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Pre-clinical Estimation of the Intraocular Lens A-Constant, and its Relationship to Power, Shape Factor, and Asphericity

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Calculating the intraocular lens power for a particular patient requires an empirical "lens constant" to estimate the final axial location after surgery. This is normally calculated from clinical results for each new lens style, but it can also be estimated without clinical data by comparing a new style to an existing style. The lenses are axially positioned in a model eye at comparable locations, and image distances are used to estimate the change in lens constant. The A-constant that is used by the SRK/T calculation method is evaluated here, but this can be easily converted for other calculations using an average eye. Raytrace calculations demonstrate the method, and also illustrate the effects that refractive index, shape factor, and asphericity have on the refractive error. Actual lens measurements at 35^{0}C in saline are preferable if details of the reference lens are uncertain.

1. INTRODUCTION

Intraocular lenses (IOLs) are used to replace the natural crystalline lens of the eye during cataract surgery, with over 3.5 million lenses implanted annually in the US alone, and millions more worldwide. The IOL power that is needed for a particular eye to give a targeted postoperative refraction has to be calculated using preoperative measurements of the eye, with corneal power and axial length being the primary parameters. These are not known with absolute accuracy, but the parameter that is the most difficult to estimate is the postoperative location of the IOL. Calculation methods that date back 30 years have estimated this using back-calculations from clinical data, using a simple model for the eye. The situation is illustrated in Figure 1 for an average eye, where the crystalline lens is replaced by a very thin IOL, and the traditional power calculations model the eye using thin lenses for the cornea and IOL. One particular calculation method is used here, the SRK/T calculation [1,2], but the "A-constant" parameter that is used to estimate the IOL location can be converted for use with other power calculation methods that use other "lens constants" [2–11]. The rationale for using the SRK/T calculation is that it is widely used, the equations are published, and it is usually included as a reference calculation when alternative methods are being evaluated. Newer calculation methods show improvements, but these are relatively modest, with the greatest improvements being for long and short eyes.

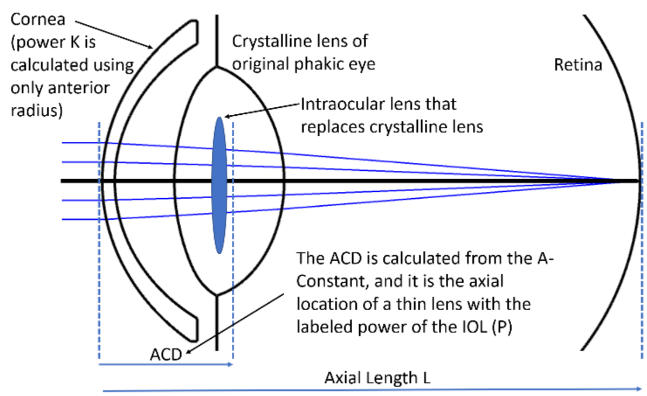

Fig. 1. Sketch of 0.7 mm thick IOL overlaid on an average phakic 70 year old eye with a 4.8 mm thick crystalline lens. A paraxial thin lens model for the eye is depicted by the vertical dotted lines. Only the ACD distance is unknown. An average value that is determined empirically from clinical data is used as an offset when calculating IOL powers for new patients.

The purpose of this paper is not to evaluate the power calculation method itself, or to explore how accurate it is over a range of IOL powers, but it is primarily about estimating the lens constant before any IOLs have been implanted, and exploring how optical design variables affect the value. There is a single lens constant for an entire range of IOL powers that has a particular design concept, and only an average eye needs to be evaluated. The evaluation is further simplified by using an object at infinity, even though in clinical practice an IOL may be chosen so that the best focus of the eye is at a closer distance. The SRK/T calculation determines the anterior chamber depth (ACD) as part of the calculation process from an empirical "A-Constant", which uses a set of equations that involve the physical sag and width of the cornea. The A-Constant terminology goes back to the earlier SRK equation that was simply IOL Power = A – 2.5*L -0.9*K, which came from a correlation, and not an optical model for the eye. The SRK/T A-Constant is slightly different to the value for that older equation, but an example A-Constant of 119 might lead to an ACD value of 5.5 mm, though this is hidden within the calculation method. And a user would not normally see it The ACD for this thin lens model includes the thickness of the cornea for IOL power calculation because the power acts at approximately the anterior corneal surface. The term ACD is sometimes used elsewhere to mean the distance from the posterior corneal surface to the anterior lens surface, and this is not always clear in publications (though aqueous depth can also be used to describe that distance).

