Integration of Heteromaterial Microelements into Plastic Microfluidic Devices by Capillary-force-assisted Micromolding

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We propose an advanced micromolding process for integrating heteromaterial microelements with complicated structures into a plastic substrate. The technique comprises the autonomous filling of dispensed liquid prepolymer into microstructures pre-engraved on a plastic substrate and subsequent curing to induce the cross-linking of the prepolymer. Specifically, the fabrication of a self-standing poly(dimethylsiloxane) elastomer membrane into a rigid plastic substrate has been investigated. Based on the experimental results, a guiding principle for the dimension control of the finally formed structure is discussed by considering the balance between the capillary force and gravity acting on the liquid prepolymer. Moreover, the present technique has been preliminarily applied to a lab-on-a-chip technology: the implementation of elastomer-based pneumatic microvalves on plastic-based microfluidic devices.

Keywords: micromolding, capillary force, poly(dimethylsiloxane), pneumatic microvalve

1. Introduction

In recent years, the use of microfabrication technology has rapidly spread from electronics to other fields including biological and/or medical applications. Lab-on-a-chip (LOC) technology that integrates micrometer-size reactors, fluidic channels, and other components on a single chip can provide revolutionary small size platforms for chemical and biological analysis, clinical testing, and so forth [1, 2]. Among the most important technologies for constructing a full-fledged integrated microfluidic system are microfluidic control elements such as microvalves and micropumps. Although extensive research has been carried out to develop various types of microvalves in the last few decades [3, 4], there is still plenty of scope for their further development with sufficient reliability and robustness for practical use. A typical success of microvalve technology now being practically used for micro total analysis systems was reported by Quake’s group in 2000 [5]. Their microvalve, named a monolithic poly(dimethylsiloxane) (PDMS) microvalve, is fabricated by multilayer soft lithography. It consists of a pneumatic channel that can be deformed under pressure to pinch off the flow of fluids in an underlying microchannel with a cross-over configuration. Although this microvalve is only compatible with the PDMS-based microfluidic platform developed by Whitesides et al., the use of an elastomer for reliable valve operation is now recognized as a “golden standard” among many researchers in the field of LOC technology.

On the other hand, the disposability of clinical diagnostic devices necessitates the use of plastics as the main substrate material to satisfy the practical requirements of a low production cost and mass producibility [6]. Therefore, several groups have explored a hybrid device approach, where soft materials such as elastomers are implemented into rigid plastic materials to fabricate microvalves or micropumps. The devices reported so far are typically composed of three layers. Zhang et al. fabricated pneumatic valves and pumps by sandwiching a PDMS membrane between fluidic channel and gas manifold layers made from
polymethyl methacrylate (PMMA) [7]. However, achieving permanent bonding between hetero-materials is often a challenge in fabricating such hybrid devices. Generally, bonding methods for polymer materials are categorized into direct bonding or indirect bonding, depending on the addition or nonaddition of an intermediate adhesive layer to the interface. In the case of indirect bonding, the use of a liquid adhesive sometimes causes the clogging of the micro-channels during the bonding step. Zhang et al. used a direct bonding approach involving UV/ozone surface treatment, but the bonding did not appear to be sufficiently strong as they reported the leakage of the valves when the fluid pressure was higher than 60 kPa. Similarly, Cortese et al. also reported a technique for bonding PDMS with cyclic olefin co-polymer (COC) by chemical surface modification and its use in microvalve fabrication. However, this technique requires several process steps [8]. In addition, it is worth noting that although the sandwich-type structure has been proposed owing to its ease of assembly, it often becomes a major obstacle in the design of the interlevel connections in multilevel fluidic channel networks. To overcome these problems, another approach for fabricating hybrid devices is required.

In this paper, we have proposed a simple and robust process for integrating complex-shaped elastomer microelements into a rigid plastic substrate. As a feasibility study, a microstructure having a thin deformable membrane was autonomously formed by directly dispensing PDMS prepolymer into a pre-engraved microstructure on a plastic substrate with subsequent curing. In addition, we present a preliminary application of the present technique to the implementation of PDMS pneumatic microvalves into a plastic-based microfluidic device.

