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Optical power of the isolated human crystalline lens

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Abstract

Purpose—To characterize the age-dependence of isolated human crystalline lens power and quantify the contributions of the lens surfaces and refractive index gradient.

Methods—Experiments were performed on a total of 100 eyes from 73 donors (average post-mortem: 2.8 ± 1.6 days) with an age range of 6 to 94 years. Lens power was measured with a modified commercial lensmeter or with an optical system based on the Scheiner principle. The radius of curvature and asphericity of the isolated lens surfaces were measured by shadowphotography. For each lens the contributions of the surfaces and the refractive index gradient to the measured lens power were calculated using an optical ray tracing software. The age dependency of these refractive powers was assessed.

Results—The total refractive power and surface refractive power both showed a biphasic age dependency. The total power decreased at a rate of -0.41D/year between age 6 and 58.1, and increased at a rate of 0.33D/year between age 58.1 and 82. The surface contribution decreased at a rate of -0.13D/year between age 6 and 55.2, and increased at a rate of 0.04D/year between age 55.2 and 94. The relative contribution of the surfaces increased by $0.17\%/year$. The equivalent refractive index also showed a biphasic age dependency with a decrease at a rate of $-3.9 \times 10^{-4}/year$ from age 6 to 60.4 followed by a plateau.

Conclusions—The lens power decreases with age due mainly to a decrease in the contribution of the gradient. The use of a constant equivalent refractive index value to calculate lens power using the lens maker formula will underestimate the power of young lenses and overestimate the power of older lenses.

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Introduction

The optical power of the crystalline lens is determined by the surface curvatures, the refractive index differences at the aqueous-lens and lens-vitreous interfaces, and the refractive index gradient distribution within the lens¹. Studying the optical properties of the lens (i.e. optical power, refractive index distribution and the surface refractive contributions) *in vivo* is difficult because of the position of the lens behind the cornea and pupil, as well as the distortions of the posterior lens surface caused by the lens refractive index gradient. Two approaches have been used to measure the lens power *in vivo*. In the first approach the curvature of the lens surfaces and lens thickness are measured using phakometry and ultrasonic or optical biometry. The lens power is then calculated assuming an equivalent uniform refractive index (typically around 1.42)^{2–3}. In the second approach the lens power is calculated from measurements of axial eye length, anterior chamber depth, corneal power and refractive state of the eye. These parameters are input into an eye model to calculate the power required for the lens to produce an optical system that matches the measurements^{3–6}. Both techniques derive the lens power from measurements of other ocular parameters. Even though recent studies have cross-validated *in vivo* lens biometry techniques^{7,8} some assumptions are required either for the refractive index of the lens, or the parameters of the eye model.

Direct measurements of the lens power (or focal length) can be obtained on *in vitro* lenses^{9–15}. Comparison of *in vivo*¹⁶ and *in vitro*¹⁷ lens shape and power suggest that the isolated lens, which is free of zonular tension, assumes a shape and power similar to those of the maximally accommodated lens *in vivo*. Measuring the shape and power of isolated lenses can therefore provide information on the optical changes occurring in the maximally accommodated lens with age, as long as special care is taken to avoid swelling and other changes caused by tissue storage and handling¹⁸. The refractive power of the isolated lens, free of zonular tension, is also an important input parameter for optical-mechanical finite element models of the lens and accommodation currently under development^{19–21} (Chen J, et al. *IOVS* 2006; 47: ARVO E-Abstract 5862).

The isolated lens power has been shown to decrease with age^{12, 13, 15, 16}. A recent study suggests that this decrease is primarily due to age-related changes in the axial profile of the refractive index distribution¹⁶. The same study suggests that the contribution of the lens surfaces to the total lens power is minor compared to the contribution of the index gradient. On the other hand, Glasser and Campbell¹³ found a significant correlation between *in vitro* lens power and lens surface curvatures, and found no age dependence of the *in vitro* lens equivalent refractive index. Their findings suggest that there are no significant age related changes in the contribution of the index gradient to the lens power. A better understanding of the respective contributions from the surfaces and the refractive index gradient to the *in vitro* lens power with age will help characterize the optical changes associated with presbyopia.

