LIQUID-CORE, PIEZORESISTIVE, FULLY POLYMER-BASED PRESSURE SENSOR

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ABSTRACT

A new, resistive transducer principle suitable for pressure sensing and related applications is presented. The technique relies on pinching of an electrolyte-filled, microfluidic channel by pressure-induced deformation of an elastic membrane forming a part of the channel wall (liquid-core strain gauge). The resulting change in resistance along the channel is directly related to the membrane deflection. A prototype of a fully polymer-based pressure sensor has been fabricated in silicone elastomer as a proof of principle, with a sensitivity between 60 and 90 ppm resistance change per Pa (equivalent to ~4 Ω/Pa). This paper describes the underlying transducer principle, the device fabrication and an analytical model for designing transducer geometries suitable for a variety of applications.

Keywords: Pressure sensor, Pressure transducer, PDMS, Microchannel

INTRODUCTION

Pressure sensors provide an electrical signal equivalent to an external absolute or differential pressure applied to the device. The most common transducer principles used in this context are piezoresistive and capacitive measurements [1], though others have been used as well. Here, we present a fundamentally new approach to this sensing problem, using elastically deformable micro- and nanochannels filled with a liquid electrolyte as sensing element. The filled sensing channel serves as electrical resistor, its resistance being defined by the electrolyte resistivity and the channel geometry. If one wall of the channel, e.g. its cover, is composed of a thin, elastic membrane that deflects under externally applied pressure, the cross section of the channel is reduced and its electrical resistance is increased. This effect is basically similar to a solid strain gauge and therefore referred to as liquid-core strain gauge in this paper.
As the ratio between deflection and sensing channel height can be defined by design, highly sensitive devices can be fabricated, in principle being capable of detecting sub-nm membrane deflections. As an application example for this new transducer concept, a new fully polymer-based pressure sensor is presented.

THEORY

Fig. 1 illustrates the operating principle in a two-dimensional simplification. A shallow microchannel with a depth, *h*, and a cross section of A_0 is covered by a thin elastic membrane of thickness *t* and width *w*. The channel is filled with an electrolyte and the electrical resistance along the channel is monitored. When an external pressure, *p*, causes the membrane to deflect, the channel is pinched off and the cross section is reduced to $A_0 - A_d(p)$. As a result, the resistance along the channel increases by $A_0/A_0 - A_d(p)$, which can be monitored electrically. For successful operation, the microchannel has to have at least one open terminal to allow for displacement of the liquid during the deflection. Also, the flow resistance of the channel should be small to assure a fast dynamic response of the transducer.

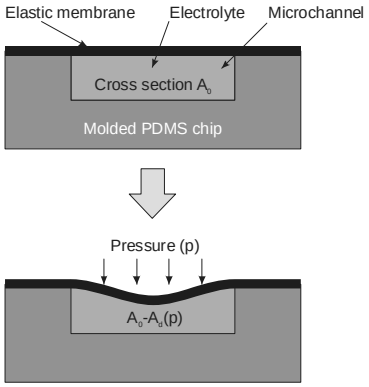


Fig. 1. The membrane deflects under pressure leading to a reduction of the channel cross section from A_0 to $A_0 - A_d(p)$. As a result, the electrical resistivity of the channel is increased by a factor of $A_0/A_0 - A_d(p)$.

For the simplified case of a square membrane with diameter *w*, the resistance along the channel filled with an electrolyte of resistivity ρ can be calculated by:

$$R(p) = \rho \int_{-w/2}^{w/2} \frac{1}{\int_{-w/2}^{w/2} h - d(x, y, p)\, dx}\, dy + R_{ch}$$

where $d(x,y,p)$ is the deflection of the membrane at a

the connection between membrane area and measurement electrodes. Figs. 2 and 3 illustrate the dependence of the transducer output on the membrane thickness and sensing channel depth. Thinner membranes increase the sensitivity of the device drastically as the deflection scales with t^{-3}. However, practical limits are imposed by the fabrication process as silicone-elastomer membranes thinner than 100 µm are difficult to handle and easily collapse during the assembly of the device. However, the sensitivity can also be increased by a reduction of the sensing channel depth as the cross-section change remains constant while the absolute cross section is reduced. Ultimately, it should be possible to detect deflections as small as a few Å with sensing channels around 1 µm deep.

