33rd Annual International Conference of the IEEE EMBS
Boston, Massachusetts USA, August 30 - September 3, 2011

AN UNPOWERED, WIRELESS CONTACT LENS PRESSURE SENSOR FOR POINT-OF-CARE GLAUCOMA DIAGNOSIS

John Yan[1]
[1]University of California, Davis USA

ABSTRACT

The development of a wireless, microfluidic pressure sensor composed of Polydimethylsiloxane (PDMS) and dyed glycerol is presented for point-of-care glaucoma diagnosis. Design, fabrication and test results are presented.

Keywords: wireless sensors, pressure sensor, point-of-care, glaucoma

INTRODUCTION

It is estimated that 67 million people suffer from glaucoma worldwide [1]. Patients with glaucoma are considered "well controlled" if their mean intraocular pressure (IOP) is lower than 21 mm Hg [2]. Owing to a rapidly aging population, it is estimated that the number of open angle glaucoma cases will increase to 3.4 million in 2020 -- making it the second leading cause of blindness and the first leading cause of irreversible blindness in the U.S [3]. Since unregulated IOP can lead to irreversible blindness by pinching the optic nerve (see Figure 1), it is exceedingly importance to monitor this pressure and make low-cost, point-of-care diagnostic tools available. The development an unpowered, wireless sensor contact lens pressure for point-of-care glaucoma diagnosis is presented.

Various wireless techniques have been conceived to continuously monitor intraocular pressure [4] such as LC wireless techniques [5], soft contact lens with microfabricated strain gauges [6] and less conventional techniques such as spiral rotation of microbourdon tubing [7]. Recently, contact lens have been developed with an integrated amperometric glucose sensor [8] and integrated RF-powered, power-harvesting LED electronics [9]. While the presented work in [14] showed amazing promise, the complicated components, usage of exotic materials and complicated manufacturing process is likely to make the contact lens solution out of reach for disposable applications not to mention a host of potential health-issues related to the high RF power needed to activate the device. The 'smart contact-lens' is constructed out of biocompatible PDMS using simple, soft lithography processing. With such a device, we hope that glaucoma can be detected at an earlier stage and steps can be taken to prevent serious vision loss. Since the device is processed through simple, batch microfabrication techniques, large-scale manufacturing will be low-cost in terms of capital goods and consumable materials. In the paper the sensing principle of the sensor is discussed, followed by device fabrication. Test results are presented along with a concluding discussion.

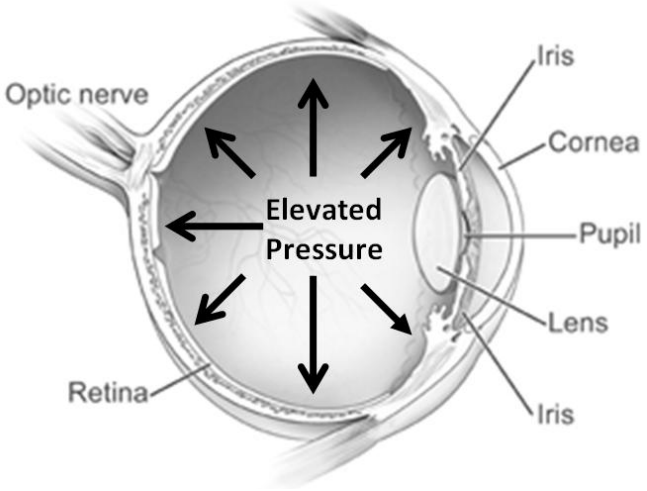

Figure 1 : Glaucoma is a result of build-up of intraocular pressure; figure adopted from [10]

SENSING PRINCIPLE

The dynamical IOP measure is based on Laplace's principle where the pressure inside the hydraulic chamber (P_t) is dynamically determined by loading pressure and measured by the built-in microfluidic pressure sensor. The sensing membrane deflects and measures the pressure difference (P_s) at each contact point through the individual surface sensing element. PDMS is used as the structural material to create the flexible, polymeric membrane and surface sensing for its elasticity and flexibility as well as excellent physical properties with a Young's modulus (E) of approximately 500kPa [11].

