ABSTRACT

KATURI, KALYAN CHAKRAVARTHI. Design and Optimization of Passive Wireless Intraocular Pressure Sensor. (Under the direction of Melur K. Ramasubramanian.)

Glaucoma diagnosis and the prevention of the disease progression depend heavily on the accuracy of intraocular pressure measurements. Continuous monitoring of intraocular pressure would make it possible to arrest the disease progression early so that the patient does not lose much of his vision. The objective of this research is to develop a design process for a minimally invasive implantable sensor for continuous intraocular pressure monitoring. The design process include identification of optimal implant location, estimation of available space for the implant, MEMS pressure sensor design, magnetic link design, and optimal implant insertion method. As a part of the pressure sensor design cycle, a micro scale capacitive pressure sensor was designed, fabricated and tested. Kapton and ParyleneC based flexible implant fabrication processes were investigated to develop sensor implant that could be inserted in a minimally invasive manner. An implant delivery system was designed for inserting the implant using incision sizes of 2.8mm-3mm on the corneal surface. A finite element analysis of the insertion process was modeled to identify stress concentration regions in the implant and for determining the orientation of implant in the delivery device so that the capacitive pressure sensor region develops minimum stress during the insertion.
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Design and Optimization of Passive Wireless Intraocular Pressure Sensor

by
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To amma, nanna
BIOGRAPHY

The author was born in a small town in the coastal region of Andhra Pradesh, India. After finishing his bachelors in mechanical engineering from Jawaharlal Nehru Technological University in Hyderabad, he enrolled as an MS graduate student at NC State. After finishing his Masters degree (concentration: Mechatronics) in 2006, he enrolled as a PhD student in mechanical engineering department at NC State. Kalyan's research interests include applied mechanics, controls and instrumentation, computational mechanics, wireless passive sensor design, bio-mimetic systems, BioMEMS, and electromechanical systems design.
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1.1 Glaucoma

Glaucoma is a group of eye diseases, usually characterized by elevated intraocular pressure (IOP). IOP is the pressure exerted by the ocular fluid called “aqueous humor” that fills the anterior chamber of the eye. Aqueous humor produced by the ciliary body in the posterior periphery of the eye chamber enters the anterior chamber through pupillary opening and drains out of the anterior chamber through two different routes. Most of the aqueous humor exits the eye via trabecular meshwork, Schlemm’s canal and episcleral veins (Figure 1.1). The remaining aqueous (about 10%-20%) leaves through the uveoscleral route where the aqueous humor passes between the ciliary muscle bundles [1].

Glaucoma damages the optic nerve which carries visual signal information from the eye to the brain. Patients with glaucoma initially lose their peripheral vision. If left untreated, the patients lose their vision completely. Glaucoma affects an estimated 67 million people worldwide, including over 3 million Americans aged 40 years and older [2]. Approximately
120,000 are blind from glaucoma; accounting for 9%-12% of all cases of blindness in the U.S. Glaucoma is the second leading cause of blindness in the U.S. and the first leading cause of irreversible blindness.

Figure 1.1: Aqueous humor flow path in a healthy eye [3].

The most common type of glaucoma is open-angle glaucoma and it accounts for about 90% of all glaucoma diagnosed cases [4]. In patients with open-angle glaucoma, the angle in the eye where the iris meets the cornea is open but the Schlemm’s canal and trabecular meshwork become clogged overtime leading to a mismatch between the inflow and outflow of intraocular fluid. Open-angle glaucoma develops over a period of time with out any noticeable symptoms. Since the visual acuity remains intact until late in the disease, it is harder for the patient to notice any change in the quality of the vision during initial periods of disease progression.
Unlike open-angle glaucoma, angle-closure glaucoma develops suddenly due to blocked drainage canals. Blocked drainage canals are a result of bunching up of outer edges of iris over the canal surface. This leads to a sudden spike in IOP. Compared to open-angle glaucoma, angle-closure glaucoma is rare and the treatment is relatively simple. Figure 1.2 shows the IOP fluid flow paths for both open-angle and angle-closure glaucoma. Other variants of glaucoma include low-tension glaucoma, congenital glaucoma, and secondary glaucoma caused by diseases such as diabetes [5].

Normal IOP is in the range of 10-21 mmHg [7, 8] but in glaucoma patients, IOP increases above the normal range because of increased resistance to the fluid flow in the drainage pathway. Elevated IOP is associated with loss of optic nerve tissue, loss of peripheral vision, and leads to blindness if not treated. IOP measurement, optic disc examination, and visual field testing are used for glaucoma diagnosis. Regular monitoring of the above three parameters is important for disease management. Early treatment helps to slow disease progression. However, early signs are detectable only by a physician. This is especially true in open angle glaucoma, which is typically symptom-free in the early stages. Current treatment is directed towards reducing the IOP, which has been shown to decrease disease progression. Medications in the form of eye drops are commonly used to lower IOP. These help either by decreasing the aqueous fluid production or by reducing the resistance to aqueous outflow via trabecular meshwork and uveoscleral route [9].

The following section reviews some of the standard tonometry techniques and wireless IOP measurement techniques. An extensive review of IOP sensing techniques were discussed by the author in an earlier publication [10].
1.2.1: Open-angle glaucoma.

1.2.2: Angle-closure glaucoma.

Figure 1.2: Open-angle and angle-closure glaucoma [6]
### 1.2 Standard Techniques for IOP Measurement

The standard way of measuring the IOP is by determining the resistance of the cornea to indentation using an instrument called tonometer. There are several indentation techniques in practice. Some of them are discussed below.

#### 1.2.1 Applanation (Goldmann) Tonometry

Goldmann Applanation Tonometer (GAT) is considered as “gold standard” for measuring IOP [11]. It is based on Imbert-Flick principle [12], [13], which states that the pressure inside the liquid filled sphere can be determined by the force required to flatten a portion of the sphere. GAT uses a probe to flatten a portion of the cornea and a slit lamp microscope is used to examine the eye. The pressure within the eye is calibrated to the weight required to flatten 3.06 \( mm^2 \) of the cornea. This is the minimal area of applanation needed to give accurate results, yet, causing only an increase of 2.5% in IOP.

Eventhough the GAT measurements are accurate to within 0.5 mmHg for IOPs of 20 mmHg or lower, they are dependent on corneal thickness. A thinner cornea than normal would applanate more thereby providing underestimation of the pressure. Similarly, a thicker cornea than normal would overestimate the IOP [14–16]. Corneal rigidity and central corneal thickness differs between patients. Several correction factors are needed to get accurate measurements.

#### 1.2.2 Non-contact Tonometry (Pneumotonometry)

In non-contact tonometry, instead of a probe, an air jet is directed at the cornea to flatten a portion of it [17–19]. Captured light reflected from the flat portion of the cornea gets is used to determine the size of the flattened area. Once a predetermined flattening diameter
is achieved, the corresponding pressure is used to estimate the IOP. This method does not require the cornea surface to be anesthetized and hence can be used to make large number of measurements.

Over the years, various portable tonometers have been developed, and these include the Perkins [20], Tono-Pen [21], Zeimer’s self-tonometer [22], Ocuton-S [23], and ProTon [24]. However, it is not possible to measure the IOP continuously by using these techniques.

1.3 Need for Continuous IOP Measurement

Patients with glaucoma can be mistakenly considered to be “well controlled” if their mean IOP is lower than 21 mm Hg. However, a well-known fact is that many glaucoma patients continue to progressively lose visual field, despite having IOPs that are considered “well controlled.” One possible explanation could be that progression is due to IOP variations within the acceptable limit. Asrani et al. [25], after monitoring 105 human eyes with home-use tonometry, reported that in glaucoma patients with office IOP in the “normal” range, large fluctuations in diurnal IOP were a significant risk factor, independent of parameters obtained in the office.

Several clinical studies [26–32] found that 24 hour monitoring may reveal IOP fluctuations that may help in better design of treatment methodologies. Treatment of glaucoma involves knowing that the pharmacologic treatment is effective and the pressure fluctuation does not exceed the allowable limits. Multiple or continuous measurement of the IOP in glaucoma patients may help in better disease diagnosis, monitoring, and management. Frequent/continuous data collection (IOP measurements) may impact glaucoma treatment similar to the impact of home glucose monitoring on the management of diabetes.
1.4 Continuous IOP Measurement

Continuous IOP sensing devices can be broadly categorized, as shown in Figure 1.3.

Implant locations of these devices are typically on the cornea, sclera, and iris. Anterior chamber implanted sensor measurements are insensitive to variations in ocular surface, cornea rigidity, and procedures that were performed on the eye such as keratoplasty and keratoprosthesis [33].

1.4.1 Inductively Coupled Telemetry

In general, Bio-telemetric systems are classified into active sensing devices and passive sensing devices. Both the systems use capacitive transducers for pressure sensing because of inherent advantages such as low power consumption, low noise, high sensitivity, low temperature drift, and good long-term stability [34]. An additional advantage is the seamless integration of capacitive sensor fabrication into standard IC fabrication techniques.
Passive Devices

The earliest passive IOP sensor was developed by Collins [35]. It has a gas bubble encapsulated in a cylindrical container consisting of glass cylindrical wall and stretched polyester circular diaphragm ends. The inner surfaces of the stretched polyester diaphragms are attached to two coaxial, Archimedean-spirals connected to form a single winding. When external pressure acts on the diaphragm, it pushes the coils closer together. This increases the mutual inductance. The frequency of an external inductively coupled oscillator is swept to continuously monitor the resonant frequency of the transducer.

Backlund et al. [36] improved Collins’ design by using a capacitive pressure sensor made using silicon fusion bonding. In silicon fusion bonding procedure, the wafer surfaces are first hydrated by boiling in $HNO_3$ followed by heating at 1000°C to fuse the wafers together with a cavity in between. A passive LC resonance technique was used for wireless pressure sensing and the resonance frequency was detected using a grid dip configuration. Subsequently, a 6 to 12 turn 50µm gold wire, 5 mm diameter coil was hand wound and bonded to the capacitive pressure transducer element [37, 38]. The entire assembly was encapsulated in silicone for bio-compatibility, resulting in an overall size of 5 mm diameter $\times$ 2 mm thickness. The sensor was tested in a cannulated rabbit eye. The resonator and total sensitivities are reported as 1 kHz/mmHg and 4 mV/mmHg, respectively.

Schuylenbergh and Puers [39] further improved this technique by patterning electrodes as flat coils to form LC tank circuit whose resonance frequency was obtained by inductive excitation. The two coils are connected such that the spirals act cooperatively as a single coil, while the capacitive coupling between the coils is changed by the pressure, which changes the gap between the coils, thereby changing the resonant frequency of the device. Later Puers et al. [40] fabricated the inductor by electro-deposition of copper on a micro-machined
chip with a pressure sensitive diaphragm. The inductor was split into halves: one placed on the movable diaphragm and the other fixed on substrate. Both halves are bonded with a small gap between the inductor parts and the inductors are connected by feed-through contacts. A diode was connected in parallel to the sensor chip to overcome weak inductive coupling between the detector coil and implant. The diode results in the generation of higher harmonics with maximum magnitude at the resonance. The sensor implant size was reported as $4 \times 4 \times 1 \text{ mm}$.

Akar et al. [41] made an absolute capacitive pressure sensor with an on-chip gold electroplated planar coil that can be used to remotely sense the pressure using LC resonance technique. The on-chip fabrication of inductor inside the sealed cavity of the capacitor minimized the implant size ($2.6 \text{ mm} \times 1.6 \text{ mm}$). The inductor structure is electroplated on the recess created in a glass substrate. The glass and silicon wafers are attached using anodic bonding. The inductor coil has a width, height, and separation of $7 \mu \text{m}, 6 \mu \text{m}, \text{and} 7 \mu \text{m}$, respectively. Pressure measurements were carried out with a hand wound external coil (10 turns, 3 mm diameter) formed by an insulated copper wire of 0.38 mm diameter. The pressure range and sensitivity of the sensor are reported as 0-50 mmHg and 120 kHz/mmHg, respectively. The coil separation distance was only 2 mm due to low Q-factor of the sensor coil.

Coosemans et al. [42] developed an inductor capacitor resonant circuit capacitive transducer and used a voltage controlled oscillator to excite the sensor over a frequency range of 20-40 MHz. The mutual coil separation distance was reported as 7.5 mm.

In addition to the above, several other passive sensors have been reported [43, 44], but they all suffer from the requirement of high coupling between the external antenna and implant, primarily claimed to be due to the size limitation limiting the coil size in the implant. This is primarily due to the approach of monolithically integrating into the package and
trying to miniaturize it as far as possible. Due to weak coupling, even small changes in the relative positions of the implant and the external device affects pressure measurement. Such passive sensors are useful where the sensor and the detection circuits can be close to each other, and the distance can be controlled. In IOP measurement, this minimum distance is governed by the anatomy of the eye.

**Active Devices:** Active sensing devices are more robust than passive devices for miniature implants like IOP sensor. As the size of implant coil becomes smaller, the transmission range of passive telemetry system decreases further. Active telemetry is necessary to transmit the power and data to larger distances, especially when using small size implants. Implant power consumption is made minimal in these systems to maximize the operating distance [45]. On-chip storage of calibration data, high signal-to-noise ratios are other advantages of active sensing devices compared to passive devices. The design of active telemetry system includes the design of inductive link for power/data transmission, arrival at optimal antenna dimensions, and understanding the effect of implant antenna dimensions on the range of the system.

