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# Comparison of through-focus image quality across five presbyopia-correcting intraocular lenses (an American Ophthalmological Society Thesis)

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# COMPARISON OF THROUGH-FOCUS IMAGE QUALITY ACROSS FIVE PRESBYOPIA-CORRECTING INTRAOCULAR LENSES (AN AMERICAN OPHTHALMOLOGICAL SOCIETY THESIS)

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## ABSTRACT

*Purpose*: To assess through-focus polychromatic image sharpness of five US Food and Drug Administration–approved presbyopia-correcting intraocular lenses (IOLs) through a range of object vergences and pupil diameters utilizing an image sharpness algorithm.

*Methods:* A 1951 US Air Force resolution target was imaged through a Crystalens AO (AO) (Bausch & Lomb Surgical, Aliso Viejo, California), Crystalens HD (HD) (Bausch & Lomb Surgical, Aliso Viejo, California), aspheric ReSTOR +4.0 (R4) (Alcon Laboratories, Fort Worth, Texas), aspheric ReSTOR +3.0 (R3) (Alcon Laboratories, Fort Worth, Texas), and Tecnis Multifocal Acrylic (TMF) (Abbott Medical Optics, Irvine, California) IOL in an anatomically and optically accurate model eye and captured digitally for each combination of pupil diameter and object vergence. The sharpness of each digital image was objectively scored using a two-dimensional gradient function.

*Results:* The AO lens had the best distance image sharpness for all pupil diameters, followed by the HD. With a 5-mm pupil, the R4 lens achieved distance image quality similar to the HD, but inferior to the AO. The R3 successfully moved the near focal point farther from the patient compared to the R4, but did not improve image sharpness at intermediate distances and showed worse distance and near image sharpness. Consistent with apodization, the ReSTOR IOLs displayed better distance and poorer near image sharpness as pupil diameter increased. The TMF lens showed consistent distance and near image sharpness across pupil diameters and exhibited the best near image sharpness for all pupil diameters.

*Conclusions:* Differing IOL design strategies to increase depth of field are associated with quantifiable differences in image sharpness at varying vergences and pupil sizes. An objective comparison of the imaging properties of specific presbyopia-correcting IOLs, in conjunction with patients' pupil sizes, can be useful in selecting the most appropriate IOL for each patient.

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## INTRODUCTION

Presbyopia[1] and cataract[2-4] represent the two most common human ocular afflictions. An effort to focus on a near object triggers the process of accommodation—an increase in the dioptric power of the eye resulting from contraction of the ciliary body associated with a change in lens shape and an increase in the lens surface curvatures.[5] With increasing age, the ability to accommodate diminishes,[6] and the onset of presbyopia affects 100% of the population over 50 years of age.[7]

In the past, cataract extraction with monofocal intraocular lens (IOL) implantation addressed the loss of contrast and forward light scatter associated with cataract, but did not remedy the effects of the age-associated loss of accommodation on limiting the range of functional vision and increasing spectacle dependence. With monofocal IOLs (and perhaps even more so in patients receiving aspheric monofocal IOLs), a functional restoration of near was limited to those patients choosing pseudophakic monovision or those patients (roughly 3% to 8% in most studies[8-12]) fortunate enough to have adequate pseudoaccommodation to readily perform near tasks (such as reading a newspaper without glasses) with both IOLs targeted to emmetropia. Pseudoaccommodative mechanisms that can conspire to improve near vision include increased instantaneous depth of focus via miotic pupil size, ptotic eyelids and squint, myopic against-the-rule astigmatism, corneal multifocality, chromatic aberration, and monochromatic aberrations (eg, spherical aberration and coma).[11,13]

The AMO Array (Advanced Medical Optics, Irvine, California) was the first multifocal IOL approved for use in the United States by the US Food and Drug Administration (FDA) in 1997. This was a zonal-progressive, distance-center dominated, refractive multizone IOL with five alternating near and far concentric rings.[14] Since that time, different approaches to IOL design have evolved in an effort to address the impact of presbyopia, and new strategies and technologies will continue to appear.[15] The multifocal IOLs are designed with refractive and/or diffractive optical properties, which focus light on multiple foci and allow the patient to see both near and distant objects.[16] Since the light energy is distributed between more than one image and some energy is lost to useless foci,[15,16] each primary image produced by a multifocal IOL is thereby fainter and defocused at any given pupil aperture, compared to monofocal or accommodating IOLs that do not split up light in this manner. In contrast to fully diffractive multifocal IOLs, where the step height of each concentric ring is uniform, apodized diffractive-refractive IOLs take advantage of progressive step heights to shift the energy distribution toward the distance foci with larger pupil diameters. Accommodating IOLs are designed to change the refractive power of the eye and/or induce higher-order aberrations to improve through-focus via the movement of the ciliary muscle and changes in vitreous pressure. Unlike multifocal IOLs, accommodating IOLs have a single point of focus and so light energy is not divided between multiple images. However, the increase in true dioptric power with current single-optic accommodating IOLs may be limited, and the clinically observed enhanced near vision may be the result of dynamic changes in spherical aberration and other higher-order aberrations resulting from accommodative effort.