The purpose of the present study was to characterize the age-dependence of the isolated human crystalline lens power and quantify the contributions of the lens surfaces and refractive index gradient.

Materials and Methods

General Description

The power and shape of *in vitro* human lenses were measured and used to calculate the refractive contributions of the lens surfaces and the gradient refractive index. The lens power data were obtained from two different experiments. In one experiment, the dioptric power of isolated lens was measured with a modified commercially available lensmeter (Lens Analyzer 350, Zeiss Humphrey, Dublin CA) (Denham DB, et al. *IOVS* 2003; 44: ARVO E-Abstract

244). In the other experiment, the dioptric power of lenses maintained in their accommodative framework (with intact zonules, ciliary body and segmented scleral rim) and mounted in the testing cell of a lens stretching system (Ex-Vivo Accommodation Simulator – EVAS) under no applied tension was measured with a custom designed optical system based on the Scheiner principle²². Both systems were calibrated on plano-convex glass lenses in the power range of 10 to 45D. The measurement error ranged from -1.8 to 2.9D (average 1.5D) for the Scheiner system and from 0.32 to 1.47D (average 0.61D) for the commercial lensmeter. To compare the two systems, a Bland-Altman analysis was performed on 14 pairs of crystalline lenses (age: 26 to 82 years; average age: 54 years) where one eye was measured with the commercial lensmeter and the contralateral eye was measured with the Scheiner system. The mean difference between the two techniques was 1.1 ± 2.2 D with the commercial lensmeter providing the higher average value and no age dependence ($p=0.374$). This difference is within the expected accuracy and variability of the measurements on *in vitro* crystalline lenses.

Donor Tissue

All eyes were obtained and used in compliance with the guidelines of the Declaration of Helsinki for research involving the use of human tissue. Experiments were performed on a total of 100 eyes from 73 phakic donors. The donor globes arrived in sealed, Styrofoam containers filled with ice and used between 1 and 5 days post-mortem (average: 2.8 ± 1.6 days). The donor ages ranged from 6 to 94 years. Any lens with visible damage or with an equatorial diameter to sagittal thickness aspect ratio less than 2.0 was considered swollen and excluded from the analysis¹⁸. In total 24 out of the 100 lenses were discarded due to swelling from storage conditions or manipulation during the experiment.

Isolated lens power measurements with a modified commercial lensmeter

For lens power measurements on isolated lenses ($n=52$), tissue dissection was performed under an operation microscope by an ophthalmic surgeon. First, for stability, a fixed ring was attached to the sclera around the anterior chamber with cyanoacrylate glue. The posterior pole was then sectioned and the cornea and iris were removed. The lens was then extracted by carefully cutting the zonules and adherent vitreous using Vannas scissors. The isolated lens was immediately immersed in a vial filled with preservation media (DMEM/F-12, D8437, Sigma, St. Louis, MO)¹⁸.

Dioptric power measurements of the isolated lens were performed with a custom modified commercial lensmeter (Lens Analyzer 350, Zeiss Humphrey, Dublin, CA) (Figure 1a). This system delivers four parallel beams arranged in a 4.24mm square pattern on the test lens. The lens refracts the incident beams and brings them to a focus. An opto-electronic system measures the focus position and calculates lens power. For the measurements, the crystalline lenses were positioned in a custom designed testing cell filled with preservation media. The testing cell was designed to offset the optical power of the test lens by -7 Diopters to increase the measurement range of the lensmeter. All power measurements on isolated lenses took less than 10 minutes.

