A pneumatically controllable flexible and polymeric microfluidic valve fabricated via *in situ* development

2005 J. Micromech. Microeng. 15 1015

(http://iopscience.iop.org/0960-1317/15/5/017)

INSTITUTE OF PHYSICS PUBLISHING JOURNAL OF MICROMECHANICS AND MICROENGINEERING

J. Micromech. Microeng. **15** (2005) 1015–1020 doi:10.1088/0960-1317/15/5/017

A pneumatically controllable flexible and polymeric microfluidic valve fabricated via *in situ* development

Ju Yeoul Baek[1], Ji Young Park[1], Jong Il Ju[1], Tae Soo Lee[2] and Sang Hoon Lee[1]

[1] Department of BioMedical Engineering, College of Medicine, Dankook University, San 29 Anseo-dong, Cheonan, Chungnam 330-714, Korea
[2] Department of BioMedical Engineering, College of Medicine, Chungbuk National University, 48 Gaesin-dong, Cheongju, Chungbuk 361-763, Korea

E-mail: dbiomed@dankook.ac.kr

Received 17 September 2004, in final form 2 March 2005
Published 30 March 2005
Online at stacks.iop.org/JMM/15/1015

Abstract

In this paper, we have developed a pneumatically controllable polydimethylsiloxane (PDMS)-based microfluidic valve. This valve regulates 'On/Off' of flow using the thick centered membrane. To integrate the freely moving thick centered membrane, regional bonding of the PDMS layer is required. Here, we integrated such a membrane employing a water soluble mask as a regional bonding method, and the mask was washed out by flowing distilled water. The fabricated valve showed good 'On/Off' operations in accordance with the applied pneumatic pressure source. The flow rate could be regulated by the pressure applied to the inlet (regulated by changes in the height of the water column) and the compression-/vacuum-period ratio (this means the ratio of 'On' and 'Off' periods in each cycle) in the range of a few microliters per minute. For the durability test, ten valves were operated simultaneously one million times, and no failed valves were observed.

(Some figures in this article are in colour only in the electronic version)

1. Introduction

Recent progress in microfluidic systems has demonstrated their many potential applications in biology, chemistry and industry, and is making possible the realization of a one-step microscale total analysis system (TAS) [1–3]. In these miniaturized analysis systems, the delivery and the regulation of sample fluids are of great importance. Typical fluidic components for these purposes involve a means to drive fluids (e.g., pumps) and a means to regulate their movement (e.g., valves and nozzles). Several miniaturized devices for fluid handling have been reported, including electrostatic actuation [4], piezoelectric actuation [5], thermopneumatic actuation [6], electromagnetic actuation [7], bimetallic actuation [8], shape memory alloy actuation [9], paraffin phase transition actuation [10], and so forth, and most of them are fabricated using silicon technology. However, the reliability, the time and labor consumption, and the economics of these devices have been major issues. It is, however, true that, in the laboratory, diverse miniaturized devices for the delivery and the regulation of biological samples in a biocompatible environment are required, and a successful realization of such devices in a very simple way would lead to commercial success. Recently, new delivery devices employing different actuating mechanisms have been developed for biomedical applications, and the approaches that have many attractive features center on electrophoretic, electroosmotic and electrochemical delivery [11–13]. But these approaches are sensitive to the physicochemical properties of fluids being pumped—such as ionic strength, pH and ionic composition—and to the presence of charged macromolecules in the fluid; electroosmotic flow is also unable to pump liquids at high flow rates in wider channels.

As an alternative way to overcome these limits, the structure of conventional miniaturized devices is transferred to a flexible polymer so that PDMS-based microscale valves or pumps can be developed [14–17]. The major advantages to this approach are fourfold: (a) the simplicity of fabrication; (b) rapid prototyping makes the optimization of designs rapid; (c) cost-effectiveness; and (d) the biocompatibility of materials, which are broadly applied to cell-culturing devices [18]. In spite of these merits, most PDMS-based valves and pumps are passive systems owing to the difficulties that characterize integration of the actuation part.

