Design and ex situ performance of a shape-changing accommodating intraocular lens

ANDRES DE LA HOZ,1,2,* JAMES GERMANN,1 EDUARDO MARTINEZ-ENRIQUEZ,1 DANIEL PASCUAL,1 NANDOR BEKESI,1 NICOLAS ALEJANDRE-ALBA,3 CARLOS DORRONSORO,1 and SUSANA MARCOS1

1Instituto de Óptica “Daza de Valdés”, Consejo Superior de Investigaciones Científicas (IO-CSIC), Madrid, Spain
2Departamento de Ciencia de Materiales, Universidad Politécnica de Madrid, Spain
3Departamento de Oftalmología, Hospital Universitario Fundación Jiménez-Díaz, Madrid, Spain
*Corresponding author: andres.delahoz@csic.es

Presbyopia, the age-related loss of the crystalline lens’s ability to dynamically focus, occurs primarily because of stiffening of lens material, making the ciliary muscle forces insufficient to reshape the lens. Despite its prevalence, there is no satisfactory solution to presbyopia. Here we present a novel accommodating intraocular lens (AIOL) able to reshape upon equatorial forces in compliance with the eye’s accommodating mechanism. The concept and design parameters are demonstrated through finite element model simulations and measurements in a manufactured AIOL prototype, using custom quantitative 3D OCT (geometrical changes) and laser ray tracing (power changes), with forces radially applied using a custom eight-arm mechanical stretcher. There was an excellent agreement between simulations and measurements (1% for the focal length and 11.4% for geometrical parameters, on average) for radial load up to 0.6 N. The developed design is expected to achieve ~2.5D of effective power change with a polymer material with 0.10–0.25 MPa Young’s modulus and $n = 1.43–1.46$. © 2019 Optical Society of America under the terms of the OSA Open Access Publishing Agreement

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1. INTRODUCTION

Accommodation is the ability of the eye to change its focusing power [1]. According to the widely accepted theory of accommodation by Helmholtz, the crystalline lens reduces its optical power to focus at infinity when the ciliary muscle relaxes, increasing the tension of the zonules, which connect the crystalline lens to the ciliary muscle. This exerts a force on the lens [2], which expands its equatorial diameter and flattens its surfaces, therefore lowering its optical power. Presbyopia is the age-related process of progressive loss of accommodation capability in individuals. The most widely accepted theory for the cause of presbyopia is the lenticular theory, which states that the crystalline lens becomes stiffer with age. A number of ex vivo studies have shown age-related increases in stiffness of the crystalline lens’s cortex and nucleus [3–5] as well as an increase in force required to change lens power for older lenses [6], giving credence to this theory. Some age-related changes in ciliary muscle function have also been reported, but the general ability of the muscle to contract and expand remains [7–9].

Presbyopia is typically treated by providing near vision aids in the form of spectacles of positive power, progressive addition spectacles, monovision (correction in one eye for far and the other for near), presbyLASIK, or multifocal contact lenses. Multifocal corrections also exist in the form of intraocular lenses (IOLs), which operate on the simultaneous vision principle (projecting both near and far focused images on the retina). These are currently on the market [10], but due to their mechanism, they do not provide dynamic change in power. Several intraocular solutions to restore dynamic accommodation have been investigated and proposed. One such proposal is replacing the stiffened contents of the aged lens with a softer, more malleable synthetic polymer refilling the capsular bag. This technique is challenged by posterior capsule opacification and has yet to reach the market [11,12]. A foreseen alternative to restore dynamic accommodation is use of accommodating intraocular lenses (AIOLs) to replace the crystalline lens in a cataract surgery, or even at an earlier age, replacing a stiff but transparent crystalline lens to correct presbyopia. The only FDA-approved AIOL (Crystallens by Bausch and Lomb, Rochester, New York) is expected to offer dynamic change in power through shifts in the position of a single-element optics. However, despite the claim, studies have shown that the axial movements of the implanted Crystallens AIOL are in fact very small and backward, instead of forward [13,14], and have suggested that the improved near vision could in fact result from pseudo-accommodation arising from increased aberrations [15]. Other single- or dual-optics axial shifting IOLs have been presented, but their mechanisms have not been shown to be successful.