Mounted lens power measurements with a Scheiner lensmeter

For the lenses measured in the lens stretcher ($n=51$) the dissection was performed under an operation microscope by an ophthalmic surgeon. The tissue preparation followed the same protocol as described in a previous study²². In this technique a band of 8 independent scleral shoes were bonded with cyanoacrylate adhesive (Duro SuperGlue, Henkel Loctite Corp, Cleveland, OH) onto the anterior sclera surface from the limbus to the equator. The band of sclera prevents deformation of the globe during dissection and the shoes provide attachment for the lens stretcher device. The posterior pole was then sectioned and the partially dissected tissue was transferred to the lens stretching chamber, which was filled with the DMEM

preservation medium. Scleral incisions were made between adjacent shoes and the cornea and iris were removed to produce 8 individual segments for stretching. Only the power measured with no tension applied (unstretched state) was used for the present study.

The optical power measurements of the lens while mounted in the lens stretcher system were performed with a custom designed optical system based on the Scheiner principle (Figure 1b). This system places four parallel laser beams (arranged in a 3mm square pattern) incident on the lens. The parallel beams are focused by the lens. A camera mounted on a vertical translation stage is positioned at the location of convergence of the four incident beams. This location corresponds to the focal plane of the lens. The dioptric power of the lens immersed in the testing chamber is calculated using a formula derived from a paraxial optical model of the system. On average the lens was immersed in DMEM for approximately 30 minutes until the power measurements were completed.

When the tissue was not needed for other experiments (n=11 out of 51 lenses), the lens was extracted by a surgeon at the end of the stretching experiment by carefully cutting the zonules and adherent vitreous using Vannas scissors. The isolated lens was immediately immersed in a DMEM-filled vial to prevent swelling^{18, 23}. A separate preliminary study showed that the shape and power of the unstretched crystalline lens are not affected by stretching experiments (Parel JM, et al. IOVS 2002; 43: ARVO E-Abstract 406).

Lens shape measurements

Upon completion of the lens power measurements the shape of the isolated lens was measured using an optical comparator (Topcon BP-30S, Tokyo, Japan)²⁴. Digital photographs of sagittal views were recorded at 20× magnification using a protocol described previously²³. During these measurements the lens was positioned in an immersion chamber filled with DMEM. The lens was supported by a meshwork of 10–0 nylon monofilament sutures. The central lens thicknesses (t), as well as the anterior and posterior surface profiles, were obtained from the shadowgraph images (Figure 2). The surface profiles were fit with conic functions over the central 6mm zone to calculate the radii of curvatures (R) and asphericities (Q)^{23, 25}. In total the lenses were immersed in preservation media on average for 20–30 minutes for isolated lenses and 1–2 hours for lenses measured in the lens stretcher.

Contributions of surface refraction to total refractive power of isolated lenses

The refractive contributions of the lens aspheric surfaces were quantified by calculating the power of each measured lens assuming a uniform refractive index equal to the cortex refractive index. Based on the data of Jones et al using MRI, the refractive index of the cortex was assumed to be $n=1.3709$ ¹⁶ independent of age. The calculations were performed with a commercial optical ray tracing software (OSLO LT, Lambda Research, Littleton, MA). The ray tracing simulation incorporated isolated crystalline lens thickness, surface curvatures and asphericity measurements. The refractive index of the surrounding medium (DMEM) was assumed to be equal to the refractive index of water ($n=1.332$ at 635nm). To simulate the experimental conditions, a ray with a height of 1.5mm (Scheiner) or 2.24mm (commercial lensmeter) was traced entering the lens. The position of the focal plane was determined by the location of the intersection point of the ray with the optical axis.

Equivalent Refractive index calculations

The equivalent refractive index (N_{eq}) is defined as the uniform refractive index value that is required inside the lens to provide a power equal to the actual crystalline lens power if all other parameters (R, Q and t) remain equal. To calculate the equivalent refractive index for each lens an optical ray tracing simulation of the power measurement setups, incorporating the chamber and the measured lens biometric data (lens radii of curvature, asphericities and thickness) was

performed with a commercial optical ray tracing software (OSLO LT). The refractive index of the simulated lens was adjusted iteratively until the power of the lens corresponded to the measured power.