Recently, we have developed a microinjection system by embedding in it a flexible and polymeric active microfluidic valve that a user can control with pneumatic force to deliver fluid into a single cell [19]. This valve is fabricated by stacking four PDMS layers onto a thick PDMS substrate. Rather than using a flat membrane, we employed a thick centered (TC) membrane for the rapid and reliable 'On/Off' operation of the valve. But one of the major drawbacks of such a polymeric valve in the fabrication process is that the regional bonding of the PDMS layer for the free motion of the TC membrane is difficult. Previously, we placed the PDMS-based micromasks onto the target area manually, and exposed both of the masked layers to oxygen plasma. After the surface treatment, the masks were removed, and both of the layers were aligned and bonded—except for the masked region—by thermal curing. However, this method requires the micromanipulation of microscale masks, which is a tedious process that consumes time and calls for skilled labor. In addition, during the curing time, the non-exposed region sometimes bonded just by the heating. In this case, it is not easy to separate the non-exposed region, and this difficulty is a serious obstruction in the fabrication of PDMS-based actuators. In this paper, we describe a simpler and more reliable fabrication method for flexible and polymeric active microvalves based on regional bonding via *in situ* development, a method according to which the aqueous solution washes out the soluble mask that is coated on a certain area. The performance and the durability of fabricated valves are evaluated via short- and long-term experiments.

2. The structure, operation and design of a valve having a TC membrane

A schematic view describing the operation of the microvalve is illustrated in figure 1. The valve consists of one bas-relief plate (thickness: 10 mm) and three PDMS layers (thickness: 200 μm), and the cover glass is stacked onto the top layer. Through the pneumatic port, a slight vacuum is applied, the membrane with a thick center is lifted upward, and the center hole (diameter: 100 μm) is opened (figure 1(*b*)). Then the fluid from the inlet port passes the center hole and travels to the outlet. In contrast, if the compressed air enters through the pneumatic port and presses the membrane, the flow path is closed as the TC seals off the center hole (figure 1(*a*)). Usually, a diaphragm with a TC membrane makes an excellent sensing element for strain gage transducers by inducing maximum output from the Wheatstone bridge network. The reason that a diaphragm with a TC is employed here is that a faster 'On/Off' operation can be accomplished, as the gap between the TC membrane and the hole layer is small.

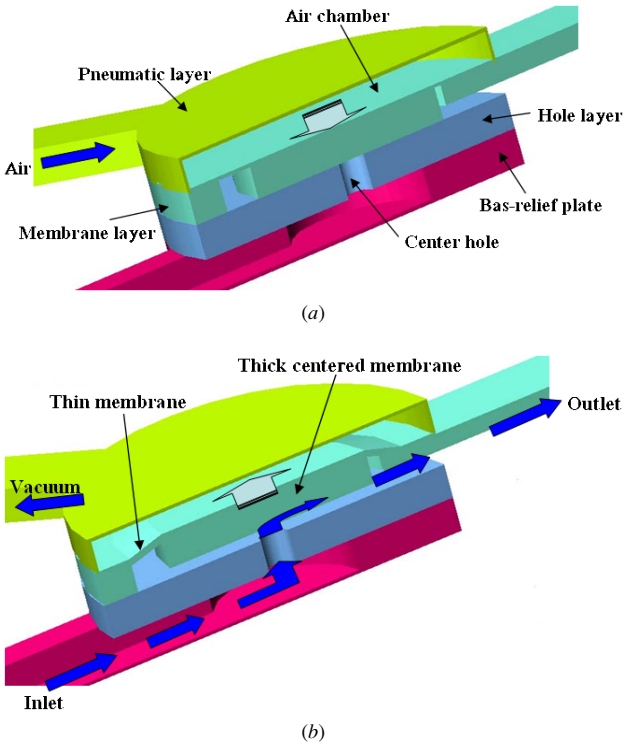


Figure 1. Schematic and operation of the thick centered valve. (*a*) Closing of the valve by compressed air, (*b*) opening of the valve by vacuum.

