RAPID PROTOTYPING OF MICROFLUIDIC COMPONENTS

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Abstract

Microfluidic devices are receiving considerable attention in the fabrication of micro-electro-mechanical systems for biotechnological applications (e.g. BioMEMS). Microscale total analysis systems can reduce cost and increase speed of analysis, especially through reduced use of reagents and reduced system size. A key factor in developing new microfluidic applications is the development of prototyping technologies for fabricating and testing new system designs that incorporate fluidic networks, microreactors, separation, and detection systems. One technique for fabricating microfluidic networks is based on replica molding of microchannels with poly(dimethyl siloxane) (PDMS). Microchannel networks with feature size down to 5 μm have been demonstrated with PDMS replica molding. PDMS replica molding can fabricate microchannel networks with a turnaround period of hours to days using inexpensive, benchtop equipment. However, valves and pump technologies are still macro-scale (e.g. external syringe pumps, peristaltic pumps, or power supplies for electro-osmotic pumps). We present a rapid prototyping technique for fabrication of microfluidic networks and active components (e.g. valves) based on PDMS replica molding combined with magnetic actuation. This technique allows embedding of magnetically actuated, mechanically active polymers within the PDMS microchannel network. By activating the microchannels, the channels may be closed or opened (e.g. valving action). Multiple valves and reservoirs may be actuated in order to create pumping via peristaltic action. Magnetic actuation may be achieved by incorporating a magnetically active material into the PDMS (whether permanent magnet, or high permeability material), and applying an external magnetic field from a printed circuit board. Both valves and pumps have been demonstrated. Using inexpensive benchtop equipment, we can fabricate actual devices with sizes on the order of a few millimeters, with channel sizes on the order of a few hundred micrometers. We have demonstrated active valves that close leak-tight, and withstand back pressures on the order of 1.5 kPa.

Introduction

Recent years have witnessed an explosion in the types and proposed uses of microfabricated fluidic systems, especially in bioanalytical applications. A significant impediment to the incorporation of microfluidic devices is the difficulty of designing, building, and testing microfluidic systems. We describe a convenient method for creating new microfluidic systems that incorporate active microfluidic components (namely valves) that combine magnetic actuation with elastomeric materials. This method can be easily extended to allow the rapid fabrication of a variety of other active microfluidic components such as pumps, injectors, flow controllers and mixers, thus allowing the rapid design, fabrication, testing and utilization of complete microfluidic systems (e.g., within a day).

This work is an extension of methods for rapid prototyping of fluidic microchannels based on replica molding of elastomers, such as poly(dimethyl siloxane) (PDMS). Microchannel networks with feature sizes down to ~5 μm can easily be formed by replica-molding of three-dimensional patterns formed by simple photolithographic methods. Silicone polymers such as PDMS have high compliance, high elongation, and good sealing properties, making them useful valving materials. Elastomer-based valving strategies have recently been reported by Unger et al., who described pneumatic methods for deforming elastomeric microchannels to form microvalves and micropumps.

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By combining elastomers with magnetic materials and using electromagnets in the substrate, active valves can be fabricated. This approach is simpler and allows miniaturization, by removing the need for macroscopic, externally switched pneumatic supplies.

Theory

Electromagnetism is frequently used for actuation. Force production is either due to the interaction between the applied field and magnetic domains in a ferromagnetic material (e.g. an electromagnet lifting scrap metal), or to the interaction between an applied field and an electric current (e.g. a voice coil speaker). A magnetic field may be generated by an electric current (e.g. a coil, or electromagnet), or may be intrinsic in a material (e.g. a permanent magnet). Modulation of force, or control of force is achieved through the control of current in an electromagnet.

There are three complications:

- the force of magnetic attraction between an electromagnet and a ferromagnetic material drops off quickly with distance,
- elastomers (being non-ferromagnetic) have magnetic permeabilities several orders of magnitude smaller than ferromagnetic materials, and
- ferromagnetic materials are typically ceramics or metals, and do not have satisfactory elastic properties.

