35th Annual International Conference of the IEEE EMBS
Osaka, Japan, 3 - 7 July, 2013

Intraocular Electro-Optic Lens with Ciliary Muscle Controlled Accommodation*

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Abstract— **In this paper a concept is proposed of an intraocular lens implant with electro-optic accommodation of a variable-focus hybrid liquid-crystal-based lens. The dioptric strength of the lens is electronically controlled by a signal that is derived from the change of inductance of a sensing coil due to a marker implanted in the nearby contracting or decontracting ciliary muscle. Analytical, numerical and experimental results are reported on the dependency of the frequency of a Colpitts oscillator circuit on the location of a nearby conductive marker. A concept is also reported on the use as an electro-optic lens of a device based on a liquid crystal in planar alignment, which is held between a flat and a curved window coated with optically transparent and electrically conductive layers.**

I. INTRODUCTION

One of the important challenges in ophthalmology is to restore perfect vision of patients that were cured from cataract. Cataract is a disease that gradually clouds the crystalline lens of an eye. This clouding results in a blurred vision and eventually in blindness. It has been estimated that with the current demographic evolution about 40 million people will need cataract surgery by the year 2020 [1] which indicates the relevance of cataract prevention, diagnosis and treatment including full vision restoration.

The standard cure is to extract the clouded lens [2] and to replace it by a non-accommodating monofocal artificial intraocular lens (IOL). Multifocal lenses have different focal distances in different regions of the lens, similar to multifocal lenses in regular glasses. People wearing such lenses have simultaneously sharp images of different distances projected on their retina. This is confusing for the visual cortex and goes along with discomfort and stress. In addition, multifocal lenses only have a limited amount of focal distances, so that objects at intermediate distances cannot be seen sharp. Multifocal lens designs also lead to a number of problems such as lowered contrast sensitivity [13], [20], [21], halos [13] and direct light intolerance [13]. Better restoration requires the IOL to fulfil a fully accommodating function. Currently available accommodative IOLs are: singular accommodative [14], dual accommodative [16] and refilled lens [19], [18]. The singular and dual IOLs are based on a lens with a fixed focal distance that is allowed to be moved back and forth in the capsular bag of the eye guided by the visual cortex controlled (de-)contracting movement of the ciliary muscle. Lens refilling is done by mimicking the properties of the natural lens by making use of an elastic transparent polymer [18], [19]. The contraction state of the ciliary muscle then controls the shape of such deformable lens, and thus its dioptric strength. The most common issue with the singular accommodative lens is the so-called Z-syndrome [7], [12]. The lens refilling techniques suffer from long term instability, insufficient image quality and insufficient accommodative power [19]. Since motion of the IOL in the capsular bag depends on the stiffness of the capsular bag, which varies in time [8], it is expected that adjustments and thus repeated surgery will be required.

Inspired by solutions that have been developed for prescription glasses, in this work, we propose a fully self-adapting electro-optic IOL, which continuously adapts its dioptric strength so that the image of the object of interest remains in focus [4]. The mechanism used for automatic accommodation is very similar to the natural function of a perfectly working eye. In section II, the concept is given of an electro-optic lens with accommodation based on a feedback mechanism involving wireless sensing of the ciliary muscle contraction that is steered by the visual cortex. In section III, the concept for an inductive sensor is given together with experimental, numerical and analytical results. In section IV, the concept is given for a hybrid electro-optic lens based on a liquid crystal in a plano-convex volume. Conclusions for this research are given in section V.

II. CONCEPT OF AN ELECTRO-OPTIC LENS WITH ACCOMMODATION BASED ON WIRELESS INDUCTIVE SENSING OF THE CILIARY MUSCLE CONTRACTION

In a perfectly working eye, focusing on an object of interest in the field of vision by accommodation is based on a feedback mechanism involving the assessment of the retinal image sharpness by the visual cortex, and control of the eye lens shape and thus dioptric strength by the ciliary muscle. In particular, if the visual cortex detects that the image projected on the retina is out of focus, then a signal is sent to the ciliary muscle to contract or decontract thus altering the shape and refractive power of the crystalline lens (Fig. 1). Since in spite of frequent malfunctioning of the natural lens with growing age, the mechanism of people's sharpness assessment and ciliary muscle contraction keeps functioning during the whole lifetime of humans, an ideal

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* One of the authors (D.D.) is grateful to FWO-V for financial support (PhD scholarship 2012-2016).