The technique to calculate the equivalent refractive index was calibrated using a 61D plano-convex glass (SF5, $n=1.6683$) lens, a 36.7D plano-convex glass (BK7, $n=1.5150$) lens and a 19.2D plano-convex glass (BK7) lens. The shape of the lenses was measured from shadowgraph images following the same procedure as that of the isolated lenses. The refractive index calculated from the measured equivalent power, thickness, diameter and radius of curvature, was within 0.006 to 0.013 of the specified value at 635nm. An analysis of the sources of error shows that the error in the value of the refractive index comes mainly from the error in the power measurements.

Data and Statistical Analysis

There was a strong correlation between the power and the refractive index of the left and right eye from the same donor. Therefore to prevent that values obtained from paired eyes skew the age dependence, the values from left and right eyes of each donor were averaged. The average values were then used as a single point for age dependence calculations. The total power, surface contribution, and equivalent refractive index were plotted as a function of age to characterize the age dependence. A single or piecewise linear regression was calculated for each data plot to determine if there was an age dependence. A p-value of 0.05 was set as the condition for statistical significance. For each donor the OD and OS average value for each parameter was used as a single data point to characterize age dependence.

Results

Over the age range of the sample set (6–94 years), the age dependence of both total and calculated surface powers as well as the anterior and posterior surface curvatures appear to be biphasic with a change in trend occurring approximately within the period of onset of presbyopia (40–60 years) (Figure 3, Figure 4a and Table 1). Given these apparently biphasic trends, the age dependences for total power, calculated surface power, relative surface contribution, equivalent refractive index and surface curvatures were analyzed with a bi-linear model (Appendix A). Applying a non-linear regression technique to this model, a break-point age was identified for each parameter. The break-point age divides the data set into two age groups: data points whose age lies below the break-point age (which will be referred to as the “lower age group” throughout this paper), and data points whose age lies above the break-point age (the “upper age group”). The non-linear regression outcome on the bi-linear model is summarized in Table 2. Predicted anterior and posterior radii of curvatures from the bi-linear model are plotted in Figure 3. Predicted total and calculated surface power from the bi-linear model are plotted in Figure 4.

For both total and surface power and for anterior and posterior radius of curvature, the bi-linear model showed very high levels of statistical significance ($p < 10^{-5}$) and low standard error of estimates (3.76D for total and 1.5D for surface power; 2.16mm for anterior and 0.63mm for posterior radius). In the lower age group lenses, there is a decrease in total and surface power; the total power decreasing at a faster rate (-0.41D/year) than the surface power (-0.13D/year). In the upper age group lenses, total lens power increased at a greater rate ($+0.33\text{D/year}$) than surface power ($+0.04\text{D/year}$). Anterior and posterior (absolute value) radii of curvature increased in the lower age range at a rate of 0.157mm/year and 0.058mm/year and decreased in the upper age range at a rate of 0.055mm/year and 0.027mm/year.

There was no statistically significant ($p=0.74$) break point age for the relative surface refractive contributions (Table 2). There was however a statistically significant increase in the relative

refractive power contribution from the lens surfaces with age (Figure 5 and Table 1). The surfaces provide from 41% (at 6 years) to 55.1% (at 82 years) of the total refractive power of the isolated lens.

There was a statistically significant ($p < 0.028$) break-point age of 60.4 years for the calculated lens equivalent refractive index (Figure 6 and Table 2). In the lower age group the equivalent refractive index decreased at a rate of -3.88×10^{-4} /year. At the break point age the equivalent refractive index value was 1.4085. In the upper age range the equivalent refractive index remains relatively constant with age.

The equations of the linear regressions are given in Table 1.

