Theoretical Comparison of Aberration-correcting Customized and Aspheric Intraocular Lenses

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ABSTRACT

PURPOSE: To assess the performance and optical limitations of standard, aspheric, and wavefront-customized intraocular lenses (IOLs) using clinically verified pseudophakic eye models.

METHODS: White light pseudophakic eye models were constructed from physical measurements performed on 46 individual cataract patients and subsequently verified using the clinically measured contrast sensitivity function (CSF) and wavefront aberration of pseudophakic patients implanted with two different types of IOLs. These models are then used to design IOLs that correct the astigmatism and higher order aberrations of each individual eye model's cornea and to investigate how this correction would affect visual benefit, subjective tolerance to lens misalignment (tilt, decentration, and rotation), and depth of field.

RESULTS: Physiological eye models and clinical outcomes show similar levels of higher order aberration and contrast improvement. Customized correction of ocular wavefront aberrations with an IOL results in contrast improvements on the order of 200% over the control and the Tecnis IOLs. The customized lenses can be, on average, decentered by as much as 0.8 mm, tilted >10°, and rotated as much as 15° before their polychromatic modulation transfer function at 8 cycles/degree is less than that of the Tecnis or spherical control lens. Correction of wavefront aberration results in a narrower through focus curve but better out of focus performance for ±0.50 diopters.

CONCLUSIONS: The use of realistic eye models that include higher order aberrations and chromatic aberrations are important when determining the impact of new IOL designs. Customized IOLs show the potential to improve visual performance. [*J Refract Surg*. 2007;23:374-384.]

When a patient undergoes cataract surgery, the cataractous lens is removed and replaced with an intraocular lens (IOL). Intraocular lens implantation following cataract removal is one of the most successful modern surgical procedures. The main optical elements of the pseudophakic eye (an eye implanted with an IOL) are the cornea, pupil, and IOL.

Degradation of images formed on the retina of the eye can come from three different sources: refractive errors, pupil diffraction, and scatter. This article examines the potential image quality improvements associated with correcting all of the monochromatic optical aberrations of the pseudophakic eye.

Each individual eye has a distinct pattern and magnitude of wavefront aberration, which at least partially defines the eye's optical performance. A reduction in ocular wavefront aberration improves optical quality, the contrast of images formed on the retina, and thus spatial vision.[1-11] A large variability exists in the ocular wavefront aberrations measured in normal populations[12-15] and therefore a large variation in the potential improvement achieved through the individualized correction of these aberrations (customized correction) exists.[16] In recent years, several approaches have been used to reduce ocular aberration, including laboratory use of adaptive optics[2,5,17-21] and phase plates,[9,22] and clinical use of customized corneal refractive surgery,[3,23-25] customized contact lens-

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es,[26-28] and wavefront modified IOLs.[6-8,10,29,30] Another focus of recent studies is the determination of metrics for predicting the improvement in visual performance based on the corresponding reduction of wavefront aberrations.[31-34] Similar techniques will be applied in this article, with the main focus being the prediction of the potential improvement associated with IOLs that provide customized correction of wavefront aberrations for a pseudophakic individual.

Positive spherical aberration exhibited by the cornea[35,36] occurs when rays entering the periphery of the pupil are focused in front of rays entering near the center of the pupil. Different IOL models induce different types of wavefront aberration patterns.[29,37-42] Conventional IOLs have spherical surfaces and positive spherical aberration,[6,29,41,42] and thus increase the total positive ocular spherical aberration of the average cataract patient. The SofPort AO lens (Bausch & Lomb, Rochester, NY) is designed to correct the spherical aberration of the isolated lens. Thus, this lens does not add spherical aberration like the spherical IOL.[30] The Tecnis IOL (AMO Inc, Santa Ana, Calif) partially corrects wavefront aberration by compensating for the corneal spherical aberration of an average cataract patient.[43] Recent studies reveal that this lens successfully eliminates spherical aberration in the average pseudophakic eye.[6,29,44] As a result of the correction provided by the Tecnis lens, the postoperative contrast vision of patients implanted with this lens has been shown to be improved.[6-8,10,11] However, neither of these IOLs addresses the variation in corneal spherical aberration that exists in cataract patients on an individual level. Piers et al[20] made use of an adaptive optics system to investigate the potential improvement in contrast sensitivity associated with the customized correction of spherical aberration in a cataract population and showed this improvement to be approximately 32% at 6 cycles/degree. From the results obtained in this study it is clear that there are potential benefits to customized correction of spherical aberration in cataract patients using an IOL.

It is well known that the correction of all higher order aberrations with an IOL will change the sensitivity to lens tilt, decentration, and rotation.[30,43-46] Correction of higher order aberrations will also have an effect on depth of focus.[42,47,48] However, it is not fully understood what the visual benefit would be of a customized IOL designed to correct all wavefront aberrations or the tolerance of such a lens to defocus and misalignments. Wang and Koch[46] studied the predicted performance of IOLs that correct all higher order aberrations of the cornea and the sensitivity of these lenses to decentration. They concluded that for a 6-mm pupil, excellent centration (within 0.48 mm) is necessary for the wavefront aberration of the average eye not to be increased by these lenses. However, these models did not contain chromatic aberration, which is present in all pseudophakic eyes. In this article, to investigate both the potential advantages and disadvantages of new theoretical designs of IOLs, white light eye models are developed from clinical data of individual cataract patients. These models are subsequently verified using the clinically measured visual performance of pseudophakic patients implanted with two different types of IOLs. The models are then used to design customized IOLs for each individual eye model and to investigate how this correction would affect visual benefit and subjective tolerance to defocus and lens displacement (tilt, decentration, and rotation).

MATERIALS AND METHODS

MODEL DEVELOPMENT

A database of >200 cases was collected from a retrospective sample of preoperative measurements performed on patients included in several clinical studies. The studies followed the tenets of the Declaration of Helsinki, and signed informed consent was obtained from every patient after the nature and all possible consequences of the study had been explained.

A random selection was made of 46 eyes from 46 patients and pseudophakic eye models were constructed from physical measurements performed on these cataract patients resulting in 46 individual eye models. Patients with corneal astigmatism >1.50 diopters (D) were not included.

For each model, corneal topography and axial length were used to determine the dimensions. Using corneal topography, individual elevation heights and the location of the pupil center relative to the corneal apex were determined. These data were input into the OSLO optical design software (Lambda Research Corp, Littleton, Mass) to describe the anterior surface of the cornea models and relative location of the pupil. The measured axial length was used as the distance between the anterior corneal surface and the retina of the eye model. The IOL power chosen for each eye to achieve emmetropia was determined using the measured axial length and K values as input for the Lens Haptic Plane formulas.[49] The IOL was positioned in the eye according to the method for determining anterior chamber depth outlined by Norrby et al.[50] The performance of three different IOL models was simulated in these models: a spherical control lens, the Tecnis IOL, and an IOL designed to correct astigmatism and all higher order aberrations of each individual eye model's cornea (customized lens).

