Microdevices for Continuous Sized Based Sorting by AC Dielectrophoresis

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Abstract In this project, various microfluidic devices are designed for microparticles and cells separation. The dominant regions of dielectrophoresis among other AC Electrokinetic transport mechanisms are predicted. Flow behaviors of particles for each design parameter are modeled and simulated using COMSOL Multiphysics 4.3a software. Lab-on-a-chip devices having a Ti interdigitated electrode layer on a glass substrate and a PDMS microchannel are fabricated to investigate the most effective design solution for separating particles based on their sizes. Polystyrene particles with different diameters of 3.2 μm and 9.8 μm are used in our experiments and experiments with circulating tumor cells (CTCs) are in progress.

Keywords: AC Dielectrophoresis, Field-Flow-Fractionation, Microfluidics, Lab-on-a-Chip

1. Introduction

Lab-on-a-Chip devices (LOCs) create inexpensive and efficient solutions by integrating several laboratory functions on a single chip of only a few cantimeters in size. LOCs are used in biomedical and environmental applications, such as, fluid transport, drug delivery, diagnosis of viruses and bacterias and manipulation of particles for focusing and separation applications. Particle movements in LOCs can be observed without labelling which puts this technique forward for biological cell and submicron scale particle separations.

Dielectrophoresis (DEP) is a force acting on a dielectric particle when it is subjected to a non-uniform electric field of either DC or AC field. Contrary to electrophoresis (EP), suspended particles do not need to have a surface charge, DEP only occurs when there are induced charges in non-uniform field. A dielectric particle polarizes under a non-uniform electric field and there is a net dipole moment acting on it [1,2].

The time averaged dielectrophoretic force on a particle is given by [3]:

$$F_{DEP} = 2\pi\varepsilon_0\varepsilon_m a^3 Re(f_{CM})\nabla E^2 \quad (1)$$

Where ∇E^2 is the gradient of the root mean square of the electric field, $f_{CM}$is the Clausius-Mossotti (CM) factor,a function of particle and fluid properties, determines the direction of dielectrophoretic force and Re(…) indicates the real part of CM factor, a is the diameter of the particle [3].

There are two ways that particles interact with non–uniform electric fields: when the local electric field lines are intense particles are trapped in these regions which is called positive dielectrıphoresis (pDEP) or particles are pushed away from these regions which is called negative dielectrophoresis (nDEP). The electrical properties of particles and the frequency of the applied electric field determine the behavior of DEP.

Within this project, various microfluidic devices are designed for separating microparticles and cells. Flow behaviors of particles for each designs are modeled and simulated using COMSOL Multiphysics 4.3a software. Polystyrene particles with different diameters of 3.2 µm and 9.8 µm are used in our experiments and the experiments are planned to be conducted with circulating tumor cells (CTCs). An array of interdigitated electrodes are used to create electric field gradients along microchannels and placed at 45° angle to repel the particles from the electrodes. Total force caused by both hydrodynamic drag force and nDEP on particles causes them to deflect along the electrode geometry. Through the length of the device, larger particles deflect more than the smaller particles according to the formula of F_{DEP} and that makes the separation of particles in different sizes possible.

2. COMSOL Simulations

Two different geometries are modeled in simulations: One, G_1, has microelectrodes only on one side of the microchannel, that is easier to fabricate but less efficient in case of separation which will be discussed further on. The other geometric model, G_2, has microelectrodes on both side of the microchannel which is distinctively more efficient but the fabrication steps are complex. Electrostatic simulations are carried out in 2D geometry which is a crossection of the microfluidic channels as seen in Figure 1, to investigate the effects of dielectrophoresis inside the microchannel. Applied boundary conditions are also given in the same figure. The width, w, and the gap, g, of the electrodes are 50µm, the height of the electrodes is 0.4µm, the height of the microchannel for G_1, Figure 1(a), is 50µm and for G_2, Figure 1(b), is 100µm. Initial potential amplitude is set to φ_e=3.546V.