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Currently, in the United States, these presbyopia-correcting IOL designs include spherical progressive zonal refractive, aspheric full-aperture diffractive (Figure 1, left), spherical, and more recently, aspheric apodized-diffractive-refractive (Figure 1, middle), spherical and aspheric accommodating (Figure 1, right), and a spherical accommodating IOL with central bispheric optic modification. In the near future, we will likely have toric versions of all of these IOLs, along with dual-optic accommodating IOLs[17,18] and shape-changing accommodating IOLs,[17,19] and others.

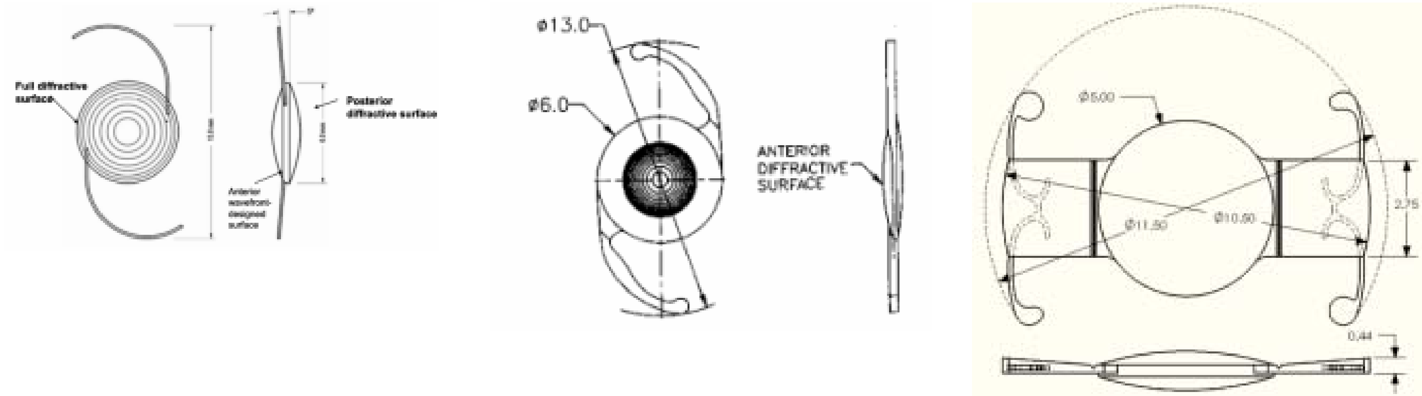


**FIGURE 1**

Left, The Tecnis multifocal IOL is an aspheric, full-aperture diffractive IOL with a wavefront-guided modified prolate anterior surface designed to offset the average positive corneal spherical aberration and a posterior diffractive surface with concentric diffractive rings of uniform step height extending across the entire optic. Middle, The AcrySof ReSTOR 3.0 and 4.0 IOLs are aspheric apodized-diffractive-refractive IOLs, with 9 and 12 concentric diffractive zones of decreasing step height from center to periphery on the anterior central 3.6-mm lens surface and an aspheric posterior surface imparting -0.1 µm of spherical aberration to a 6-mm wavefront. Right, The Crystalens AO is a silicone accommodating IOL harboring terminal polyimide loops on two plate haptics, which are hinged to a central 5-mm aspheric, zero higher-order aberration optic. The Crystalens HD is a spherical, modified plate haptic accommodating IOL of similar overall design specifications, distinguished by a central 1.5-mm bispheric modification of the optic that increases the effective add with miosis.

In this study, we compared five FDA-approved IOLs for the correction of aphakia and presbyopia, as follows:

- The Crystalens AO (model AT-50AO, Bausch & Lomb Surgical, Aliso Viejo, California) silicone multipiece accommodating IOL is a modified plate haptic lens. The plate haptics are hinged to the 5.0-mm optic and have terminal small looped polyimide haptics. The overall length of the lens from loop tip to loop tip is 11.5 mm, and the overall length from the ends of the plate haptic is 10.5 mm. This aspheric lens has uniform power across the optic and has zero higher-order aberrations.
- The Crystalens HD (model HD-500, Bausch & Lomb Surgical, Aliso Viejo, California) is a spherical accommodating modified plate haptic lens with a bispheric modification of the central 1.5-mm zone of the 5-mm optic (Figure 2), increasing the effective add during miosis.
- The AcySof ReSTOR +3.0 (model SN6AD1, Alcon Laboratories, Fort Worth, Texas) is a single-piece, foldable, aspheric multifocal combining the functions of a central apodized diffractive region on a refractive platform. This lens has been modified such that the apodized diffractive optics within the central 3.6-mm zone on the anterior surface are composed of nine concentric steps of gradually decreasing height, creating multifocality from near to far, with two primary foci. The refractive region of the optic surrounds the apodized diffractive region. This area directs light to a distance focal point for larger pupil diameter and is dedicated to distance vision. The IOL has a symmetric biconvex design with an anterior aspheric optic designed to reduce whole-eye spherical aberration (imparts -0.1 µm spherical aberration to a 6-mm wavefront entering the eye). The IOL incorporates a +3.0 D near add at the IOL plane, which translates into approximately +2.4 D at the spectacle plane.
- The aspheric AcrySof ReSTOR +4.0 (model SN6AD3, Alcon Laboratories, Fort Worth, Texas) is an apodized, diffractive, single-piece, foldable hydrophobic posterior chamber IOL with a biconvex 6-mm optic. The posterior surface of this biconvex optic has an aspheric design, and the lens is designed to impart -0.1 µm of spherical aberration to a 6-mm wavefront entering the eye. It has a central 3.6-mm, apodized diffractive optic region, where 12 concentric diffractive zones on the anterior surface of the lens divide light into two primary diffractive orders to create two lens powers. The add power of this ReSTOR is 4.0 D at the IOL plane, which provides approximately 3.2 D at the spectacle plane. Unlike the uniform step heights of the fully diffractive Tecnis Multifocal IOL, the ReSTOR lens uses step heights that decrease with increasing distance from the lens center via the process of apodization. While the diffractive steps introduce phase delay of light at the zone boundaries, the benefit of apodization is that it greatly increases the proportion of light energy directed to the distance focus with larger pupil diameters, such that the amount of light directed at distance