A solution to these problems is to create microfluidic systems that contain areas of "magnetic elastomer" which serve as actuation points for processes such as valving, pumping, and mixing. The magnetic polymer can be permanently magnetized or not, and can be created by blending ferromagnetic powder (such as magnetically soft materials such as iron or silicon iron; or high hysteresis, magnetically hard alloys such as NdFeB, strontium oxides or iron oxides) with the elastomer precursor at the desired locations before curing. Thus, the magnetic elastomer will have a combination of the necessary magnetic permeability and the needed elastic properties. Alternatively, a nonmagnetic elastomer forming a microfluidic channel can be over-coated with a magnetic material; however, this increases the distance from the electromagnet, and hence, decreases the magnetic force that can be applied to it.

For materials without permanent magnetization, reversing action is due to the elastomeric nature of the composite material: When the electromagnet is deactivated, the elastomer returns to its initial, undeformed shape, thus opening the channel for fluid flow. If an elastomeric magnetic composite is permanently magnetized, it should be possible to control both forward and reverse action in an actuator by switching the direction of the current in an external electromagnet.

Forces on ferromagnetic materials are based on aligning the magnetic moments within the materials. The magnetic pressure F/A (force per unit area)between an electromagnet and a ferromagnetic material is approximately

$$\frac{F}{A} = \frac{B^2}{2\mu_0}$$

where B is the flux density, and μ_0 is the permeability of free space. The flux density B can be calculated using finite element analysis software, estimated by using magnetic circuit theory, or measured empirically. Permanent magnet materials such as NdFeB alloys can support flux densities on the order of 1 tesla at the magnet face, and transformer laminations can support about 1 T before saturation occurs; therefore, the practical upper limit on the amount of pressure which may be generated with readily available materials is approximately 400 kPa. In practice, we would expect to see less pressure, since the magnetic elastomer needs to work against the elastomer modulus, and the magnetic circuits are not optimized.

Experimental procedure

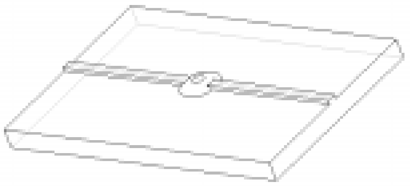
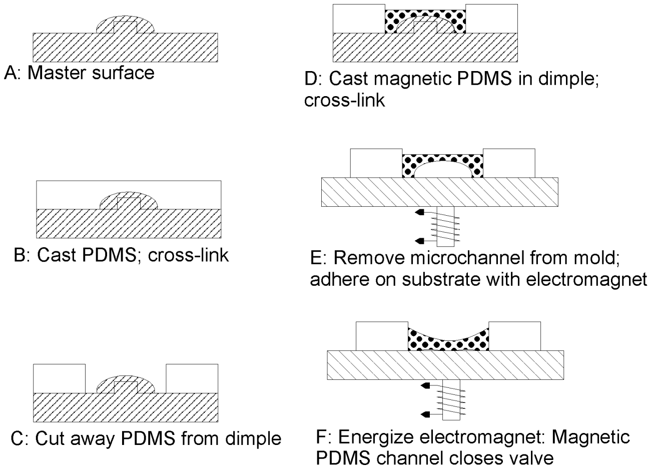


Figure 1. Master form, which serves as replica mold for a microfluidic channel and valve.

The fabrication of components is based on replica molding of a master form (which is a negative of the microfluidic networks). Figure 1 shows a sample master form. These masters may be fabricated by any convenient means, such as machining, rapid prototyping, or photolithography. For "small" (μm size) microchannels, typical master fabrication is by spinning photoresist on a flat surface (e.g. a silicon wafer), and generating a pattern by exposing the photoresist through a laser-printed or linotronic-printed transparency. For convenience, the masters described here were fabricated by machining aluminum, with channel widths of 250 μm and height of 100 μm.

The typical PDMS elastomer used for replication in these experiments is Dow Sylgard 184, though many silicone elastomers may be used. Standard Sylgard 184 preparation is mixing 10:1 elastomer to curing agent by weight, degassing, then, pouring on the master form. After cross-linking (either 24-48 hours room temperature cure, or 2 hours to 30 minutes at 70-150 C in a convection oven), the replica may be peeled off of the master, and adhered to a flat substrate to form microfluidic channels and networks. The resulting elastomer is quite compliant, with a Young's modulus measured at 0.48 MPa.

