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Optomechanical Response of Human and Monkey Lenses in a Lens Stretcher

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Abstract

Purpose—To quantify the forces necessary to change the shape and optical power of human and monkey lenses.

Methods—Cynomolgus monkey ($n = 48$; age: 3.8–11 years), rhesus monkey ($n = 35$; age: 0.7–17 years) and human ($n = 20$, age 8–70 years) eyes obtained postmortem, including the lens, capsule, zonules, ciliary body, and sclera were mounted in an optomechanical lens-stretching system. Starting at zero load, the lenses were symmetrically stretched in a stepwise fashion in 0.25- or 0.5-mm steps. The load, lens diameter, inner ciliary body diameter, and lens power were measured at each step and the diameter- and power-load responses were quantified.

Results—The diameter- and power-load responses were found to be linear in the physiologically relevant range of stretching. The average change in cynomolgus, rhesus, and human lens diameter, respectively, was 0.094, 0.109, and 0.069 mm/g in young lenses, and 0.069, 0.067, and 0.036, mm/g in older lenses. For the same lenses, the average change in lens power was -3.73 , -2.83 , and -1.22 D/g in young lenses and -2.46 , -2.16 , and -0.49 D/g in older lenses.

Conclusions—The force necessary to change the lens diameter and lens power increases with age in human and monkey lenses. The results agree with the Helmholtz theory of accommodation and with presbyopia theories that predict that the force required to disaccommodate the lens increases with age.

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Several intraocular lens designs and surgical approaches have been proposed to restore dynamic accommodation in the presbyopic eye.^{1–5} One of the fundamental hurdles in the development of such approaches is the absence of a definitive explanation for the progressive decrease in the accommodative ability with age, which leads to presbyopia. Most likely, presbyopia results from a combination of factors affecting several of the anatomic structures involved in accommodation.^{6–9} Most of the current theories of presbyopia attribute the loss of accommodation directly or indirectly to age-related changes in the lens itself.^{7,10–14} Alternatively, Strenk et al.⁶ recently suggested that presbyopia may also be due to an inward movement of the ciliary muscle that occurs with age and reduces the tension on the zonules when the ciliary muscle is relaxed. Croft et al.¹⁵ proposed that a reduction in the forward movement of the ciliary body could contribute to the decrease in lens accommodation with age.

The relation between the changes in zonular tension and the resultant changes in lens shape and dioptric power is a key characteristic of these current theories of presbyopia and of the accommodation mechanism in general. The relation between the lens biometric and optical responses has been relatively well characterized in vivo.^{16–23} In contrast, the relation between the forces of accommodation and the resultant biometric and optical responses remains to be quantified. Unfortunately, there is currently no suitable technique for direct measurements of the accommodation force in absolute, physical units in an in vivo setting.

Direct quantitative measurements of the force–power relation have been obtained on human tissue postmortem by simulating the force exerted on the zonules during disaccommodation. The first studies of this kind appear to be those of Pau,²⁴ who simulated the force of accommodation on the lens by attaching weights to the ciliary body. Fisher²⁵ later designed a device that mimicked the pull on the zonules by spinning isolated lenses to produce a centrifugal force and a second, more sophisticated, system²⁶ that simulated accommodation by radial stretching and recorded the changes in lens focal length, curvature, and thickness. Fisher estimated the stretching force indirectly, by comparing the thickness measured in the stretching device with the thickness produced when the same lens was submitted to a spinning experiment. Around the same time, Sunderland and O’Neill²⁷ developed a system that simulated accommodation by applying circumferential forces to isolated feline lenses, using an approach similar to that of Pau.²⁴ More recently, Van Alphen and Graebel²⁸ relied on uniaxial loading experiments to estimate the relative force required to stretch the lens, zonules, ciliary body, and choroid. Pierscionek^{29,30} and Glasser and Campbell³¹ have developed experimental setups that stretch lenses to simulate accommodation. Their studies focused on quantifying biometric changes and did not include force measurements. Several other lens-stretching or -spinning systems have been designed recently^{32,33} (Michail H et al. *IOVS* 2004;45:ARVO E-Abstract 1726) to study the optomechanical response of the lens.

