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DESIGN AND ANALYSIS OF NONINVASIVE INTRAOCULAR PRESSURE SENSOR

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ABSTRACT

This paper presents a design of noninvasive capacitive pressure sensor using micro electromechanical system (MEMS) for the measurement of Intraocular pressure (IOP). The normal IOP of a human eye ranges from 12mmHg to 20mmHg. IOP is the important factor in the diagnosis of glaucoma. The corneal curvature of eye changes with the change in IOP. Thus the IOP can be measured by measuring the variation in the corneal curvature. Since it is going to be used for human eye, it has to measure the pressure of range 5-40 mmHg. The sensor consists of a poly silicon layer which gets deflected when pressure is applied. This simulation is done in COMSOL software and it gives good sensitivity and linearity in the IOP range. Here, the changes in the sensitivity according to the number of perforations has been shown. The transfer of sensed information is often achieved through a wireless interface established between transmission/reception antennas of the involved parts.

Keywords: intraoccular pressure sensor, MEMS, capacitive pressure sensor, glaucoma, monitoring device, wireless transmission.

1. INTRODUCTION

Intraocular pressure (IOP) is the important risk factor for glaucoma and also the primary factor in the diagnosis and treatment of glaucoma [1, 2]. Glaucoma is the second leading cause of blindness; it also damages the optic nerve and loss of visual field with increased intraocular pressure (IOP). In some of the studies it is shown that IOP peaks fluctuates, so it has to be measured frequently and IOP measurement taken over 24 hours are accurate factors for vision loss in glaucoma patients [3,4]. Since single time measurement IOP measurement is often lower than the peak value of IOP, glaucoma patients suffering from vision loss exhibiting below threshold IOP, so the patient is unidentified with glaucoma.

Glaucoma is a neuro degenerative state that is leading cause of irreversible blindness. An aqueous humor is secreted by ciliary body in the anterior chamber. In the normal eye, this aqueous humor circulates in the eye and drains through the trabecular mesh work. The rate of secretion and discharge should be in balanced condition. Any obstruction in this mesh.

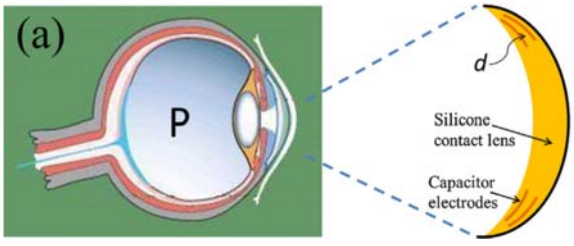

Figure-1. Contact lens sensor on eye for normal range of IOP.

work will result in increase of Intraocular pressure (IOP). The outflow of aqueous is lower than the inflow because of the resistance due to obstruction in the drainage pathway. The elevated IOP will compress the optic nerve tissue and the blood vessels and in turn leads to loss of vision if not treated. There won't be any symptoms at this point of time and also cannot be detected without a pressure measuring device. Therefore, it is essential to have an accurate, sensitive, continuous, comfortable IOP measuring device.

In this work, a capacitive membrane with an inductor has been designed. As it is going to be used in human eye, the outer diameter of the sensor was taken as 14 mm, and operates in the range between 10 and 40 mmHg. A change in the curvature of the soft cornea displaces the capacitive membrane. When the IOP is increased, shape of the cornea in the eye changes. As there is change in the shape of cornea the capacitive sensor which is going to be placed on the cornea displaces. For the purpose of wireless signal transmission, the capacitor is coupled with the inductive coil. This forms a LC circuit to facilitate wireless signal transmission, the capacitor (C) is coupled with an inductive coil (L) with fixed inductance to form a LC resonant circuit. The resonant frequency produced by LC circuit is then transmitted [5]. So as an initiative, in this paper the capacitive membrane made up of poly silicon material is given pressure ranging from (10-40) mmHg with different number of perforations in it. According to the number of perforations the sensitivity of the membrane varies.

When the capacitor's parallel electrodes are embedded into a contact lens, changes in the corneal curvature can lead to changes in the capacitive gap spacing and change the capacitance, which results into change of the resonance frequency (Figure-1).

2. REVIEW OF EXISTING SOLUTIONS

A brief overview of the existing methods for measuring intraocular pressure are as Goldmann Applanation Tonometry This method was introduced in 1950's has been the 'gold standard' for measuring IOP.

