Microfluidic rectifier based on poly(dimethylsiloxane) membrane and its application to a micropump

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A microfluidic rectifier incorporating an obstructed microchannel and a PDMS membrane is proposed. During forward flow, the membrane deflects in the upward direction; thereby allowing the fluid to pass over the obstacle. Conversely, during reverse flow, the membrane seals against the obstacle, thereby closing the channel and preventing flow. It is shown that the proposed device can operate over a wide pressure range by increasing or decreasing the membrane thickness as required. A microfluidic pump is realized by integrating the rectifier with a simple stepper motor mechanism. The experimental results show that the pump can achieve a vertical lift height of more than 2 m. Moreover, it is shown that a maximum flow rate of 6.3 ml/min can be obtained given a membrane thickness of 200 μm and a motor velocity of 80 rpm. In other words, the proposed microfluidic rectifier not only provides an effective means of preventing reverse flow but also permits the realization of a highly efficient microfluidic pump. © 2013 AIP Publishing LLC. [<http://dx.doi.org/10.1063/1.4818905>]

I. INTRODUCTION

Microfluidic and Lab-on-Chip (LoC) systems are widely used for biochemical and biomedical applications such as food safety inspection and quality control, environmental monitoring, and drug discovery.^{1–7} Microfluidic devices are capable of performing many different functions, including sample injection and separation, cell sorting and counting, polymerase chain reaction (PCR), species mixing, micro pumping, and microfluidic rectification.^{8–13} Compared to their large-scale counterparts, miniaturized systems have many important advantages, including a reduced sample and reagent consumption, a more rapid response, an enhanced efficiency, a lower power consumption, an improved sensitivity, lower fabrication and operating costs, greater portability, and the potential for integration with other miniaturized devices.

The performance of microfluidic devices depends on a highly precise manipulation and control of tiny volumes of sample and reagent. Moreover, in applications such as micropumping and drug delivery, it is essential that the fluid flows in one direction only. Thus, the literature contains many proposals for microfluidic rectification devices. Broadly speaking, the proposed devices can be categorized as passive microvalves, active microvalves, or valveless microvalves.^{14–20} Park *et al.*²¹ presented a microfluidic rectifier incorporating a PDMS membrane and an *in-situ* polymerized seal. During the fabrication process, photocurable oligomer was introduced into the fluidic channel and a gas pressure was then applied in order to deform the membrane. The polymer seal was then locally polymerized using a photolithography technique in order to produce a structure with the same curvature as that of the deformed

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membrane. Singh and Kumar²² performed computational fluid dynamics (CFD) simulations to investigate the effects of the flap parameters on the fluid rectification performance of a microfluidic diode. The flap opens during forward flow and seals against a stopper during reverse flow. This allows flow in the forward direction and prevents it in the reverse direction. Li *et al.*²³ presented a finger-squeeze-based rectification device comprising a cascaded arrangement of check-valves and balance tunnels between two layers of microchannels. Tsai *et al.*²⁴ proposed a microfluidic rectifier consisting of a microchannel and a sudden expansion channel containing an embedded block structure. It was shown that the block induced two vortex structures in the microchannel immediately upstream of the expansion channel entrance; thereby impeding backflow. The experimental results showed that the rectifier achieved a maximum diodicity of $Di = 1.54$.

Microfluidic pumps can be categorized as either displacement micropumps or dynamic micropumps, depending on the manner in which the fluid is displaced.^{25–34} In displacement pumps, a pumping force is generated periodically by applying a force to one or more movable chamber walls. By contrast, in dynamic microfluidic pumps, a continuous energy source is used to exert a steady driving force on the fluid. Nabavi and Mongeau³⁵ performed a numerical investigation into the effects of the divergence angle and maximum pressure amplitude on the characteristics of high-frequency (10–30 kHz) pulsating flows through a micro-diffuser. Wang *et al.*³⁶ presented a valveless micropump incorporating a nozzle-shaped actuation chamber, which functioned as both a pumping chamber and a flow rectification structure. It was shown that the pump developed a high backpressure and flow rate at frequencies in the range of 20–100 Hz. Ezkerra *et al.*³⁷ proposed a micropump comprising a deflectable diaphragm and a set of out-of-plane cantilever check valves. The experimental results showed that the micropump achieved a maximum flow rate of 177 $\mu\text{L}/\text{min}$ given an effective area of 10 mm², an actuating frequency of 6 Hz and a driving pressure of 200 kPa. Ni *et al.*³⁸ presented a pneumatic PDMS-based micropump consisting of two in-plane low-leakage check valves. It was shown that a maximum flow rate of 34 $\mu\text{L}/\text{min}$ could be obtained by applying a driving frequency of 3 Hz and a driving pressure of 25 kPa.

