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On-chip pressure generation using a gel membrane fabricated outside of the microfluidic network

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Abstract

On-chip generation of pressure gradients via electrokinetic means can offer several advantages to microfluidic assay design and operation in a variety of applications. In this article, we describe a simple approach to realizing this capability by employing a polyacrylamide-based gel structure fabricated within a fluid reservoir located at the terminating end of a microchannel. Application of an electric field across this membrane has been shown to block a majority of the electroosmotic flow generated within the open duct yielding a high pressure at the channel–membrane junction. Experiments show the realization of higher pressure-driven velocities in an electric field-free separation channel integrated to the micropump with this design compared to other similar micropumps described in the literature. In addition, the noted velocity was found to be less sensitive to the extent of Debye layer overlap in the channel network, and therefore more impressive when working with background electrolytes having higher ionic strengths. With the current system, pressure-driven velocities up to 3.6 mm/s were realized in a 300-nm-deep separation channel applying a maximum voltage of 3 kV at a channel terminal. To demonstrate the separative performance of our device, a nanofluidic pressure-driven ion-chromatographic analysis was subsequently implemented that relied on the slower migration of cationic analytes relative to the neutral and anionic ones in the separation channel likely due to their strong electrostatic interaction with the channel surface charges. A mixture of amino acids was thus separated with resolutions greater than those reported by our group for a similar analysis previously.

Keywords

Gel membrane; Integrated device; Ion chromatography; Microfluidic pump; Pressure-driven flow

1 Introduction

The generation of pressure gradients on microchip devices is desirable for both realizing novel functionalities as well as improving the performance of a variety of routine liquid phase assays on this platform. Use of such pressure-driven flows (PDFs) for enabling efficient sample transport, preconcentration and separation methods are perhaps some important applications of this technology, which have the potential to significantly enhance

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the analytical capabilities of microfluidic systems [1-4]. Although the simplest approach to generating PDF within a microchannel is to integrate it to an external syringe pump, this strategy offers poor dynamic control over the PDF velocity due to the large dead volumes involved in the tubings/flow adaptors at the microchip–pump interface. In addition, leakage of liquid at this junction tends to be an issue with such a design when fluidic ducts shallower than 5 μm are employed producing large pressures at the interconnects. Although on-chip generation of pressure gradients presents a practical path to addressing some of these challenges, more research and development is currently needed for maturing this capability [5-8].

A variety of approaches to generating pressure gradients on-chip have been demonstrated in the literature over the past two decades. These include employing electrokinetic, mechanical, thermopneumatic, and electrochemical means among others to realize PDFs within microchannels with high precision [5, 7]. Liu et al., for example, have reported on the generation of PDF velocities in the range of cm/s by expanding air in a heated chamber as well as via electrochemical gas generation for driving liquid samples through a network of polycarbonate-based microchannels [9]. However, while the noted approach allowed the realization of large PDF velocities, it offered limited dynamic control over this transport rate. To accomplish rapid switching of PDFs within microchannels both in terms of their magnitude and direction, our research group has mostly focused on developing pressure gradients on-chip via electrokinetic means, which then have been successfully applied to improving the resolving power of microchip-based electrophoretic and chromatographic separation methods [10-16]. The strategy to realizing the pressure-generation capability in all our previous works involved the construction of a microfabricated structure, for example, a membrane [11, 12] or a shallow segment [10, 13-16], inside of a microchannel next to an open duct. Application of an electric field across this structure then enabled us to at least partly block the electroosmotic flow generated within the open channel leading to the development of a high pressure at the junction of the membrane/shallow region and the open duct. Notably, this approach to on-chip pressure generation offered tremendous dynamic control over the pressure gradients in our system allowing the rapid injection of sample zones into a separation channel. In addition, it was shown that the PDF velocity thus realized was nearly independent of the channel depth in the fluidic network enabling pressure generation in micro- and submicrometer scale separation columns with equal ease. Unfortunately, the noted approaches to pressure generation also suffered from some significant drawbacks, which may limit their usage in practical devices. In the case of a membrane based micropump, the consistent fabrication of the porous structure poses a challenge, whose structural integrity has been further observed to be frequently compromised under the influence of a large electric field. The pumping units involving a shallow channel segment on the other hand were observed to generate limited pressures when working with background electrolytes having ion concentrations greater than 1 mM [10, 15].

