Concentrating molecules in a simple microchannel

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A simple method is proposed and tested to concentrate sample molecules from a dilute solution in a microchannel by electrokinetic means. The microfluidic chip has a straight microchannel connecting two wells and three electrodes. This method uses electrokinetic trapping and flow control simultaneously to concentrate a charged species of interest. A numerical model of the sample concentration process is presented in this paper. Using a fluorescent dye as the sample molecules, experimental investigation into the concentration process was performed. The 90 times of the concentration increase was achieved in 110 s. The numerical simulations of the concentrating and the subsequent dispensing processes agree well with the experimental results.

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1. Introduction

In microfluidic devices, the concentration of sample molecules directly affects the effectiveness of the biochemical reaction and the ability to detect the sample molecules (such as DNA and proteins) in the solution [1,2,4]. The ability of concentrating the sample molecules in one specific area is key to the success of many on-chip biochemical assays. For example, DNA detection has become a key technology for various biomedical applications. Currently, polymerase chain reaction (PCR) has to be used to amplify the amount of DNA from the initial sample, before any DNA detection techniques can be applied. If the small amount of the DNA molecules from the initial sample can be concentrated and collected, this could enable many biomedical detection methods directly on a microfluidic chip.

There are a number of papers reporting studies of sample concentrating processes in microchannels. Numerical simulation of DNA sample pre-concentration in a microdevice by electrophoresis was introduced by Srivastava et al. [3]. This study presented a computational tool for understanding and designing sample-injectors and electrophoretic protocols for DNA separation in a double-T injector. In another paper, electro-pre-concentration process in a microchannel by electrokinetic means. The microfluidic chip has a straight microchannel connecting two wells and three electrodes. This method uses electrokinetic trapping and flow control simultaneously to concentrate a charged species of interest. A numerical model of the sample concentration process is presented in this paper. Using a fluorescent dye as the sample molecules, experimental investigation into the concentration process was performed. The 90 times of the concentration increase was achieved in 110 s. The numerical simulations of the concentrating and the subsequent dispensing processes agree well with the experimental results.

Various stacking methods and focusing methods have been developed for concentrating samples by transport the sample to a local equilibrium state electrokinetically. Isoelectric focusing (IEF), which is also known as electrofocusing, is a technique for separating different molecules by their electric charge differences. This method uses pH gradients along a microchannel to focus charged samples at their isoelectric points (pI) at which the electrophoretic force exerted on the samples vanishes. Cabrera and Yager used natural pH gradients without the use of an ampholyte to concentrate bacteria [5]. Huang and Pawliszyn used thermally driven pH gradients in a tapered microchannel to perform IEF [6]. Li et al. generated pH gradients using a commercially-available carrier ampholytes [7]. It should be noted that IEF has also been used to separate samples according to their pI values [8]. Despite the high resolution of IEF in discerning analytes, as fine as 0.01 pH, applications in biochemical analysis are limited, partially because of the low solubility of proteins at their pIs and because not all analytes have a specific pI. Field amplified sample stacking (FASS) and isostaphophoresis (ITP) use two or more buffers with different conductivity to generate electric field variation along a microchannel. A key to performing FASS or ITP is to inject a sample plug into the system without an electrokinetic injection bias. As a result, a hydrodynamic sample-injection scheme is often used. Wainright et al. coupled ITP and zone electrophoresis on a microchip to concentrate and subsequently separate samples, resulting in an approximately 400-fold improved sensitivity [9].
Jung et al. reported an 1100-fold signal increase on using FASS combined with a porous polymer structure for improved sample injection and flow control [10]. Beard et al. implemented FASS to detect 20-pM concentrations of biogenic amines [11].

Shaikh et al. [12] achieved simultaneous concentration, focusing, and metering capabilities with current-generation sample-injection technology. They used microfluidic chips incorporating arrays of individually addressable microfabricated electrodes and demonstrate that DNA can be sequentially concentrated, focused into a narrow zone, metered, and injected into an analysis channel. Their technique transports charged biomolecules between active electrodes upon application of a small potential difference (1 V) and is capable of achieving orders of magnitude concentration increases within a small device footprint. The collected samples are highly focused, with sample zone size and shape defined solely by electrode geometry. Despite their good results, fabricating such chips is complicated and needs high accuracy.

