Andrew Duffy
Department of Chemical Engineering, University of Illinois at Chicago
aduffy3@uic.edu
(847) 624-4555

Development of a Meso-scale Drift-free Pressure-sensor for Glaucoma Patients

A. Duffy
Department of Chemical Engineering, University of Illinois at Chicago

K. Punchihewa and A. Feinerman
Department of Electrical and Computer Engineering,
University of Illinois at Chicago, Chicago, IL 60607

Abstract

Glaucoma is a devastating disease effecting millions of people, and is the second leading cause of blindness worldwide. The need for a drift-free biocompatible pressure sensor to monitor the intraocular pressure of glaucoma patients is ever present. The purpose of this project is to design and test a Meso-scale biocompatible pressure sensor that can measure small variations in intraocular pressure. The sensor will be fabricated from metalized thermoplastics and polyethylene using a novel process of welding developed at the University of Illinois at Chicago. The sensor will be drift-free and biocompatible. Once a prototype of the device is fabricated it will be tested in-vitro to insure that it can properly measure applied pressure. Finally the device will be implanted in laboratory animals to test for biocompatible.

Introduction

Glaucoma is a debilitating disease currently effecting 2.2 million Americans, and is estimated to affect 3.3 million Americans by 2020. Glaucoma is a condition referring to several eye diseases that damage the optic nerve. The ultimate result of the disease is irreversible destruction of retinal cells, which leads to blindness. This condition is the second leading cause of blindness worldwide. Increased intraocular pressure has been identified as a major risk factor for developing the disease.

The need for an accurate biocompatible pressure sensor that can monitor intraocular pressure is evident. One major problem facing traditional pressure sensors implanted into living tissue is pressure reading drift. This occurs when proteins and tightly woven collagen fibers form around the foreign object as an immune response. Buildup of material on the compliant membrane restricts its ability to deform and therefore produces inaccurate pressure readings.

In this study a drift-free biocompatible pressure sensor will be fabricated and tested. The first step of this project is the design and fabrication of a large-scale prototype sensor. Once complete the prototypes will be tested in-vitro in a custom built vacuum chamber.

After successful testing of the prototypes, fabrication of the actual scale sensors will begin. Three pressure sensors will then be implanted into laboratory rabbits to test for biocompatibility.

Theory

A drift-free sensor is essential in obtaining accurate pressure readings when implanted into living tissue. For this project a meso-scale drift-free pressure sensor will be constructed from metalized thermoplastic polyester films, high-density polyethylene (HDPE), and low-density polyethylene (LDPE). All of these materials have been shown to be biocompatible.

Two HDPE electrodes will be welded inside of a compliant thermoplastic fluid reservoir filled with saline solution. As positive gauge pressure is applied to the sack fluid will drain into a rigid gas reservoir thus compressing a small air column. This contraction of the compliant reservoir will move the electrodes closer causing a change in electrical impedance. Impedance readings can be correlated to an applied pressure, and thus an accurate intraocular pressure reading can be taken. Because this sensor does not rely on the deformation of a compliant membrane, it is hypothesized that protein and collagen deposition will not cause the impedance readings to drift.

Fig. 1: Schematic of Meso-Scale pressure sensor. Compliant liquid reservoir is made of Mylar film with HDPE electrodes situated inside the sack on the top and bottom. When positive gauge pressure is applied fluid drains into the LDPE rigid reservoir compressing the gas column and reducing the distance between the electrodes.

A novel process of welding thermoplastic films together using a vacuum frame and carbon dioxide laser has been previously developed in Dr. Feinerman's laboratory at UIC. This procedure will be used to create leak proof welds and fluid channels in Mylar films.

Fig. 2: 10x magnification of compliant liquid reservoir of prototype sensor. Two circular HDPE electrodes are located on the outer edge of the fluid sack. The separation of the two electrodes can be seen in the upper right corner.

Fig. 3: 10x magnification of LDPE rigid gas reservoir in prototype sensor. The Fluid-gas interface is located in the center of the image. This interface will advance or recede with varying applied pressure.

