High Quality Factor Parylene-Based Intraocular Pressure Sensor

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Abstract—A new concept of the intraocular pressure (IOP) sensor design and its implantation approach are presented in this paper. A parylene-based sensing part with about 30 µm in thickness was fabricated, and then integrated with an implantation tube attached to sensor’s backside pressure access hole. During the implantation, only the implantation tube was implanted into the anterior chamber to fulfill minimally invasive implantation. The IOP sensor membrane is thin and flexible so that it can attach to the cornea. Because the sensing area was exposed outside to the air all the time, the quality factor can be kept at 27-30 to maintain the sensing distance during the whole testing pressure range. The sensitivity is obtained as high as 542 ppm/mmHg while the responsivity is about 205 kHz/mmHg, which is suitable for biomedical applications.

Keywords—Intraocular implant, Intraocular pressure, pressure sensor, Glaucoma, Parylene

I. INTRODUCTION

People have been working on intraocular pressure (IOP) sensors measuring human’s eye pressure (or IOP) for years to prevent and treat eye pressure-related diseases. For example, glaucoma is an eye disease caused by elevated eye pressure in patients’ eyes. Without proper treatment, the elevated IOP would damage patients’ optic nerves in the backside of the eye, and causes the blindness in the end. Statistics show that glaucoma is the second leading cause of blindness in the world according to World Health Organization [1]. Because there could be no symptoms of pain and human’s eye tend to compensate a small peripheral vision loss, glaucoma patients in early stage usually don’t know they are developing glaucoma and turns out to treat the disease late. It is shown that only half of the glaucoma patients in the U.S are aware of their having glaucoma. Thus an early stage diagnostic becomes important to find and treat glaucoma.

To diagnose the glaucoma, people usually measure a person’s IOP to determine if the person is having glaucoma or not. Currently there are several ways to measure IOP. One of the most common ways is puff air onto or directly touch person’s eye to calculate the current IOP by measuring the bouncing force of the eye ball. The drawbacks of these approaches are not just only the fact that some of them are contact methods, but also there exists some limitations of it. For example, as it is reported that the IOP can actually fluctuate depending on patients’ daily lives [2], it is necessary to monitor patients’ IOP 24 hours to understand eye pressure progress.

In order to fulfill the goal of monitoring IOP automatically, continuously and wirelessly, in the past we have presented a flexible parylene-based IOP sensor[3]. The sensor has one inductor and one capacitor combined in series to fulfill a passive LC-tank resonance circuit. The IOP sensor was implanted into the anterior chamber and anchored on the iris. The resonant frequency shift was registered by an external reader coil through a wireless inductive coupling link, as shown in Fig.3. In Fig.4-(a), when the sensor’s surrounding pressure increases, the capacitance increases due to the sensing plate’s deforming concavely, causing the resonant frequency shifts to the lower range. However, with the high loss tangent of the eye fluid in the anterior chamber and the blockage of the cornea, the quality factor measured was seriously degraded and the sensing distance reduced [3-5]. To overcome this problem, the new IOP sensor structure presented here has a very similar sensing part design, but the new implantation approach leaves the sensing part exposed to the air keeping its high quality factor, as shown in Fig.2. The sensing part is still composed of one sensing inductor and one sensing capacitor, as shown in Fig.1-(a). A pressure access hole connecting the sensing capacitor was created during the fabrication. An implantation tube, which inner diameter is larger than the pressure access hole was mounted onto the back of the device to cover the pressure access hole (Fig.1-(b)). The biocompatible epoxy was applied to seal the gap between the implantation tube and the sensing part to make it airtight and ensure the biocompatibility.

Fig.1. Thenew IOP sensor design: (a) Top view of the sensing part; (b) AA’ cross-section view of the IOP sensor.
Fig. 2: The concept of minimally invasive implantation. The IOP sensor was implanted through the cornea and the flexible parylene-based sensing part is lying on the cornea.

![Image](https://via.placeholder.com/150)

**Fig. 3.** The concept of wireless inductive coupling link: The frequency shift is registered through an external oil reader.

(a) $\Delta p>0 \text{ mmHg}$  (b) $\Delta p=0 \text{ mmHg}$  (c) $\Delta p>0 \text{ mmHg}$

**Fig. 4:** Resonant frequency shift corresponds to the applied pressure. (a) Frequency decreases as the capacitance increases. (b) No frequency shift when no pressure difference exists. (c) Frequency increases as the capacitance decreases.