The integration of PCR with DNA separation on a single microchip was experimentally demonstrated by Lettieri et al. [13]. They described the trapping, pre-concentration, and release of large DNA molecules in microfluidic devices using controlled micro recirculating flows. Their method was used to extract and purify DNA from mixtures containing other substances. A convenient method capable of providing highly resolved, adequately sensitive, and reproducible separation of DNA fragments was reported by Chen et al. [14]. They demonstrated an online pre-concentration and separation of DNA fragments by using uncoated capillaries together with UV detection. This technique involved a high degree of automation, used minimum quantities of the samples and reagents, and was able to make reproducible separations of PCR products. The usefulness of this method was demonstrated by detecting exogenous genes in transgenic oilseed rape.

Svarnas et al. [15] presented the fabrication of a hybrid micro and nanofluidic device to study electro-pre-concentration of biomolecules. In their study, they applied a bulk-like machining method and developed a glass nanopore-based system in which four reservoirs are connected to four microchannels, respectively, in an “H”-shaped channel network via nanochannels. The sharp contraction from a microchannel to a nanochannel creates a strong local electric field. This field is used for electrokinetic trapping of negatively charged molecules. They investigated concentration of charged molecules under the effect of concentration polarization. Huang and Yang [16] experimentally investigated the effect of concentration polarization using a hybrid microchannel and nanochannel interface. They controlled the flow by applying a two-step voltage. Their system contains two microchannels (one V-shape and one cross-form) connected by a short nanochannel. They tried to produce a low-conductivity region within the loading channel by creating an ionic depletion effect at the anodic side of the nanochannel. In practical applications, manufacturing nanochannels is not simple and hence limits these methods.

du et al. [17] experimentally evaluated a microfluidic device for trapping and concentrating a trace amount of DNA molecules. They used a quadrupole electrode platform to generate a nonlinear electroosmotic flow by charge polarization under high-frequency AC fields. Their trapping approach was validated over a large range of DNA concentrations. Using this method led to significant and more efficient concentration enhancement within a few seconds. They could switch between trapping and release of DNA, showing potential in concentrating and transporting biomolecules in a continuous fashion. However, using such a quadrupole electrode device and high-frequency AC fields brings complexity to many applications of microfluidic lab-on-a-chip devices.

The methods reviewed above generally require complicated fabrication of the microscale electrodes or nanoscale channel structures and complex process control. It is highly desirable to be able to concentrate sample molecules from a dilute solution in a simple microchannel using a simple process. In this study, an electrokinetically-driven concentration method in a straight, closed-end microchannel is developed. This method utilizes simple electric field applied via three electrodes along a microchannel to concentrate the sample molecules from a dilute solution in a well and then transport the concentrated sample molecules along the channel. This microfluidic chip is simple to make and easy to operate, and can increase the sample concentration significantly.

2. The microfluidic chip design

Fig. 1 illustrates the microchannel-electrodes system which is used to study the concentration process. The system consists of two Reservoirs, A and B, each with a diameter of D, and one microchannel of width 2a and length L. The reservoirs and the channel are initially filled with a buffer solution. A dilute sample solution is added to the Reservoir A. Three electrodes, E1, E2, and E3, are used in this system, to control the electric field in the different processes. Electrodes 1 and 3 are placed at the far-end of the Reservoirs A and B, respectively. Electrode 2 is mounted at the entrance of the microchannel from Reservoir A. First, we wish to concentrate the negatively-charged sample molecules placed in the dilute solution in Reservoir A by applying an electric field. During the concentration process, an electric field gradient is created between E1 and E2, so that the negatively-charged sample molecules will migrate to the vicinity of E2. In the meantime, an electric field between E2 and E3 is produced to ensure zero net flow at the channel cross-section where E2 is located. After the sample molecules are concentrated in the region of E2, a different electric field between E2 and E3 will be used to produce a flow to transport the concentrated sample plug along the microchannel towards Reservoir B.

For numerical studies, Cartesian coordinate system is chosen, as indicated in Fig. 1. The liquid in the microchannel is assumed to be

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**Fig. 1.** Two-dimensional (x–y plane) illustration of the closed-end microchannel used in this study of sample concentration and transport.
an incompressible and Newtonian, with symmetric electrolytes, constant density, \( \rho \), and viscosity, \( \mu \). The channel wall is uniformly charged with a zeta potential, \( \zeta \). The temperature is assumed to be uniform and constant.