Fabrication of Prototype

All fabrication of the sensors was carried out at the University of Illinois at Chicago. The following is a step-by-step description of the fabrication process of the prototype sensors. These sensors measure 20mm by 50mm with a height varying based on applies pressure but not exceeding 7mm.

1. “P” shaped electrodes are cut from sheets of HDPE using a CO_2 laser and templates created on AutoCAD software.

2. Two HDPE electrodes are welded to Mylar sheets by compressing them between steel blocks and baking them for 25 minutes at 190 degrees Celsius in a furnace.

3. The two Mylar sheets are placed together with the electrodes touching in a custom built vacuum frame.

4. A 15-psi vacuum is applied between the sheets causing them to adhere to each other.

5. Using a template created on AutoCAD 2002 and a 100W CO_2 laser manufactured by *Universal Laser Systems* the two films are selectively welded together, and fluid channels are created to guide draining solution into the rigid reservoir.

6. Excess Mylar film is trimmed from around the sensor and a 1mm diameter LDPE tube is inserted into the center channel.

7. The ends of the sensor are sealed using acrylic strips and glue leaving a small amount of exposed electrode to be used for measurement.

8. 1.2 ml of saline solution is inserted through the LDPE tube using a medical syringe. The solution used in the experiment is BSS Sterile Irrigating Solution with a NaCl concentration of 0.64%. Care is taken to ensure no air pockets form inside of fluid sack, and the LDPE tube is sealed shut.

Fig. 4: 10x Magnification of center of prototype pressure sensor. The compliant fluid reservoir is located in the center of the image. Selective welding created using CO_2 laser is present in the lower right corner.

Experiment

In-vitro testing of the prototype sensors is carried out in a custom built acrylic vacuum chamber measuring 120 mm in length and 80 mm in height and width. The sensor is sealed in the airtight box with the two ends of exposed electrodes wired to an Inductance Capacitance Resistance (LCR) meter. The LDPE tube running from the sensor is connected to a vertical wound column of identical diameter tube that acts as a ~5ml reservoir for saline to drain into. The chamber is connected to a small aquarium pump, and an adjustable valve is connected in between the pump and the chamber.

When the aquarium pump is turned on the valve is slowly opened allowing pressure to build up in the chamber. A pressure gauge displays the gauge pressure that is applied to the sensor. When the pressure inside of the chamber reaches a desired value the valve is closed. Although the pressure is now constant fluid initially still drains from the sensor.

This is most likely because the tube used has such a small diameter that the fluid draining from the sensor lags behind the pressure. Once the fluid has come to equilibrium we record the volume of fluid drained from sensor, the impedance, and the gauge pressure. This process in continued from atmospheric pressure to ~13 kPa above atmospheric pressure. The same process is then carried out in reverse by slowly bleeding pressure out of the vacuum chamber and taking measurements.

Results and Discussion

Fig. 5: Electrical impedance in ohms as a function of applied gauge pressure in kPa. Three trials obtained during in-vitro testing of prototype sensor

Fig. 5 includes three in-vitro test trials, and describes the change in impedance of a prototype sensor as pressure in the vacuum chamber is varied. The lower line describes the impedance values as pressure in the chamber is increasing. The upper line describes the values for decreasing pressure. It is apparent that the two paths exhibit nearly linear behavior, however prominent hysteresis is present. A maximum impedance reading of between 1706 and 1709 ohms is recorded at a gauge pressure of ~13 kPa. At the maximum pressure of ~114 kPa the electrical impedance of the sensor increased by an average of 45 ohms.

Fig. 6: Volume of the air column located in the rigid gas reservoir in Liters as a function of applied gauge pressure in kPa. Three trials obtained during in-vitro testing of prototype sensor.

Fig. 6 shows the same three in-vitro test trials as Fig. 5, and describes the change in volume of the air column in the rigid gas reservoir as pressure is varied. Once again the lower line describes the path of increasing pressure, and the upper line is that of decreasing pressure. These hysteretic curves also show nearly linear behavior in the independent paths.