2. Experimental Methods
2.1. Formation of self-standing PDMS membranes

Figure 1(a) shows the process flow of our proposed method of forming self-standing PDMS membranes on a plastic substrate. The main substrate material, poly(methyl methacrylate-co-styrene) (P(MMA-co-S)), was purchased from Japan Acryace. Using a computer numerical control (CNC) milling machine (MDX-540, Roland DG), a hole of the coaxial cylindrical tube structure as shown in Figs. 1(b) and (c) was engraved on a plastic substrate with 100×80 mm² area and 5 mm thickness. The inner and outer tubes were separated by a tubular wall suspended from the top with the gap of 0.5 mm from the hole bottom. The typical radii of the cylindrical tube structures used in this study were the inner tube radius \( R_1 = 1.4 \) mm, the outer tube radius \( R_2 = 2.5 \) mm, the thickness of the separation wall \( R_2 - R_1 = 0.4 \) mm, and the gap width of the outer cylinder \( R_3 - R_2 = 0.7 \) mm. A temporary surface protection film (SPV-P-6030, Nitto Denko) was softly laminated on one side of the plastic substrate to avoid the leakage of the liquid prepolymer during the filling step. Using an automatic fluid dispensing system comprising a precision dispenser (Nano Master SMP-III, Musashi Engineering) and a robot (SHOTMASTER 200DS, Musashi Engineering), a mixture of PDMS prepolymer and curing agent with 15:1 (v/v) ratio (Sylgard 184, Dow Corning Co., MI, USA) was poured into the pre-engraved inner cylindrical hole on the plastic substrate. The volume of dispensed PDMS prepolymer used in this study was 35-50 \( \mu \)L. After being left to rest for 10 min, sufficient time for the prepolymer to reach the steady state, the filled prepolymer was cured in an oven at 80°C for 1 h. Finally, the protective film was delaminated from the plastic substrate.

![Figure 1](https://example.com/figure1.png)

Figure 1. (a) Process sequence for fabrication of a self-standing elastomer membrane by capillary-force-assisted micromolding. (b) Coaxial double cylindrical structure engraved into a plastic substrate. A partition wall is suspended from the top plane via four supporting bridges. (c) Cross-sectional view of the hole structure.
2.2. Bulge test of self-standing PDMS membranes
To evaluate the mechanical properties of self-standing PDMS membranes as well as the strength of bonding between the PDMS sheet and the P(MMA-co-S) substrate, a pressure bulge test was conducted using the experimental setup schematically shown in Fig. 2. A uniform pressure was applied to one side of the circular PDMS membrane using pressurized nitrogen and the deflection of the membrane was measured as a function of the applied pressure, which was measured using a digital pressure sensor (AP-C33, Keyence).

Figure 2. Experimental setup used for bulge test of the self-standing PDMS membrane.

3. Results and Discussion
3.1 Integration of self-standing PDMS membrane into plastic substrate
Soon after the PDMS prepolymer was dispensed in the inner tube, it began to flow into the outer tube. When the space of the outer tube was relatively narrow, the level of liquid prepolymer in the outer tube exceeded that in the inner tube due to the strong capillary effect, consequently leaving a thin membrane at the bottom of the hole as shown in Fig. 3. The final state of the liquid prepolymer in the hole of the coaxial double tube structure can be predicted by considering the balance between the capillary force and gravity (see Fig. 4). When the effect of the meniscus is neglected for simplification, the difference in the liquid level between the inner and outer tubes \( H-h \) can be described by the following formula.

\[
(H - h) = \frac{2 \gamma \cos \theta}{\rho g} \left( \frac{1}{R_3} - \frac{1}{R_2} + \frac{1}{R_1} \right)
\]

(1)

Figure 3. Side-view photo of PDMS filled inside the hole of the cylindrical double tube structure pre-engraved on a plastic substrate.

Figure 4. Schematic model that describes the balance of forces acting on the liquid prepolymer dispensed in the coaxial double tube structure engraved in a plastic substrate. The radii \( R_1, R_2, \) and \( R_3 \) of the coaxial double tube structure are geometrical parameters used to determine the degree of the capillary effect.

Here \( \rho, \gamma, \) and \( \theta \) are the density, surface tension, and contact angle of PDMS, respectively. \( R_1, R_2, \) and \( R_3 \) are the radii of the valve structures as denoted in Fig. 4.

To verify the validity of the model, the \( H-h \) value was experimentally investigated by varying the volume of dispensed PDMS prepolymer as shown in Fig. 5. Also, the effect of the geometrical parameters of the engraved hole was examined by considering two different inner tube diameters, \( R_1 = 1.0 \text{ mm} \) and \( 1.4 \text{ mm} \). From equation 1, it was expected that the liquid level difference \( H-h \) would depend on the radii \( R_1, R_2, \) and \( R_3 \) of the tube structure but not on the volume of dispersed
Indeed, the $H-h$ values were not affected by the volume of dispensed PDMS and were almost constant at $1.87\pm0.0261$ (1.4%) mm and $2.49\pm0.0962$ (3.9%) mm for $R_1=1.0$ mm and $R_1=1.4$ mm, respectively.

Figure 5. Difference in the liquid level between the inner and outer tubes ($H-h$) as a function of the dispensed PDMS volume. The difference in the liquid level is constant irrespective of the dispensed volume of PDMS ($N=3$) when the dimensions of the engraved holes are the same. $H-h$ for two different diameters of the inner tube ($R_1$) of 1.0 mm and 1.4 mm was examined with the dimension of the outer tube fixed; the tube thickness ($R_3-R_2$) was 0.7 mm. Closed and open circles indicate the results for $R_1=1.0$ mm and 1.4 mm, respectively.