The purpose of the present study was to quantify the power- and diameter-force characteristics of primate lenses by using a lens-stretching system (EVAS; Ex Vivo Accommodation System, designed and constructed in-house at the Ophthalmic Biophysics Center) that allows direct measurements of the force required to stretch the crystalline lens to its unaccommodated state while simultaneously measuring the associated change in lens power and equatorial diameter. Results obtained with rhesus monkey, cynomolgus monkey, and human crystalline lenses are presented.

Materials and Methods

Donor Tissues

Experiments were conducted on 35 eyes from 20 rhesus monkeys (*Macaca mulatta*), 48 eyes from 28 cynomolgus monkeys (*Macaca fascicularis*) obtained from the University of Miami

Division of Veterinary Resources through an institutionally approved tissue-sharing protocol and on 20 human eyes from 12 donors, provided by the Florida Lions Eye Bank. All animal experiments adhered to the ARVO Statement for the Use of Animals in Ophthalmic and Vision Research. The human donor eyes were obtained and used in compliance with the guidelines of the Declaration of Helsinki for research involving the use of human tissue. The mean ages of the donors were 3.8 to 11 years for the cynomolgus monkeys, 0.7 to 17 years for the rhesus monkeys, and 8 to 70 years for the humans. For the analysis, the eyes were divided into separate groups based on the age of the donor (Table 1): group 1 included lenses from young donors; group 2, lenses from older prepresbyopic donors; and group 3, presbyopic lenses (human only).

The monkey's eyes were enucleated immediately after euthanatization, wrapped in gauze, placed in a closed jar, and transported in a cooler. The human eyes were received from the eye-bank in closed jars. On arrival at the laboratory, all eyes (human and monkey) were either directly prepared for the stretching experiment or refrigerated at 4°C. Cynomolgus monkey eyes were used between 1 and 72 hours after enucleation (average 26 hours; median: 24 hours). Rhesus monkey eyes were used between 1 and 48 hours after enucleation (average: 11 hours; median: 5.5 hours). Human eyes were used between 1 and 5 days postmortem (average, 2.8 days; median, 3.0 days).

Dissection

The conjunctiva, tenon, episcleral tissue, and extraocular muscle insertions were removed from the globe to expose a clean scleral surface. After the sclera was dried, eight custom-made annular segments of PMMA ("scleral shoes") were bonded with cyanoacrylate adhesive (Duro SuperGlue; Henkel Loctite Corp., Cleveland, OH) onto the anterior scleral surface to form a circumferential band of the globe from 1 mm posterior to the limbus to the equator (Fig. 1). The shoes were custom-designed to match the anatomy of the eyes. They prevent deformation of the globe during dissection and provide attachment points to mount the tissue in the lens stretcher (EVAS). During the bonding process, the shoes are held in place with fixating pins in an alignment ring to ensure proper alignment and centering (Fig. 1).

After bonding the scleral shoes and giving time for the glue to dry to avoid the emission of fumes, the posterior segment of the eye was circumferentially dissected and excess vitreous was cut with Vannas scissors until the vitreous surface was flush with the scleral shell. Special care was taken to leave the anterior vitreous near the lens and ciliary body untouched. The alignment ring and fixation pins holding the globe were then removed, and the tissue was transferred to the tissue chamber and connected to the lens stretcher (EVAS) with the anterior surface facing upward. The cornea was dissected at the level of the limbus, and the iris was carefully pulled off at its root. A diamond knife was used to make full-thickness meridional incisions in the sclera between adjacent shoes, producing eight independent segments (Fig. 1). Special care was taken to avoid damaging the ciliary body, zonule, and lens capsule during the incision process. The tissue chamber was then filled with preservation medium (D-MEM/F-12, D8437, Sigma-Aldrich, St. Louis, MO).