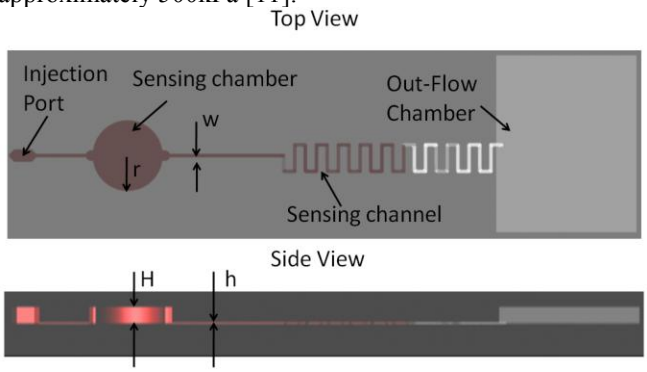

Figure 2. Schematic Illustration of Calibration Device

The device used to characterize the sensing principle is illustrated in Figure 2. It consists of a large, circular sensing chamber network with height H and radius r. This is followed by sensing channels of width, w and height h

which act as the sensing elements. As localized pressure is applied at the sensing chamber network of height H and radius r, a resulting strain is induced on the elastomer housing generating an internal pressure gradient in the microfluidic network. As the sensing chamber is compressed, fluid is outwardly displaced to the sensing channel. Due to the geometric difference between the large and tall sensing chamber and narrow and short sensing channel, mechanical displacement amplification occurs as a result of the conservation of mass. As stress is released from the sensing chamber, the elastomer recovery properties of PDMS create a negative pressure to withdraw fluid from the sensing channel.

The microfluidic network is carefully designed to prevent air bubble formation at the sensing chamber during the injection process. This is done by installing an injection port before the sensing chamber network. Laplace valves at the entrance and exit of the sensing chamber are designed to prevent bubble cavitation as the fluid flows from the fluidic interconnects to the sensing chamber. To minimize compressive effects of displacement of air in the sensing channel during fluidic displacement, a large out-flow chamber is designed to have a volume a thousand times larger than the sensing channel, acting as a pressure relief conduit. A long, rectangular straight sensing channel is used to characterize the sensing principle of the device since its laminar flow profile characteristics are well understood. By optically observing the magnitude of fluidic displace, the resulting pressure on the sensing chamber can be determined.

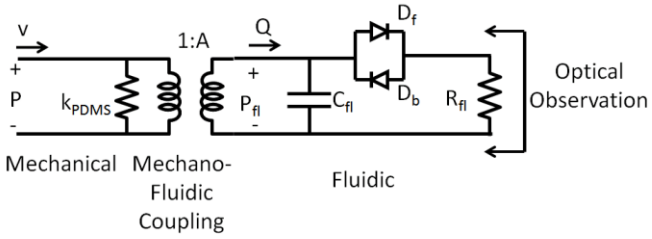

Figure 3. Equivalent Circuit Model of Transducer

The mechano-fluidic transduction can be modeled as a circuit network as shown in Figure 3. As a pressure is applied to the sensing chamber with a set velocity, a corresponding flow of fluid will result due to the change in the internal pressure displacement. A transformer is used to model the conversion from solid mechanical displacement of the elastomer and the fluidic displacement in the microfluidic network. The displacement amplification, A, is set by the geometry of the sensing chamber and sensing channel. The stiffness of the sensing chamber is:

$$k_{PDMS} = \frac{stress}{strain} \quad (1)$$

According to the strain-stress relationship, the change of the micro-chamber height (H) can be expressed as, $\Delta H/H=\sigma/E$, where E is Young's Modulus of PDMS elastomer, and, σ is the loading pressure. The displaced fluidic volume enters the incompressible microchannel, where the geometry (i.e., the cross-section area) determines the wetted length (Δl) of the colored fluid. Herein, the sensitivity (S) can be calculated as:

