Design, simulation and experiment of electroosmotic microfluidic chip for cell sorting

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Abstract

A microfluidic cell sorting chip has been developed using micromachining technology, where electroosmotic flow (EOF) is exploited to drive and switch cells. For this electroosmotically driven system, it is found that the effect of induced hydrostatic pressure caused by unequal levels in solution reservoirs is not negligible. In this work, the numerical simulation of EOF and opposing pressure induced flow in microchannels is presented and the velocity profiles in the microchannels are measured experimentally using microparticle imaging velocimetry (PIV) system. The result shows that the final resulting velocity is the superposition of the two flows. A total volume of 0.305 l is transported in the 50 μm microchannel and the back flow occurs after 240 s transportation. The task of sorting cells is realized at the switching structure by adjusting the electric fields in the microchannels. The performance of the cell sorting chip is optimized by investigating the effect of different switching structures. A Y-junction switching structure with 90° switching angle is highly recommended with simulated leakage distance of 53 μm and switching time of 8 ms.

Keywords: Microfluidics; Cell sorting; Electroosmotic flow

1. Introduction

Fluorescent activated cell sorting (FACS) is a crucial instrument for measurement of the fluorescence and light scattering of microscopic particles and biological cells and separating them based on their size and fluorescence color. Since 1990s, a number of micromachining methods [1–3] have been developed to replace the conventional flow chambers in FACS with microfluidic devices that allow more sensitive optical detection, easier mechanical set-up, and innovative sorting schemes. Several miniaturized cell sorting devices have been reported using different microfabrication materials such as metal [4,5], silicon [6–8], glass [9,10] and plastic materials [11–14]. Focusing of a sample stream, counting and sorting of cells on chip have been demonstrated. The latest results of a miniaturized flow cytometer, which can perform the functions of detection, enumeration and sorting of fluorescent species have been demonstrated [14]. Additionally, a microfluidic device for on-chip reaction, high-sensitivity detection and sorting of fluorescent cells and particles has been reported [15]. A microfabricated high-throughput fluorescent-activated cell sorter has also been presented [16]. Sample dispensing, interrogation, sorting and recovery have been demonstrated.

Despite of the rapid development of microfabricated cell sorters, several problems such as difficulties in microfluidic control and slow switching are still observable in the existing FACs devices. Most often, pressure is used in these devices to induce controllable movement of the fluid in a microchannel. As a result, pressure pumps and valves are necessary to regulate the flow properties. However, the small scale of the microchannels associated with microfluidic chips restricts the task of incorporating a mechanical pump or valve into a single chip. The various interfaces of microfluidic channels, tubing systems, and valves also improve the potential difficulties in the design of the microchip. As a result, electroosmotic pump using electric field gains more popularity because it is more suitable for miniaturization as an electrical field is relatively easy to be generated and structured in a microscale. However, in electroosmotic pumping, hydrostatic pressure-induced flow, caused by unequal levels in solution reservoirs, has an adverse effect.
on electroosmotic transportation. The counter pressure-induced flow, which is largely related to the cross-section of microchannel has yet been studied in details before. In this paper, one of the objectives is to investigate the transport capability of electroosmotic pumping in microchannels with various cross-sections.

Slow switching is another challenging issue for microfluidic cell sorter. A few examples of cell sorting on microfabricated devices have emerged where electroosmotic, hydrodynamic, and dielectrophoretic (DEP) forces have been exploited. Among them, electroosmotic switching as a valveless switching process is the most popular method due to its easy implementation and fast response. The fastest electroosmotic switching is reported by Dittrich et al. where the sorting time of the system is 33 ms per switch and the throughput of the device is about 30 cells per second [15]. However, this is still considered relatively low. Therefore, the second objective in this paper is to optimize the design of switching structure to improve the switching speed and accuracy. Leakage distance and switching time are simulated for various switching structures and experimental results are compared with the simulation results.

2. Microfluidic cell sorting chip

The schematic of the microfluidic cell sorting chip is shown in Fig. 1. The chip consists of two parts, a microfluidic system for cell manipulation and an optical system for cell detection. The microfluidic system includes two other units, a focusing structure and a switching structure. The optical system consists of laser diode, microfocusing lens and photodetector.