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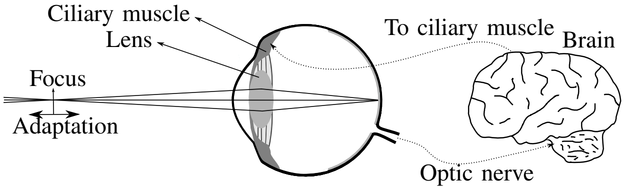

Fig. 1. Feedback principle in a human eye. The light rays from an object are refracted by the crystalline lens and form an image on the retina of the eye. The visual cortex processes the image and if it is not in focus, a signal is sent to the ciliary muscle to (de-)contract, thereby altering the shape of the crystalline lens and thus the refraction of the light and the image formed on the retina.

design of an accommodative IOL should keep making use of the well-performing feedback. As mentioned above, a scheme involving a mechanical connection and control of the IOL by the contraction state of the ciliary muscle is not sustainable due to the stiffness parameters of the eye lens and capsular bag holding the lens deteriorating over time [8]. Another option is to use the natural change of the contraction of the pupil [11] that goes along with the intentions of the visual cortex to change the dioptric strength. However, this feedback mechanism is fragile due to the pupil diameter responding even more strongly to the amount of light reaching the retina. Here we propose the use of a non-contact, induction based sensing principle in which a marker is implanted in the ciliary muscle and moves together with it during contraction, resulting in a change of distance of the marker with respect to a sensing coil placed on the statically mounted base that holds an electro-optic IOL in its centre. In this way the distance between the marker and the sensor is a measure for the contraction state of the ciliary muscle and for the degree of accommodation desired by the visual cortex. The measured inductance of the sensor coil is monotonically related with the marker distance, and thus serves as a suitable control parameter for electronically adapting the dioptric strength of the IOL, e.g. by changing the magnitude of an AC voltage applied to a liquid crystal based IOL.

Fig. 2 depicts the changes of the distance between the marker and the sensor coil that go along with changes of state of contraction of the ciliary muscle. Depending on the anatomy of the eye, the maximum distance between the detector and the marker is between 2 and 4 mm. During accommodation the ciliary muscle radially contracts over about 1 mm, bringing the marker correspondingly closer to the sensor. Together with the dioptric strength of about 40 diopters of the cornea the IOL forms a lens with a dioptric variation between 59 and 71 diopters in a young eye [5].

III. INDUCTIVE SENSOR

As mentioned above, the proposed principle for determining the distance between the marker and the sensor coil makes use of the monotonic dependence of the inductance L of the sensor coil on that distance. The here implemented detector circuit is a simple Colpitts oscillator, which is based on a feedback loop involving a resonant LC configuration and a transistor. The oscillation frequency f is given by:

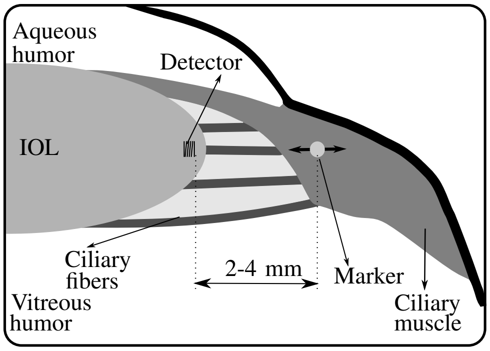

Fig. 2. Marker implanted in the ciliary muscle at a distance between 2 and 4 mm. When the eye accommodates the ciliary muscle moves the marker closer or further away from the sensor (maximum range of 1 mm). In this way the signal from the sensor is a measure for the contraction state of the ciliary muscle.

$$f = \frac{1}{2\pi\sqrt{L\frac{C_1C_2}{C_1+C_2}}}, \quad (1)$$

with L the inductance and C_1 and C_2 capacitances. The inductance L is given by :

$$L = \frac{N^2A}{l}\mu, \quad (2)$$

with N the number of turns, A the effective cross-section of the coil, l the length of the coil and μ the magnetic permeability of the medium in and around the coil.

The marker can exert an effect on the inductance via both μ and A. A marker made of permeable material ($\mu > \mu_0$) has an increasing effect on L when approaching the coil. When it is made of highly conductive material ($\sigma >> 0$), then it can conduct eddy currents and thus shields a region in the surroundings of the coil for the fluxlines created by the coil currents. This effect is equivalent to lowering A and thus decreases L as the marker comes closer to the coil. In view of the complex relation between the marker-coil distance d on one hand, and the effective coil cross-section A and medium permeability μ on the other hand, in the following, we verify the correspondence between experimentally determined $L(d)$ data and results from analytical and numerical simulations. The goal of this comparison is to validate the numerical simulation tool for this application, in order to safely use it in future for finding optimal coil and marker parameters that maximize the sensitivity of L to d in the distance range of interest [17] [15].

