Continuous particle focusing in a waved microchannel using negative dc dielectrophoresis

This article has been downloaded from IOPscience. Please scroll down to see the full text article.
2012 J. Micromech. Microeng. 22 095001
(http://iopscience.iop.org/0960-1317/22/9/095001)

View the table of contents for this issue, or go to the journal homepage for more

Download details:
IP Address: 202.40.139.167
The article was downloaded on 02/08/2012 at 09:47

Please note that terms and conditions apply.
Continuous particle focusing in a waved microchannel using negative dc dielectrophoresis

Ming Li¹, Shunbo Li², Wenbin Cao³, Weihua Li¹, Weijia Wen²,³ and Gursel Alici¹

¹ School of Mechanical, Materials and Mechatronic Engineering, University of Wollongong, Wollongong, NSW, 2522, Australia
² Department of Physics, The Hong Kong University of Science and Technology, Clear Water Bay, Kowloon, Hong Kong
³ Nano Science and Technology Program and KAUST-HKUST Micro/Nanofluidic Joint Laboratory, The Hong Kong University of Science and Technology, Clear Water Bay, Kowloon, Hong Kong

E-mail: weihuali@uow.edu.au and phwen@ust.hk

Received 5 May 2012, in final form 1 June 2012
Published 26 July 2012
Online at stacks.iop.org/JMM/22/095001

Abstract
We present a waved microchannel for continuous focusing of microparticles and cells using negative direct current (dc) dielectrophoresis. The waved channel is composed of consecutive s-shaped curved channels in series to generate an electric field gradient required for the dielectrophoretic effect. When particles move electrokinetically through the channel, the experienced negative dielectrophoretic forces alternate directions within two adjacent semicircular microchannels, leading to a focused continuous-flow stream along the channel centerline. Both the experimentally observed and numerically simulated results of the focusing performance are reported, which coincide acceptably in proportion to the specified dimensions (i.e. inlet and outlet of the waved channel). How the applied electric field, particle size and medium concentration affect the performance was studied by focusing polystyrene microparticles of varying sizes. As an application in the field of biology, the focusing of yeast cells in the waved microchannel was tested. This waved microchannel shows a great potential for microflow cytometry applications and is expected to be widely used before different processing steps in lab-on-a-chip devices with integrated functions.

(Some figures may appear in colour only in the online journal)

1. Introduction

Recently, lab-on-a-chip (LOC) devices have been increasingly used for clinical and biological management and analysis in the fields of medicine, environment, biochemistry and biotechnology [1–3]. One of the applications involves focusing bioparticles into a narrow stream, which is usually necessary for subsequent analytical and processing steps (i.e. counting, detecting and sorting, etc). Various techniques have been developed to focus particles in the aqueous solution, such as hydrodynamic [4–7], inertial [8], hydrophoretic [9], acoustic [10], magnetic [11], optical [12], electrophoretic (EK) [13–16], electrophoretic [17] and dielectrophoretic [18–30] methodologies. Among those focusing methods that have been applied in microfluidic devices, dielectrophoresis (DEP) may be the most popular one due to its great advantages, including label-free nature, compatibility with LOC devices, ability to manipulate neutral bioparticles, easy and direct interface with electronics and the analysis of high selectivity and sensitivity [31].

Originally described by Pohl [32], DEP is the resultant motion of particles in a non-uniform electrical field due to the polarization effect induced on the particles, which has been employed in microfluidic systems for wide applications [33–35]. There are two main strategies to generate the inhomogeneous electric fields required for DEP effect:
microelectrodes and insulating topographical structures. Compared to traditional electrode-based DEP devices, insulator-based DEP devices have following advantages [36]: (1) as there are no electrodes embedded in the channel, the device with insulating structures is robust and chemically inert, accompanied by reduced surface fouling and no gas evolution due to electrochemical reactions; (2) without the metal deposition process, device fabrication techniques such as injection molding and hot embossing are relatively simple and applicable for mass production; (3) particle movement taking advantage of the combined electrophoretic and electroosmotic effects eliminates the requirement of additional hydrodynamic flow.