Lens Stretcher

The tissue assembly was connected to the stretcher (EVAS) via eight hooks that fit through a hole in each shoe (Fig. 1). Each hook is attached to a suture wire (6-0 prolene, P/N8304; Ethicon, Somerville, NJ) connected at its other end to a T-shaped bar (T-bar) mounted on a motorized translation stage (MS33-LXB-L200; Servo-Systems Co., Montville, NJ; Fig. 2). Each wire traverses two sets of pulleys designed to produce a radial stretching force equally distributed among the eight wires. The translation stage has a 1- μ m resolution, a 50-mm

maximum displacement, and a translation speed adjustable from 0.02 to 100 mm/s. A load cell with a range of 100 g and resolution of 0.01 g (GSO-100; Transducer Techniques, Temecula, CA) is mounted on the T-bar to record the total load applied on the wires.

Stretching Experiments

The translation stage was programmed to move a total of 2 mm in 0.25 mm steps at a speed of 0.5 mm/s, corresponding to a maximum diameter increase of the outer sclera of 4 mm. The translation stage is stopped for 10 seconds at the end of each step to allow measurements of the lens power and lens and inner ciliary body diameters (described later). At the end of the last step, the translation stage is returned to its initial position at the maximum speed setting. In a few initial experiments, the increment was 0.5 mm ($n = 6$), or the maximum displacement was 1.5 or 1.75 mm ($n = 17$). The load was continuously recorded during a stretching experiment. Successive stretching experiments on the same tissue were used to measure the load-power and -diameter responses separately. Each response was recorded a minimum of three times. An additional set of stretching cycles was performed to obtain slit-lamp images of the lens cross-section during stretching. These experiments helped ensure that stretching did not produce any significant axial movement or tilting of the lens. One preconditioning stretching cycle was performed on each lens before the actual stretching experiments were started, to ensure that there were no problems with the tissue or with the attachment of the shoes and to ensure that stretching was symmetric. Before the preconditioning cycle was started, the position of the motorized translation stage was adjusted by using a joystick control until the wires are under tension. At the end of the preconditioning cycle, the load varied from -0.53 to 1.29 g. This served as the starting point for the actual stretching experiment. The zonules are expected to be under minimal tension at the beginning and end of each stretching cycle. The contribution of gravity to the load applied on the immersed lens was negligible (<0.03 g, assuming a density of 1.096 g/cm³ and a maximum volume of 0.3 mL³⁴). Before each series of experiments, the lens stretcher was calibrated by using a $70\text{-}\mu\text{m}$ -thick annulus with a rectangular cross-section and 17-mm inner diameter and 19-mm outer diameter made of the silicone elastomer, poly(dimethyl siloxane) (PDMS; elastic modulus, 1.7 MPa).

Lens and Inner Ciliary Body Diameter Measurement

At the beginning of each cycle and at the end of each step, digital images of the lens were recorded at $10\times$ magnification. Diffuse retroillumination provided a sharp and high-contrast image of the lens and inner ciliary body circumference (Fig. 3). The lens and inner ciliary body diameters were measured along two perpendicular meridians (horizontal and vertical) in units of pixels by using commercial graphics software (Canvas 9.0; ACD Systems, Miami, FL). A digital image of a drafting ruler with 0.5-mm graduations was recorded before each experiment, to convert from pixel units to length units. The measurement resolution was approximately 0.03 mm in the plane of the tissue. The inner ciliary body diameter was measured from valley to valley of the ciliary processes. The average of the horizontal and vertical diameters was used as a measure of the diameter.

Lens Dioptric Power Measurement

The dioptric power of the lens and its changes during stretching were measured at the beginning of each cycle and after each step by using an optical system based on the Scheiner principle (Fig. 4). The Scheiner system had a measurement range of 12.5 to 125 D. The accuracy and the precision of the measurement were quantified by measuring four immersed lenses of known paraxial power ranging from 15.0 to 39 D. From these experiments, the worst-case accuracy of the lens power measurements was estimated to be within ± 1.5 D.