3. Mathematical modeling

Patankar et al. [18] has proved that the velocity component in the z-direction (the direction of the channel depth) is very small, one can apply a two-dimensional (the \( x \)-\( y \) plane) study at any given height and obtain a highly accurate description of the flow field with a lower computational cost. Therefore, we will consider the transport processes in two-dimensions (\( x \)-\( y \) plane). Based on the assumption of a steady state, incompressible, laminar flow, the motion of the electrolyte solution under the application of an external electric field, \( E \), in the closed-end microchannel can be described by the following equations.

3.1. The electric field

According to the theory of electrostatics, the applied electrical potential, \( \phi \), in the reservoirs and in the microchannel can be described by Poisson equation,

\[
\nabla^2 \phi = 0
\]

(1)

Boundary conditions are required to solve this equation. We impose the insulation condition to all the walls of the reservoirs and the microchannel, and the specific potential values at each electrodes. Once the electrical field is known, the local electric field strength can be calculated by \( E = -\nabla \phi \).

3.2. The flow field

The flow field is defined by the continuity and momentum equations.

\[
\nabla \cdot \vec{V} = 0
\]

(2)

\[
\rho \vec{V} \cdot \nabla \vec{V} = -\nabla P + \mu \nabla^2 \vec{V}
\]

(3)

where \( \vec{V} \) is the net velocity determined by both the electroosmotic flow and the induced pressure-driven flow. \( P \) is the pressure, \( \mu \) is the kinematic viscosity of the fluid, and \( \rho \) is the density of the liquid. Assuming the gravity effect is negligible; then, the body force term will be the electric force which is \( \vec{F} = -\vec{E}_r \rho_e \phi(x) \). Here, \( \rho_e \) is the local net charge density and \( \vec{E}_r \) is the electrical field strength. Generally, the microchannel wall surface is negatively charged, the net charge in the electric double layer (EDL) is positive. Since the local net charge density is not zero only in the electrical double layer (EDL), the driving force for the electroosmotic flow, \( \vec{E}_r \rho_e \), exists only in EDL. The electroosmotic flow velocity changes sharply in a thin layer of liquid near the wall, from zero at the channel wall surface to an approximately constant value at the outer edge of the EDL. Considering thin EDL and for the purpose of modeling the bulk liquid flow outside the EDL, the body force term \( \vec{E}_r \rho_e \) in Eq. (3) will be dropped off and the electroosmotic effect is considered by the slip velocity boundary condition: \( V_{\text{slip}} = \mu_{\text{slip}} \vec{E} \) on the channel wall according to Smoluchowski equation. Eq. (3) is subject to the following boundary conditions on the wall of system in the two-dimensional (\( x \)-\( y \) plane) system:

\[
0 < x < D + b \quad V_{\text{slip}} = \mu_{\text{slip}} E_{1-2}
\]

(4a)

\[
D + b < x < 2D + L \quad V_{\text{slip}} = \mu_{\text{slip}} E_{3-2}
\]

(4b)

where \( E_{1-2} \) and represents the electric field between electrode E2 and E1, same as \( E_{3-2} \) means the electric field which is applied between electrode E3 and E2.

3.3. The concentration field

The distribution of the sample molecules or the concentration field can be described by the mass concentration law in the form of transient mass transport equation,

\[
\frac{\partial (C_i)}{\partial t} + \nabla \cdot \left( (\vec{V} + \vec{V}_{\text{epi}}) C_i - D_i \nabla C_i \right) = 0
\]

(5)

where \( C_i \) is the concentration of the \( i \)th species, \( D_i \) is the diffusion coefficient of the \( i \)th sample molecules. \( \vec{V}_{\text{epi}} = \mu_{\text{epi}} E \) is the electrophoretic velocity of the \( i \)th species, where \( \mu_{\text{epi}} \) is the electrophoretic mobility. Eq. (5) is subject to the following initial and boundary conditions:

\[
0 < x < D + b \quad C_i \big|_{x=0} = 0
\]

(6a)

\[
D + b < x < 2D + L \quad C_i \big|_{x=L} = 0
\]

(6b)

\[
x = 2D + L \quad C_i \big|_{x=0} = 0
\]

(7)

\[
y = 0 \quad \frac{\partial C_i}{\partial n} = 0
\]

(8)

At walls \( \frac{\partial C_i}{\partial n} = 0 \)

(9)

It should be realized that the above three equations are coupled together by the net flow velocity. Therefore, the process of concentrating the sample molecules in a closed-end micro-channel involves the electroosmotic flow, the counter flow caused by the induced pressure, and the electrophoretic motion of the charged sample molecules.