Schuylenbergh et al. [46, 47] described a telemetric tonometer for IOP measurements. The sensor was implanted in an artificial intraocular lens by creating a cylindrical space of \(3.5\, \text{mm}ID \times 8\, \text{mm}OD \times 0.5\, \text{mm}\) thickness. A power of \(100\, \mu\text{W}\) is delivered by a tuned coil system. A switch capacitor circuit, which runs at 35\(\mu\text{A}\) was used to make a differential pressure measurement between the reference capacitor and the pressure sensor. McLaren et al. [48] implanted a commercial telemetric pressure transducer subcutaneously on the back of the neck of the rabbits and measured the IOP by using a catheter that conducts pressure to the transducer from the anterior chamber. This is an invasive technique because of the catheter tip in the anterior chamber. This can cause irritation in the eye and also might cause scars on the cornea surface.
Eggers et al. [49] developed an active multichip module of size $6.5m \times 9m$ mounted on a $100 \mu m$ thin substrate using flipchip technology. The multichip module consisted of a telemetry chip, coil, ceramic surface mounted capacitor, pressure sensor, and a readout chip. The chip was mounted on a modified intraocular lens and the pressure sensor cavity was sealed by low-pressure chemical vapor deposition (LPCVD) sealing with low temperature oxide. The implanted coil has several windings made of $31 \mu m$ thin wire. The system operates at a frequency of 125 Hz with negligible absorption of RF field by the ocular fluid.

Schnakenberg et al. [50, 51] described the work of a pressure sensor and transponder integrated in the haptic of an artificial lens. The integrated electronics generate pulse width modulated signals representative of the measured pressure. An external RF field activates and powers the sensor chip without using a battery for the implant transponder. The RF activated sensor measures the pressure and transmits the data digitally to a remote reader unit located on the spectacles using absorption modulation technique. The sensor performance was evaluated in rabbit eyes and it showed promising results. The sensor disc size after the encapsulation was reported as 15 mm in diameter and 4.5 mm thick. Stangel et al. made significant improvements in the design of this system [52–54]. A temperature sensor was added along with pressure sensor array. The sensor chip consisted of pressure independent reference capacitors with oxide passivation on top along with capacitive pressure sensor to reduce parasitic effects. The receiver antenna of the implant has a diameter of 10.5 mm and was flipchip bonded to the implant chip. Integrated circuits were used to convert the pressure signals into pulsewidth modulated (PWM) signals. The telemetry link was tested using a transmitter coil of 40 mm diameter and two windings. The working distance between the coils was 3 cm when the sensor was implanted in rabbit eyes for six months.

Though active devices have better performance when compared with passive devices, the design is more complex. There is a need to come up with batch manufacturing techniques
to attach the antenna to sensor chip. In addition, almost all the devices are implanted after removal of the native lens of the eye. In patients who have not yet developed a cataract, this is a major irreversible surgical procedure in that the patient will end up with an intraocular lens implant just to measure IOP.

**Other IOP Measurement Techniques**

Fink et al. [55] described an optically powered and optically data transmitting wireless IOP sensor device consisting of a set of threshold switches that becomes activated sequentially when the IOP is increased above a predetermined threshold value. The device essentially consists of an IR photodiode, capacitor, resistor, pressure switch and a power source. An implanted solar cell with a light receiving area of $2-4 \text{mm}^2$ supplies $300-600 \mu W$ and acts as power source for the pressure switch system. The pressure switch is made of two electrodes mounted onto a compressible enclosure filled with either gas or vacuum. Whenever the IOP increases above a predetermined threshold value, the pressure switch closes the circuit thereby discharging the capacitor. An external readout circuit charges the capacitor. The external photo detector queries the wavelength specific photoreceptor periodically to check whether the capacitor remains charged or not. When the switch is activated, the charge in the capacitor emits light through the indicator LED detected by an external photo detector, if the IOP did not exceed the preset critical value since the last readout. This technique does not give a continuous IOP measurement. For each pressure level, there needs to be one pressure switch. Furthermore, it is not clear from the patent if this has been reduced to practice. No specific data is available. However, it is a different idea from the rest of the implantable sensors. Bae et al. [56, 57] conducted in vitro tests to measure the performance of an IOP sensor valve system. Piezoresistive sensor with constant excitation voltage was used for pressure sensing along with an electromagnetically actuated valve that controls fluid
flow. The sensor is made on a valve membrane suspended over silicon substrate. Wheatstone quarter bridge circuit is used to sense the change in resistance and relate it to the pressure. Noise is the limiting factor for piezoresistive sensors. The final size of sensor-valve system (9\text{mm} \times 9\text{mm} \times 2\text{mm}) is quite large to be implanted in the anterior chamber. Chen et al. [58–60] fabricated a passive pressure transducer that is completely mechanical. The device is a microbourdon parylene-C polymer tube fabricated using the micromachining process in the form of a 1 mm radius spiral. The device is pressurized to 1 atmosphere internally and sealed. When placed in a fluid, the external pressure on the tube causes the tube to elongate circumferentially. The end of the tube is optically tracked externally to read the pressure. Linear correspondence between the pressure and the angular displacement of the tube end has been reported. This device, although is simple and promising, may still require electronics if a continuous readout is required for tracking diurnal variations. Furthermore, the location of the external sensing optics relative to the implant will be crucial in measuring relative change in the orientation. Chen [61] fabricated a flexible parylene based capacitive pressure sensor that has a form factor of 4\text{mm} \times 1\text{mm}.

Table 1.1: Summary of coil based (wireless) design specifications

<table>
<thead>
<tr>
<th>Reference</th>
<th>Pressure Sensing Element</th>
<th>Reader to Implant Distance</th>
<th>Implant Size (mm)</th>
<th>Implant Location</th>
</tr>
</thead>
<tbody>
<tr>
<td>[38, 39]</td>
<td>Capacitor</td>
<td>22mm</td>
<td>5\times2</td>
<td>Hard PMMA intraocular lens</td>
</tr>
<tr>
<td>[41, 50]</td>
<td>Capacitor</td>
<td>Not reported</td>
<td>4\times4\times0.7</td>
<td>Artificial IOL</td>
</tr>
<tr>
<td>[42]</td>
<td>Capacitor</td>
<td>2mm</td>
<td>2.6\times1.6</td>
<td>Not reported</td>
</tr>
<tr>
<td>[43]</td>
<td>Capacitor</td>
<td>7.5mm</td>
<td>3\times3</td>
<td>Not reported</td>
</tr>
<tr>
<td>[49]</td>
<td>Capacitor</td>
<td>Not reported</td>
<td>6.5\times9</td>
<td>Not reported</td>
</tr>
<tr>
<td>[50–54]</td>
<td>Capacitor</td>
<td>30mm</td>
<td>10.3\phi</td>
<td>Soft PDMS IOL</td>
</tr>
</tbody>
</table>

The important design parameters of IOP sensing techniques are shown in Table 1.1 and
Table 1.2: Non-telemetry Techniques

<table>
<thead>
<tr>
<th>Reference</th>
<th>Pressure Sensing Element</th>
<th>Implant Size, mm</th>
<th>Implant Location</th>
</tr>
</thead>
<tbody>
<tr>
<td>[58, 59]</td>
<td>Tip rotation of a Parylene bourdon tube structure</td>
<td>$2\phi$</td>
<td>on top of Iris secured in place by Parylene anchors</td>
</tr>
<tr>
<td>[60]</td>
<td></td>
<td>$1.1\phi$</td>
<td></td>
</tr>
<tr>
<td>[56, 57]</td>
<td>Piezoresistive membrane</td>
<td>$9\times9\times2$</td>
<td>Not reported</td>
</tr>
<tr>
<td>[62, 63]</td>
<td>Strain gage</td>
<td>$11.5\phi$</td>
<td>Contact lens</td>
</tr>
</tbody>
</table>

II. In summary, there are two important factors that have not been given enough significance in existing designs.

1. Once implanted, the sensor is harder to retrieve without damaging the original state of the tissue surrounding the implant. This poses a problem when the implant needs to be removed either in the case of device failure or at the end of treatment/monitoring.

2. Attention to the spatial constraints in the eye, surgical complexity, and reliability will significantly improve the chances of design being successful.

To address the requirement of reversibility, recognition that the implant is not permanent is very important. A least intrusive method would not involve procedures such as penetrating the choroidal region or replacing the intraocular lens with an artificial one. While monolithic ultra miniaturization seems to be the general trend to address least intrusiveness requirement, the resulting weak coupling between the implant and the external coil seems to be the trade
off. However, an optimal miniaturization strategy, taking into account the anatomy of the anterior chamber is likely to succeed.

1.5 Thesis Outline

Chapter-2 discusses the design requirements of IOP sensor system. The implant location, available space for implant, and dimensional limits on implant size are also described in Chapter-2.

In Chapter-3, a simplified lumped element model of passive LC resonance based circuits is discussed. The model captures important aspects of the IOP sensor design. A POLYMUMPS process flow based MEMS capacitive pressure sensor design and the fabrication process is explained in Chapter-3.

Chapter-4 discusses a multicoil based IOP sensing. Multicoil based design eliminates the need for connecting the sensor coil with the sensor capacitor by making use of two sensor coils that are oppositely wound and separated by a small distance between them. In multicoil design approach, the via mask can be eliminated during the sensor implant fabrication process making the sensor fabrication process flow simple and less expensive. It also makes it possible to minimize the implant thickness. Flexible substrate based fabrication processes were developed and the sensor implant behavior is modeled in the same chapter. Analytical models of sensor coil inductances and parasitic capacitances were derived based on literature survey. Finally, sensor implant sensitivities for different configurations of the sensor diaphragm and coil sizes were quantified.

In Chapter-5, sensor and reader coils misalignment analysis was carried out to get estimates of coupling coefficient. Coupling coefficients were calculated for coils of different sizes and orientations. The numerical method was tested using test coil setup and the results
were discussed. Limiting angle of view of eyes were measured and the coupling coefficient for angularly misaligned sensor and reader coils was calculated.

Sensor implant insertion mechanics are covered in Chapter-7. An FEA model of 50µm thick ParyleneC based sensor implant was developed to model implant stress variation during the insertion process. Stress concentration regions of the sensor implant are identified. A way to insert the implant surgically without developing plastic strains in the implant is also explained.
REFERENCES


In addition to being biocompatible, the sensor implant should meet the following constraints that ultimately decide the overall effectiveness of the implant. The design constraints are listed in order of priority.

1. The implant and its location should not pose any interference with the normal functions of the eye.

2. In case of patient discomfort or if the doctor decides that there is no longer any need for implanted sensor, it should be easily removable from the patient’s eye via a minor surgery.

3. The form and size factor of the implant should be small enough so that the implant insertion procedure does not differ much from the standard surgical procedures such as cataract surgery.

4. The range of operation of the sensor implant should be long enough so that the reader coil circuitry does not have to be positioned very close to the patient’s eye which might
interfere with patient's daily activities

5. Pressure sensitivity of the sensor should be high enough so that it can capture diurnal variations of the intraocular pressure accurately.

6. The operation of the sensor should be made tolerably insensitive to the eye movement

7. The implant fabrication process should be adoptable for batch fabrication processes so that fabrication tolerances can be met consistently. Also, ease of batch fabrication decides the overall cost of the sensor. By making the sensor fabrication process tuned to standard batch fabrication processes, the sensor can be made affordable to larger section of patients.

All the above mentioned requirements are tried to be addressed in the current design flow.

2.1 Implant Location

The workable implant space available in the anterior chamber of the eye is considerably small. The suitable region for surgically placing and accessing the implant is the anterior chamber of the eye that has not been used efficiently by current devices for IOP sensing. Implants in the vitreous cavity have a higher risk of infection, retinal detachment and encapsulating fibrosis, thus despite the larger available space, this approach is not used. An Optical coherence tomography (OCT) image of the eye was obtained to estimate three dimensional volume model of the space available in the eye. The anatomy and size scales for the purposes of this discussion are shown in Figure 2.1. It can be seen that the space available in the anterior chamber is very small yet may be efficiently utilized. The thickness of the cornea is about
500\(\mu m\) along the optical axis and at the extremities it is about 1\(\mu m\). The overall diameter of the lens is about 16 mm.

![Figure 2.1: In vivo cross-sectional view of a live human eye anterior segment obtained using optical coherence tomography.](image1)

![Figure 2.2: An illustration of implant location on top of iris inside the anterior chamber](image2)

There are axial symmetries inside the eye that can be effectively utilized for device placement. A cylindrical region along the optical axis must be subtracted out, as it is not available for any device placement due to it being in the visual axis. The implant should not be close to
the irido-corneal angle, the junction where the iris, the cornea and ciliary body meet. Also, it should not be too wide as it might obstruct the iris movement. The flatter the device, better it is for minimal impact on the aqueous humor flow process and the danger of adhering the iris to the cornea. OCT image of eye revealed that the space available in the eye is in the form of a hollow open cylinder, 7 mm ID, 8 mm-12mm OD, and a cylinder height of less than 500 \( \mu \) m. The diameters of the sensor and reader coils are decided so that the sensor coil can fit within the allowed space in the anterior chamber, Table 2.1.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Specification</th>
</tr>
</thead>
<tbody>
<tr>
<td>Minimum inner diameter of the sensor coil ( S_{ID} )</td>
<td>7 mm</td>
</tr>
<tr>
<td>Maximum outer diameter of the sensor coil ( S_{OD} )</td>
<td>8 mm to 12mm</td>
</tr>
<tr>
<td>Minimum outer diameter of the reader coil ( R_{OD} )</td>
<td>8 mm</td>
</tr>
<tr>
<td>Minimum separation distance between the coils</td>
<td>20 mm</td>
</tr>
</tbody>
</table>

The minimum separation distance between the coils is estimated by measuring the average distance of separation between the outer surface of the eyeball and inside face of the standard edition eyeglasses worn by patients. A distance of 2cm is considered to be in the comfort zone of patients. A 3D model of implant inside the anterior chamber is shown in Figure 2.2. The sensor implant can be made non obstructive to normal fluid flow path of intraocular fluid by minimizing the thickness of the implant as much as possible.
3.1 Single Coil Passive IOP Sensor

Majority of passive LC pressure sensors that were discussed in Chapter 1 have single coil connected to a pressure sensitive capacitor that form sensor implant. In this setup, the sensor coil is electrically connected to the capacitor using a “via” mask. A time domain and frequency domain lumped parameter models of passive pressure sensor system are shown in Figure 3.1 and Figure 3.2.