The working principles of the cell sorting chip are as follows. Samples containing different types of blood cells are prepared off chip where targeted cells are labeled with fluorescent marker. The mixture of labeled and unlabeled cells is injected into the input reservoir by micropipette. These cells are transported in microchannel by electroosmotic flow (EOF), which is controlled by platinum electrodes at the input and output reservoirs. The chip comprises a branched channel system for focusing of the sample solution and cells are focused into a narrow stream using neighboring sheath fluids such that the cells can line up and move forward one by one. The sheath fluid flows are pumped by electric field to flow at the required velocity so that the width of the sample flow can be well controlled. The pre-focused sample moves down to the detection region where fluorescence is excited by the focused laser beam. Laser diode is integrated onto the chip as the light source, which makes the optical detection system more compact. The light is focused by a micro-lens array and fluorescent emission is collected by a photo-detector. The targeted cells are identified and counted by their distinct fluorescent responses. In the subsequent switching region, the cells can be sorted into two different output channels at the second intersection. The electroosmotic switching is exploited to direct the cells into the appropriate outlet channels. Sample solution is defaulted to flow into the waste reservoir. When the fluorescence of the cell passes a preset threshold, active sorting occurs, which is realized by adjusting the direction of the electroosmotic flow through electrically switching the voltages at the output reservoirs. Since the electroosmotic flow responds to the direction of the applied field instantaneously, the electrical manipulation can shorten the switching times significantly.

2.1. Simulation of opposing electroosmotic and pressure-induced flows

In this study, EOF is adopted as the mechanism to transport sample solution through the microchannels, because it is a convenient tool for tuning liquid flow into individual channels, and it responds instantaneously to the direction of the applied field [17].

EOF is the motion of bulk liquid caused by the application of an electric field to a channel with a charged wall [18,19]. If the channel surface is negatively charged (such as in the case of deionized water), the positive ions are attracted towards the surface, and the negative ions are repelled from the surface, keeping the bulk of the electrically neutral liquid far from the wall. The distance from the wall, where the electroosmotic potential energy is equal to the thermal energy is known as the Debye length. The ion redistribution within the Debye length results in a net electric charge, which is characterized by the electroosmotic potential. This region is defined as the electric double layer (EDL) that consists of two distinct zones, Stern layer and diffuse layer [20]. The Stern layer is formed owing to strong attraction of ions onto the oppositely charged wall surface. The ions are immobilized near the charged surfaces and the electroosmotic potential drops rapidly across the Stern layer to a value at the edge of the Stern layer, which is known as the zeta potential \( \zeta \).

When the EDL thickness is significantly smaller than the channel dimensions, a plug-like velocity is observed in the bulk of the channel, which is called the Helmholtz–Smoluchowski electroosmotic velocity and is defined as:

\[
\nu_{\text{eof}} = \frac{-\zeta \varepsilon E_x}{\mu}
\]

(1)

where \( E_x \) is the applied electric field in the stream-wise direction, \( \varepsilon \) the electric permittivity and \( \mu \) is the absolute fluid viscosity.

For the ideal case, EOF velocity has a flat profile. However in practice, several factors induce velocity gradients in the bulk liquid [20]. One particular problem is the coexistence of unwanted
pressure-induced flow with EOF as shown in Fig. 2. A typical structure of a microfluidic system is shown in Fig. 2(a) where two reservoirs are connected by a microchannel. Initially, the liquid in two reservoirs are at the same level as shown by the dashed line, and only EOF exists inside the microchannel. As fluid continuously flows from the inlet to the outlet, the liquid level drops in the inlet reservoir while rises in the outlet reservoir, resulting in an appreciable height difference. This height difference in turn introduces a pressure driven flow, which perturbs the EOF. The linear flow rate due to height differences in a microchannel, $v_p$, is given as:

$$v_p = -\frac{\Delta h \rho g D_h^2}{2 f \text{Re} \mu L}$$  \hspace{1cm} (2)

where $\Delta h$ is the difference in liquid levels, $\rho$ the density of the solution, $g$ the acceleration due to gravity, $f \text{Re}$ the fanning friction factor multiplied by the Reynold’s number and depends on channel geometry, $\mu$ the solution viscosity, $L$ the channel length and $D_h$ is the hydraulic diameter of the channel. Wherever electroosmosis and pressure-induced counter flow co-exist in microchannels, the average flow rate $v_\text{ave}$ consists of the sum of the electrokinetic linear velocity $v_\text{eof}$, and the average pressure-induced linear velocity $v_p$, that can be expressed as:

$$v_\text{ave} = \frac{-\zeta E_x}{\mu} - \frac{\Delta h \rho g D_h^2}{2 f \text{Re} \mu L}$$  \hspace{1cm} (3)

Eq. (3) predicts that the resulting velocity is the superposition of plug profile of the EOF and parabolic profile of Poiseuille flow as shown in Fig. 2(b).