Researchers have generated spatial non-uniformities using in-channel insulating obstacles, including posts [23], one pair of oil droplets [24] and a single constriction in the channel [25, 26], to achieve the focusing of microparticles. However, these designs suffer from locally amplified electric field, significant shear stress, trans-membrane voltage and Joule heating, which cause adverse effects on biological samples. Furthermore, the probability of device fouling due to particle clogging or adhesion at the manufactured obstacles in channels is relatively high.

These drawbacks are relieved or even overcome by curving the microfluidic channel to create non-uniformities of the electric field. Xuan and his co-workers [27, 29, 35] introduced a serpentine microchannel with 90° turns to focus particles into a stream along the channel centerline; however, the area where electric gradient can be generated is limited, that is, only within the corner of channel turns, while the electric field is uniformly distributed in the straight section of the microchannel. Moreover, a spiral microchannel has been demonstrated by Zhu et al [30] for focusing polystyrene particles. However, it is necessary that either the applied dc electric field is sufficiently large or the curved channel is sufficiently long, especially for effectively focusing smaller particles. Additionally, this design is more sensitive to contaminations or particle adhesion on the channel wall, as particles are deflected and focused to near the outer sidewall of the microchannel.

In this paper, we proposed a waved microchannel for the focusing of microparticles and cells using negative dc DEP in a continuous flow. This design achieves continuous pumping and focusing by taking advantage of EK and dielectrophoretic effects, which are responsible for particle streamwise transport and particle cross-stream deflection, respectively. The overall result is the movement of particles in a confined and concentrated stream along the channel centerline at the outlet of the curved section. Compared to the serpentine channel with 90° turns, this design extends the region creating the DEP effect to the full width of the waved microchannel, allowing a greater control over particle motion. More importantly, the stagnation regions and locally amplified electric field due to sharp turns are eliminated, which contribute to a better protection against particle adhesion to sidewalls and cell damage, respectively. This design also shows advantages over spiral channels, as it achieves the focusing of particles in the channel’s center region, and the magnitude of applied dc fields and the total length of the curved channel required for desirable deflection are reduced. In addition, the proposed waved microchannel is more flexible in the structure design, in other words, the large/small radius of the curvature, width and length of the microchannel could be optimized depending on the required focusing performance, even the entire microchannel could be wound up to reduce the device footprint. The effects of applied dc electric field, particle size and medium concentration on the focusing performance are examined. Moreover, the functionality of the device was demonstrated by focusing yeast cells continuously and effectively. Numerical simulations were also performed to predict and verify the focusing behavior of both microparticles and cells, the results of which indicate a reasonable agreement with the experimentally obtained particle trajectories.

2. Theory and mechanism

In dc electric fields, suspended particles experience the EK effect, which is a combination of fluid electroosmosis (EO) and particle electrophoresis (EP). The resulting motion is determined by the EK velocity written as [37]

$$u_{\text{EK}} = \mu_{\text{EK}} E = u_{\text{EO}} + u_{\text{DEP}} = (\mu_{\text{EO}} - \mu_{\text{DEP}})E,$$  \hspace{1cm} (1)

where $\mu_{\text{EK}}$ is EK mobility, $\mu_{\text{EO}} = -e_m \zeta_w / \eta$ and $\mu_{\text{DEP}} = -e_p \zeta_p / \eta$ are electroosmotic and electrophoretic mobility, respectively, $e_m$ is the permittivity of the suspending medium, while $\eta$ is the dynamic viscosity of the suspending medium. $\zeta_w$ and $\zeta_p$ represent the zeta potentials of the channel wall and particle, respectively. $E$ is the electric field vector. Equation (1) shows that the EK velocity is linearly proportional to the local electric field and directionally flows the field lines.