Discussion

Several previous studies have shown that the isolated lens power decreases with age. Jones *et al* (2005)¹⁶ found that the decrease in lens power followed a linear trend over the entire age range of their samples (7 to 82 years). Our study included a larger number of lenses, particularly in the presbyopic age range. With the larger sample size, we find that the age related changes in total power of the *in vitro* lens is biphasic, with the transition occurring around or towards the end of the age range of presbyopia onset. The power initially decreases approximately linearly. After this initial linear decrease there is a trend for both the total power and the calculated surface power to increase in the older presbyopic lenses. This same trend has been reported by Glasser and Campbell (1999)¹³ for lenses older than 65 years in a study on 19 isolated human lenses with ages ranging from 5 to 96 years of age (Figure 7A). Glasser and Campbell¹³ attribute this increase to the inclusion of lenses with early stages of cataract. In our study 11 out of 40 lenses over the age range 58–94 had some stage of cataract, as evaluated subjectively by grading the color and structure of the lenses in the shadowgraph images. On average the power and the surface contribution of the cataractous lenses was higher even though the difference was not statistically significant. This result confirms that the increase in power in older lenses may be due to the effect of cataract formation.

Even though the general trends in our study were similar to those observed by Jones *et al* (2005)¹⁵ the isolated lens power values obtained by Jones *et al* (2005)¹⁶ were significantly higher than our values (by 6.8D for a 20 year old lens and by 8.3D for a 60 year old lens) as well as the values of Glasser and Campbell (1999)¹³ (by 7.8 for a 20 year old lens and by 8.3D for a 60 year old lens) and of Schachar (2004)¹⁴. Jones *et al.* (2005)¹⁶ recognized that their lens power is higher than expected. They attributed this difference to a potential increase in lens power beyond maximal accommodation when the lens is removed from the eye. The fact that our results are in excellent agreement with the measurements of Glasser and Campbell, as well as with *in vivo* measurements^{2, 26–31}, suggests that the values obtained by Jones *et al.* (2005)¹⁶ overestimated the power of the isolated lens. Our findings show that removing zonular tension does not increase the power of the lens significantly compared to the maximally accommodated *in situ* lens.

According to our results, the dioptric contribution of the lens surfaces to the total lens power decreases only slightly with age. The decrease of the *in vitro* lens power with age is therefore, due mainly to changes of the internal structure of the lens including the cortex. These findings confirm the results of Jones *et al.* (2005)¹⁶. In absolute values the contribution of the surfaces in the present study is similar to their value (around 10D). The relative contribution (in percent) of the surfaces in our study is higher, as expected since their values seem to overestimate total lens power. We find that the surface refractive contribution increases with age from 26 to 55% of the total lens power (Table 1).

The contribution of the gradient refractive index of the lens to the total lens power is often quantified using the concept of an equivalent refractive index. In a longitudinal study of school age children, Mutti *et al.* (1995)⁶ showed that the equivalent refractive index between 5 and 16 years of age is approximately constant with an average value of 1.427. Dubbelman and Van der Heijde (2001)³² found that the equivalent refractive index decreases with age in the adult eye with values ranging from approximately 1.441 to 1.418 in the age range of 16 to 65 years. Glasser and Campbell calculated the equivalent refractive index of isolated lenses from ray tracing experiments^{12, 13}. They did not find a change with age. Their mean equivalent refractive index value was 1.4257 ± 0.0163 ¹³. Glasser and Campbell may not have been able to find an age dependence in equivalent refractive index due to their smaller sample size ($n=19$). We find that the equivalent refractive index decreases with age from a maximum of 1.437 for a 14 year old lens to a minimum of 1.396 for an 82 year old lens.