4. Numerical method

The present numerical study is conducted by using FLUENT and employing a mesh independent structure to make sure that the results are unique and will not change if any other grid distribution applies. The flow field is solved as a steady state flow and coupled to the transient concentration field equation via the velocity term in Eq. (5). In order to discretize the solution domain, the unstructured quadratic meshes were applied. The solution domain is broken into quadratic small meshes to allow that the meshes fully cover the solution domain without overlapping. In order to find reliable results which are grids independent, we examined different number of grids. Finally we found the number of grids with which the numerical results will not change if we further increase the number of grids. In this study, the overall flow is driven by the electroosmotic flow along the microchannel walls. In order to verify the numerical method, the flow field of a two-side wall driven flow in a cavity was solved and compared with the result published by Kalita et al. [19]. Fig. 2a represents the streamline patterns of such a cavity flow. Fig. 2b shows the corresponding velocity vectors. The flow field results of our numerical study demonstrated excellent agreement with that of Kalita et al. [19], which verifies our numerical method.

5. Experimental

5.1. Chip and electrode fabrication

The polydimethylsiloxane (PDMS) microchip was fabricated on a glass substrate using soft lithography protocol. Details can be found elsewhere [20]. Briefly, the master was prepared by first spin coating the film of SU-8 25 negative photo-resist onto a silicon wafer. After pre-baking, a photomask bearing certain microchannel geometry, i.e. a straight channel with two electrodes (E2), was placed on the top of the film. The photo-resist film covered with
the mask was then exposed to UV light. After post-baking and developing, the master could be obtained. Then, a 10:1 (w/w) mixture of PDMS polymer base and curing agent was poured over the master and cured at 75 °C for 3 h after being degassed under low vacuum. The layout of the microfluidic chip is shown in Fig. 3. In this design, the microchannel’s width and the depth are 200 μm and 20 μm, respectively; the length of the channel is 10 mm. The diameter of both Reservoirs (A and B) is 5 mm. Branch C and D are chambers for placing the electrodes E2.

The thickness and the length (in the flow direction) of the embedded copper microelectrode are 18 μm and 500 μm, respectively. The fabrication procedure of the copper microelectrode is developed from the soft lithography. First, The Pyralux AP single-sided copper-clad laminate AC 182000R (Dupont Electronic Materials, Research Triangle Park, NC) was spin-coated with a layer of SU-8 at 2000 rpm. After the baking treatment, a photomask bearing the electrode geometry was placed on the top of the copper film and exposed to UV light. After post-baking, the unexposed photo-resist is dissolved by the chemical developer (MicroChem, Newton, MA). The copper sheet covered with a layer of exposed photo-resist of the desired pattern is emerged in the copper etchant CE-100 (Transene Company Inc., Danvers, MA) at 150 °C until all the uncovered copper is removed. Finally, the copper electrodes are placed in NaOH (30%) bath at 130 °C for 60 min to remove the polyimide substrate and the photo-resist. The microelectrodes are inserted carefully into the electrode chambers on the PDMS chip manually under a microscope (AZ100, Nikon, Japan). After that, the PDMS slab was plasma treated and bonded with a glass slide (VWR) to form the microchannel with embedded electrodes. Because the thickness of the microelectrode (18 μm) was slightly smaller than the depth of the PDMS electrode chamber (20 μm), a droplet of liquid PDMS was used to seal the gap between the PDMS slab and the glass substrate by capillary action, and followed by heating on a hotplate at 200 °C for several seconds to make it solidified.

6. Materials and experimental setup

In order to visualize the concentration process, a fluorescent dye was used as the sample molecules. The working solution was diluted from Dyelight™ 488 NHS-Ester (Fisher, Canada) in Dimethyl sulfoxide (DMSO) buffer. The dye concentration is 0.1 mg/mL. The excitation and emission light wavelength of the fluorescence dye are 488 nm and 518 nm, respectively. The experimental system consists of the following major units: the microfluidic chip, a fluorescence microscope and image system, a power supply, and a data acquisition system, as shown in Fig. 4. The microfluidic chip was fixed on the observation platform of the microscope. A DC-regulated power supply (CSI12001X, Circuit Specialists Inc., USA) was used to control the voltages to the electrodes. A fluorescent microscope (AZ100, Nikon) with a high-intensity illumination system and a CCD camera (DS-QiMC, Nikon) was used to monitor the alignment and visualize the dye motion inside the microchannel. The images were captured by the digital camera and digitized by the computer software (NIS-Elements BR 3.0).