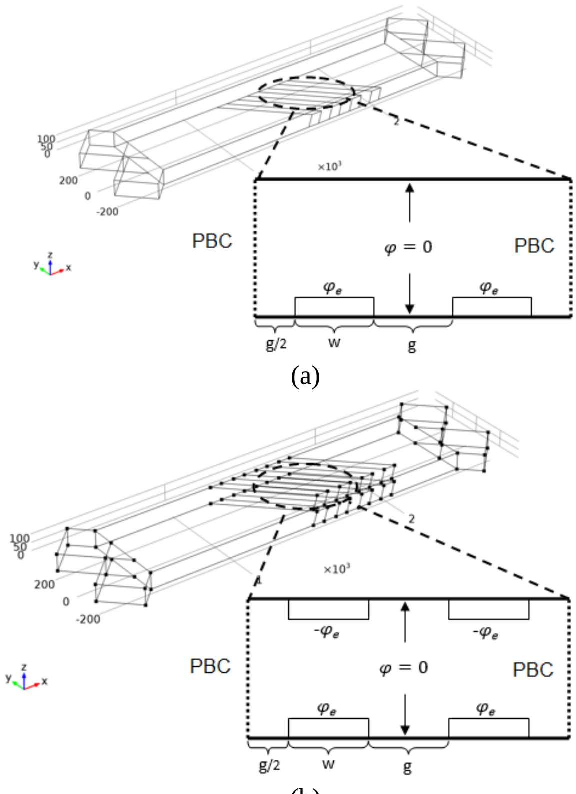
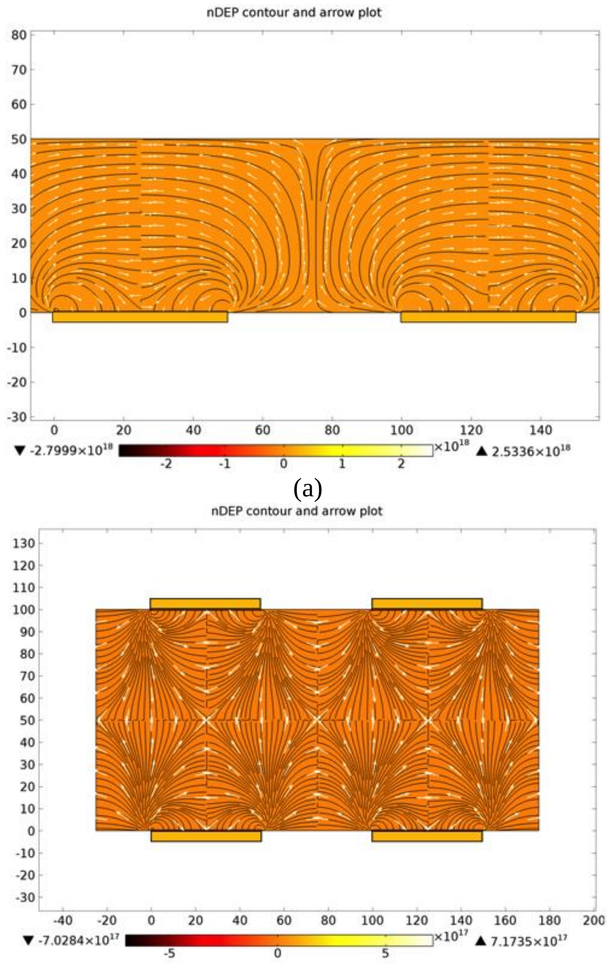


Figure 1: Boundary conditions for 2D electrostatic simulations: a) Electrodes only on one side and the height of microchannel is 50µm. b) Electrodes on both sides and the height of microchannel is 100µm. Periodic boundry conditions are applied for both geometries since they repeat along the x direction inside the microchannel.

In Figure 2, ∇E^2 demostrates one of the most important arguments of dielectrophoretic force is shown when nDEP is effective on particles. As expected particles are pushed away from the regions where ∇E^2 is maximum at the end points of the electrodes. For G_1, Figure 2(a), particles subjected to nDEP aresticked to the upper wall during the flow which both cause contamination inside the channel and decrease the efficieny of separation. For G_2, particles are condensed in the middle of the channel height which results minimum interaction with the channel walls and electrodes.



(a)

(b)

Figure 2: Color bar: the amplitude of ∇E^2, contour and arrow plot shows the direction of ∇E^2 when nDEP is effective on particles. Yellow boxes outside the channel represents the positions of microelectrodes.