The A-constant is expected to be determined empirically by analyzing the postoperative results for a series of patients, and it may also need to be "personalized" to address unique details of the overall process. Primarily, however, the A-constant relates the physical postoperative location of the lens component of the overall IOL to the effect that it has on the final refraction. If the A-constant is already known for one lens style, then it can be estimated for another one, which is particularly useful for pre-clinical work on a new IOL style. Many new IOLs are evaluated in the clinic every year, with the continuous development of various designs for monofocal, toric, multifocal, trifocal, and extended depth of focus concepts. This evaluation also provides insight into the effects that the "shape factor" and asphericity of the IOLs have on the final refraction.

Ophthalmology has a long history of empirical adjustments in response to clinical data, with the aim of minimizing the postoperative refractive error for various procedures. The calculations are sometimes set up as "nomograms" for other situations, and are only created after clinical data are available. This was also necessary initially with IOLs because there were many things that were unknown, but over time, the development of phacoemulsification and small incision surgery removed many of the sources of variability, and improvements in measurement instrumentation reduced the effect of other parameters. Early multi-piece IOLs also contributed to the uncertainty in refractive error, because they were made from rigid PMMA, with a variety of angulated haptic designs. IOLs are now primarily foldable, and many are planar, and haptic forces are now more consistent with the strength of the capsular bag.

The postoperative axial location of the physical lens is the greatest unknown, and although it was initially difficult to predict, modern IOLs are primarily constrained by the lens capsule [12–14]. With planar haptics this is straightforward to envisage, but even with multipiece haptics, if they are made from PMMA then they initially support and center the lens after implantation, but then thermally reform over a few hours so that it is the capsule that determines the final location of the IOL, and not the haptics. With angulated haptics made from other materials, or lens styles that have an offset between the haptics and the optic, particularly if the mechanical forces are unusually large, then additional judgement may be needed. In many cases though, envisaging the capsule collapsing down around the entire IOL, with an anterior capsulotomy, and with the capsule stretched flat by the zonules, an estimate can be made of the physical location of a new lens style in comparison to an existing lens style. This allows an older IOL style to be replaced in a model eye with the new lens style, at the best estimate for the equivalent physical location, and then the labeled power that the new IOL would require to image at the same focus can be determined.

Estimating the physical IOL location is an important step, but the A-constant also compensates for many other parameters. One parameter is the shape factor, which alters the effective plane where the IOL power appears to act. Another is the lens asphericity, which causes an optical shift of the focus because of aberrations. These factors are evident in the design, but many other variables are also addressed automatically if physical lenses are measured, such as any small systematic differences in power values that are a consequence of the test methods. One detail that may not be widely known is that although the refractive index of aqueous is standardized to be 1.336 at 546 nm and 35°C, this is not actually the refractive index of balanced salt solution (BSS) under those conditions, which is closer to 1.3345 (determined from an estimate by Pearson [15] of 1.3362 at 546nm at 20C, and a change with temperature of -0.0001 /°C). The IOL standards also refer to *in situ* conditions, in addition to specifying the 1.336 value. "Asphericity" and "spherical aberration" are also mentioned in the standards, but there is no discussion about how these might alter the effective lens power. IOL power measurements would also not normally be made at 35°C during manufacturing because that adds complexity and potential inaccuracy to the measurement, and power is also both defined and measured in collimated light, even though IOLs are always used in converging light. These small details are never an issue clinically, because it is never power alone that is used to specify the choice of IOL power, with the clinical A-constant (or equivalent lens constant for other power calculation methods) always being used to make any compensation that is necessary.