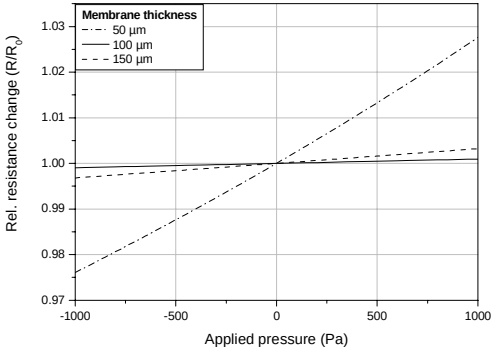


Fig. 2. Simulated relative resistance change as a function of membrane thickness (sensing channel depth: 10 µm, membrane diameter: 200 µm).

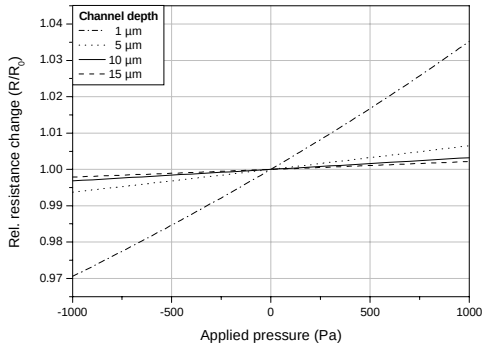
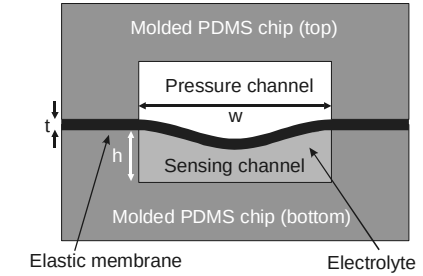


Fig. 3. Simulated relative resistance change as a function of the sensing channel depth (membrane thickness: 100 µm, membrane diameter: 200 µm).

DEVICE FABRICATION

A test device working as a pressure transducer has been fabricated by permanently laminating three layers of poly(dimethyl siloxane) (PDMS): a lower part with the sensing channel (200 µm wide, 12 µm deep), an upper part with a pressure channel (1 mm wide at the inlet, 200 µm wide in the sensing region, 12 µm deep) and a 115-µm-thick membrane in between (Fig. 4). Sensing and pressure channel form a cross to prevent collapsing of the membrane. Membranes thinner than 100 µm could not be successfully integrated for this channel geometry, yet, due to mechanical instability.



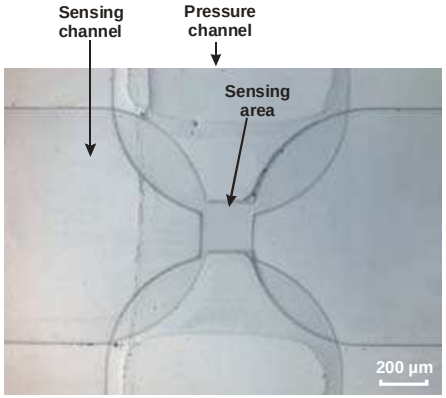


Fig. 4. Above: Schematic cross section of the device. The membrane is sandwiched between sensing channel (bottom) and a pressure channel (top) connected to a pressure controller. Below: Micrograph of the assembled device (view from bottom). The supported areas of the membrane were held in place by bonding while the membrane could deflect freely in the sensing area.

All PDMS components were fabricated by casting from a photopatterned mold [1]. Briefly, a 100-mm-diameter silicon wafer was coated with a 12-µm-thick layer of SU-8 2015 (Microchem, USA), exposed and developed to fabricate the mold for top and bottom parts of the device. Before casting, the mold surface was passivated in dimethyloctadecylchlorosilane dissolved in toluene to facilitate demolding. The elastomer components were cast using Sylgard 184 (Dow Corning, USA) with a 4-hour curing at 65 °C.

The PDMS membranes, the actual transducer elements, were fabricated by spin-coating on untreated, polished silicon wafers. After coating, the wafers remained on the level spin coater at room temperature for 15 min to reduce the edge bead of the viscous material, before they were transferred to a hotplate and cured as described before. Fig. 5 illustrates the experimentally determined relationship between membrane thickness (after curing) and spinning speed. If thicknesses below 50 µm are necessary, the PDMS precursor can be diluted in chloroform to reduce its viscosity [3].