If the electric field is non-uniform, the motion of the suspended particle is also affected by dielectrophoretic force. The time-average of this force, $F_{\text{DEP}}$, on an insulating spherical particle in dc electric fields, and the induced dielectrophoretic velocity, $u_{\text{DEP}}$, respectively, are given by [38]

$$F_{\text{DEP}} = \langle 1/2 \rangle \pi e_p d^3 \zeta_{\text{CM}} (E \cdot \nabla E)$$ \hspace{1cm} (2)

and

$$u_{\text{DEP}} = \mu_{\text{DEP}}(E \cdot \nabla E) = (e_m d^2 \zeta_{\text{CM}} / 6 \eta) \cdot (E \cdot \nabla E),$$ \hspace{1cm} (3)

where $d$ is the particle diameter, $f_{\text{CM}} = (\sigma_p - \sigma_m) / (\sigma_p + 2\sigma_m)$ is known as the Clausius–Mossotti (CM) factor, $\sigma_p$ and $\sigma_m$ are the electric conductivities of particle and the suspending medium, respectively. If the particle is less conductive than the suspending medium ($\sigma_p < \sigma_m$), then the CM factor will be negative ($f_{\text{CM}} < 0$), resulting in a negative DEP force, which repels the particle away from the region of the strong electric field.

Unlike traditional electrode-based DEP devices, in which the non-uniform electric field required for DEP effect is created by applying the ac electric field via the metallic microelectrodes embedded inside the microfluidic channel, insulator-based DEP devices create the spatial non-uniformities of the electric field by insulating topographic structures (i.e. posts, hurdles, constriction in channels and curved channels), and dc or ac electric field is applied via remote electrodes outside the microchannel. In the case of
exposed to negative DEP forces. Microchannel are shown (darker region has a stronger electric field). The electric field in the channel's center region. A schematic illustration of the focusing mechanism is shown in figure 1. The electric field lines and contours of the electric-field strength $E$ within the microchannel are shown (darker region has a stronger electric field). Particles moving through the waved channel electrokinetically are exposed to negative DEP forces.

In the proposed s-shaped curved channels, the particle experiencing negative DEP force will be deflected toward the region with the lowest local electric field, namely the outer wall of each semicircular microchannel. The outer wall and inner wall alternate between the counterclockwise and clockwise semicircular microchannels, the negative dielectrophoretic force exerting on the particle (always from the inner wall to the channel centerline) also alternates as illustrated in figure 1. Moreover, the repulsive negative DEP force is stronger if the particle is closer to the inner wall where the electric field gradient is larger. At appropriate and moderate electric fields, particles that electrokinetically move through the waved microchannel will be gradually deflected toward the channel's center region under the influence of such negative DEP force, leading to a focused particle stream along the channel's centerline in the exit region. However, if the applied electric field is significantly high, the induced negative DEP forces will be so strong that deflect particles past the centerline to the outer wall of each semicircular microchannel. Therefore, the particles will be forced to bounce between sidewalls and finally move out of the waved section close to one sidewall. This phenomenon has also been demonstrated by Church et al. [39] for size-based particle separation in a serpentine microchannel.

Three features are combined in such design of a waved microchannel to control the particle motion using negative DEP forces: (1) curved channel (or semicircular channel) generates electric field gradient across the streamline when a dc electric field is applied; (2) s-shaped channel causes the induced negative DEP forces alternating directions between two adjacent semicircular channels; (3) waved channel accumulates the focusing effect of negative dielectrophoretic forces in each s-shaped section.

3. Experimental details

3.1. Device fabrication

The microchannels patterned in polydimethylsiloxane (PDMS) were fabricated using standard photo- and soft-lithography techniques. The microdevice layout was designed and drawn using AutoCAD (Autodesk Inc., USA), and printed onto a transparent plastic thin film with dark field at a resolution of 20,000 dpi serving as the mask. Photoresist (SU-8 2025, MicroChem Corp., Newton, MA) was spun on a clean 4″ silicon wafer by a two-step coating cycle (with the setting of 800 rpm for 15 s and 2000 rpm for 30 s), resulting in a nominally 30 μm thick film. After relaxing in air at room temperature for 2 h to level the photoresist, a two-step soft bake (65 °C for 10 min and 95 °C for 5 min) was performed on hotplates. Contact lithography was then conducted by 25 s exposure to UV light through the printed mask using a mask aligner system (ABM, San Jose, CA). Following another two-step hard bake (65 °C for 5 min and 95 °C for 6 min), the wafer was immersed and gently vibrated in SU-8 developer solution for 4 min and rinsed with isopropyl alcohol (IPA), leaving a positive replica (exposed photoresist) of the microdevice pattern on the wafer. After developing, the SU-8 master was treated by trichlorosilane (97%, Sigma-Aldrich, USA) in an encapsulated chamber under a pressure of about −0.5 bar to deposit a mono-layer of silane onto the surface. The purpose of this treatment was to reduce the adhesion of polydimethylsiloxane (PDMS) to the SU-8 master surface.

