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Continuous Intra Ocular Pressure Measurement Sensor for Glaucoma Diagnostic

Gaëlle Lissorgues¹, Lionel Rousseau¹, Patrick Poulichet¹, Laurie Valbin¹, Serge Picaud², Laurent Chicaud², Serge Bernard⁴, Philippe Bergonzo³, Francois Dedieuleveult³, and Philippe Auvray⁵

¹ ESYCOM, ESIEE, Noisy le Grand, France

² Institut de la Vision, INSERM, Paris, France

³ LIST, CEA, Saclay, France

⁴ LIRMM, Université II Montpellier, Montpellier, France

⁵ OPHTIMALIA, SME, Colombelles, France

Abstract— Glaucoma is an ocular pathology usually associated with an increase in Intra Ocular Pressure (IOP). In this study, we are developing disposable eye lenses including a specific micro fabricated pressure sensor to measure IOP all day long. The information data will be wireless transferred via magnetic coupling to an external receiver. Our first work deals with the sensor design and fabrication. Simulation results based on classical electronic circuit tools will be presented and lead to several sensor solutions working at different RF frequencies. The fabrication process of the first sensors will also be described and an early IOP characterisation set-up will be presented to try to later quantify the sensor sensitivity, with IOP variations measured in the range 20 to 70mmHg.

Keywords— Biomedical engineering systems, microfabricated sensors, IOP measurement, Glaucoma diagnostic.

I. INTRODUCTION

Glaucoma being an ocular pathology, which leads to the second cause of blindness in people over the age of 50, the aging of the world population will lead to further increasing the number of patients greatly visually impaired by this disease. In most cases, glaucoma is associated with an increase in Intra Ocular Pressure (IOP) [1]. Different hypotheses have been put forward to explain the degeneration of ganglion cells, which often occurs after an increase in intraocular pressure. One of these hypotheses is that the compression of the optic nerve and associated blood vessels could result in retinal ischemia with excitotoxicity of the retinal tissue and current treatments are essentially based on diminishing intraocular pressure [2].

In this study funded within the project ANR TecSan MATEO, we are developing disposable eye lenses including a specific pressure sensor to measure IOP all day long, and the instrumented lens will communicate by radio frequency to an electronic chip located on a glass branch. The daily information would thus be available for ophthalmologists to improve diagnostics. Indeed, ophthalmologists can only

measure the intra-ocular pressure punctually, when the patient is visiting the ophthalmologist's practice, most often using the Goldmann applanation tonometer method. Moreover, the intraocular pressure can only be well controlled at the time of the visit due to proper medication but remains partly uncontrolled during the remaining part of the day. Depending on the advancement of glaucomatous damage to the optic nerve, closer monitoring seems to be necessary. Detailed knowledge of the behavior of IOP in patients with glaucoma is of paramount importance, as increased IOP and wide diurnal IOP variations are considered major risk factors for the progression of glaucoma [3, 4].

This work will offer a way to monitor continuously the IOP, on the typical basis of one measurement every 15 minutes, sufficient to monitor daytime variations, or every 5 minutes on diagnostic mode to try to associate the relationship with the physical environment of the patient (posture, activity, environment etc).

II. GENERAL DEVICE DESCRIPTION

A. Operation Principle

The final device will be a disposable lens containing a pressure or deformation sensor which transfers the information via magnetic coupling to the reusable receiver fixed on the glasses. Then data are stored in a portable device and transferred to a computer for further processing and analysis, see Fig. 1. Moreover this lens will be disposable to avoid any sterilization problems between patients or during long-term use. Therefore, our aim is also to limit electronic content on the lens to minimize its individual cost and complexity level.

The key element of this innovative measurement method is a soft contact lens with an embedded microfabricated sensor that allows the measurement of changes in cornea curvature correlated to IOP, as previously studied by [5, 6].