Data Analysis

The average value and range of the lens diameter, inner ciliary body diameter, and lens power in the unstretched and maximally stretched (4-mm diameter) states and of the load at maximum stretch (4-mm diameter) were calculated for each group. The lens diameter, inner ciliary body diameter, lens power, and load measured at the beginning of each cycle and at each step were plotted to produce diameter- and power-load response curves. Data from the preconditioning cycle were not included in the analysis. For each eye, the average value of the slope of the initial linear portion of the diameter-load (millimeter per gram) and power-load (diopters per gram) plots was calculated. For each species, the average value of these slopes was calculated for each group. An unpaired *t*-test was performed for each parameter to determine whether there were statistically significant differences between the different age groups. For all parameters, values from paired eyes of the same donor were averaged and counted as one sample in the statistical test.

Results

Unstretched Diameter and Power

The values of the lens diameter, inner ciliary body diameter, and lens power for each age group are listed in Table 1. The values of the human lens diameters are similar to values obtained by shadow photography.³⁴

Changes in Diameter, Power, and Load

Representative examples of the change in load, lens diameter, inner ciliary body diameter and lens power during stretching are shown in Figure 5. The figure provides an example of an experiment with a human eye for which the load at the end of the preconditioning experiment was slightly below 0. The first three steps (from 0- to 0.75-mm displacement) were needed to place the lens under tension. The load at 2-mm stretch and the accompanying change in lens diameter, inner ciliary body diameter, and lens power for each group are listed in Table 2.

Diameter- and Power-Load Responses

The diameter- and power-load plots include a linear phase that started once the tissue was under tension and lasted until the diameter increased by 5% to 10% of its original value (Fig. 6). Beyond this linear region, there was an increase in the slope of the load-diameter curve and a decrease in the slope of the power-load curve. The slopes of the linear phase of the load-diameter and power-load responses are shown in Table 3. Data points recorded before the tissue was under tension were not included in the linear regression.

Discussion

Our results show that the lens power continuously decreased as the tension on the zonules increased and vice versa. This result provides further support for the basic principles of the Helmholtz theory of accommodation. The results also indicate that the force necessary to change the shape and power of the prepresbyopic lens increases with age in both human and monkey lenses. Since the change in inner ciliary body diameter per unit force was independent of age in rhesus monkey and human lenses, the age-related changes in the force required to stretch these lenses can only be due to changes in the properties of the lens and potentially the zonules. Our findings therefore agree with the theories of presbyopia that predict that the tension that must be applied to the zonules to change the shape of the lens increases with age.⁶

The response of human and monkey lenses in the EVAS lens stretcher were found to be qualitatively similar. The most notable difference in the age-related changes is that there was a statistically significant decrease in the load-diameter slope of the inner ciliary body in the cynomolgus lenses that was not observed in the rhesus monkey and human lenses. This finding may indicate that there are differences in the age-related changes of the accommodative structures between cynomolgus monkey lenses and human and rhesus monkey lenses. There were larger changes in the diameters and dioptric power with age in the rhesus monkey lenses than in the cynomolgus monkey and human lenses. The larger change in the rhesus monkeys reflects the fact that the younger rhesus monkey group (group 1) was within the fast growth period (average age, 2 years). On average, the lens and inner ciliary body diameters of the older (group 2) rhesus monkey and human lenses were approximately equal. Overall, with the exception of the response of the inner ciliary body of cynomolgus eyes, our data suggest that rhesus and cynomolgus monkey lenses are suitable qualitative models for the response of human lenses in the EVAS system but that the rhesus monkey may be a better anatomic model overall than the cynomolgus monkey.