Where, $I_1$ and $I_2$ are the loop currents. $V_I$ is the voltage of the reader coil. $L_p, L_s$ are the self-inductances of the reader and sensor coils. $C_s$ is capacitive pressure sensor with a fixed bottom electrode and a flexible top electrode and a sealed chamber between the two electrodes. $M$ is the mutual inductance between the two coils. $R_s$ and $R_p$ are the resistances of the sensor coil and primary coil. In Figure 3.1 and Figure 3.2, parasitic capacitances of the coils were not included. But the included circuit diagram captures significant behavior of the passive LC based resonant circuit. A much more detailed multi-coil based sensor implant
Figure 3.1: Time domain lumped parameter model of passive IOP sensor

Figure 3.2: Frequency domain lumped parameter model of passive IOP sensor
circuit will be discussed in Chapter-4. The resistance of the coil is given by equation (3.1).

\[
R = \frac{\rho l}{w \delta(1 - e^{-h/\delta})}
\]  

(3.1)

Where \(\rho\) is the electrical resistivity, \(l\) is the length of the coil, \(w\) is the width of inductor coil line, \(h\) is the height of the inductor coil line, and \(\delta\) is the skin depth given by

\[
\delta = \sqrt{\frac{\rho}{\pi f \mu}}
\]

\(\mu\) is the magnetic permeability of the coil material and \(f\) is the frequency in Hertz. Skin depth of two commonly used metals Copper and Gold is shown in Figure 3.3
Using KVL and mesh analysis,

\[ V_1 = R_p I_1 + I_1 j \omega L_p - I_2 j \omega M \]  
\[ 0 = I_2 R_s - I_2 \frac{j}{\omega C_s} + I_2 j \omega L_s - I_1 j \omega M \]

By solving Eq. 3.2 for the loop currents, the equivalent input impedance as seen by the primary voltage source can be derived as follows:

\[ Z_{in} = \frac{V_1}{I_1} \]  
\[ Z_{in} = R_p + j \omega L_p r + \frac{\omega^2 M^2}{R_s + j \omega L_s + \frac{1}{j \omega (C_s + \Delta C_s)}} \]

The membrane of the sensor deflects because of the pressure acting on the top side of it. This deflection changes the capacitance between the two electrodes. The change in capacitance affects the resonance frequency of the sensor circuit given by the expression (3.7).

\[ f_0 = \frac{1}{2\pi} \sqrt{\frac{1}{L_s (C_s + \Delta C_s)}} - \frac{R_s^2}{L_s^2} \]  
\[ f_0 = \frac{1}{2\pi} \sqrt{\frac{1}{L_s (C_s + \Delta C_s)}} \text{ if } R_s^2 \ll \frac{L_s}{C_s} \]

The sharpness of the resonance frequency is measured by quality factor \((Q)\) of sensor circuit given by the equation (3.19).
\[ Q_s = \frac{2\pi f_0 L}{R_s} = \frac{1}{2\pi f_0 R_s C_s} = \frac{1}{R_s} \sqrt{\frac{L_s}{C_s}} \]  

(3.9)

When the reader coil voltage is varied, at certain frequency, the implant sensor circuit goes into resonance which get reflected on the primary coil circuit through the interaction of the magnetic fields of the reader and sensor coils.

The impedance as seen from the reader coil can also be derived in terms of sensor coil quality factor, \( Q_s \) and coupling coefficient \( k \) [1]:

\[
\begin{align*}
Z_{in} &= \frac{V_i}{I_i} \\
Z_{in} &= R_r + j \omega L_r + \frac{\omega^2 k^2 L_r L_s}{R_s + j \omega L_s + \frac{1}{j \omega (C_s + \Delta C_s)}} \\
Z_{in} &= R_r + j 2\pi f_0 L_r \left[ 1 + \frac{k^2 (\frac{L_r}{f_0})^2}{1 - (\frac{L_r}{f_0})^2 + j \frac{1}{Q_s f_0}} \right]
\end{align*}
\]

(3.10 - 3.12)

In equation (3.10), \( f \) is the frequency of reader coil voltage and \( k \), the coupling coefficient defined by equation (3.13), the maximum value of which is equal to one.

\[ k = \frac{M}{\sqrt{L_r L_s}} \]  

(3.13)

At resonance, \( f = f_0 \) and input impedance of the reader coil becomes:

\[ Z_{in} = R_r + j 2\pi f_0 L_r (1 + j k^2 Q_s) \]  

(3.14)

The change in phase (\( \Delta \phi \)) of input impedance at resonance is approximated by equa-
Table 3.1: Reader Coil Impedance Design Parameters

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Specification</th>
</tr>
</thead>
<tbody>
<tr>
<td>Reader coil inductance, $L_R$</td>
<td>1 $\mu H$</td>
</tr>
<tr>
<td>Coupling coefficient, $k$</td>
<td>0.1</td>
</tr>
<tr>
<td>Implant circuit resonant frequency, $f_0$</td>
<td>35, 38, 40 MHz</td>
</tr>
<tr>
<td>Sensor coil quality factor, $Q_s$</td>
<td>40</td>
</tr>
</tbody>
</table>

It is preferable to measure the phase of input impedance instead of the magnitude because of measurable change in the phase in the sensor circuit resonance regime. This can be seen in Figure 3.4 and Figure 3.5. The circuit parameters that were used to calculate the phase and amplitude of reader coil impedance are shown in Table 3.1.

Coupling coefficient "k" is the key factor in limiting the wireless range of inductive telemetry sensor system. As it can be seen from Eq. (3.15), the change in the phase angle ($\Delta \phi$) is proportional to square of the coupling coefficient ($k$) which itself is a function of mutual inductance and self inductances of sensor and reader coils. When the reader and sensor coil axes are aligned, k can be approximated by equation (3.16) [3, 4].

$$k(z) \cong \frac{r_s^2 r_r^2}{\sqrt{r_s^2 r_r^2} \sqrt{d^2 + r_f^2}^3}$$

(3.16)

By measuring the phase of the input impedance, one can see a drop in the phase angle when operating frequency equals the sensor resonant frequency. Fonseca [1] calculated the relationship between the measured frequency when the phase of the equivalent impedance
becomes the lowest to the resonant frequency of the implant circuit as

$$f_{min} = f_0 \left( 1 + \frac{k^2}{4} + \frac{1}{8Q^2} \right)$$

(3.17)

Where,

$$Q = \frac{\omega_0 L_s}{R_s}$$

(3.18)

Q can also be defined as [5]

$$Q = 2\pi \frac{maximum \ energy \ stored}{total \ energy \ lost \ per \ period}$$

(3.19)

From Eq. 3.18, it can be seen that sharpest drop in the phase angle occurs when the
implant inductor coil has high inductance and low series resistance. Figure 3.6 illustrates the
effect of Q on the phase of reader circuit impedance. The design parameters needed for the
calculation of the impedance are shown in Table 3.2.

Slight misalignment in the axes changes mutual inductance between the coils. As the
distance between coils increases, the coupling coefficient decreases which in turn lowers
the phase dip. This makes it difficult to identify the resonance frequency of sensor circuit
because of smaller value of reflected impedance in the primary circuit because of weak

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Specification</th>
</tr>
</thead>
<tbody>
<tr>
<td>Reader coil inductance, $L_p$</td>
<td>2.5 $\mu$H</td>
</tr>
<tr>
<td>Coupling coefficient, $k$</td>
<td>0.12</td>
</tr>
<tr>
<td>Implant circuit resonant frequency, $f_0$</td>
<td>23 MHz</td>
</tr>
</tbody>
</table>
coupling. This problem becomes more acute in the design of intraocular pressure (IOP) sensor because of the movement of the eyeball relative to the external coil. In order to design a robust passive telemetry system, the designer needs to identify tolerable limits of coupling coefficient so that the sensor data can be accurately measured even when the reader and sensor coil axes are slightly misaligned.

### 3.2 Surface micromachined MEMS sensor fabrication

Capacitive pressure sensor is well suited to measure intraocular pressure compared to other kind of pressure sensors such as Piezo-resistive pressure sensors. This is because capacitive sensors exhibit monotonic response under increased applied pressure [6]. Polysilicon was selected to be used as structural material for the sensor diaphragm because of low stress (< 1MPa) compared to other materials [7]. The low stress in the films avoids sensitivity clustering.
A capacitive pressure sensor was fabricated by utilizing a commercially available three layer polysilicon surface micromachining process called POLYMUMPS [8]. Making use of commercial MEMS fabrication process cuts down sensor unit cost and also provides reproducibility because of well studied process parameters. In this process, Polysilicon is used as the structural material, oxide layer (PSG) is used as the sacrificial layer, and silicon nitride is used as electrical isolation between the polysilicon and the substrate. Non-planar surface micromachined pressure sensor has reference pressure cavity above the silicon wafer surface. The reference cavity is formed by etching away the sacrificial oxide. Since the IOP range is very small (10 mmHg to 50 mmHg above atmospheric pressure), surface micromachining fabrication process was deemed to be more suitable compared to bulk micromachining process. Surface micromachining based sensor fabrication does not require additional fabrication steps such as wafer bonding, chemical mechanical polishing (CMP). Surface micromachined sensor also has small form factor compared to bulk micromachined sensor.

Using the fabrication process data, a finite element model of the MEMS pressure sensor was developed to calculate the maximum deflection of the movable plate. A plane strain model of the capacitive pressure sensor is shown in Figure 3.7. On the top surface of sensor, a pressure load of 50 mmHg above atmospheric pressure was applied. The stress distribution
Figure 3.8: Stress distribution in a 100$\mu$m $\times$ 100$\mu$m cap sensor at a pressure load of 50mmHg above atmospheric pressure [9]
Table 3.3: Movable electrode plate deflection

<table>
<thead>
<tr>
<th>Plate size, µm × µm</th>
<th>Maximum Deflection, µm</th>
</tr>
</thead>
<tbody>
<tr>
<td>100µm × 100µm</td>
<td>0.07</td>
</tr>
<tr>
<td>140µm × 140µm</td>
<td>0.20</td>
</tr>
<tr>
<td>180µm × 180µm</td>
<td>0.54</td>
</tr>
</tbody>
</table>

in a capacitive sensor of plate size 100µm × 100µm is shown in Figure 3.8. The deflections of cap sensors of different sizes were calculated from finite element analysis (Table 3.3). As expected, 100µm × 100µm cap sensor is stiffest among the three.

A wide variety of capacitive pressure sensors were fabricated by following a three layer Polysilicon surface micromachining process. This process has the general features of a standard surface micromachining process: (1) polysilicon is used as the structural material, (2) deposited oxide (PSG) is used as the sacrificial layer, and silicon nitride is used as electrical isolation between the polysilicon and the substrate.

The process begins with 150 mm n−type (100) silicon wafers of 1–2 Ω−cm resistivity. The surface of the wafers are first heavily doped with phosphorus in a standard diffusion furnace using a phosphosilicate glass (PSG) sacrificial layer as the dopant source. This helps to reduce or prevent charge feedthrough to the substrate from electrostatic devices on the surface. Next, after removal of the PSG film, a 600 nm low-stress LPCVD (low pressure chemical vapor deposition) silicon nitride layer is deposited on the wafers as an electrical isolation layer. This is followed directly by the deposition of a 500 nm LPCVD polysilicon filmPoly0.

Poly0 is then patterned by photolithography, a process that includes the coating of the wafers with photoresist, exposure of the photoresist with the appropriate mask and developing the exposed photoresist to create the desired etch mask for subsequent pattern transfer into the underlying layer (Figure 3.9.1).

After patterning the photoresist, the Poly0 layer is then etched in a plasma etch system.
Figure 3.9: MEMS capacitive sensor fabrication step sequence [9]
A 2.0\textit{\mu}m phosphosilicate glass (PSG) sacrificial layer is then deposited by LPCVD (Figure 3.9.3) and annealed @1050°C for 1 hour in argon. This layer of PSG, known as First Oxide, is removed at the end of the process to free the first mechanical layer of polysilicon. The sacrificial layer is lithographically patterned with the DIMPLES mask and the dimples are transferred into the sacrificial PSG layer in an RIE (Reactive Ion Etch) system, as shown in Figure 3.9.4. The nominal depth of the dimples is 750 nm. The wafers are then patterned with the third mask layer, ANCHOR1, and reactive ion etched (Figure 3.9.5). This step provides anchor holes that will be filled by the Poly 1 layer. After etching ANCHOR1, the first structural layer of polysilicon (Poly 1) is deposited at a thickness of 2.0\textit{\mu}m. A thin (200 nm) layer of PSG is deposited over the polysilicon and the wafer is annealed at 1050°C for 1 hour (Figure 3.9.6). The anneal dopes the polysilicon with phosphorus from the PSG layers both above and below it. The anneal also serves to significantly reduce the net stress in the Poly 1 layer.