with a 5-mm pupil is more than double that provided by a full diffractive optic. Depending on pupil size, a variable proportion of light energy is not directed to either near or far foci, but rather is lost to higher diffractive orders with the ReSTOR IOL (Figure 3). For example, with a very small pupil, this lens distributes approximately 41% of light at near and 41% at distance and 18% is lost to useless foci, whereas at a 5-mm pupil, 84% of light is distributed at distance and 10% at near and only 6% is lost.

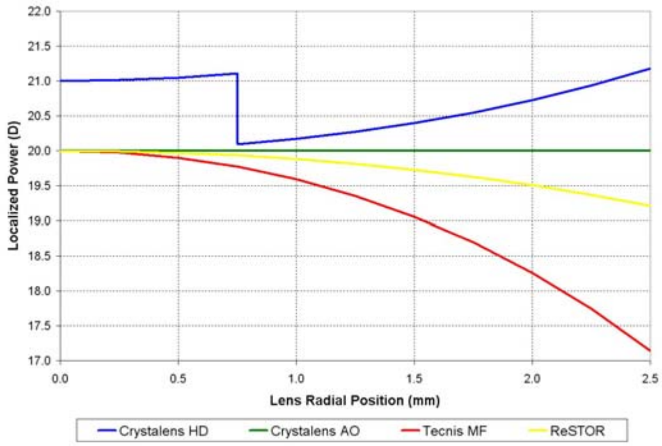


**FIGURE 2**

The spatial dioptric power profile of different presbyopia-correcting IOLs.

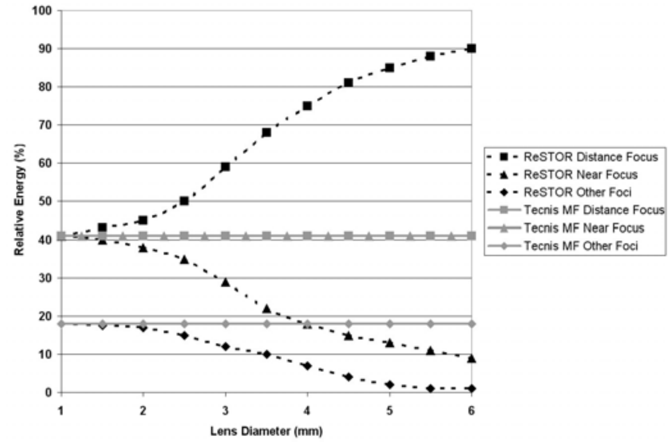


**FIGURE 3**

Relative distribution of light energy at varying pupil apertures with two multifocal IOLs.

• The Tecnis Multifocal foldable hydrophobic acrylic IOL (Model ZMA00, Abbott Medical Optics, Irvine, California) is an aspheric ultraviolet light–absorbing posterior chamber IOL. The lens imparts -0.27 µm of spherical aberration to a 6-mm wavefront entering the eye, which is intended to offset the average positive spherical aberration of the cornea. Light passing through the lens is evenly distributed between the distance and near foci at all pupil sizes (Figure 3), with 41% of light energy at distance, 41% at near, and 18% lost to higher-diffraction orders. The near power represents a +4.0 D add in actual lens power at the IOL plane and approximately 3.2 D at the spectacle plane. Higher-diffraction orders cause focus locations at integrals of the 4 D add (eg, 8 D, 12 D, -4 D, -8 D), and the loss of 18% of light to these useless foci is a consequence of the overall interaction of light at the diffraction steps The biconvex lens has a modified prolate anterior surface and a fully diffractive bifocal posterior surface. The diffractive pattern consists of 32 concentric rings with equal step heights of approximately 0.25 µm. The diameter of the central zone is 1 mm.

With each presbyopia-correcting IOL design strategy, there are inherent tradeoffs with regard to contrast sensitivity, loss of light energy to useless foci, night glare and photic phenomenon, and near, intermediate, and distance image quality at any given pupil diameter.[10,13,16,20-28] The better we can assess and model the optical performance of each IOL at different vergences and pupil diameters, the more information we have by which to custom-match different presbyopia-correcting IOLs for individual patients who may have marked differences in pupil size, shape, and dynamics, all of which may further change with advancing age.[29,30]

In photography, autofocus is the process by which the camera lens position (ie, focal length) is automatically adjusted to maximize the focus (ie, sharpness) of the image. Active autofocus uses a sensor or measuring device to determine the distance between the object and the lens, whereas passive autofocus relies on image information alone, using different autofocus functions that evaluate pixel information to determine focus.[31] Autofocus algorithms have been applied in various medical specialties, including digital retinal[32] and optic nerve photography.[33] While not the only image quality metric, image sharpness is an important element of image quality. As per International Organization for Standardization (ISO) standard 11979, the US Air Force (USAF) 1951 resolution target is routinely employed to subjectively determine the spatial resolution efficiency of the IOL.[13,34] In this study, we utilized an image sharpness algorithm similar to that commonly used in photography, to provide an objective computation of the image sharpness of the 1951 USAF target imaged through an anatomically and optically accurate model eye incorporating each of five presbyopia-correcting IOLs.