Many on-chip pumping mechanisms for point-of-care applications have been proposed in recent years.^{39–44} Gervais and Delamarche⁴⁵ presented an integrated device for performing one-step immunoassays, in which the blood serum sample was drawn continuously from an open reservoir by means of a capillary pump placed at the opposite end of the device. Dimov *et al.*⁴⁶ proposed a microfluidic device in which vacuum pressure was used to deliver a few microliters of whole blood from the inlet ports to the downstream analysis region. Li *et al.*⁴⁷ developed a “place n play” modular pump consisting of a degassed particle desorption PDMS slab with an integrated mesh-shaped chamber designed not only to absorb the air in the microfluidic system, but also to create a negative pressure to drive the fluid. The results showed that the performance of the degassed PDMS pump was strongly dependent on the surface area of the pumping chamber, the exposure area, and the volume of the PDMS pump slab. Wu *et al.*⁴⁸ developed a portable plastic pump for on-chip polymerase chain reaction (PCR) comprising an air-filled syringe connected to a continuous-flow PCR component via highly gas-permeable tubing. In the proposed device, the sample plug was driven by means of the pressure gradient created by the compressed air in the syringe and the air escaping from the tubing.

This study proposes a high-performance microfluidic rectifier comprising an obstructed microchannel and a thin PDMS membrane. The PDMS membrane deflects in the upward direction during forward flow; thereby permitting the fluid to flow over the obstacle. However, during reverse flow, the membrane seals against the obstacle; thereby closing the channel and preventing fluid flow. The relationship between the membrane deformation and the operating pressure is investigated as a function of the membrane thickness. In addition, the rectification performance of the proposed device is compared with that of a device with equivalent geometric parameters but no obstacle within the microchannel. Finally, the performance of a micropump consisting of a simple stepper motor mechanism and the proposed microfluidic rectifier are examined for different PDMS membrane thicknesses and stepper motor speeds.

FIG. 1. (a) Photograph of microfluidic rectifier and (b) SEM image of obstruction within microchannel of rectifier.

II. FABRICATION AND EXPERIMENTAL DETAILS

Figure 1(a) presents a photograph of the microfluidic rectifier developed in the present study. The device has a sandwich structure comprising a rectangular PMMA upper substrate containing a circular rectifier chamber, a thin PDMS middle layer, and a PMMA lower substrate containing an obstructed microchannel. The fluid flow through the rectifier is controlled by two hand-operated syringe pumps located at either end of the microchannel. The microchannel has dimensions of $200\ \mu\text{m} \times 150\ \mu\text{m}$ (width $\times$ depth) and contains an obstructive block with a length of $300\ \mu\text{m}$ at the midpoint position (see Fig. 1(b)).

Figure 2 illustrates the major steps in the fabrication process used to realize the proposed microfluidic rectifier. As shown, the process commenced by placing two steel wires with a diameter of $200\ \mu\text{m}$ on the surface of a PMMA substrate with dimensions of $30\ \text{mm} \times 10\ \text{mm} \times 2\ \text{mm}$ (Fig. 2(a)). Note that to ensure a precise alignment of the two steel wires, two notches were first ablated on the PMMA substrate using a CO_2 laser (V-12, Laser Pro Venus, Taiwan) with a power of 3 W. The steel wires were then impressed into the PMMA substrate by means of a hot press procedure.⁴⁹ In the hot press process, the PMMA substrate was sandwiched between two glass plates and heated to a temperature of 90°C . The pressure was then increased to $10\ \text{kg}/\text{cm}^2$ for 5 min while maintaining the temperature at 90°C (Fig. 2(b)). The two wires were then stripped from the PMMA substrate to create the obstructed microchannel (Fig. 2(c)). To form the PDMS membrane, Sylgard 184 (Dow Corning, MI, USA) and its curing agent were mixed in a weight ratio of 10:1 and were degassed in a vacuum oven for 1 hr. The PDMS mixture was spin coated on a glass slide at a speed of 750 rpm

FIG. 2. Schematic overview of fabrication process used to realize microfluidic rectifier.