All three IOLs were silicone lenses with a refractive index of 1.46 for 545 nm. The control lens is biconvex with two spherical surfaces resulting in positive spherical aberration. The Tecnis IOL has one spherical surface (posterior) and one modified prolate surface (anterior), the shape of which has been described in detail by Holladay et al.[43] One customized lens was designed for each of the 46 individual pseudophakic eye models. To design these lenses, the anterior surface of the IOL was modified so that its shape was described using a set of Zernike polynomials (up to the 4th order). This surface was then optimized to correct the astigmatism and monochromatic (λ=545 nm) higher order aberrations (up to the 4th order) of the eye model using a damped least squares optimization procedure in OSLO for a 4-mm pupil.

All models were white light models, meaning that the refractive indices and dispersion values of all media except the lens were taken from Legrand[51] and all optical performance calculations were performed for 36 wavelengths (from 385 to 745 nm). Polychromatic image quality metrics were calculated by weighting each wavelength contribution with the corresponding value for the wavelength dependent luminosity function of the eye.[52-56] Theoretical calculations of pseudophakic optical quality (average radial polychromatic modulation transfer function [PMTF] and ocular wavefront aberration) were performed in each of the 46 pseudophakic eye models containing each of the three lens models using OSLO and a 4-mm pupil diameter in the best-focused position with the lens centered. The sensitivity to lens displacement was then determined by calculating the PMTF for varying degrees of lens decentration, tilt, and rotation. The eye models are not refocused following lens misalignment. The sensitivity to defocus was determined by performing a through-focus PMTF calculation.

To simulate the best-focusing correction procedure conducted in clinics, a spectacle lens was placed 12 mm in front of the anterior corneal surface and the spherical and cylindrical power were corrected in increments of 0.25 D. Best spectacle correction (spherical and cylindrical power) was determined for a 3-mm aperture as the focus position that maximized the volume under the radial PMTF. For the customized lenses, astigmatism was corrected in the IOL and therefore, best spectacle correction was used only to correct spherical refractive error. The amount of spherical correction provided by the spectacles for the customized models is the same as the values used for the same models containing the Tecnis lens to keep these models as similar as possible for comparison purposes. Defocus is further corrected during the process of customization.

Model Verification

To verify that the models provide relevant information with respect to what we could expect in terms of clinical performance, we compared the values predicted by the theoretical calculations described above to actual clinical measurements collected in a second clinical database of patients implanted with the control and Tecnis IOL models. All patients included in the verification clinical sample were bilateral cataract patients implanted with one silicone spherical control lens (AMO 911A or SI40NB) and one Tecnis IOL. All patients were preoperatively selected from patients with bilateral cataracts and with otherwise healthy eyes. Patients with any ocular pathology, optically irregular eyes, or a history of ocular surgery were excluded. Cataract surgeries were performed by three different surgeons, all of whom performed small-incision surgery, continuous curvilinear capsulorrhexis, and phacoemulsification, followed by implantation of the foldable IOL into the evacuated capsular bag. A total of 79 patients were included. Three months postoperatively, the best-corrected contrast sensitivity function (CSF) was assessed under mesopic lighting conditions (6 cd/m^2) using sine wave grating targets (FACT chart: 1.5, 3, 6, 12, 18 cycles/degree) and the VSRC CST 1500 view-in tester. In the same population, the wavefront of the pseudophakic eyes was measured using a Hartmann-Shack wavefront sensor (λ=780 nm) with a fully pharmacologically dilated pupil. Pupils were dilated to enable a measurement to be performed for a 4-mm pupil. The principles associated with this technique for measuring the wavefront aberration of the eye have been described in detail in previous literature.[57,58] Successful wavefront measurements were performed for a 4-mm pupil in both eyes for 51 of the 79 patients.

For each of the individual pseudophakic eye models, the area under the radial PMTF curve between 1.5 and 12 cycles/degree was calculated for each lens design. Correspondingly, for each pseudophakic patient the area under the CSF was calculated for the same spatial frequencies for both eyes. Simpson's 3/8 rule (4-point closed formula) was applied to calculate the approximate areas under the PMTF and CSF curves. (The values measured and predicted for 18 cycles/degree were not used in the comparison because >50% of the patients could not see the highest contrast pattern presented by the FACT chart at this spatial frequency with at least one of their eyes.) Similar techniques have been applied in the past to determine one global figure for performance of an eye.[31-34,59] For each individual eye model, the ratio of the area under the radial PMTF for the model containing the Tecnis lens to the area under the radial PMTF for the model containing the con-

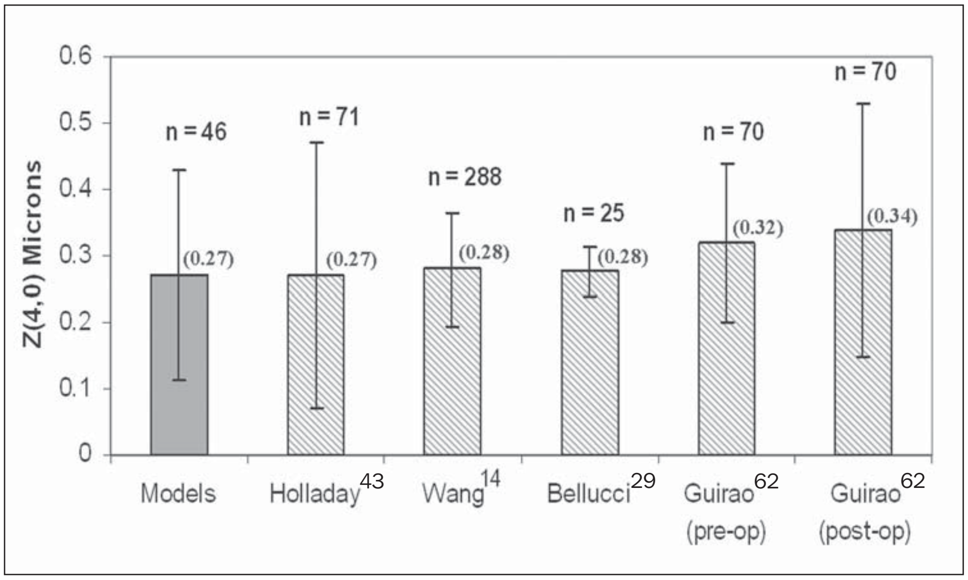


Figure 1. A comparison between corneal spherical aberration values (Z[4,0] measured in microns) in published literature and in models used for the current study. Values are for a 6-mm pupil.

trol lens was calculated and used to estimate a percent of improvement in predicted optical quality (optical improvement factor). For each individual patient, the ratio of the area under the CSF for the eye implanted with the Tecnis lens to the area under the CSF for the eye implanted with the control lens was calculated and used to determine a percent of improvement in measured contrast vision (visual improvement factor). The average optical improvement factor and the average visual improvement factor were then compared to determine whether the models provided a reasonable prediction of the contrast improvement measured in real eyes.

