WIRELESS INTRAOCULAR PRESSURE SENSOR USING STRETCHABLE VARIABLE INDUCTOR

Mohammad Hossein Mazaheri Kouhani[1]*, Arthur Weber*[2]*, and Wen Li*[1]
[1] Electrical and Computer Engineering Department, Michigan State University, East Lansing, USA
[2] Physiology Department and Neuroscience Program, Michigan State University, East Lansing, USA

ABSTRACT

This paper reports the design, fabrication, and characterization of a wireless, flexible, passive pressure sensor that enables real-time intraocular pressure (IOP) monitoring. It employs a novel approach that measures the strain on the sclera of eye-ball using integrated planar stretchable variable inductor. Major reduction in fabrication complexity is achieved in contrast to the state-of-the-art solutions which are based on variable capacitors. Our device potentially improves the stability of long-term measurement by eliminating baseline drift due to leaks within sealed capacitive chambers. Implantation in the sclera reduces energy absorption by tissue, thereby improving inductive coupling efficiency between the implanted sensor and the external coil. This will significantly facilitate the commercialization of minimally invasive, *in-vivo* IOP sensing devices for Glaucoma patients. Devices are packaged using Parylene-C, which ensures the mechanical flexibility and biocompatibility of the sensor.

INTRODUCTION

Emergence of advanced bioinformatics calls for new technologies monitoring circadian patterns on a 24-h fashion for each individual patient reliably, accurately, and mobile. Glaucoma is the second leading cause of blindness in the world [1]. It is an asymptomatic, progressive, and irreversible disease that is usually associated with elevated IOP. Accurate, real-time, and portable monitoring of IOP can be of great use while coping with Glaucoma.

Most previous implantable devices, reported over the past two decades, can be categorized into three groups in terms of their energy transferring mechanisms: active, passive, and radio-frequency-powered devices.

Active devices usually employ application-specific-integrated-circuits (ASIC) that can store, process, and transmit data. However, they make the overall implant device large, heavy and inflexible. Also, active systems require energy either from an integrated battery or a power-receiving coil, which both add to the weight and size of the final implant. Implantdata Eyemate is a relatively successful example of its kind offering a wireless intraocular transducer (WIT) [2] and yet suffering from disadvantages discussed above.

Among fully-passive implants, optically read Bourdon tube is a well-designed representing example of its category [3]. It requires no external power delivery since the sensing implant is completely passive. However, it requires high precision surgery to anchor the device on the iris, which requires highly skilled operators. Also, the fabrication process of its spiral tube is complex and expensive. Furthermore, its optical external reading mechanism requires a large module that interferes with the vision of the patient, therefore, making it unsuitable for wearable, long-term measurements.

To date, most devices that are excited using electromagnetic (EM) coupling employ variable capacitors [4-6] as a pressure sensitive element. Variable capacitors need a pressurized reference chamber that hardly sustain their baseline pressure over-time due to packaging imperfections. This can cause significant signal drift mainly to their leaking. For instance, the Implantdata Eyemate suffers from 3.47 mmHg of drift on average [2].

Our solution to the above challenge is a passive strain gauge that serves as both a pressure sensitive element and a wireless interface. In our design, the strain gauge is configured as a variable inductor instead of a variable capacitor. The variable inductor will be implanted in the sclera of the Glaucoma eye. The EM response of the sensory implant change enabling us to correlate those changes with pressure fluctuations.

The method reported here differs from most previous works. In contrast to variable capacitive sensors and the variable inductor sensor [7] that is based on a movable ferrite magnet, our variable inductor does not employ any sealed chamber, which prevents the inaccuracy of measurement due to baseline drift over time. It is simple, stable, and cheap to fabricate, which makes it a great candidate for chornic implantation in Glaucoma eyes. If it is embedded in a contact lens, the inductive strain gauge sensor has the potential to become a non-invasive device.

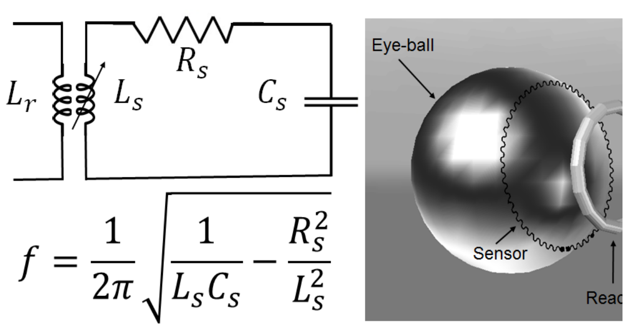

Figure 1: The conceptual diagram of the device and an equivalent circuit model.

