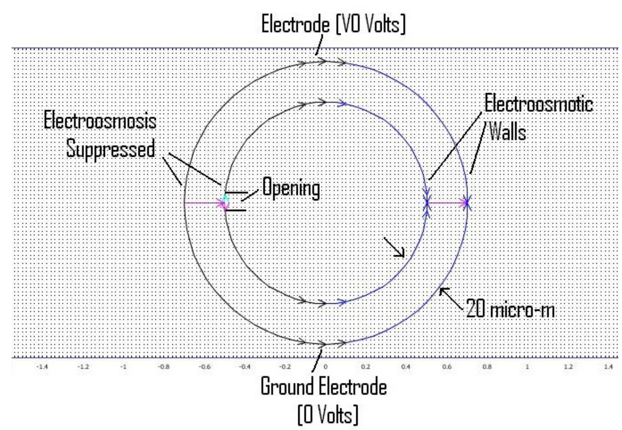

Fig. 1. Schematic of the recirculation system. Aspect ratio of the channel is unity.

The net effect of such an arrangement is that there is a continuous flow of fluid in a closed circle, a phenomenon which is difficult to obtain through simple pressure driven arrangements or otherwise and yet has very important applications as explained earlier.

2.2 Physics of EOF

Electroosmotic flow is caused by the movement of the electric double layer in the direction of an applied electric field. The direction as well as speed of flow is dependent upon the Zeta Potential of the electric double layer. From the point of view of modeling of electroosmotic flow [3], the end outcome of this coupled electrostatic and fluid flow phenomena is the movement of the shear layer in along the applied electric field. This results in a 'slip' boundary condition at the electroosmotically active walls, with a non-zero velocity tangential to the wall, given as[4][5][6]:

$$U_{eof} = -(\quad / \quad)\times E \qquad (1)$$

Where, is the electric permeability, is the dynamic viscosity of fluid, is the Zeta Potential and E is the applied electric field. As is seen that the resulting slip velocity is proportional to the applied electric field, the proportionality term is called as the electroosmotic mobility [2] μ_{eof}. Given as,

$$ _{eof} = - \quad / \tag{2}$$

Wherein from (1) and (2) we have,

$$U_{eof} = \ _{eof}(\quad) \times E \tag{3}$$

From (3), it seems advantageous to exploit the dependence of slip velocity on the zeta potential as well as the electric field. It implies that under the action of the same electric field, a number of parallel microchannels can have different slip velocities. The zeta potential in usually settles near to about -0.1 V [2].

3. MODELING & SIMULATION

The device idea is modeled and simulated on COMSOL Multiphysics software. A brief description of the simulation methodology in the context of the above proposed design is presented next.

3.1 Geometry & Mesh

The simulation was setup using a 2D solver with a third axis depth of 20 micron. The geometry was a fairly simple annular region as shown in fig. 1. Meshing was done using triangular elements. A boundary layer mesh was setup near the walls so that the shear layers due to electroosmotic velocity could be resolved. The channel width is 20 microns, and the inner diameter of the annular flow domain is 100 microns.

3.2 Material Properties

The standard fluid properties for the simulation were of water. Some important properties pertaining to electroosmotic behavior of water are noted below.

Material Properties	
Electrical Conductivity [sigma]	0.01 [S/m]
Relative Permeability [eps_r]	78.5
Density [Ro]	1000 [kg/m^3]
Viscosity [eta]	0.001 [Pa-s]

Table 1: Material Properties data for multiphysics simulation

3.3 Solution Scheme

To simulate the EOF boundary condition the value of zeta potential or electroosmotic mobility is entered as parametric value. The solver calculates a slip velocity at the walls boundaries from (1) or (3). The slip velocity obtained is used for bounding the Navier Stokes solution and obtaining the flow properties. The electric field used in (1) or (3) is calculated by solving the second order differential equation for electric potential or the Poisson’s Equation. This equation is bounded by assigning fixed voltages on domain boundaries.

3.4 Boundary Conditions

As seen in fig. 1, two regions in the domain are assigned fixed voltage values, 0 (ground) and V0. V0 has been varied as a parameter from 0 to 30 Volts. Further, the electroosmotic mobility has been set to zero to model suppressed EOF on the left arm. The right arm has a non-zero electroosmotic mobility given the table 2 along with other relevant boundary conditions. The opening on the left, inner walls is set at atmospheric pressure. Though the opening receives negligible flow, it is essential for a numerical simulation to converge the solution to steady state.

Boundary Conditions	
Applied Voltage [V0]	12 [V]
Back Pressure at Outlet [p0]	0 [Pa]
Electroosmotic Mobility [mu_eo]	0.06 [mm^2/V.s]
Device Thickness in third dimension [d]	20 [μm]

Table 2: Default boundary condition values. Used if variable is not parametrically varied.

4. RESULTS & DISCUSSION

4.1 Preliminary Results

A parametric simulation was run for values of V0 between 0 and 30 V with steps of 3 V each. Contour plots, and other data was extracted for value of V0 = 12 V. Considering a potential drop of 12 V within 140 micron (distance between diametrically opposite ends), the average electric field in the channels would be in the range of 86 kV/m. Fig. 2 shows the contours of the derived velocity field.