RESULTS

MODEL VERIFICATION

To determine whether the randomly selected pseudophakic eyes chosen for the eye models were representative of a typical cataract population, the corneal spherical aberration was calculated using a ray tracing procedure, described by Guirao et al,[60] for a 6-mm corneal aperture. The resulting average corneal spherical aberration (shown here as the Zernike coefficient of the wavefront aberration expansion Z(4,0) represented using the double-index format for describing ocular aberrations explained by Thibos et al[61]) was compared in Figure 1 to values for corneal spherical aberration found in the literature for the same aperture size.[14,29,43,62] The average axial length of the eyes used in the models is 23.41 mm.

The average wavefront aberration predicted by the pseudophakic eye models (n=46, based on preoperative measurements) was compared to the clinical measurements of ocular wavefront aberration (n=51, postoperative measurements) for a 4-mm pupil in Figure 2A for the Tecnis lens and Figure 2B for the control lens. For each lens design, there was only one wavefront term that had a statistically significant difference between the eye model calculations and the clinical measurements (determined using a *t* test and $P<.05$; a statistically significant difference is represented by an asterisk).

An important aspect of any wavefront-modified lens is what we expect to gain in terms of visual performance. We have examined the ability of our model to predict this by relating improvements in predicted radial PMTF to improvements in measured contrast sensitivity for the Tecnis IOL. Figure 3A shows that for a 4-mm pupil, based on the PMTFs of the Tecnis and control lenses, it is predicted that the Tecnis IOL will provide an improvement in contrast vision (n=46). This was also clinically measured as improved contrast sensitivity, as shown in Figure 3B (n=79). Figure 3C compares the average optical improvement factor (model prediction) and the average visual improvement factor (clinically measured). These average improvement factor figures translate to approximately 20% to 30% contrast improvement for both the measured clinical data and the predictions of the models. There is no statistically significant difference between the optical improvement factor and the visual improvement factor (determined using a *t* test and $P<.05$).

CUSTOMIZED LENSES

Figure 4 shows the average radial PMTF calculated for the pseudophakic eye models with the wavefront aberration customized IOLs. Based on these calculations, it is predicted that fully customized lenses will provide a significant contrast improvement over the

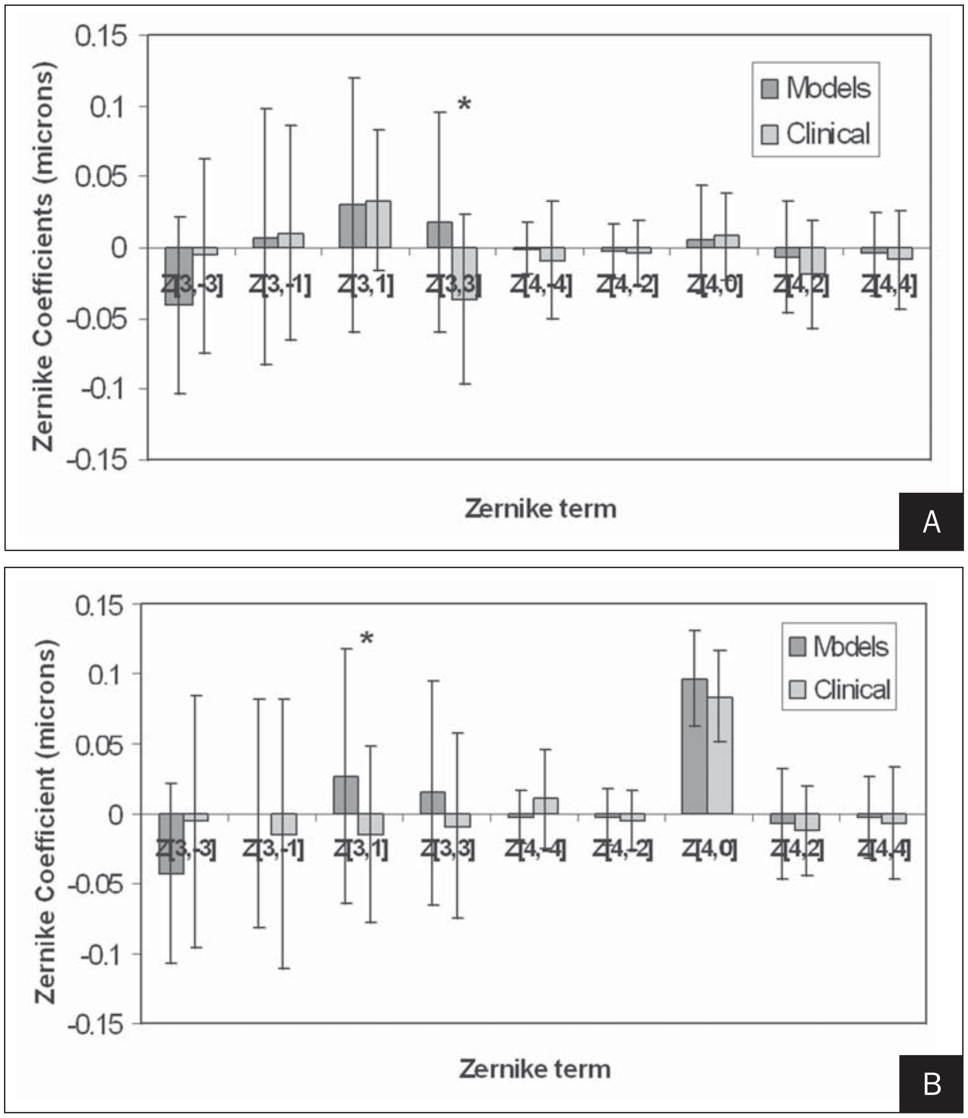
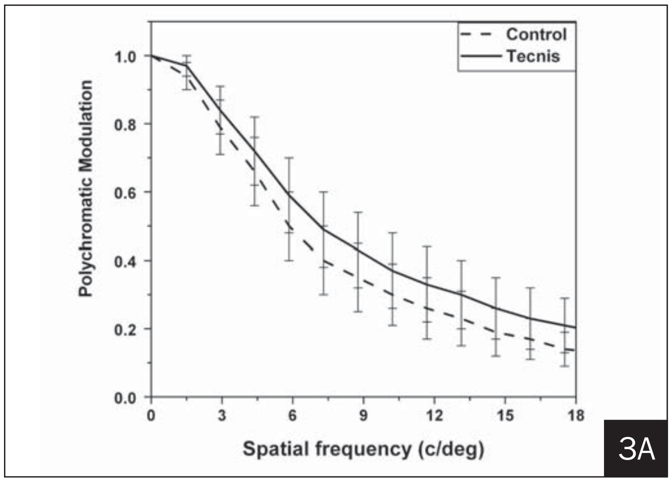


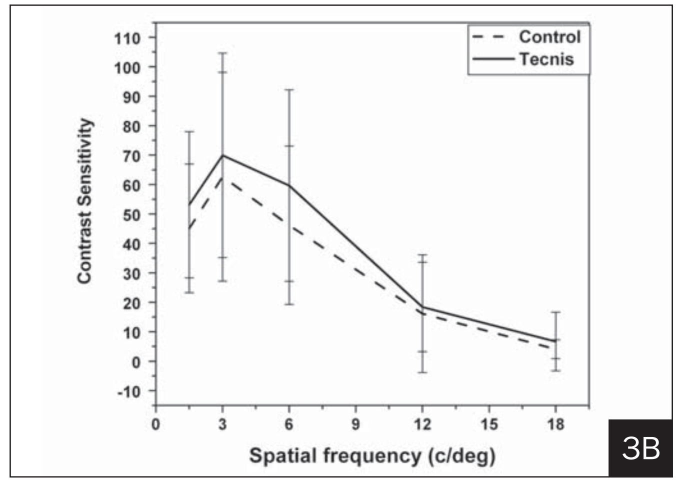
Figure 2. The predicted wavefront aberrations from the eye models were compared to clinical measurements of wavefront aberration for a 4-mm pupil. **A)** Comparison between eye models and eyes with the Tecnis lens. **B)** Comparison between eye models and eyes with the control lens. Statistically significant differences (*P*<.05) are indicated with an asterisk (*).