Our results show the same trend, with a similar decay rate as the study of Dubbelman and Van der Heijde (2001)³² (Figure 7B), but our values are slightly lower. The difference could be due to the fact that we are measuring the *ex-vivo* lens, which is expected to correspond to the maximally accommodated state, whereas Dubbelman and Van der Heijde (2001)³² measured the refractive index of the relaxed unaccommodated lens *in-vivo*. However, Dubbelman *et al.* (2005)¹⁷ find that the equivalent refractive index slightly increases with accommodation. If the difference is due to a change in accommodative state then our values should be higher. Most likely the difference is due to differences in methodology. In our study we calculated the refractive index of the lens from direct measurements of the lens power and high resolution measurements of the lens shape including lens asphericity. Even though *in vitro* lenses allow direct measurement, they are subject to alterations in lens shape caused storage and handling. On the other hand, accurate *in-vivo* measurements of the lens equivalent refractive index are difficult to obtain because of the uncertainty of the shape of the posterior lens surface which is imaged through the internal structure of the lens^{17, 32}. Despite these potential sources of error in both *in vitro* and *in vivo* measurements there is good agreement between our *ex-vivo* results and the *in-vivo* measurements of Dubbelman and Van der Heijde³² (2001).

Our results demonstrate a biphasic age dependent trend for both anterior and posterior radii of curvatures, which confirms the findings of two previous studies^{9,13}. Since power and radii of curvature exhibit a biphasic age-dependency, we expect to find the same trend in the equivalent refractive index and in the relative contribution of the surfaces to the lens power. There is a statistical significant breakpoint in the equivalent refractive index, but not in the relative contribution of the surface power. A larger number of samples is needed in the upper age range (above 55 years) to demonstrate the biphasic trend of these two parameters with a higher confidence level.

Like Glasser and Campbell (1998 and 1999)^{12, 13}, we find that the decrease of *in vitro* lens power matches well with the decrease of *in vivo* accommodation amplitude³⁴. This result is expected, since it is assumed that the power of the lens when accommodation is relaxed remains approximately constant with age in the adult^{1, 28} and the isolated lens is expected to correspond approximately to the maximally accommodated state. The decrease in accommodation amplitude in the adult must therefore correspond directly to a decrease in the power of the maximally accommodated lens (Figure 8). The relation between *in vitro* lens power and *in vivo* accommodation amplitude may not hold in children because there is evidence that the unaccommodated lens power decreases with age in children^{27, 30, 31, 35}.

In summary our measurements are consistent with the findings of Dubbelman *et al.*^{17, 32} and of Jones *et al.*¹⁶ that there is a decrease in the equivalent refractive index with age or, in other words the contribution of the gradient refractive index distribution decreases. One of the implications of the decrease in equivalent refractive index with age is that the use of fixed

equivalent refractive index value to calculate lens power using the lens maker formula will underestimate the power of young lenses and overestimate the power of old lenses. Furthermore an error may be induced in central lens thickness and posterior radius of curvature calculations when a fixed equivalent refractive index is used^{3, 36} during analysis of *in vivo* phakometry or Scheimpflug images. Our measurements confirm that the power and shape of the *in vitro* lens correspond to that of the *in vivo* maximally accommodated lens.

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APPENDIX A

Fitting of a Bi-Linear Model to Apparently Biphasic Data Set using an Iterative Computational (Non-Linear Regression) Technique

Given the apparent biphasic nature of the relationships for measured total power and calculated surface power versus age, we tested the data set against a bi-linear model. Mathematically, the bi-linear model can be represented (given here in parametric, column matrix form as it best illustrates the concept) by:

$$\begin{bmatrix} X \\ Y' \end{bmatrix} = \begin{cases} \begin{bmatrix} X_{bp} \\ Y_{bp} \end{bmatrix} + k_L \cdot \begin{bmatrix} L_x \\ L_y \end{bmatrix} & \text{for } X \leq X_{bp} \\ \begin{bmatrix} X_{bp} \\ Y_{bp} \end{bmatrix} + k_U \cdot \begin{bmatrix} U_x \\ U_y \end{bmatrix} & \text{for } X > X_{bp} \end{cases} \quad (1)$$

where

X	is the x value (age)
Y'	is the predicted y-value (total power, calculated surface power)
X_{bp}	is the break-point age
Y_{bp}	is the predicted y-value at the break-point age
k_L, k_U	are scalar parameters
L_x, L_y	are direction cosines for points with age below break-point age
U_x, U_y	are direction cosines for points with age above break-point age