7. Experimental operation

Initially, the Reservoirs A and B and the microchannel were filled with DI water. To start the experiment, 20 μL of the dye solution was loaded to the Reservoir A; and another 20 μL of water was
added to Reservoir B at the same time, in order to maintain the same level of liquid in all reservoirs to minimize differential pressure effects. During the concentration process, 25 V was applied at electrode E2, zero at the electrode E1 in the Reservoir A. In this electric field the negatively-charged dye molecules migrated to the vicinity of the electrodes E2. In the meantime, 22.5 V was applied at the electrode E3 in the Reservoir B, generating a voltage difference of 2.5 V between the electrodes E2 and E3. This small electric field is used to ensure zero net flow at the channel cross-section where the electrodes E2 were located. After the dye molecules were concentrated in the region of the electrodes E2, 20 V was applied between electrode E1 in Reservoir A and electrode E3 in Reservoir B. This electric field generated a net flow that transported the concentrated dye molecules along the microchannel towards the Reservoir B. Sequential images were taken during the experiments, and the image analysis was done by using the Nikon Elements BR program.

8. Results and discussion

In the concentration process, one electrical field is applied between electrodes E1 and E2, so that the negatively-charged sample molecules will move to a small region close to E2; while another electrical field between electrodes E2 and E3 is setup in such a way that at the end of electrode E2 position there is no liquid flow crossing that cross-section. This can be seen clearly by the numerically simulated flow field as shown in Fig. 5. This is to ensure the sample molecules migrating to the vicinity of E2 will not be carried away by the flow. In the dispensing process, after the sample molecules are concentrated in the region of E2, a different electric field between E1 and E3 will be used to transport the plug along the microchannel towards Reservoir B.

When the dye molecules are concentrated, the fluorescent light intensity is increased proportionally. As an example, Fig. 6 shows the sequential images of the fluorescent intensity in the region near the electrode E2. In this study, we use the fluorescent light intensity change as the indication of the concentration change of the dye molecules. In order to calculate the efficiency of the concentration process, a fixed detection area (the area of the microchannel between the two electrodes E2) was used to calculate the average intensity. First of all, several images of the Reservoir A were taken before the electric field was applied. The intensity value, \( I_0 \), was recorded as the indicator of the initial bulk concentration of the fluorescent dye in the Reservoir A. After the electric field was applied and the concentration process began, sequential images were taken at a fixed time interval. For a given moment, several images were taken to obtain an average light intensity. The average intensity of these images, \( I(t) \), in Table 1, was determined by the Nikon Elements BR program. Before computing the concentrating effect, several images of the detection area were also taken without using the excitation light of the fluorescent micro-

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Fig. 4. Illustration of the experimental system.

Fig. 5. Velocity vector distributions in the reservoirs and along the microchannel during the concentrating process.
scope. The average light intensity of these images was considered as the background noises, $I_{bn}$ due to the CCD camera and non-specific light effects in the lab. The times of the concentration increase was then calculated by:

$$
\text{Times of the concentration increase} = \frac{I(t) - I_{bn}}{I_0 - I_{bn}} \quad (10)
$$

Using the measured light intensity data and Eq. (10), it was found that the dye concentration in the area near the electrodes E2 increased by 90 times after applying the electric field for 110 s, as shown in Table 1.

The same concentration process was simulated numerically and compared with the experimentally measured concentration/light intensity change. The comparison is shown in Fig. 7. The concentration field and the average concentration of the dye molecules in the region near the electrodes E2 as a function of time was determined by numerically solving the mathematical model presented above. In the simulation, the electrophoretic mobility and the diffusion coefficient of this dye molecules are $-3.3 \times 10^{-8} \text{ m}^2 \text{ V}^{-1} \text{ s}^{-1}$ and $4.37 \times 10^{-10} \text{ m}^2 \text{ s}^{-1}$, respectively [21]. The average zeta potential of the PDMS-glass microchannel is $-60 \text{ mV}$. The concentration change in comparison with the initial concentration in the Reservoir A was then calculated by using Eq. (5). The results are shown in Table 2. Clearly, the numerical simulation is in a good agreement with the experimental data.

