SINGLE-LAYER MICROFLUIDIC CURRENT SOURCE
VIA OPTOFLUIDIC LITHOGRAPHY

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ABSTRACT
This work marks the first use of in-situ photopolymerization to create single-layer microfluidic devices which serve as ultra-low Reynolds Number (Re) current sources to regulate fluid flow rate independent of operating pressures. Autonomous fluidic components are an emerging aspect of micro/nanofluidic circuits and applications; however, many existing fluidic applications require specific pressure and/or flow rate conditions to perform optimally, and many require complex and expensive fabrication procedures. Here we introduce single-layer microfluidic system which utilize a spring and piston system – fabricated in situ via optofluidic lithography – to passively constrain fluid flow rate to a value independent of operating pressure. Experimental results revealed controlled flow rates of 29.2 ± 0.8 μl/min (from P = 50-100 mbar) and a maximum small-signal resistivity of 141.1 mbar-min/μl, which represents the highest performance for a low-pressure microfluidic current source.

INTRODUCTION

Microfluidic Current Control Mechanisms
Microfluidic systems offer numerous benefits for biochemical applications, such as low cost, rapid reaction times, low reagent volumes, and the ability to mix, transport, and/or array chemical processes involving beads, cells, or other biomolecules [1-4]. The ability to dynamically control microfluidic systems in a self-regulating manner is therefore critical for the long-term development of lab-on-a-chip (LOC) technology; at present, the majority of integrated microfluidic systems require substantial external regulation during device operation [5-8]. In recent years, research has shifted toward developing new, self-regulating microfluidic technologies [9-15]. However, producing dynamic microfluidic control systems in an efficient, cost-effective manner has remained elusive and the focus of widespread research efforts. For example, optofluidic “domino diodes” are capable of passive flow rectification only [16]. While many biological experiments require precise reagent flow rates unavailable through such passive controls, building devices that function as hydraulic analogs to electronic “current sources” has proved a considerable challenge. Previously microfluidic current sources have required three PDMS layers and pressure > 1 ATM for current regulation [17], or have been single-layer devices PDMS devices that only function at high Reynolds number (i.e., Re > 80) [18].

Optofluidic Lithography for Microfluidic Circuitry
Previous works have demonstrated optofluidic lithography in Teflon-based systems do fabricate freely moving pistons and check-valves in two- to four-layer microfluidic systems. However, the efficacy of such components remains limited due to the high pressures required for operation [19-20]. Maskless optofluidic lithography has been used to create railed systems which allow free movement at low pressure [21-23]. Single-layer optofluidic regulatory devices have been developed which require changes in fluid properties, such as Temperature and pH, but do not respond to flow parameters (e.g., pressure and flow velocity) [24].

In earlier works, we presented microfluidic “gain” valves – which can use low pressures to close valves at higher pressures – using free-floating microstructures (fabricated via optofluidic lithography) [25]. However, these valves exhibit motion based only on flow pressure and do not respond directly to fluidic current, and thus lack some of the flexibility found in analogous electronic devices. To expand the capabilities of these optofluidic circuit elements, we introduce a hybrid current source utilizing optofluidic lithography to fabricate a single-layer microfluidic system with a moving piston element which can passively control fluid flow rates independent of operating pressures.

Figure 1: Illustrations of the current source concept. (a) The device consists of a spring-mounted piston which slides within a microchannel according to the fluid forces exerted on it. As pressure increases, the increased flow rate forces the piston into the high-resistivity portion of the microchannel, increasing the hydrodynamic resistance of the system. The increased resistance then lowers the flow rate in a negative feedback loop, causing the flow rate to auto-stabilize. (b) Behavior of an ideal current source. Once the pressure passes a critical threshold value, the flow rate remains constant regardless of the pressure applied to the device.
CONCEPT

Figure 1 a-c highlights the concepts behind the single-layer microfluidic current source. The device consists of a freely-moving piston mounted to a spring which moves in and out of a narrow microchannel according to pressure and shear hydrodynamic forces exerted on it. The motion of the piston enables a negative feedback system within the current source; as pressure and flow increase, the piston’s motion increase the resistance of the system, which then to stabilizes the flow rate. According to Poiseuille’s Law for low Re systems, flow rate \((Q)\) is proportional to pressure drop \((\Delta P)\) divided by hydrodynamic resistance \((R)\). In general, \(R\) is proportional to the length of a channel and its resistivity \((r)\), a factor which is highest where the piston is located in the narrow part of the channel. For this system, resistance is a roughly linear function the piston’s position within the narrow channel, approximated by

\[
R(x) = R_0 + rx
\]

where \(R_0\) is a constant parasitic resistance which does not depend on channel position, \(r\) is piston resistivity, and \(x\) is the length of the piston contained within the narrow channel. By balancing spring restoring force \((k x)\) with pressure force \((\Delta P)\) and shear force \((\sigma)\), we can determine the pressure and flow rate as parametric functions of \(x\):

\[
(P(x), Q(x)) = \left( \frac{k x (R_0 + rx)}{\sigma + rx}, \frac{k x}{\sigma + rx} \right)
\]

These equations demonstrate that \(Q\) tends toward a flat value \(k/r\) while the spring is free to extend, and resistance reaches a constant value \(R_{\text{max}} = R_0 + rL\) when the piston reaches its maximum deformation length \(L\).