$$S = \frac{P}{\Delta l} = \frac{whE}{\pi r^2 H} \quad (2)$$

Another important characteristic measure of the sensor is the frequency response. The microfluidic sensing system can be modeled as a first-order linear circuit, in which the micro-chamber compliance C and the microchannel resistance R are:

$$C = \frac{\pi r^2 H}{E} \quad (3)$$

$$R = \frac{12\mu\Delta l}{wh^3} \quad (4)$$

where μ is the viscosity of the sensing fluid. D_f and D_b represent the effect of surface tension and the resulting difference in internal pressure that needs to be overcome before the fluid can displace in the sensing channel. This surface tension is set by the surface properties of the material.

To analyze the fluidic response, we assume that flow is dominated by pressure driven flow in the laminar region so the lubrication theory approximation can be used for the Naveir-Stokes equation. The cutoff frequency, f_c, is set by the fluidic resistance and capacitance [12] as follows:

$$f_c = \frac{1}{2\pi\tau_{fl}} = \frac{1}{2\pi RC} = \frac{1}{2\pi\left(\frac{12\mu\Delta l}{wh^3}\right)\left(\frac{\pi r^2}{E}\Delta H\right)} \quad (5)$$

The optical detection can be realized using a low power, infrared LED source and a corresponding photodiode. To characterize the sensitivity of the pressure sensor, we employed an off-the-shelf, digital microscope and ambient light. Or simpler, colored, non-evaporative fluid can be injected into the microfluidic network. A promising, low-cost candidate is glycerol. To accommodate the human cornea with an approximately a diameter of 7.8mm, the microfluidic sensing chambers and corresponding meandered sensing channels are designed on the peripheral of the contact lens at the sclera area.

FABRICATION

To realize such a polymeric pressure sensor, we use bottom-up micromachining, replica molding, oxygen plasma and fluidic injection. The strategy for material selection focuses on biocompatible materials which are optically transparent, highly mechanically deformable under applied pressures and utilize low-cost, microfabrication processing. Polydimethylsiloxane (PDMS) has well-characterized soft lithography processing [11, 13], mechanical soft elastomeric properties with a low mechanical loss tangent [11], and is electrically isolative, chemically inert, non-toxic, and optically transparent.

These properties make it an ideal candidate for housing the microfluidic network. The process flow is shown in Figure 4.

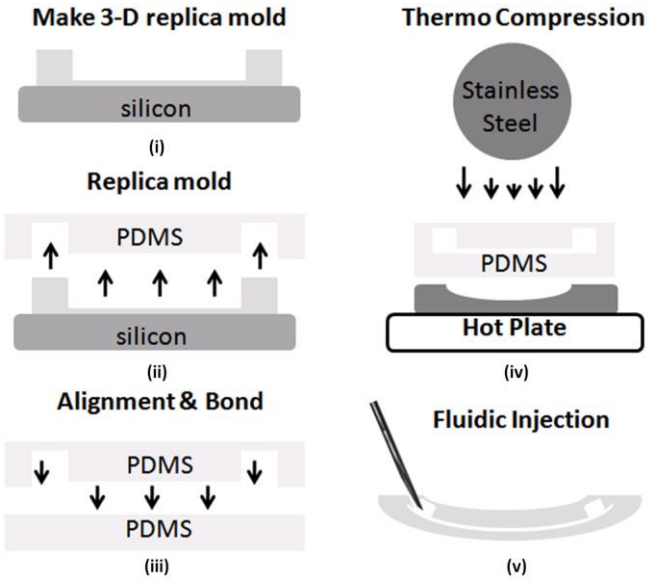

Figure 4. Fabrication Process (i) replica mold (ii) soft lithography (iii) oxygen plasma (iv) thermocompression (v) fluidic injection