In the implemented experimental configuration, a spiral coil with a cross-sectional area of 14.14 ± 0.05 mm^2 and $n = 4$ windings and capacitances $C_1 = C_2 = 10$nF were embedded in a Collpitts oscillator circuit, featuring a resonance frequency of 8.6 MHz without marker. For the used copper marker (cylinder with diameter 2.67 ± 0.01 mm and hight 2.11 ± 0.01 mm), the corresponding skin depth δ, to which the flux can penetrate the marker, is only a few μm, which is much smaller than the marker's dimensions. As a consequence, the influence of the marker on the inductance was dominated by the eddy current effect. The oscillation frequency f of the circuit was monitored by means of a phase locked loop circuit, while adapting the distance between the centre of the marker and the coil between 1 and 5 mm. In order to be able to compare the data with simulation and the analytical results (fig: 3), a dimensionless quantity F_S is defined:

$$F_s = \frac{\frac{1}{f^2} - \frac{1}{f_*^2}}{\frac{1}{f_0^2} - \frac{1}{f_*^2}}, \tag{3}$$

with f_* the frequency without marker and f_0 the frequency with a marker at 0 mm. In order to model the dependence of the inductance on the marker distance, we make a number of simplifications. Given an electromagnetic wavelength at 8.6 MHz of about 35 m, much larger than the dimensions of the detection configuration, we use the quasi-static approximation of the Maxwell equations, so that the interaction between the marker and the coil essentially happens via the magnetic induction $\boldsymbol{B}$ created by the coil. The magnetic induction is related to the magnetic energy density η by [6]:

$$\eta = \frac{\boldsymbol{B}^2}{2\mu_0}. \tag{4}$$

For a single loop coil in empty space with radius R, carrying a current I, the magnetic induction along the central axis x of the coil is analytically given by [6]:

$$\boldsymbol{B}_x = \frac{\mu_0 I R^2}{2(R^2 + x^2)^{3/2}} \hat{\boldsymbol{x}}. \tag{5}$$

The pre-factors are scaled out by defining a dimensionless quantity η_S (Fig. 3) which can be compared to experimental and numerical simulation:

$$\eta_S = \frac{\eta - \eta_\infty}{\eta_0 - \eta_\infty}, \tag{6}$$

with η_∞ and η_0 the magnetic energy density at infinity and 0 mm respectively. Additionally a numerical simulation was carried out, assuming a single loop circular coil with radius R_{model}, and a spherical marker with radius r_{model}. Their sizes were chosen such that the ratio of their surfaces is the same as the ratio of the cross-sections of the spiral coil A_{coil} and of the cylindric marker A_{marker} used in the experiment:

$$\frac{A_{\text{marker}}}{A_{\text{coil}}} = \frac{r_{\text{model}}^2}{R_{\text{model}}^2}. \tag{7}$$

The magnetic energy M of the system is rescaled, defining a dimensionless quantity M_S (Fig. 3) in order to be comparable to the experiment and the analytical derivation:

$$M_S = \frac{M - M_*}{M_0 - M_*}. \tag{8}$$

with M_* the magnetic energy without marker and M_0 the magnetic energy at 0 mm. In Fig. 3 the marker position dependencies on the scaled experimental data F_S, the scaled simulation data M_S and the scaled analytical data η_S are compared. It turns out that, after proper normalizations, the marker-coil distance dependency on the experimentally determined oscillator frequency, corresponds very well with the analytically derived magnetic energy density and the numerically simulated magnetic energy. It is reasonable to assume that these normalized quantities can thus be exploited to predict $f(d)$ curves in more complicated geometries, enabling numerical simulations of $\eta(d)$ to find parameters optimizing their sensitivity to the marker position.

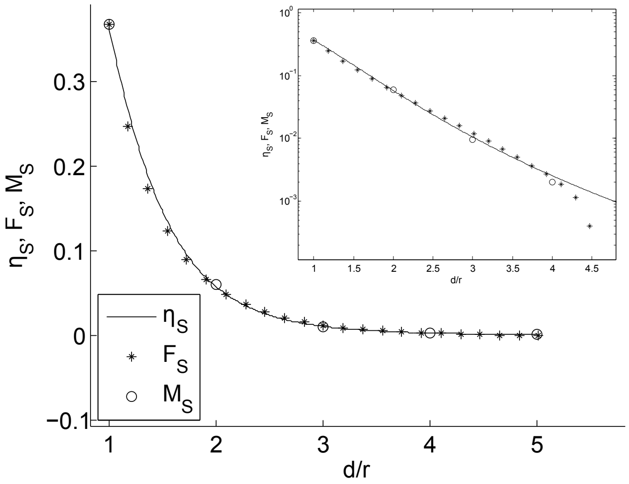

Fig. 3. Rescaled values for the distance dependent influence of a copper marker on the inductance of the detection circuit. A linear-log plot is included in the inset. It is clear that theory η_S, model M_S and experiment F_S are in good agreement.