The goal of the EVAS lens stretcher is to mimic on postmortem lenses the changes in zonular tension produced by relaxation (stretching) and contraction (unstretching) of the ciliary muscle during natural accommodation in vivo. Our lens stretcher combines various elements of previous stretching devices.^{24,26,27,29–31} The dissection technique preserves the sclera–ciliary muscle junction, the anterior vitreous, and the relation of the posterior zonules with the hyaloid membrane.³⁵ Independent of the design of the stretcher, there are some key differences between lens stretching and in vivo accommodation. First, it is necessary to use postmortem lenses because accommodation forces cannot be measured in vivo. Special care must be taken to prevent osmotic swelling of the lens during storage and during the experiment. The tissue protocol was based on our previous in vitro lens preservation and biometry studies,^{34,36} which produced values for the lens curvature, diameter, and thickness that were in good agreement with previously published in vivo and in vitro data. In addition, to avoid using data obtained from swollen lenses, all lenses used in the stretcher were imaged by shadow photography^{34,36} at the end of the experiments. Lenses with capsular separation or an abnormal equatorial diameter-to-sagittal thickness ratio were removed from the study. In total, data on four human lenses and two monkey lenses were discarded. Also, the EVAS stretcher relies on radial forces whereas the movements of the ciliary body during in vivo accommodation include a small anteriorly directed component.¹⁵ The posterior vitreous and choroid, the iris, and the anterior structures, which may influence the in vivo lens response,^{37,38} are not present in the tissue preparation of the EVAS stretcher. The lens and accommodative structures are taken out of the eye, and their natural in vivo orientation as well as the anterior eye dynamics may be altered. Despite these differences, we found that the EVAS stretcher produces biometric and optical changes that are within the range of previously published in vivo data acquired from monkey or human eyes.^{21,39} One limitation of the current version of the biometric component of the EVAS system is that it does not allow measurement of the lens curvature. Curvature measurements are needed to confirm that the system produces changes in lens shape that are similar to the changes occurring in vivo.

The displacement of the translation stage during an experiment produced a 4-mm diameter expansion of the outer scleral shell, in 0.25-mm steps. This total scleral stretch may appear to be excessive, but it was found to produce changes in lens diameter (<0.9 mm) and inner ciliary body diameter (<1.6 mm) that are within the physiological range. Glasser et al.³⁹ found that the maximum lens diameter increase during accommodation induced by brain stimulation was $7.0\% \pm 0.6\%$ on average in four eyes of two rhesus monkeys, 5 and 6 years of age. In our two groups of rhesus monkey eyes we find 7.5% and 5.3% on average for the same parameters (from the data of Tables 1, 2). Croft et al.⁴⁰ found that the average

centripetal movement of the lens and inner ciliary body at maximum accommodation in six rhesus monkey eyes with age ranging from 5.8 to 9.5 years was 0.33 ± 0.02 and 0.40 ± 0.02 mm, respectively, corresponding approximately to a 0.66 ± 0.04 - and 0.80 ± 0.04 -mm change in diameter. In our rhesus monkey eyes, with all eyes pooled, we found an average 0.54 ± 0.17 and 0.81 ± 0.27 mm, respectively, for the same parameters. A similar correspondence was found when comparing the change in human lens and ciliary body diameter produced by stretching with published in vivo data.²¹ No tissue damage was observed in any of the stretching experiments, and successive load-stretch and power-load responses acquired on the same tissue were found to be repeatable. We are therefore confident that the forces produced by the 4-mm diameter scleral stretch did not produce any irreversible tissue changes.

In all eye groups, the change in inner ciliary body diameter induced by stretching was larger than the change in lens diameter, which suggests that stretching produced an elongation of the zonules. The zonular elongation can be estimated by calculating the difference between the change in inner ciliary body diameter and the change in lens diameter, and by dividing the result by two. With this method, the estimated average zonular elongation ranges from 0.075 mm (cynomolgus monkey, group 1) to 0.325 mm (human, Group 2). The average values obtained for the rhesus monkey eyes are slightly larger than the values of the change in circumlental space measured by Croft et al.⁴⁰ during in vivo monkey supramaximum accommodation (0.16 ± 0.08 mm vs. 0.11 ± 0.02 mm).