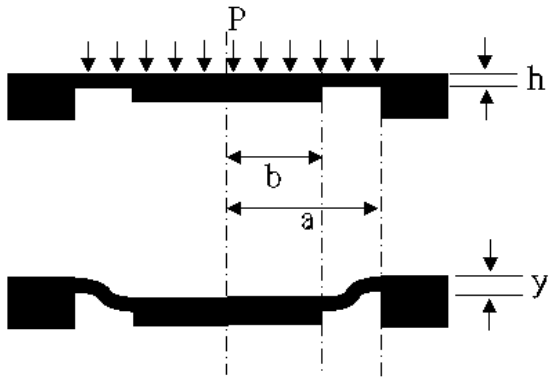


Figure 2. Modeling of the thick centered valve.

The operation of the TC membrane can be modeled following Di Giovanni's book [20], as shown in figure 2. The pressure applied to the TC membrane for any deflection is given by

$$P = \frac{Eh^3}{A_p a^4}(y) + B_p \frac{Eh}{a^4}(y^3) \tag{1}$$

$$A_p = \frac{3(1-\nu^2)}{16}\left(1 - \frac{b^4}{a^4} - 4\frac{b^2}{a^2}\ln\frac{a}{b}\right) \tag{2}$$

and

$$B_p = \frac{\left(\frac{7-\nu}{3}\right)\left(1+\frac{b^2}{a^2}+\frac{b^4}{a^4}\right) + \frac{(3-\nu)^2}{(1+\nu)}\left(\frac{b^2}{a^2}\right)}{(1-\nu)(1-(b^4/a^4))(1-(b^2/a^2))^2} \tag{3}$$

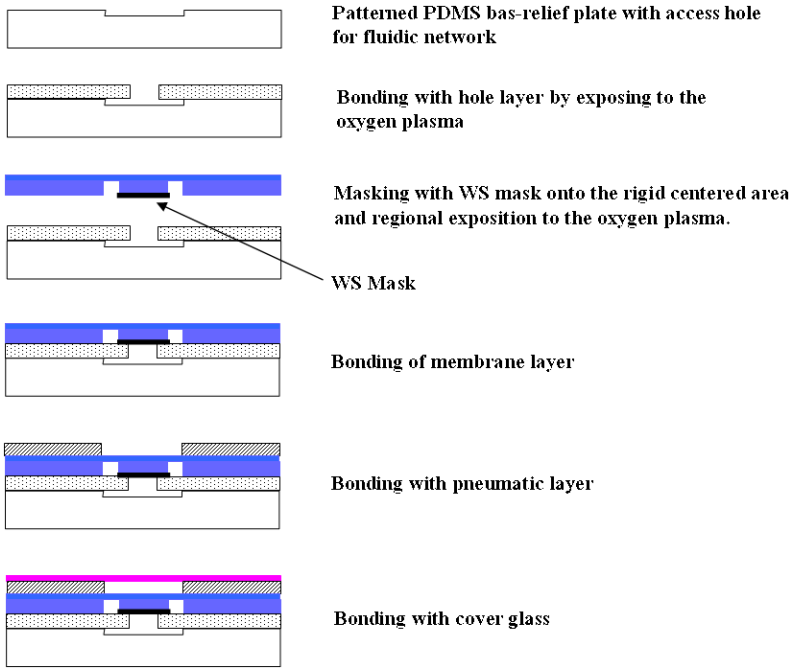


Figure 3. Process of valve fabrication.

where y is the deflection, E is Young's modulus, h is the membrane thickness, ν is Poisson's ratio and a and b are the outer and inner membrane radii.

