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Soft wearable contact lens sensor for continuous intraocular pressure monitoring

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ABSTRACT

Intraocular pressure (IOP) is a primary indicator of glaucoma, but measurements from a single visit to the clinic miss the peak IOP that may occur at night during sleep. A soft chipless contact lens sensor that allows the IOP to be monitored throughout the day and at night is developed in this study. A resonance circuit composed of a thin film capacitor coupled with a sensing coil that can sense corneal curvature deformation is designed, fabricated and embedded into a soft contact lens. The resonance frequency of the sensor is designed to vary with the lens curvature as it changes with the IOP. The frequency responses and the ability of the sensor to track IOP cycles were tested using a silicone rubber model eye. The results showed that the sensor has excellent linearity with a frequency response of ~8 kHz/mmHg, and the sensor can accurately track fluctuating IOP. These results showed that the chipless contact lens sensor can potentially be used to monitor IOP to improve diagnosis accuracy and treatment of glaucoma.

1. Introduction

Intraocular pressure (IOP) generally measured during office hours is used as a primary indicator of glaucoma [1]. However, peak IOP occurs at night during sleep [2], and IOP measured during a single visit during office hours would miss the peak. Hughes et al. [3] examined the use of 24-h IOP profile in glaucoma treatment. They found that the use of 24-h IOP profile is useful in facilitating early detection of glaucoma which changed glaucoma treatment in 79% of the cases.

IOP profile can be measured using telemetric pressure sensors [4–7] implanted invasively into the eye. Alternately, periodic IOPs can be measured in a hospital sleep lab, where the patient is waked periodically for IOP measurement using standard tonometry. In addition to applanation, IOP can also be determined non-invasively by measuring the corneal curvature using Contact Lens Tonometry (CLT) [8–10]. In these approaches, the local strain change in the contact lens with IOP is sensed by the piezoresistive sensing elements embedded in the sensor. However, the local deformation on the cornea is sensed by the rigid metallic sensing element only after it is mechanically attenuated by the soft lens. After the attenuation, the resultant strain in the embedded sensing element is inherently small. A silicon chip is typically integrated as part of the sensor to amplify and transmit the signal wirelessly. With an encapsulated chip, sensors with embedded chip can be as thick as 583 μm [11], while commercial contact lenses for vision correction can be as thin as 50–100 μm. The added thickness of the chip based sensor is needed for proper encapsulation of the chip, but the added thickness also would increase the mechanical attenuation and degrade the strain sensitivity of the sensor. Alternate designs using flexible non-metallic conductive film elements to increase the sensitivity has been proposed [12], but the thickness remains relatively thick since silicon chip is still required.

To eliminate the silicon chip from the sensor, position-sensitive dyed glycerol-filled micro-chamber was proposed as sensing elements [13]. Instead of sensing local strain changes, the dye is designed to move with the global changes in curvature. The corneal curvature can be determined by tracking the dye movement using a camera. The camera requires a clear line of sight to the eye, which makes it unsuited for use with closed eye during sleep when the peak IOP occurs. Alternately, the change in corneal curvature as a function of IOP can be sensed by an inductive coil (Fig. 1). Changes in the corneal curvature change the inductance of the embedded coils. By coupling the inductive coil (L) with a capacitor (C) to form a resonator, changes in the coil inductance can be detected wirelessly by an external reader as changes in the LC resonance [14]. By establishing the correlation between LC resonance and IOP, the changes in IOP can be read and monitored dynamically by a wireless reader mounted near the eye.

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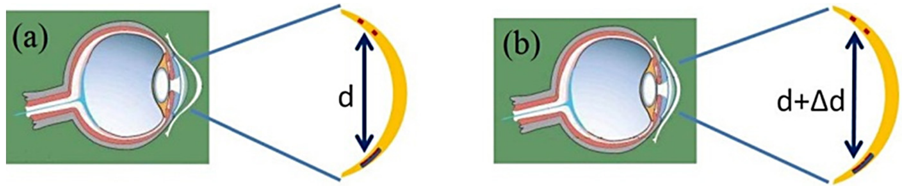

Fig. 1. Sensing mechanism of the contact lens sensor for continuous IOP monitoring. (a) Contact lens sensor configuration on an eye at low IOP. (b) Contact lens sensor configuration on an eye at high IOP.