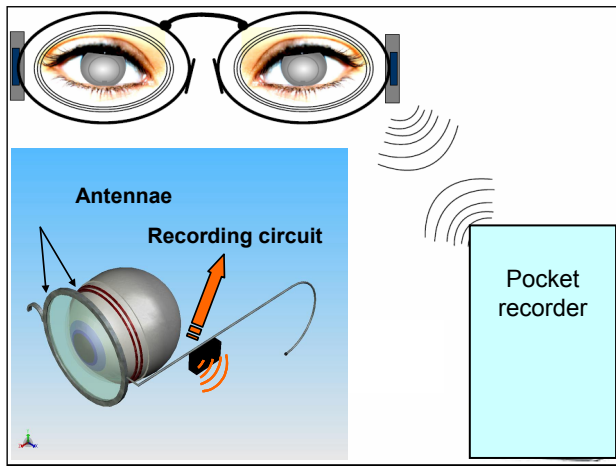


Fig. 1 Conceptual view of the system

Because of the short distance between the eyes and a pair of glasses, typically less than 3cm, the RFID protocol to measure the deformation of the lens can be used or any contact less communication protocol. As well as offering comfort to the patient, a wireless link protocol offers also the opportunity not to require any battery on the lens. The system will consist of an implanted sensor (corresponding to the RFID tag) and an external reader on the glasses, see Fig. 2. The system is therefore based on inductive coupling between the coils L1 and L2 respectively representing the reader and the embedded sensor.

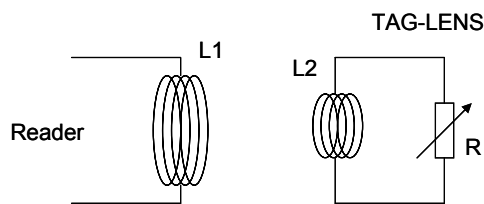


Fig. 2 Principle of the wireless communication (sensor in the lens and reader on glasses)

B. Electrical Simulations

The operation principle relies on resonant circuit modifications which can be simulated using classical electronic circuit tools applied on a transformer structure. The schematic of fig.2 can be modified in Fig.3 to extract the equivalent impedance Z1 in which appear 3 parameters: the coupling coefficient k, the quality factor Qs, and the resonance frequency $\omega_s = 2\pi f_s$ [7].

$$Z1 = j.L_r.\omega. \left[1 - k^2 \frac{\left(j \frac{\omega}{\omega_s} \right)^2}{1 + j \frac{1}{Q_s} \frac{\omega}{\omega_s} + \left(j \frac{\omega}{\omega_s} \right)^2} \right] \text{ with}$$

$$k = \frac{M}{\sqrt{L_r.L_s}} \quad Q_s = \frac{1}{R_s} \sqrt{\frac{L_s}{C_s}} \quad \omega_s = \frac{1}{\sqrt{L_s.C_s}}$$

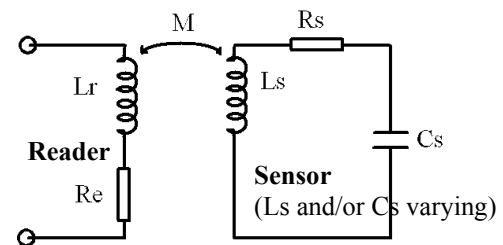


Fig. 3 Equivalent circuit of the sensor in the lens and reader on glasses based on magnetic coupling

The relative frequency variation is directly related to the LC component variations, as expressed below.

$$\frac{\Delta f}{f} = -\frac{1}{2} \frac{\Delta L}{L} - \frac{1}{2} \frac{\Delta C}{C}$$

Then any component variation induced by a mechanical deformation at the cornea will be converted into a frequency shift, either detectable while measuring the impedance magnitude or phase, Fig. 4.

