Published in final edited form as:

J Micromech Microeng. 2010 ; 20(9): 095033–. doi:10.1088/0960-1317/20/9/095033.

A planar PDMS micropump using in-contact minimized-leakage check valves

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Abstract

We present a micropump with a simple planar design featuring compliant in-contact check valves in a single layer, which allows for a simple structure and easy system integration. The micropump, based on poly(dimethylsiloxane) (PDMS), primarily consists of a pneumatically driven thin membrane, a pump chamber, and two in-plane check valves. The pair of check valves is based on an in-contact flap–stopper configuration and is able to minimize leakage flow, greatly enhancing the reliability and performance of the micropump. Systematic experimental characterization of the micropump has been performed in terms of the frequency response of the pumping flow rate with respect to factors including device geometry (e.g. chamber height) and operating parameters (e.g. pneumatic driving pressure and backpressure). The results demonstrate that this micropump is capable of reliably generating a maximum flow rate of $41 \mu\text{L min}^{-1}$ and operating against a high backpressure of up to 25 kPa. In addition, a lumped-parameter theoretical model for the planar micropump is also developed for accurate analysis of the device behavior. These results demonstrate the capability of this micropump for diverse applications in lab-on-a-chip systems.

1. Introduction

Lab-on-a-chip microsystems offer significant advantages such as miniaturization, physical and functional integration and automation, and have broad applications in biology, medicine, environmental monitoring and defense [1–3]. In these systems, easily integrated micropumps capable of precise and safe transportation of minute volumes of reagents have become crucial components. Generally, micropumps can be divided into two categories: mechanical pumps and non-mechanical pumps. Non-mechanical pumps [4, 5] typically afford simple designs and can be highly miniaturized, but in general they strongly depend on fluid properties and thus are often incompatible with biochemical protocols. In contrast, mechanical micropumps are not limited by the properties of liquids, and provide a more versatile pumping solution.

The earliest mechanical micropumps were usually based on silicon micromachining techniques [6–10], which required rather complicated fabrication processes and resulted in high fabrication costs. Alternatively, polymer materials, such as polyimide [11], polyester [12], parylene [13], silicone rubber [14] and poly(dimethylsiloxane) (PDMS) [15–17] have also been used to fabricate micropump devices. Polymer materials, particularly PDMS, have the significant advantages of low cost, high compliance, ease of fabrication and excellent biocompatibility over semiconductor materials. However, current PDMS micropumps are relatively complicated since they are typically based on non-planar hybrid structures that usually encompass a multilayered check valve configuration. More importantly,

micropumps with out-of-plane check valve structures prove difficult to integrate with other planar microfluidic modules and sensing components, and thus have limited application in lab-on-a-chip systems. Recently, PDMS valveless micropumps based on diffuser/nozzle structures [18–20] have been developed with a simple planar configuration and a relatively easy fabrication process. However, micropumps of this type are in general affected by the low diodicity of the diffuser/nozzle elements (i.e. the ratio of the forward flow to reverse flow rates). Overpressure at the outlet causes reverse flow, and hence potential cross contamination between the upstream and downstream fluids [21].

To date, micropumps with simple in-plane check valves and reliable microfluidic handling have been scarce. A planar micropump with no-moving-part (NMP) check valves [22] capable of transporting particle-loaded fluids such as blood samples was reported; however, it suffered from the limited effectiveness of the NMP valve. Another planar micropump utilizing thermopneumatic actuation and in-plane flap valves [23] used a ‘hole-in-the-wall’ process to fabricate valves that significantly reduced the leakage. However, the fabrication process used complicated silicon micromachining methods, and thus its usefulness was limited. Recently, a polymer planar micropump based on two *in situ* photopolymerized check valves [24] demonstrated relatively good mechanical performance and biocompatibility.

Aiming to address these limitations, this paper presents a planar PDMS micropump featuring compliance-based in-contact check valves which minimize the leakage of reverse flow for reliable operation. Based on the recent finding that the planar in-contact check valve with a flap–stopper configuration is able to limit the leakage flow with a high diodicity (i.e. the ratio of the forward flow to reverse flow rates) of up to 10^5 [25], we integrate such check valves, a pump chamber and a pneumatically actuated PDMS membrane into a single micropump. This device offers a fully PDMS-based planar design which allows for easy fabrication and integration with other microfluidic components. Furthermore, although pneumatic actuation is chosen for demonstration in this paper, this planar micropump can be readily extended to other pumping principles, such as electromagnetic and piezoelectric actuation methods for various applications in lab-on-a-chip systems. Systematic characterization of this micropump device has been performed, including the frequency response of the pump flow rate with respect to various choices of device geometry (e.g. chamber height) and operating parameters (pneumatic driving pressure, back pressure). The results demonstrate that this micropump is highly effective and capable of continuously generating fluid flow. Furthermore, a lumped-parameter dynamic model considering the coupled interaction of the fluid flow and the membrane movement is also developed for accurate analysis of the micropump behavior. Validated by the experimental data, the model is demonstrated to be an efficient tool to generate optimal device designs.