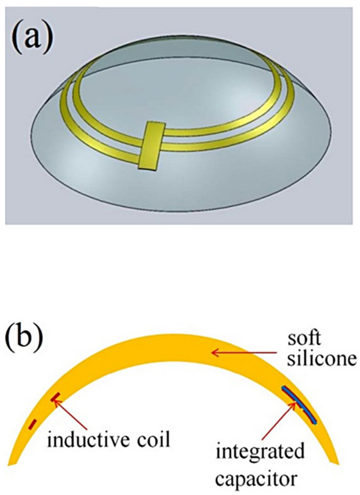

Fig. 2. Structure of the contact lens sensor with an embedded LC resonance circuit. (a) 3-D view of the contact lens sensor. (b) Cross-section view of the contact lens sensor.

In this study, the feasibility of this IOP sensing concept is explored. A soft contact lens sensor with embedded LC resonator is designed and fabricated. The contact lens sensor is tested on silicone rubber model eye and its performance in the human IOP range is examined.

2. Design and fabrication

2.1. Design of the contact lens sensor

Elevated IOP will change the corneal curvature [15–17], and the curvature of the soft contact lens on top. A soft deformable LC resonance circuit (Fig. 2) embedded inside the lens was designed to track the change of the corneal curvature with IOP (Fig. 3). The inductance L_S can be calculated using a simple planar spiral coil model with coil spacing s and average coil diameter d_{avg} and $s << d_{avg}$ [18]. The inductance L_S is given as,

$$L_S = \frac{1}{2}\mu n^2 d_{avg}\left[\ln\left(\frac{2.46}{\rho}\right) + 0.2\times\rho^2\right], \tag{1}$$

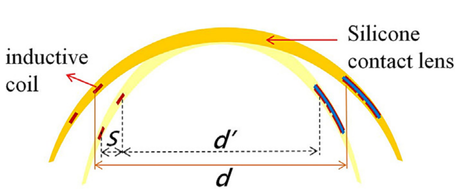

Fig. 3. Schematic of the change in inductive coil diameter with contact lens curvature change. The change is exaggerated for clarity, while the typical contact lens curvature is ~8 mm.

where

$$d_{avg} = \frac{d_{in} + d_{out}}{2}, \tag{2}$$

$$\rho = \frac{d_{out} - d_{in}}{d_{in} + d_{out}} \tag{3}$$

is the fill ratio of the coil, d_{out} is the outer coil diameter, d_{in} is the inner coil diameter, μ is the magnetic permeability and n is the number of turns of the coil. The change of the coil diameter Δd_{avg} is,

$$\frac{\Delta d_{avg}}{d_{avg}} = \frac{(r_o + \Delta r)\cdot\sin(\alpha\cdot r_o/(2\cdot(r_o+\Delta r)))}{r_o\cdot\sin(\alpha/2)} - 1, \tag{4}$$

where r_o is the initial radius of curvature of the contact lens, and α is the opening angle of the circumferential inductive coil, Δr is the change of the radius of curvature which changes with IOP. For $\Delta r \ll r_o$, the expression can be simplified to,

$$\frac{\Delta d_{avg}}{d_{avg}} = a\cdot\Delta r, \tag{5}$$

where

$$a = \frac{1}{r_o} - \frac{\alpha}{2r_o}\cot\left(\frac{\alpha}{2}\right). \tag{6}$$

In earlier studies [16], investigators reported that an IOP change of 1 mmHg generates a change of ~3 μm change in corneal radius (for an eye with its corneal radius of curvature as 7.8 mm) such that the change in IOP can be modeled as,

$$\Delta\text{IOP} = \frac{\Delta r}{c}, \tag{7}$$

where c is dependent on the biomechanical properties of the eye. From Eqs. (5) and (7), it can be determined that the change in coil diameter due to human IOP change is typically less than 0.15 mm, and is small. The change in L_S as a function of d_{avg} is shown in Fig. 4. The plot showed that the dependence between 11 mm and

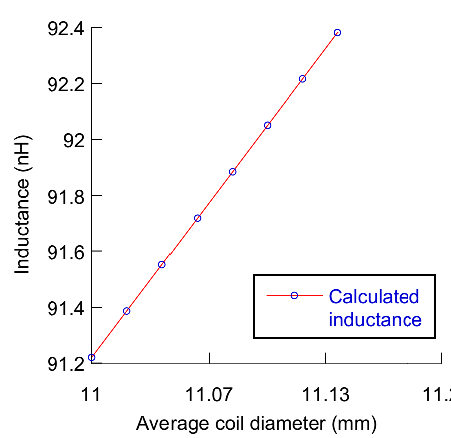

Fig. 4. The inductance of the coil as a linear function of the average coil diameter.