3. Fabrication and *in situ* development

The patterned bas-relief plate and the layers were constructed by replica molding against a master consisting of photolithographically patterned photoresist (SU-8, MicroChem) on a 3 inch silicon wafer. We fabricated the bas-relief plate by pouring a 10:1 mixture of the PDMS prepolymer and a curing agent (Sylgard 184 silicone elastomer kit, Dow Corning, Midland, MI) onto the master mold placed in a Petri dish. The mixture was thermally cured for 2 h on a hot plate at 80 °C, and separated from the mold. A 12-gauge needle was used so that the holes for the fluidic network could be cored out. The layers were fabricated by compression micromolding of the PDMS elastomer, as previously reported [18]. With this plate and these layers, the microvalve was constructed by an aligning and bonding process, as demonstrated in figure 3. For the bonding, both of the surfaces of each layer were exposed to an oxygen plasma, and aligned with a homemade aligning system by using methanol as a lubricant. Both of the aligned layer's surfaces were bonded by 30 min curing on a hot plate at 80 °C.

For the 'On/Off' motion of the valve, the surface of the TC membrane should be separated from the hole layer, and regional oxygen plasma exposure is required. We developed a new, simple and effective method for regional bonding, and the procedure is illustrated in figure 4(a). First, we modified the surface of the TC membrane by exposing it to the oxygen plasma. Then, only a few nanoliters of the mixture of salt (NaCl) and polyvinylalcohol (PVA) solution (the salt was supersaturated) were dropped onto the surface of the TC membrane. As the surface is hydrophilic, the solution spreads uniformly over the surface. But at the edge of the TC membrane, the expansion is stopped due to the surface tension. So the mixed solution covered just the surface of the TC membrane and dried, and this dried salt and PVA layer were utilized as a mask in the prevention of the oxygen plasma exposure. The covered mask is rapidly soluble in water and does not exert any chemical attack on the PDMS surface. The masked membrane layer was exposed to the oxygen plasma and bonded onto the hole-layer, and the pneumatic layer was aligned and bonded sequentially in accordance with the same procedure. Finally, the cover glass was stacked onto the pneumatic layer, and the constructed valve was cured for 8 h on the hot plate at 80 °C. To remove the salt and the PVA mask, we slowly operated a syringe pump (2 μl min^{-1}) to introduce distilled water into the inlet of the fabricated valve. Then, the fluid infiltrated the microgaps on the mask and melted the mask (figure 4(b)). By this procedure, the mask covering the surface of the TC membrane was completely washed out within 2 min.

4. Experiment

The fabricated microvalve was evaluated under various experimental conditions, and figure 5 shows the experimental configuration. The water column provided the microvalve's input flow with constant pressure, and the controllable compression and vacuum source was utilized as a pneumatic driving force. The pressure controller, which consisted of solenoid valves, a microcontroller-based (80C196K, Intel) electronic circuit, an LED display, and so forth, switched the solenoid valve alternatively to the compressor and the vacuum pump and, thus, provided the pneumatic port of the microvalve with compressed and vacuum force. Then, the compression-/vacuum-period (C/V) ratio could be adjusted by the pressure controller. The pressures at both the