When the average velocity equals to 0, i.e., when the magnitude of EOF is identical to the pressure-induced flow, back flow will start. The height difference at this point is called threshold height difference that can be derived as:

$$\Delta h = -\frac{2 f \text{Re} L \zeta E_x}{\rho g D_h^2}$$  \hspace{1cm} (4)

The transport capability of an electroosmotically driven microchannel can be given as:

$$Q = \frac{\pi D^2 \Delta h}{8}$$  \hspace{1cm} (5)

where $Q$ is the threshold transported volume, $D$ the diameter of the reservoir and $\Delta h$ is the threshold height difference between the inlet and outlet reservoirs.
Based on Eq. (3), the average flow rate is related to hydraulic diameter of the channel. Therefore, several straight microchannels with different cross-sections are designed and simulated. For convenience, channel width is fixed at 100 μm. Channel depth is allowed to vary from 20 to 100 μm and all the channel lengths are fixed at 5 mm. The two reservoirs with diameter of 4 mm and height of 2 mm are connected by the microchannel as shown in Fig. 2(a). Initially, the buffer in the two reservoirs is set at an equal level. The electrodes are inserted into the reservoirs. An electric field of 300 V/cm is applied.

The microfluidic structure is operated in laminar flow regime. The working buffer is H₂O and its density equals to 1000 kg/m³, the viscosity is 10⁻⁹ m²/s and the relative permittivity is 78.5. The zeta potential is obtained from experiment and the value is ~0.0295 V. Simulations are performed on a multiphysics software package CFD-ACE+ (ESI Research Corporation) based on the finite volume method.

As the theory predicts that the plug profile of EOF and the parabolic profile of Poiseuille flow are linearly independent, the simulation of the straight channel is divided into three models. The first model is a pure EOF simulation without pressure disturbance, the second model is a pure pressure driven flow, and the last model combines EOF and pressure-driven flow to determine the threshold height difference between the two reservoirs at which backward flow will occur. On the y–z plane of the channel as shown in Fig. 2(a), a line along the z direction at the middle width is defined as the chordlength. Velocities along the chordlength are studied.

For the simulation of the EOF model, the CFD software shows that the EOF velocity is independent of the channel size, and there is no corresponding increase in the average velocity with increased channel cross-section. For all channel sizes, the EOF velocities equal to 695 μm/s; then through the intersection of this horizontal line and the plotted curve, a vertical line is drawn to intersect with the pressure axis. Thus, the pressure value is read out and this is the threshold pressure. For example, when the channel depth is 40 μm, the horizontal line A–B intersects with the plotted curve at point O, and through point O, a vertical line C–D is drawn, intersecting with the pressure axis at 14.7 Pa, therefore the threshold pressure for 100 × 40 cross-sectional channel is 14.7 Pa. The threshold pressure can be transformed to critical height difference according to \( \Delta P = \rho g \Delta h \), where \( \Delta h \) is the height difference between the buffer reservoirs. Total volume transported \( Q \), is calculated by Eq. (5). The threshold pressure, height difference and total volume transported for various cross-sections are summarized in Table 1.

As noted from Table 1, the threshold height difference decreases with channel size. This implies that it is easier for the backflow to occur in a big channel. Based on the value of \( Q_{\text{threshold}} \), to transport a total volume of 3 μl sample from one reservoir to another, \( D_b \) of the microchannel should not be larger than 100 μm.