To address these limitations, we describe an alternate approach to designing an electrokinetically driven micropump, which relies on a membrane structure constructed within a fluid reservoir at the terminating end of a microchannel. In addition to reducing the fabrication effort in preparing the device, the construction of the membrane outside of the

fluidic network considerably improves the stability of this structure under the influence of an electric field. Moreover, our experiments show that the noted design yields higher PDF velocities for a given applied electrical voltage, and is relatively insensitive to the ion concentration in the background electrolyte compared to other similar micropumps described in the literature. The functionality of our micropump has been further demonstrated by integrating it to an electric field-free separation channel to drive a nanofluidic ion-chromatographic analysis based on the electrostatic interaction of analyte molecules with the channel surface charges. Moreover, this pressure-driven ion-chromatographic separation was observed to yield a higher peak resolution compared to similar separations reported in one of our previous works [15].

2 Materials and methods

2.1 Device fabrication

For fabricating the micro- or nanofluidic devices employed in this work, bottom substrates and cover plates made from borosilicate glass were purchased from Telic Company (Valencia, CA). Although the purchased cover plates had both their faces unprotected, the bottom substrates came with a thin layer of chromium and photoresist (AZ 1518) laid down on one of their surfaces. Custom-designed photomasks created through Fineline Imaging (Colorado Springs, CO) were used to pattern the desired channel layout onto the bottom substrate using standard photolithographic methods [11, 17]. The channel network shown in Fig. 1 was used in our experiments with all of its segments being 200 μm wide and uniformly deep. The length of the separation channel (segment F) was chosen to be 3.5 cm in all our devices. After completion of the photo-patterning process, the photoresist layer was cured in microposit developer MF-319 (Rohm and Haas) and the chromium layer removed along the channel network with a chromium etchant (Transene). The fluidic ducts were then etched to a depth between 0.3 and 5 μm using a solution of buffered oxide etchant purchased from Transene. The protective photoresist and chromium layers on the bottom substrate were subsequently removed using acetone and chromium etchant solutions, respectively. Access holes were drilled at the channel terminals using a micro-abrasive powder blasting system (Vaniman) to allow the introduction of liquid reagents/sample into the micro-/submicrometer deep channels. Finally, the micro- or nanofluidic network was sealed off by bringing a glass cover plate in contact with the bottom substrate in deionized water and allowing the two plates to bond at 550°C for 12 h [18]. Cylindrical glass reservoirs with internal volumes of about 150 μL were affixed over the access holes using a UV light curable glue (Norland Products) to serve as a source or sink for the liquid flow within the glass channels. We would like to point out that a 5- μm -deep fluidic network was chosen in many of our experiments as it offered a good balance between the efforts needed to fabricate the microchip while minimizing unwanted PDF in the system generated due to unequal hydrostatic heads at the liquid reservoirs. In general, glass substrates with a shallower fluidic network are more challenging to bond to a cover plate but produce smaller unwanted PDFs in the channels.

2.2 Membrane fabrication

Prior to the membrane fabrication process, the microchannels in our fluidic network were thoroughly rinsed with 0.1 M sodium hydroxide for about 10 min, followed by extensive rinsing with deionized water and 1 mM sodium tetraborate (pH 9.4) buffer. The glass surface within reservoir 3 and around the terminating end of channel segment C was then silanized to allow its covalent attachment to the membrane structure in order to minimize leakage around the glass–membrane interface. To this end, a 25 μL solution containing deionized water, methanol, glacial acetic acid, and 3-(trimethoxysilyl) propyl methacrylate at a ratio 3:3:2:2 was aliquoted into reservoir 3 while filling the remaining reservoirs with 25 μL of deionized water. After allowing the silanization reaction to occur for 30 min, the entire fluidic network was sequentially rinsed with methanol, deionized water, and 1 mM sodium tetraborate buffer three times each applying vacuum at reservoir 3. With the fluidic network now filled with the 1 mM sodium tetraborate buffer, about 75 μL of a 8–50% (v/v) premixed solution of acrylamide and bisacrylamide prepared in the same 1 mM sodium tetraborate buffer was introduced into reservoir 3 as the gel precursor material [19]. The premixed solution noted above contained 40% acrylamide/bisacrylamide in the ratio 37:1 (Sigma–Aldrich) along with 0.2% (w/v) of a photo-initiator, VA-086 (Wako Chemicals). The gel precursor solution was degassed prior to its introduction into reservoir 3 in order to prevent bubble generation in it during its photopolymerization by UV light. The remaining reservoirs at this point were filled with about 25 μL of the 1 mM sodium tetraborate buffer allowing the gel precursor solution to slowly flow into segment C for about 5 min due to a higher hydrostatic head at reservoir 3. The microchip was then placed under a UV lamp for 5 min before filling all fluid reservoirs with another 75 μL of the 1 mM sodium tetraborate buffer. This additional buffer was pipetted into the reservoirs to prevent them from drying out during their subsequent exposure to the UV lamp. The microchip device was now again exposed to the UV lamp for another 15 min to complete the photopolymerization process.