To estimate a new A-constant, the best method would be to make measurements using a model eye at body temperature, with the IOL fully hydrated in saline, in order to include the effects of as many of the unknown parameters as possible, since that situation is as close as possible to that of the eye itself. This paper uses calculations instead, where all the details of the different IOLs are perfectly known, in order to evaluate the method using known parameters. The evaluations also provide insights into the relationship between the IOL design and the lens constant.

2. METHOD

An emmetropic pseudophakic eye is modeled by the ISO2 eye model only for the region of the IOL itself and beyond [16](Figure 2(a)), but this can be used to measure the change in refractive effect for different IOLs because the image distance moves. Differences in focus for either the ISO2 model eye, or the SRK/T calculation method, are proportional. The "A-constant" is needed in order to position the IOL at the best estimate for its location in the eye, and it is essentially optimized clinically so that the largest number of patients have the smallest difference from their targeted refractive result. The ISO2 model eye is also emmetropic, which is a useful simplification for this modeling, and there is no need to include a targeted refractive error.

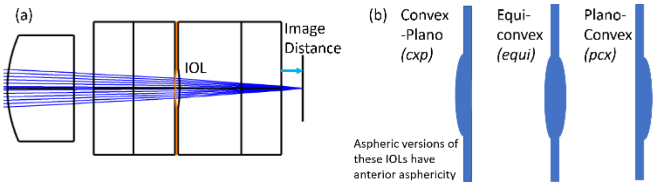

Fig. 2. (a) Image distances for the ISO2 model eye characterize A-Constant differences between lens styles. (b) Sketches of conceptual planar single-piece IOLs with three shape factors.

Different IOL styles can be compared using calculations with the schematic eye in Figure 2(a), and the same method can be used with measurements of actual lenses (with several lens powers being measured in order to average out the effect of the power tolerance). Different levels of corneal asphericity can be simulated for different ISO2 eye models by modifying the anterior surface of the front lens, which is made from PMMA. IOLs with different powers will image at different distances beyond the final window of the model eye, and if two IOLs that are labeled with the same power image at different distances when the IOLs are positioned appropriately, then the difference in the image distances can be used to adjust the A-constant for a lens that has no clinical data.

Conceptual lens designs were created using two different refractive index values of 1.55 and 1.46, with 3 types of optical design for each of the shape factors. One set of lenses has spherical surfaces, where the only criterion is that the power calculated using the thick lens equation matches the labeled power (where the nominally flat surfaces of the example designs actually have large radius values of 1000 mm, to simplify evaluations). Another set started with the same designs, but the design was optimized to have an aspheric anterior surface when positioned in the ISO2 model eye with a spherical aberration (SA) value of 0 microns with the anterior apex at the nominal position of 6.25 mm in the wet cell. A third set was also designed to be aspheric, but with a spherical aberration value of 0.3 microns. The SA terminology is described in the ISO 11919-2 standard [16], where the value is given as the Zernike spherical aberration value in microns for a 6 mm entrance pupil of a real eye. The coefficients represent the aberration of the eye, and a positive sign is used, with the IOL asphericity acting in the opposite direction to correct the aberration. The values of 0 microns and 0.3 microns SA are at the extremes of the typical values commercially available for aspheric IOLs. The Zemax raytrace software was used to optimize the designs over a 5 mm diameter region (Zemax, Kirkland, WA), using the coefficients for the r4 and r6 terms of the standard Zemax even asphere polynomial. Five lens powers were created for each set of variables (19, 20, 21, 22, and 23 diopters (D)), and the designs for the 21 D and 23 D powers are listed in Table 1. The labeled power was always calculated using the thick lens equation and the radius values, ignoring the asphericity.