PDMS gel prepared by thoroughly mixing Sylgard 184 and its curing agent (Dow Corning, Midland, MI) in a weight ratio of 10:1 was poured over the master, and then put in a vacuum pump to get rid of bubbles. Following the degassing, it was put into an oven at 70 °C for 1.5 h to solidify the PDMS gel, and the cured negative PDMS cast containing the microdevice pattern was cut and removed from the master. Two holes serving as the inlet and outlet reservoirs were punched through the PDMS cast. Immediately after oxygen plasma treatment in a plasma cleaner (PDC-002, Harrick Plasma,
Ossining, NY) for 2 min, the channel side of the PDMS and a clean glass slide were bonded together and thus an enclosed microchannel was constructed. Once sealing, the channel was filled with working buffer by capillary action to maintain its surface properties.

Figure 2 depicts the structure and dimensions of the microfluidic device consisting of a waved microchannel at the center, inlet/outlet reservoirs at the end and two straight connecting microchannels. The device was filled with red dyed oil for a clear demonstration. The microchannel has a uniform width and depth of 100 and 30 μm, respectively. The 1 cm long waved section at the center is composed of 17 units of identical s-shaped curved channels in series, each of which has a small and large curvature of 100 and 200 μm, respectively, as shown in the inset. The total length of the microchannel (including the straight section) is 2 cm, and the diameter of the two reservoirs is 6 mm.

3.2. Microparticle and yeast cell preparation

Polystyrene particles with 10 and 15 μm in diameter (Fluosphere, Invotrogen, CA, USA) were employed in our experiments, the sizes of which are comparable to red and white blood cells. Both particles, which originally suspended in 0.15 M NaCl solution, were diluted by deionized (DI) water 150 times. 5 μm-diameter polystyrene microparticles (Singma-Aldrich, USA) that originally suspended in pure water were re-suspended in either 1 or 10 mM NaCl solution, and the final concentration was around 10⁶ beads per milliliter. Baker’s yeast cells (Saccharomyces cerevisiae) were cultured at 37 °C in the YEP broth (MP Biomedicals, LLC). After about 24 h, the cells were diluted with DI water and then re-suspended in the five times diluted phosphate buffer solution (PBS, pH 7.0, Radiometer Analytical A/S, Denmark) at a concentration of about 10⁶ cells per milliliter. The average size of the yeast cells was around 5 μm in diameter. The suspensions of particles and yeast cells were introduced into the inlet reservoir using a pipette, while the outlet reservoir was filled with the corresponding working solution.

3.3. Experimental technique

The electric field was generated by a dc power supply (SL10P300/200, Spellman High Voltage Electronics Corp., Hauppauge, NY), and used to drive the fluid flow via two platinum electrodes submerged in each reservoir. The motion of particle through the waved microchannel was monitored by an inverted microscope (Olympus IX71, Tokyo, Japan) and recorded by a CCD camera (DXC-390P, Sony, Japan). The camera was run in the video mode at 25 frames per second. The acquired digital images had a resolution of 576 × 720 pixels, with representing an approximated 1 mm long channel for 10 × objective. All the videos and images were post-processed by MATLAB R2011b (Mathworks Inc., Natick, MA), and the particle trajectory images were obtained by superimposing consecutive images converted from videos.