To make valves, we need to make a magnetically active elastomer. This is done by making a more elastic polymer (32:1 elastomer to curing agent; measured modulus 0.29 MPa), and mixing this with a magnetic material (in this case, 50% by weight of iron powder. Adding the filler can result in increasing the modulus by as much as a factor of 4). This magnetic elastomer is cured in the locations where the dimples are on the master. Our procedure is to actually cut away the cured non-magnetic elastomer from the dimple area; pouring in magnetic elastomer over the dimple; and curing the magnetic elastomer. The resulting elastomer network with magnetic elastomer is then peeled from the master, and adhered to a flat substrate. Electromagnets (extracted from Radio Shack relays) are placed under the substrate, which can then be used to turn the valves on or off. The measured magnetic field with the electromagnets energized, at the level of the elastomer, is 87 mT (870 gauss) A schematic of the process is shown in Figure 2, and photographs of the master part and the resulting microfluidic network with two valves are shown in Figure 3.



A: Master surface

D: Cast magnetic PDMS in dimple; cross-link

B: Cast PDMS; cross-link

E: Remove microchannel from mold; adhere on substrate with electromagnet

C: Cut away PDMS from dimple

F: Energize electromagnet: Magnetic PDMS channel closes valve

Figure 2. Process of making magnetic elastomeric microvalves

PDMS tends to adhere more to itself than to other materials before cross-linking, this makes mold releases unnecessary. The downside is that the microchannels will not adhere very well to the substrate either, unless the substrate is carefully cleaned, or unless a layer of PDMS is also applied to the flat substrate prior to adhering the molded microchannel. This, and the low physical strength of the elastomer, will limit the microchannel fluidic systems to low pressure applications.

Results

Experimental results with these microchannels show leak-tight seals to water and to air. When an external pressure of greater than 1.5 kPa is applied to one side of a closed micro-valve, the PDMS fails. When the valve is open, flows of 250 µL/min are accommodated with a pressure drop of 0.55 kPa across a length on the order of 100 mm of microchannel and valve. Figure 4 shows some results from two-valve switching to select between a fluorescent and a clear flow.

Figure 3. (Left) Master aluminum part, which has raised channels, wells, and dimples. (Right) After replica molding, the elastomer microchannel is peeled from the master. Two magnetic valves are shown.

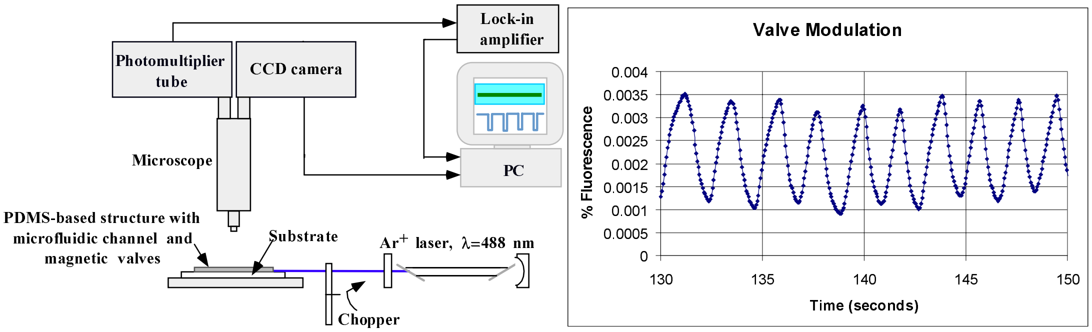


Figure 4. Valve modulation and lifetime tests; one side of the T network has fluorescent dye; the other, plain water. The fluorescence is detected as the two valves on the T are switched on/off off/on every second for several hours

Conclusions

We have demonstrated a simple, inexpensive approach for rapid prototyping of microfluidic components. Although the pressure capability of our current process is not very high, we are optimizing material properties, magnetic designs, and processing methods. Complementing development of microfluidic fabrication methods, collaborations with sensor development, assay development, fluid dynamics analysis, and biological tests are ongoing.

References

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