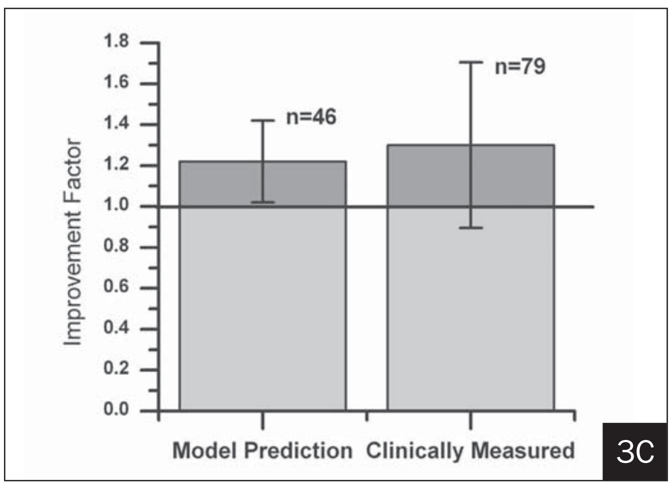
control and the Tecnis IOLs. This improvement is on the order of 200% (based on ratios determined from the calculated area under the radial PMTF). All eye models can be considered well corrected as they all satisfy the Marechal criterion and as such the monochromatic (λ=545) Strehl ratios are >0.82. Figure 5 shows the predicted average radial PMTF at 8 cycles/degree for varying degrees of decentration, tilt, and rotation. Because the slopes of these curves are greatest for the customized lenses, it can be concluded that these lenses are the most sensitive to lens displacement. However, the customized lenses can be, on average, decentered by as much as 0.8 mm, tilted >10°, and rotated as much as 15° before their performance is less than that of the Tecnis or spherical control lens. (Note that the control and Tecnis lenses are rotationally symmetrical and therefore insensitive to rotation.) Through-focus curves for the fully customized lens, the Tecnis lens, and the spherical control lens for 8 cycles/degree and a 4-mm pupil are shown in Figure 6. Based on these calculations, the fully customized lens has a narrower depth of focus than the other two types of lenses. However, within ±0.50 D of defocus, the performance of the customized lenses is better than or equal to that of the control and Tecnis IOLs.

DISCUSSION

During the verification of the eye models it was determined that average values of corneal spherical aberration found in the literature and for the models used herein were surprisingly consistent (see Fig 1), especially considering that these data were from studies of different types of populations and were measured with different instruments. The average axial length of the eyes used in the models was 23.41 mm, whereas the average axial length measured in a population of







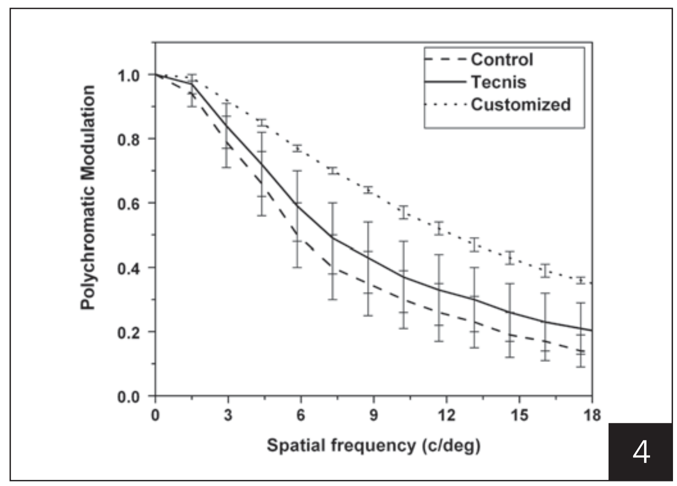


Figure 3. Comparison of **A)** radial polychromatic modulation transfer function (PMTF) and **B)** contrast sensitivity function for the Tecnis eyes and the control eyes. **C)** Comparison between average improvement factors predicted by the PMTF values of the models and the average improvement factor measured in the clinic with contrast sensitivity testing. **Figure 4.** The average radial PMTF of the customized lenses calculated in the 46 eye models.

>15,000 cataract patients was 23.48 mm (Prof. W. Haigis, personal communication, 2006). Because of the closeness of these measurements to the average values used in the pseudophakic eye models, it can be said that the selected population represents an average group of cataract patients in terms of corneal spherical aberration and axial lengths. When comparing the average predicted wavefront aberration to the average aberration measured in patients implanted with two different IOL designs (Tecnis and control), we found that for each design one wavefront term differed significantly (see Fig 2). These differences may be due to lens misalignments. They may also be due to surgically induced changes, as the models are based on preoperative corneal descriptions and the clinical measurements were performed postoperatively. Ideally, we should compare models based on individual physical measurements with the clinical performance of the same individual. From Figure 2, it is also important to note that although there is significant variation in most of the wavefront terms, few of the terms are on average significantly different from 0. One notable exception is the spherical aberration Z(4,0) of the eyes and eye models with the control lens.

All calculations performed in this study were done for a 4-mm pupil. Gobbe[63] determined that for individuals aged >55 years, typical cataract patients, the average pupil size under low luminance conditions (2.5 cd/m^2) was 5.02 mm and the average under high luminance conditions (50 cd/m^2) was 3.22 mm. The pupil size used in the pseudophakic eye models is therefore considered to be a realistic pupil size for cataract patients. The potential benefits associated with the customized correction of wavefront aberration are reduced when the pupil is small. As a result, the improvement in visual performance would be largest in younger patients where the pupil is large, or in situations where the pupil is dilated due to low light conditions.