IV. ELECTRO-OPTIC LENS

A. *Electro-optic lens for IOL*

Accommodation of an eye requires an IOL to control its dioptric strength and in this way the refraction of the incoming light according to the focal distance desired by the visual cortex. The natural mechanism to achieve this, is by changing the convexity of the eye lens via contraction of the ciliary muscle. Soon after placing a lens implant, the elasticity of the capsular bag is degrading, so that the mechanical coupling between ciliary muscle and lens deteriorates [8]. In view of this, we here propose to replace the concept of shape control by electro-optic control of the dioptric strength, while still making use of the (typically lifelong intact) feedback control information, i.e. the state of contraction of the ciliary muscle. A common way to electro-optically control the dioptric strength of an optical element is by making use of electrowetting effects for electrically tuning the convexity of a liquid or gel droplet. However, this solution requires rather high voltages and power consumption, and has to cope with evaporation of the liquid, and typically limited optical window diameter [10] [9].

Also electro-optic control of a rectangular or concentric geometry Fresnel or holographic type liquid crystal matrix is an interesting candidate for electro-optic lens accommodation, with the disadvantage that the optical response is different depending on the polarization of the incoming light [3].

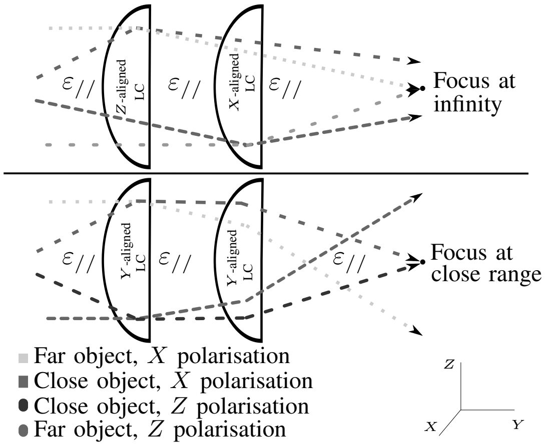

Fig. 4. Serial system of two plano-convex, liquid crystal lenses embedded in a medium with $\varepsilon = \varepsilon_{//}$.Without electric field, the first lens only affects light with a x-polarisation while the second lens only affects light with a z-polarisation. By rotating the liquid crystal molecules in the lenses the refraction for respectively z and x-polarisations change.

B. Optical device containing a convex / concave liquid crystal compartment as an electro-optic lens

Our approach, which aims at avoiding the complications mentioned above, makes use of a convex or concave shaped liquid crystal volume, with electrical control of the nematic director alignment, via a combination of surface treatment and optically transparent, electrically conductive electrodes. Nematic liquid crystals are characterized by uniaxial anisotropy of the dielectric permittivity ($\varepsilon_{//}$ for the long axis and $\varepsilon_{\perp}$ for the short axis) so that the refractive index for incoming light depends on the orientation of the polarization with respect to the nematic director. In a curved geometry, this results in different focal points for the ordinary and extraordinary polarization components respectively. The focal distances are determined by the radius of curvature of the liquid crystal compartment surfaces, and by the mismatch between the refractive indices of the liquid crystal and the compartment material. The compartment surfaces are treated such that without external electric field the alignment is planar, with the director parallel to the surfaces (z-direction). An electric field applied in the normal (y-) direction then rotates the director in the yz-plane, thus changing the index of refraction for the z-polarisation component, and leaving the one for the x-polarisation component unaffected. This configuration allows to move the extra-ordinary focal point f_e without moving the ordinary focal point f_o. For proper functioning in an accommodation scheme, the focal point of the electro-optic assembly should be polarization independent. This can be achieved by a configuration of two serial lenses with orthogonal planar orientations, each refracting the respective extraordinary polarization component to the same focal point (Fig. 4) [22].

V. CONCLUSION

Responding to the clear need for self-accomodating intraocular lenses, in particular for restoring peoples natural vision by implantation after cataract surgery, this paper proposes a concept based on an electro-optic lens, whose dioptric strength is controlled by a steering signal on the basis of an inductive detection of a marker implanted in the ciliary muscle. The IOL does not make use of any moving components inside the capsular bag, which is a major advantage, considering the changing elastic properties of the ageing human eye over time. The inductive sensor provides sufficient resolution for distance measurement of the marker. Two convex or concave liquid crystal compartments act as an electro-optic accommodating lens and focus both polarisations of light.

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