FIG. 3. Working principle of rectifier.

(Chemat KW-4 A, USA) and was then cured at a temperature of 75 °C for 3 h. The upper PMMA substrate was drilled with via holes using a CO₂ laser (V-12, Laser Pro Venus, Taiwan) with a power of 12 W. The inlet and outlet ports each had a diameter of 3 mm, while the microfluidic rectifier chamber (located in the center of the substrate) had a diameter of 5 mm (Fig. 2(d)). The PDMS membrane was stripped from the glass slide and inserted between the upper and lower PMMA substrates. The three-layer sandwich structure was then sealed by means of four screws (Fig. 2(e)). Finally, the two hand-operated syringe pumps were screwed into the inlet and outlet ports, respectively.

Figure 3 illustrates the working principle of the proposed device. As shown in Fig. 3(a), during forward flow, the PDMS membrane deflects in the upward direction; thereby allowing the fluid to flow over the obstacle and enter the downstream microchannel. Conversely, during reverse flow, the membrane deflects in the downward direction; thereby sealing against the obstacle and closing the channel (see Fig. 3(b)).

FIG. 4. Schematic illustration of experimental setup.

Figure 4 shows the experimental setup used to characterize the microfluidic rectifier. During the experiments, sample fluid (DI water containing green die) was injected continuously into the rectifier using a commercial syringe pump (KDS 200 series, KD scientific, USA). The fluid pressure within the microchannel was measured using a pressure meter (DPI 103, GE, USA) connected to the inlet side of the device via a T-junction connector.

III. RESULTS AND DISCUSSION

A. Microfluidic rectifier

The deflection (δ) of the PDMS membrane in the proposed rectifier varies as a function of the membrane and rectifier chamber geometry, the membrane properties, and the applied fluid pressure, i.e.,⁵⁰

$$\delta = \frac{3}{16} \frac{r^4}{h^3} \frac{1 - \nu^2}{E} p, \quad (1)$$

where h is the membrane thickness, r is the radius of the rectifier chamber, E is the elastic modulus of the PDMS membrane, ν is the Poisson ratio of the PDMS membrane, and p is the applied fluid pressure. In the present study, the rectifier chamber had a radius of 5 mm; and the elastic modulus and Poisson's ratio of the PDMS membrane were equal to 6×10^5 Pa and 0.48, respectively. Moreover, three different rectifiers were fabricated incorporating PDMS membranes with thicknesses of 200 μm , 400 μm , and 800 μm , respectively.

Figure 5 shows the experimental and theoretical results for the variation of the membrane displacement with the fluid pressure in each of the three rectifiers. It is seen that for a given membrane thickness, the membrane deflection increases approximately linearly with an increasing pressure. Moreover, a good qualitative agreement is observed between the experimental results and the theoretical results. It is observed that for a membrane thickness of 200 μm or 400 μm , a membrane displacement of 4 mm can be achieved using a pressure of less than 50 kPa. Meanwhile, for a membrane thickness of 800 μm , a displacement of 3 mm can be obtained by applying a driving pressure of around 200 kPa. In general, the results presented in Fig. 5 show that the proposed rectifier device can operate over a wide pressure range by simply increasing or decreasing the membrane thickness as required.

FIG. 5. Variation of membrane deformation with operating pressure given different membrane thicknesses.

FIG. 6. Experimental images of fluid column height in microfluidic rectifier during (a)–(c) forward flow and (d)–(f) reverse flow.