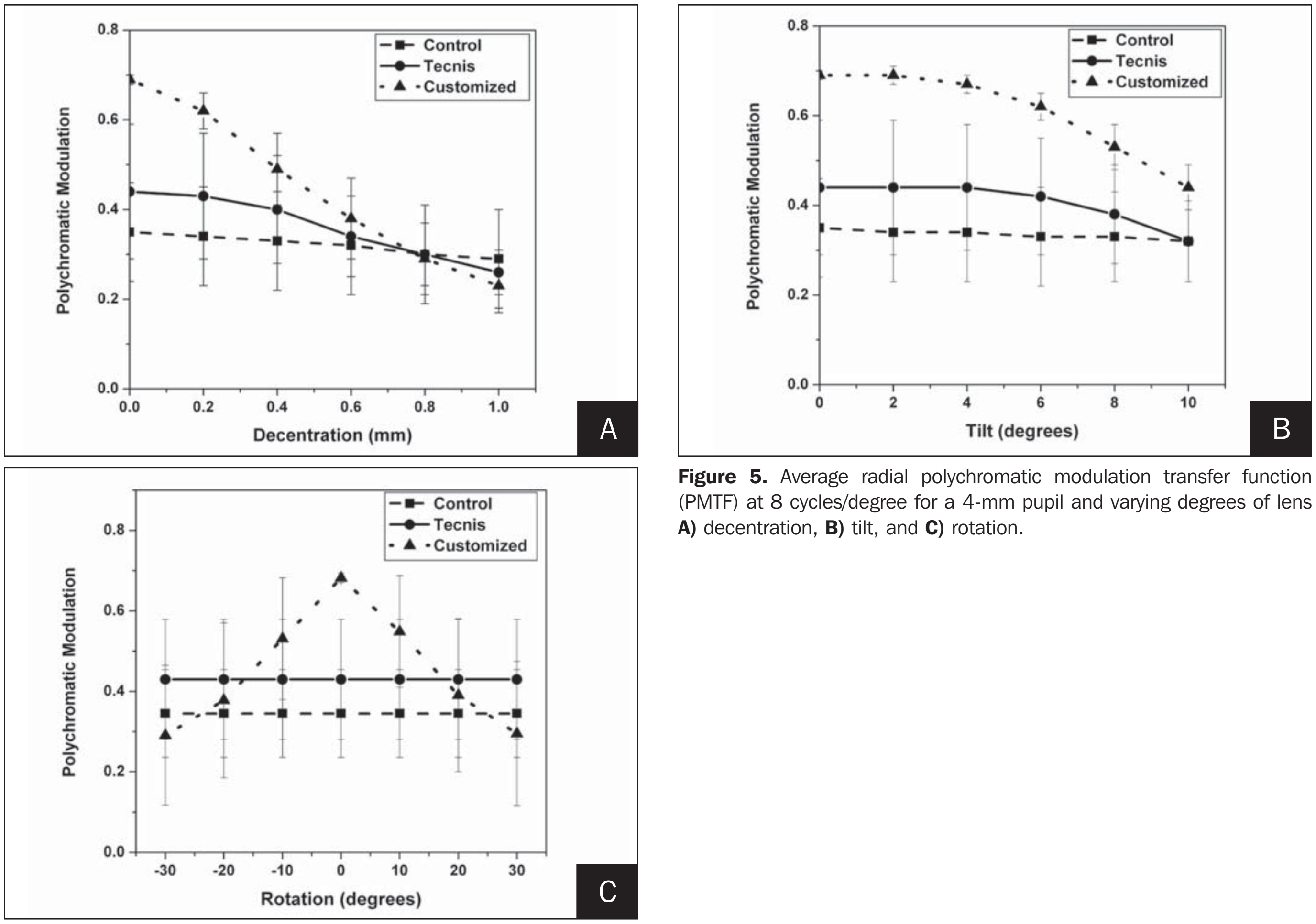


Figure 5. Average radial polychromatic modulation transfer function (PMTF) at 8 cycles/degree for a 4-mm pupil and varying degrees of lens **A)** decentration, **B)** tilt, and **C)** rotation.

In this study, we examined the optical limitations and improvements provided by customized lenses. Ocular optics provide only one limitation to vision. Neural limitations also determine how well any patient implanted with the lenses examined here will perform.

Figure 5 illustrates the importance of using realistic eye models. Previous articles discussing the sensitivity of the Tecnis lens to tilt and decentration state that for a 5-mm pupil, this lens can be decentered 0.3 or 0.4 mm and tilted as much 7° before its performance drops below that of a spherical lens.[30,43] These articles calculate MTF in monochromatic eye models that are rotationally symmetric (ie, models that contain only spherical aberration and use only green light for calculations). In this study, we found that when polychromatic eye models that are asymmetric and reproduce the odd aberrations as well as the spherical aberration of the cornea are used, the Tecnis lens can be decentered as much as 0.8 mm and tilted as much as 10° before its performance is less than that of the spherical control. The more an eye model approaches a quantitative description of a physiological eye, the more realistic its predictions of performance. Because all eyes have higher order aberrations and chromatic aberrations, realistic eye models that include higher order aberrations and chromatic aberrations are important when considering the impact of new designs.

As mentioned above, when wavefront aberrations are corrected with an IOL, the patient's optical performance will be dependent on the degree of misalignment of the lens. The customized lenses designed and examined in this study can be, on average, decentered by as much as 0.8 mm, tilted >10°, and rotated >15° before their performance is less than that of the Tecnis or spherical control lens. Recent studies of tilt and decentration of foldable IOLs have found average decentration values of 0.15 mm, 0.28 mm, and 0.30 mm and average tilts of 1.13°, 2.83°, and 2.41°.[64-66] The median value of rotation of lenses with loop haptics has been measured to be 6.8°.[67] The optical performance of the customized lens remains superior for a clinically normal range of misalignments. Thus, these studies confirm that, on average, modern IOL implantation provides lens centration within the misalignment tolerances needed to achieve improved optical performance with a customized IOL.

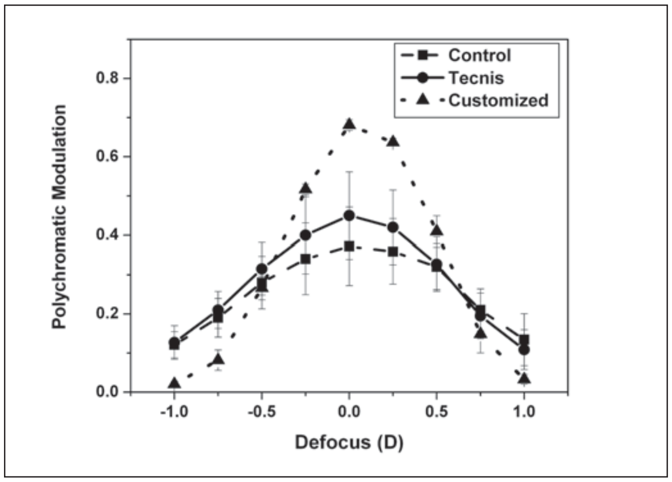


Figure 6. Average through-focus radial polychromatic modulation transfer function (PMTF) at 8 cycles/degree for a 4-mm pupil.