The polysilicon (and its PSG masking layer) is lithographically patterned using a mask designed to form the first structural layer POLY1. The PSG layer is etched to produce a hard mask for the subsequent polysilicon etch. The hard mask is more resistant to the polysilicon etch chemistry than the photoresist and ensures better transfer of the pattern into the polysilicon. After etching the polysilicon (Figure 3.9.7), the photoresist is stripped and the remaining oxide hard mask is removed by RIE.
After Poly 1 is etched, a second sacrificial PSG layer (Second Oxide, 750nm thick) is deposited and annealed (Figure 3.9.10). The Second Oxide is patterned using two different etch masks with different objectives. The \textit{POLY1POLY2VIA} level provides for etch holes in the Second Oxide down to the Poly 1 layer. This provides a mechanical and electrical connection between the Poly 1 and Poly 2 layers. The \textit{POLY1POLY2VIA} layer is lithographically patterned and etched by RIE (Figure 3.9.9). The ANCHOR2 level is provided to etch both the First and Second Oxide layers in one step, thereby eliminating any misalignment between separately etched holes. More importantly, the ANCHOR2 etch eliminates the need to make a cut in First Oxide unrelated to anchoring a Poly 1 structure, which needlessly exposes the substrate to subsequent processing that can damage either Poly 0 or Nitride. The ANCHOR2 layer is lithographically patterned and etched by RIE in the same way as \textit{POLY1POLY2VIA}.

The second structural layer, Poly 2, is then deposited (1.5\,\mu m thick) followed by the deposition of 200 nm PSG. As with Poly 1, the thin PSG layer acts as both an etch mask and dopant source for Poly 2. The wafer is annealed for one hour at 1050 °C to dope the polysilicon and reduce the residual film stress. The Poly 2 layer is lithographically patterned with the seventh mask (POLY2). The PSG and polysilicon layers are etched by plasma and RIE processes, similar to those used for Poly 1. The photoresist then is stripped and the masking oxide is removed (Figure 3.9.10).

\subsection{Sensor Release}

The final deposited layer in the PolyMUMPs process is a 0.5\,\mu m metal layer that provides for bonding, electrical routing. The wafer is patterned lithographically with the eighth mask (METAL) and the metal is deposited and patterned using lift-off. The final, unreleased structure is shown in Figure 3.9.10. The wafer was then diced for sacrificial release and
Figure 3.11: Poly0 line after HF etch

test. Figure 3.10 shows the device after sacrificial oxide release. The release is performed by immersing the chip in a bath of 49% HF (room temperature) for $5 - 12$ minutes. This is followed by several minutes in DI water and then alcohol to reduce stiction followed by 10 minutes in an oven at 110°C. An SEM image of Poly0 line after HF etch showed that there was no significant etching of Polysilicon (Figure 3.11).

After HF etch, the sensor chips were transferred to a CO$_2$ critical point dryer chamber so that any liquid that is remaining between Poly0 layer and Poly1 layer can be evaporated. Critical point drying avoids stiction issues caused by any remaining traces of liquid between the sensor electrodes. The patterned Poly0 layer acts as a bottom electrode and Poly1 together with Poly2 act as top electrode plate of the capacitive pressure sensor. In order to increase the bending stiffness of the movable top electrode, Poly1 and Poly2 layers were deposited one on top of another. After the critical point drying of the sensor, the etch channels that provide access to the HF solution through the sensor side walls are closed by a blanket coating of Parylene. Pressure sensors of different plate sizes were fabricated on a die size of 5 mm x 5 mm (Figure 3.12). The etch channels and the bottom and top electrode lines are shown in Figure 3.13. During the etching process, the HF solution goes under the top electrode
through the etch channels and etches the sacrificial oxide. The dimple in the center of the diaphragm in Figure 3.13 prevents the top electrode from getting in contact with the bottom electrode when excessive pressure is applied on the top face of the diaphragm.

The thickness of the sensor plate was increased in different combinations. Poly1Poly2Via mask was used to deposit Poly2 on top of poly1 layer. In addition to the individual variable capacitors, some were connected in parallel. The geometry of the capacitive pressure sensors can be seen clearly in Fig 3.14. The diaphragm thickness were also increased by sandwiching a layer of Oxide between Poly1 and Poly2 layers. The capacitances of these sensors for 8 minute etch time are shown in Table 3.4.

To study the effect of residual sacrificial oxide on the sensor performance, two sets of sensors were released differently by using different etch times of 8 minutes and 12 minutes. Capacitance at atmospheric pressure was measured for both sets of sensors. The results are
Figure 3.13: Capacitive pressure sensors with etch channels in the side walls. [9]

Figure 3.14: Parallel MEMS capacitive sensors [9]
Table 3.4: Capacitor combinations

<table>
<thead>
<tr>
<th>Plate Dimension, µm</th>
<th>Layers</th>
<th>etch time, mins</th>
<th>capacitance, pF</th>
</tr>
</thead>
<tbody>
<tr>
<td>100</td>
<td>Poly1</td>
<td>8</td>
<td>3.404</td>
</tr>
<tr>
<td>100</td>
<td>Poly1+Poly2</td>
<td>8</td>
<td>2.554</td>
</tr>
<tr>
<td>100</td>
<td>Poly1+Poly2+Oxide2</td>
<td>8</td>
<td>NA</td>
</tr>
<tr>
<td>120</td>
<td>Poly1+Poly2</td>
<td>8</td>
<td>3.076</td>
</tr>
<tr>
<td>120</td>
<td>Poly1+Poly2+Oxide2</td>
<td>8</td>
<td>4.071</td>
</tr>
<tr>
<td>140</td>
<td>Poly1+Poly2</td>
<td>8</td>
<td>3.076</td>
</tr>
<tr>
<td>140</td>
<td>Poly1+Poly2+Oxide2</td>
<td>8</td>
<td>3.247</td>
</tr>
<tr>
<td>160</td>
<td>Poly1+Poly2</td>
<td>8</td>
<td>4.105</td>
</tr>
<tr>
<td>160</td>
<td>Poly1+Poly2+Oxide2</td>
<td>8</td>
<td>4.117</td>
</tr>
<tr>
<td>180</td>
<td>Poly1+Poly2</td>
<td>8</td>
<td>4.481</td>
</tr>
<tr>
<td>180</td>
<td>Poly1+Poly2+Oxide2</td>
<td>8</td>
<td>4.1529</td>
</tr>
</tbody>
</table>

Table 3.5: Capacitance measured at Atmospheric Pressure, 8mins etch time

<table>
<thead>
<tr>
<th>Plate Side Length, µm</th>
<th>Plate Layer</th>
<th>Capacitance, pF</th>
</tr>
</thead>
<tbody>
<tr>
<td>100</td>
<td>Poly1</td>
<td>3.34</td>
</tr>
<tr>
<td>100</td>
<td>Poly1+Poly2</td>
<td>2.55</td>
</tr>
<tr>
<td>140</td>
<td>Poly11+Poly2</td>
<td>3.08</td>
</tr>
<tr>
<td>160</td>
<td>Poly1+Poly2</td>
<td>4.11</td>
</tr>
<tr>
<td>180</td>
<td>Poly1+Poly2</td>
<td>4.48</td>
</tr>
</tbody>
</table>

Table 3.5 shows no residual sacrificial oxide left after the HF etch, the sensor chips were studied under an X-Ray photoelectron spectroscopy machine to obtain chemical state information for compound identification. As the etch time was increased, the residual oxide count became smaller (Figure 3.15, Figure 3.16). This explains the decrease in capacitance between the plates of equal dimensions from 8 minutes etch time to 12 minutes etch time. Twelve minutes in HF bath seemed to remove all the oxide between bottom Poly0 layer and the top Poly1 layer.
Table 3.6: Capacitance measured at Atmospheric Pressure, 12mins etch time

<table>
<thead>
<tr>
<th>Plate Side Length, µm</th>
<th>Plate Layer</th>
<th>Capacitance, pF</th>
</tr>
</thead>
<tbody>
<tr>
<td>100</td>
<td>Poly1</td>
<td>2.15</td>
</tr>
<tr>
<td>100</td>
<td>Poly1 + Poly2</td>
<td>2.12</td>
</tr>
<tr>
<td>140</td>
<td>Poly11</td>
<td>2.22</td>
</tr>
<tr>
<td>160</td>
<td>Poly1 + Poly2</td>
<td>2.37</td>
</tr>
<tr>
<td>180</td>
<td>Poly1 + Poly2</td>
<td>2.42</td>
</tr>
</tbody>
</table>

Figure 3.15: 8 minute etch spectra

Figure 3.16: 12 minute etch spectra. The oxide count is found to be statistically insignificant.
3.3 Effectiveness of Silicon based Sensor for IOP measurement

Even though, surface micromachined cap sensors have better sensitivity with respect to pressure, sacrificial oxide etching process becomes unreliable for diaphragms of larger size. Any residual oxide between the plates after the release process will alter the load-deflection behavior of the cap sensor and results in loss of accuracy. This is evident from the data in Table 3.5 and Table 3.6. Etch channel widths can be increased to provide better access to the HF etchant during the release process. Since Silicon is not bio-compatible, it has to be coated with a bio-compatible material such as Parylene. High surface conformity of Parylene causes the Parylene layer to get under the top electrode during the vapor deposition process.

Even after successful release, the cap sensor has to be attached to the implant coil via microbonding processes. This makes it difficult to release both sensor and coil together. Any bending of the coil would increase the stress concentration at the Coil-Sensor bondpads and there is a chance that bondpad could get dislodged from the sensor substrate. The thickness of the overall device will be more than 300$\mu m$ because of the Silicon substrate on which the cap sensor is built. This makes it harder for the device to be inserted into the anterior chamber through a small incision.
REFERENCES


4.1 Multi-coil Sensor Implant Design

The need for flexible substrate based sensor design is evident from the sensor insertion related constraints. Therefore, instead of focusing on integrating MEMS coil to a surface micromachined silicon pressure sensor, fabrication processes that will result in flexible implant were investigated. Elimination of bonding of individually diced capacitive pressure sensor to a MEMS coil makes sensor much reliable because there will be no uncertainty in the quality of the electrical bond region after the sensor is implanted inside the eye.

In this chapter, a multi-coil based sensor implant design is discussed. In this approach, the sensor has two coils instead of one and the capacitance is distributed between the two coils. Distributed capacitance based circuits were discussed in by several groups [1–3]. In this implementation of distributed LC sensor, the sensor coils are not connected using a “Via” mask. The distributed capacitors and inductors eliminate the need for a physical metal contact between the two layers. This concept is explained using Figure 4.1. It shows two
spiral coils with a circular diaphragm in the middle. A sealed chamber under the circular diaphragms allows them to deflect when external pressure is applied on them.

The distributed electrical equivalent circuit for the sensor shown in Figure 4.1 is shown in Figure 4.2. In Figure 4.2, $L_s$, $R_s$ are the self-inductance and resistance of the spiral coils. $C_{par}$ is the parasitic capacitance of the coil and $L_{ss}$ is the mutual inductance between the two spiral coils. The two coils need to be wound in opposite directions for the mutual inductance...
to act in a complimentary manner.

\[ C_{oveeq} = \frac{1}{2} (C_{ove} + C_{sensor}) \]

Where, \( C_{ove} \) is the overlap capacitance between the spiral coil turns and \( C_{sensor} \) is the pressure sensitive capacitance [1]. The spiral inductor by itself is usually using a \( \pi \) circuit as shown in Figure 4.3. The substrate capacitances and resistances are ignored for the spiral inductor model. The imaginary part of the impedance of the inductor circuit can be set to zero to get the self-resonant frequency of the inductor. The inductor resonant frequency and the quality factor (Q) are shown in Eq. 4.1:

\[
f_r = \frac{1}{2\pi \sqrt{L_s C_{par}}} \sqrt{1 - \frac{C_{par} R_s^2}{L_s}} \]
\[
Q_{coil} = \frac{\omega L_s}{R_s} \left( 1 - \frac{C_{par} R_s^2}{L_s} - C_{par} L_s \omega^2 \right)
\]

Low values of parasitic capacitance offset the sensor resonant frequency away from the overall sensor resonant frequency. In tightly wound spiral coils, the high value of parasitic capacitance makes the resonant frequency of coil go down and reach close the resonant frequency of the sensor. This makes it harder to detect the pressure value using the reader circuitry. The multiplication factor in the quality factor equation is the reduction in Q due
to an increase in electrical energy with frequency. At the resonant frequency of the coil, the quality factor becomes zero. The inductance of circular spiral coils can be calculated by using concentric circle approximation of planar spiral coils. According to this method, the self-inductance of singular circular loop of wire radius “R” and loop radius “a” is given by:

\[
L(a,R) = \mu_0 a \left( \ln \frac{8a}{R} - 2 \right)
\]  

(4.3)

The above equation is valid when \( \frac{R}{a} \ll 1 \). For a coil composed of \( N_a \) concentric circular loops with different radii ranging from \( a_i = (i = 1,2,..N_a) \), and with wire-radius R,

\[
L_a = \sum_{i=1}^{N} L(a_i,R) + \sum_{i=1}^{N_a} \sum_{j=1}^{N_a} M(a_i,a_j)(1 - \delta_{ij})
\]  

(4.4)

The second method uses a current sheet approximation for calculating self-inductance of planar spiral coil [4, 5]:

\[
L \approx \frac{\mu_0 n^2 d_{avg} C_1}{2} \left[ \ln \frac{C_2}{F} + C_3 F + C_4 F^2 \right]
\]  

(4.5)

Where \( n \) is the no.of coils, \( d_{avg} \) and \( F \) are defined as:

\[
d_{avg} = 0.5(d_{out} + d_{in})
\]  

(4.6)

\[
F = \frac{d_{out} - d_{in}}{d_{out} + d_{in}}
\]  

(4.7)

\( d_{out} \) and \( d_{in} \) are the outer and inner diameters of the coil. \( C_1, C_2, C_3, C_4 \) are the geometry based constants.
The overlap capacitance \( C_{ove} \) is given by the expression [6]:

\[
C_{ove} = \frac{\varepsilon_0 \varepsilon_r A_o}{t_d}
\]  

(4.8)

where \( A_o \) is the overlap area of the sensor coils and \( t_d \) is thickness of the dielectric between the coils.

