Electrical microfluidic pressure gauge for elastomer microelectromechanical systems

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We report on an electrical microfluidic pressure gauge. A polydimethylsiloxane microvalve closes at a characteristic applied pressure determined by the material's properties and the valve's dimensions. Hence, when the same pressure is applied to all valves of a heterogeneous valve array, some valves close while others remain open. The state of the array is combined with knowledge of the respective characteristic closing pressures of the individual valves to yield an estimate of the applied pressure. The state of each valve is obtained by electrical measurements, since the electrical resistance of the respective underlying fluid-filled channel increases by at least two orders of magnitude as the valve closes and its insulating elastomer material interrupts the electrical circuit. The overall system functions as a pressure gauge with electrical readout. This device would be a critical component in active pressure-regulation loops in future integrated microfluidic systems. © *2007 American Institute of Physics*. [DOI: 10.1063/1.2801008]

INTRODUCTION

The past ten years have witnessed the rapid development of polydimethylsiloxane (PDMS) microfluidic technology from the simplest channels[1] to basic valves and pumps[2] to an extended family of devices integrated by the thousands within the same chip.[3] Many exciting specialized chips have emerged to address particular applications and offer new capabilities,[4] most recent examples being microfluidic immunoassays,[5] autoregulators,[6] and digital polymerase chain reaction (PCR) chips.[7,8]

After the development of elastomer microfluidics and the subsequent vigorous pursuit of its biological applications, it has become clear that the next generation of microfluidic devices would combine the fundamental technology with integrated electrical and optical systems for control and measurement. Such integration would offer new analytical and functional capabilities, as well as true overall device miniaturization.[4] A few steps in this direction have already been taken, producing a capacitance cytometer,[9] a thermal cycler,[10] and a spectrophotometer.[11] Moreover, such devices can be arranged as independent but interconnected modules functioning within a single chip.[12]

As part of this emerging wave of devices, we present an electrical elastomeric pressure gauge. It utilizes the basic effect that an electrolyte-filled microchannel experiences a large increase in electrical resistance when it is completely pinched off by a microfluidic valve whose material is an electrical insulator. This effect establishes an electrical means of reporting valve status. On the other hand, each valve remains open up to some characteristic pressure determined by its dimensions.[13,14] Therefore, the status of a particular valve offers an inequality for the magnitude of the applied pressure—either larger or smaller than the characteristic closing pressure of the valve. Then, a heterogeneous valve array produces a system of such inequalities that limit the magnitude of the applied pressure to a particular interval of values. The overall system acts as a pressure gauge. Its compatibility with standard PDMS microfluidic technology and electrical measurement and control makes it ideally suitable for integration in the future PDMS microelectromechanical systems (MEMS).

Other pressure-sensing schemes and devices have been developed: a microfluidic differential manometer,[15] a ce-

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ramic piezodielectric sensor,[16] a piezoelectric thin-film aluminum-nitride sensor,[17] a piezoelectric bimorph microcantilever sensor,[18] a piezoelectric gallium-arsenide sensor,[19] piezoelectric thin-film zinc-oxide sensors,[20] vacuum-sealed silicon sensors,[21] a metal oxide capacitor sensor,[22] and an optical sensor.[23] However, so far, our method seems the one most suitable for elastomeric microfluidic applications.

MATERIALS AND METHODS

Substrate fabrication. DWL66 direct mask writer (Heidelberg Instruments GmbH, 69126 Heidelberg, Germany) with a 20 mm head is used to create the respective pattern on glass slides coated with 90 nm nonoxidized chrome and 530 nm Az1500 photoresist (Telic Co., Valencia, CA 91355). The photoresist is developed for 20 s in Microposit MF-322 (Rohm and Haas Electronic Materials). The sample is washed with de-ionized water. The exposed chrome is etched for 3 min in CR-7 Chromium Photomask Etchant, Surfactant Added (Cyantek Corp.). The sample is washed with de-ionized water, blown dry, and stored as is. The substrate is washed with acetone (to remove the remaining protective photoresist), washed with ethanol, and blown dry immediately before assembly to PDMS.

Chip fabrication. A previous recipe[5] is employed to produce the molds and PDMS devices.

Experimental setup. A microfluidic microscopy station[5] was equipped with a digital pressure gauge from TIF Instruments Inc. and a multimeter MiniRangemaster from Extech Instruments.

Basic scheme. Multilayer soft lithography[2] is used to fabricate a PDMS microfluidic chip. A comblike array of 100-μm-wide channels in the lower layer is crossed with a single channel in the upper layer to produce an array of pushdown microfluidic valves of varying length (55–200 μm). This two-layer device is then bound to a metallized glass substrate in such a way that each prong of the comb is aligned with its own pair of electrical contacts, thereby completing the chip fabrication (Fig. 1).