DESIGN AND FABRICATION

A concept diagram of the sensor is shown in Fig. 1 along with its equivalent circuit. The wireless passive sensing used in this work is based on the mutual inductance between two coils: the reader and the sensor. The reader (transmitter) coil is connected to an impedance analyzer. The diameter of the reader coil is comparable to the diameter of the sensor coil which is coupled with. The sensor can be modeled as a planar and circular LC passive resonator that has a constant capacitor and a stretchable variable inductor. The self-inductance of the inductor, the parasitic capacitance between the segments, and the Q-factor are all changeable by the expansion of the sensor diameter. This phenomenon is employed to measure the strain in the eye tissues induced by IOP elevation. The change of self-inductance and parasitic capacitance result in the change of resonance frequency of the LC loop. Therefore, we are able to read the pressure variance by means of frequency drift.

Finite element simulation is performed in the Magnetic and Electric Fields (*mef*) module of COMSOL Multiphysics (Ver. 5.2 COMSOL, Inc.). The shift of the mutual inductance between the reader and sensory coils is studied in response to the diameter expansion of the sensory coil (Fig. 2). The geometries of the reader and sensory coils are provided in Table I. Simulation results show a decreasing mutual inductance as the diameter of the sensory coil expands, suggesting an upshift of the resonance frequency based on the resonance frequency equation shown in Fig. 1. Table I details the inductance decrease by the increase of diameter of the sensory coil.

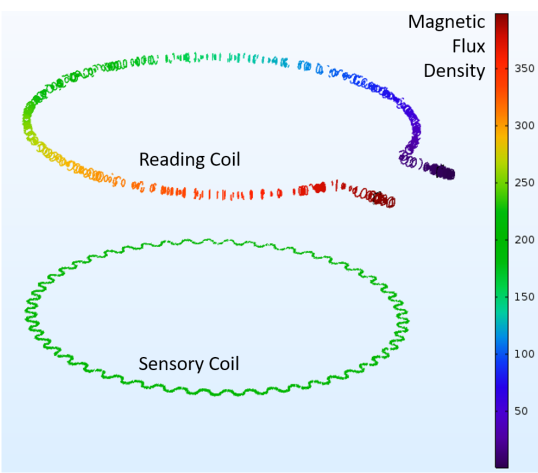

Figure 2: Simulation in COMSOL Multiphysics. The Inductance variation of reading-coil is measured with respect to diameter variation of the sensory coil.

Table I: Simulation Results for Inductance variation

Reading Coil (mm)	**15.5**	**15.5**	**15.5**
Sensory Coil Diameter (mm)	**15.5**	**19.5**	**23.5**
Inductance (pH)	**43.34**	**43.20**	**42.89**

The fabrication process is schematically shown in Fig. 3. The fabrication of the sensor prototype is carried out using two masks. First, a 3-inch glass wafer is coated with 5 µm Parylene-C, a 20 nm layer of titanium and 700 nm of copper (Fig. 3A and 3B) respectively. Parylene-C is deposited in a chemical vapor deposition (CVD) system *(*PDS2010 Parylene Coater, Specialty Coating System*)*. Metal is evaporated using a thermal evaporator. After that, mask one is used to pattern the geometric shape of the device metal core using ultra-violet (UV) photolithography and then wet etching (Fig. 3C and 3D). At this point, surface mounted (SMD) capacitor is soldered to the contacts of the inductor using silver epoxy. After the full wafer is coated with another 5 µm Parylene-C, aluminum is deposited on top of Parylene-C and patterned using the second mask to form an aluminum mask to protect the sealed metal core during plasma dry etching of Parylene-C (Fig. 3 E-G). Eventually, devices are released from the glass substrate before implantation (Fig. 3H). Fig. 4 show the photo image of a fabricated device and the microscope image of the "S-shape" stretchable wire.

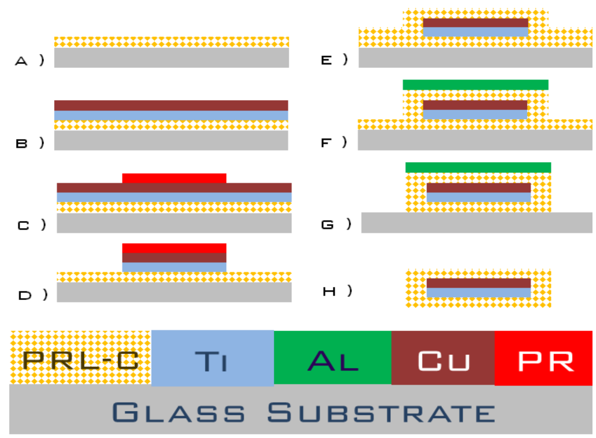

Figure 3: Fabrication process flow of the sensor. A. Parylene-C deposition on full wafer. B. Metal deposition. C. First UV Photolithography. D. Wet etching metal. E. Full wafer Parylene-C Deposition. F. Forming aluminum mask using the 2nd mask. G. Plasma dry etching Parylene-C. H. Releasing device.