The force measured by the EVAS stretcher is the force necessary to stretch the entire tissue, including the ciliary body, zonules, and lens. The 4-mm diameter scleral stretch produces a change in the inner ciliary body diameter ranging from 0.62 to 0.91 mm on average (Table 2). The ciliary body is extended on average by a maximum of $(4.00-0.91)/2 = 1.55$ mm to $(4.00-0.62)/2 = 1.69$ mm radially. The measured force includes the force necessary to produce this stretching of the ciliary body. In a series of separate preliminary experiments designed to quantify the contribution of the ciliary body, we stretched the ciliary body of three cynomolgus eyes after dissection of the zonules and removal of the lens. The force needed to stretch the ciliary body alone was found to range between 22% and 29% of the force required to stretch the ciliary body with the lens attached. These values are in good agreement with the results of Van Alphen and Graebel,²⁸ which suggest that the ciliary body accounts for approximately 23% of the total force necessary to stretch human lenses.

There are few reports on the accommodation forces available for comparison with our results. Van Alphen et al.⁴¹ and Lograno and Reibaldi⁴² measured the contraction force of isolated ciliary muscle strips stimulated with various pharmacological agents. Because there were significant differences between the tissue geometry and experimental approach in their study and ours, their results cannot be directly compared with the stretching forces produced by the EVAS system. The results obtained by Sunderland and O'Neill²⁷ in feline lenses may also be inappropriate for comparison, since the mechanism of accommodation, lens growth, and lens shape are not the same in cats and primates. Nevertheless, the diameter of cat lenses was found to increase linearly with the applied load, with an average value of 0.095 mm/g (calculated from the first 12 g of their Table 1 data), which is similar to the results that we obtain on monkey lenses. Fisher's data^{25,26} suggest that a force of approximately 1 mN is necessary per diopter of disaccommodation for human lenses with an average age of 22 years (calculated from the linear portion of the data of Fig. 8 in Fisher²⁶). On average, for the young (8–19 years) and older (38–41 years) human groups, we find a dioptric change of 1.22 and 0.49 D/g, which corresponds to 8 and 20 mN/D, respectively. There is a difference of approximately one order of magnitude between these values and Fisher's results. Others have previously found that Fisher's predictions of the force of contraction of the ciliary muscle are questionable.^{43,44}

In summary, we found that the human and monkey lens diameter increases and the lens power decreases with the applied force in the EVAS lens stretcher, at rates that are age dependent. In the physiologically relevant range, the diameter-and power-load responses can be modeled with a linear response. The change in lens diameter and lens power per unit force decreases with age in both human and monkey lenses. These findings are consistent with the Helmholtz theory of accommodation and with the theories of presbyopia that predict that the amount of tension that must be applied to the zonules to change the shape of the lens increases with age.

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Figure 1.

Preparation and mounting of the tissue. (A) During the preparation, the eight shoes are held upside-down with pins in a custom-made Teflon ring to allow precise alignment of the globe with the shoes. (B) After the shoes bond, the posterior segment is sectioned, the posterior vitreous is carefully removed, and the tissue with anterior vitreous and hyaloid membrane intact is placed on a platform. (C) After removal of the cornea, iris, and vitreous, the sclera is cut between the segments. (D) The resultant sample is mounted in a chamber and connected to the lens stretcher via stainless-steel hooks.

Figure 2.

General view of the lens-stretching system (*top*) and close-up view of the tissue chamber (*bottom*). The sclera is stretched using suture wires that connect the hooks inserted in the shoes to a T-shaped bar (T-bar). The T-bar is displaced by using a motorized translation stage. A system of pulleys transforms the axial displacement of the T-bar into a radial displacement of each lens segment. The optical power is measured using an optical system that relies on the Scheiner principle. The lens diameter is measured on digitized video images acquired using a frame grabber. The bottom of the chamber is a transparent optical-grade window that allows optical measurements as well as retroillumination of the tissue during stretching.

Figure 3.

Composite of two stills from a typical digital recording of a cynomolgus monkey lens acquired during a stretching experiment. The upper half shows the lens in the unstretched state, and the lower half shows the lens at maximum stretch. The lens and inner ciliary body diameter are measured directly from these digital photographs. To produce high-contrast images, the lens is illuminated from below using a custom-made fiber-optic diffuse retroillumination system.

Figure 4.