![Fig. 6. Sequential images of the concentration process of a fluorescent dye in the microchannel.](image)

<table>
<thead>
<tr>
<th>Time (s)</th>
<th>$I(t)$</th>
<th>$I_0$</th>
<th>$I_{bn}$</th>
<th>Times of the concentration increase</th>
</tr>
</thead>
<tbody>
<tr>
<td>15</td>
<td>649.16</td>
<td>122.906</td>
<td>95.676</td>
<td>20.3</td>
</tr>
<tr>
<td>30</td>
<td>1264.82</td>
<td>122.906</td>
<td>95.676</td>
<td>42.9</td>
</tr>
<tr>
<td>60</td>
<td>2116.66</td>
<td>122.906</td>
<td>95.676</td>
<td>74.2</td>
</tr>
<tr>
<td>100</td>
<td>2411.52</td>
<td>122.906</td>
<td>95.676</td>
<td>85.0</td>
</tr>
<tr>
<td>110</td>
<td>2553.66</td>
<td>122.906</td>
<td>95.676</td>
<td>90.3</td>
</tr>
</tbody>
</table>

After the concentration process, the sample (the fluorescent dye in this case) molecules are concentrated and formed a plug in the region of the electrodes E2. To transport this concentrated sample plug to the downstream of the microchannel, a different electric field was applied between E1 and E3: 20 V at E1 and 0 V at E3. This electric field produces the electroosmotic flow of the liquid in the channel from the left to the right and also the electrophoretic motion of the fluorescent dye molecules from the right to the left. In addition, it should be noted that this channel has two closed ends (Reservoirs A and B). To meet the flow continuity requirement, there is an induced pressure-driven flow from the Reservoir B towards Reservoir A. Therefore, the transport of the concentrated dye plug is determined by net flow velocity $V_{\text{net}} = V_{\text{eo}} + V_{\text{p}} + V_{\text{eph}}$. In this study, the averaged (over the channel cross-section) electroosmotic flow velocity $V_{\text{eo}}$ and the averaged (over the channel cross-section) pressure-driven flow velocity can be determined by the numerical solution of the flow field; their values are $174.6 \mu \text{m/s}$ and $-82.7 \mu \text{m/s}$, respectively (here the negative sign means the direction of the velocity is opposite to the x direction). The electrophoretic velocity of the dye molecules can be calculated as $V_{\text{ep}} = \nu_{\text{ep}} E = -66 \mu \text{m/s}$. Thus, the net flow velocity is $V_{\text{net}} = V_{\text{eo}} + V_{\text{p}} + V_{\text{eph}} = 25.9 \mu \text{m/s}$, and the moving direction of the concentrated dye sample is from left to right. Fig. 8 shows the comparison of the numerically simulated sample dispensing process and the experimentally observed dispensing process.
The 90 times of the concentration increase reported here is definitely not the limit of this method. There is no doubt that this method can achieve much higher concentration, for example by using a higher electric field. However, it should be realized that in this paper we used a fluorescent dye to visualize the concentration change; this approach significantly limited our ability to show any higher concentration change. It is well-known that the fluorescent dye will be photo-bleached under electric field for sufficiently long time. That is why we used only a lower electric potential, 25 V, between E1 and E2 during the concentration process. In fact, while more and more fluorescent dye molecules were accumulating in the E2 region, some of the dye molecules in this region were fading. We observed that the light intensity in the E2 region reached maximum at about 110 s. After that, the fluorescence intensity began to decrease.

Also, it should be pointed out that the objective of this work is to proof the concept, that is, the sample molecules can be concentrated in such a simple microchannel. One of our future research works will be how to utilize the concentrated samples on the chip. For example, additional channels can be introduced at the downstream; the concentrated sample plug will be transported to a side channel where the sample molecules mix with other reagents to have a biochemical reaction such as DNA hybridization or immunoassay.

9. Conclusion

A simple electrokinetic microfluidic method is demonstrated in this paper to concentrate sample molecules from a dilute solution. The experimental study shows that 90 times increase in concentration can be achieved in 110 s. This method does not require any complicated microfabrication of electrodes and nanochannels. The process is easy to operate by simple change of the applied electric field via three electrodes.
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