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A magnetically driven PDMS micropump with ball check-valves

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Abstract
In this paper, we present a low-cost, PDMS-membrane micropump with two one-way ball check-valves for lab-on-a-chip and microfluidic applications. The micropump consists of two functional PDMS layers, one holding the ball check-valves and an actuating chamber, and the other covering the chamber and holding a miniature permanent magnet on top for actuation. An additional PDMS layer is used to cover the top magnet, and thereby encapsulate the entire device. A simple approach was used to assemble a high-performance ball check-valve using a micropipette and heat shrink tubing. The micropump can be driven by an external magnetic force provided by another permanent magnet or an integrated coil. In the first driving scheme, a small dc motor (6 mm in diameter and 15 mm in length) with a neodymium–iron–boron permanent magnet embedded in its shaft was used to actuate the membrane-mounted magnet. This driving method yielded a large pumping rate with very low power consumption. A maximum pumping rate of 774 µL min−1 for deionized water was achieved at the input power of 13 mW, the highest pumping rate reported in the literature for micropumps at such power consumptions. Alternatively, we actuated the micropump with a 10-turn planar coil fabricated on a PC board. This method resulted in a higher pumping rate of 1 mL min−1 for deionized water. Although more integratable and compact, the planar microcoil driving technique has a much higher power consumption.

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

1. Introduction

Microvalves and micropumps are considered to be key functional components in microfluidic systems. Over the past decade with an increasing interest in the development of biochemical microsystems, there have been numerous reports on the design and fabrication of a variety of microvalves and micropumps. Microvalves can be classified into two types: passive and active, depending on whether they use an external energy source for their operation [1, 2]. Passive valves operate based on the pressure gradient and are often used as check-valves for micropumps, while active valves require outside actuation (e.g., electrostatic, electromagnetic, thermal expansion, thermopneumatic, pneumatic, piezoelectric and shape memory alloy). Micropumps have generally been developed in three different categories, i.e., electroosmotic, positive-displacement and peristaltic [1–13]. Depending on working fluid type, required pumping rate and desired backpressure, various actuation mechanisms and fabrication techniques have been employed. Electroosmotic pumps have the advantage of high backpressures and simple design, but they require very high voltages and a charged working liquid (alternatively, one can pump an uncharged liquid if a charged contact surface can be provided). Peristaltic pumps have also been demonstrated using different fabrication and actuation techniques such as thermopneumatic, piezoelectric,
electrostatic and magnetic. However, they usually have relatively small pumping rates, low backpressures and require a rather complex driving circuitry. Positive-displacement pumps, with the exception of nozzle-diffuser designs, require complex fabrication methods for creating two check-valves and an actuation chamber. Overall, three main drawbacks associated with current micropumps include: (1) fabrication complexity, (2) cost and (3) power consumption. A recent paper on micropumps by Laser and Santiago provides a more detailed review on various pumping schemes and structures [14].

In this paper, we report on the design, fabrication and testing of a magnetically actuated positive-displacement PDMS micropump addressing all of the above concerns. Figure 1 shows a perspective view of the PDMS micropump along with its equivalent circuit diagram [15]. In this diagram, the two diodes simulate the one-way performance of the passive check-valves. The ac voltage represents the periodic pumping performance of a positive-displacement pump, e.g., the micropump drive has the advantage of being more compact. However, this is at the expense of much higher power consumptions. For the purpose of clarity, in the following sections, we call the two aforementioned driving schemes ‘micromotor-drive’ and ‘microcoil-drive’.

As will be shown in the following sections, the pumping rates achieved in our designs range from 100 µL min\(^{-1}\) to 1 mL min\(^{-1}\). Micropumps operating at these rates are particularly useful in applications such as (1) handheld devices for environmental sampling, e.g., water pollution measurement and bacterial detection, (2) infusion pumps for bedside and ambulatory drug delivery and (3) disposable lab-on-a-chip systems for chemical and biological analysis. Many of these applications can benefit from a low cost, low power consumption and high flow rate micropump.

2. Design and fabrication

2.1. Assembly of one-way ball check-valves

The one-way behavior of check-valves significantly affects pumping performance of a positive-displacement pump, e.g., valve leakage reduces backpressure and pumping rate. A novel micropipette–microtube molding technique is used to fabricate the ball check-valves. Figure 2 illustrates the basic components, assembling process, and final packaging of the valves. First, a tapered tube having an inner diameter (ID) of 0.7 mm at the narrow end and 1.2 mm at the wide end is cut from a plastic micropipette. Then, a small stainless steel microsphere (diameter = 0.8 mm, grade = 24, and sphericity = 0.6 µm, Small Parts Inc., FL) is introduced into the tube. A short piece of a Teflon™ tubing (ID = 0.56 mm, OD = 1.1 mm) is cut at a 45° angle at one end and is then inserted into the wide end of the tapered tube. This piece acts as a stopper for the micropipette. Finally, two simple microtubes are connected to each side of the tapered micropipette and sealed by polyolefin heat shrink jackets at 150°C. Using this technique, one can easily achieve a one-way ball valve in a tapered tubing structure with microfluidic tube connections, utilizing its low cost and easily reproducible design.