Figure 6 presents a sequence of experimental images showing the fluid flow within the rectifier during the forward flow and reverse flow stages. As shown in Figs. 6(a)–6(c), during the forward flow stage, the membrane deflects away from the obstacle, and thus the fluid level in Syringe 1 (attached to the inlet) reduces while that in Syringe 2 (attached to the outlet) rises. During the reverse flow stage (Figs. 6(d)–6(f)), the suction force produced by raising the plunger in Syringe 1 causes the membrane to seal against the obstacle in the microchannel. Consequently, reverse flow through the microchannel is prevented, and thus the fluid level within Syringe 2 remains unchanged. In other words, the rectification ability of the proposed microfluidic device is confirmed.

For comparison purposes, a second rectifier was fabricated with the same dimensions and structure as that described above but with an unobstructed microchannel (see Fig. 7). Figure 8 presents a series of experimental images showing the flow of the sample fluid through the

FIG. 7. Photograph of microfluidic device with unobstructed microchannel.

FIG. 8. Experimental images of fluid column height in microfluidic rectifier with unobstructed microchannel during (a)–(c) forward flow and (d)–(f) reverse flow.

device during the forward and reverse flow stages. During the forward flow stage (Figs. 8(a)–8(c)), the membrane deflects in the upward direction; allowing the sample to flow through the microchannel and into Syringe 2. Consequently, the fluid in Syringe 2 increases from an initial height of around 3 cm to a final height of around 4.6 cm. During the reverse flow stage, the PDMS deflects toward the base of the microchannel. However, the channel is not sealed completely, and thus the fluid flows back through the microchannel and enters the inlet port. As a result, the fluid in Syringe 2 reduces from an initial height of approximately 4.2 cm to a final height of around 3 cm. Comparing the results presented in Figs. 6 and 8, respectively, it is clear that the proposed microfluidic rectifier incorporating an obstructed microchannel provides an improved flow rectification performance

B. Micropump

In the present study, the microfluidic rectifier was integrated with a simple stepper motor mechanism (K27, AVIOSYS, Taiwan) in order to realize a microfluidic pump (see Fig. 9). Note that the performance of the micropump is independent of the head pressure since the PDMS membrane effectively prevents reverse flow during the suction stage. Figure 10 presents the experimental results obtained for the vertical lift capacity of the micropump given a stepper motor velocity of 40 rpm. The results show that the micropump has a vertical lift capacity of at least 2 m. Once a vertical lift height of 2 m had been attained, the stepping motor was turned off, and the experimental apparatus was allowed to stand for 24 h. The sample height was

FIG. 9. (a) Photograph of microfluidic pump system and (b) close-up view of micropump.

FIG. 10. Experimental images showing vertical lift performance of micropump.

FIG. 11. Variation of flow rate with stepper motor velocity given different membrane thicknesses.

found to be unchanged after the 24-h observation period. Thus, the effectiveness of the rectifier device was confirmed.

Figure 11 shows the variation of the flow rate developed by the micropump with the stepper motor velocity given membrane thicknesses of 200 μm and 800 μm , respectively. It is seen that for both membranes, the flow rate increases linearly with an increasing motor velocity. Furthermore, for a given motor velocity, the flow rate decreases with an increasing membrane thickness due to the greater pressure required to deflect the membrane in the upward direction. From inspection, the micropump incorporating a rectifier with a 200- μm -thick membrane achieves a maximum flow rate of 6.3 ml/min given a motor velocity of 80 rpm.

IV. CONCLUSIONS

This study has presented a high-performance, low-cost PMMA-based microfluidic rectifier comprising a PDMS membrane and an obstructed microchannel. The rectifier is easily fabricated using conventional hot embossing and laser ablation techniques and has a planar structure; thereby facilitating its simple integration with other microfluidic devices. In the proposed device, the application of a driving pressure to the inlet port causes the membrane to deflect in the upward direction; thereby allowing the sample to flow over the obstacle and through the microchannel. By contrast, the application of a suction pressure causes the membrane to seal against the obstacle; thereby sealing the channel and preventing reverse flow. The experimental results have shown that the proposed rectifier can be adapted to various working pressure ranges by increasing or decreasing the membrane thickness as required. A microfluidic pumping system has been realized by integrating the proposed rectifier with a simple stepper motor mechanism. It has been shown that the micropump can attain a vertical lift of at least 2 m given a stepper motor rotation speed of 40 rpm. Finally, the results have shown that the pump can achieve a flow rate of 6.3 ml/min given a PDMS membrane thickness of 200 μm and a motor velocity of 80 rpm. Overall, the results presented in this study indicate that the proposed microfluidic rectifier and micropump represent ideal solutions for microfluidic applications.