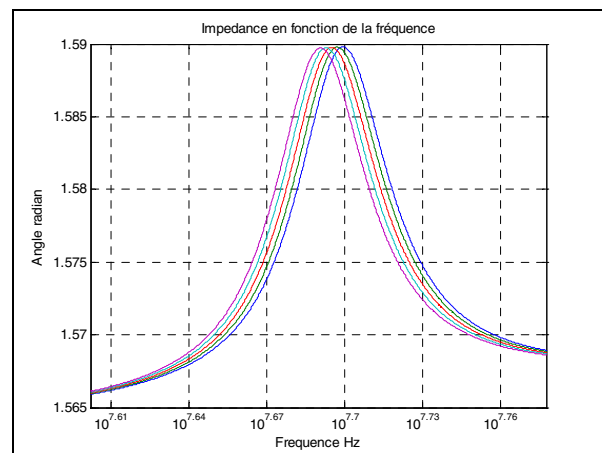


Fig. 4 Induced phase shift of Z1 versus the frequency, for 1 to 5% inductance variations – example at 50MHz

To optimize the global sensor operation, the coupling coefficient k must be as high as possible, meaning that the

distance between the instrumented lens and the glasses should be reduced. The quality factor must also be enhanced at the lens level, meaning high inductance values with low resistive loss and small capacitance values, as for example at 868MHz, $L = 40\text{nH}$, $R < 10\Omega$, with $C = 0.8\text{pF}$. Based on this transformer structure, the wireless link can use RFID standards: at 125kHz with maximal range of 10cm using a piezoelectric sensor, at 13.56MHz / 27MHz with maximal range of 1m using a first kind of LC based sensor or at 868MHz with maximal range of 2m using a second kind of LC based sensor.

III. DEVICE FABRICATION

To perform the monitoring IOP, we proposed to develop a disposable lens that follows the deformation of the cornea caused by the modification of IOP. We need a technology that offers the possibility to measure a small deformation of the cornea (typically about $3\mu\text{m}$) through a transparent and disposable lens. Traditional MEMS technology proposes several solutions for pressure or deformation sensing, either based on piezoresistive or piezoelectric gauges, or on capacitance variations. As previously explained, a derivative solution has been chosen based on inductance and capacitance variations using a resonating circuit. And the sensor will be coupled to an integrated antenna required for the magnetic coupling within a reduced surface corresponding of the iris diameter (typ. 1cm).

Concerning the fabrication, as it is required to adapt the technology onto an optically transparent soft material, i.e. based on a polymer lens, not to affect the vision of the patient, a low temperature process and low stress technology has been defined and tested, see Fig.5. Several prototypes are currently being fabricated, with examples shown on fig.6, and fig.7 for an embedded LC sensor.

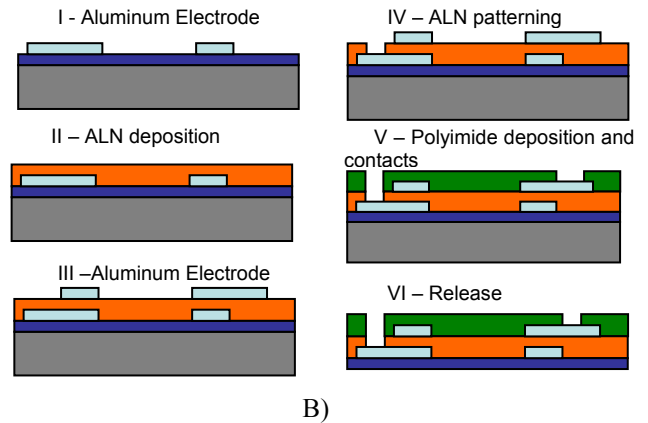


Fig. 5 Fabrication process A) LC sensors B) piezoelectric sensors

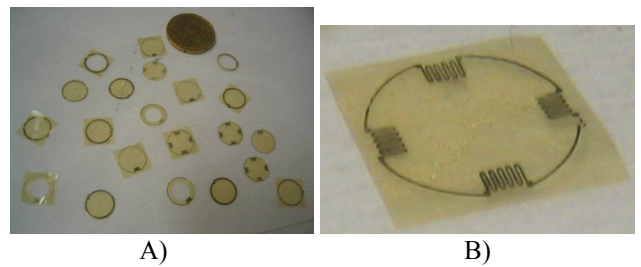


Fig. 6 Example of fabricated sensors A) metal TiW/Au on polyimide of $10\mu\text{m}$ thickness B) AIN based sensor on similar polyimide