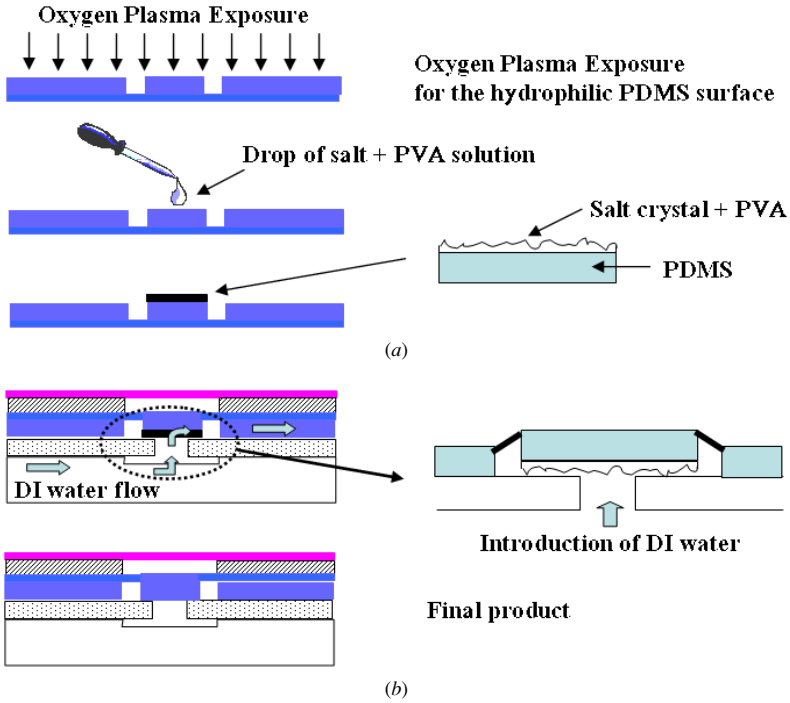


Figure 4. Process of *in situ* development. (*a*) Masking process with the salt (NaCl) and PVA solution, (*b*) development process with DI water.

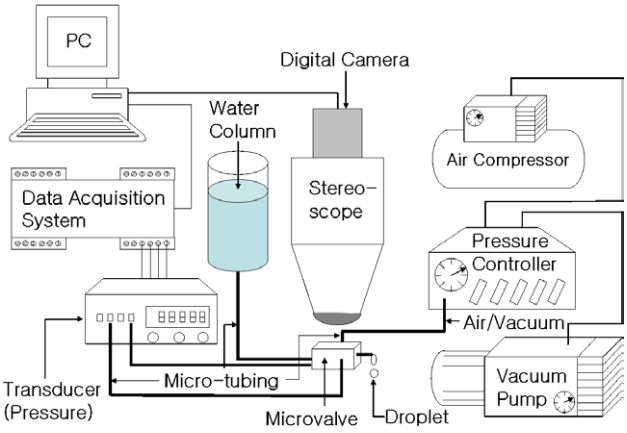


Figure 5. Experimental configuration for the static and dynamic operation of the valve.

inlet port and the microvalve's pneumatics port were measured with a modular pressure transducer (PX217A, OMEGA), and the pressure signal was sent to the personal computer via a data acquisition system. The operation of the valve was visually monitored under a stereoscope equipped with a digital camera, and the captured images were sent to the personal computer.

5. Results and discussion

The effectiveness of the water-soluble (WS) mask was evaluated first. Figure 6(*a*) shows an optical micrograph of the WS mask on the TC membrane. The mixture of salt and PVA solution uniformly covered the surface of the TC membrane. We visually inspected the surface of the mask by using a scanning electron microscope (SEM) (S-4300,