2.2. Molding of micropump

Figure 3 shows an expanded diagram of the PDMS micropump with two driving schemes. Silicone elastomer (Silgard 184, Dow Corning Corp., MI) was the main structural material used for the body of the pump. The bottom layer was molded to incorporate one actuating chamber and two assembled ball check-valves with microfluidic connectors. The top layer was simply a thin sheet used as the actuation membrane and did not require any mold. Two ball check-valves were fabricated as mentioned in the previous section and horizontally placed into a mold which also held a 1.0 mm thick metallic disc (i.e., the mold of actuation chamber). This configuration results in a lower flow resistance and smaller overall dimensions, as compared with the other alternative, i.e., vertical placement. The valves were orientated so that they would both allow liquid to flow in one direction and block the back flow in the opposite direction. In addition, smooth surfaces of steel ball
A magnetically driven PDMS micropump with ball check-valves

![Diagram](image)

**Figure 2.** Illustration of (a) basic components, (b) assembling process and (c) final packaging of the micro-ball valve (not to scale).

![Diagram](image)

**Figure 3.** Expanded diagram of the PDMS micropump with two driving schemes: (a) micromotor-drive and (b) microcoil-drive.

and tapered tube provide little resistance to the movement of the ball. Subsequently, PDMS (Silgard 184, 10:1 ratio) was poured over the mold of 1.5 mm thick and cured at 100 °C in an oven for 15 min. After curing, the metallic chamber mold could be easily removed (PDMS is not adhesive to metallic surface), leaving the micro-ball valves in place and creating the actuation chamber. A 100 µm thick layer of PDMS was prepared and cured onto a flat surface to serve as the actuation membrane. To bond the actuation membrane to the bottom layer, two different methods were employed. In the first approach, a very thin film of uncured PDMS was applied to the bonding surface of the molded PDMS substrate which was then flipped over and bonded to the thin actuating membrane at a 100 °C oven for 3 min. Alternatively, oxygen plasma treatment (100 W, 13.3 Pa for 30 s) followed by a short curing step (a relative pressure of 100 kPa at 200 °C) was used as a bonding strategy [5]. Both of these methods yield a very strong bond between the two PDMS layers preventing any possible leakage. A NdFeB magnetic disc having a diameter of 3.1 mm and 1.6 mm thick (residual induction $B_r = 12,900$ Gauss) was finally attached to the actuating membrane right above the center of the actuation chamber using a thin layer of adhesive (0.9 g in weight of the magnetic disc has negligible effect on the membrane deflection compared with external driving forces). The top encapsulating layer and planar coil used in the microcoil-drive scheme (10-turn planar copper coil of 200 µm line width machined on a PC board of 35 µm thick copper film) were also bonded using the first technique. The magnetic field strength ($B$) at 2 mm above the center of the integrated coil, which is the rest position of the actuation magnet, can be also estimated using Biot–Savart’s law:

$$B = \frac{\mu_0}{2\pi} \frac{I}{z^2 + R^2}.$$  

In addition, a dead volume of 42 µl was measured from the difference in the weight between the ‘dry’ (without water) and ‘wet’ (filling with water) states. Finally, an ideal stroke volume ($V$, i.e., chamber size) of 36 µl was calculated from the dimensions of the mold disc. Figure 4 shows a photograph of the micropump.

3. Experimental setup

In the micromotor-drive scheme a miniature dc coreless motor of 6 mm diameter and 15 mm length (MicroMo Electronics Inc., FL) was used to actuate the micropump. For this purpose, a polyvinylsiloxane shaft with an embedded NdFeB permanent magnet (3.1 mm in diameter and 1.6 mm thick, $B = 500$ Gauss at 2 mm above its surface) was constructed for the micromotor. The pump was then placed on a glass slide that separated it from the micromotor resulting in a 2 mm distance between the external and internal magnets. In the microcoil-drive scheme, the pump was powered by a Crown PB-2 power amplifier triggered by a HP 33120A function generator. Using Biot–Savart’s law, a magnetic flux density of 10 Gauss is calculated at 2 mm above the center of the integrated coil,
Figure 4. A photograph of the micropump (a US quarter coin is placed for size comparison).

assuming a current of 1.5 A. Although the micromotor-drive scheme resulted in a much higher magnetic flux density \( B = 500 \) Gauss than its microcoil-drive counterpart \( B = 10 \) Gauss, since the magnetic field orientation constantly rotates as the motor shaft spins, the actuation time is somehow limited (see the measurement result section). Figure 5 shows the test and measurement setup for both driving configurations.

Deionized (DI) water (density \( = 1.0 \) g ml\(^{-1}\) and viscosity \( = 1.025 \times 10^{-3} \) Pa s at room temperature) and olive oil (density \( = 0.92 \) g ml\(^{-1}\) and viscosity \( = 81 \times 10^{-3} \) Pa s at room temperature) were chosen as the pumping liquids to represent a wide viscosity range. The priming process involved submerging the inlet tube in liquid, and using a mild suction on the outlet to draw-in the fluid and eliminate any air bubble. This priming process is critical to avoid air bubble trapped inside, since any air bubbles in the fluidic channels or actuation chamber will induce surface tension effects and significantly compromise micropump and valve performance.