4. Modeling

In order to study and predict the effect of negative dc DEP on particles/cells transport through the waved microchannel, a numerical model first developed by Kang et al [39] was used, which has also been applied by other researchers to simulate the dielectrophoretic (i.e. dc, dc-biased ac) focusing in different channel structures [26, 27, 30]. A correction factor, c, was introduced to account for the perturbation of particle size, coupled particle–particle interaction, etc on the dielectrophoretic velocity, which has been found to decrease with the increase in particle size, and depends on channel geometry, but is insensitive to the applied electric field [27, 30, 40]. By using this correction factor, the particle velocity can be rewritten as

\[ u_p = u_{EK} + u_{EOF} = \mu_{EK}E + c\mu_{EOF}(E \cdot \nabla E). \]  \hspace{1cm} (4)

In the modeling, the eletrokinetic mobility, \( \mu_{EK} \), was determined by equation (1) with zeta potential values obtained from references [41–44]. The dynamic viscosity, \( \eta = 0.9 \times 10^{-3} \) kg (m s)⁻¹, and permittivity, \( \varepsilon_m = 6.9 \times 10^{-10} \) C (v m)⁻¹ for pure water at 25 °C were used to calculated the EK and dielectrophoretic mobility by equations (1) and (3). The CM factor, \( f_{CM} \), which depends on electrical conductivities of both the particle/cell and the suspending medium was set to be −0.5. The reason is that the electric conductivities of polystyrene particles and live yeast cells at dc electric field are much smaller than those of suspending media used in our experiments (i.e. 1 or 10 mM NaCl solution, and PBS). The velocity in equation (4) was employed in a streamline function in the electrostatics module of COMSOL 4.0 (COMSOL Inc., Burlington, MA) to simulate the particle trajectories. As the dielectrophoretic focusing of particle remained the same along the channel depth, a 2D model was employed. At the inlet, particles were assumed massless and uniformly distributed along the channel width. The correction factor, c, was determined by matching the predicted particle trajectories to the width of focused stream in the exit region in the experiment.

Figure 2. Photograph of the microfluidic chip used in the experiment for particle focusing. The inset indicates the structure and dimensions of the waved microchannel.
5. Results and discussion

Following the experimental and numerical approach previously presented, a parametric study was performed to understand the focusing performance of particles in the proposed waved microchannel. The particle motion in insulator-based DEP microfluidic devices is dependent on the three effects of EP, EOF and DEP. Thus, the effects of a variety of parameters, including the applied electric field, particle size, zeta potentials of particle and suspending medium on the particle motion were studied via both experiments and simulations. In addition, the focusing of yeast cells under an electric field with magnitudes of 100 and 200 V cm\(^{-1}\) was examined.

5.1. Effect of electric fields

Experimentally observed trajectories of the 10 \(\mu\)m particles under the effect of applied dc electric fields are presented in figures 3(a)–(d). In order to demonstrate the focusing effect as the particle transport through the waved channel, the superimposed particle image at the inlet of the waved channel (figure 3(a)) is included, from which it is found that particles are unfocused and almost covering the entire width of the straight microchannel before entering the waved channel.

Figures 3(b)–(d) are superimposed images of focused particle streams at the outlet of the waved channel obtained under nominal dc electric fields varied from (b) 100 to (c) 200 and (d) 300 V cm\(^{-1}\), which correspond to the actual applied potentials across the 2 cm long microchannel of 200, 400 and 600 V, respectively. At every electric field, particles are focused to move in a tight stream along the channel centerline at the outlet of the waved microchannel. By comparing the width of the focused stream, it is obvious that 10 \(\mu\)m particles obtain a better focusing performance when the electric field magnitude is increased, as the stream width at the outlet becomes narrower. This is mainly due to the repulsive dielectrophoretic motion that directs the particle away from the inner wall to the channel’s center region. According to equations (1) and (3), EK motion is linearly proportional to the electric field, while dielectrophoretic motion depends on the gradient of the square of the electric field; thus dielectrophoretic motion grows faster than EK motion as the magnitude of electric field increases, which contributes to a more effective focusing effect.