It can be seen from Equation (1) that conceptually, the bi-linear model defines a 'break-point' age value that divides the data set into two sub-sets; one sub-set of data points whose age values lie below the break-point age, a second sub-set with age values lying above the break-point age. A line of best fit is then found for each of the two sub-sets with the additional constraint that the two lines intersect at the break-point (X_{bp}, Y_{bp}). Values for the best fitting (X_{bp}, Y_{bp}), (L_x, L_y) and (U_x, U_y) are found by the least-squares method (minimizing the total sum of squares of residuals for both sub-set of data combined). This was achieved computationally using the Solver tool in Microsoft Excel (version 2003). Results are given in Table 2.

In consideration of the degrees of freedom in this regression model, although equation (1) is written with six variables, effectively, the model involves the fitting of two lines, each with its independent slope and intercept. Once these two lines are defined, their intersection defines, without further degrees of freedom, the break-point. Hence, a total of four degrees of freedom are assignable to this model and used for the F-test analyses in Table 2.

FIGURE 1.

A. Isolated lens refractive power measurement with a commercial lensmeter (Lens Analyzer 350, Zeiss Humphrey, Dublin CA). The isolated lens testing cell consisted of a -20D plano-concave lens (PLCC 01LPK021, $f = -50\text{mm}$, $\varnothing = 40\text{mm}$, $t_c = 2\text{mm}$, Melles Griot, CA.) in a custom designed lens chamber, filled with DMEM and covered with a glass cover slip (to remove the surface meniscus), giving a total combined optical power of -7D. **B.** *In vitro* lens power measurement with a Scheiner system (Manns 2007). Four parallel beams are directed by a 45° mirror into the vertical direction coaxially with the optical axis of the *in vitro* lens. A CCD chip mounted on a vertical position is used to visualize and record the location of the focus. In both cases the anterior of the surface of the lens is up.

FIGURE 2.

A sample sagittal (side view) shadowgraph image of a 53 year old isolated lens 2 days postmortem with the anterior and posterior surface profiles overlaid. The surface profiles were fit with conic sections to calculate the anterior and posterior radii of curvature (R_a and R_p) and asphericities (Q_a and Q_p). The thickness was measured directly from the images using the ruler as a calibration. The image shows the crosshairs of the optical comparator and the supporting suture mesh of the immersion cell. During the measurement, the camera is inverted. In this image the lens was photographed with the anterior surface resting on the suture wires.

FIGURE 3.

In vitro lens anterior (n=47) and posterior radii of curvature (n=48). The equations for the regression lines are given in Table 1.

FIGURE 4.

In vitro lens refractive power (n=65) and calculated surface refractive power (n=48). Surface power was calculated from surface curvature measurements obtained during isolated lens biometry. The equations for the regression lines are given in Table 1.

FIGURE 5.

The relative surface refractive contributions to the total *in vitro* lens refractive power as a function of age (n=41). The equation for the regression line is given in Table 1.

FIGURE 6.

The age dependence of the calculated equivalent refractive index of *in vitro* human lenses (n=40). The equation for the regression line is given in Table 1.

FIGURE 7.

Comparison of our results with previously published data on the age dependence of the lens power (A.) and the refractive index (B.).

FIGURE 8.

Calculated maximum possible accommodative amplitude of each measured lens power data point (n=65). The age dependent data for the *in vivo* amplitude of accommodation from Duane 1912 (corrected to place the reference plane at the anterior corneal surface instead of the spectacle plane) is superimposed on the graph. There is good agreement between the age dependent loss of accommodation and the maximum amplitude of accommodation calculated from the isolated lens power.