Fully accommodating intraocular lenses capable of dynamically changing their optical power through a change in their surfaces as a response to stimuli from the ciliary muscle would be an
ideal solution for presbyopia, mimicking the physiological accommodative process. These types of lenses require a mechanism to translate the forces and actions from the ciliary muscle to the accommodating lens. Some of the methods proposed involve the use of the entire capsular bag to reshape the lens and thus depend on capsular integrity [16]. An example of a capsular bag proposal is FluidVision (PowerVision), in which the capsular bag is expected to squeeze peripheral fluid reservoirs, which then displace fluid into or out of a central optic, which becomes rounder or flatter in response. Some proposed methods depend on the natural fibrosis process in order to properly attach a structure to the capsule [17,18]. Although the need for capsular engagement with the lens haptics has been acknowledged by some authors, and suggestions for using biocompatible adhesives such as fibroin [19] or mechanical means [20] have been made in this context, to our knowledge, those have not been materialized, except for the Z capture haptics, where the engagement of the haptic platform to the capsular bag relies on natural fibrosis [21].

We have previously proposed and demonstrated the use of photochemical bonding as a method for guaranteeing engagement between an IOL and the lens capsule. The method involves using a photosensitizer and light irradiation to create chemical bonds between different types of materials, which we have demonstrated using Rose Bengal, an FDA-approved dye already used in ophthalmology, as a photosensitizer, and a green light source for irradiation. Rose Bengal photobonding (RBPh) had been shown to be effective for sealing corneal incisions [22] and for bonding amnion to the cornea [23]. We have shown that it is possible to use this photo-activated technique to bond polymers such as those typically used for contact lens and IOL manufacturing to capsular tissue [24] as well as full intraocular lenses following phaco-emulsification in a porcine eye model [25].

The ability to bond polymers to the capsule through RBPh opens a path for AIOLs that fully engage with the accommodative apparatus and mimic the crystalline lens’s behavior more closely. Furthermore, this concept does not require full capsular bag integrity to operate. We have proposed a surface-changing AIOL in which the haptics, shaped as transverse curved plates of approximately 1 mm height, are photochemically bonded to the equatorial capsular region, allowing for the transmission of forces from the ciliary muscle and the reshaping of the AIOL [26].

This concept requires a material or materials with specific optical and mechanical properties in order for the accommodative apparatus to successfully change the AIOL power to a relevant degree. In this paper, we explore the viability of this concept by estimating the power and surface changes in mechanical and optical simulations of this AIOL design, as well as experimentally in prototypes manufactured using a commercial ophthalmic polymer.

2. METHODS

A. Surface-Changing Accommodating Intraocular Lens

The proposed AIOL consists of two elements. The first is a deformable optic, which responds dynamically to the forces transmitted by the ciliary muscle in order to change the AIOL’s shape and thus its power. The deformable optic has eight haptics with a curved external surface, shaped to adapt to the equatorial region of the capsular bag. The external surface of these haptics would be coated with Rose Bengal, allowing them to be photochemically bonded to the region of the capsular bag where the zonulæ connect, and for the force of the ciliary muscle to be transmitted to the AIOL and deform its surfaces. The second element is a thinner (0.35 mm) optic that is not directly engaged to the capsular bag, and its refractive power remains constant. This optical element is placed in front of the deformable optical element, and its posterior surface is in contact with the deformable optic’s anterior surface. The role of this element is to compensate for the subject’s refractive error, with the sum of the two optics’ powers ensuring a consistent focal length in the fully accommodated state.

A prototype of the deformable component of the AIOL was designed and manufactured in a five-axis lathe machine. The prototype has an overall diameter of 11 mm and an optical region diameter of 5 mm, a central thickness of 1.433 mm, and anterior and posterior surface radii of curvature of 12 mm and 5 mm. The material used for the prototype was a contact lens material (Contaflex 42%, Contamac Ltd.) with a refractive index of 1.42 and an equivalent Young’s Modulus of 0.55 MPa. The material’s stress-strain behavior was measured experimentally with a uniaxial extensometer (Instron 5543A, Instron). The stress-strain curve was used to model the material as a hyperelastic, two-term Mooney–Rivlin material. The haptic structures of this prototype, unlike those for the proposed AIOL, are flat rather than curved. This is because the purpose of the prototype was to evaluate change after direct force application in a mechanical stretcher, requiring the clamping of the haptics on a flat surface.