2. Design and fabrication

2.1. Design

The planar micropump employs an actuated thin membrane in a pump chamber situated between two in-contact, minimized-leakage check valves that provide unidirectional fluid flow. As shown in figure 1, the micropump consists of five structural layers from bottom up: a plain PDMS or glass substrate, a thin PDMS functional layer in which through-openings define the check valves and pump chamber, a thin PDMS flexible membrane, a glass plate with a through-hole aligned to the pump chamber to define the actuation area on the thin membrane and a PDMS top layer with a channel to introduce the pneumatic driving pressure. For the nominal dimensions of the micropump, the radius and depth of the pump chamber are $2200\text{ }\mu\text{m}$ and $200\text{ }\mu\text{m}$, respectively. The check valves consist of a microfluidic channel ($400 \times 200\text{ }\mu\text{m}^2$ in width and height), a flap ($350 \times 30\text{ }\mu\text{m}^2$ in width and length) and

a stopper channel (the thinner channel section defined by the stopper of the check valve and $60 \times 500 \mu\text{m}^2$ in width and length). The thickness of the PDMS membrane is $150 \mu\text{m}$.

Note that each check valve in the micropump exploits a simple in-plane flap–stopper-based configuration, where the highly compliant flap is reversibly in contact with the stiff stopper, allowing virtually no leakage [25]. Under forward pressure (from left to right in figure 1), the flap is deflected away from the stopper, allowing flow passage with reasonable resistance (i.e. the ratio of the pressure drop to the flow rate). Under reverse pressure (from right to left), however, the flap remains in firm contact with the stopper, preventing flow passage. The in-contact relative flap–stopper position minimizes the reverse leakage flow.

With this simple planar minimized-leakage check valve to direct the fluid in the inlet and outlet, the micropump is able to achieve continuous fluid flow, while the PDMS membrane oscillation in the pump chamber generates two-mode pump cycles. During pumping mode, a pneumatic pressure is applied on the membrane, forcing the membrane to deflect downward and decreasing the volume of the pump chamber. This generates positive pressure, pushing the fluid out through the outlet valve while the inlet valve checks the reverse flow to the inlet. Next, the pneumatic pressure is removed or reversed, and the membrane moves upward, creating a negative pressure in the pump chamber. This opens the inlet valve and draws the fluid through the inlet to refill the pump chamber, while the outlet valve checks the backflow from the outlet. By periodically applying pneumatic pressure, this planar micropump is capable of continuously producing forward fluid flow.

2.2. Fabrication process and experimental setup

The PDMS top layer of the micropump was first fabricated using the PDMS replica molding technique [26], in which SU-8 photoresist (SU-8 2150, MicroChem Corp.) was used to fabricate the molding master. Next, a clamp-assisted replica molding technique was exploited to obtain the through-opening thin PDMS functional layer (figure 2(a)), a detailed description of which is provided elsewhere [27]. Briefly, the technique used a transparency film to facilitate the demolding of the PDMS functional layer from the SU-8 mold, while the thin functional layer could be removed from the mold along with the transparency film after curing, maintaining its dimensional stability and flatness. Then, by carefully peeling off the PDMS layer from the transparency film, a desired through-opening PDMS functional layer with minimum distortion was obtained, with a small designed gap existing between the flap and stopper (figure 2(b)). Only restricted at one side of the channel, the highly compliant flap tended to deflect and adhere to the stopper spontaneously (figure 2(c)). Alternatively, using methanol as a surfactant, the flap was more efficiently drawn toward and remained in contact with the stopper [25]. The flap–stopper contact persisted because of the spontaneous adhesion between the surfaces (figure 2(d)).

In parallel, a PDMS prepolymer with a curing agent at a weight ratio of 10:1 was spin-coated on a flat, transparent film at 500 rpm and cured for 1 h at 70°C to obtain an approximately $150 \mu\text{m}$ thick PDMS membrane. Following an oxygen plasma treatment by using a reactive ion etching (RIE) system (O_2 gas flow, 180 mTorr chamber pressure, 60 W power, 10 s exposure time), the five individual structural layers were bonded together in the bottom-up sequence shown in figure 1. The micropump device was finally interfaced at the inlets and outlets with Tygon PVC tubing and sealed with epoxy glue (figure 2(e)).