The 1951 USAF resolution target was created in response to the need to assign numerical tolerances to optical systems and photographic processes. In the target, the change in pattern size is in a geometric progression based on the sixth root of 2. The number of lines per millimeter doubles with every sixth target element. These six elements are known as a group, and the group heading is the power to which the first element is raised to express the number of lines per millimeter in that element. For example, for a group heading of -2, 2 taken to the -2 power is equivalent to 0.25 lines per millimeter. Following the ISO standards to determine the resolution efficiency of an IOL,[34] the USAF 1951 target group and the smallest element of the group in which the image pattern can be resolved are recorded and used to determine the spatial frequency of the resolution limit. In terms most familiar to ophthalmologists, the group 3/element 2 and group 2/element 2 components of the resolution target in this setup approximate the Snellen letter size of

20/20 and 20/40, respectively. To our knowledge, this is the first study to utilize the standard 1951 USAF target linked to an objective measure of image sharpness by which to compare five presbyopia-correcting IOLs in a model eye system.

A number of different metrics besides image sharpness have been frequently utilized to characterize optical quality. For example, the modulation transfer function of individual IOLs can be determined over a range of spatial frequencies, thereby providing another objective image quality metric. The point spread function of the IOL is determined, and then by Fourier transformation an optical transfer function can be determined, which describes how various spatial frequencies are transmitted by the optical system. The absolute value of the optical transfer function refers to the modulation (contrast) transfer function, and the phase transfer function describes spatial shifts of the image relative to the object.[35]

Neither the image sharpness function nor modulation transfer function of IOLs in a model eye represents or reflects the patient's perception of an image, which is beyond the scope of this study. While these aforementioned metrics are useful in describing important components of simulated retinal image quality, they do not incorporate or in any way account for the neural component of vision, which relies on the ultimate image processing system—the human brain. Although some have suggested that the other specific quality of vision metrics, such as the visual optical transfer function, may more closely relate to subjective vision,[36] nevertheless, good correlation has been found among predictive modulation transfer function testing of IOLs with physiological eye models, wavefront aberrometry, and mesopic contrast sensitivity testing.[37]

The optics of both refractive and diffractive multifocal IOLs present substantial challenge to the fidelity of wavefront sensors. The diffractive multifocals utilize concentric steps that induce discrete repetitive phase jumps to make the light interfere constructively at more than one foci. The square microlens array of the Hartmann-Shack sensor may not effectively sample the locally distorted wavefronts and, depending on the lenslet number, may result in the apparition of some additional centroids straying inside or outside their pixel subarrays.[38] The double-pass technique records images of a point source after their reflection off the retina and double-pass through the ocular media. Dual-pass systems that directly sample the retinal point spread function may be more sensitive to the impact of scatter and higher-order aberrations that may be "smoothed out" by mathematical decomposition of Shack-Hartmann centroid images.[39] However, the measured point spread cannot be reconstructed to provide the magnitude or direction of specific higher-order aberrations.

While providing important tools for modeling the optical effect of specific IOLs in the human eye, the results of optical bench studies, as presented herein, must ultimately be coupled with and compared to the results of clinical studies to support or refute their predictive value. In addition, the optical bench does not allow the accommodating IOLs to be tested in configurations other than for distance. Nevertheless, the model system that we describe and the application of objective sharpness metrics represent a useful simulation and means to objectively compare the characteristics and quality of the retinal image produced by each IOL at different pupil apertures and object vergences.

## METHODS

### TEST ARTICLES

Tognetto and associates[38] investigated the optical quality of three lenses each of 23 different IOL models and found the optical quality among IOLs of the same manufacturer and model had almost no variance. Hence, in the current study, one representative lens of each of the following five presbyopia-correcting IOL models was measured:

Crystalens AO (model AT-50AO), referred to herein as AO
Crystalens HD (model HD-500), referred to herein as HD
ReSTOR +3.0 (model SN6AD1), referred to herein as R3
ReSTOR +4.0 (model SN6AD3), referred to herein as R4
Tecnis Multifocal acrylic (model ZMA00), referred to herein as TMF

The key optical properties of the five different presbyopia-correcting IOL models are listed in the Table. The Tecnis Multifocal had a labeled dioptric power of +21.0 diopters, and all other IOLs had a labeled dioptric power of +20.0 diopters.

### MEASUREMENT APPARATUS

The apparatus used to measure the through-focus imaging quality of different IOL models is shown schematically in Figure 4. The apparatus consisted of the following key components: target assembly, Badal relay assembly, and model eye/camera assembly. The target assembly consisted of a light source, a diffuser, a 1951 USAF resolution target, and an achromatic collimating lens. The light source was a high-intensity broadband halogen lamp. The Badal relay assembly consisted of two achromatic lenses, and the distance between the two lenses was adjusted to achieve different object vergences. The model eye/camera assembly consisted of a circular aperture simulating the entrance pupil, a model cornea, the test IOL, a glass window, a microscope objective, and a CCD camera. The 3-mm space between the model cornea and the glass window contained balanced salt solution (BSS) at ambient temperature. The model cornea had a focal length of 23 mm (43.5 D) and spherical aberration of +0.27 µm over the central 6-mm zone. These values were chosen because they approximate average values of measured human corneas. Four fixed aperture diameters (3, 4, 5, and 6 mm) were used.