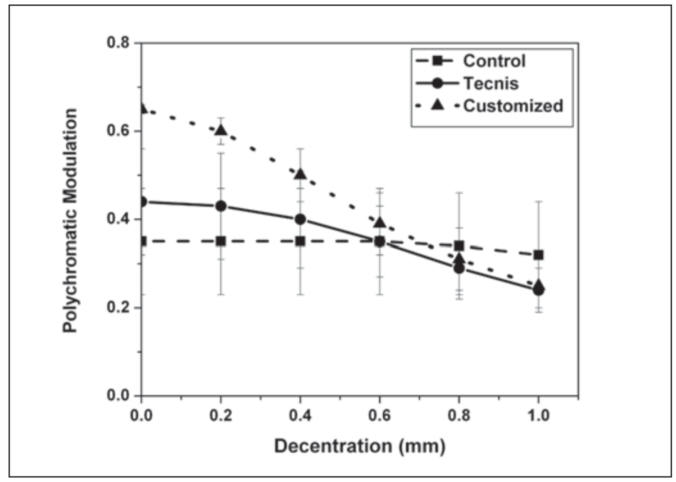


Figure 7. Average radial polychromatic modulation transfer function (PMTF) at 8 cycles/degree for a 4-mm pupil for 3° of temporal lens tilt and varying degrees of lens decentration.

It is a well-known fact that when toric lenses that correct for corneal astigmatism are rotated 15° the resulting astigmatism correction is reduced by 50%.[68] The lenses studied herein are not simple toric lenses that correct only lower order astigmatism, they also correct coma, trefoil, and other higher order aberrations, which will also be influenced by rotation of the lens. This fact explains why there is variation in the rotational limit per patient. However, it can be assumed from the degree of similarity in the limits to which the customized lens can be rotated that astigmatism plays a major role in determining these limits.

The polychromatic performance of the pseudophakic eye models (PMTF) was determined by calculating the monochromatic MTF at 36 different wavelengths, best-focused for green, and then calculating the PMTF by weighting each wavelength contribution by the corresponding spectral sensitivity of the eye (V[λ]). This calculation technique does not include the transverse chromatic aberration of the eye but only the longitudinal chromatic aberration. To make the eye models even more representative of real world optical quality, transverse chromatic aberration could also be included in the models. This could be considered as a next step in this research.

Decentration and tilt rarely exist in isolation.[28] When an IOL is decentered in the eye it is often tilted as well. Due to this fact, sensitivity to decentration in combination with tilt was evaluated. Tabernero et al[44] have shown that the IOLs studied herein tilt predominantly in the temporal direction and recent studies of the magnitude of tilt of foldable IOLs report average values <3°.[64-66] Figure 7 shows the predicted average radial PMTF at 8 cycles/degree for varying degrees of decentration in 3° of temporal tilt. Under these conditions, the customized lenses can be decentered, on average, as much as 0.6 mm before performance is less than that of the Tecnis or spherical control lens.

Using the calculated Strehl ratio for varying degrees of defocus, Marcos et al[42] and Nio et al[47,48] determined theoretically that aspheric lenses that correct ocular spherical aberration showed a decreased tolerance to defocus. Piers et al[20] used adaptive optics simulation to show that when spherical aberration was corrected with an IOL, the visual performance, in terms of visual acuity and contrast sensitivity, was as good as or better than that of a standard spherical IOL for defocus values as large as ±1.00 D. The extent of the depth of field of the visual system depends on what criterion is used to determine what is "acceptably sharp." In this study, we used through-focus PMTF at 8 cycles/degree. This spatial frequency was chosen because it is near the peak of the contrast sensitivity function for normal healthy eyes.[69] Based on these calculations, the fully customized lens has a slightly narrower depth of focus than the other two types of lenses, making it more important to achieve a postoperative target refraction of emmetropia.

Correction of the spherical aberration has profound influence on the power calculation, as evidenced by the fact that the CeeOn 911A and Tecnis Z9000 lenses (AMO Inc) have the same design except for their anterior surfaces, which are spherical and modified prolate, respectively. However, they have A-constants[70] of 118.3 and 119.0, respectively, indicating that on average, a 0.70-D stronger lens is needed for the latter to achieve the same refractive outcome. With spherical aberration present, as in the case of CeeOn 911A or any other spherical lens, best-focus is anterior to paraxial focus, whereas for the Tecnis Z9000 lens, the best-focus position coincides with the paraxial focus. Because currently used power calculation formulas[70-72] are based on thin lens theory, they are incapable of treating spherical aberration. The formula constants are instead adjusted to give, on average, zero postoperative refraction for a large number of cases, so-called personalization. For customized IOLs, personalization is no longer an option, rather new power calculation schemes need to be developed as other aberrations may also influence the power calculation.

Power calculation has two elements: prediction of the postoperative position of the IOL and the optical calculation. An IOL is held in place by contact between its haptics and the tissue of the eye. With the IOL placed in the capsular bag, the plane of contact can be envisaged as the equatorial plane of the bag. This location has been termed the lens haptic plane.[49,73] The axial position of this plane can be determined by regression formulas in terms of preoperatively measured parameters[50] or by direct measurement. In principle, the latter is possible with techniques such as ultrasound biomicroscopy and magnetic resonance imaging, but is not currently used for this purpose. The position of the IOL in relation to the lens haptic plane is given by its mechanical design.[49] The optical calculation must take asphericity into account. This means that the current concept of characterizing the optical power of a lens must then be abandoned, as it is no longer sufficient to characterize the cornea by K-values in the steep and flat meridians and the IOL must be described by its exact optical design.

We have determined that customized IOLs show the potential to improve visual performance while neither significantly limiting depth of focus nor being sensitive to lens decentration or tilt. On the other hand, their potential may be limited by the following factors: preoperative measurement accuracy, manufacturing accuracy, the ability to position the lens in the correct plane such that little rotation occurs, and surgically induced aberrations. In a study published by Guirao et al,[62] it was shown that for small-incision surgery, on average, the wavefront aberrations are unchanged following cataract surgery; on an individual basis, their magnitude and orientation may be susceptible to surgically induced changes. Until this issue is better addressed, our ability to provide customized aberration correction with conventional lenses will be limited.

Optimally, the aberration of the pseudophakic eye should be corrected after lens implantation at a time when the position of the lens in the eye is stable, thereby avoiding the issues described above. This could be achieved through the use of a lens that can be adjusted postoperatively, such as the innovative light adjustable lens being developed by Calhoun Vision.[74] This or similar technologies remain exciting prospects for customized cataract and refractive lens surgery.