The first step to realize the elastometric device is the 3-D replica mold. The replica mold consists of a two-step SU-8 (Microchem, Inc.) process on a silicon substrate. First, the sensing channel, drain channel, buffer channel and microfluidic interconnects are defined, with photolithography, at a height of 15µm using traditional SU-8 processing [8]. Next, the injection port, sensing chamber and out-flow chamber are defined to have a height of 200µm. A two-step soft-bake is used for the thick SU-8 to minimize the internal stress during the soft-bake and post-exposure bake process. Next, PDMS (10:1 base to agent) is mixed, degassed and poured onto the 3-D master mold to cure at 80 ^{o}C overnight. The cured PDMS mold is then exposed to oxygen plasma treatment to create radical –OH on the surface to initiate the covalent bonding onto another PDMS replica mold. The two bonded layers of PDMS are then thermocompressed in a heated platform of 300C for 5 minutes against two rigid stainless steel molds in the shape of a contact lens. Because of the porous nature of PDMS, the fluid must not evaporate under normal room temperature and pressure conditions. Glycerol was chosen, as opposed to DI or PBS, to satisfy the above property in addition to its low-cost and biocompatibility. A BD 30 ½ G needle is inserted into injection port of the elastomer housing and a controlled volume of dyed glycerol (100:1 glycerol to color dye) is infused into the microfluidic network from a glass syringe with a syringe pump at a calibrated flow rate.

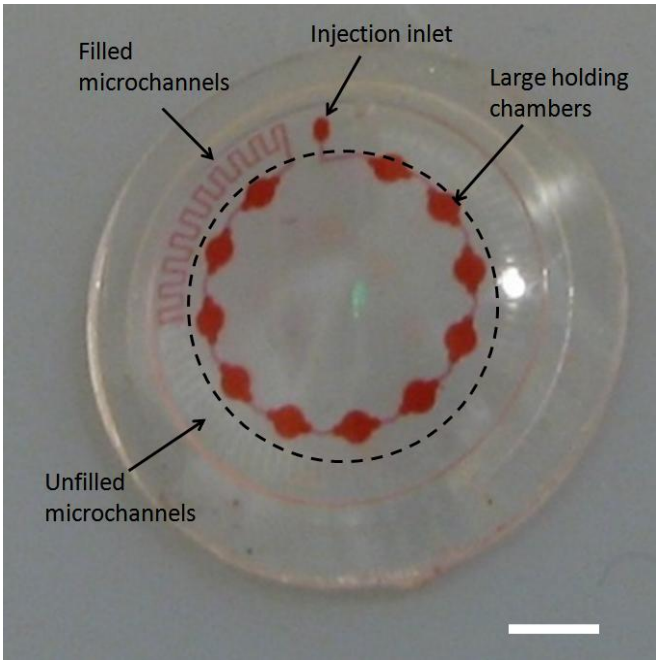

Figure 5. Optical photograph of the microfabricated device; scale bar is 5mm

TEST RESULTS

To test the device, a force gauge connected to a step motor is used to exert pressure on the sensing chamber. A digital microscope is used to record the fluidic displacement as the step control moves to exert pressure on the sensor. Replica molds with sensing channel widths of 20, 40, 80, 160µm and total length of 6mm are constructed and characterized. The test results are summarized in Figure 6. Table 1 summarizes the sensitivity and dynamic range results of the test structures. As the sensing channel width increases, the corresponding sensitivity increases at the cost of dynamic range. For the prototyped contact lens, the sensing channel length is over 80mm with a sensing channel width of 20µm. This corresponds to a dynamic range of 130mmHg – more than sufficient for measuring the IOP.

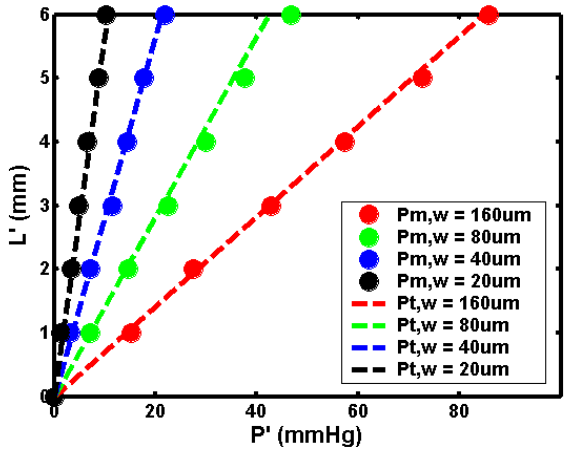

Figure 6. Measured results compared to computational analysis of elastomeric housing on top of rigid substrate