The image distance for the model eye was then calculated for each IOL using Zemax, with the IOLs all positioned with the haptics at the same location. This assumes that the haptic location will be the main characteristic that positions these particular IOLs in the axial direction, though for other haptic types additional considerations may be needed. Calculations were done for model eyes with two different asphericity values, with SA = 0 microns and SA = 0.3 microns. The aspheric IOL designs correct the aberrations of either one or the other of the model eyes, but the spherical IOLs do not match either. A 3mm diameter illumination region at the anterior IOL surface was used, and the image distance was optimized to give the best modulation transfer function (MTF) for a 50 lp/mm spatial frequency. The image distance for the simple equiconvex lens was used as the reference value when evaluating the results. Defocus values were also rescaled to approximate diopters at the IOL using the image distance values calculated for the equibiconvex spherical lenses.

3. RESULTS

Figure 2 plots the distances to the image from the posterior window of the ISO2 eye model with SA = 0 microns, for the two different IOL material refractive index values. When this method is used to generate a pre-clinical A-constant, there would normally be just two curves, one for the reference IOL whose clinical A-constant is known, and the other for an IOL for which there are no clinical data. The approximate power change at the IOL, relative to the reference lens, is given at the right, and purely by chance the reference equi-biconvex spherical IOL approximately lines up with the gridlines (solid green line) for the 21 D IOL. The plot shows that if a plano-convex (pcx) aspheric SA = 0 microns IOL was used instead (dotted blue line), then for the image to be in focus at the same distance, the IOL power would have to be about 22.25 D rather than 21 D.

The relationship between the effective power change and the SRK/T A-constant for an average eye with an axial length of 23.5 mm and a corneal power of 43.7 D (using the 1.3375 keratometer constant) is plotted in Figure 4, using the published equations for the SRK/T calculation method [1]. A simple linear fit indicates that 0.8 of the power difference can be used to modify the A-constant, though a quadratic curve is a slightly better fit to the data. If the reference lens has a clinical A-constant value of 119.0, then the new lens for the example above would have a value of 120.0 (119 + 0.8*1.25). The equivalent calculation for the original SRK equation is also plotted in Figure 4, but this is solely for historical interest, and the SRK calculation is no longer recommended. That equation was based on a regression calculation during a period where anterior chamber lenses and contact ultrasound were used, and although it gave good results for central IOL powers for many years, more modern equations provide much better outcomes.

The main part of the power difference for the plano-convex example is due to the shape factor change, which moves the power of the lens to the posterior surface. This is physically closer to the retina, and the lens power has to be increased in order to image at a shorter distance. If asphericity was then also added, it would have an additional effect, because a simple paraxial plano-convex lens would have a lot of spherical aberration in this configuration, making a spherical IOL appear to be more powerful than it is, causing a lens with a weaker labeled power to be needed. The asphericity corrects the underlying IOL aberration, and with an aspheric IOL a higher labeled power would be required.

Table 1. Conceptual IOL designs, for 2 of the 5 powers in each group. D=labeled power in diopters. Equi = equibiconvex. Sph= spherical. Asph =aspheric on anterior surface. Cxp=convex-plano. Pcx = plano-convex. SA0 = SA 0 microns. SA0.3 = SA 0.3 microns. R1 = anterior radius. R2 = posterior radius. Ct = center thickness. The aspheric terms are in mm.