A salty buffer (Tris 10 mM, NaCl 10 mM, $MgCl_2$ 0.1M, pH 8) is introduced in the lower (flow) layer by applying pressure and letting the preceding air escape through the polymer matrix. Filtered distilled water is similarly introduced in the upper (control) channel. The flow channels are left at atmospheric pressure, while pressure is applied to the control channel. For each value of the applied pressure, resistance is measured between the contacts of each prong. The status of the respective valve is also confirmed visually.

Ionic redistribution in response to the applied dc field tends to increase the resistance of the channel over time. While this is a small effect, the polarity of applied voltage is reversed at every pressure setting to decrease it further.

RESULTS AND DISCUSSION

As PDMS and glass are electrical insulators, it is a reasonable expectation that an uninterrupted microchannel filled with a salty buffer would drastically increase its electrical resistance when it is pinched off by a closed valve, since the insulating material of the valve would interrupt the electrical circuit of the channel. Simple preliminary experiments confirmed that prediction.

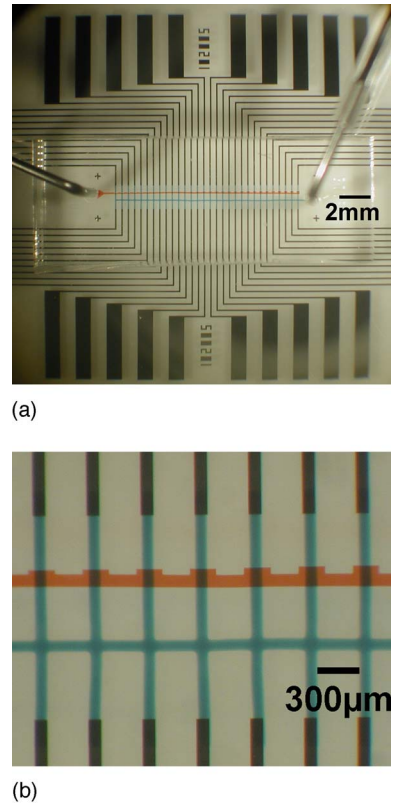

FIG. 1. (Color online) Device Architecture. A two-layer PDMS device is bound to a glass slide with chrome contacts in such a way that each pair of electrodes is matched to a short transverse segment of the lower-layer microchannel (blue dye, normally filled with salty buffer) (A). The upper-layer channel (red dye, normally filled with de-ionized water) forms an array of microfluidic pushdown valves of different dimensions (note valve length increasing from left to right in the picture) (B) and thus different characteristic closing pressures (Ref. 14).

The observed effect establishes a 1-1 correspondence between channel resistance (low/high) and valve status (open/closed). The mechanical properties of the valve establish another 1-1 correspondence, between the valve status (open/closed) and the applied pressure (insufficient/sufficient). The status of a *single* valve does not report the value of the applied pressure, but reports if the applied pressure is below or above the valve's characteristic pressure. In essence, a single valve status produces a single inequality.

Valves of different dimensions have different characteristic pressures.[14] Therefore a heterogeneous array of valves, acted upon by the same applied pressure, would report a set of inequalities, whose bounds are the different characteristic pressures. Thus the magnitude of the applied pressure would be limited to one of a set of intervals delineated by these bounds.

This analysis suggested the basic scheme described above. The scheme was implemented for applied pressures of 0.5 to 19 psi above atmospheric, in steps of 0.5 psi. The maximal value was set by the pressure at which all valves were closed. Further increase in the pressure would not change the result, while pressures in excess of 20 psi signifi-

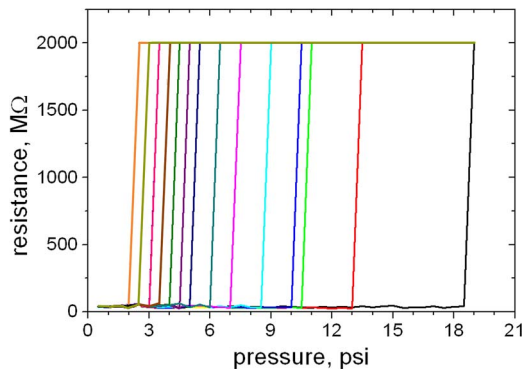

FIG. 2. (Color online) Device Function. Each valve closes when the applied pressure exceeds the valve's characteristic pressure. Thus the status of each valve determines applied pressure to an upper or lower bound. Then the status of a heterogeneous set of valves produces an interval estimate of the applied pressure. The electrical resistance of the electrolyte-filled channel segment under each valve [see Fig. 1(b)] increases drastically if and only if the valve is closed, so valve status is measured electrically. The overall device is utilized as a microfluidic pressure gauge with electrical readout. The experimentally measured resistances of the elements of the array are plotted as a function of applied pressure. (The upper limit of the multimeter scale is 2 GΩ, so the actual resistance of the fluidic seal is higher.)

cantly increase the risk of layer delamination. Each flow-channel prong had low resistance (26–64 MΩ) when its valve was open, while the resistance increased beyond the dynamic range of the multimeter (2 GΩ) when the respective valve closed completely. Figure 2 shows the resistance pattern versus applied pressure. Since in certain cases multiple valves closed at the same pressure, their resistance patterns overlapped and are represented by the same color.