The reader coil input impedance for a multi-coil sensor is given by [1]:

\[
Z_{readercoil} = s L_r - \frac{2 C_{eq} M^2 s^3}{1 + s C_{eq} R_s + C_{eq} L_{eq} s^2}
\]  

(4.9)

where

\[
C_{eq} = C_{oveeq} + C_{par}
\]

(4.10)

\[
L_{eq} = L_s (1 + k_s)
\]

(4.11)

\[
M = k \sqrt{L_r L_s}
\]

(4.12)

\( k_s \) is the coupling coefficient of the sensor coils. The reader coil impedance can be expressed in terms of sensor resonant frequency and quality factor as

\[
Z_{readercoil}(f) = j 2 \pi f L_r \left( 1 + k^2 \left( \frac{2}{1 + k_s} \right) \left( \frac{f^2}{f_0^2} \right) \left( \frac{f}{f_0} \right) \right)
\]

(4.13)
Two design methodologies are presented based on multicoil sensor circuit. In each methodology, the fabrication process flow of the sensor implant, and the implant performance is described in detail. The first type of sensor design uses copper clad Kapton® as the sensor substrate. The second sensor design approach uses ParyleneC as the sensor substrate. Both the design flows allow the sensor implant to be flexible so that it can be easily implanted inside the eye.

### 4.2 Kapton Based Sensor Implant Design

Kapton® is a polyimide film manufactured by Dupont. Excellent biocompatibility of polyimide makes it a good choice for sensor substrate. Kapton films are manufactured with electro deposited copper cladding on either single side or both sides of the film. For the purpose of IOP sensor design, a single sided copper clad Kapton film of thickness 12µm was chosen. The sensor fabrication process flow is shown in Figure 4.4.

The process starts with two single side copper clad Kapton films (*Polyimide 12µm + Copper 9µm*). The films are patterned so that the spiral traces along with the capacitor plates are transferred from the mask to the film. Standard photolithography techniques along with wet chemical etching of copper can be used for pattern transfer. The two patterned Kapton films form the two layers of the multicoil design explained in the previous section.
The inner layer bonds the two layers together and has a cutout using laser to create embedded cavity. This approach results in an intrinsically packaged structure. The Bondply layer is made of Kapton with adhesive coating on both sides \((Adhesive_{13\mu m} + Kapton_{13\mu m} + Adhesive_{13\mu m})\). Laminating conditions are 190\(\text{degreeC}\) with a pressure of 14-28 \(kg/cm^2\) for 1-2 hours at the specified temperature and pressure. After the curing of adhesive, only outer surfaces of polyimide layer are exposed to the environment. A cross-sectional view of sensor is shown in Figure 4.5.

Kapton based coils fabricated using screen printing technique are shown in Figure 4.6. Better tolerances of the sensor coils can be obtained by using cleanroom processes.

### 4.2.1 Kapton Pressure Sensor Design

The implant design process starts with the design of pressure sensor. The goal is to obtain smaller form factor of the sensor without compromising the sensitivity of the sensor. The deflection of circular diaphragm of MEMS capacitors operating in small deflection regime is
Figure 4.5: Cross-sectional view of Kapton based IOP sensor

Figure 4.6: Coils made using screen printing technique
given by equation (4.17) [7].

\[
    w(r) = \frac{\Delta P a^4}{64D} \left(1 - \left(\frac{r}{a}\right)^2\right) \tag{4.17}
\]

Where \( \Delta P \) is the difference in pressure on both sides of the diaphragm, \( a \) is the radius of the diaphragm, and \( D \) is the flexural rigidity of the diaphragm. A more accurate measure of the deflection can be obtained by using large deflection theory. When the center deflection \( d_0 \) becomes greater than half the thickness of the plate, the plate acts like a shallow shell and supports the lateral load as a membrane rather than a flexural structure. Large deflections of clamped circular plates under lateral load is [8]

\[
    P = \frac{64D}{R^3} \left(\frac{d_0}{R}\right) + \frac{8}{3(1-\nu)} \left(\frac{t}{R}\right) \left(\frac{d_0}{R}\right)^3 \tag{4.18}
\]

Where, \( P \) is the pressure acting on the plate, \( D \) is the flexural rigidity of the plate, \( R \) is the radius, \( t \), the thickness, and \( \nu \) is the poisson's ratio. The flexural rigidity of homogeneous plate of uniform thickness “t” and Young's modulus “E”is given by

\[
    D = \frac{Et^3}{12(1-\nu^2)} \tag{4.19}
\]

In the case of a two layer stacked plate of thicknesses \( t_1, t_2 \) and Young's moduli \( E_1, E_2 \), the equivalent flexural rigidity can be calculated by using the following equations [9]:

\[
    D = E_1 t_1^2 \left(\frac{t_1}{3} - \frac{Z_n}{2}\right) + E_2 t_2^2 \left(t_1^2 + t_1 t_2 + \frac{t_2^2}{3} - \left(t_1 + \frac{t_2}{2}\right)Z_n\right) \tag{4.20}
\]
where,

\[ Z_n = \frac{E_1 e t_1^2 + E_2 e t_2(2t_1 + t_2)}{2(E_1 e t_1 + E_2 e t_2)} \]  \hspace{1cm} (4.21) \\
\[ E_{ie} = \frac{E_i}{1 - v_i^2} \]  \hspace{1cm} (4.22)

Even though the above mentioned equations are applicable for stacked plates with clamped edges, they can be used to estimate the plate deflection at certain load. Since the relationship between the deflection and applied load is nonlinear, the plate deflection should be within 20%-25% of the enclosed chamber height so that the sensitivity of the capacitor stays linear [10, 11]. For given maximum pressure load, Eq. 4.18 can be used to get an estimate capacitor plate radius that has a center deflection of value equal to 25% of the chamber height. The updated equation for radius estimation is

\[ R = \sqrt[3]{\frac{1}{P} \left( 64Dd_0 + \frac{8}{3(1 - v)} t d_0^3 \right)} \]  \hspace{1cm} (4.23)

Using Eq. 4.23, estimates of radii of cap sensor for a range of maximum applied pressure from 20mmHg to 150mmHg above atmospheric pressure were obtained (Figure 4.7). The thickness of the Bondlayer decreases after the lamination from 39\( \mu m \) to anywhere between 34\( \mu m \) to 29\( \mu m \) because of the applied compressive pressure during the lamination process. The adhesive reflows slightly at elevated temperatures during the bonding process thereby reducing the height of the enclosed chamber. In this design, Kapton pressure sensors were modeled for different Bondlayer thicknesses. Since the total thickness of the diaphragm at the center is (Kapton12\( \mu m \) + Copper9\( \mu m \)) is 21\( \mu m \), the deflection of the sensor diaphragms is small at low pressures. Therefore, an etched Kapton layer of 4\( \mu m \) along with copper layer of 9\( \mu m \) was also studied.
In case of closed form solutions, the capacitance between the sensor plates can be calculated by using the equation

$$C = \int_S \frac{\varepsilon_0 \varepsilon_r dS}{h - d(r)}$$

(4.24)

Analytical solution of exact center deflection is not available in the literature because of non-standard boundary condition along the diaphragm edge. Therefore, the sensors were modeled using CoventorWare, a multi-physics based finite element software. After obtaining the nodal deflections of the plate, an electrostatics simulation was performed to obtain the capacitance between the plates. In the following discussion, the copper thickness is modeled to be equal 9µm unless otherwise stated. Also, Kapton diaphragm means a diaphragm that has Kapton on top and Copper on the underside of it. Only edges of the Kapton layer are clamped. The copper radius was set smaller than that of Kapton so that it can bend freely at the edges. This lowers the stiffness of the diaphragm and helps increase the sensitivity of the sensor.
4.8.1: Deflection of 4micron thick Kapton diaphragm.

4.8.2: Radial pattern on Copper to lower bending stiffness.

4.8.3: Deflection of 4micron thick grooved Kapton diaphragm.

Figure 4.8: Kapton diaphragm, 250 microns radius
In the first run, a Kapton diaphragm of 250µm radius was modeled and deflection is shown in Figure 4.8. In order to increase the deflection of the diaphragm, radial grooves were patterned on copper. The length of the radial grooves was kept less than that of the radius of the copper plate to maintain it as a single conductor. The sensor deflection increased slightly but not large enough to obtain good sensitivity in the pressure range of 10mmHg-30mmHg. In the next set of simulations, the radius of Kapton diaphragm was increased to 600µm. The center deflection and the resulting capacitance change was calculated for diaphragm thicknesses 12µm and 4µm and Bondply thickness values of 34µm and 29µm Figure 4.9.

Figure 4.9: Capacitance Vs Pressure of Kapton pressure sensor of 600µm radius

Modal analysis on the sensor was performed to extract natural resonant frequencies and the mode shapes of the movable diaphragm structure. At these resonant frequencies, the system transfer function becomes infinite. Frequencies and their associated mode shapes are of particular interest in design because they closely resemble the corresponding characteristics of an underdamped mechanical system, and they indicate when the system will have its maximum response to an intended or unintended (noise) input.
Figure 4.10: Mode shapes and frequencies of 600 micron Kapton Sensor

mode shapes were normalized to a maximum deflection of $1\mu m$ since they do not represent the actual amplitudes of the structure. Two parameters that cause a shift in the resonant frequencies of the cap sensor structure are the residual stress in the membrane and applied external pressure load. An external pressure of 50mmHg above atmospheric pressure was applied on top of the diaphragm. The natural frequencies and mode shapes are shown in Figure 4.10. The first natural frequency is found to be equal to 30 kHz which is well beyond the possible frequency of intraocular pressure changes in the eye.

Since the lamination process is done under pressure and at elevated temperature, the sensor diaphragm will have residual in plane tensile stress. Since Copper layer is bonded on to the Kapton layer, any residual stress in the Copper layer will alter bending characteristics of the sensor and thereby changes the sensitivity of the sensor. To model the effects of residual stress on sensor performance, an initial in-plane tensile stress equal to 10 MPa was applied in the copper layer region and a parametric study was done to obtain the sensor deflection at
residual stresses of values up to 50 MPa. The results are shown in Figure 4.11. As the stress increases, the effective bending stiffness of the membrane increases, which in turn lowers the capacitance at the same applied pressure as before.

![Graph](image-url)

4.11.1: Kapton 4micron thick, 600microns radius

4.11.2: Kapton 4micron thick, 600microns radius

Figure 4.11: Effect of in-plane residual stress on sensor performance

Spring-softening is another problem that is common in capacitance based sensors. As the external pressure increases, the plate deflection increases, which in turn increase the strength of the electric field between the plates for a given applied voltage to the plates. In order to identify any possible spring softening effect, the voltage on the plates was increased and the resulting deflection behavior was observed Figure 4.12. The voltage has negligible effect on the deflection of the membrane and the voltage in a passive LC sensor will not be very high because of coupling loss between the reader and sensor coils.
4.2.2 Kapton Implant

The diameter of the pressure sensor influences the overall dimensions of the Kapton implant. Unlike traditional passive LC sensors, the pressure sensor diaphragm cannot be situated at the center of the coils because it would obstruct the vision of the patient. Therefore the sensor has to be located along the periphery. The sensor diaphragms can be located either on the inside periphery or the outside periphery of the sensor coils. Placing the pressure sensor plate on the inside periphery of the coils has an added advantage in that the sensor coil diameter can be larger than that of the other case. Larger sensor coils have better coupling related performance. A 3D view of the completed sensor implant is shown in Figure 4.13.

Based on the above discussed configuration, the exact dimension of the Kapton implant were designed. The dimensions are shown more detail in Figure 4.14.

The inside diameter of the implant is 7mm and the outside diameter is 12mm. The width of the implant is 2.5mm. The area of the implant is divided into concentric circular ring areas of different widths. The bonding area is the region where the bondply makes contact with the polyimide film. Since the quality of the bonding is dependent on the re-flow characteristics of the bondply adhesive during the lamination, the bonding area is made to be equal to...
400μm. Adjacent to the bonding area is the sensor area. This ring area can be made useful to connect multiple capacitive pressure sensors to increase the overall plate capacitance and the pressure sensitivity of the implant Figure 4.13.

A multi-diaphragm parallel connected pressure sensors were modeled using FEA. The results are shown in Figure 4.15

As expected, the plate capacitance has increased significantly but the overall sensitivity of the sensor implant did not differ much from the case when individual sensor was used. This is probably because of the higher bending stiffness of annular Kapton diaphragm than
that of a circular diaphragm. The sensor area is followed by another section of bonding area that bonds polyimide and bondply layers. The outermost region is the coil region of 500 \( \mu m \) width.