REFERENCES

1. Tang CY, Charman WN. Effects of monochromatic and chromatic oblique aberrations on visual performance during spectacle lens wear. *Ophthalmic Physiol Opt*. 1992;12:340-349.
2. Liang J, Williams DR, Miller DT. Supernormal vision and high-resolution retinal imaging through adaptive optics. *J Opt Soc Am A Opt Image Vis Sci*. 1997;14:2884-2892.
3. Seiler T, Mrochen M, Kaemmerer M. Operative correction of ocular aberrations to improve visual acuity. *J Refract Surg*. 2000;16:S619-S622.
4. Marcos S. Aberrations and visual performance following standard laser vision correction. *J Refract Surg*. 2001;17:S596-S601.
5. Yoon GY, Williams DR. Visual performance after correcting the monochromatic and chromatic aberrations of the eye. *J Opt Soc Am A Opt Image Sci Vis*. 2002;19:266-275.
6. Mester U, Dillinger P, Anterist N. Impact of a modified optic design on visual function: clinical comparative study. *J Cataract Refract Surg*. 2003;29:652-660.
7. Kershner RM. Retinal image contrast and functional visual performance with aspheric, silicone, and acrylic intraocular lenses. Prospective evaluation. *J Cataract Refract Surg*. 2003;29:1684-1694.
8. Packer M, Fine IH, Hoffman RS, Piers PA. Improved functional vision with a modified prolate intraocular lens. *J Cataract Refract Surg*. 2004;30:986-992.
9. Yoon G, Jeong TM, Cox IG, Williams DR. Vision improvement by correcting higher-order aberrations with phase plates in normal eyes. *J Refract Surg*. 2004;20:S523-S527.
10. Kennis H, Huygens M, Callebaut F. Comparing the contrast sensitivity of a modified prolate anterior surface IOL and of two spherical IOLs. *Bull Soc Belge Ophtalmol*. 2004;294:49-58.
11. Bellucci R, Scialdone A, Buratto L, Morselli S, Chierego C, Criscuoli A, Moretti G, Piers P. Visual acuity and contrast sensitivity comparison between Tecnis and AcrySof SA60AT intraocular lenses: a multicenter randomized study. *J Cataract Refract Surg*. 2005;31:712-717.
12. Porter J, Guirao A, Cox IG, Williams DR. Monochromatic aberrations of the human eye in a large population. *J Opt Soc Am A*. 2001;18:1793-1803.
13. Thibos LN, Hong X, Bradley A, Cheng X. Statistical variation of aberration structure and image quality in a normal population of healthy eyes. *J Opt Soc Am A Opt Image Sci Vis*. 2002;19:2329-2348.
14. Wang L, Dai E, Koch DD, Nathoo A. Optical aberrations of the human anterior cornea. *J Cataract Refract Surg*. 2003;29:1514-1521.
15. Castejon-Mochon JF, Lopez-Gil N, Benito A, Artal P. Ocular wave-front aberration statistics in a normal young population. *Vision Res*. 2002;42:1611-1617.
16. Guirao A, Porter J, Williams DR, Cox IG. Calculated impact of higher-order monochromatic aberrations on retinal image quality in a population of human eyes. *J Opt Soc Am A*. 2002;19:1-9.
17. Vargas-Martin F, Prieto PM, Artal P. Correction of the aberrations in the human eye with a liquid-crystal spatial light modulator: limits to performance. *J Opt Soc Am A*. 1998;15:2552-2562.
18. Hofer H, Chen L, Yoon GY, Singer B, Yamauchi Y, Williams DR. Improvement in retinal image quality with dynamic correction of the eye's aberrations. *Optics Express*. 2001;8:631-643.

19. Fernandez EJ, Manzanera S, Piers P, Artal P. Adaptive optics visual simulator. *J Refract Surg*. 2002;18:S634-S638.

20. Piers PA, Fernandez EJ, Manzanera S, Norrby S, Artal P. Adaptive optics simulation of intraocular lenses with modified spherical aberration. *Invest Ophthalmol Vis Sci*. 2004;45:4601-4610.

21. Vdovin G, Loktev M, Simonov A, Kijko V, Volkov S. Adaptive correction of human-eye aberrations in a subjective feedback loop. *Opt Lett*. 2005;30:795-797.

22. Navarro R, Moreno-Barriuso E, Bara S, Mancebo T. Phase plates for wave-aberration compensation in the human eye. *Opt Lett*. 2000;25:236-238.

23. Mrochen M, Kaemmerer M, Seiler T. Clinical results of wavefront-guided laser in situ keratomileusis 3 months after surgery. *J Cataract Refract Surg*. 2001;27:201-207.

24. Sarkisian KA, Petrov AA. Clinical experience with the customized low spherical aberration ablation profile for myopia. *J Refract Surg*. 2002;18:S352-S356.

25. Kohnen T, Buhren J, Kuhne C, Mirshahi A. Wavefront-guided LASIK with the Zyoptix 3.1 system for the correction of myopia and compound myopic astigmatism with 1-year follow-up: clinical outcome and change in higher order aberrations. *Ophthalmology*. 2004;111:2175-2185.

26. Lopez-Gil N, Castejon-Mochon JF, Benito A, Marin JM, Lo-a-Foe G, Marin G, Fermigier B, Renard D, Joyeux D, Chateau N, Artal P. Aberration generation by contact lenses with aspheric and asymmetric surfaces. *J Refract Surg*. 2002;18:S603-S609.

27. de Brabander J, Chateau N, Marin G, Lopez-Gil N, Van Der Worp E, Benito A. Simulated optical performance of custom wavefront soft contact lenses for keratoconus. *Optom Vis Sci*. 2003;80:637-643.

28. Dietze HH, Cox MJ. Correcting ocular spherical aberration with soft contact lenses. *J Opt Soc Am A Opt Image Sci Vis*. 2004;21:473-485.

29. Bellucci R, Morselli S, Piers P. Comparison of wavefront aberrations and optical quality of eyes implanted with five different intraocular lenses. *J Refract Surg*. 2004;20:297-306.

30. Altmann GE, Nichamin LD, Lane SS, Pepose JS. Optical performance of 3 intraocular lens designs in the presence of decentration. *J Cataract Refract Surg*. 2005;31:574-585.

31. Cheng X, Thibos LN, Bradley A. Estimating visual quality from wavefront aberration measurements. *J Refract Surg*. 2003;19:S579-S584.

32. Applegate RA, Marsack JD, Ramos R, Sarver EJ. Interaction between aberrations to improve or reduce visual performance. *J Cataract Refract Surg*. 2003;29:1487-1495.

33. Piers P, Norrby NE, Mester U. Eye models for the prediction of contrast vision in patients with new intraocular lens designs. *Opt Lett*. 2004;29:733-735.

34. Marsack JD, Thibos LN, Applegate RA. Metrics of optical quality derived from wave aberrations predict visual performance. *J Vis*. 2004;4:322-328.

35. Oshika T, Klyce SD, Applegate RA, Howland HC. Changes in corneal wavefront aberrations with aging. *Invest Ophthalmol Vis Sci*. 1999;40:1351-1355.

36. Guirao A, Redondo M, Artal P. Optical aberrations of the human cornea as a function of age. *J Opt Soc Am A*. 2000;17:1697-1702.

37. Taketani F, Matsuura T, Yukawa E, Hara Y. High-order aberrations with Hydroview H60M and AcrySof MA30BA intraocular lenses: comparative study. *J Cataract Refract Surg*. 2004;30:844-848.

38. Vilarrodona L, Barrett GD, Johnson B. High-order aberrations in pseudophakia with different intraocular lenses. *J Cataract Refract Surg*. 2004;30:571-575.

39. Casprini F, Balestrazzi A, Tosi GM, Miracco F, Martone G, Cevenini G, Caporossi A. Glare disability and spherical aberration with five foldable intraocular lenses: a prospective randomized study. *Acta Ophthalmol Scand*. 2005;83:20-25.