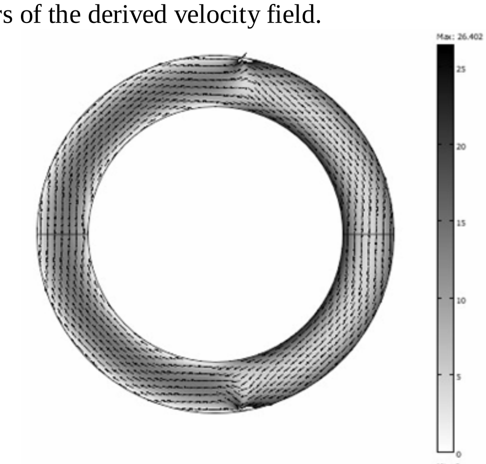

Fig. 2. Contours of Velocity field.

Notice that the region near to the electrodes reaches very high velocity values about 5 times the average velocity in the rest of the channel. This is because of the intense electric field prevalent in that area. Placement of electrodes is a critical aspect in EOF pump fabrication.

4.2 Design Issues& Performance

The applied voltage is the driving force for the flow and thus is an important parameter in the design of the perpetual flow system. Fig. 3 shows the variation of flow rate measured at a cross-section in the right arm (see fig. 1) plotted against different values of applied voltage [V0].

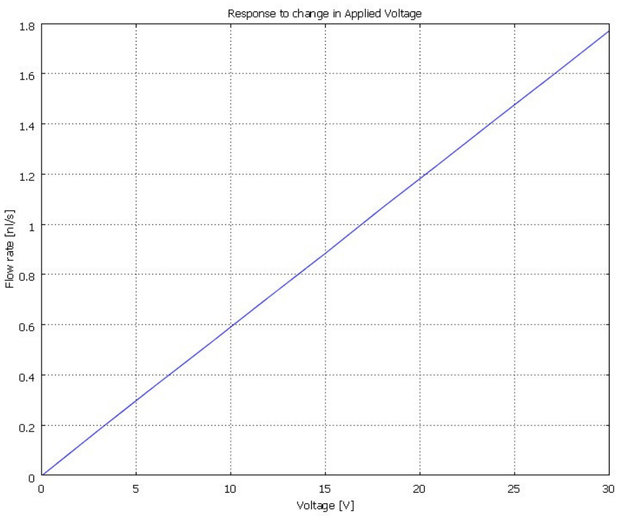

Fig. 3. Flow rate (nl/s) at cross-section Vs. Applied Voltage (V0 Volts)

The restriction on applied voltage is safety and possibility of dielectric breakdown of the walls, etc. Hence an optimum value has to be chosen according to requirements. Second consideration is of the velocity profile. Fig. 4 plots the velocity profile at cross-section A for different applied voltages. The value of 1 for normalized channel width pertains to the outer boundary. The asymmetric behavior can be easily explained by considering the distribution of electric field. The electric field will be most intense along the line joining the electrodes. The farther the point away from this line, the lesser is the intensity. Since the slip velocity at the walls is proportional to electric field, the outer walls experience a lower slip.

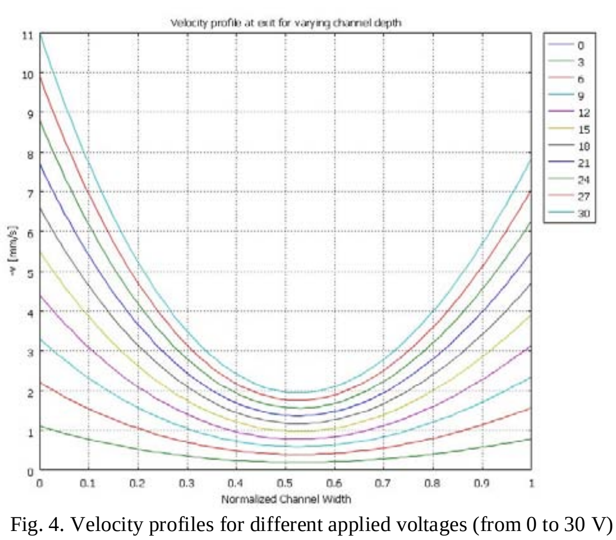

Fig. 4. Velocity profiles for different applied voltages (from 0 to 30 V)

4.3 Applications

It can be seen from fig. 2 that the arm on the left side has a flow with velocity maximum in the center of the channel, as in the case of pressure driven flow, whereas on the right arm, the velocity peaks near the walls and vanishes towards the center. This phenomenon occurs since the electroosmosis has been suppressed in the left arm. Also, in the region near the electrodes, the velocity vanishes in the center of the channel. This is due to the fact that there is a sudden stoppage to the electroosmotic effects and fluid accumulates in this region. A pressure gradient (drop) is formed from the ground electrode to the electrode V0 which drives the liquid in the left arm.

As an end result, a perpetual-looking flow of fluid has been obtained, wherein the same quantity of fluid moves in a closed circle. Applications of such flows can be found for mixing [8] and reacting a finite quantity of liquid with another liquid or for surface reactions which are slow in nature and require high residence time. Since liquid does not leave the reaction chamber, but is fed back to the chamber, a high residence time is obtained and at the same time provides turbulence and recirculation.

5. CONCLUSION

Since electroosmotic flow is a phenomenon that is shear driven, it has peculiar properties that can be exploited to create unique flow scenarios. A novel design was conceptualized for a perpetual flow microsystem that is driven by electroosmotic effects.

The perpetual flow microsystem has applications in various fields especially micro-reaction engineering. The current design allows flow rates in the 1 nl/s range which is a significant achievement.

6. REFERENCES

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