B. Mechanical and Optical Simulations

The ANSYS Workbench platform was used to develop a finite element model of the AIOL prototype. In order to reduce the computational load of the simulations, and taking advantage of the design symmetries, the lens was modeled with symmetry along the X-Y and Y-Z planes and represents 1/4th of the total volume of the prototype [Fig. 1(a)]. Radial forces of up to 0.6 N were applied to the AIOL haptics [Fig. 1(b)]. Nodal surface data for the AIOL’s anterior and posterior surfaces were extracted for different substeps of the simulation as well as the corresponding force. The nodal surface data was fitted to a circle in order to obtain numerical values for the anterior and posterior surface radii and to calculate the thickness. With these numerical values and the applied force, a force-deformation relationship was obtained.

The thickness and the anterior and posterior radii values obtained from ANSYS were then used as inputs in ZEMAX [Fig. 1(c)] in order to create a model of the laser ray tracing (LRT) experimental setup (discussed below). The focal lengths of the AIOL for various stretching states were obtained from the model.

C. Ex vivo Accommodation Simulator

A custom-developed eight-arm mechanical stretcher was used to apply a uniform deformation to the AIOL prototype (Fig. 2). The design of the mechanical stretcher is based on the EVAS II system, which has been used to characterize crystalline lenses [27]. The device consists of eight stretching arms arranged radially around a 10 cm plastic chamber with a 5 x 5 mm glass floor. The base of each arm is a computer-controlled translation stage (M-111 Compact Micro-Translation Stage, Physik Instrumente) with a displacement range of 15 mm and a minimum step size of 50 nm. Each motor has been mounted with a force transducer (Fort 10 g, WPI). These force transducers use semiconductor strain gauges to
produce a linear output force voltage with an applied force, through a sensing leaf, with a force range of 10 g and a resolution of <1 mg. An aluminum structure with an L-shaped arm up front is also mounted on the base at the same height as the sensing leaf of the force transducer and held in place through spring leafs. Fixation shoes are mounted on the arm. A small rod passing through the sensing leaf and the structure connects the sensor and arm and allows for a force measurement when the AIOL is stretched via the translation stage. The force measurement for each arm was calibrated using 13 μm thick glass fibers with a known linear stress-strain relationship, previously measured with the uniaxial extensometer. The lens is mounted directly onto the stretcher, and each of its eight flat haptics is clamped with the fixation shoes.

D. Laser Ray Tracing: Optical Power Measurements

To characterize the focal length of the AIOL, a custom ray tracing system was used [LRT in Fig. 2(a)]. A schematic of the system can be seen in Fig. 3. A He–Ne laser (λ = 632 nm) is used as a light source. A 2D galvanometric scanning system (Mod. 6210, Cambridge Technology, Medford, Massachusetts) is used to generate a ring pattern of selected diameter. This ring pattern is reflected by various mirrors and scanned onto the surface of the AIOL, which is mounted on the mechanical stretcher. Below the chamber, a CMOS camera (UC480, Thorlabs) attached to a translating stage (UniSlide, Velmex Inc.) is used to capture through-focus images at the focal plane of the AIOL. The images are analyzed using a MATLAB program that calculates ring size based on the number of pixels with maximum intensity for every image; it finds the minimum ring size and its corresponding height. The system was calibrated using commercial glass lenses of known focal length (Edmund Optics Inc. Barrington, New Jersey, USA).

E. Optical Coherence Tomography: Surface Shape Measurements

A custom-built spectral-domain OCT system was specifically built for this study and used to image the anterior and posterior surfaces of the AIOL [OCT in Fig. 2(a)]. A schematic of the system can be seen in Fig. 4. The setup was based on a fiber optics...
Michelson interferometer configuration. The light source was a superluminescent diode (Superlum SLD-371, Superlum) of central wavelength of 840 nm and bandwidth of 56.2 nm. A spectrometer consisting of a holographic Bragg transmission grating (HD 1800 l/mm @ 840 nm, Wasatch Photonics) and CCD line camera (spl4096-140 km Basler Sprint, Basler) was used to record the interference fringes. The axial acquisition rate of the system was 50,000 A-scans/s, with an axial range of approximately 5 mm and a theoretical pixel resolution of $5.2 \mu m$, and a roll-off sensitivity of $-2.4$ dB/mm. The output of the system is in the form of B-Scans, cross-sectional images of the AIOL, which can be used to reconstruct its volume as a fully three-dimensional object.