Figures 3(e)–(h) illustrate the numerically predicted 10 \(\mu\)m particle trajectories at both the inlet and outlet of the waved section under the same conditions as those in the experiments. In the simulations, the zeta potential values of particles and PDMS-glass channel wall were set to \(-33\) and \(-89\) mV, respectively [41, 42]. The correction factor was set to be 0.5 by matching the width of simulated stream to that of experimentally obtained result at 200 V cm\(^{-1}\), which remained constant in all the cases. In general, the simulated trajectories show good agreement with the experimentally obtained particle focusing behaviors.

5.2. Effect of particle size

In order to study how particle size affects the focusing performance, the negative dielectrophoretic focusing of 15 \(\mu\)m particles through the waved channel was examined. The left column of figure 4 shows the experimentally obtained superimposed images of focused particles at the outlet of the waved channel under a nominal electric field of (a) 100 and (b) 200 V cm\(^{-1}\). The width of the focused particle stream decreases with the increase of applied electric field, indicating 15 \(\mu\)m particles obtained a better focusing performance at higher applied electric field, which shows the same trend as 10 \(\mu\)m particles. Additionally, focusing of larger particles is more effective than that of smaller particles. This is demonstrated by comparing the stream width of focused 10 and 15 \(\mu\)m particles under the same condition (see figure 5): both the experimental and numerical values of the focused stream width for 15 \(\mu\)m particles were smaller than those for...
Figure 5. The effects of the electric field and particle size on the particle focusing performance. Both the experimental and numerical values of measured width of focused particle stream at the outlet are shown.

10 μm particles in all the three different applied electric fields. Given by equation (2), DEP force which is responsible for the particle cross-stream transpose depends on the cube of particle diameter. In comparison with smaller particles, larger particles experiencing larger repulsive negative DEP forces tend to be deflected further toward the channel center, resulting in a narrower focused stream along the channel centerline. Figure 5 shows that excellent focusing performance was achieved when focusing 15 μm particles at applied electric field strength of 200 and 300 V cm\(^{-1}\) (the measured stream widths were around 20 and 16 μm, respectively). However, the measured stream widths of numerical results were narrower (less than 15 μm). This discrepancy may be attributed to the neglect of particle size effect.

The numerically predicted trajectories for 15 μm particles under the same experimental conditions are depicted in figures 4(c) and (d). The zeta potentials of 15 μm particles and wall with the values of −33 and −89 mV, respectively, were the same as those for 10 μm particles given before. The correct factor was set to be 0.4.

5.3. Effect of medium concentration

Figures 6(a) and (b) compare the focusing of 5 μm particles in the waved microchannel in two different medium solutions: (a) 1 and (b) 10 mM NaCl solution under the same applied electric field of 200 V cm\(^{-1}\). It was observed that increasing the medium concentration improves the focusing performance, because the width of the particle stream in the 10 mM NaCl solution is narrower than that in the 1 mM solution. This is mainly due to the impact of the difference in electroosmotic mobility on the overall EK mobility of particles: the higher the ionic strength of the suspending medium, the more the double layer is compressed, so the lower the resulting EOF [45]. The increase in the medium concentration reduces the EOF, which in turn causes the decrease in the EK mobility. At lower EK velocity, the time for the DEP force to deflect particles as they progress through the waved channel is longer, the focusing performance is therefore enhanced.

In the simulations to predict the trajectories of focused particles, the zeta potential values of microchannel walls in 1 and 10 mM NaCl buffer solutions were set to be −89 and −54 mV, respectively [42]. The correct factor, \(c\), was fixed at 0.8. The numerically predicted cell trajectories (figures 6(c) and (d)) are in reasonable agreement with the experimentally obtained superimposed images (figures 6(a) and (b)) for both medium concentrations.

5.4. Focusing of yeast cells

The focusing behaviors of yeast cells suspending in 0.2 × PBS in the waved channel at different applied electric field were studied. Both the experimentally observed (a)–(c) and numerically predicted (d)–(f) images under two different nominal electric fields with the magnitude of 100 (middle row) and 200 V cm\(^{-1}\) (bottom row) are presented in figure 7. Under both conditions, yeast cells that initially unfocused and almost uniform distributed (covering almost the channel width) at the inlet of the waved channel (figure 7(a)) were moved into focused streams along the channel centerline at
the exit region. By comparing the stream width within the straight section at the outlet, it is found that the performance of cell focusing is improved with the rise of field magnitude. This expected trend is the same as that previously obtained for both 10 and 15 μm particles, and is because of the faster growing dielectrophoretic motion as well.