**TABLE. CHARACTERISTICS OF THE PRESBYOPIA-CORRECTING INTRAOCULAR LENSES STUDIED IN THE MODEL EYE SYSTEM**

| | TYPE OF INTRAOCULAR LENS | | | | |
|---|---|---|---|---|---|
| **CHARACTERISTIC** | **Crystalens AO** | **Crystalens HD** | **ReSTOR +3.0** | **ReSTOR +4.0** | **Tecnis Multifocal** |
| Optic material | Silicone | Silicone | Acrylic | Acrylic | Acrylic |
| Optic diameter | 5.0 mm | 5.0 mm | 6.0 mm | 6.0 mm | 6.0 mm |
| Anterior optic design | Aberration-free aspheric | Bispheric | Apodized diffractive bifocal, Add = 3 D | Apodized diffractive bifocal, Add = 4 D | Modified prolate aspheric |
| Posterior optic design | Aberration-free aspheric | Spherical | Aspheric | Aspheric | Diffractive bifocal, Add = 4 D |
| Spherical aberration | 0 | N/A due to zone discontinuity at 1.5-mm radius | -0.1 µm | -0.1 µm | -0.27 µm |

N/A, not applicable.

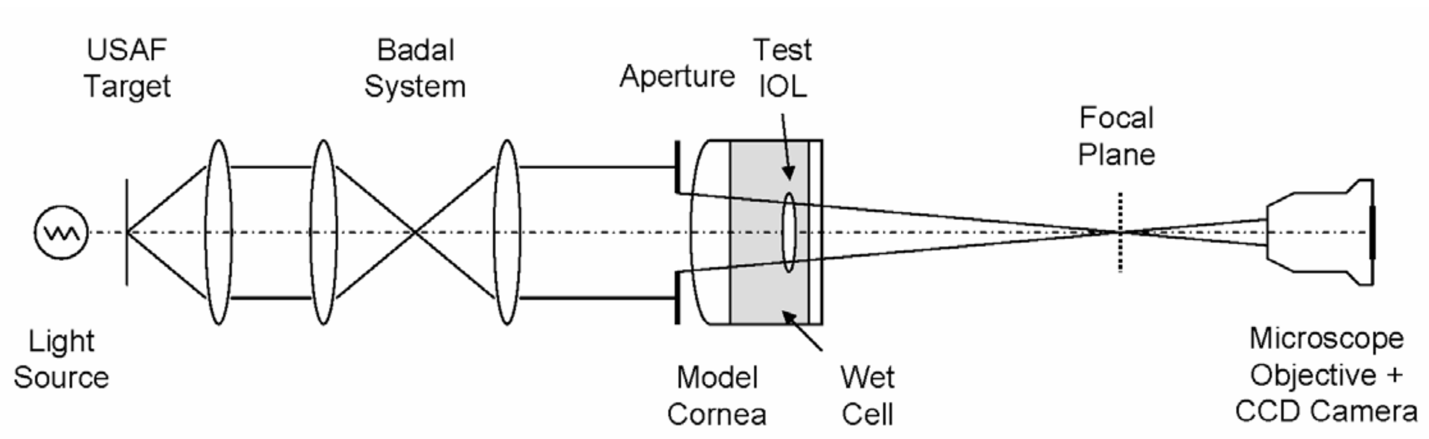


**FIGURE 4**

Schematic representation of the apparatus used to measure through-focus image sharpness of five presbyopia-correcting IOLs.

## MEASUREMENT PROCEDURE

The test IOL and fixed aperture were assembled together to ensure precise centration. Then the IOL/aperture assembly was accurately positioned and centered within the model eye. The space between the model cornea and glass window was filled with BSS (Alcon Laboratories, Fort Worth, Texas), and the model eye/IOL assembly was allowed to equilibrate for at least 1 hour. With the Badal assembly set for an object vergence of zero, the distance between the glass window and the camera assembly was adjusted to achieve best focus. The projected image of the USAF resolution target was captured for each object vergence between -1.0 D and +4.0 D in 0.125 D steps. The level of the light source was set such that each image had the same total intensity. Although the Crystalens models are accommodating lenses, the Crystalens lenses were measured in a static, nonaccommodated position for all object vergences. Once all images were captured for each combination of IOL model, aperture diameter, and object vergence, the sharpness score of each image was objectively determined and plotted vs object vergence.

## RESULTS

In Figure 5, the best-focus distance USAF target images for each IOL are shown for the 3-mm aperture. The AO has the highest image sharpness, followed by the HD. For a 3-mm pupil, the sharpness metric for AO at zero vergence (distance) position is close to twice that of the multifocal lenses (Figure 6). The multifocal IOLs are closely clustered, with TMF having the sharpest image, followed by R4 and R3.