### Table 1

<table>
<thead>
<tr>
<th>Depth (μm)</th>
<th>20</th>
<th>40</th>
<th>60</th>
<th>80</th>
<th>100</th>
</tr>
</thead>
<tbody>
<tr>
<td>( P_{\text{threshold}} ) (Pa)</td>
<td>61.5</td>
<td>14.7</td>
<td>8.8</td>
<td>6.2</td>
<td>4.8</td>
</tr>
<tr>
<td>( \Delta h_{\text{threshold}} ) (mm)</td>
<td>6.15</td>
<td>1.47</td>
<td>0.88</td>
<td>0.62</td>
<td>0.48</td>
</tr>
<tr>
<td>( Q_{\text{threshold}} ) (μl)</td>
<td>38.6</td>
<td>9.2</td>
<td>5.5</td>
<td>3.9</td>
<td>3</td>
</tr>
</tbody>
</table>

2.2. Design and simulation of switching structures

For cell sorting application, switching structure is designed in order to sort two different types of cells. During the operation of the cell sorting chip, after focusing and optical detection, the sample is then guided into the appropriate outlet port through switching structure. The performance of the switching structure can be evaluated based on two important parameters, switching speed and switching accuracy. The switching speed refers to how fast the fluid is switched from one direction to another. The switching accuracy means how many percentage of the targeted cells can be successfully manipulated into the predetermined reservoir. Theoretically, when a cell is supposed to move into the collection reservoir, the fluid is switched towards that reservoir and electroosmotic flow towards the waste reservoir is totally blocked. However, in practice, switching error due to the leakage of fluid cannot be avoided. In this paper, three-arm “Y-junction” switching structure intersecting at a junction is designed as shown in Fig. 4. To achieve satisfactory switch-

![Fig. 3. Simulation results of threshold pressures for various cross-sections.](image-url)
The leakage distance is determined for Y-junction switching structures with different angles using steady state simulation. During the simulation, electric potential at the input reservoir “I” is fixed at 100 V, the collection reservoir “C” is set to 0 V and the waste reservoir “W” is set to float state. As a result, sample solution is defaulted to flow to collection reservoir “C”. Fig. 5 illustrates the definition of leakage distance. In Fig. 5(a), the sample solution is directed into collection channel “C”, and fluid inside waste channel “W” is supposed to be blocked totally. The dark grey represents zero velocity. However, due to the inertia of fluid, there is some velocity in channel “W” near the junction, which is represented by the light grey color. The velocity along the centerline line $m-n$ is plotted as shown in Fig. 5(b) to determine the leakage distance. The critical velocity is chosen at 10% of the maximum velocity, which equals to 220 $\mu$m/s in this case, and the distance between $m$ and the position at which velocity reaches 22 $\mu$m/s is defined as the leakage distance.

By performing the leakage analysis at different angles, the relationship between the leakage distance and the switching angle $\theta$ is established and shown in Fig. 6. When the angle is small, the inertia from the inlet channel to the undesired channel is large, resulting in a large leakage distance. Therefore, the leakage distance decreases with the increase of angle, $\theta$.

In order to study the effect of the switching angle $\theta$ on the switching speed, a series of transient simulations are carried out at different angles. The initial condition is shown in Fig. 7(a). The electric potential at the input reservoir “I” is fixed at 100 V, the voltage at the collection reservoir “C” is set to 0 V and the waste reservoir “W” is set to float state. As EOF always flows from anode to cathode, sample solution is defaulted to flow to collection reservoir “C”. By switching float state to reservoir “C” and ground state to reservoir “W”, the fluid is switched from the direction I $\rightarrow$ C to I $\rightarrow$ W as shown in Fig. 7(b). For the switching time measurement, the velocity at the intersecting point O as a function of time is plotted as shown in Fig. 7(c). It is sensible to see an increase in the velocity magnitude, as during the switching period, fluid in the waste channel transforms from static state to dynamic. The switching time is defined as the time when the velocity at point O reaches the final steady velocity. The smaller the switching time, the faster the switching speed.
Switching time analysis is performed at different angles and the relationship between the switching time and the switching angle $\theta$ is shown in Fig. 8. The graph reveals that the switching time becomes longer as the switching angle increases. This is because as the switching angle increases, the change in the velocity vector from one output channel to another is larger, resulting in longer switching time.

From Figs. 6 and 8, it is noted that there is a trade off between the leakage effect and the switching speed. If the switching angle is small, such as 60°, the switching speed is faster but larger leakage phenomenon is observed, inducing lower accuracy in cell sorting. If the switching angle is large, such as 160°, the leakage phenomenon is reduced, but the switching speed is slower. For the application of blood cell sorting, both switching accuracy and switching speed are important. Thus, an angle should be neither too large nor too small. In this case, the switching angle $\theta$ is chosen to be 90°.