TABLE 1
Age dependence of in vitro human lens optical properties

In vitro human lens age dependent linear regression equations of the measured optical properties. The values from the left and right eyes of each donor were averaged and used as a single point for the age dependence linear regressions.

Age Range (years)	Total age range	Lower age range	Upper age range
Measured Equivalent Power (D)		41.59 – 0.41·Age (6 to 58.1 years) n=39	–1.34 + 0.33·Age (58.1 to 82 years) n=26
Calculated Surface Power (D)		14.09 – 0.13·Age (6 to 55.2 years) n=33	4.58 + 0.04·Age (55.2 to 94 years) n=15
* Relative Surface Refractive Contribution (%)	30.67 + 0.17·Age p=0.0031 n=41	35.44 + 1.8×10 ⁻³ ·Age (6 to 57 years) n=29	7.37 + 0.49·Age (57 to 82 years) n=12
Equivalent Refractive Index		1.4320 – 3.90×10 ⁻⁴ ·Age (6 to 60.4 years) n=30	1.4096 – 1.82×10 ⁻⁵ ·Age (60.4 to 82 years) n=10
Anterior Radius of Curvature (mm)		4.46 + 0.14·Age (6 to 50 years) n=26	13.83 – 0.05·Age (50 to 94 years) n=21
Posterior Radius of Curvature (mm)		–3.47 – 0.06·Age (6 to 57 years) n=34	–7.97 + 0.02·Age (57 to 94 years) n=14

All ages in years

* The break point analysis was not statistically significant

Table 2

Summary of statistics for fitting of bi-linear model against total power and surface power versus age (see Appendix A for details).

Measured Total Power					Calculated Surface Power				
Group	Break-Point Age (year)	Break-Point Power (D)	Direction Cosines (Age, Power)		Break-Point Age (year)	Break-Point Power (D)	Direction Cosines (Age, Power)		
Lower Age Upper Age	58.07	17.60	-0.9242 0.9510	0.3819 0.3092	55.26	7.09	-0.9956 0.9991	0.0941 0.0414	
ANOVA	Df	SS	MS	F	Sign	SS	MS	F	Sign
Model	4	1502.89	375.72	26.61	1.2E-12	37.6	9.41	11.14	5.5E-06
Residual	60	847.07	14.12	(SSE=3.76)		30.4	0.9	(SSE=1.5)	
Total	64	2349.7			40	68.1			

Relative Surface Refractive Contribution					Equivalent Refractive Index				
Group	Break-Point Age (year)	Break-Point Value (%)	Direction Cosines (Age, %)		Break-Point Age (year)	Break-Point Refractive Index	Direction Cosines (Age, Neq)		
Lower Age Upper Age	57	35.54	-1.00 0.8932	-0.0018 0.4414	60.4	1.4085	-1.00 1.00	0.0004 0.0000	
ANOVA	Df	SS	MS	F	Sign	SS	MS	F	Sign
Model	4	118.29	29.57	0.49	0.741	0.0013	0.0003	3.08	0.028
Residual	36	2161.93	60.05	(SSE = 7.75)		0.0038	0.0001	(SSE=0.01)	
Total	40	2280.21			39	0.0051			

Anterior Radius of Curvature					Posterior Radius of Curvature				
Group	Break-Point Age (year)	Break-Point Radius (mm)	Direction Cosines (Age, Radius)		Break-Point Age (year)	Break-Point Radius (mm)	Direction Cosines (Age, Radius)		
Lower Age Upper Age	49	11.47	0.9879 -0.9985	0.1553 0.0547	57	-6.75	-0.9983 0.9999	0.0582 0.0274	
ANOVA	Df	SS	MS	F	Sign	SS	MS	F	Sign
Model	4	418.04	104.51	22.43	2.3E-09	40.39	10.10	25.35	6.4E-10
Residual	36	167.75	4.66	(SSE = 2.16)		13.94	0.40	(SSE=0.63)	
Total	40	585.80			39	54.33			