The number of turns of coils that can be fit within the coil region for different values of coil pitch, trace width, and space between the turns are shown in Figure 4.16

The inductance and resistance of coils of different pitches were calculated using the analytical equations that were discussed in section 4.1. Fasthenry [12] and Fastcap software were also used to extract inductance, resistance, and parasitic capacitances of the coils of different configurations. The results are shown in Figure 4.17
Figure 4.16: Number of coil turns

4.17.1: Inductance of coils.

4.17.2: Resistance of coils.

4.17.3: Parasitic capacitance of the coils.

Figure 4.17: Kapton implant coil electrical properties
The coil inductance remains almost constant during the frequency range of 1MHz-250MHz. The inductance is maximum for a sensor coil of 20$\mu$m pitch because of more number of turns. The increase in inductance is offset by the increase in coil series resistance. The resistance is a function of the length of the coil. Therefore, as the number of turns increase, the resistance also increases. The parasitic capacitance of the coils are calculated both in air and in Kapton. The parasitic capacitance of coils embedded in Kapton is about 3 times higher than that of the coils in air. Also, the change in the parasitic capacitance with respect to the pitch of the coils is more for coils in Kapton than those in air.

The extracted inductor parameters are used to calculate the quality factors of coils of different types Figure 4.18. The quality factor is high for coils in air because of low parasitic capacitance. Also, the coil with least number of turns has highest quality factor among all the remaining coil types in both air and Kapton. This is because of low parasitic capacitance and low series resistance. But choosing a coil of 3 turns will lower the coupling coefficient between the sensor and reader coil. Therefore, the trade-off is between the quality factor, $Q$, and the coupling coefficient, $k$. Increasing one lowers the value of the other. Since the magnitude of impedance phase dip is proportional to square of the coupling coefficient, the coupling coefficient will be chosen as the parameter that needs to be maximized.

The resonant frequencies of all sensor implant combinations were calculated for different applied pressure values. The results are shown in Figure 4.19.

Finally, the Kapton sensor pressure sensitivity with respect to the resonant frequency of the implant was calculated for both “Via” and “Vialess” sensor implants. The results are shown in Table 4.1.
### Table 4.1: Sensor implant pressure sensitivity

<table>
<thead>
<tr>
<th>Implant Type</th>
<th>Coil Pitch, µm</th>
<th>Pressure Sensitivity, kHz/mmHg</th>
</tr>
</thead>
<tbody>
<tr>
<td>Kapton(4µm);Bondply(4µm);Via</td>
<td>60</td>
<td>-119.88</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>-88.25</td>
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<td></td>
<td>40</td>
<td>-69.12</td>
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<tr>
<td></td>
<td>30</td>
<td>-49.24</td>
</tr>
<tr>
<td>Kapton(4µm);Bondply(34µm);Vialess</td>
<td>60</td>
<td>-23</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>-15.8</td>
</tr>
<tr>
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</tr>
<tr>
<td></td>
<td>30</td>
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</tr>
<tr>
<td></td>
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<td>40</td>
<td>-16.39</td>
</tr>
<tr>
<td></td>
<td>30</td>
<td>-11.43</td>
</tr>
</tbody>
</table>

4.18.1: Quality factor of coils in air.

4.18.2: Quality factor of coils in Kapton.

Figure 4.18: Quality factors of coils
4.19.1: Thickness: Kapton (4\mu m), Bondply 34\mu m; Single coil with via.

4.19.2: Thickness: Kapton (4\mu m), Bondply 34\mu m; Vialess.

4.19.3: Thickness: Kapton (12\mu m), Bondply 34\mu m; Vialess.

4.19.4: Thickness: Multi-Kapton (4\mu m), Bondply 34\mu m; Vialess.

Figure 4.19: Sensor implant resonant frequencies
4.3 ParyleneC based pressure sensor

In the previous section a flexible Kapton based IOP sensor process flow was developed and the sensor implant behavior was modeled. The final thickness of Kapton Implant is between 76μm to 71μm depending on the thickness of the Bondply layer. The width of the sensor implant was 2.5mm. The width of the sensor cannot be reduced much further because of performance constraints on Bondply layer. Therefore, the bond regions need to be at least 400 μm on either side of the sensor diaphragm to prevent debonding of top and bottom polyimide layers. Ways to lower the width of the sensor implant were investigated as part of developing a flexible substrate based IOP sensor. Parylene C has a unique combination of electrical and physical properties and a very low permeability to moisture that makes it suitable for IOP sensor fabrication. It can be deposited in vapor forum and the uniqueness of ParyleneC is that structurally continuous films of tenths of microns to several mils can be formed with good surface conformity [13–15].

A parylene micromolding technique developed Noh et al. [16] was adapted to develop a flexible IOP sensor implant fabrication process flow. The fabrication process flow is shown in Figure 4.20.

Unlike Kapton based sensor, the Gold coils will be embedded in the parylene layers in this process. The process starts with micromould formation in silicon substrate by deep reactive ion etching(DRIE). A 5μm thick parylene layer is vapor deposited on the silicon substrate. Ti/Au film of 1-3 μm is deposited on the 1st parylene layer using E-beam evaporator. The deposited metal is patterned by liftoff process. Thicker layers of metal cause surface conformity problems during the liftoff process. Therefore the maximum thickness of the Gold layer is limited to 3μm. After the patterning of the Gold layer, a 2nd parylene layer of 5μm thickness is deposited using vapor deposition process. Then the 2nd parylene layer
Figure 4.20: Parylene based sensor fabrication process flow
is patterned. This patterning step is to decrease the thickness of the sensor diaphragm of the sensor implant.

As a part of the lower electrode and coil fabrication, a Teflon coated stainless steel sheet is deposited on the top with parylene. The sheet is coated with Teflon because of low adhesion of Teflon to parylene. A 2nd Ti/Au layer is then e-beam evaporated on the parylene layer. The metal is then patterned using liftoff process. A conformal parylene coating 5µm is then applied on top of the 2nd metal layer. Then the two separated substrates are aligned and uniformly compressed by applying a pressure close 24 MPa. The setup is then transferred to vacuum oven. The temperature of the oven is ramped up to 200 degreesC for an hour and maintained at that temperature for another hour. Then the temperature is slowly ramped down.

Bonding between the outer parylene layers of Teflon and silicon substrates due to interdiffusion of parylene layers at temperature close to its melting point (290 degrees C). Free standing parylene structure with embedded gold layers obtained by etching the silicon substrate using KOH solution and peeling off the Teflon coated stainless steel sheet.

An advantage of ParyleneC based process flow over Kapton process flow is that the bonding between the parylene layers is much more robust and parylene deposition happens at lower temperature resulting in stress free membranes.

The design process of ParyleneC based implant follows the same approach as Kapton based sensor design. An FEA model of 600µm radius ParyleneC based pressure sensor was developed. The deflection of the sensor over an applied pressure range of 10mmHg-50mmHg was obtained. A coupled electro-mechanical simulation was done to extract the capacitance of the sensor. The results are shown in Figure 4.21. As it can be seen from Figure 4.21, the capacitance of the sensor with 1µm embedded gold layer and 5µm thick parylene layers, varies linearly with applied pressure.
The design dimensions of ParyleneC based IOP implant are shown in Figure 4.22. The dimensions of implant design are similar to those of Kapton based implant design except the width of the bonding area. The space allocated to bonding area can be considerably reduced because there are no adhesive flow issues. A 100µm wide bonding area is sufficient to provide hermetic seal chamber between the sensor plates. The overall width of the sensor is 1.8mm which is 0.7 mm less than that of the Kapton based sensor.

The sensor coil inductance and resistance are calculated following the same approach as in Section 4.2.2. The results are shown in the Figure 4.23. As it can be seen from the figures, the inductance values of coils are lower than Kapton implant coils because of smaller coil size. The series resistance of the coils is much greater than Kapton implant coils. This is due to smaller trace thickness of gold layers. Because of these reasons, the quality factor of ParyleneC based implant is lower than that of the Kapton implant.
Figure 4.22: Dimensions of ParyleneC IOP implant design

Figure 4.23: Inductance and resistance of ParylenC based IOP implant coils

4.23.1: Gold coil inductance.

4.23.2: Series resistance of gold coils embedded in parylene.
REFERENCES


In implanted wireless sensor systems, the sensor coil and the reader coil are separated either by skin layer or muscle or tissue or a combination of all three. Due to the movement of the either the implant or the reader coil, the coils can be misaligned, thus reducing the coupling efficiency. In this chapter, the effects of misalignment and the coupling coefficient, $k$, on the detection of IOP sensor resonant frequency by the external reader circuitry are modeled.

The basic premise of misalignment analysis based design approach is to get upper and lower bounds on coupling coefficient, $k$. Coupling coefficient is a function of the geometry of the implant and reader coils, and also the relative position of the two coils with respect to each other. Including the upper and lower bound values of “$k$ ” helps the designer in designing a robust wireless sensor that takes into account coil misalignment. The maximum distance at which a reader coil can be placed from the sensor coil and still get an accurate sensor data is function of “$k$”. Therefore, from the known lower and upper bounds of $k$, one can accurately predict the maximum sensing distance at which the sensor operation is not severely compromised.
5.1 Misalignment Analysis of Primary and Sensor Coils

Zierhofer et al. [1] identified the issues concerning coupling coefficient variations in inductively powered systems. One of the conditions they listed is that the voltage amplitude at the secondary coil should be insensitive to variations of the relative position of the coupling coils. Induced voltage insensitivity at the secondary is important in applications where the receiver coil is used to derive an implant supply voltage which has to be kept within particular limits. This property is also important in IOP measurements. The reflected impedance on the primary coil circuit is a function coupling coefficient, which varies because of the frequent movement of the eye ball. The movement of the eye changes the relative position of the sensor coil with respect to the reader coil and hence changes the coupling coefficient.

Unlike other implantable wireless sensors such as “cochlear implants” [2], the chances that the primary and secondary coil are misaligned are very high in sensor implants in the eye. This is because of the rapid eye movement during the course of the day and the rotation of the eye ball towards the top of the head during the time of sleep of the patient. In order to account for the misalignment caused by the above mentioned reasons, the reader coil circuit should be designed in such a way so that it can compensate for the loss of coupling caused by coil misalignments.

5.1.1 Mutual Inductance

There have been several published works on misalignment analysis of circular loops [3–6]. Grover [3] derived integral equations for the calculation of mutual inductance between laterally and angularly misaligned circular loops. Soma et al. [6] presented a detailed theoretical analysis of misalignment effects in RF coil systems. Expressions for self inductances, mutual inductance, and coupling coefficient of two on-axis circular coils of radii, $a_1$ and $a_2$, where
Babic et al. [4] derived a formula for calculating the mutual inductance between inclined circular filaments arbitrarily positioned with respect to each other. The formula was derived using the method of vector potential, as opposed to Grover's approach, which was based on the Neumann's formula. Using Babic's method, one can calculate mutual inductance between both laterally and angularly misaligned coils provided one knows radii of primary \((R_p)\) and secondary coils \((R_s)\), parameters that define the plane of the secondary coil \((a, b, c)\), and the center of the secondary coil \((x_c, y_c, z_c)\).

According to the integral equation [4], mutual inductance is given by:

\[
M = \frac{\mu_0 R_s}{\pi} \int_0^{2\pi} \frac{[p_1 \cos \phi + p_2 \sin \phi + p_3] \Psi(k) d\phi}{k \sqrt{V_0^3}}
\]  

(5.5)
where:

\[
\alpha = \frac{R_s}{R_p}, \quad \beta = \frac{x_c}{R_p}, \quad \gamma = \frac{y_c}{R_p}, \quad \delta = \frac{z_c}{R_p},
\]

\[
l = \sqrt{a^2 + c^2}, \quad L = \sqrt{a^2 + b^2 + c^2},
\]

\[
p_1 = \pm \frac{\gamma c}{l}, \quad p_2 = \mp \frac{\beta l^2 + \gamma ab}{lL}, \quad p_3 = \frac{\alpha c}{L},
\]

\[
p_4 = \mp \frac{\beta ab - \gamma l^2 + \gamma bc}{lL}, \quad p_5 = \mp \frac{\beta c - \delta a}{l},
\]

\[
A_0 = 1 + \alpha^2 + \beta^2 + \gamma^2 + \delta^2 + 2\alpha(p_4 \cos \phi + p_5 \sin \phi),
\]

\[
V_0^2 = \alpha^2((1 - \frac{b^2c^2}{l^2L^2}) \cos^2 \phi + \frac{c^2}{l^2} \sin^2 \phi + \frac{abc}{l^2L} \sin 2\phi)
\]

\[
+ \beta^2 + \gamma^2 \mp 2\alpha \frac{\beta ab - \gamma l^2}{lL} \cos \phi \mp \frac{2ab \gamma c}{l} \sin \phi,
\]

\[
k = \sqrt{\frac{4V_0}{A_0 + 2V_0}}, \quad \Psi(k) = (1 - \frac{k^2}{2})K(k) - E(k)
\]

K(k) and E(k) are the complete elliptic functions.