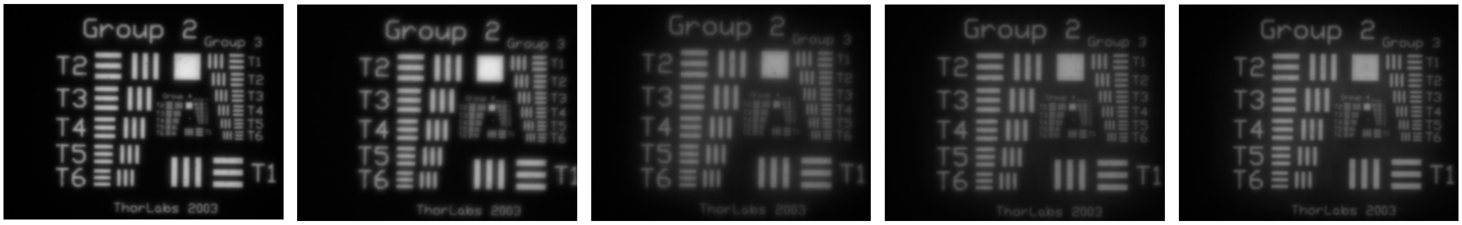


**FIGURE 5**

Digital image of the 1951 USAF resolution target imaged in a model eye with a 3-mm pupil through the following presbyopia-correcting IOLs, from left to right: AO, HD, R3, R4, and TMF.

The through-focus sharpness curves for the five IOLs are shown for each aperture diameter in Figures 6 through 9. The monofocal AO and HD lenses had a single peak in the through-focus sharpness curves, whereas the R3, R4, and TMF had two peaks. For all pupil diameters, the AO had the highest sharpness score for the zero-vergence position, followed by the HD. The depth of field of the HD was asymmetrically broader than that of the AO and was skewed in favor of positive defocus (ie, toward better near vision).

The three multifocal lenses had similar zero-vergence sharpness scores for all pupil diameters. The positions of the second peak for the multifocal lenses differed. The second peaks of the R4 and TMF were positioned about 3.25 D and 3 D away from the primary focus, respectively. The second peak of the lower-add R3 was positioned about 2.25 D away from the primary focus. The second peak of the nonapodized diffractive TMF lens was higher than those of the apodized ReSTOR lenses for all pupil apertures, consistent with a sharper near image metric. The through-focus sharpness curves for all tested lenses dampened as pupil diameter increased.

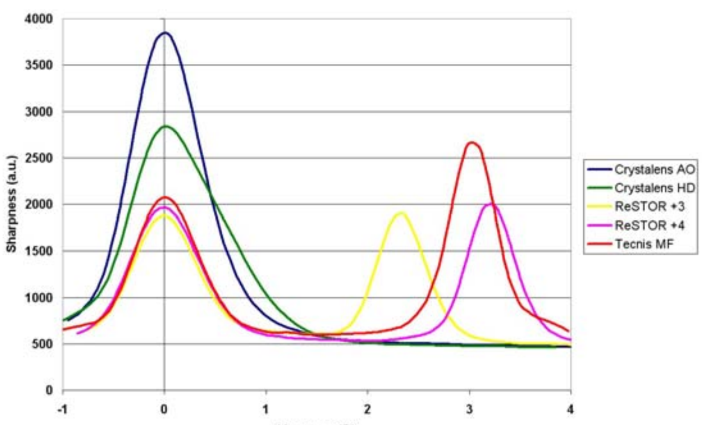


**FIGURE 6**

Through-focus image sharpness of the 1951 USAF resolution target imaged in a model eye through five presbyopia-correcting IOLs with a 3-mm pupil.

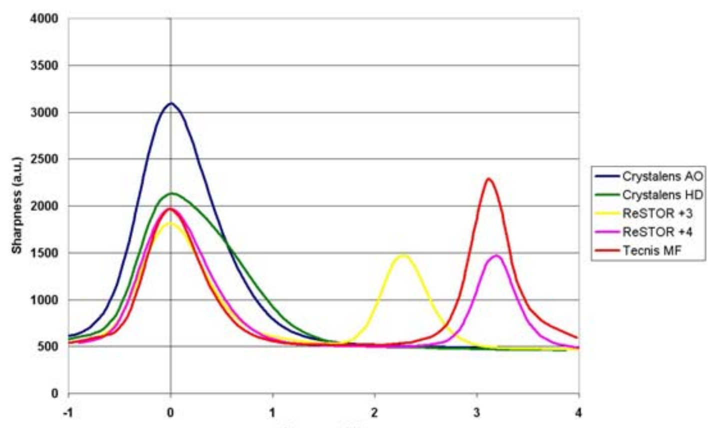


**FIGURE 7**

Through-focus image sharpness of the 1951 USAF resolution target imaged in a model eye through five presbyopia-correcting IOLs with a 4-mm pupil.

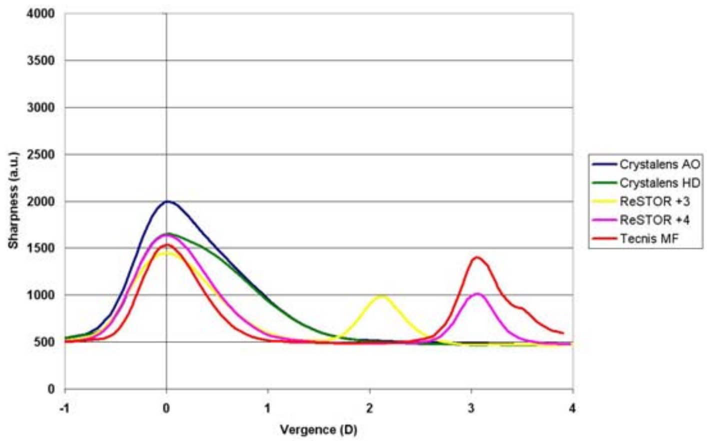


**FIGURE 8**

Through-focus image sharpness of the 1951 USAF resolution target imaged in a model eye through five presbyopia-correcting IOLs with a 5-mm pupil.