The acquired data is processed to correct for fan and optical distortion; the former is produced by the scanning architecture of the spectral OCT (sOCT) system [28] and the latter is produced by refraction caused by the optical surfaces being characterized [29]. The automatic surface segmentation is based on algorithms described previously [30,31]. Specific routines that improve the AIOL detection for the experimental conditions have been designed for this study. The system's quantification capacities were evaluated using glass spheres of known radii.

Further analysis of the 3D lens included fits of the anterior and posterior surfaces to conic surfaces (described by radii of curvature $R$ and asphericity $Q$), comparison to the surface profile across different meridians, and virtual ray tracing analysis to estimate the wave aberration (and its change with accommodation).

**F. Experimental Protocols**

The three experimental components (mechanical stretcher, LRT, and sOCT) were placed on an optical table and co-aligned [Fig. 2(c)]. The AIOL prototype is mounted in the mechanical stretcher with each haptic clamped in each mechanical arm. Care is taken to ensure that no excessive compression is being applied onto the haptic. The chamber is filled with distilled water in order to conduct the test in immersion and maintain the AIOL consistently hydrated. A small initial force (0.01 N approx.) is applied to the AIOL in order to reduce slack from the mounting.

After mounting the AIOL, the mechanical stretcher was used to apply a consistent displacement to each of the 8 arms, stretching the AIOL radially, and the resulting forces were recorded. For each trial done, displacement was linearly applied in five substeps, and this cycle was repeated five times, with measurements (image capture of the light spot) taken at each substep of each cycle, for a total of 25 measurements. Each LRT measurement consisted of a series of spot images taken sequentially with a motor step size of 0.25 mm. The images are processed using MATLAB in order to calculate the spot size relative to distance and to calculate the focal length of the AIOL in the system.

The same trial structure of five substeps and five cycles was done to obtain cross-sectional images using sOCT. For these measurements, 300 B-Scans (of 300 A-Scans per B-Scan) were taken for a measurement space of 6 mm by 6 mm by 5 mm across the x, y, and z axes. These images were then processed in MATLAB to obtain numerical values for surface radii and thickness after reconstructing the AIOL volume.

Two AIOL prototypes of the same dimensions were measured multiple times in the system. Surface curvature and thickness results from sOCT were compared to predicted results from mechanical simulations. Focal length results from the LRT system were compared to predicted results from optical simulations.

**3. RESULTS**

**A. AIOL Optical Changes upon Accommodation**

Figure 5 shows the results from the LRT experiments. Three trials of the experiment are shown on the graph, presented as the relationship between stretching force and the measured change in focal length. Each trial consists of a stretching cycle divided into five substeps (a zero measurement and force applied in four
substeps). This stretching cycle is repeated 5 times per trial for a total of 25 measurements. The average standard deviation of the change in focal length, per trial, is of 0.067 mm. The average standard deviation of the force is of 0.0124 N. The maximum force represents an increase of 0.31 mm in overall diameter.

A linear fit of the experimental data is displayed, as well as the predicted results calculated from the finite element results and optical simulations. The predicted focal length change is linear, and the experimental results show a good linear fit. The average $R$ for a linear fit of each trial is 0.978. The average slope for a linear fit of each trial is 3.03 mm/N, compared to a predicted slope of 3.00 mm/N. The experimental results show a $y$ intercept for the linear fit, which has an average magnitude of 0.1 mm (standard deviation of 0.016 mm). This magnitude is below the height resolution of the experiment (0.25 mm) and is reduced when the AIOL is pre-stretched before the trial.

B. AIOL Surface Curvature Changes upon Accommodation

Figure 6 shows (a) raw cross-sectional images, (b) en face images, and (c) 3D reconstruction of the AIOL mounted in the stretcher, obtained from sOCT. Visualization 1 shows a dynamic visualization of the changes undergone by the AIOL during stretching from OCT. The AIOL flattens, the equatorial lens diameter increases, and the lens thickness decreases upon disaccommodation; while it becomes more curved, the equatorial lens diameter decreases and the lens thickness increases when equatorial forces are released, similarly to the young crystalline lens function in the eye.

Figure 7 shows results from OCT experiments comparing experimental and predicted results for AIOL thickness and anterior and posterior surface radii [Figs. 7(a)–7(c)]. Results are presented for trials in which the anterior AIOL surface faced the OCT beam.