Design	Index	D	r1 mm	A4* 10^6	A6* 10^6	Ct mm	r2 mm
equi sph	1.55	21	20.333	0	0	0.695	-20.333
equi sph	1.55	23	18.558	0	0	0.738	-18.558
cxp sph	1.55	21	10.294	0	0	0.701	-1000
cxp sph	1.55	23	9.391	0	0	0.747	-1000
pcx sph	1.55	21	1000	0	0	0.701	-10.294
pcx sph	1.55	23	1000	0	0	0.747	-9.391
equi asph SA0	1.55	21	20.334	-151.5	-0.8	0.682	-20.334
equi asph SA0	1.55	23	18.559	-174.8	-1.1	0.723	-18.559
cxp asph SA0	1.55	21	10.294	-43.0	-0.5	0.698	-1000
cxp asph SA0	1.55	23	9.391	-50.6	-0.7	0.742	-1000
pcx asph SA0	1.55	21	1000	-403.6	-1.7	0.667	-10.294
pcx asph SA0	1.55	23	1000	-488.6	-2.6	0.705	-9.391
equi asph SA0.3	1.55	21	20.336	-553.8	-6.0	0.646	-20.336
equi asph SA0.3	1.55	23	18.561	-576.2	-6.5	0.687	-18.561
cxp asph SA0.3	1.55	21	10.294	-431.5	-8.1	0.661	-1000
cxp asph SA0.3	1.55	23	9.391	-436.3	-8.7	0.705	-1000
pcx asph SA0.3	1.55	21	1000	-820.8	-4.5	0.632	-10.294
pcx asph SA0.3	1.55	23	1000	-906.5	-5.3	0.669	-9.391
equi sph	1.46	21	11.766	0	0	1.028	-11.766
equi sph	1.46	23	10.736	0	0	1.105	-10.736
cxp sph	1.46	21	5.940	0	0	1.068	-1000
cxp sph	1.46	23	5.420	0	0	1.160	-1000
pcx sph	1.46	21	1000	0	0	1.068	-5.940
pcx sph	1.46	23	1000	0	0	1.160	-5.420
equi asph SA0	1.46	21	11.767	-353.1	-5.0	0.996	-11.767
equi asph SA0	1.46	23	10.737	-420.6	-6.9	1.066	-10.737
cxp asph SA0	1.46	21	5.940	-214.8	-6.5	1.046	-1000
cxp asph SA0	1.46	23	5.420	-295.3	-11.1	1.129	-1000
pcx asph SA0	1.46	21	1000	-1300	-22.1	0.946	-5.940
pcx asph SA0	1.46	23	1000	-1607	-33.8	1.006	-5.420
equi asph SA0.3	1.46	21	11.770	-1038	-16.1	0.932	-11.770
equi asph SA0.3	1.46	23	10.740	-1103	-18.6	1.002	-10.740
cxp asph SA0.3	1.46	21	5.940	-853.5	-25.7	0.980	-1000
cxp asph SA0.3	1.46	23	5.420	-925.3	-31.7	1.062	-1000
pcx asph SA0.3	1.46	21	1000	-2037	-24.7	0.885	-5.940
pcx asph SA0.3	1.46	23	1000	-2349	-35.7	0.944	-5.420