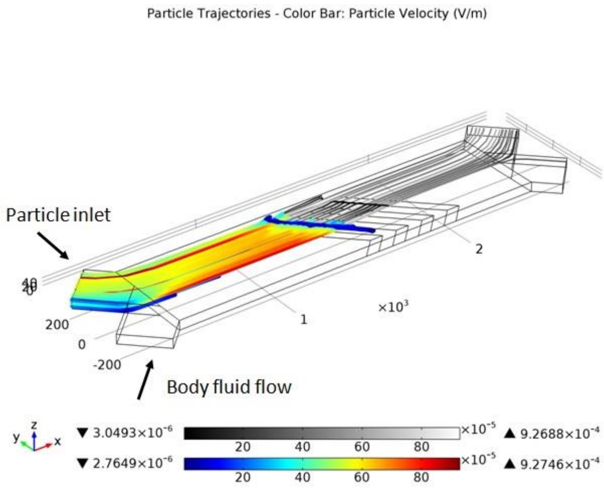
The results of 2D models are valid when the electrodes are parallel to each other but the direction of total force is different when the electrodes are oblique. Furthermore, since there is a drag force on particles caused by hydrodynamic forces we also modeled 3D geometries to investigate the particle flow under continuous fluid flow and similar electrostatic conditions which is applied to 2D models previously. For both geometries the length of the microchannel is 2mm and the width is 0.5mm and there are 2 inlets, one for the particle mixture which includes 3.2µm and 9.8µm polystyrene particles and the other for the body fluid flow. For G_1, there are only 4 microelectrode fingers on one side where G_2 has 4 electrodes on one side and 4 electrodes on the opposite wall to focus the particles in the middle of the microchannel. There is also a "v" curve on electrodes which focuses the particles through the desired outlet and prevents the interaction with sidewalls.

Electrostatic forces and fluid flow are analyzed in both of the models and particle flow is integrated into the same study to investigate the time-dependent particle velocities and positions. Initial and continuous fluid flow is set to 0.8 [µl/min] and no slip boundary conditions are applied to walls. Material properties used in our models are given in Table 1. Considering the CM factor, as the permittivity of medium is greater than the particles, nDEP is effective in our models.

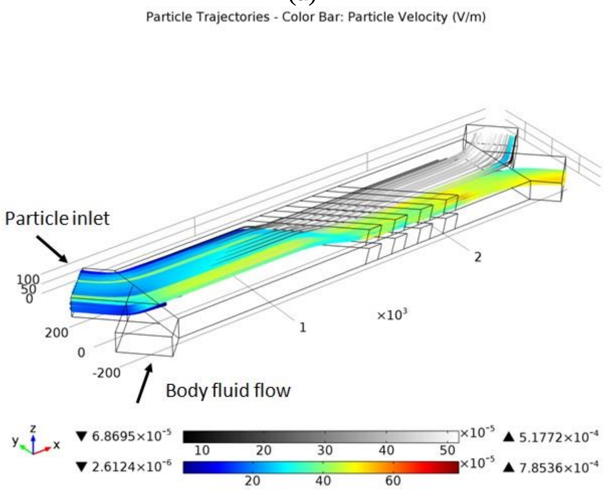
Table 1: Material properties used in our models.

Material	Properties	Value	Unit
DI-Water	Density	1000	kg/m^3
DI-Water	Dinamic viscosity	0.001	Pa.s
DI-Water	Permittivity	120	
Polystyrene Particle	Particle radius	3.2 - 9.8	µm
Polystyrene Particle	Density	1050	kg/m^3
Polystyrene Particle	Permittivity	2.55	

Final results of 3D models is given in Figure 3. Grayscale trajectories represents 3.2µm polystyrene particles for 10 seconds after initial fluid flow and rainbow color trajectories represent 9.8µm polystyrene particles for the same amount of time.