TABLE I. MEASURED SENSITIVITY AND DYNAMIC RANGE

Sensing Channel Width	Sensitivity	Dynamic range
20 μm	600 μm/mmHg	10mmHg
40 μm	285 μm/mmHg	21mmHg
80 μm	130 μm/mmHg	46mmHg
160 μm	70 μm/mmHg	85mmHg

CONCLUSION

This paper presents the development of a wireless, unpowered pressure sensor, composed of PDMS and dyed glycerol, for point-of-care glaucoma diagnosis. With such a device, glaucoma can be potentially detected at an earlier stage and steps can be taken to prevent serious vision loss.

ACKNOWLEDGEMENTS

This work is financially supported by the Graduate Assistance in Areas of National Need (GAANN) Fellowship, Kirchestein-NRSA Fellowship and the National Science Foundation (CAREER ECCS-0846502). The author would like to thank Tingrui Pan for technical discussions and Chaoqi Zhang for help in prototyping the device.

REFERENCES

[1] H. A. Quigley and A. T. Broman, "The number of people with glaucoma worldwide in 2010 and 2020," British Journal of Ophthalmology, vol. 90, pp. 262-267, Mar 2006.

[2] J. M. Tielsch, et al., "A Population-Based Evaluation of Glaucoma Screening - the Baltimore Eye Survey," American Journal of Epidemiology, vol. 134, pp. 1102-1110, Nov 15 1991.

[3] D. S. Friedman, et al., "Prevalence of open-angle glaucoma among adults in the United States," Archives of Ophthalmology, vol. 122, pp. 532-538, Apr 2004.

[4] K. C. Katuri, et al., "Intraocular pressure monitoring sensors," IEEE Sensors Journal, vol. 8, pp. 12-19, Jan-Feb 2008.

[5] R. M. H. a. K. D. Wise, "An Intraocular Pressure Sensor Based On A glass Reflow Process," Workshop on Solid State Sensors and Actuators, Hilton Head, pp. pp. 49-52 June 6-10, 2010 2010.

[6] M. Leonardi, et al., "First steps toward noninvasive intraocular pressure monitoring with a sensing contact lens," Investigative Ophthalmology & Visual Science, vol. 45, pp. 3113-3117, Sep 2004.

[7] P. J. Chen, et al., "Unpowered spiral-tube parylene pressure sensor for intraocular pressure sensing," Sensors and Actuators a-Physical, vol. 127, pp. 276-282, Mar 13 2006.

[8] H. Yao, et al., "A contact lens with embedded sensor for monitoring tear glucose level," Biosens Bioelectron, Dec 31 2010.

[9] J. Pandey, et al., "A Fully Integrated RF-Powered Contact Lens With a Single Element Display," IEEE Transactions on Biomedical Circuits and Systems, vol. 4, pp. 454-461, Dec 2010.

[10] Available: http://www.nei.nih.gov/health/glaucoma/

[11] J. C. Lotters, et al., "The mechanical properties of the rubber elastic polymer polydimethylsiloxane for sensor applications," Journal of Micromechanics and Microengineering, vol. 7, pp. 145-147, Sep 1997.

[12] D. Kim, et al., "A method for dynamic system characterization using hydraulic series resistance," Lab on a Chip, vol. 6, pp. 639-644, 2006.

[13] M. A. Unger, et al., "Monolithic microfabricated valves and pumps by multilayer soft lithography," Science, vol. 288, pp. 113-116, Apr 7 2000.

[14] Y. Liao, et al., "A 3μW Wirelessly Powered CMOS Glucose Sensor for an Active Contact Lens," IEEE ISSCC Dig. Tech. Papers, Feb. 2011