Eq. (5.5) calculates the mutual inductance between a pair of single turn coils. To calculate the mutual inductance between a pair of flat spiral coils, it was assumed that the primary and secondary coils are composed of \(N_a\) and \(N_b\) circular coils, respectively (Figure 5.1) [6]. The connection between these circular coils is assumed to be in such a way that the direction of the current is equal in all the turns of a coil thereby each turn enhances the magnitude of the mutual inductance between primary and secondary coil. The other assumption is that the overall length of the connection paths is sufficiently shorter than the geometric dimensions of the coils, so that the influence of the current through the paths on the self and mutual inductances is negligible.
Then the mutual inductance between a pair of planar spiral coils can be written as:

\[
M_{total} = \sum_{i=1}^{N_a} \sum_{j=1}^{N_b} M(a_i, b_j, (a, b, c), (x_c, y_c, z_c))
\]  

(5.6)

Eq.(5.6) is used to calculate the mutual inductance between the reader and sensor coils.

The self inductances \( L_s \) and \( L_p \) of planar spiral coils can be approximated in different ways as mentioned in Section 4.1.

### 5.2 Coil Alignment

A perfectly aligned coil system is one in which the axes of two coils are collinear and the planes of the spiral coils are parallel to each other. In case of perfectly aligned coils, the
coupling coefficient can be approximated as \[ k(z) \approx \frac{r_s^2 r_p^2}{\sqrt{r_s r_p} \sqrt{z^2 + r_p^2}} \] (5.7)

Where, \( r_s, r_p \) are the equivalent radii of secondary and primary coils and \( z \) is the axial distance between the coil centers.

Zierhofer et al. [1] identified the issues concerning coupling coefficient variations in inductively powered systems and proposed a geometric approach for the coupling enhancement. They found that it is possible to exploit the area within the outer circumferences of primary and secondary coils to enhance the coupling coefficient. The coupling coefficient can be significantly improved, if the turns of the coils are not concentrated at the outer circumferences, but distributed across the radii. This approach is not possible on the secondary side of IOP system because region in the center of the implant needs to be hollow so that the implant does not block patient’s field of vision. But the primary side does not have those requirements to certain extent.

In general, secondary coil with a diameter smaller than that of the primary coil reduces the coupling at a particular distance between the coils compared to the case when the diameter of the secondary coil is equal that of the primary coil. But the loss in absolute value of the coupling coefficient is offset by increase in insensitivity of coupling coefficient to lateral displacement of the secondary coil remains.

In active implant systems, the receiver RF-amplitude is made insensitive to varying coil coupling by employing resonant circuits in both the transmitter and the receiver. The drive frequency is equal resonant frequency of the transmitter and receiver sides of the sensing system. This approach will not work in passive LC resonance based sensing since the resonant frequency of secondary coil side is the variable of interest and hence can not be
designed to be of constant value.

The three possible coil misalignment conditions are:

1. Lateral misalignment

2. Angular misalignment

3. General misalignment i.e., a combination of lateral and angular misalignments

Laterally misaligned coils are in parallel planes but their centers are separated by some distance (Figure 5.2.1). In the case of angularly misaligned coils, the planes of the coils are tilted to form an angle and the axis of one coil passes through the center of the other (Figure 5.2.2). Generally misaligned coils have a combination of lateral and angular misalignment (Figure 5.2.3).

In the case of IOP implant, the coils will be lateral misaligned when the eye glasses with external reader coil on them are offset by some distance in the lateral direction from the center of the secondary coil. Angular misalignment does not usually happen because none of the axes of the coils goes through the center of other coil when the coil planes are at an angle to each other. This is because of the location of the secondary coil on the near spherical surface of the eyeball.

In IOP sensing system, most frequently, the primary and secondary coils are generally misaligned. This is because of the rotation of the eye ball in the eye socket. To explain the coupling variations better, from here onwards in this chapter, angular misalignment is referred to be the misalignment caused when the primary coil is held stationary and the eye with the secondary coil in it rotates by an angle.
5.2.1: Laterally misaligned coils.

5.2.2: Angularly misaligned coils.

5.2.3: Generally misaligned coils.

Figure 5.2: Possible coil misalignments
5.3.1: Perfectly aligned coils.  
5.3.2: Laterally misaligned coils.  
5.3.3: Angularly misaligned coils.  

Figure 5.3: Possible coil misalignments in IOP sensing system
5.3 IOP Sensor Reader Coil Design

For coils of turns $N_p$ and $N_s$ the mutual inductance is scaled by $N_s N_p$ and the self-inductance of the coils is scaled by $N_s^2$ and $N_p^2$. Based on this fact, Soma et al [6] proposed a design procedure for coupling coefficient optimization for the case of RF coils. The procedure starts by determining the coil parameters of the implant and the worst-case misalignment of the implant. Then the coupling coefficient is calculated for single coil loops of sensor and reader circuits. The arithmetic mean radius of the coils is used as the radius of the coil single coil loop. The actual parameters of the coils are calculated using empirical shape factors applied to the single turn based coupling coefficient.

In this work, instead of obtaining the mutual inductance for single turn coils of effective radii, the actual mutual inductance of the coils was calculated directly using the numerical integration of the Eq. 5.5 and Eq. 5.6. Coupling coefficient between IOP implant coil and the external reader coil was calculated for different dimensions of primary coil at various coil orientations using Eq. 5.5 and Eq. 5.6. The derivation of equations that extract coil orientation parameters are shown in the Appendix.

As a part of designing the reader coil, the radius of the reader coil was made to be equal to that of the sensor coil and the coupling coefficient was calculated for perfectly aligned coils. The distance between the coils was varied from 5mm to 20mm. Both the sensor and reader coil pitch was set to be equal to $30\mu m$ and the sensor coil inner and outer radii were set to be equal to 5.5mm and 6mm. The radii values correspond to any Kapton based IOP implant with a pitch of $30\mu m$ and 7 turns. The outer radius of the primary coil was increased gradually and the corresponding coupling coefficients were obtained. The results are shown in Figure 5.4.

As the radius of the primary coil is increased, the coupling coefficient increases. The rate
Figure 5.4: Coupling coefficient of IOP sensor system as a function of reader coil outer diameter
of increase is higher at distances close to the secondary coil and the rate decrease as the primary is moved away from the secondary. Also the coupling coefficient curves get close to each other as the radius increases, this implies that the coefficient will not increase beyond certain value as the reader coil outer diameter keeps on increasing.

As a next part of the design, the outer radius of the primary coil was set to be equal to that of the implant coil and the inner radius of the primary coil was reduced to get the coupling coefficient for primary coils with progressively increasing distribution of coils towards the center. The results of the numerical simulation are shown in Figure 5.5

As it can be seen in Figure 5.5, the coupling coefficient does increase slightly with primary
coil with smaller inner diameter than that of the implant coil. But the rate of increase is lower than the previous case. This is evident from the closely grouped curves for the coupling coefficient for increasingly lower values of primary coil inner radius. Based on the observations from coupling coefficient analysis for different inner and outer radii of the primary coil, the optimum primary coil dimension were chosen to be equal to 9mm ID and 16mm OD for the case of sensor implant coil of ID 11mm and 12mm OD. Increasing the OD of primary coil more than 16mm for the selected sensor implant might increase the coupling coefficient but it will also increases the parasitic capacitance and resistive losses on the primary side.

The effect of coupling coefficient on the primary impedance phase dip was calculated for different values of coupling coefficient (Figure 5.6). For these simulations, the capacitance of the sensor was set to be equal to 3pF which corresponds to a 50mmHg intraocular pressure acting on a multi-diaphragm Kapton implant of radius 600µm with 34µm Bondply layer thickness. The phase dip was 0.2 degrees for k equal to 0.04. This corresponds to a distance of 1.3 cm between the sensor and implant coil, which is only 7mm less than that of the targeted distance of 2 cm. Still higher coupling can be obtained by using a sensor implant coil of 20µm pitch with 12 turns.

The variation of coupling coefficient to the lateral misalignment of the coils was obtained by increasing the lateral distance between the axes of the coils and the distance between the coil planes. The coupling coefficient decreases to a value of 0.02 when the lateral misalignment is equal to 10mm. This is unlikely to happen since bridge of the eye glasses will not let the glasses move that far off the center. A more a realistic value of lateral misalignment would be any where between 0 mm to 5 mm. In that case, the coupling coefficient decreases from 0.06 to 0.05 at a distance of 1.25 cm from the sensor implant.

Numerical accuracy of coupling coefficient simulation model results were tested by using
Figure 5.6: Effect of coupling coefficient on the primary impedance phase dip

Figure 5.7: Coupling coefficient variation with lateral misalignment
fabricated test coils of 10mm ID and 34 OD on FR4 substrate. The coils were both made of equal dimensions. The pitch of the coil turns was set to be equal to 2mm and the trace width and coil spacing were set to be equal to 1mm. A capacitor of 10pF capacitance was connected in parallel to one of the test coils. The test setup is shown in Figure 5.8. The phase angles and the amplitudes of the reader coil was measured at a distance of 10mm and 15mm. The results are shown in Figure 5.9. The numerical simulations of coupling coefficient accurately captures phase dip and magnitude of reader coils. The resonant frequency of the test coils and the model are offset from each other by 8 MHz. This is because the parasitic capacitance of the coils was into included in the coupling coefficient estimation and impedance calculation of the numerical model. Since, the phase dip of the reader coil is a function of the coupling coefficient and the phase dip magnitudes of both test coils and numerical model are close to each other, the accuracy of the model is verified.

Before modeling the coupling coefficients of coils that are angularly misaligned, the maximum angular misalignment that might actually occur in a human eye are measured by using the test setup shown in Figure 5.10.
5.9.1: Reader coil impedance phase angle.

5.9.2: Reader coil impedance amplitude.

Figure 5.9: Comparison of numerical and test results

Figure 5.10: Test setup to measure angle of view of left and right eyes
Different subjects were asked to stand against a line that was at a fixed distance from a white screen with a scale on the wall. After covering one eye of the subject with an eye patch, the subject was asked to align his other eye to a dotted line on the middle of the screen. The subject was then asked to follow an object on the screen without rotating his head until he felt that he could no longer view the object comfortably without rotating the head to get a clear view of the object. By noting the limiting distance on either side of the eye, the limiting angles were calculated for different test subjects. The results are shown in Figure 5.11, Figure 5.12 and Table 5.1.

The test results indicate that the angle is not symmetric for both the eyes. The limiting angle for the right eye is smaller on the left side than that of limiting angle on the right side. The
Table 5.1: Limiting angle of view

<table>
<thead>
<tr>
<th>Right Eye</th>
<th>Left Eye</th>
</tr>
</thead>
<tbody>
<tr>
<td>Left angle</td>
<td>Right angle</td>
</tr>
<tr>
<td>Mean</td>
<td>44.74</td>
</tr>
<tr>
<td>Std.</td>
<td>9.29</td>
</tr>
</tbody>
</table>

Figure 5.12: Limit angle of each eye.
Limiting angle for the left eye is smaller on the right side than that on the left side. Also, the limiting values varied for each subject. The test results gave an estimate of the actual angular misalignment that might occur during the daily activities of a patient.

Finally, as a part of coil design process, the angular misalignment effect on the coupling coefficient was modeled using the numerical model that estimates the coil mutual inductance. A surface plot of coupling coefficient as a function of eye angle and distance between the coils was obtained. Since the lateral misalignment is negligible in reality, it was not included in the numerical model. The results are shown in Figure 5.13.

Since, coupling coefficient is the main limiting parameter for increasing the range of the sensor, a multi-primary coil approach can be adopted to increase sensor performance insensitivity to the coupling coefficient. This approach is shown in Figure 5.14. This setup has
Figure 5.14: Multi-Coil based Sensing
three primary coils that are located at angle of 45 degrees to each other. A micro-controller based switching circuitry can be used to switch the three coils consecutively. The primary circuit current flows through only one among the three coils at any instant of time. The switching frequency can be set to below 1kHz since the intraocular pressure varies slowly over the course of the day.
REFERENCES


6.1 FEA modeling of Sensor Implant Insertion

Careful design of implant delivery device is necessary to avoid crimping of the sensor implant and plastic deformations during folding. Avoiding folding of the sensor during the entire process of insertion will be beneficial.

The insertion process of the sensor implant is designed to be similar to the insertion process of intraocular lens (IOL). The incision sizes that are allowed during implant delivery are assumed to be of the same size as the incisions made on the cornea during a cataract surgery. The incisions made on cornea for the purpose of acrylic intraocular lens insertion are generally of size 2.8mm to 3mm. Multiple incisions of the same size are made along the periphery of the corneal surface so that surgical devices can have access to the anterior chamber and intraocular lens.

A 3D model of the implant delivery tube is shown in Figure 6.1. A close up view of the tip of the device is shown in Figure 6.2
Figure 6.1: Implant delivery device

Figure 6.2: Close up view of device tip
The outer diameter the tube at the end is 3mm. A guide wire runs along the length of the delivery device. As shown in Figure 6.2, the hooked tip of the guide wire is used to latch on to the hollow inner portion of the implant. During the surgery, the surgeon can load the implant through the access hole provided on the top of the device. The delivery device diameter is made to be slightly more than the outside diameter of the implant. Once the implant is loaded, the surgeon can align the hooked tip of the guide wire to the inner surface of the hollow portion of the implant. Then the guide wire is slowly inserted into the rectangular shaped groove that runs along the inner surface of the device. The groove helps keep the guide wire stationary during the insertion procedure. The sensor implant can then be pulled slowly through the inner diameter of the tube at the tip of the device. After more than half the length of the implant enters the anterior chamber of the eye, the guide can be slowly retracted and the remaining portion of the implant can be pushed through the tube. This way there will not any crimping of the implant inside the smaller tube.