2.3 Device operation

Following the membrane fabrication step, all microchannels were thoroughly rinsed with deionized water or the phosphate buffer to be used in the experiment to condition the microchip. This was accomplished by first filling all reservoirs with the relevant liquid (deionized water or phosphate buffer) before applying vacuum to reservoir 5 for 10 min. Although this approach allowed the rinsing of most of our microfluidic network, no measurable PDF was observed in segment C upon vacuum application at reservoir 5 due to the presence of the gel membrane at reservoir 3. In this situation, a voltage drop of 500 V was applied across reservoirs 2 (high voltage end) and 3 to enable electrokinetic rinsing of segment C in addition to washing out any excess unpolymerized gel precursor material from the system. After completion of the rinsing procedures, the sample solution was placed in reservoir 1 while filling the remaining reservoirs with the corresponding background electrolyte (deionized water/phosphate buffer). A sample flow profile suitable for gated band injection [20, 21] was then created by applying high voltages to reservoirs 1 and 2 (0.5 and 2 kV, respectively) and electrically grounding reservoirs 3 and 4 (see Fig. 1, stage A). Reservoir 5 was electrically floated in all our experiments by not placing an electrode in it to realize an electric field-free condition in the separation channel (segment F). In this situation, any liquid/sample flow in this segment was entirely pressure driven irrespective of

the voltages applied at the different channel terminals. The gated injection scheme was implemented by electrically grounding reservoir 2 for 1 s during which the sample was allowed to fill up the entire injection cross (see Fig. 1, stage B). Following this step, the electric potentials at reservoirs 1 and 2 were switched back to the operation voltages (ϕ and 4ϕ , respectively) to be used in the experiment, which also forced a fraction of the sample that had filled up the injection cross to be swept into segment F (see Fig. 1, stage C). Notice that the voltage applied at terminal 2 was always chosen to be 4 times the voltage applied at terminal 1 in all stages of our microchip operation, as it yielded a clean injection of the analyte band into the analysis channel. Also, while these voltages were always set at 2 kV (reservoir 2) and 0.5 kV (reservoir 1) during stage A of microchip operation in order to minimize variations in the injected sample volume, their magnitudes were varied in the ratio 4:1 in stage C to manipulate the migration velocity of the sample band injected into segment F. The analyte bands injected into segment F were detected 2.5 cm downstream of the injection cross via the laser induced fluorescence technique. For these measurements, a laser beam (125 mW Argon ion tunable laser, Melles Griot) of wavelength 514 nm was used to excite the fluorescent sample and the resulting signal collected using a photomultiplier tube (Hamamatsu Photonics) after passing it through a 585 ± 20 nm bandpass optical filter (Semrock). The sequence of voltages applied to the different channel terminals during the sample injection/separation stages and the data acquisition process by the photomultiplier tube were controlled using a LABVIEW program. The peak shapes were analyzed in our work with numerical data analysis tools available in MATLAB. The visualization of the analyte bands was realized in this study using an epifluorescence microscope system (Nikon) with bandpass excitation (528–543 nm) and emission (590–650 nm) optical filters. The fluorescence images of the sample injection process were recorded using a CCD camera (Roper Scientific) in our experiments.

3 Results and discussion

3.1 Micropump Characterization

Prior to assessing the pressure-generation capability of the micropump, we attempted to visually characterize a functional gel membrane in our device. Unfortunately, this membrane was visually indistinguishable from a buffer solution both in the fluid reservoir and the channel segment C. In this situation, we indirectly established its existence in the fluid reservoir by introducing a colorful solution of 10 μ M Rhodamine B base (prepared in a 1 mM sodium phosphate buffer, pH 7.2) into the cylindrical well at terminal 3 (see Fig. 2A). We later discovered that imaging the gel membrane within channel segment C was even more challenging. We attempted including the Rhodamine dye in the gel precursor material prior to photopolymerization to this end, but this approach failed to provide us with a clear image of the membrane as this structure extended only about a millimeter into the microchannel. This region unfortunately was located around the area where the cover plate had to be powder-blasted to create the access hole making the glass surface in its vicinity highly rough and light-scattering. In this situation, we ended up visualizing the membrane by drying the gel structure within reservoir 3/segment F in an oven for 60 min at 80°C before imaging it under a microscope. We have included an optical image of the dried gel structure in Fig. 2B, which shows its axial extent to be about 1 mm. Our experiments further