Figure 6. Pressure-deflection curve obtained from bulge test of self-standing PDMS membrane fabricated on plastic substrates. The deflection was defined as the height of the PDMS membrane at the center. The vertical error bars resulted from the resolution limit of the CCD camera used to image the deflected membrane. Side-view photographs of the deformed PDMS membrane at 0.1, 0.15, and 0.2 MPa are inset in the figure.

3.2. Bulge test

As shown in Fig. 6, the deflection of the membrane showed a quadratic increase with increasing applied gas pressure; it was 0.5 mm at 0.1 MPa and 2 mm at 0.2 MPa. In addition, the membrane ruptured when a pressure higher than 0.225 MPa was applied, while no failure was observed at the interface between the PDMS and the plastic substrate. These results illustrate the following two aspects: the successful formation of a thin PDMS membrane at the bottom of the hole and the strong anchoring of the PDMS in the outer tube via the large contact area.

3.3. Discussion and preliminary application to microfluidic devices

The process known as micromolding in capillaries (MIMIC) was originally developed by Whitesides et al. in 1995 [9], after which there were several reports on its use in producing complex polymeric microstructures supported on different substrates and the applications of these microstructures in microfabrication [10, 11]. MIMIC is a type of soft lithography, which is a collective term for several micropatterning techniques: microcontact printing (µCP), replica molding (REM), microtransfer molding (µTM), MIMIC itself, and solvent-assisted micromolding (SAMIM) [12]. In MIMIC, an elastomeric micromold typically made from PDMS having a patterned relief structure on its surface is used. After filling the channels and curing the prepolymer to form a solid cross-linked polymer, the PDMS mold is removed and a network of the polymeric material remains on the surface of the substrate [11]. The technique reported in this paper is also a type of MIMIC by its nature, but it has rather different technical aspects since it does not involve the preparation of negative replicas on the solid surface. In addition, since the prepolymer is partially filled in the engraved structures, the use of a dispenser and the optimal structure design with due consideration of the effect of the capillary force and gravity are essential.

As a preliminary study towards the application of our technique to the fabrication of LOC devices such as elastomer-based microactuators, a normally open pneumatic microvalve was implemented in a plastic-based microfluidic device. As shown in Fig. 7(a), the microvalve was composed of a fluidic channel layer and a gas manifold layer. A straight fluidic channel with a
rectangular cross section of 0.5×2 mm² area and inlet/outlet holes with 3 mm diameter on both ends was engraved on a plastic substrate with dimensions of 100×80×5 mm³. Using the micro-molding technique, a self-standing PDMS membrane was formed on another plastic substrate having a gas manifold channel. Finally the fluidic channel layer and the gas manifold layer were assembled by vapor-phase solvent bonding using toluene [13].

The prototype microvalve was operated as follows. The flow channel and the valve were filled with a fluorescent aqueous solution of 0.05 w/w% sulforhodamine B (Kanto Chemical). To monitor the valve actuation, fluorescence emission from the space with variable volume in the microvalve was measured. The light was delivered through an iris (Thorlabs ID15/M) and a lens (LA1805, Thorlabs) to an avalanche photodiode (APD; C5460, Hamamatsu Photonics). The analog signal from the APD was also converted to a digital signal using a data acquisition module (NIUSB-6008, National Instruments). As plotted in Fig. 7(b), the fluorescence signal decreased with increasing pneumatic pressure applied to the PDMS membrane. A rapid drop in the signal caused by the deformation of the PDMS membrane was observed from 0.1 MPa, and complete closure of the valve was reached at approximately 0.2 MPa. In this test, no rupture of the PDMS membrane was observed even at pressures of up to 0.4 MPa in contrast to the bulge test because the deformation of the PDMS membrane was limited by the walls of the fluidic channels.

4. Conclusion

A process for integrating elastomeric microelements into plastic substrates by the micro-molding of a dispensed liquid prepolymer with the aid of the capillary force was studied. Specifically, a self-standing elastomer membrane was fabricated by dispensing PDMS prepolymer into a coaxial cylindrical tube microstructure pre-engraved on a plastic substrate with subsequent curing. In the practical design of implemented microelements, prediction of their final shape is necessary with the consideration of the force balance between the capillary force and gravity. A strong and reliable mechanical joint was obtained without using an adhesive glue by adopting a suitable structure that ensures a large contact area between the heteromaterials. Moreover, the developed molding process was applied to the fabrication of a pneumatic microvalve on a plastic-based microfluidic device and its successful operation was demonstrated. This robust and scalable process is expected to be useful for manufacturing disposable hybrid microfluidic devices that can be used in various biomedical applications.
Acknowledgements

This research is partially supported by the Japan Society for the Promotion of Science (JSPS) through the “Funding Program for World-Leading Innovative R&D on Science and Technology (FIRST Program)” and the Center of Innovation Program from Japan Science and Technology Agency, JST.

References