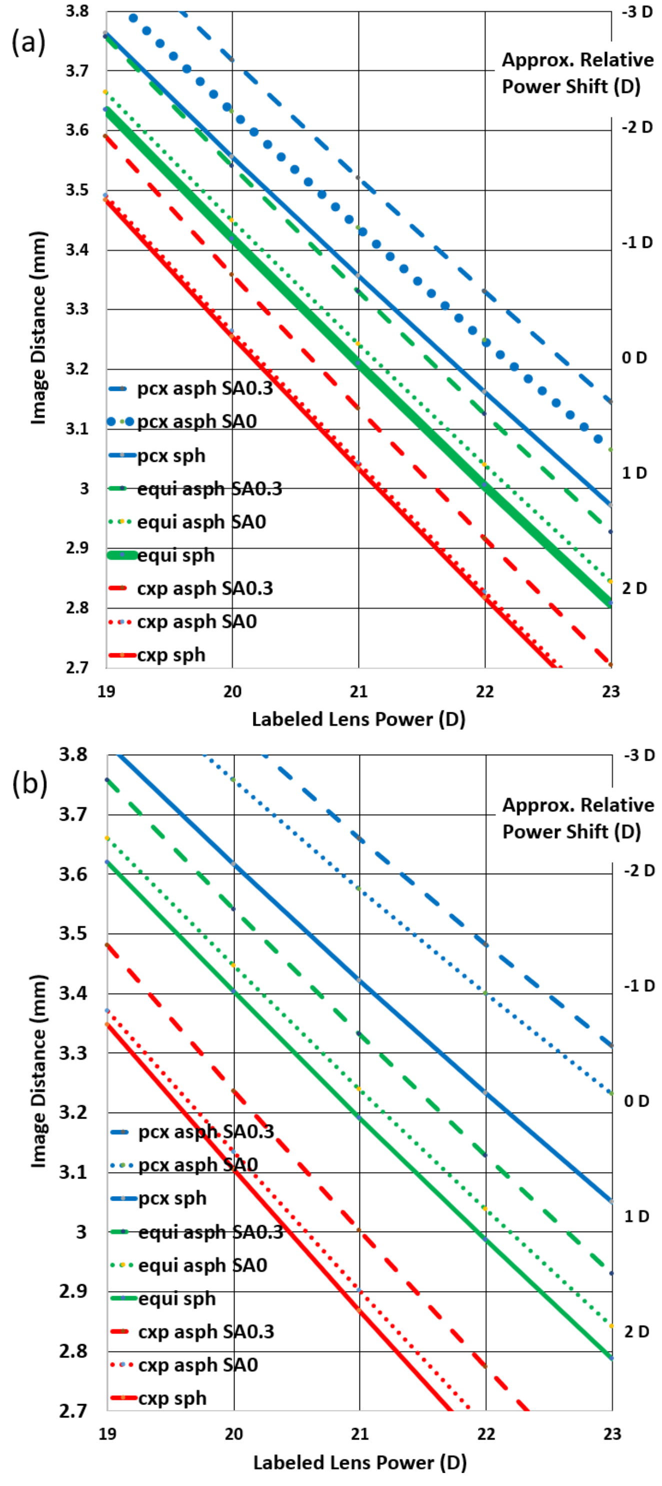

Fig. 3. Image distance values using the SA = 0 microns ISO2 model eye for refractive index values of (a) 1.55, and (b) 1.46. The approximate change in power value at the IOL is given at the right.

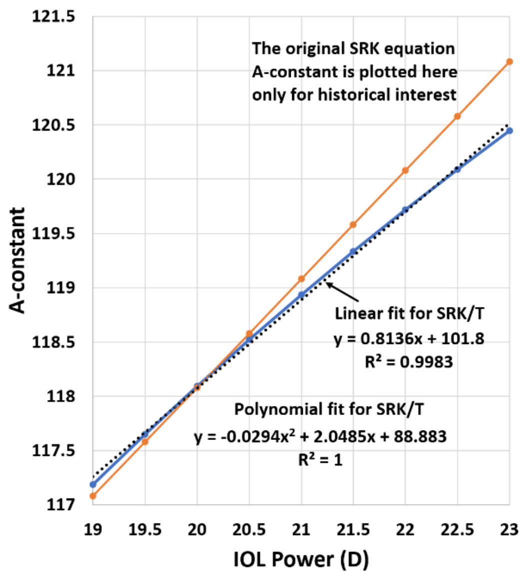

Fig. 4. The relationship between the A-constant and IOL power for an emmetropic eye with an axial length of 23.5 mm and a corneal power of 43.7 D (blue curve). The orange line plots the equivalent A-constant for the original SRK equation (P=A- 2.5*L – 0.9*K).

Figure 5 plots the same data as Figure 3, but with the image distance for the spherical surface design subtracted. The plots show that there is a small trend that is related to the shape factor of the IOL. The change in IOL power here corresponds to a change in axial length, and the calculation method should compensate for the trend, but this evaluation will be the most accurate for a model that represents an average eye. With both sets of plots, the effects are larger for the IOLs with the lower refractive index, which for these examples are also thicker because both optical surfaces extend to a 6 mm diameter (though this is not necessarily the case for IOLs that are routinely used).