The experimental setup of the micropump is illustrated in figure 3. Pressurized nitrogen was used to actuate the PDMS membrane deflection. The applied pneumatic pressure was controlled by a waveform generator (33220A, Agilent) through a pressure regulator (ITV 2030, SMC). Sine waveform voltages with various amplitudes and frequencies were applied to the tested device, where different voltage amplitudes were used to regulate the actuated

pneumatic pressure amplitudes used in the test, which were 30, 45 and 60 kPa. During testing, de-ionized water was used as a sample liquid for the characterization of the check valves and micropumps. The flow rate was averaged (calculated by measuring the moving speed of the water front in the outlet tubing). In backpressure-related micropump tests, a pressure difference was obtained by placing the inlet and outlet tubes at a different height H , and could be calculated as $\Delta p = \rho g H$, where ρ is the density of water (10^3 kg m^{-3}), and g is gravitational acceleration.

3. Lumped-parameter model

To further understand the dynamic behavior of the planar micropump, a lumped-parameter model considering the coupled interaction of the liquid flow and the membrane movement is developed. The schematic of the planar micropump model is shown in figure 4. The inlet and outlet valves are represented by two unidirectional diodes with the ability to completely eliminate the reverse flow. The pressure and the flow rate at the entrance of the pump chamber are p_1 and Q_1 , respectively; those at the exit are p_2 and Q_2 , respectively, while the pressures at the micropump inlet and outlet are p_{in} and p_{out} . The pump chamber height in the absence of membrane deflection is h .

In general, the vibration of the membrane is influenced by the coupled interaction of the driving force, the hydrodynamic force of the sample liquid and the membrane inertia. As the pump usually operates at frequencies ($< 20 \text{ Hz}$) much lower than the natural frequency of the membrane (estimated to be $\sim 8 \text{ kHz}$), the inertia of the membrane can be ignored. While the actual membrane deflection varies continuously with its location in the pump chamber, for a lumped-parameter representation the membrane can be further approximated as a rigid plate attached to an elastic spring with the spring constant K and displacement δ (figure 4).

The pump chamber height h is small compared with the chamber radius a (i.e. $h/a \ll 1$), so the fluid motion in the pump chamber can be represented by the use of lubrication theory [28]. Thus, a lumped-parameter model for the planar check valve micropump can be represented using the following equations:

$$K\delta = F_{\text{pneu}} + F_{\text{hydro}}, \quad (1)$$

where $K\delta$ represents the spring force on the membrane with K being the spring constant defined based on the average membrane deflection. The pneumatic actuation force F_{pneu} is given by

$$F_{\text{pneu}} = -Ap_{\text{max}}(1 - \cos\omega t), \quad (2)$$

where t is time, $\omega = 2\pi f$ with f being the actuation frequency, and p_{max} is the pneumatic pressure amplitude, and $A = \pi a^2$ is the pump chamber area. The hydrodynamic force F_{hydro} [29] takes the form

$$F_{\text{hydro}} = -\frac{\mu A^2}{(h+\delta)^3} \dot{\delta} + \frac{1}{2}(p_1 + p_2)A, \quad (3)$$

where the first term accounts for the squeeze-film damping effects caused by the interaction of the membrane motion and viscous fluid flow, where μ is the liquid viscosity, and the second term denotes the hydrostatic effects of the entrance and exit pressures in the pump chamber.

The flow rates at the entrance and exit of the pump chamber are given by

$$Q_1 = \pm \frac{1}{2} A \left(\delta + \frac{(h+\delta)^3}{12\mu} (p_1 - p_2) \right), \quad (4)$$

which can be obtained by the integration of the lubrication theory solution over the cross section. Additionally, the behavior of the check valves is represented as follows:

$$Q_i = \begin{cases} C_i \Delta p_i & \text{if } \Delta p_i > \Delta p_{\text{crit}} \\ 0 & \text{if otherwise,} \end{cases} \quad (5)$$

where C_i is the forward flow coefficient for the upstream ($i = 1$) and downstream ($i = 2$) check valves, and $\Delta p_1 = p_{\text{in}} - p_1 - 12\mu L_{\text{in}} Q_1 / (WH^3)$ and $\Delta p_2 = p_2 - p_{\text{out}} - 12\mu L_{\text{out}} Q_2 / (WH^3)$ are the pressure drops across the check valves and Δp_{crit} is the cracking pressure (i.e. the critical pressure required to open the check valves). Here the check valve pressure drops reflect the compensation for the pressure drops over the inlet and outlet channels, which have lengths L_{in} and L_{out} , respectively, with uniform cross-sectional width W and height H .