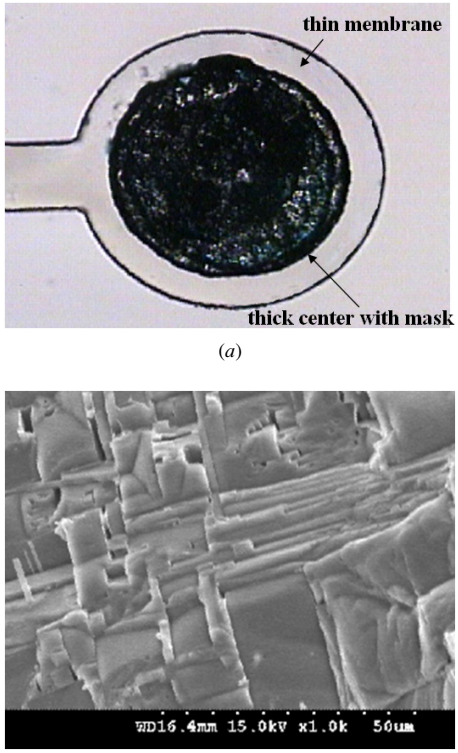
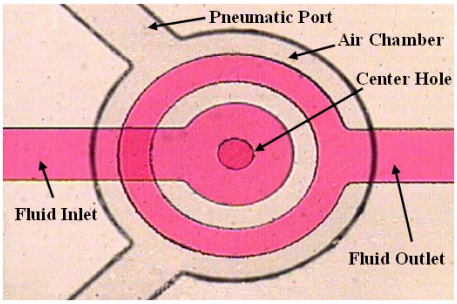
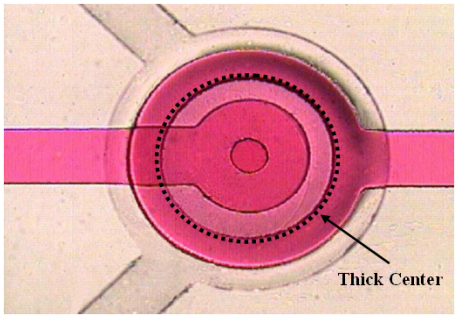


Figure 6. Optical and SEM micrographs of the WS mask. (*a*) Micrograph of the WS mask, (*b*) SEM image of the mask surface.

Hitachi), and figure 6(*b*) illustrates the captured microimage. As expected, the salt crystallized during the thermal curing process and its surface was rough, and the PVA was spread



(*a*)



(*b*)

Figure 7. Operation of fabricated valve: (*a*) closed valve, (*b*) open valve.

between the salt crystals. In general, this mask prevents exposure of the TC membrane's surface to oxygen plasma and helps DI water travel easily through the channel to wash out the WS mask. The coarseness of the mask's surface is one of the key characteristics of our method. If the surface is smooth, the TC membrane closely attaches to the hole-layer during the curing process, and this situation makes it difficult to separate the TC membrane from the hole-layer, even though the TC membrane is not exposed to the oxygen plasma. But if the surface is rough, then lots of microgaps form at the WS mask, and the DI water can infiltrate these gaps and enable the salt crystals and the PVA to be melted gradually. By this method, the TC membrane was easily separated from the hole-layer, and the WS masks were clearly washed out within 2 min. Via the regional WS masking, the controllable PDMS-based microvalve was successfully fabricated, and this method can be applied broadly in the fabrication of PDMS-based microstructures. This soluble mask-based fabrication method does not chemically attack the PDMS surface, nor does it leave behind any harmful chemicals; therefore, the method is very advantageous in the production of bio-applicable microfluidic valves.

The operation of the fabricated valve was tested through various experiments. The 'On/Off' operation as a microvalve was evaluated, and the results are displayed in figure 7. Water that was dyed red was injected into the inlet with a constant pressurized driving force, and the pneumatic force was applied using a 5 cc syringe. Figure 7(*a*) demonstrates the picture of the 'Off' state, and the well-aligned overall structure appears clearly. The red-colored fluid that was introduced into the fluid channel passed through the center hole and moved to the outlet. Owing to the compression of the TC membrane, the thick center stuck closely to the hole-layer (the black-circled region

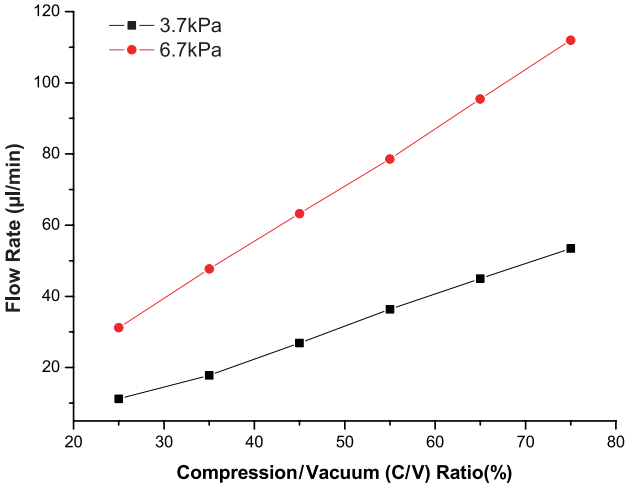


Figure 8. Change of the flow rate relative to the C/V ratio and the inlet pressure.