40. Pesudovs K, Dietze H, Stewart OG, Noble BA, Cox MJ. Effect of cataract surgery incision location and intraocular lens type on ocular aberrations. *J Cataract Refract Surg*. 2005;31:725-734.

41. Barbero S, Marcos S, Jimenez-Alfaro I. Optical aberrations of intraocular lenses measured in vivo and in vitro. *J Opt Soc Am A Opt Image Sci Vis*. 2003;20:1841-1851.

42. Marcos S, Barbero S, Jimenez-Alfaro I. Optical quality and depth-of-field of eyes implanted with spherical and aspheric intraocular lenses. *J Refract Surg*. 2005;21:223-235.

43. Holladay JT, Piers PA, Koranyi G, van der Mooren M, Norrby NE. A new intraocular lens design to reduce spherical aberration of pseudophakic eyes. *J Refract Surg*. 2002;18:683-691.

44. Tabernero J. How robust to misalignments are IOLs correcting spherical aberration. Presented at: 6th International Congress on Wavefront Sensing and Optimized Refractive Corrections; February 13, 2005; Athens, Greece.

45. Dietze HH, Cox MJ. Limitations of correcting spherical aberration with aspheric intraocular lenses. *J Refract Surg*. 2005;21:S541-S546.

46. Wang L, Koch DD. Effect of decentration of wavefront-corrected intraocular lenses on the higher-order aberrations of the eye. *Arch Ophthalmol*. 2005;123:1226-1230.

47. Nio YK, Jansonius NM, Fidler V, Geraghty E, Norrby S, Kooijman AC. Spherical and irregular aberrations are important for the optimal performance of the human eye. *Ophthalmic Physiol Opt*. 2002;22:103-112.

48. Nio YK, Jansonius NM, Geraghty E, Norrby S, Kooijman AC. Effect of intraocular lens implantation on visual acuity, contrast sensitivity, and depth of focus. *J Cataract Refract Surg*. 2003;29:2073-2081.

49. Norrby S. Using the lens haptic plane concept and thick-lens ray tracing to calculate intraocular lens power. *J Cataract Refract Surg*. 2004;30:1000-1005.

50. Norrby S, Lydahl E, Koranyi G, Taube M. Clinical application of the lens haptic plane concept with transformed axial lengths. *J Cataract Refract Surg*. 2005;31:1338-1344.

51. Legrand Y. *Form and Space.* Millidot M, Heath GG, trans-ed. London, United Kingdom: Indiana University Press; 1967.

52. *CIE Proceedings*. Paris, France: Bureau Central de la CIE; 1951.

53. Vos JJ. Colorimetric and photometric properties of a 2-deg fundamental observer. *Color Research and Application*. 1978;3:125-128.

54. Kokoschka S, Adrian WK. Influence of field size on the spectral sensitivity of the eye in the photopic and mesopic range. *Am J Optom Physiol Opt*. 1985;62:119-126.

55. Stockman A, MacLeod DI, Johnson NE. Spectral sensitivities of the human cones. *J Opt Soc Am A*. 1993;10:2491-2521.

56. van den Berg TJ, Tan KE. Light transmittance of the human cornea from 320 to 700 nm for different ages. *Vision Res*. 1994;34:1453-1456.

57. Liang J, Williams DR. Aberrations and retinal image quality of the normal human eye. *J Opt Soc Am A*. 1997;14:2873-2883.

58. Prieto PM, Vargas-Martin F, Goelz S, Artal P. Analysis of the performance of the Hartmann-Shack sensor in the human eye. *J Opt Soc Am A*. 2000;17:1388-1398.

59. Cornelissen FW, Bootsma A, Kooijman AC. Object perception by visually impaired people at different light levels. *Vision Res*. 1995;35:161-168.

60. Guirao A, Artal P. Corneal wave aberration from videokeratography: accuracy and limitations of the procedure. *J Opt Soc Am A Opt Image Sci Vis*. 2000;17:955-965.

61. Thibos LN, Applegate RA, Schwiegerling JT, Webb R. Standards for reporting the optical aberrations of eyes. *Vision Science and its Applications, OSA Trends in Optics and Photonics*. 2000;35:110.

62. Guirao A, Tejedor J, Artal P. Corneal aberrations before and after small-incision cataract surgery. *Invest Ophthalmol Vis Sci*. 2004;45:4312-4319.

63. Gobbe M. Higher-order ocular aberrations do not increase with age. Presented at: 6th International Congress on Wavefront Sensing and Optimized Refractive Corrections; February 13, 2005; Athens, Greece.

64. Akkin C, Ozler SA, Mentes J. Tilt and decentration of bag-fixated intraocular lenses: a comparative study between capsulorhexis and envelope techniques. *Doc Ophthalmol*. 1994;87:199-209.

65. Hayashi K, Harada M, Hayashi H, Nakao F, Hayashi F. Decentration and tilt of polymethyl methacrylate, silicone, and acrylic soft intraocular lenses. *Ophthalmology*. 1997;104:793-798.

66. Mutlu FM, Bilge AH, Altinsoy HI, Yumusak E. The role of capsulotomy and intraocular lens type on tilt and decentration of polymethylmethacrylate and foldable acrylic lenses. *Ophthalmologica*. 1998;212:359-363.

67. Patel CK, Ormonde S, Rosen PH, Bron AJ. Postoperative intraocular lens rotation: a randomized comparison of plate and loop haptic implants. *Ophthalmology*. 1999;106:2190-2195.

68. Morlet N, Minassian D, Dart J. Astigmatism and the analysis of its surgical correction. *Br J Ophthalmol*. 2001;85:1127-1138.

69. Nio YK, Jansonius NM, Fidler V, Geraghty E, Norrby S, Kooijman AC. Age-related changes of defocus-specific contrast sensitivity in healthy subjects. *Ophthalmic Physiol Opt*. 2000;20:323-334.

70. Retzlaff JA, Sanders DR, Kraff MC. Development of the SRK/T intraocular lens implant power calculation formula. *J Cataract Refract Surg*. 1990;16:333-340.

71. Holladay JT, Prager TC, Chandler TY, Musgrove KH, Lewis JW, Ruiz RS. A three-part system for refining intraocular lens power calculations. *J Cataract Refract Surg*. 1988;14:17-24.

72. Hoffer KJ. The Hoffer Q formula: a comparison of theoretic and regression formulas. *J Cataract Refract Surg*. 1993;19:700-712.

73. Norrby NE. The lens haptic plane (LHP) a fixed reference for IOL implant power calculation. *European Journal of Implants and Refractive Surgery*. 1995;7:202-209.

74. Sandstedt CA, Chang S, Schwartz DM. The Calhoun light adjustable lens: a postinsertion method for the correction of refractive errors. In: Krueger RR, Applegate RA, MacRae SM, eds. *Wavefront Customized Visual Correction: The Quest for Super Vision II*. Thorofare, NJ: SLACK Inc; 2004:291-297.