4. Measurement results

Figure 6 shows leakage measurement results for the ball check-valves. As can be seen, the valves perfectly seal at pressures lower than 5 kPa. A leakage of less than 1 \( \mu \)L min\(^{-1}\) is measured between 5 and 30 kPa. Above this threshold pressure of 30 kPa, the valves show a nonlinear behavior with a leakage rate increasing faster than pressure. This can be attributed to the non-ideal circular cross-section of the micropipette and its deformation at high pressures, and the surface roughness and spherical deviation of the steel balls.

Equation (1) is used to obtain the actual rotation speed and frequency of the micromotor-drive:

\[
V_0 = (I \times R) + V_e = (I \times R) + (\omega \times k_e),
\]

where \( V_0 \) and \( I \) are the supply voltage and current, \( V_e \) is the back EMF, \( k_e \) is a EMF constant and \( R \) is the terminal resistance. The values for \( k_e \) \( (0.212 \) mV rpm\(^{-1}\)) and \( R \) \( (37.7 \) \( \Omega \)) can be obtained from the manufacturer.

To perform the pumping rate test, two additional silicone tubes (ID = 1.5 mm and OD = 1.9 mm) were used to connect the inlet valve to a reservoir and the outlet valve to a measuring cylinder. Laminar flow in microfluidic devices allows the inlet and outlet fluidic resistances to be directly calculated using Poiseuille’s equation:

\[
R = \frac{8 \eta L}{\pi r^4} \quad (R \text{ is the flow resistance, } \eta \text{ is the fluid viscosity and } L, r \text{ are length and radius of the flow channel}).
\]
microcoil were primarily designed to illustrate the pumping capability and improved performance is achievable using an electromagnetic drive. Power consumption and heating effects can be easily reduced through the fabrication of a multi-layer low-profile solenoid (to increase the magnetic field), although it will be extremely challenging to compete with the micromotor-driven pump in power dissipation and efficiency.

We also conducted several tests to measure the pumping backpressure. For these measurements, the inlet needle was connected to a reservoir of water via a primed tube while the outlet needle was connected to a vertical tube used to measure the pumped liquid height. A graph of pumping backpressure (at zero flow rates) versus motion frequency of the actuating membrane for both driving schemes is shown in figure 8. A maximum backpressure of 7.5 kPa (at 4.0 Hz) for DI-water in the microcoil-driven pump was measured, while 3.6 kPa (at 4.8 Hz or 286 rpm) and 1.7 kPa (at 6.2 Hz or 373 rpm) were the peak backpressures measured for DI-water and olive oil in the micromotor-driven pump, respectively. As can be seen, the actual backpressure is lower than the valve leakage threshold pressure (30 kPa). This is due to the back flow associated with the lag in the closure of the ball check-valve.

It is instructive to compare the performance of the micropump described in this paper with several similar devices in the market and some previously reported micropumps in the literature. Table 1 shows the comparison of key characteristics of several micropumps. Bartels mP5 and ThinXXS MDP1304 are positive-displacement micropumps manufactured by two German companies [16, 17]. The former is made of plastic layers, while the latter one consists of metal and plastic materials. EsoxPump™ V01 is using a deformed elastomeric membrane to drive the micropump [18]. In addition, Tai’s and Quake’s groups at CalTech have fabricated a variety of micropumps from silicon and polymer (Parylene and PDMS) [4, 19]. Compared with these pumps, one can point to several advantages associated with our micropumps. These include (1) they are fabricated from inexpensive building materials such as PDMS, micropipette tubing, and stainless steel microspheres, (2) the microcoil-driven pump total dimensions are $12 \times 10 \times 5$ mm$^3$ which is one of the smallest among those reported and marketed pumps, and (3) the actuation frequency can be controlled from 1 to 6 Hz (60 rpm to 360 rpm) for DI-water and olive oil which is lower than the valve leakage threshold pressure (30 kPa). This is due to the back flow associated with the lag in the closure of the ball check-valve.
(it still provides a comparable maximum performance such as pumping rate and backpressure) and (3) the micromotor-driven pump has one of the lowest power consumption among the micropumps with hundreds of \( \mu \text{L min}^{-1} \) pumping rates, although some other pumps, e.g., Bartels mP5, are targeting higher backpressures (50 kPa) and pumping rates (3000 \( \mu \text{L min}^{-1} \)).

5. Conclusions

In this paper, we reported on a PDMS membrane micropump with two one-way ball check-valves. The fabrication process includes one simple PDMS molding step with an additional layer of actuating membrane mounted with small permanent magnets. Two driving schemes, micromotor-drive and microcoil-drive, were investigated. Large pumping rates were achieved in both schemes (hundreds of \( \mu \text{L min}^{-1} \)) with the micromotor-drive having the lowest power consumption reported in literature. The microcoil-driven pump was fully integrated with an even higher pumping rate (1 mL min\(^{-1}\)) and back pressures (7.5 kPa, a human heart can generate 13 kPa) at the expense of having much higher power consumption.

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