Finally, the three PDMS components were surface-activated in an oxygen plasma at 100 W for 30 sec and immediately bonded together to yield a permanent seal. The alignment of the layers was done manually under an inspection microscope. During assembly, care must be taken to avoid collapsing of the thin sensing membrane due to excessive pressure applied on the stack.

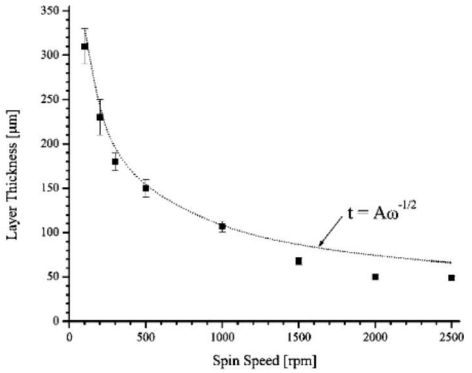


Fig. 5. Thickness of a spin-coated PDMS layer after curing. The constant A in the formula was experimentally determined to be A=3300 µm$\sqrt{\text{rpm}}$.

EXPERIMENTAL RESULTS

The sensing channel was filled with phosphate-buffered saline solution (PBS) at pH 7.4. The resistivity was experimentally determined to be 53.2 Ω·cm at

trodes due to electrolysis, the resistance was monitored in an alternating current (AC) mode using platinum wire electrodes inserted into the open reservoirs of the chip. The chip was filled with the electrolyte by applying a vacuum on one reservoir of the sensing channel due to the hydrophobicity of the material. The electrical measurements were performed using an HP4284A precision LCR meter operating at 500 Hz and 500 mV excitation.

The pressure channel was opened only on one side and connected to a Druck DPI 520 pressure controller via a syringe needle to apply the test pressure. With this setup, pressures up to 40 kPa could be applied to the device without detectable leakage at the chip connection. While this setup is suitable for short-term measurements, it is recommended to fill the pressure channel with a liquid, e.g. water or silicone oil, to prevent gas diffusion into the PDMS during longer experiments.

Fig. 6 compares measurement data for the resistance change of two devices with a simple analytical approximation for circular membranes. The pressure sensitivity, that is the resistance change per applied pressure, is 60 and 90 ppm/Pa for device 1 and 2, respectively. Given the resolution and measurement noise of the setup used here, the limit of detection is estimated to be 3 Pa at three times RMS noise. The difference in sensitivity between the two devices tested is assumed to be caused by imprecisions in the mold-making process, especially in terms of the sensing-channel height, and by alignment errors during assembly. This would also explain the slight difference in the absolute resistance of both devices at zero pressure, which were 44.3 and 48.7 kΩ.

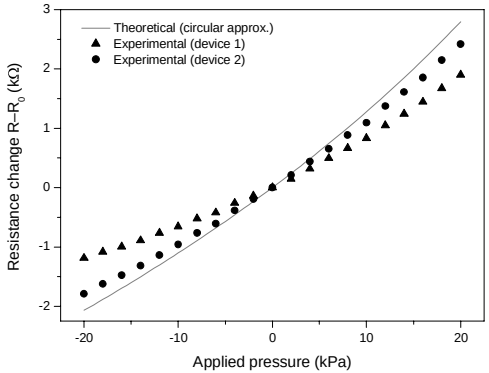


Fig. 6. Measured resistance change in the sensing channel under applied pressure between -20 and 20 kPa. The sensing channel was filled with phosphate-buffered saline solution at pH 7.4 with a resistivity of 53.2 Ω·cm.

CONCLUSION AND OUTLOOK

Resistive monitoring of a pressure-induced microchannel deformation has a high potential for a variety of sensing applications including strain and displacement measurements. The straightforward readout and simple, polymer-based fabrication process results in a robust, efficient sensing principle. The performance can be further enhanced by a Wheatstone-bridge scheme or by synchronous detection for read-out. The long-term stability of the transducers still needs to be assessed in detail, especially as evaporation of the electrolyte through the porous PDMS polymer might pose a problem [4]. Possibly, alternative electrolytes, such as gels, could be used to circumvent this issue.

References

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