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- ¹X. Weng, C. H. Chon, H. Jiang, and D. Li, *Food Chem.* **114**, 1079 (2009).
- ²N. V. Panini, E. Salinas, G. A. Messina, and J. Raba, *Food Chem.* **125**, 791 (2011).
- ³H. H. Hou, Y. N. Wang, C. L. Chang, L. M. Fu, and R. J. Yang, *Microfluid. Nanofluid.* **11**, 479 (2011).
- ⁴W. J. Ju, L. M. Fu, R. J. Yang, and C. L. Lee, *Lab Chip* **12**, 622 (2012).
- ⁵C. H. Lin, W. N. Wang, and L. M. Fu, *Biomicrofluidics* **6**, 012818 (2012).
- ⁶Y. N. Wang, R. J. Yang, W. J. Ju, M. C. Wu, and L. M. Fu, *Biomicrofluidics* **6**, 034111 (2012).
- ⁷L. M. Fu, W. J. Ju, Y. N. Wang, and R. J. Yang, *Microfluid. Nanofluid.* **14**, 479 (2013).
- ⁸M. M. Gong, B. D. MacDonald, T. V. Nguyen, and D. Sinton, *Biomicrofluidics* **6**, 044102 (2012).
- ⁹F. Yang, Z. Chen, J. Pan, X. Li, J. Feng, and H. Yang, *Biomicrofluidics* **5**, 024115 (2011).
- ¹⁰H. C. Lee, H. H. Hou, R. J. Yang, C. H. Lin, and L. M. Fu, *Microfluid. Nanofluid.* **11**, 469 (2011).
- ¹¹D. Puchberger-Enengl, S. Podszun, H. Heinz, C. Hermann, P. Vulto, and G. A. Urban, *Biomicrofluidics* **5**, 044111 (2011).
- ¹²K. E. Jensen, P. Szabo, F. Okkels, and M. A. Alves, *Biomicrofluidics* **6**, 044112 (2012).
- ¹³L. M. Fu, W. J. Ju, C. H. Tsia, H. H. Hou, Y. N. Wang, and R. J. Yang, *Chem. Eng. J.* **214**, 1 (2013).
- ¹⁴N. T. Nguyen, Y. C. Lam, S. S. Ho, and C. L. N. Low, *Biomicrofluidics* **2**, 034101 (2008).
- ¹⁵J. Liu, Y. F. Yap, and N. T. Nguyen, *Phys. Rev. E* **80**, 046319 (2009).
- ¹⁶C. Y. Lee, J. C. Leong, Y. N. Wang, L. M. Fu, and S. J. Chen, *Jpn. J. Appl. Phys., Part 1* **51**, 047201 (2012).
- ¹⁷K. Yang, I. Chen, C. Wang, and J. Shyu, *Microsyst. Technol.* **16**, 1691 (2010).
- ¹⁸P. C. Sousa, F. T. Pinho, M. S. N. Oliveira, and M. A. Alves, *J. Non-Newtonian Fluid Mech.* **165**, 652 (2010).
- ¹⁹C. H. Tsai, C. P. Yeh, C. H. Lin, R. J. Yang, and L. M. Fu, *Microfluid. Nanofluid.* **12**, 213 (2012).
- ²⁰E. Biffi, F. Piraino, A. Pedrocchi, G. B. Fiore, G. Ferrigno, A. Redaelli, A. Menegon, and M. Rasponi, *Biomicrofluidics* **6**, 024106 (2012).
- ²¹W. Park, S. Han, and S. Kwon, *Lab Chip* **10**, 2814 (2010).
- ²²K. P. Singh and M. Kumar, *Biomicrofluidics* **4**, 034112 (2010).
- ²³W. Li, T. Chen, Z. Chen, P. Fei, Z. Yu, Y. Pang, and Y. Huang, *Lab Chip* **12**, 1587 (2012).
- ²⁴C. H. Tsai, C. H. Lin, L. M. Fu, and H. C. Chen, *Biomicrofluidics* **6**, 024108 (2012).