showed that if this structure was significantly shorter or longer than the noted length scale (~ 1 mm), the pressure-generation capability of the device was significantly compromised. This observation may be explained arguing that as the penetration length of the gel membrane increases, the electrical resistance of segment C becomes dominated by it reducing the electric field strength, and thereby the electroosmotic flow velocity, generated in the open region of channel segment C. In the opposite scenario, if the gel structure does not penetrate enough into segment C, significant leakage of the PDF can occur into reservoir 3 compromising the pressure gradient generated across segment F.

Following this visualization effort, we proceeded to characterize the pressure-generation ability of our device by measuring the flow speed of the carrier fluid in the separation channel, that is, segment F. Note that this flow was entirely pressure driven in our experiments due to the absence of an electric field in segment F. In order to make these flow measurements, 10 μM samples of a neutral fluorescent dye (Rhodamine B base) were injected into the separation channel using the operating procedure described in the previous section. In Fig. 3A, we have presented data obtained from one such set of measurements made using a microchip that had a channel depth of 5 μm in the entire fluidic network. As may be seen from the figure, the flow speed in segment F was observed to have a linear dependence on the voltage applied at reservoir 1 (ϕ). This result is an expected one as the pressure gradient in our system was generated using electroosmotic flow, which is known to scale linearly with the applied voltage or electric field in the system. Moreover, for a given value of ϕ , the flow velocity in the separation channel was seen to increase with a decrease in the ionic strength of the mobile phase. This again is a consequence of the fact that the electroosmotic mobility of a fluid varies inversely with an increase in the ion concentration in it due to a reduction in the zeta potential at the channel walls [22]. Previous work in our laboratory has shown that the application of an electric field across a membrane structure interfacing with an open microchannel leads to the blockage of electroosmotic flow generated in the open region. This in turn raises the fluid pressure at the channel–membrane interface yielding a pressure gradient over the entire fluidic network. In our device, a fraction of this PDF was guided to an electric field-free analysis channel (segment F) in order to realize the desired pressure-driven assay, that is, ion chromatography. As may be seen from Fig. 3, we were able to generate pressure-driven velocities up to 3.9 mm/s in segment F of our device using deionized water as the carrier fluid upon application of 3 kV across reservoirs 2 and 3. Notice that this flow velocity is about 26% higher than that reported by our group previously for a similar micropump employing a shallow channel segment instead of a gel membrane [15]. It must be also noted that for the current device, the pressure-driven velocity in segment F reduced by less a factor of 2 going from deionized water to a 10 mM sodium phosphate buffer as the carrier fluid. This reduction compares well with the ratio of the electroosmotic mobility for the two carrier fluids suggesting that the fraction of electroosmotic flow blocked by our membrane structure to be relatively insensitive to the ionic strength of the mobile phase. On the other hand, our previous work has shown that the pressure gradient generated by a micropump involving a shallow channel segment to be significantly more sensitive to the ionic strength of the carrier fluid usually reducing the pressure-driven velocity in segment F by over a factor of 4 going from deionized water to a 10 mM sodium phosphate buffer [15]. This improvement in

performance in our current device likely occurs due to a nonreliance of the pressure gradients developed in it on the extent of Debye layer overlap in the channel network, which is not the case for the micropump design involving a shallow channel segment. With our current system, the monomer concentration in the gel precursor solution nevertheless had a strong influence on its pressure-generation capability. In particular, for acrylamide/bisacrylamide concentrations below 8% (v/v), no measurable PDF was noticed in segment F. However, this velocity increased sharply upon raising the acrylamide/bisacrylamide concentration from 8 to 20% and then leveled off gradually at higher gel precursor concentrations (see Fig. 3B). This observation is consistent with an expected decrease in pore size within the gel structure upon increasing the acrylamide/bisacrylamide concentration in the precursor solution, which allows greater blockage of the electroosmotic flow generated in the open channel region interfacing the membrane. Based on this result, we carried out all experiments here onwards with gel membranes prepared using a 50% precursor solution. The authors would like to point out that the pH of the background electrolyte decreased from 7.2 to 7.05 going from a 10 mM sodium phosphate buffer to deionized water in the current work. Although it is known that such changes in electrolyte pH alter the zeta potential at the channel walls and thereby the electroosmotic mobility in the system [23], we anticipate the small reduction in electrolyte pH measured in our experiments to lead to <1% decrease in the electroosmotic mobility of the liquid, and therefore to contribute insignificantly to any trends seen in the experimental data.