Portable Hand-held tonometers [6] are costly and require a skilled operator. Unfortunately there is no direct measure of rigidity of the cornea in live patients and thus IOP cannot be accurately determined with existing techniques. Studies directed to tonometry strongly recommend not to compare results from patient to patient, but rather use it as a relative determinant of IOP in the same patient, as a precaution [7].

3. INTRAOCULAR IMPLANTS FOR PRESSURE SENSING

As continuous monitoring of intraocular pressure has now become important, several groups had then started working towards development of an implantable device that transmits this information continuously or whenever required to an external portable device. The earliest implantable intraocular pressure sensor was developed by Collins [8]. He placed a small passive resonant transensor inside the eye that absorbed energy from an external oscillator at different frequencies depending on the IOP. A more recent fully developed system that is embedded in an intraocular lens was described by Mokwa et al [9]. This device is designed to be implanted following removal of the native lens of the eye. Other techniques involve drilling the cornea to place a pressure sensor into the anterior chamber. Several research works in implantable pressure transducers have been reported [Michigan, Minnesota [10]. The state of the art is possibly best represented by the work of Stangel et.al [4]. They describe an IOP sensor and transponder where the silicon integrated sensor and electronics is implemented in a 2.6 x 2.6 mm die. They also describe a foldable antennae concept but do not propose how it might be built. The long term goal of this project is also to design a relief valve that actuates based on the sensed pressure. Thus the sensor and the actuator together would represent a complete solution for glaucoma treatment. Several other sensors types have been developing including PZT and other capacitor sensors. However, none is available for actual use due to problems of irreversibility (too intrusive). A contact lens based sensor system with a wire coming out of the eye for data transmittance, has been reported in animal models [11]. Having a wire leading out of the eyelids limits its use due to issues of comfort, frequency with which patient has to place and remove the lens and ability of patient to insert the lens. The method is still under refinement and is being developed with wireless telemetry. At present, there is no fully developed intraocular pressure sensor for continuous monitoring.

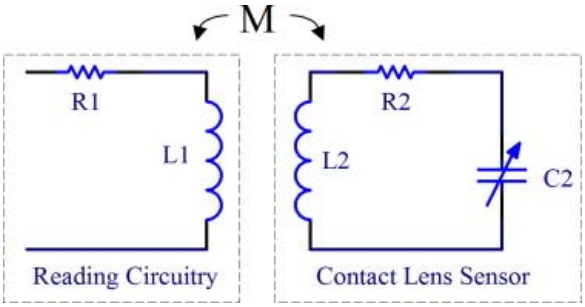

Figure-2. Reading circuitry schematic of the contact lens sensor

As indicated above, the the drawbacks that plague the existing devices are Irreversibility, Intrusiveness and invasiveness, Lack of a mechanism to monitor pressure, High power, comfortable in wearing. Easy usage etc.

4. SENSOR DESIGN

A. Sensing and electrical design overview

The design of sensor is based on micro electro mechanical system in which capacitor attached to an inductor. Piezo-resistive strain gauge and capacitive pressure sensors have widely been used for pressure sensing in many biomedical applications [12]. However, capacitive pressure sensors are more appropriate for low pressure application due to their high sensitivity to pressure changes and low power consumption [13]. When a capacitor is coupled with an inductive coil, it forms an inductor–capacitor (LC) resonance circuit; the resonance frequency is given as,

$$f = \frac{1}{2\pi\sqrt{L_2 C_2}} \quad (1)$$

where L2 and C2 are the inductance and capacitance of the resonance circuit, respectively. The consequent reading circuitry and electrical circuit of the contact lens sensor are shown in Figure-2. Here, when the capacitance changes, the resonant frequency in the secondary circuit changes [14].

B. Sensor physical design

The sensing element of the contact lens sensor is a variable gap sensing capacitor that can sense changes in curvature. An electrode with a soft gap is fabricated in a soft silicone rubber sensing layer on the corneal side of the lens, while a reference electrode and an inductive coil are fabricated in a hard silicon rubber layer on the air side of the lens (Figure-3). The sensing capacitor is electrical coupled with the inductive coil with fixed geometry.

The sensor is designed to be worn on humans. For a typical person, the curvature change for typical IOP variation between 5 and 40 mmHg is 0.12 mm for a typical cornea (radius of curvature 8 mm). To maximize linearity, the difference in

Table-1. Contact lens sensor design parameters.