(a)

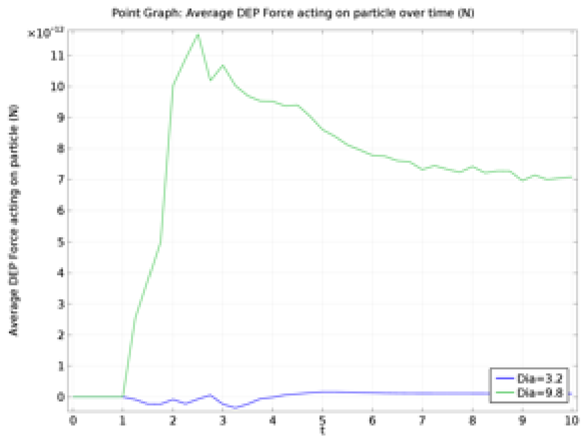


(b)

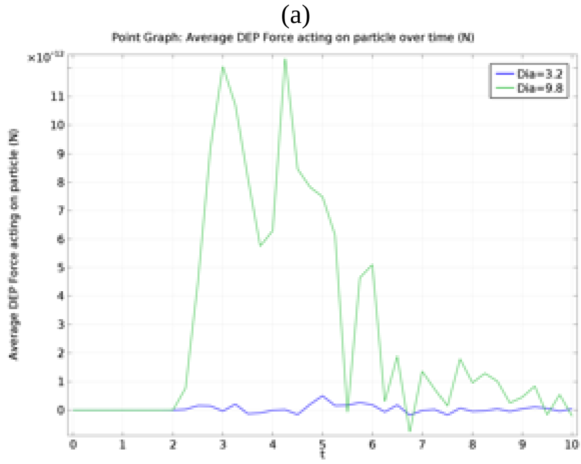
Figure 3: Time dependent particle trajectories for the 10th second. a) for G_1 and b) G_2.

For both geometries, particles having smaller diameters exit the microchannel without changing their linear path. In contrast, particles having larger diameters are deflected along the electrode geometries and separated from the particle mixture. In Figure 3(a), particles having 9.8µm diameter in G_1 could not reach to channel outlet since they stick or disappear under the command given by COMSOL itself when they interact with the walls, but still one can clearly see that they are deflected along (–y) direction. Since the interaction of the particles with the walls are prevented in G_1, both particles exit the channel from different outlets without any problems.

For further investigation, the time dependent average DEP force acting on particles is given in Figure 4. Since the channel of G_1 is narrower than G_2, the fluid and the particle velocities are higher for this model and particles interact with the DEP force one second earlier compared to particles in G_2. In Figure 4(a), 9.8µm particles are deflected until they reach to upperwall and at that point DEP force acting on them remains nearly linear. In Figure 4(b), 9.8µm particles are continuously deflected along the electrode geometries and DEP force reaches to zero when particles enter the channel outlet. The average force on 3.2µm particles for both geometries remains nearly zero along the microchannel which results with no deflection.



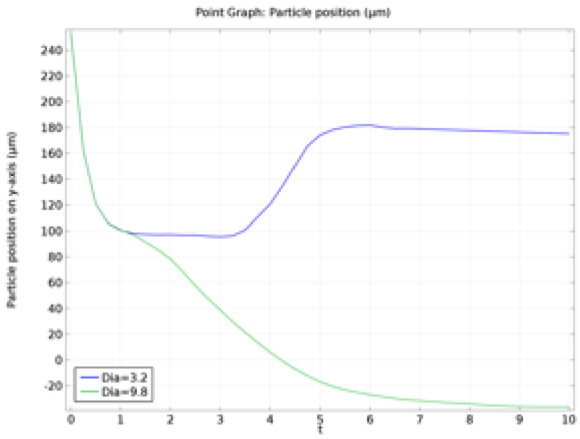
(a)



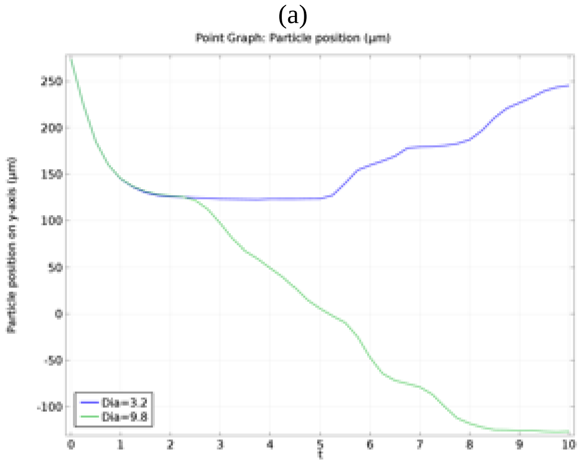
(b)

Figure 4: Average DEP force acting on particles over time for a) G_1 and b) G_2.