4. Results and discussion

We have systematically characterized the micropump system. The investigation started with the characterization of the planar in-contact check valves, verifying its function as a unidirectional valve with virtually no leakage. We then systematically studied the planar micropump characteristics by varying the pneumatic driving pressure, pump geometry and backpressure. The dynamic model described in the previous section is also applied to the device to simulate its performance using MATLAB (Mathworks Inc.) and compared with the experimental results. All the measurements were performed five times with error bars representing the standard deviation.

4.1. In-contact check valve characterization

The flow characteristics of the valve were investigated (figure 5) by studying the dependence of the flow rate on the applied static pressures within a 30 kPa range across the valve. A detailed description of the check valve experimental setup can be found in [25]. It can be seen that the forward flow rate first remained nearly negligible despite the increase of the pressure until a pressure of approximately 2 kPa. This phenomenon was most likely caused by the spontaneous adhesion between the flap and the stopper, resulting in a critical pressure of 2 kPa for opening the valve. The flow rate then increased almost linearly with the applied pressure, indicating that the forward flow resistance of the valve was not influenced significantly by pressures within the range of 2–30 kPa. A flow rate of $180 \mu\text{L min}^{-1}$ was achieved at a pressure of 30 kPa and larger flow rates were expected when the pressure further increased. In addition, this linear characteristic of the forward flow indicates that the forward flow coefficient, C_i (equation (5)), could be experimentally obtained as approximately $C_i = 6.70 \mu\text{L (min kPa)}^{-1}$ by measuring the ratio of the forward flow rate to the pressure drops. Thus, the forward fluidic resistance, as the reciprocal of C_i , was $0.15 \text{ kPa } (\mu\text{L min}^{-1})^{-1}$. For the reverse flow, we note that virtually no flow was detected throughout during the observation of at least half an hour, suggesting that this check valve configuration is capable of almost completely eliminating leakage.

4.2. Micropump characterization

We have further systematically characterized the micropump with respect to various parameters such as the pneumatic driving pressure, the pump's geometry, i.e. the pump chamber height and backpressure. Using the dynamic model we developed, we have also numerically simulated these experiments and compared the numerical results with the experimental data. For simplicity, the microchannels and check valves in the inlet and outlet were assumed to have identical dimensions with $C_i = 6.70 \mu\text{L} (\text{min kPa})^{-1}$ as determined above. Also, both the inlet and outlet pressures, p_{in} and p_{out} , were set to zero when the micropump device was tested at zero backpressure. Furthermore, a consistent set of nominal operation parameters were used throughout the tests, which include a pneumatic driving pressure of 45 kPa, a deflected membrane radius of 2.2 mm and a pump chamber height of 200 μm .

The frequency response of the micropump flow rate at varied pneumatic pressures was first studied as shown in figure 6 at zero backpressure. It can be seen that the tested pumping flow rates show a consistent trend with the driving frequency at different pneumatic pressures. Specifically, the flow rate initially increased with the frequency until it reached a peak value at a frequency of 2 Hz due to the increasing membrane vibration frequency. The pumping flow rate then steadily decreased as the frequency further increased due to the fact that the membrane vibration cannot synchronize with the pneumatic actuation at higher frequencies, which accordingly resulted in reduced membrane stroke volume. Additionally, a nonlinear relationship between the pumping flow rate and the frequency can be observed, which could have resulted from the complex nonlinear interaction between the membrane deflection, the pneumatic driving pressure and the hydrodynamic forces present in the fluid of the pump chamber. Finally, the pumping flow rate was observed to increase with the pneumatic pressure amplitude. The maximum flow rates of $19.3 \mu\text{L min}^{-1}$, $25.8 \mu\text{L min}^{-1}$ and $41.0 \mu\text{L min}^{-1}$ were obtained under the pneumatic pressure amplitudes of 30 kPa, 45 kPa and 60 kPa, respectively. This is because the higher driving pressures cause greater displacements of the membrane, and thus result in larger stroke volumes in the pump chamber.

Results from simulation are also compared with the experimental data in figure 6. It can be seen that the simulation results agree well with the experimental data within relative errors of 6.8%, 8.3% and 7.6%, respectively, for driving pressures of 30 kPa, 45 kPa and 60 kPa. The error, though small, may result from the uncertainty in the material properties of the PDMS. For example, Young's modulus of PDMS can vary from 50 kPa to 4 MPa depending on process conditions [30]. The measurement error of the geometric dimensions of the micropump device may also contribute to the simulation error. These simulation results confirm that the lumped-parameter model is a promising tool for gaining quantitative insight into the dynamic performance of the planar micropump.