This system can be used as a pressure gauge. For example, if all valves up to and including "blue" in Fig. 2 are in "high resistance" state, while all valves above it (light green, red, black) are in "low resistance state," then the applied pressure is between 10.5 and 11 psi.

The precision of the measurement is dependent on the spacing of curves in Fig. 2 and thus, on the spacing of characteristic pressures. In the demonstrated device, valve width, thickness, and material were kept constant, while valve length was varied between 55 and 200 μm. However, if that restriction is relaxed, many more values for characteristic pressure become available.[14] Then the pressure spacing can be shrunk accordingly, thereby improving the precision of the measurement.

Meticulous mappings of the phase space of closing pressures versus valve dimensions[14] showed that individual valve behavior is robust, reliable, and reproducible, thereby attesting to the robustness, reliability, and reproducibility of its derivative devices, such as the presented pressure gauge. Hence, the accuracy of the measurement is set by the accuracy of the digital gauge and the size of the pressure step used in calibration.

The response time of the overall system is governed by the mechanical actuation time of the valves and the time taken by the electrical measurements. Micropump characterization[24] sets the former at the scale of milliseconds, while the latter typically are orders of magnitude faster. Thus the overall sensor response time would be milliseconds. Such speeds are adequate both for isolated measurements and for continual monitoring in pressure regulation loops.

In the presented embodiment, the pressure gauge achieves large signal-to-noise in valve status measurements, through the use of a salty buffer. Hence, there are concerns that salt ions could diffuse through the elastomer, thereby adversely changing the ambient conditions in particular biochemical applications. However, biochemistry typically works with salinities at physiological level or above, which is essentially what we used here. Furthermore, in any particular case, the electrical measurement fluid can be set at the same salinity as the biochemical environment to prevent nonzero net ionic flux. Finally, potential losses of signal to noise in valve status measurements can be offset by optimizations of the geometry of the conducting channels.

Other pressure-sensing schemes and devices have been proposed and demonstrated. A microfluidic differential manometer has been used to study cells.[15] However, it does not produce an absolute measurement of the pressure, while the readout is done visually and thus cannot be straightforwardly automated or miniaturized.

Piezodielectric[16] and piezoelectric[17–20] sensors have convenient electrical readout and are miniaturizable. However, it may prove challenging to adapt these devices for inexpensive mass production, robust performance, and integration into particular systems, especially since many biological and biomedical applications already place tight constraints on the properties of the substrate, e.g., due to binding surface chemistries[5] and/or cell toxicity. In addition, some of the devices[18,20] have inadequate dynamic ranges for typical elastomer microfluidics.

Vacuum-sealed silicon sensors[21] utilize capacitance, piezoresistivity, or resonance measurements involving a silicon microdiaphragm. These sensors are compatible with mass production, are electrically readout, and are at the correct physical scale. However, the capacitance version requires voltages that are impractical for overall system miniaturization, the piezoresistive version is limited to silicon substrates, and the resonator versions are difficult to fabricate and are also limited to silicon substrates.

A pressure sensor,[22] involving an interdigitated capacitor with metal oxide dielectrics, has high sensitivity but excessive physical size and insufficient dynamic range. An optical sensor[23] measures the deflection of a silicon membrane by the intensity of light reflected from it; however, a light source, waveguides, and a photodetector are necessary, and so, overall system miniaturization is problematic.

By contrast, our device is simple, inexpensive, easy to fabricate, and straightforwardly integrable within elastomer chips. In principle, it is compatible with any substrate, since buffer-filled microchannels and vias[6] can be used as three-dimensional electrical connections to access the array. In addition, the system is reliable, robust, reproducible, and appropriately sized. It requires simple electric circuitry to function and offers adequate dynamic range, accuracy, and precision. So far, it seems to be the best overall solution for a pressure sensor integrated in elastomer microfluidics. The envisioned particular application is to monitor pressure and electrically report the result to logic circuitry that controls

the overall PDMS MEMS. This capability would be essential in completing the pressure-control loop when pressure generation (for valve actuation and fluid transport) is achieved within the integrated chip of the future.

CONCLUSIONS

An electrical elastomeric pressure sensor is described herein. Applied pressure determines the status of an array of PDMS microfluidic valves of different characteristic closing pressures. The electrical resistance of each valved channel is high when the respective valve is closed. Thus valve array status and, therefore, an interval estimate of the pressure, are reported electrically. The overall microdevice functions as a pressure gauge with electrical readout. Its compatibility with both electrical circuitry and the basic PDMS microfluidic technology make it a suitable subsystem for the next generation of integrated PDMS MEMS.

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