A detailed view of displacement of particles in y coordinates are given in Figure 5.



(a)



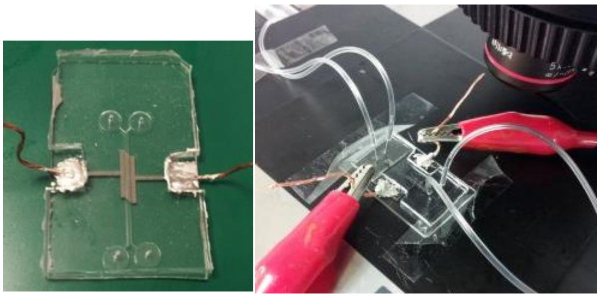
(b)

Figure 5: Displacement of particles in y coordinates over time for a) G_1 and b) G_2.

Figure 5 correlates with previous results and indicates that both designs successfully separates particles based on their sizes.
LOCs having a Ti interdigitated electrode layer on a glass substrate and a PDMS microchannel are fabricated to investigate the most effective design for separating particles based on their sizes. The standard photolithography techniques and lift off processes are used in order to obtain the LOC devices.

3. Experimental

In this study, standard photolithography techniques and lift off processes are used to obtain the geometry G_1. The final device which has a 200nm thick Ti interdigitated electrode layer on a glass substrate and a PDMS microchannel are fabricated and given in Figure 6.



(a) (b)

Figure 6: a) Fabricated LoC b) mounted for experimental study under an optical microscope.

A signal generator, “81150A Pulse Function Arbitrary Noise Generator, Agilent Technologies, (USA)” is connected to LoC and 5Vsin(wt) and 5Vsin(wt+π) applied to connection wires using alligator clips shown in Figure 6(b). Fluid flow is provided by two “Harvard Pump 11 Plus Single Syringe Pump” for each inlet and set to 2µl/min which is the lowest limit we could get. 3.2µm particles are collected from the outlet without any displacement along y direction but 9.8µm particles are deflected along the electrode geometry. Experimental results for 9.8 particles are given in Figure 7. The direction of fluid flow is from right to left and the white dashed line represents the channel walls. In Figure 7 from (a) to (d) the movement of the same particle can be seen step by step.

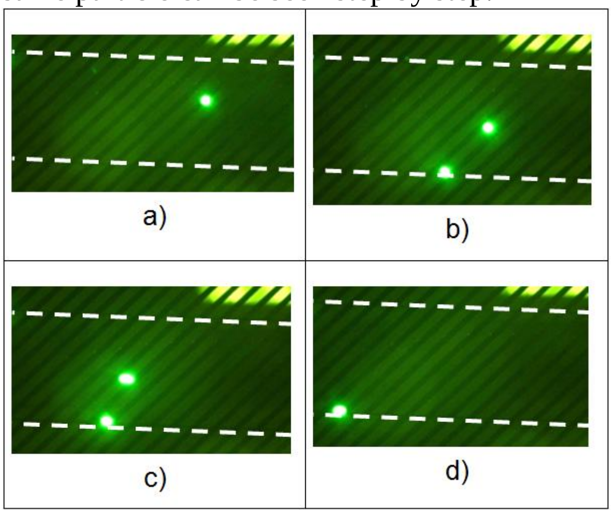


Figure 7: The displacement of a 9.8µm polystyrene particle along y direction step by step. The direction of fluid flow is from right to left.

4. Conclusion

Our models and simulation results show that efficient separation of particles based on their sizes is achievable with our designs. Our designs are applicable for sorting 9.8 µm polystyrene particles and can be easily optimized by levelling the electrode geometries or microchannel height for other type of particles or cells such as CTCs which have a diameter range between 10 to 20µm. Further steps of our study will proceed with decreasing the fluid flow as in our COMSOL simulations and experimenting with living cells such as Jurkat and Daudi cells. We also plan to fabricate other geometries such as G_2 to create stronger electric field gradients which results with more efficient separations.

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