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**Fig. 6.** (a) Central B-scan of AIOL mounted on mechanical stretcher. (b) 3D reconstruction of the AIOL. Top view. (c) Snapshot of the 3D reconstruction of the AIOL, unstretched state. $X - Y$ dimensions are 6 mm by 6 mm.

**Fig. 7.** AIOL geometrical parameters as a function of the applied force obtained from OCT system. (a) AIOL thickness. (b) Anterior surface radius of curvature. (c) Posterior surface radius of curvature.
anterior surface and dians). The measured surface asphericities were the topographic profile of the deformed lens (average difference minor, with a difference of only around 5 μm). The anterior surface curvature increases. These observations match the behavior of the crystalline lens to previously measured values, either accounting only for the lens [32] or including the zonular and ciliary muscle elasticity and movement [33]. Experimentally, estimates of the ciliary muscle forces are obtained by stretching the crystalline lens, recording applied force using a mechanical setup, and comparing the changes in power and surface shape to those obtained by other methods [6]. There is experimental evidence that the changes in power and shape the crystalline lens can be accurately represented through the application of an equatorial force, without requiring the input of additional forces such as IOP [34].

The forces used in our stretching experiment are higher (up to 0.6 N) than those expected from the accommodative mechanism in order to evaluate the experimental platform at its full capacity. For the forces available in the eye, the prototype, manufactured with off-the-shelf hydrophilic material, achieved 0.6D of change, slightly lower than the 1D accommodative range required by the FDA to claim accommodation. In order to assess the mechanical and optical properties necessary for this AIOL concept to show a larger change in power (>2D) for reduced equatorial forces, mechanical and optical simulations were conducted for a range of material properties between 0.1 and 1.0 MPa equivalent Young’s modulus, and a 1.40–1.46 range of refractive index. The force applied to the simulated AIOL was compliant to a conservative value of the forces presumably available (0.08 N). For this analysis, unlike the prototype simulations, the power change was obtained from a simulation of the AIOL in a model eye in order to assess implanted AIOL performance.

The results of these simulations are presented in Fig. 8. The simulations show that the range of suitable mechanical and optical properties required for adequate change in power due to change in the shape of the AIOL surface is between 0.1 and 0.2 Young’s modulus and 1.43–1.46 refractive index. Due to the reliance on deformation, this AIOL concept is stiffness driven: as can be observed in the figure, there is an inflection point (around 0.3–0.35 MPa) after which the material is insufficiently deformable and thus unable to result in any meaningful change in power despite higher refractive index values. Before this point, the AIOL deforms, changes its surface shape, and results in a change in power, which increases as the refractive index increases. The optimal material would have an equivalent Young’s modulus of material properties between 0.1 and 0.2 Young’s modulus and a 1.40–1.46 refractive index.

4. DISCUSSION

The experimental behavior of the AIOL, measured using quantitative techniques (LRT and 3-D OCT) experiments confirms the predictions of the simulation. Upon stretching (disaccommodation), the AIOL power decreases, thickness and anterior surface radius decrease, and the posterior radius of curvature increases. Upon accommodation, the AIOL thickness and anterior surface increase, the posterior surface curves, and the power of the AIOL increases. These observations match the behavior of the crystalline lens during accommodation, except the direction of changes in anterior surface curvature.

The force exerted on the crystalline lens during the accommodative process is estimated to be in the range of 0.08–0.1 N. This estimate has been obtained through different approaches, computational and experimental. Computationally, finite element models of the crystalline lens have been used to estimate the force necessary to change the diameter, thickness, and curvature of the lens to previously measured values, either accounting only for the lens [32] or including the zonular and ciliary muscle elasticity and movement [33]. Experimentally, estimates of the ciliary muscle forces are obtained by stretching the crystalline lens, recording applied force using a mechanical setup, and comparing the changes in power and surface shape to those obtained by other methods [6]. There is experimental evidence that the changes in power and shape the crystalline lens can be accurately represented through the application of an equatorial force, without requiring the input of additional forces such as IOP [34].

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between 0.15 and 0.25 MPa for a refractive index between 1.44 and 1.46.