Contact lens diameter (mm)	14
Radius of curvature of sensing layer (mm)	6.5
Inner radius of curvature (mm)	4.9
Thickness of each contact lens layer (Î¼m)	100
Dynamic range (mmHg)	35

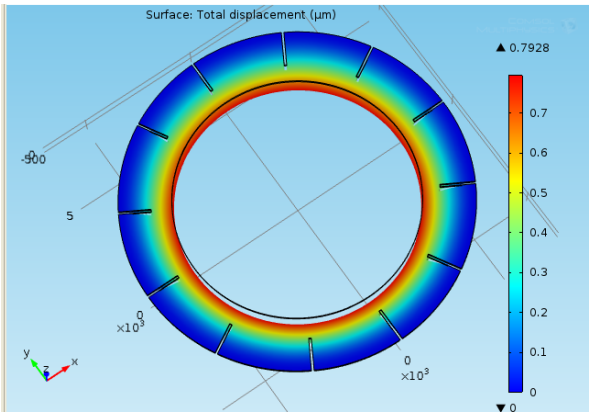

Figure-3. Simulated results.

radius between the sensing layer and the reference layer should be maximized, but an overly large difference would reduce the IOP sensitivity. A linear, but reasonably comfortable range of 0.5 mm curvature change was implemented in for the capacitor. In addition to the capacitor, a circular spiral multi-turn inductive coil designed to have a high Q factor to maximize the reading resolution of the LC resonance circuit. The design parameters of the contact lens sensor are detailed in Table-1.

5. SIMULATION

This simulation is carried out in COMSOL. The work done here is on a movable capacitive circular plate which has perforations in it. It has been studied that as the perforations in the membrane increases, the sensitivity is increased. So for the accurate measurement of IOP a circular plate with greater perforations can be used as shown in Figure-3.

Table-2. Pressure vs Displacement with 6 perforations.

Pressure in Pascal	Displacement in µm
1500	0.2138
2000	0.285
3000	0.4275
4000	0.57
5000	0.7125

Table-3. Pressure vs Displacement with 12 perforations.

Pressure in Pascal	Displacement in µm
1500	0.2162
2000	0.2883
3000	0.4325
4000	0.5766
5000	0.7208

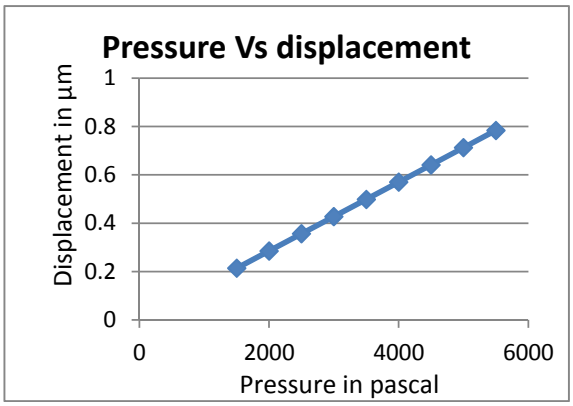

Figure-4. Pressure Displacement relationships.

6. RESULTS

A circular electrode membrane has been designed with different number of perforations. As the number of perforation increases, the sensitivity of the material increases which is shown in the Tables 3 and 4

The graph for pressure and displacement for the membrane having 6 perforations and 12 perforations respectively is shown below. When the displacement changes the capacitance will also change according to the displacement

7. BIOMEDICAL DESIGN CONSTRAINTS

Several sensor designs were developed and characterized to advance flexible passive pressure sensors in biomedical applications. However, first it is necessary to define the design space and constraints associated with operation in such an environment. The list below describes the research objectives for designing pressure sensors for biomedical applications

Use parallel-plate capacitors fabricated on and integrated into pressure-deflectable plates or membranes. Leverage standard microelectronics flex-circuit packaging technology used for ICs to achieve batch-fabrication.

Fabricate devices from biocompatible flexible materials.

Leverage multi-layer fabrication approaches to achieve embedded passives.

Develop designs without via interconnects between planar layers to increase robustness as a flexible device.

Use materials with the potential for micro-fabrication processes to create micro-scale mechanical and electrical features.

Demonstrate applicability in biomedical applications.

8. CONCLUSIONS

The measurement of IOP is most important; if the IOP increases beyond 20 mmHg then it leads to glaucoma. So the IOPO has to be monitored continuously. In this part of work on IOP sensor design, one layer of the capacitive sensor is taken and has been designed. Here the important factor is that IOP sensor must have high sensitivity. So here the sensitivity has been increased by increasing the number of perforations. The future work could be to extend the length of the transmission distance between the display device and the sensor.

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