A significant advantage to generating PDF via electroosmotic means is the straightforward miniaturization of such micropumps down to the submicrometer length scale. This is because electroosmotic transport rates being relatively insensitive to the lateral dimension of a channel lead to pressure-driven velocities that weakly depend on this geometric parameter as well [12,23]. To highlight this aspect of our device, we investigated the effect of the depth in our channel network (d) on the PDF velocity generated in segment F. In Fig. 4A, data obtained from such a study have been included that show the hydrodynamic flow speed in our separation channel to be reduced by only about 10% going from a 5 μ m-deep fluidic network to a 0.3 μ m-deep one. This reduction likely occurs due to a lower electroosmotic flow in the shallower fluidic network resulting from an overlap of the electrical Debye layers around adjacent channel walls [24, 25]. It must be noted, however, that because the pressure drop across segment F varies inversely with the square of the channel depth (d) for a fixed hydrodynamic flow speed (U), this quantity is estimated to increase by over 2 orders of magnitude as the channel depth is reduced from 5 mm to 300 nm. These estimates have been included in Fig. 4B and were obtained based on the flow equation governing the transport of a Newtonian fluid through a parallel-plate device, that is, $P = 12U\eta L/d^2$, where η denotes the viscosity of the carrier fluid in the system and L represents the length of the separation channel. For a maximum pressure-driven velocity of 3.6 mm/s realized in a separation duct of depth 300 nm, the noted equation predicts a pressure drop of about 16.8 MPa across the 3.5-cm-long separation channel for a water-based electrolyte of viscosity of 1 cp.

3.2 Nanofluidic ion chromatography

The functionality of the integrated microchip device described above was demonstrated in our study by performing an ion-chromatographic separation of two 5-TAMRA, SE-labeled

amino acid species. For this separation, the 300-nm-deep microchip was chosen as it yielded the largest differential in transport rate for the two amino acid species under consideration as well as one of the lowest plate heights for the sample peaks as was determined in one of our previous works [15]. In Fig. 5A, we have included an ion chromatogram (the top one) from this study, which shows a clean separation between the fluorescently labeled arginine and glutamic acid peaks under PDF conditions. The separation resolution in these experiments was seen to increase with the operating voltage reaching values as large as 2.6 for $\phi = 0.75$ kV. Also, the use of 1 mM sodium phosphate buffer as the background electrolyte was observed to yield better resolutions than upon employing deionized water as the carrier fluid. This result is consistent with the larger plate heights measured for water-based assays as seen in our previous work [15]. Interestingly, the separation between the two amino acid species could not be accomplished in our device upon driving the mobile phase via electrokinetic means as may be seen from the bottom chromatogram in Fig. 5A. The electrokinetic separation was realized by electrically floating reservoir 3, that is, not placing an electrode in this reservoir, which prevented any pressure generation in the system while electrically grounding reservoir 5 to realize an electric field in segment F. Moreover, during stage A of microchip operation in this analysis, the voltages applied to reservoirs 1 and 2 were modified to 0.5 and 1.25 kV (from 0.5 and 2 kV), respectively, in order to accomplish a clean injection of the analyte band into segment F, which were then switched to 0.75 kV (reservoir 1) and 1.875 kV (reservoir 2) during stage C to drive the chromatographic assay. The lower resolution (~ 0.75) realized in the above experiment was unsurprising as electrophoresis counteracted the ion-chromatographic separation process in this case. Moreover, because a 1 mM sodium phosphate buffer was used in this study, we expected the electroosmotic flow profile in our electrokinetically driven separation to be relatively flat compared to the parabolic PDF distribution further deteriorating the desired separation. The authors would like to clarify that while the experimental data included in Figs. 1 and 3B, 4A, 4B, and 5A were all obtained for $\phi = 0.75$ kV, similar trends as seen in these figures have been also noticed for other choices of ϕ . However, because the pressure-driven velocity and separation resolution in our system were observed to increase for larger values of ϕ , we opted to mostly include experiments performed at $\phi = 0.75$ kV in our figures, which corresponded to the maximum voltage applicable in our setup.