Measurement of lens power using an optical system relying on the Scheiner principle. Four narrow (~ 0.3 -mm diameter) parallel laser beams are produced by transmitting a collimated 635-nm, low-power red diode laser beam through a custom-built aperture (Scheiner disk) with four 0.3-mm diameter openings separated by 3 mm. A 45° mirror directs the four beams into the vertical direction coaxially with the optical axis of the microscope. The optical system is mounted under the objective of the operation microscope between the two observation channels. The beams are refracted by the lens to a focus. The position of the focus is detected by using a miniature board-level camera (510×492 pixels, 15-BB13; Jameco Electronics, Belmont, CA) mounted on a vertical translation stage under the lens chamber and connected to a video monitor. The camera chip is translated manually until the four spots corresponding to the four beams are superimposed. An optical model was developed to derive the relation between the focus position and the effective power of the immersed lens.

Figure 5.

Typical changes in load (*left*) and lens and inner ciliary body diameter (*right*) as a function of the displacement of the translation stage for a Rhesus monkey (*top*), cynomolgus monkey (*middle*), and a human (*bottom*) eye. Each curve corresponds to the average of three or five successive stretching cycles. Error bars show the SD.

Figure 6.

Typical load-diameter (*left*) and power-load characteristics (*right*) of a Rhesus monkey (*top*), cynomolgus monkey (*middle*), and human (*bottom*) eye. The initial portion of the response was fit with a linear regression line. The responses were characterized by the slope of the regression line.

Table 1

Biometric and Optical Parameters of the Unstretched Lenses

	Cynomolgus			Rhesus			Human		
	Group 1	Group 2		Group 1	Group 2		Group 1	Group 2	Group 3
<i>n</i>	31	17		23	12		8	7	5
Age range (y)	3.8–7.5	9.0–11.0		0.7–4.3	7.1–17.0		8–19	38–41	55–70
Mean (SD)	5.6 (1.4)	9.4 (0.7)		2.0 (1.1)	10.5 (3.8)		14.0 (3.9)	39.5 (1.3)	62.7 (7.5)
Lens diameter (mm)									
<i>n</i>	31	17		20	12		8	7	5
Range	7.0–8.0	7.2–7.8		7.3–8.4	8.5–9.6		8.4–9.3	8.9–9.3	8.7–9.7
Mean (SD)	7.6 (0.3)	7.5 (0.2)		7.9 (0.4)	8.8* (0.4)		8.8 (0.4)	9.0 (0.2)	9.2 (0.5)
Inner ciliary body diameter (mm)									
<i>n</i>	31	17		20	12		8	7	5
Range	8.5–9.6	8.6–9.4		9.0–10.0	9.8–12.5		11.1–11.7	10.9–11.5	10.4–11.3
Mean (SD)	9.1 (0.3)	8.9 (0.2)		9.3 (0.3)	10.6* (1.0)		11.4 (0.2)	11.2 (0.3)	11.0 (0.5)
Power (D)									
<i>n</i>	23	17		20	12		6	6	3
Range	42.8–55.5	39.6–53.2		45.8–54.3	32.9–40.7		32.6–40.0	22.2–26.0	20.4–22.8
Mean (SD)	47.7 (3.6)	46.7 (5.2)		50.5 (3.2)	36.7* (3.1)		35.6 (3.6)	24.5† (1.7)	21.6 (1.7)

The biometric or optical parameters for which there was a statistically significant difference between age groups are shown. The lenses were divided into separate age groups (groups 1–3). Group 1, lenses from young donors; group 2, from older pre-presbyopic donors; group 3 (for human lenses only), from presbyopic donors. Power measurements with the Scheiner system were performed in 23 of 31 lenses in the group 1 cynomolgus lenses and in 20 of 23 lenses in the group 1 rhesus lenses. Diameter measurements were performed on 20 of the 23 group 1 rhesus lenses. Power data are not available on 3 of 20 human lenses due to technical problems.

* $P < 0.001$.

† $P = 0.0015$.