![Figure 2: The optofluidic fabrication process. (a) A photocurable liquid – PEGDA with 1% photoinitiator – is loaded into a PDMS microchannel. (b) A photomask is placed in contact with the device which is then (c) exposed to UV light, causing PEGDA to photopolymerize. The piston does not stick to the PDMS due to contraction of the cured PEGDA and oxygen layers in to the PDMS. (d) The mask is removed and the remaining photocurable liquid is drained and replaced by non-photocurable PEGDA.](image)

FABRICATION

The bulk microfluidic system was fabricated using standard soft lithography processes, as described in previous work [16,25]. Briefly, micromolded poly (dimethyl-siloxane) (PDMS), with microchannel heights of 110 μm, was thermally bonded to glass slides coated with 80 μm of PDMS (via PDMS-PDMS bonding techniques) to prevent the optofluidic components from sticking to the glass slides.

Figure 2 shows conceptual illustrations of the optofluidic lithography process to create freely moving pistons within a PDMS microchannel. First, a solution of poly(ethylene glycol) diacylate (PEGDA) with 1% photoinitiator (2,2-dimethoxy-2-phenylacetophenone) was loaded into the device (Fig. 2a). A photomask was placed in contact with the device and aligned (Fig. 2b). The device was then exposed to UV, causing the PEGDA to photopolymerize. Due to an oxygen layer in the PDMS, the cured PEGDA does not bond to the PDMS, and due to contraction of the PEGDA during exposure, a thin gap forms, which allows for lubricated piston motion. Figure 3 shows optical images of the fabricated gain device, illustrating the piston in both ‘open’ and ‘closed’ positions.

![Figure 3: Images of fabricated current source device in (a) relaxed spring configuration (under no flow conditions) and (b) extended spring configuration (under forward fluid flow). Micrographs depict images stitched together with photo-editing software. Scale bar = 500 μm](image)

RESULTS AND DISCUSSION

Simulations

Figures 4 shows the results of three-dimensional COMSOL Multiphysics simulations of the microfluidic current source. In Figure 4a, fluid pressure drop across the piston occurs primarily when the piston is located within the narrow channel. This is due to the much higher resistivity in this region, as shown by the numerical results in Figure 4b. Note that the resistivity of the narrow channel plus piston is more than 60x greater than the resistivities of the other sections of the current source, demonstrating that the resistance of this device becomes dominated by this region as the piston moves farther into the channel.

Figure 4c shows an important limitation for current sources using an asymmetric spring design: lateral distortion forces arise even when the spring is stretched in a purely axial direction. These lateral forces in turn push the piston against the channel wall, causing drag forces that prevent the piston from withdrawing smoothly from the channel creating the hysteresis effect seen in Figure 5.
Device hysteresis can be limited in several ways. First, the optofluidic exposure time can be reduced, which prevents overexposure of the piston sides, reducing overall friction. Second, the process can be controlled to prevent movement of the microfluidic device during exposure. Third, a symmetric spring design can be used to prevent the lateral distortions entirely – this method is sub-optimal because *ceteris paribus*, a symmetric spring has a 16x larger spring constant than does an asymmetric spring, resulting in devices that are too stiff to move. Finally, micropillars within the channels can act to constrain the piston to axial motion within the channel.

**Flow Rate-Pressure Experiments**

To investigate the overall performance of the current source, experiments were performed where the current source was subjected to a varying pressure sweep at its input and output terminals. Pressure differential applied and resultant flow rates were measured (*via Fluigent MFCS and Flowell sensors*) for the duration of the sweeps and the resultant flow rate-pressure (QP) curves were fitted using the aforementioned theoretical model and plotted.

In Figure 5, the QP curve for a single current source is compared with the linear response of an equivalent empty channel. During the forward sweep (red), flow rate levels off at a maximum value of 29.2 ± 0.8 μl/min (from \( P = 50-100 \) mbar) due to the action of the current source. When the piston reaches maximum extension, the device begins to act as an ordinary resistor, thereby limiting the device’s dynamic pressure range. This range can be extended primarily by increasing the length of the optofluidic piston.

An extreme hysteresis curve (green) shows that during the reverse pressure sweep, the piston remained partially extended. An empty channel (blue) yields a linear response, as is expected from the Poiseuille-flow resistive model.

In 5, the QP curves are plotted for current sources with differing spring amplitudes and spring stiffness. Note that as the spring constant increases, the current stabilization is not reached until higher pressures.

**CONCLUSION**

Autonomous, low-cost microfluidic circuit components offer significant potential to improve the functionality of lab-on-a-chip applications. In this work, we presented a single-layer system for implementing microfluidic current control, with fabrication *via in situ* optofluidic lithography. The methodology presented has several adaptations over previous methods; for example it implements the current source using a fabrication method that requires only a single layer of PDMS substrate and two photolithographic processing steps, has enabled controlled flow rates as low as 16 μl/min (*i.e.* for \( Re < 0.15 \)). The high pressure stability of these results suggest that the presented methodology could numerous Lab-on-a-Chip applications in situations where precise reagent delivery is essential.

Several adaptations of this method could be used to improve regulate fluid flows in a variety of conditions. For example, decreases piston width lowers the steady-state flow rate, and increasing piston length increases the operating range. For future work, we plan to investigate hysteresis-reduction techniques to make the device behave more like an ideal fluidic current source. Due to its ease of manufacture and low flow rates available, this microfluidic current source can be adapted to offer a simple, yet versatile, component for microfluidic and lab-on-a-chip circuits and systems.
ACKNOWLEDGEMENTS
The authors thank all the members of the Liwei Lin Laboratory, the Luke P. Lee Laboratory, and the Micro Mechanical Methods for Biology (M 3B) Laboratory Program. This research is supported in part by the DARPA N/MEMS program under the Micro/Nano Fluidics Fundamentals Focus (MF3) center and by the National Science Foundation.

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