4 Concluding remarks

To summarize, we have successfully demonstrated on-chip pressure generation via electrokinetic means employing a polyacrylamide-based gel membrane fabricated within a fluid reservoir located at the terminating end of a microchannel. This approach to membrane fabrication not only reduced the fabrication effort in preparing the device, but was also seen to considerably improve the mechanical stability of the gel under the influence of a large electric field. Our experiments further showed that the noted design yields a higher PDF velocity for a given applied electrical voltage, and is relatively insensitive to the ion concentration in the background electrolyte compared to other similar micropumps described in the literature. The functionality of our micropump was further demonstrated by integrating it to an electric field-free separation channel to drive a nanofluidic ion-chromatographic analysis based on the electrostatic interaction of analyte molecules with the

channel surface charges. Separation resolutions as large as 2.6 were realized using our current design for a separation length and time of 2.5 cm and 15 s, respectively, which is significantly more impressive than the results reported by us previously for similar assays [15].

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Abbreviation:

PDF pressure-driven flows

5 References

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Figure 1.

Schematic of the operating procedure for performing a pressure-driven assay on our microchip device. In stage A, appropriate electric voltages are being applied at the channel terminals to set up a flow profile that is suitable for gated injection of sample plugs into segment F. In stage B, the electric voltage at reservoir 2 is grounded allowing the sample to fill up the injection cross as well as parts of segment D and F. In stage C, the electric voltage at reservoir 2 is elevated back to sweep a part of the sample that occupied the injection cross and segments D/F into segment F where it is transported via pressure drive. Notice that while the voltages applied at reservoirs 1 and 2 were always set at 0.5 and 2 kV, respectively, during stage A of microchip operation, their magnitudes were varied in the ratio 1:4, that is, as ϕ and 4ϕ , in stage C to manipulate the migration velocity of the sample band injected into segment F. The fluorescence images presented above were obtained using a 10 μ M solution of Rhodamine B sample prepared in a 1 mM sodium phosphate buffer (pH 7.2). A microchip with a 3- μ m-deep fluidic network was employed in this experiment and operated at a voltage of $\phi = 750$ V. The membrane structure photopolymerized in reservoir 3 for this experiment was prepared using a 50% gel precursor solution.

Figure 2.

An optical image of the membrane in (A) fluid reservoir 3 and (B) channel segment F prepared using a 50% gel precursor solution. The image in subfigure (b) was obtained for a 5- μm -deep fluidic network after drying the membrane in an oven for 1 h at 80°C. Notice the parallel thread-like structure in the gel region of the microchannel in this image, which was found to be a characteristic of membranes that blocked electroosmotic flow very effectively. We would also like to point out that this structure was only seen in the microchannel area and not within the circular region in the image that corresponds to an access hole.

Figure 3.

(A) Effect of operating voltage on the PDF speed in the separation channel. The microchip used in this experiment had a $5\mu\text{m}$ -deep channel network with the polyacrylamide membrane prepared in reservoir 3 using a 50% gel precursor solution. (B) Observed variation in the pressure-driven velocity in the separation channel with the gel concentration in terminal 3. For these experiments, a $5\mu\text{m}$ -deep fluidic network and an operating voltage of $\phi = 750\text{ V}$ was chosen. The error bars included in the figures are based on five measurements.

Figure 4.

(A) Variation in the PDF speed in segment F with the depth of the fluidic network when operated at $\phi = 750$ V. The membrane in reservoir 3 was prepared using a 50% gel precursor solution for these experiments. (B) Estimated pressure drop across the separation channel for the measurements shown in subfigure (A). The error bars included in the figures are based on five measurements.

Figure 5.

(A) Nanofluidic ion chromatography of 5-TAMRA, SE-labeled arginine, and glutamic acid samples under pressure- and electrokinetically driven conditions in a 300-nm-deep channel performed using a 1 mM sodium phosphate buffer (pH 7.2) as the background electrolyte.

(B) Effect of operating voltage on the separation resolution of the two fluorescently labeled amino acid samples considered here. The reported experiments were performed in a 300-nm-deep channel with the polyacrylamide membrane prepared in reservoir 3 using a 50% gel precursor solution. The error bars included in the figure are based on five measurements.