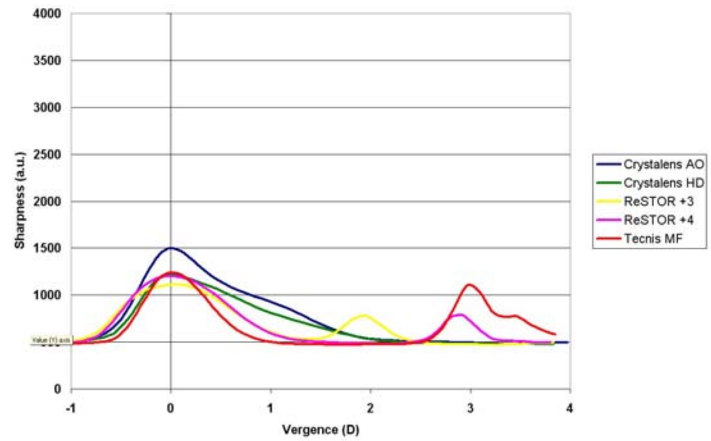


**FIGURE 9**

Through-focus image sharpness of the 1951 USAF resolution target imaged in a model eye through five presbyopia-correcting IOLs with a 6-mm pupil.

## DISCUSSION

Many factors can affect retinal image quality. These include image contrast, sharpness, brightness, optical scatter, and higher- and lower-order aberrations.[13] Differing IOL design strategies to increase depth of field are associated with quantifiable differences in image sharpness at varying vergences and pupil sizes. We compared one component of image quality—image sharpness[31,32]—at selected vergences and pupil sizes using an anatomically and optically accurate eye model, incorporating an average corneal power and spherical aberration representative of the cataract population. Although other studies of image quality through various IOLs incorporated in model eye systems have been reported,[40-42] to our knowledge this is the first investigation of through-focus image quality of presbyopia-correcting IOLs using a two-dimensional image sharpness gradient.

The multifocal IOLs achieve an extended depth of field by splitting light between two major energy foci. The Tecnis Multifocal lens splits the light evenly between near and far (41% each) at all pupil diameters while losing 18% of light to useless higher diffractive orders.[28] Although the simultaneous near and far retinal images reduce optical quality and image sharpness, the even distribution of light energy regardless of pupil size makes for better near image sharpness for all pupil diameters than with the ReSTOR lens (Figures 6 through 9). The ReSTOR IOLs utilize apodization to shift the amount of light predominantly toward distant foci with larger pupils and also show less light energy lost to higher diffractive orders with larger pupil aperture.[15,16] As an accommodating IOL, the Crystalens models do not split light among multiple foci[8,17] and have the highest distance image sharpness among the IOLs tested. Our through-focus sharpness curves show that it has a greater tolerance of distance image quality to defocus than any of the multifocal IOLs tested, but the eye model does not allow configuration changes to the IOL shape that may mimic the effects of accommodative effort.

Our study showed that the multifocal IOLs have two distinct peaks of image sharpness and basically perform primarily as bifocal IOLs. None of the three multifocal IOLs showed good image sharpness in the intermediate object vergences, although the near peak of the R3 borders the intermediate range. Although clinical studies show a mean bilateral distance-corrected intermediate vision around 20/40 with the Tecnis Multifocal IOL, this was worse in patients with larger pupils.[28] The reported mean binocular intermediate visual acuity[43,44] with the R3 was between 20/28 and 20/30. It is possible that intermediate vision may be influenced by pseudoaccommodating mechanisms not present in our eye model, as well as neural blur interpretation and other psychovisual mechanisms not measured by the sharpness metric alone. The positions of the second peaks (near) were different for ReSTOR and TMF models and corresponded with the add power of each. The secondary peak (near) was closer to the primary peak (distance) for R3 than for R4, because its inherent add was lower (3 D vs 4 D). The positions of the secondary peaks of the R4 and TMF were similar, because they both have a 4 D add. However, the secondary peak for TMF was a little closer to the primary peak, because the multifocal optic is located on the posterior surface and so the effective add is less. This difference in the distance best-corrected near focus between the TMF and R4 is confirmed in clinical studies.[28]

It is of interest that the introduction of the aspheric ReSTOR +3.0 successfully lengthened the focus of maximum near image sharpness to around +2.4 D at the image plane compared to around +3.2 D with the aspheric ReSTOR +4.0. This is confirmed in clinical studies where the preferred binocular near point was 38.4 to 40 cm vs 30.6 to 33 cm (equivalent to 2.5 to 2.6 D near point for the R3 and 3.0 to 3.2 D for the R4, respectively, at the spectacle plane).[43,44] However, by reducing the separation between distance and near foci, this closer superimposition of two images was associated with a decrease in both distance and near image sharpness for R3 compared to R4, as shown in Figures 6 through 9. Along these lines, it is interesting that studies have shown that the modulation transfer function cutoff and point spread function width as measured with a dual-pass system, along with uncorrected distance vision and distance-corrected near vision of patients implanted with the R4 lens, were worse with higher dioptric powers than low.[45] Petermeier and colleagues[46] also found that the preoperative refractive status had an influence on the postoperative outcomes of the ReSTOR IOL. This is again consistent with the notion that near and far image foci are more separated in eyes with a low-power IOL than higher power, reducing image overlap and degradation. For every 0.25 D that the two foci are brought closer together, the intensity or brightness of the out-of-focus image increases by 18%. So, the intensity of the out-of-focus image with ReSTOR +3.0 is approximately 72% brighter (and hence potentially more bothersome) than that of the ReSTOR +4.0.