- ²⁵J. R. Du and H. H. Wei, *Biomicrofluidics* **4**, 034108 (2010).
- ²⁶J. M. Li, C. Liu, K. P. Zhang, X. Ke, Z. Xu, C. Y. Li, and L. D. Wang, *Microfluid. Nanofluid.* **11**, 717 (2011).
- ²⁷E. M. Melvin, B. R. Moore, K. H. Gilchrist, S. Grego, and O. D. Velev, *Biomicrofluidics* **5**, 034113 (2011).
- ²⁸R. Zhang, C. Dalton, and G. A. Jullien, *Microfluid. Nanofluid.* **10**, 521 (2011).
- ²⁹A. R. Tovar, M. V. Patel, and A. P. Lee, *Microfluid. Nanofluid.* **10**, 1269 (2011).
- ³⁰E. Diller, S. Miyashita, and M. Sitti, *RSC Adv.* **2**, 3850 (2012).
- ³¹A. B. Wang and M. C. Hsieh, *Lab Chip* **12**, 3024 (2012).
- ³²A. Jahanshahi, F. Axisa, and J. Vanfleteren, *Microfluid. Nanofluid.* **12**, 771 (2012).
- ³³J. Wang, C. Wang, C. Lin, H. Lei, and G. Lee, *Microfluid. Nanofluid.* **10**, 531 (2011).
- ³⁴X. Wang, J. A. Hagen, and I. Papautsky, *Biomicrofluidics* **7**, 014107 (2013).
- ³⁵M. Nabavi and L. Mongeau, *Microfluid. Nanofluid.* **7**, 669 (2009).
- ³⁶S. S. Wang, X. Y. Huang, and C. Yang, *Microfluid. Nanofluid.* **8**, 549 (2010).
- ³⁷A. Ezkerra, L. J. Fernández, K. Mayora, and J. M. Ruano-López, *Lab Chip* **11**, 3320 (2011).
- ³⁸J. Ni, B. Li, and J. Yang, *Microelectron. Eng.* **99**, 28 (2012).
- ³⁹M. Moscovici, W. Chien, M. Abdelgawad, and Y. Sun, *Biomicrofluidics* **4**, 046501 (2010).
- ⁴⁰G. H. Kwon, G. S. Jeong, J. Y. Park, J. H. Moon, and S. H. Lee, *Lab Chip* **11**, 2910 (2011).
- ⁴¹A. R. Abate and D. A. Weitz, *Biomicrofluidics* **5**, 014107 (2011).
- ⁴²Y. Q. Fu, L. Garcia-Gancedo, H. F. Pang, S. Porro, Y. W. Gu, J. K. Luo, X. T. Zu, F. Placido, J. I. B. Wilson, A. J. Flewitt, and W. I. Milne, *Biomicrofluidics* **6**, 024105 (2012).
- ⁴³Y. H. Chang, C. J. Huang, and G. B. Lee, *Microfluid. Nanofluid.* **12**, 85 (2012).
- ⁴⁴H. Zhang, G. Li, Lingying Liao, H. Mao, Q. Jin, and J. Zhao, *Biomicrofluidics* **7**, 034105 (2013).
- ⁴⁵L. Gervais and E. Delamarche, *Lab Chip* **9**, 3330 (2009).
- ⁴⁶I. K. Dimov, L. Basabe-Desmonts, J. Garcia-Cordero, B. M. Ross, A. J. Ricco, and L. P. Lee, *Lab Chip* **11**, 845 (2011).
- ⁴⁷G. Li, Y. Luo, Q. Chen, L. Liao, and J. Zhao, *Biomicrofluidics* **6**, 014118 (2012).
- ⁴⁸W. Wu, T. L. Kieu, and N. Y. Lee, *Analyst* **137**, 983 (2012).
- ⁴⁹T. F. Hong, W. J. Ju, M. C. Wu, C. H. Tai, C. H. Tsai, and L. M. Fu, *Microfluid. Nanofluid.* **9**, 1125 (2010).
- ⁵⁰W. K. Schomburg, *Introduction to Microsystem Design* (Springer-Verlag, Berlin, Heidelberg, 2011), pp. 29–52.