3. Experimental results and discussions

3.1. Microfabrication and experiment set-up

The microfluidic device is fabricated in polydimethyl siloxane (PDMS) using soft lithography technique. The mask layout of microfluidic channels (diameters >20 μm) is designed using a CAD program (Freehand 8.0). The master for molding is fabricated in SU-8 50 (MicroChem) by standard photolithography. Once the SU-8 master is fabricated, channels in PDMS are formed by replica molding. A 10:1 mixture of PDMS prepolymer and curing agent (Sylgard 184) is stirred thoroughly.
and then the prepolymer mixture is poured onto the master and cured for 1 h at 65 °C. After curing, the PDMS replica is peeled from the master. Reservoirs are created by punching PDMS with a 4 mm diameter puncher. The two PDMS substrates are sealed by plasma bonding. Within 30 s after the removal from the plasma chamber, the substrates are brought into conformal contact where an irreversible seal formed spontaneously. The exposure to plasma oxidation renders the surface hydrophilic because of the silanol groups. Aqueous solutions can easily make these oxidized channels wet. In order to protect the surface, the microchannels are filled with deionized (DI) water 3 min after bonding.

The velocity profiles in microchannels are measured using a microscopic particle image velocimetry (µ-PIV) system (Dantec). It enables spatially resolved measurements of instantaneous flow velocity field within a very short time and allows the detection of small scale spatial structures in the flow velocity field. In this study, the liquid flow in the microchannels is monitored from the fluorescence of the seeded dye molecules. One-micrometer fluorescent micro-spheres (Molecular Probes, Inc.) with an excitation peak at 490 nm and an emission peak at 515 nm are used as seedings. An illumination system – microstrobe (Dantec Pte Ltd.) – is used to illuminate the seeding particle. Blue light beam passed through a beam expander to enlarge the beam diameter, and then illuminated the test section of the microchannel. The fluorescence signal is recorded with an interline transfer CCD monochrome camera (Hisense MKII, Strobe double frame).

3.2. Velocity profiles in straight microchannels

Experiments are carried out to study the velocity profiles and transport capabilities of straight microchannels with different dimensions. An external 300 V/cm electric field is applied axially along the microchannels. The straight microchannels are 5 mm long with two 4 mm-diameter reservoirs at each end. The height of the reservoir is the same as the thickness of the PDMS layer, which is 2.5 mm. The buffer surfaces are at the two equal elevations initially. The depths of the microchannels are 26 and 50 μm, respectively, while the widths of both channels are 100 μm. The electrodes are placed at the two reservoirs and H2O is used as buffer liquid.

The measured velocity profile with time for a 100 μm-wide, 50 μm-deep microchannel at an applied electric field of 300 V/cm is shown in Fig. 9. The profiles were taken at \( t = 1 \) s, \( t = 20 \) s and \( t = 40 \) s, respectively. Electroosmotic flow has a plug-like velocity profile initially. Due to frictional drag, the electroosmotic flow at the wall is slightly slower than the flow through the rest of the microchannel. In the fully developed electroosmotic flow, average velocity has a peak speed of \( \sim 695 \mu m/s \).

The velocity profile after 20 s represented superposition of the plug-like electroosmotic profile and the Poiseuille profile due to the adverse pressure gradient, which is a result from the downstream reservoir height rise with flow. By inducing adverse pressure gradients into the microchannel, electroosmotic flow cannot maintain the plug-like profile and the center velocity is reduced to 559 μm/s. The velocity at \( t = 40 \) s is much slower compared to the fully developed electroosmotic flow and center velocity of 418 μm/s is observed.

The instantaneous flow rates at different instances are plotted for the straight microchannels with widths of 26 and 50 μm as shown in Fig. 10. It shows that the 50-μm channel has higher initial flow rate. This is due to its bigger cross-sectional area. However, the instantaneous flow rate of the 50-μm channel decreases faster than that of the 26-μm channel because the backward pressure has larger effect in the microchannels with bigger hydraulic diameters. Therefore, the fluid transport in 26-μm channel is slower but last longer.

Fig. 11 plots the total transported sample volumes calculated by integrating instantaneous flow rate with time. It can be seen that the 50 μm microchannel managed to transport total volume of 0.305 μl and the back flow occurs after 240 s transportation. For 26 μm microchannel, it transports the volume of 0.33 μl within 425 s, which is much slower compared to the 50 μm microchannel, but back flow occurs much later. This observation confirms that the occurrence of back flow is due to the adverse pressure gradients and this back flow is larger for microchannels with bigger cross-sections.