Table 2
Load and Accompanying Changes in Lens Biometry and Power at Maximum Displacement

	Cynomolgus			Rhesus			Human		
	Group 1	Group 2		Group 1	Group 2		Group 1	Group 2	Group 3
Load at 2 mm stretch (g)									
<i>n</i>	23	16		14	10		8	7	5
Range	2.2–8.0	2.9–19.7		4.5–20.1	5.3–9.0		4.1–13.8	4.2–10.5	8.5–23.0
Mean (SD)	6.1 (1.7)	8.2 (5.6)		8.4 (5.0)	7.2 (1.4)		8.3 (3.8)	8.2 (2.9)	13.6 (8.1)
Lens diameter change (mm)									
<i>n</i>	24	16		10	10		8	7	5
Range	0.21–0.79	0.25–0.88		0.37–0.89	0.36–0.73		0.21–0.63	0.14–0.31	0.24–0.29
Mean (SD)	0.53 (0.15)	0.42 (0.18)		0.59 (0.18)	0.47 (0.15)		0.34 (0.18)	0.25 (0.08)	0.26 (0.02)
Inner ciliary body diameter change (mm)									
<i>n</i>	25	16		10	10		6	7	5
Range	0.28–1.10	0.38–1.32		0.49–1.35	0.66–0.98		0.40–1.56	0.53–1.20	0.58–0.73
Mean (SD)	0.68 (0.26)	0.63 (0.28)		0.84 (0.35)	0.80 (0.12)		0.76 (0.48)	0.91 (0.28)	0.64 (0.08)
Lens power change (D)									
<i>n</i>	19	16		14	10		6	6	3
Range	7.3–21.6	10.8–20.3		10.1–21.6	10.1–13.8		6.0–12.1	0.8–4.0	–0.4–0.2
Mean (SD)	16.5 (4.0)	15.3 (3.6)		14.6 (3.8)	12.0 (1.6)		8.4 (2.6)	1.9* (1.5)	–0.3 (N/A)

The biometric or optical parameters for which there was a statistically significant difference between age groups are shown. A 2-mm displacement corresponds to A 4-mm diameter scleral stretch. In some groups, the number of samples is smaller than in Table 1 because not all lenses were stretched by 2 mm (see the Methods section) or because the experiment could not be completed due to technical problems.

* $P=0.005$

Table 3

Load Diameters and Power Load Slopes

	Cynomolgus			Rhesus			Human		
	Group 1	Group 2		Group 1	Group 2		Group 1	Group 2	Group 3
Lens diameter-load slope (mm/g)									
<i>n</i>	25	16		19	12		7	7	4
Range	0.079–0.113	0.048–0.090		0.076–0.144	0.044–0.083		0.049–0.101	0.027–0.043	0.029–0.063
Mean (SD)	0.094 (0.011)	0.069* (0.013)		0.109 (0.021)	0.067* (0.016)		0.069 (0.020)	0.036 [†] (0.007)	0.045 (0.017)
Inner ciliary body diameter-load slope (mm/g)									
<i>n</i>	30	16		20	11		6	7	3
Range	0.082–0.222	0.077–0.125		0.098–0.190	0.089–0.199		0.097–0.161	0.112–0.159	0.093–0.139
Mean (SD)	0.131 (0.031)	0.097 [‡] (0.018)		0.142 (0.031)	0.133 (0.038)		0.133 (0.031)	0.129 (0.021)	0.115 (0.023)
Lens power-load slope (D/g)									
<i>n</i>	12	8		19	12		6	6	5
Range	–4.34–3.00	–3.46–1.69		–3.66–1.69	–3.40–1.46		–1.64–0.52	–1.33–0.11	0.02–0.11
Mean (SD)	–3.73 (0.49)	–2.46* (0.56)		–2.83 (0.56)	–2.16 (0.74)		–1.22 (0.50)	–0.49 (0.58)	0.05 [§] (0.05)

The slopes for which there was a statistically significant difference between age groups are shown. For some groups the number of samples is smaller than in Table 1 because the experiment could not be completed due to technical problems.

* $P < 0.001$.

[†] $P = 0.006$.

[‡] $P = 0.02$.

[§] $P = 0.026$, group 3 vs group 1.