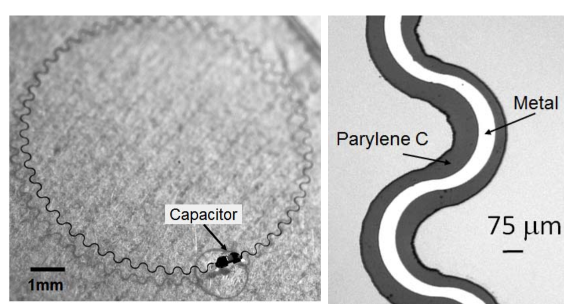

Figure 4: A photo of a fabricated stretchable sensor and a close-up microscope image showing the "S-shape" wire geometry.

CHARACTERIZATION AND RESULTS

Device is characterized under two circumstances. One uses a latex balloon as a model for eye. The pressure of the balloon is controlled by adjusting the air pressure inside of it using a syringe pump. Fig. 5 shows the experimental setup for device characterization using the balloon. The detected resonant frequency versus the separation between the coils and the chamber pressure are shown in Fig. 6 and Fig. 7, respectively. The elasticity the balloon membrane is much higher than the eye tissue. As a result, the strain gauge sensor expands way larger, faster, and easier in response to lower pressure elevations comparing with the eye. Therefore, the pressure sensitivity measured using the balloon is higher than the one characterized later using a pig eye. A pressure sensitivity of 2667 KHz/mmHg is observed in case of balloon. The Q-factor of the sensory coil is measured as the sharpness of the resonant peak. As shown in Fig. 6, the Q-factor drops as the separation of the two coils increases, mainly due to the weakening of the electromagnetic coupling between the two coils.

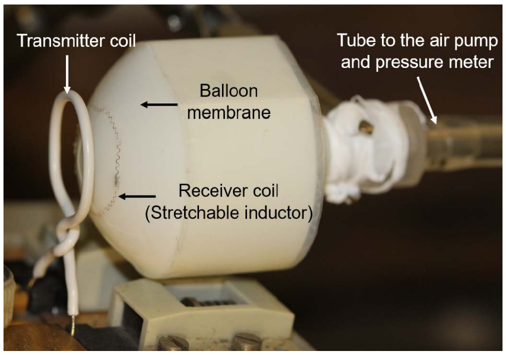

Figure 5: Setup for testing the stretchable inductor on a balloon membrane

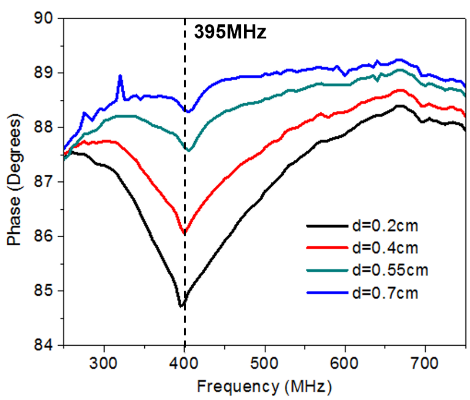

Figure 6: Results showing the dependence of signal strength on the distance between the reading coil and sensor.

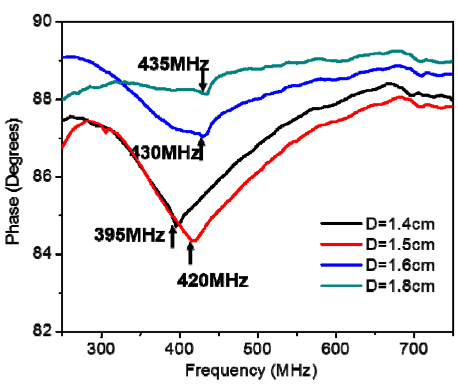

Figure 7: Resonance frequency shift due to pressure elevation.