Numerous studies[47,48] show that the mean corneal spherical aberration is around +0.27 µm at a 6-mm zone, ranging from +0.055 to +0.57 µm. Aspheric IOLs have been developed to fully or partially offset the positive corneal spherical aberration or to be aberration-neutral in an effort to improve image quality and contrast sensitivity. Given the location of the IOL posterior to the iris, ray tracing demonstrates that offsetting the spherical aberration at a 6-mm corneal optical zone corresponds to around a 5-mm pupil. Our study showed that the through-focus sharpness curves diminished for all lenses as pupil diameter increased, due to the increasing impact of corneal spherical aberration and IOL aberrations with larger pupil apertures.

There is a balance between enhancing image quality (by reducing whole-eye spherical aberration) and reducing depth of field and tolerance to defocus.[13] The earlier FDA-approved multifocal IOLs (ie, AMO Array, and ReZoom and spherical ReSTOR +4.0) have positive spherical aberration, which increases in magnitude with IOL power.[15] The Crystalens HD, similarly, has inherent positive spherical aberration that increases with IOL power and would add to the corneal positive spherical aberration, reducing image quality while expanding depth of field. The central bispheric modification of the HD allows for further expansion of depth of field by adding effective add power with pupil constriction below a 3-mm aperture.[49] This particular lens design increases the tolerance to defocus in an asymmetric manner, skewed in the direction of positive defocus.

Newer-generation accommodating and multifocal IOLs have become aspheric in design, thereby enhancing image quality. There is a higher image sharpness score for the aspheric ReSTOR +4.0 in comparison to the spherical ReSTOR +4.0 at zero vergence (data not shown) and a higher sharpness metric for the Crystalens AO vs the Crystalens HD at all pupils tested (Figures 6 through 9). The Crystalens AO has zero aberration, which makes it immune to the effects of IOL decentration with respect to the visual axis and less sensitive to tilt.[50] It neither adds to nor subtracts from the inherent corneal spherical aberration. While this small, variable amount of residual whole-eye spherical aberration may have resulted in a slight decrease in image quality when compared to total offset (eg, Tecnis Multifocal; -0.27 µm spherical aberration) or partial offset (aspheric ReSTOR +3.0 and +4.0; -0.1 µm spherical aberration) of average corneal aberration, it serves to increase depth of field,[50] helps to offset chromatic aberration[51] and residual hyperopia, and may mitigate the effects of some other higher-order aberrations.[52]

There are a number of limitations to this study. First, we did not explore the effect of different corneal powers, ranges of spherical aberration, or other symmetric or asymmetric corneal aberrations on the performance of these IOLs. These can be a source of retinal image degradation in addition to the IOL.[53] This would be an area for future investigation.

Second, there may be pseudoaccommodative mechanisms not incorporated or reflected in this model. Patients' pupils may not be round and may have various dynamic ranges that may require further analysis in the eye model. In addition, we did not investigate the effects of IOL tilt or decentration with respect to the visual axis, or the effect of angle kappa, on image quality using these IOLs in the eye model. The pseudophakic eye is not a centered optical system.[54] Numerous studies have shown that significant decentration of the IOL from the visual axis averages around 0.5 mm, which can impact image quality through various lens designs.[52-54]

Finally, as discussed earlier, IOL power may also influence IOL performance, particularly with multifocal IOLs, where higher IOL dioptric powers may have closer near and far focal points than lower, impacting retinal image quality.[45,46]

A number of clinical conditions that can affect IOL performance were not incorporated into this model. For example, we did not model the potential effects of dry eye or other ocular surface diseases in this model, which can have a major impact on image quality given the large difference in index of refraction between air and tears. Similarly, the impact of posterior capsular fibrosis can lead to light scatter and affect the retinal point spread function.

A final, important difference between this optical bench eye model and the clinical situation is that of neural processing.[13] While our studies showed that different presbyopia-correcting IOLs have differences in image sharpness at varying vergences and pupil apertures, this does not take into account the neural processing of the retinal image, nor the ability to potentially influence neural image processing by visual training, which has been demonstrated in patients with some multifocal and accommodating IOLs, improving both contrast sensitivity and near visual acuity.[55,56] Finally, image sharpness is just one of many metrics that reflect overall image quality, and further studies will be required to see how this compares to subjective scores of image quality in the clinical setting. Despite these shortcomings, the eye model is useful in demonstrating many of the effects of variations in pupil aperture and object vergence on simulated retinal image quality. With additional modifications, the eye model could be helpful in predicting important aspects of IOL performance within a range of pupil sizes and shapes, object vergences, and other ocular characteristics that can be quantified and modeled.

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