The experiment proves the linear dependence of electroosmotic velocity on an imposed hydrostatic counter pressure as discussed above, however, a big discrepancy is observed between the simulated and experimental results of total volume transported. The maximum volume of fluid in each reservoir is 12 μl.
However, at the end of the experiment, the inlet reservoir was empty even though the highest total volume transported was only 0.33 L. Evaporation and electrolysis are the main reasons for the fluid loss. Evaporation is the change of liquid to vaporize at temperatures below the liquid’s boiling point as the molecules near the liquid surface escape from the liquid into the atmosphere. Evaporation is faster at higher temperature, lower pressure and with forced convection. Electrolysis is the electrochemical process in which an electrolyte is decomposed by a current passing through it. For this EOF experiment, the current decomposed water to hydrogen gas at the cathode and oxygen gas at the anode. These two elements may lead to 90% loss of the fluid. Evaporation can be decreased by putting cover lids on reservoirs and electrolysis can be reduced by applying smaller electric field or using lower conductivity buffer.

3.3. Switching structures

Two different designs of Y-junction switching structures with switching angles of 90° and 180° are fabricated in PDMS. All the dimensions remain the same as those shown in Fig. 4. The potential at the inlet Reservoir “I” is held constant at 100 V and the voltages applied at the collection and the waste reservoirs are switched between 0 V and floating state to achieve proper switching. H2O is used as the buffer. The flow velocities and the leakage distances are measured by micro-PIV system.

The flow velocities of the Y-junction structures with switching angle θ = 90° and 180° are measured as shown in Fig. 12. For θ = 90°, the flow velocities at the inlet, the collection and the waste channels are 257, 229 and 21.8 μm/s, respectively. For θ = 180°, the velocities at the inlet, the collection and the waste channels are 265.1, 254.7 and 9.6 μm/s, respectively. The velocity in waste channel for θ = 90° is bigger than that for θ = 180°, implying larger leakage distance for θ = 90°. Table 2 summarizes the simulation and experimental results for the leakage distance.

Based on the simulation and experimental results of the switching accuracy and the switching speed, it can be concluded that when the switching angle increases, the switching accuracy is higher while the switching time is longer. The design of the switching structure largely depends on the application. When the accuracy is considered as the first priority, large switching angle is desired. If throughput is more important, smaller switching angle is preferred. For the objective of blood cell sorting in this project, both accuracy and throughput are essential parameters. Therefore, the Y-junction switching structure with 90° switching angle is recommended.

Table 2: Comparison of simulation and experimental results of leakage distances

<table>
<thead>
<tr>
<th>Y-junction switching structure</th>
<th>Leakage distance (μm) (simulation)</th>
<th>Leakage distance (μm) (experiment)</th>
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</thead>
<tbody>
<tr>
<td>θ = 90°</td>
<td>53</td>
<td>55.2</td>
</tr>
<tr>
<td>θ = 180°</td>
<td>25</td>
<td>25.4</td>
</tr>
</tbody>
</table>

Fig. 12. Experimental results of flow velocities for Y-junction structure: (a) θ = 90° and (b) θ = 180°.
4. Conclusions

The induced hydrostatic pressure and switching effect in the electroosmotically driven microfluidic cell sorting system have been extensively studied in this paper. The effect of the induced hydrostatic pressure is investigated for straight microchannels with different cross-sections. The result shows that the resulting velocity in microchannels can be viewed as the superposition of electroosmotic velocity and opposing induced pressure-induced velocity. For the 50 μm microchannel, the total volume of 0.305 μL is transported and the back flow occurs after 240 s transportation; for 26 μm microchannel, the total volume of 0.33 μL is transported within 425 s, which is significantly slower compared to the 50 μm microchannel, but back flow occurs later.

For the switching structure design, Y-junction switching structures with different angles are simulated and tested. The effects of the switching angles on the switching accuracy and speed are investigated. It is found that for a larger switching angle, the switching accuracy is higher while the switching time is longer. For the objective of blood cell sorting in this work, the Y-junction switching structure with 90° switching angle is highly recommended. The leakage distance is 53 μm and the switching time is 8 ms.

References