### II. SENSING SCHEME AND THE DEVICE DESIGN

#### A. Sensing scheme

The concept of wireless sensing scheme is shown in Fig. 3. The right RLC circuit represents the implanted IOP sensor and the resonant frequency can be expressed as [4]

$$f_s = \frac{1}{2\pi} \sqrt{\frac{1}{L_c C_s} - \frac{R_s^2}{L_s^2}} \approx \frac{1}{2\pi \sqrt{L_c C_s}}$$

(1)

where $R_s$, $L_c$, and $C_s$ represent the sensor’s resistance, inductance, and capacitance, respectively. The equivalent impedance viewed from the external coil reader and apparatus is derived as [5-7]

$$Z_{eq} = j2\pi f L_c \left[ 1 + k^2 \frac{\left(\frac{f_s}{f}\right)^2}{1 - \left(\frac{f}{f_s}\right)^2 \frac{L_s}{L_c}} \right]$$

(2)

where $f$ is the excitation frequency, $k$ is the coupling coefficient of the inductive link depending on the physical geometries of the sensor [5-9]. $Q_s$ is the quality factor of the sensor at the resonance and can be represented as

$$Q_s = \frac{1}{R_s \delta^2 C_s}$$

(3)

When the sensor is excited at the resonant frequency, $Z_{eq} (2)$, becomes

$$Z_{eq} = j2\pi f L_c (1 + jk^2 Q_s)$$

(4)

and its phase dip magnitude can be approximated as

$$\Delta \phi \approx \tan^{-1}(k^2 Q_s)$$

(5)

When the capacitance of the IOP sensor changes, it can be shown from (3)-(5) that the impedance phase dip shifts to either lower or higher frequency and can be detected by a network analyzer.

#### B. Electrical and mechanical design of the device

The electrical design of the IOP sensor can be designed by the well-developed equations as follows [3, 10, 11]. The inductance of the spinal coil can be represented as:

$$L_s \approx \frac{\mu_0 n^2 d_{avg} c_s}{2} \left[ \ln \left(\frac{c_s}{r}\right) + c_3 F + c_4 F^2 \right]$$

(6)

where $n$ is the number of turns of the inductor, $d_{avg}$ is the averaged diameter of the coil windings, $F = (d_{out} - d_{in})/(d_{out} + d_{in})$ is the fill factor of the coil windings, and $c_3$, $c_4$ are constant coefficients determined by the winding geometry. The coil inductor inherently comes with a resistance which can be calculated as:

$$R_s = \frac{\rho}{\omega d (1 - e^{-h/\delta})}$$

(7)

where $\rho$ is the electrical resistivity of the metal, $w$ and $h$ are the metal line width and height, respectively. $\delta$ is the frequency-dependent metal skin depth which can be written as:

$$\delta = \frac{\rho}{\sqrt{\pi f \mu}}$$

(8)

where $\mu$ is the magnetic permeability of the metal. The capacitance of the IOP sensor is given by

$$C_s = C_{s,g} + C_{s,p}$$

(9)

where $C_{s,g}$ is the capacitance of the parallel metal plates capacitor at the center of the IOP sensor and $C_{s,p}$ is the parasitic stray capacitance introduced by other components in the entire device.

To have a detectable impedance phase dip shifts, a deformable circular metal plate was designed at the center of the IOP sensor. Once the plate deforms either downward or upward, the capacitance of the parallel metal-plates diaphragm capacitor changes and the impedance phase dip shifts can be registered. The deformation of the metal plate corresponding to the pressure difference can be predicted as [12]:

$$w(r) = \frac{\Delta P a^4}{640 \left[ 1 - \left(\frac{a}{r}\right)^2 \right]^2}$$

(10)

where $\Delta P$ is the pressure difference, $r$ is the radius calculated from the center of the plate, $a$ is the diaphragm radius and $D$ is the flexural rigidity of the diaphragm.

In our new sensor implantation approach, the sensing metal plate deforms convexly with higher surrounding pressure transmitted to the metal diaphragm capacitor chamber through the implantation tube. It is shown that according to (1), this higher eye pressure causes the capacitance to reduce and thus the resonant frequency shift to the higher range, as described in Fig. 4-(c).

### III. DEVICE FABRICATION AND CHARACTERIZATION

#### A. Device fabrication

The fabrication procedure is shown in Fig. 5. The sensing part was made of parylene-3 µm gold-parylene sandwich structure. The first layer of parylene was first deposited and...
the pressure access hole with 180 µm in diameter was opened by oxygen plasma. A 3 µm Ti/Au was deposited on top of the first layer parylene and patterned. The distance between 2 capacitor metal plates was designed as 10 µm which was fulfilled by spin coating a 10 µm photosresist and patterned. The second layer of parylene was deposited to cover and protect the 3 µm Ti/Au, followed by a 0.5 µm Ti/Au and then the third parylene layer with 7 µm in thickness. The final sensing part was released from the substrate by soaking in the acetone. The final completed sensing part is shown in Fig.6-(a).