In order to produce accurate measurements the hysteretic behavior of the pressure sensors must be reduced as much as possible. Two main causes of hysteresis have been proposed. One being contact angle hysteresis, where by advancing and receding contact angles give rise to path dependent behavior. And the other source being mechanical hysteresis caused by stress and strain applied to the Mylar from continuous expansion and contraction. Reducing the rate at which pressure is varied and thus reducing flow rate of solution has thus far shown to decreases hysteretic behavior by most likely decreasing contact angle hysteresis. Further it has been proposed that decreasing the volume of the fluid used will moderate the contribution of mechanical hysteresis.

It is important that the compliant reservoir efficiently translates applied pressure without mechanical resistance. To insure that the reservoir is truly compliant we developed a model of expected pressure-volume behavior and compared this against our measurements. Using the Ideal Gas and Van Der Waals equations of state we were able to predict what the volume of the air columns should be at given pressures.

$$PV = nRT \qquad \ldots\text{Eqn. 1}$$

$$P = \frac{nRT}{V - nb} - \frac{n^2 a}{V^2} \qquad \ldots\text{Eqn. 2}$$

Eqn. 1 is the Ideal gas equation of state. Where P = pressure in kPa, V = volume in Liters, n = number of moles, T = thermodynamic temperature in Kelvin, and R = gas constant 8.314 $Jmol^{-1}k^{-1}$.

Eqn. 2 is the Van Der walls equation of state. Where a = correction for attractive forces between gas molecule 0.1402 Jm^3mol^{-2}, and b = correction for non-zero particle size $3.1764*10^{-5}$ m^3mol^{-1}. The constants were calculated from known values of oxygen and nitrogen, and the assumption that air is roughly 80% nitrogen and 20% oxygen.

Fig. 7: Volume of air column as a function of applied gauge pressure compared against a gas behavior model. The behavior model line was obtained using the Ideal Gas and the Van Der Waals equations of state.

Fig. 7 shows pressure-volume behavior for trial two compared against the model of predicted behavior. The linear brown line represents the expected volume of air as a function of pressure. Both the Ideal Gas and Van Der Waals equations of state overlap showing that the air column can be treated as ideal at the given pressures and temperature. From this graph it can be seen that the measured values obtained during in-vitro testing exhibit good agreement with the predicted model.

Summary and Future Work

To date several large-scale prototype sensors have been fabricated and tested in-vitro. Very promising nearly linear data has been recorded, and the pressure-volume data shows good agreement with predicted results. Further work is being done to reduce the hysteric behavior of the sensor during filling and draining.

An efficient method of fabricating the prototype sensors has been established and carries over well for the actual scale sensors. Thus far the previously developed production method has been used to fabricate a small number of actual size sensors. These sensors will measure 6mm by 10 mm with a height varying with pressure but not exceeding 2mm. A 1-3 micrometer outer coating of Parylene C will also be applied to improve biocompatible Once three working sensors are made they will be implanted into laboratory rabbits to test for biocompatibility.

References

1. Quigley HA, Broman AT. The number of people with glaucoma worldwide in 2010 and 2020 Br J Ophthalmol. 2006 Mar;90(3):262-7.
2. “Thin Welded Sheets Fluid Pathway,” A. Feinerman, patent filed 5/4/06.

3. N. Paya, Tatjana Dankovic, and A. Feinerman, “A Microfluidic Mixer Fabricated From Compliant Thermoplastic Films,” Journal of Undergraduate Research 2, pp. 1-5 (2008). http://jur.phy.uic.edu/issue2/JUR-REU0801005.pdf
4. Bélanger MC, Marois Y. Hemocompatibility, biocompatibility, inflammatory and in vivo studies of primary reference materials low-density polyethylene and polydimethylsiloxane: a review. J Biomed Mater Res. 2001;58(5):467-77. Review.
5. L. M. Raff: ‘Principles of Physical Chemistry’, 1st edn, 27; 2001Upper Saddle River