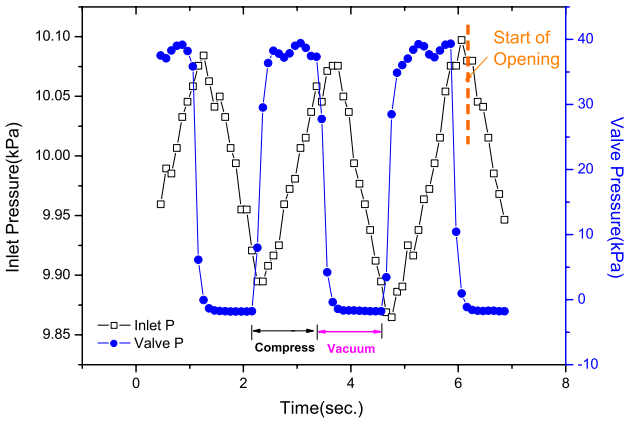


Figure 9. Response of valve operation inferred from the pressure curve.

in figure 7(*b*)), and the flow stopped. The slight vacuum force through the pneumatic port elevated the TC membrane, and the fluid started to flow through the center hole. The red fluid filled the space where the TC membrane was located, as illustrated in figure 7(*b*), and the basic operation of a microvalve was proved via this experiment.

Using the experimental setup (figure 5), we measured both the outlet flow variation as it responded to inlet pressure (regulated by changes in the height of the water column) and the C/V ratio; the results are illustrated in figure 8. As expected, the flow altered linearly with the C/V ratio and the inlet pressure, and this indicates that we can regulate the flow rate in the range of a few microliters per minute by adjusting the C/V ratio and the inlet pressure. More precise regulation of flow (e.g. sub-microliter per minute) could be achieved by the reduction of the channel size.

When the microvalve is operated by pneumatic force, the response time is slow. Figure 9 demonstrates the pressure change (at the inlet port) that corresponds to the variation of pneumatic pressure (the C/V ratio was fixed at 50%). Owing to the air compliance at the pneumatic port, a delay of approximately 500 ms in the opening of the valve occured. We expect that magnetic or other driving mechanisms could greatly reduce such a delay. A durability test of the valve was

carried out. A total of 10 valves were operated one million times simultaneously with a 50% C/V ratio and a 3.7 kPa inlet pressure, and no failures were observed. This result indicates that our valve is sufficiently durable for long-term applications.

6. Conclusion

In this paper, we fabricated a PDMS-based controllable valve via *in situ* development using a WS mask that is suitable for regional bonding. The TC membrane freely moved up and down according to the driving force generated through the pneumatic force, and this valve can provide fluid with the volume of a few tenths of a microliter per minute. Owing to the many advantages attributable to PDMS, such as biocompatibility and excellent transparency, this valve can be used in the delivery of biological samples or cell handling. And, with this valve, we can integrate diverse actuating forces. For example, if a microscale magnet disc is mounted on the membrane, a magnetically driven valve is possible. In this case, we expect that a faster 'On/Off' operation of the valve is possible and that a more compact driving system is realizable. The fabrication method is comparatively simple, and the time spent in accomplishing the total process is within 20 h. Owing to the simplicity of the fabrication process, this valve can be broadly applied at the laboratory level, which requires diverse dimensions, shapes and only small amounts of fluid delivery.