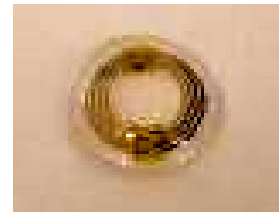
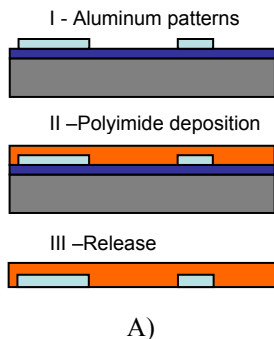


Fig. 7 Example of a fabricated LC sensor embedded into a soft lens



IV. IOP CHARACTERISATION SET-UP

In parallel, early characterization of the IOP increase versus the eye deformation has been performed to try to quantify the sensor sensitivity, with IOP variations measured in the range 20 to 70mmHg.

A. Set-Up Description

An experimental test bench has been set up to demonstrate the operation principle of the system using a proof of concept LC based prototype. Impedance measurements

have been performed with an HP4195 Analyser, an inductive antenna representing the reader (on glasses) being connected to the analyzer and the embedded sensor being put on an isolated pig eye 2cm under the reader.

In parallel, early characterization of the eye's deformation has been developed. The principle relies on micro injection of fluid (Ringer) inside the eye's inner chamber, with a volume step of 10µL, correlated to IOP measurement performed using a Tonolab TV02 (accuracy of ± 2mmHg).

B. First Experimental Results

The relation between the injected fluid and the measured IOP is shown on Fig.8. We will then use this simple principle to increase the intra ocular pressure in isolated pig eyes on which the sensor is electrically tested. We can observe on Fig.9 that the impedance magnitude and phase are slightly changing while the IOP is increased from 10 to 70mmHg, demonstrating the feasibility of such a sensor.

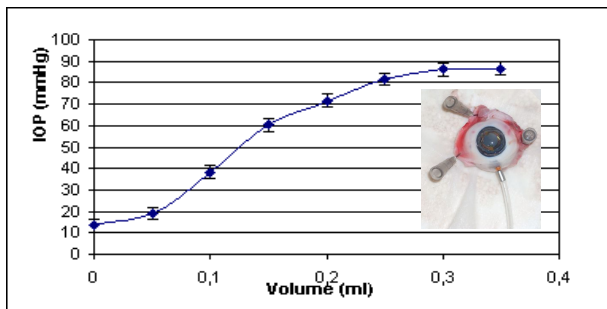


Fig. 8 Measured IOP vs. injected fluid volume

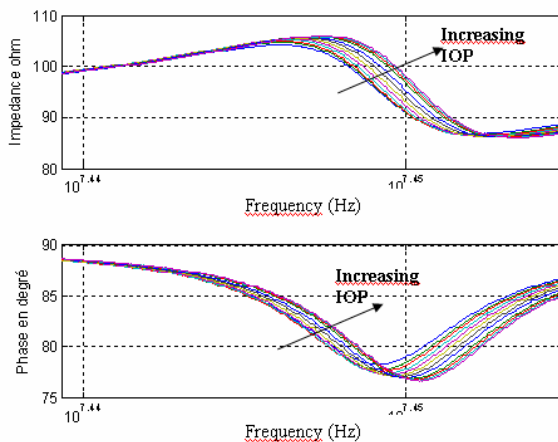


Fig. 9 Measured Impedance vs. frequency with IOP varying from 10 to 70mmHg

V. CONCLUSIONS

We proposed a sensor embedded into a transparent lens to measure the intra ocular pressure. First proof of concept prototypes have been fabricated and tested to assess the operation principle of the system. Further work is focusing on several ways. First, one study concerns the optimization of the existing sensors, and the development of alternative piezoelectric sensors to compare the performances with LC based sensors. Second, the electronic and communication protocol is under development in parallel. And third, the experimental set-up is being improved.

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Author: Lissorgues Gaëlle
 Institute: ESYCOM - ESIEE
 Street: Cité Descartes BP 99, 2 Bd Blaise Pascal
 City: 93162 Noisy le Grand
 Country: France
 Email: g.lissorgues@esiee.fr