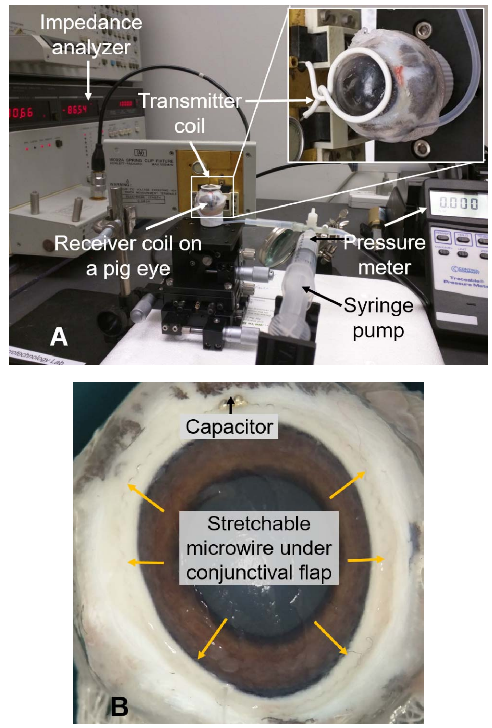

Figure 8: A. Setup for testing the stretchable inductor on a pig eye. B. Top-view of the pig eye with a device implanted around the cornea

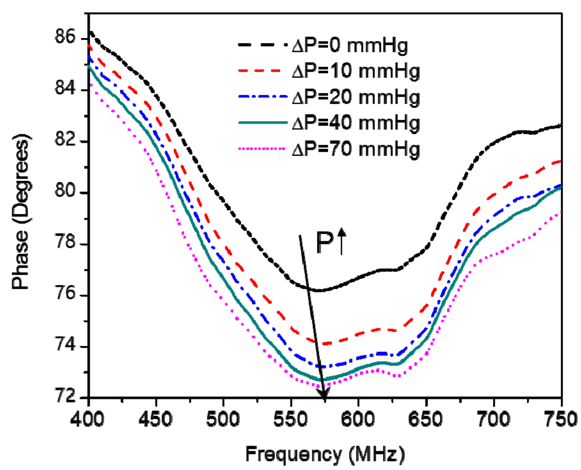

Figure 9: Resonance frequency drift due to pressure elevation.

Table II: Summary of device design and pressure sensing performance.

Properties	Device on balloon	Device on pig eye
Coil diameter (cm)	**1.4**	**1.4**
Overall thickness (μm)	**10.8**	**10.1**
Capacitance (pF)	**2.2**	**1.8**
Inductance (nH)	**50**	**50**
Parasitic resistance (Ω)	**50**	**25**
Theoretical resonant freq. (MHz)	**411**	**551**
Pressure sensitivity (kHz/mmHg)	**2667**	**57**

In order to become one step closer to *in-vivo* studies, we report a characterization experiment on a pig eye. Fig. 8 illustrates the experimental setup for device testing and Fig. 9 shows the frequency shift versus the pressure elevation. In this case, a small conjunctival flap is cut around the cornea. The inductor wire is embedded under the conjunctival flap of the eye. The eye is pressurized with an initial IOP of around 15 mmHg. A small needle is inserted to the bottom of the eye in order to inject saline to the interior chamber of the eye through a syring pump. This enables us to mimic the pressure elevation that occurs in the human eye. Since the outflow of the liquid is limited, we are able to control he pressure by only increasing the inflow of saline. This experiment shows a pressure sensitivity of 57 KHz/mmHg based on the frequency shift correlated with the eye pressure in Fig. 9.

Table II summarizes the performance of the sensor characterized under different conditions. Results from both balloon and pig eye are in agreement with each other, in a way that, they both present an upshift of resonant frequency when the pressure elevates (Fig. 7 and Fig. 9). The elasticity of the balloon membrane is significantly higher than the eye tissue, therefore, exhibiting a much higher sensitivity. The Q-factor of the sensor drops when moving from the balloon to the pig eye due to the electromagnetic absorption of the tissue. The change in Q-factor may act as an additional indicator signal correlated with pressure variation in our further studies.

When the sensor expands, the upshift of the resonant frequency is speculated to be caused by two main factors. The mutual inductance between the reader and sensory coils decreases as indicated in Table I. This might be due to the variation of the overall magnetic flux density through the coil loops, as a result of the change in the air core area. Another important factor is the parasitic capacitor between adjacent winding segments, which decreases as the separation of the segments increases. The ruling analytical equation has not yet completely understood and will be investigated in the future work.

CONCLUSION

In this work, we design and implement a variable inductive sensor that can non-linearly respond to pressure variation by changing its geometry due to the strain caused by the pressure elevation in the eye. We demonstrate that an upshift of resonant frequency occurs in the sensing LC tank when the pressure is elevated. We analytically speculate and verify with simulation results that this frequency shift is caused by the diameter increase of the sensory coil. However, we do not yet exactly know the detailed equation ruling this relationship. Indeed, the parasitic capacitance between the “S-shape” winding segments of the sensory coil also play an important role, along with the variations of the overall Q-factor and mutual inductance. Further work is in progress to study various other designs and their EM response to precisely explain the fundamentals behind this relationship and optimize the sensing mechanism.

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CONTACT

*M. H. M. Kouhani, tel: +1-517-9748327;
mhmk@msu.edu