To identify the stress concentration regions during the insertion process, a finite element model of the sensor implant was built (Figure 6.3). ParyleneC was used as implant material and the drug delivery device is modeled as discrete rigid body. The strain rate dependent material data is used for material modeling of Parylene.

Contact between the implant and the delivery device is modeled by specifying frictionless behavior in the tangential direction and hard contact condition in the normal direction. Just like in cataract surgery, a viscous bio-compatible fluid can be applied on the inner surface of the delivery device so that the tangential friction can be reduced to a minimum value.

A displacement boundary condition was prescribed on the section of the implant that goes under the hooked tip of the guide wire. This boundary condition simulates the forces acting on the implant under the hooked region during the insertion procedure. The results are shown in Figure 6.4
As is seen in Figure 6.4 and Figure 6.5, most of the stress built up during the insertion process is relieved naturally after the implant exits the delivery tube. Plastic deformations do not occur in most regions of the implant except in the midsection of the implant that contacts itself in the delivery tube (Figure 6.5.2). Before the insertion, the capacitive sensor region of the implant can be aligned in such a way that it is in the low stress region of the implant. This can accomplished easily by rotating the implant to the desired orientation in the outer delivery tube section. Once the sensor is in place, the guide wire tube can be hooked for the device insertion.
6.4.1: Implant in the outer tube.

6.4.2: Implant moving towards inner tube.

6.4.3: Implant contact the entry surface of inner tube.

6.4.4: Implant folding naturally to move through the tube.

Figure 6.4: Stress history of the implant during the insertion process
6.5.1: Starting of stress buildup.

6.5.2: Maximum stress point.

6.5.3: Implant exiting the tube.

6.5.4: Implant unfolding by itself.

Figure 6.5: Stress history of the implant during the insertion process
As a part of developing a minimally invasive intraocular pressure sensor, sensor implant location has been identified and space availability for the implant has been mapped. A MEMS capacitive pressure sensor is designed, fabricated, and tested. Kapton and ParyleneC based biocompatible flexible substrate based IOP sensor implant fabrication processes are developed. Fabrication process flows are used as a basis to develop mechanical and electrical models of the sensor implants. Pressure sensitivity of capacitive pressure sensors and the overall sensitivity of various sensor implants of different sizes and scales are quantified. Reader coil that maximize the coupling coefficient for a given sensor coil size is designed by calculating the mutual inductance between the coils for different inner and outer diameters of reader coils. Limiting angle of view of eyes are estimated. Lateral, angular, and general misalignment issues are discussed and coupling coefficient is estimated for all the cases. The developed numerical model is verified using test results. Implant stress minimization techniques during insertion process are developed by performing a finite element analysis of Parylene implant insertion process.
7.1 Future Direction

In future, mask sets that follow the fabrication process flows of Kapton and Parylene sensors outlined in this work can be generated. Since the optimum dimensions of the sensor implants, and process parameters of fabrication process are already identified, sensors fabrication can be started without having to design the IOP system first.
CALCULATION OF COIL ORIENTATION PARAMETERS

A.1 Derivation

The following three variables are needed for the calculation of mutual inductance between the spiral coils using Eq. 5.5 and Eq. 5.6.

1. the primary and secondary coil radii, $R_p$ and $R_s$;

2. the parameters $a$, $b$, and $c$ defining the normal of the secondary coil plane;

3. the coordinates $(x_c, y_c, z_c)$ defining the center of the secondary coil that is on top of iris surface.

The coil setup is shown in Figure A.1

The center of the primary coil is at $(0, 0, 0)$. $(x_c, y_c, z_c)$ are the coordinates of the center of the secondary coil and $(x_e, y_e, z_e)$ are the coordinates of the center of the eyeball that sits on top of iris. Angle, $\theta$, is the counter-clockwise rotation about $y'$ axis and $\phi$ is the
counter-clockwise rotation about $z'$ axis. $R_e$ is the radius of the eye ball, $S$ is the distance along $z$ axis between the centers of primary and secondary coils.

The coordinates of $(x_c, y_c, z_c)$ are calculated using the following equations:

$$x_c = x_e + R_e \sin(\theta)\cos(\phi)$$  \hspace{1cm} (A.1)

$$y_c = y_e + R_e \sin(\theta)\sin(\phi)$$  \hspace{1cm} (A.2)

$$z_c = z_e + R_e \cos(\theta)$$  \hspace{1cm} (A.3)

Assuming the eyeball has spherical shape, the equation for the surface of the eyeball is

$$(x - x_e)^2 + (y - y_e)^2 + (z - z_e)^2 - R_e^2 = 0$$  \hspace{1cm} (A.4)

$$\Rightarrow x^2 + y^2 + z^2 - 2zz_e + z_e^2 - R_e^2 = 0$$  \hspace{1cm} (A.5)
The general equation of any quadratic surface in xyz rectangular coordinate system is

\[ Ax^2 + By^2 + Cz^2 + 2A'yz + 2B'zx + 2C'xy + 2A''x + 2B''y + 2C''z + D = 0 \] \hspace{1cm} (A.6)

Where \( A, B, C, A', B', C', A'', B'', C'', \text{and} \ D \) are constants. The equation of the tangent plane of the surface, with \((x_c, y_c, z_c)\) as the point of tangency is

\[ Ax_cx + Ay_cy + Az_cz + + A'(z_cy + y_cz) + B'(x_cz + z_cx) + C'(y_cx + x_cy) + A''(x + x_c) + B''(y + y_c) + C''(z + z_c) + D = 0 \] \hspace{1cm} (A.7)

Comparing Eq. A.4 with Eq. A.6, the non zero constant values are

\[ A = 1; \quad B = 1; \quad C = 1; \quad C'' = -z_c; \quad D = z_c^2 - R^2_e; \]

Substituting the constant values in Eq. A.7, we get the equation of spiral coil plane at \((x_c, y_c, z_c)\), which is

\[ xx_c + yy_c + zz_c - z_c(z + z_c) + z_c^2 - R^2_e = 0 \] \hspace{1cm} (A.8)

The unit normal \( \hat{n} \) to a plane specified by

\[ f(x, y, z) = ax + by + cz + d = 0 \] \hspace{1cm} (A.9)

is given by

\[ \hat{n} = \nabla f(x, y, z) = [abc]' \] \hspace{1cm} (A.10)

Comparing Eq. A.10 with Eq. A.8, we get

\[ a = x_c; \quad b = y_c; \quad c = z_c - z_e \] \hspace{1cm} (A.11)
B.1 Sprial Coil Parameter Extraction Code

```matlab
1 %********* File to Generate primary and secondary coils for an ... inductive link*************
2 clear all
3 close all
4 clc
5 % disp('Please enter the radius of the secondary coil in mm');
6 % ri = input('');%coil inner dia
7 ri=5.5; %fixing the inner radius of the coil to be 5.5mm
8 h=0.009; %thickness of copper in mm
9 disp('Please enter the space between the traces of secondary in ... mm');
```
s=input(''); %space between turns in coil  
disp('Please enter the width of trace in mm');  
w=input('');

%Determining number of possible turns  
nt = (0.5 - (s))/(s + w);  
ntr = int8(nt) ;  
if (ntr > nt)  
ntr = ntr-1;  
end

disp(' Due to restricted space available in the eye ');  
disp(' The total number of turns that can be accomodated in a ...  
      single layer is ');  
disp (ntr);

thetaMax=double(ntr)*2*pi;  
pitch=s+w;

disp('Please enter number of points per turn');  
N=input('');

theta=0:2*pi/N:thetaMax;

xs=(ri+(pitch.*theta./(2*pi))).*cos(theta);  
ys=(ri+(pitch.*theta./(2*pi))).*sin(theta);  
%plot(xs,ys)
fid = ...
    fopen('c:/Users/kckaturi/Desktop/CoilDesign/Kaptoncoil.inp', ...  
    'w' );
fprintf(fid,'**********Generating the primary and secondary ...
    coils made of copper*****************
');
fprintf(fid,'**space=%6.3f width=%6.3f,number of turns=%d ...  
    \n',s,w,ntr)
fprintf (fid,'.units mm\n');
fprintf( fid,'.default z=0 sigma=5.8e4 nhinc=1 nwinc=3\n');
for i = 1:length(theta)
    fprintf(fid , 'N%d x = %6.2f y = % 6.2f\n','i, xs(i) , ys(i));
end
for i = 1:length(theta)-1
    j =i+1;
    fprintf(fid,'E%g N%g N%g w=%6.2f h=%6.3f \n' , i,i, j, w, h);
end
fprintf(fid,'.external N1 N%g \n' , j);
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
fmin = input(' Enter the minimum frequency ');
fmax = input(' Enter the maximum frequency ');
ndec = input('Enter the number of frequencies to be analysed in ...  
    a decade ');
fprintf(fid,'.freq fmin=%d fmax=%d ndec=%d\n',fmin,fmax,ndec);
fprintf(fid,'.end');
fclose('all')
B.2 Mutual Inductance of Generally Misaligned Coils

```matlab
%% MUTUAL INDUCTANCE CALCULATION OF GENERALLY MISALIGNED COILS
%% BASIC INFO: Secondary coil dimensions: 11mm ID, 12mm OD
%% Minimum Separation Distance = 20mm
%% Minimum Primary coil dimension = 12mm OD

%% Algorithm
% 1. Generate the coils
% 2. Calculate the self-inductance of the coils
% 3. Vary the position of the secondary coil from 0 deg to 90 deg while
%    keeping the primary coil constant
% 4. Calculate the mutual inductance of the coils
% 5. Calculate the coupling coefficient: k
clc
clear all

% Physical constants, SI units
mu0 = 4 * pi * 1e-7; % permeability of vacuum, H/m

% COIL DESIGN PARAMETERS; Constants in SI units
Wp = 30e-6; % width of the primary conductor line
Cp = 30e-6; % clearance between the primary coil lines
Ws = 30e-6; % width of the secondary conductor line
Cs = 30e-6; % clearance between the secondary coil lines
sid = 11e-3; % secondary coil inner diameter
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sod=12e-3; %secondary coil outer diameter
pid=11e-3; %primary coil inner diameter
pod=12e-3; %primary coil outer diameter

S=0.005:0.001:0.02; %Distance between the center of primary coil ... and the closest point
    %on the spherical surface of the eye
    %thee=linspace(pi,pi/2,20);
thee=pi;
coupf=zeros(length(S),length(thee),length(pod));
for radii=1:length(pid)
    %Radii of the concentric loops of primary and secondary coils
    Rp=pod/2:-(Wp+Cp):pid(radii)/2;
    Rs=sod/2:-((Ws+Cs):sid/2);
for dis=1:length(S)
    %Coordinates for the center of the eyeball: [xe ye ze]
    Re=0.01225; %Eyeball radius
    xe=0.0;
ye=0;
ze=Re+S(dis);
    %FOR LOOPS THAT ARE USED TO CALCULATE L, M, k, etc.
    %1. (xc,yc,zc) are the coordinates of the center of the ... secondary coil
    %that sits on top of iris
    %2. theta is the c.c. rotation about y-axis of the xyz cs that ... is at the
    %center of the eye ball; theta = 0 on +ve z-axis;
%3. phee is the c.c. rotation about z-axis at the center of the eye
% phee can be varied to include total coil misalignement affects.
%4. Varying theta only moves the coil along the perimeter of a ...
circle on
%the sphere

for k=1:length(thee)
    theta=thee(k);
    phee=0;
    xc=xe+Re*sin(theta)*cos(phee);
    yc=ye+Re*sin(theta)*sin(phee);
    zc=ze+Re*cos(theta);

%Equation of normal to the secondary coil plane in the outward ...
direction
%from the spherical surface
    a=xc-xe;
    b=yc-ye;
    c=zc-ze;

%SELF INDUCTANCE OF PRIMARY COIL
    Lp1=0;
    for i=1:length(Rp)%self inductance of singular circular loop
        Lp1=Lp1+mu0*Rp(i)*(log(8*Rp(i)/Wp)-2);
    end
    Mp=0;
    for i=1:length(Rp)
        for j=1:length(Rp)
if i==j
    Mpt(i,j)=0;
else
    Mpt(i,j)=MCoil(Rp(i),Rp(j),[0 0 0],[0 0 1]);
end
Mp=Mp+Mpt(i,j);
end
end
Lp(k)=Lp1+Mp;

%SELF INDUCTANCE OF SECONDARY COIL
Ls1=0;
for i=1:length(Rs)%self inductance of singular circular loop
    Ls1=Ls1+mu0*Rs(i)*(log(8*Rs(i)/(0.5*Ws))-2);
end
Ms=0;
for i=1:length(Rs)
    for j=1:length(Rs)
        if i==j
            Mst(i,j)=0;
        else
            Mst(i,j)=MCoil(Rs(i),Rs(j),[0 0 0],[0 0 1]);
        end
        Ms=Ms+Mst(i,j);
    end
end
Ls(k)=Ls1+Ms;
%%% MUTUAL INDUCTANCE BETWEEN THE PRIMARY AND SECONDARY COILS %%

Mt=0;
for i=1:length(Rp)
    for j=1:length(Rs)
        Mtotal(i,j)=MCoil(Rp(i),Rs(j),[xc yc zc],[a b c]);
        Mt=Mt+Mtotal(i,j);
    end
end
M(k)=Mt;
cc(k)=M(k)./sqrt(Lp(k).*Ls(k));
end %END OF THE FOR LOOP USED FOR CALCULATION OF M, L, k, etc.
coupi(dis,:)=cc(k);
end %END of distance for loop
coupf(:,:,radii)=coupi;
end %END of radii for loop