The prototype evaluated in this work is optically and mechanically homogeneous, unlike the real crystalline lens, which has heterogeneous mechanical and optical properties. The refractive index of the crystalline is not constant, but rather a gradient composed of layers with varying crystallin concentrations. Maximum concentration of crystallins in the central zone of the lens results in a higher refractive index, which is then reduced as the concentration of crystallins decreases towards the periphery of the lens. This gradient, in which the maximum ranges from 1.40 to 1.443 and the minimum from 1.362 to 1.39 [35,36], results in an “equivalent refractive index” that is higher than the refractive index values measured throughout the lens [37].

The crystalline lens is also believed to have spatially varying stiffness, in particular between the two zones that are classified as the nucleus and cortex. Measurements of the crystalline lens during accommodation show thickness changes occurring primarily on the nucleus [38]. Mechanical testing of the lens using indentation [39] and spinning tests [4] as well as optical methods such as Brillouin microscopy [40,41] have shown variations in lens stiffness across the lens. These spatial variations in stiffness can affect the overall performance of the lens during accommodation, changing the distribution of deformations and the curvature of the surfaces [42]. It is possible, then, that processes by which mechanical and optical properties of polymers can be targeted in a localized manner can reduce refractive index targets and/or increase stiffness targets.

Besides its mechanical performance, we found that the surface asymmetries produced by the finite number of haptics are minor. The optical performance of the prototype is not very different from the natural lens or standard monofocal IOLs (negative asphericities in the lens surfaces in the accommodating state; RMS wavefront error dominated by spherical aberration). The effect of an increase in magnitude of the RMS with accommodation also occurs in the natural crystalline lens with similar order of magnitude [43], although the effect of a positive (rather than negative) shift in spherical aberration should be further investigated in a model eye (with a cornea in front of the lens) and considering pupillary miosis. The optical zone of the lens (5 mm) complies with ISO 11979 standards regarding optical zone diameter of IOLs.

A strong advantage of the proposed AIOL is the reduced reliance on posterior capsular bag integrity. Some proposed accommodating IOLs, such as FluidVision or LensGen, require fuller capsular bag integrity in order to function [44]. In the proposed design, only the equatorial region of the capsule is relevant to the lens function; any further issues with the capsule (including posterior capsule opacification) can be resolved by removing its central regions. While for the experimental tests presented here the haptics were manufactured as flat plates (to facilitate engagement with the stretcher), the devised haptic platform with curved plates spanning a 1 mm transversal height region in the equator will facilitate stability and strong contact with the peripheral region of the capsular bag. While a peripheral engagement of the haptics to the capsular bag still needs a relatively good match between the haptic platform dimensions and the equatorial diameter of the capsular bag, the dependency on bag dimensions is largely attenuated for the following reasons: (1) the lens is implanted at the maximally unaccommodated state (using a capsular ring-like device to stretch the lens, with the diameter of the ring being easily adjustable), (2) the lens is not reshaped by the capsular bag itself and is therefore relatively independent of it. The size variability can be accounted for by having a range of slightly varying diameters and selection of the refractive component according to the achieved power of the flexible component in its maximally stretched diameter. Our group has previously developed a method for estimating volume and equatorial diameter of crystalline lenses using OCT [45], which would help in the selection of an appropriate diameter and for calculation of required refractive lens power.

Besides the optical component of this concept, further work is required on the mechanical engagement of AIOL haptics to the crystalline lens capsule. We have previously shown that these types of materials can be bonded to capsular tissue and have demonstrated IOL–capsular bag bonding both ex situ and intraocularly, with the forces required to break the bond being higher than those expected during accommodation [24]. However, current approaches have involved dyeing the polymer and capsule with Rose Bengal indiscriminately. AIOL implantation would require a controlled quantity of Rose Bengal, concentrated on the external surface of the haptics. This could be delivered in different ways, such as microfluidic channels within the haptics, deposition of RB through chemical processes, or even a simple soaking procedure. The photobonding procedure would also need to be optimized to minimize the quantity of light used while obtaining bonds sufficiently strong to withstand the forces in the accommodative process. Approaches for this problem involve haptic design tuning, a design of a light probe specifically tailored to deliver light to the haptics.

5. CONCLUSION

We have shown a prototype that is found to change its surface shape, thickness, and focal length as predicted from optical and mechanical simulation. The design is expected to achieve 2–2.5D of effective power change with a material of $E = 0.10 \text{ MPa} \text{ and } n = 1.43 - 1.46$. Further work will focus on material tuning and haptic design and implantation.

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