After the sensing part fabrication was done, an implantation tube was attached onto the backside of the sensing part, as shown in Fig.6-(b). The inner diameter of the implantation tube was chosen as 320 µm to fully cover the pressure access hole. The outer diameter of the implantation tube is 450 µm. The implantation tube was manually mounted onto the sensing part. A precision XYZ stage was used to control the position of implantation tube, maneuvered to be concentric with the pressure access hole, as shown in Fig.1-(b). The implantation tube and sensing part were glued together by putting few drops of biocompatible epoxy.

B. Device Characterization

The completed IOP sensor was then integrated to a bigger testing capillary tube and sealed by photosresist, as shown in Fig.6-(c). The inner diameter of the testing tube was chosen as 500 µm to accommodate the implantation tube. The complete sensor with testing capillary assembly was stayed overnight to dry the photosresist.

The device characterization setup is shown in Fig.7. The whole assembly was mounted onto a pressure characterization setup. During the characterization, a HP 4195A Network/Spectrum analyzer was hooked up with a 1.5-mm-diameter hand-wound coil serving as the reader coil. The characterization signal was accessed via a data acquisition system and then analyzed in personal computer. The qualified IOP sensor was released by soaking the whole assembly in the acetone to remove the photosresist. The final complete IOP sensor is shown in Fig.6-(d) and is ready for the next in vivo/ex vivo test.

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IV. CHARACTERIZATION RESULTS AND DISCUSSION

The bench top characterization results are shown in Fig.8. The results showed that the resonant frequency is 379 MHz when the applied pressure difference is 0 mmHg. When the applied pressure difference increases, the resonant frequency shifts to the right because the metal plate deforms convexly as expected. The IOP sensor’s electrical parameters are obtained and shown in Table 1. As the sensing part can always be maintained exposed to the air, the problem of quality factor drop caused by the lossy medium is solved in our new IOP sensor implantation approach. Therefore, the sensing distance can always be maintained as far as 2.5 cm, which was originally designed for the raw sensing part. The 2.5 cm
sensing distance can fulfill the concept of glass reader paradigm to accomplish the autonomous, continuous, and wireless IOP monitoring.

The result of sensitivity analysis is shown Fig.9. The sensitivity of the IOP sensor is defined as: [4]

\[
\text{IOP sensor sensitivity} = \frac{\partial P}{\partial (\Delta P)}|_{\Delta P=0} 
\]

where \( R \) is the frequency ratio defined as:

\[
R = \frac{f_{\text{min}}}{f_{\text{min}(\Delta P=0)}} \quad (12)
\]

The sensitivity of the IOP sensor is obtained as 542 ppm/mmHg, corresponding to the responsivity as 205 kHz/mmHg. With a proper designed high resolution external coil reader, the IOP sensor can resolve the pressure difference < 1 mmHg, which is suitable for glaucoma diagnostics.

V. CONCLUSION

We have successfully demonstrated the feasibility of the new concept and design of IOP sensor implant and its implantation approach. In this new implant, the IOP sensor can be implanted with sensing part exposed to the air, which maintains the high quality factor. This benefit makes the glass reader paradigm in reality, achieving autonomous, continuous and wireless IOP monitoring. The characterized IOP sensor is ready for use, and the ex vivo test is scheduled in the near future to verify the biological feasibility.

VI. ACKNOWLEDGMENT

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![Fig.8. Characterization results.](image)

Table 1. Measured IOP sensor’s electrical parameters.

<table>
<thead>
<tr>
<th>Pressure (mmHg)</th>
<th>0</th>
<th>20</th>
<th>40</th>
<th>60</th>
<th>80</th>
<th>100</th>
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<tr>
<td>Frequency (MHz)</td>
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<td>381</td>
<td>385</td>
<td>390</td>
<td>395</td>
<td>402</td>
</tr>
<tr>
<td>Q Factor</td>
<td>27</td>
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<td>28</td>
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<td>30</td>
<td>29</td>
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<tr>
<td>Sensitivity</td>
<td>542 ppm/mmHg</td>
<td></td>
<td></td>
<td></td>
<td></td>
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<tr>
<td>Responsivity</td>
<td>205 kHz/mmHg</td>
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</tbody>
</table>

![Fig.9. Sensitivity analysis of the IOP sensor.](image)

VII. REFERENCES