The simulated results in the right column of figure 7 show a reasonable agreement with the experimentally obtained superimposed images, in which the zeta potentials of channel wall and yeast cells were set to be $-46$ and $-16$ mV, respectively [43, 44], and the correction factor was set to be 0.8 under both conditions.

In the experiments of yeast cell focusing, no dc electric fields stronger than 200 V cm$^{-1}$ were applied. This is due to the concern of Joule heating of the buffer solution and shear stress on the cell membrane associated with the high dc voltage, both of which may destroy the yeast cells. The cell viability was examined using a staining method: living cells can produce the enzyme which breaks down methylene blue, leaving the cells colorless, while non-viable cells cannot produce this enzyme and appear to be colored. By staining samples of the yeast cells from both the inlet and outlet reservoirs, it was found that more than 85% of the yeast cells were still alive after the experiment was conducted with the highest applied dc electric field of 200 V cm$^{-1}$.

6. Conclusions

We reported a waved microchannel to achieve continuous and effective focusing of polystyrene microparticles and yeast cells into a confined and concentrated stream along the channel centerline. This design utilizes the cross-stream negative dielectrophoresis force induced by the electric field gradient in the waved channel consisting of successive semicircular microchannels with alternative directions. Particles that continuously transport through the waved channel by EK effect are gradually directed toward the center region of the microchannel. The effects of parameters, including the applied electric field, particle size and medium concentration, on the focusing efficiency were investigated by both experimental and numerical methods. The results indicated that the focusing performance increases with the rise of the applied electric field, particle size and medium concentration.

The advantages of the microfluidic device proposed herein are obvious: (1) the induced inhomogeneous electric fields cover the whole waved microchannel, resulting in a decrease of the total length of the curved section and the applied electric field necessary for the desired focusing performance; (2) the elimination of the stagnation regions due to sharp turns and effective focusing of particles along the channel’s centerline reduce the possibility of particle’s adhesion on the channel wall; (3) the focusing performance of particles flowing through the waved channel is controlled by simply adjusting the applied electric field or suspending medium concentration. Furthermore, the waved channel (i.e. its width, length, radius and configuration) can be optimized according to the requirement of various applications; (4) without external pressure pumping, in-channel microelectrodes and insulating obstacles, the proposed microfluidic device is simple in terms of operation and fabrication, and allows a better control of surface fouling and chemical reaction; (5) this design eliminates the locally amplified electric field, and induced large shear stress and Joule heating, indicating a great potential for biological applications, such as cell manipulation. The proposed waved channel for the continuous particle and cell focusing is anticipated to integrate different functionalities, such as detection, calculation, separation and analysis, into a single LOC device for a widespread use in the field of biology, medicine and chemistry.

Acknowledgment

The authors wish to thank the University of Wollongong funding support to promote the collaboration with The Hong Kong University of Science and Technology through an international linkage scheme.

References

[15] Xuan X and Li D 2005 Focused electrophoretic motion and selected electrophoretic dispensing of particles and cells in cross-microchannels *Electrophoresis* **26** 3552–60


[25] Xuan X, Raghhibizadeh S and Li D 2006 Wall effects on dielectrophoretic motion of spherical polystyrene particles in a rectangular poly(dimethylsiloxane) microchannel *J. Colloid Interface Sci.* **296** 743–48


[33] Song H, Mulukutla V, James C D and Bennett D J 2008 Continuous-mode dielectrophoretic gating for highly efficient separation of analytes in surface micromachined microfluidic devices *J. Micromech. Microeng.* **18** 125013


[40] Kang K, Kang Y, Xuan X and Li D 2006 Continuous separation of microparticles by size with dc-dielectrophoresis *Electrophoresis* **27** 694–70