Acknowledgment

This study was supported by a grant of the Korea Health 21 R&D Project, Ministry of Health & Welfare, Republic of Korea (0405-ER01-0304-0001).

References

[1] Shoji S 1998 Fluids for sensor systems *Top. Curr. Chem.* **194** 163–88

[2] Kamholz A E, Weigl B H, Finlayson B A and Yager P 1999 Quantitative analysis of molecular interaction in a microfluidic channel: the T-sensor *Anal. Chem.* **71** 5340–7

[3] Hong J W, Studer V, Hang G, Anderson W F and Quake S R 2004 A nanoliter-scale nucleic acid processor with parallel architecture *Nature Biotechnol.* **22** 435–9

[4] Francais O and Dufour I 1998 Dynamic simulation of an electrostatic micropump with pull-in and hysteresis phenomena *Sensors Actuators A* **70** 56–60

[5] Andersson H, van der Wijngaart W, Nilsson P, Enoksson P and Stemme G 2001 A valve-less diffuser micropump for microfluidic analytical systems *Sensors Actuators* B **72** 259–65

[6] Kataoka D E and Troian S M 1999 Patterning liquid flow at the microscopic scale *Nature* **402** 794–7

[7] Feustel A, Krusemark O, Lehmann U and Müller J 1996 Electromagnetic membrane actuator with a compliant silicone suspension *Sperling T. Actuator'96 (Bremen, 26–28 June)* p 76

[8] Wang X, Vincent L, Yu M, Huang Y and Liu C 2003 Architecture of a three-probe MEMS nanomanipulator with nanoscale end-effectors *IEEE/ASME Int. Conf. on Advanced Intelligent Mechatronics (Chicago, 22–24 Apr.)* pp 891–6

[9] Peirs J, Reynaerts D and Van Brussel H 2001 The 'true' power of SMA micro-actuation *Proc. MME* pp 217–20

[10] Klintberg L, Karlsson M, Stenmark L, Schweitz J and Thornell G 2002 A large stroke, high force paraffin phase transition actuator *Sensors Actuators* A **96** 189–95

[11] Hua S Z, Sachs F, Yang D X and Chopra H D 2002 Microfluidic actuation using electrochemically generated bubbles *Anal. Chem.* **74** 6392–6

[12] Lazar I M and Karger B L 2002 Multiple open-channel electroosmotic pumping system for microfluidic sample handling *Anal. Chem.* **74** 6259–68

[13] Polson N A and Hayes M A 2000 Electroosmotic flow control of fluids on a capillary electrophoresis microdevice using an applied external voltage *Anal. Chem.* **72** 1088–92

[14] Unger M A, Chou H-P, Thorsen T, Scherer A and Quake S R 2000 Monolithic microfabricated valves and pumps by multilayer soft lithography *Science* **288** 113–6

[15] Jeon N L, Chiu D T, Wargo C J, Wu H, Choi I S, Anderson J R and Whitesides G M 2002 Design and fabrication of integrated passive valves and pumps for flexible polymer 3-dimensional microfluidic systems *Biomed. Microdevices* **4** 117–21

[16] Liu R H, Yu Q and Beebe D J 2002 Fabrication and characterization of hydrogel-based microvalves *J. Microelectromech. Syst.* **11** 45–53

[17] Eddington D T, Liu R H, Moore J S and Beebe D J 2001 An organic self-regulating microfluidic system *Lab Chip* **1** 96–9

[18] Walker G M, Ozers M S and Beebe D J 2002 Insect cell culture in microfluidic channels *Biomed. Microdevices* **4** 161–6

[19] Lee S H, Jeong W J and Beebe D J 2003 Microfluidic valve with cored glass microneedle for microinjection *Lab Chip* **3** 164–7

[20] Di Giovanni M 1982 *Flat and Corrugated Diaphragm Design Handbook* (New York: Dekker)