The effect of the pump chamber height on the frequency-dependent pumping flow rate was investigated next (figure 7). The test results show similar trends in terms of the frequency response as in figure 6, which is that the pumping flow rate first increases with the driving frequency and then decreases, exhibiting a maximum at an optimal frequency of approximately 2 Hz. Additionally, an increase of the pumping flow rate with the pump chamber height was also observed, possibly due to the reduced squeeze-film damping effects on the membrane vibration. Compared with the test data, simulation results show relative errors of 11% and 8.3%, respectively, for pump chamber heights of 170 μm and 200 μm , again showing good agreement with the experimental results.

Finally, we explored the relationship of the pumping flow rate with the backpressure at different driving frequencies (figure 8). It can be observed that the pumping flow rate has an

almost linear relationship with the backpressure, which is consistent with the linear relationship of the forward flow rate with the applied pressure for the check valves. At zero flow rates, a maximum backpressure of 25.0 kPa was achieved at the optimal pneumatic driving frequency of 2 Hz, while backpressures of 20.5 kPa, 17.7 kPa and 13.6 kPa were obtained at frequencies of 3 Hz, 4 Hz and 5 Hz, respectively. This decrease of the backpressure with the driving frequency is most likely a result of the reduced membrane stroke volume at frequencies outside the optimal pumping frequency as previously mentioned. Note that the backpressure performance of this micropump is better than the current reported polymeric pneumatically based micropumps [14, 31, 32], in which the largest reported backpressure was approximately 6 kPa. The high backpressure operation of this planar micropump can be explained by the use of the reliable in-contact minimized-leakage check valves capable of sustaining high flow pressure.

5. Conclusions

This paper presents a planar in-contact check valve-based PDMS micropump, in which a pair of check valves utilizing a planar stopper-flap configuration allows for the incorporation of the pump chamber into a single layer. This results in a simple pump structure and permits easy system integration. With the unidirectional flow regulation realized by the pair of check valves, the device exhibits excellent performance, demonstrating its ability to drive fluid continuously and efficiently by using pneumatic actuation on the membrane for demonstration. This micropump was fabricated using simple PDMS replica and bonding techniques. In particular, the critical planar check valves were fabricated using a clamp-based replica molding technique and then manipulated to establish the flap–stopper contact.

We first characterized the check valves in this micropump device. Virtually no reverse flow was detected during the test, suggesting that these check valves are capable of almost completely eliminating the leakage flow. We then further systematically characterized the micropump device under varying parameters, including the applied pneumatic pressure, the varied pump geometry (i.e. the pump chamber height) and backpressure. Experimental results demonstrate that this micropump is able to efficiently generate a maximum flow rate of $41 \mu\text{L min}^{-1}$ and reliably operate against high backpressures up to 25 kPa. Furthermore, in order to analyze the dynamic operation of the planar pump, a lumped-parameter model considering the coupled interaction of the fluid flow and the membrane vibration is also developed and validated by the agreement of the theoretical prediction with the experimental data on the micropump performance. With its simple planar design and reliable operation, this in-plane micropump can be readily integrated with various actuation methods and potentially applied in various lab-on-a-chip systems, such as automated implantable drug delivery systems.

Acknowledgments

We gratefully acknowledge financial support from the National Science Foundation (grant no DBI-0650020 and CBET-0854030), the National Institutes of Health (grant no RR025816-01A1) and the Columbia Ophthalmology Department. JN also appreciates the support from Shanghai Leading Academic Discipline Project (project no B602) a National Scholarship from the China Scholarship Council.

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Figure 1. Schematic of the micropump design integrating two planar in-contact check valves and pump chamber into one single layer.

Figure 2.

The micropump fabrication process. (a) Fabrication of the PDMS functional layer by use of the clamping method. (b) Retrieval of the PDMS functional layer from the SU-8 mold (a gap exists between the flap and stopper). (c) Manipulation of the flap to make it in contact with the stopper. (d) Image of a check valve with in-contact flap and stopper. (e) Image of a packaged micropump device.

Figure 3.
Schematic of the micropump experimental setup.

Figure 4.
Schematic of the dynamic model for the planar micropump.

Figure 5. Characterization of the in-contact check valve in terms of its forward and backward flow rates as a function of the pressure drops. The leakage indicated by the backward flow at reverse pressures is negligible.

Figure 6. Measured and simulated pumping flow rate as a function of the driving frequency for varying pneumatic pressures.

Figure 7. Measured and simulated pumping flow rate as a function of the driving frequency for different pump-chamber heights.

Figure 8.
Measured pumping flow rate as a function of the backpressures at varying frequencies.