When the other ISO2 model eye with SA = 0.3 microns was used instead for all the same lenses, the results were almost identical, relative to their own reference (with image distances of 3.634 mm and 3.547 mm for the SA = 0 micron and 0.3 micron eyes respectively). Changing the cornea asphericity refocused the eye, so the image distance for the reference lens changed, but the relative focus distances were broadly similar for all the IOLs, with differences less than 0.005 mm. This indicates that either of these model eyes could be used for the evaluation, though the values all have to be measured using the same model. The original ISO1 model eye could also be used, which is the same as the ISO2 model when SA = 0 microns, using a glass doublet instead of a PMMA lens.

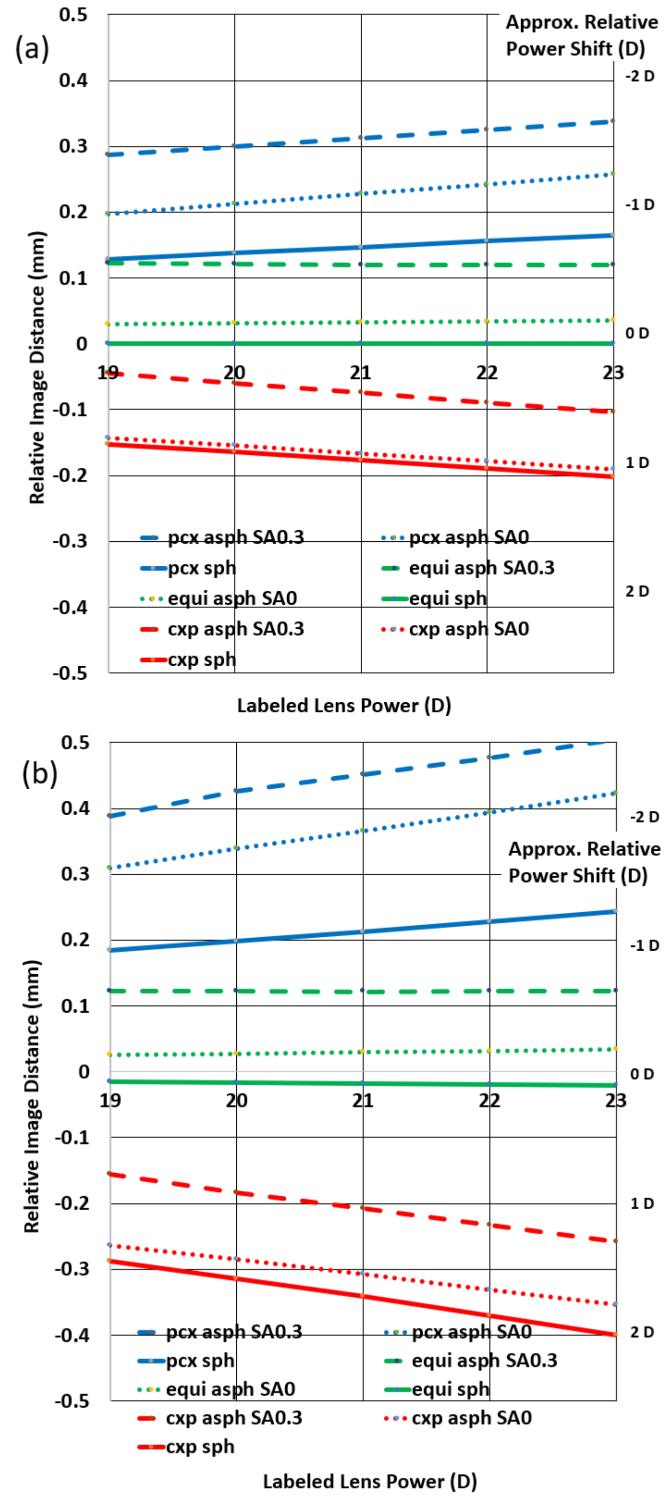

Fig. 5. Image distance values using the SA = 0 microns model eye with the value for the equi-biconvex IOL subtracted, for refractive index values of (a) 1.55, and (b) 1.46. The approximate power value change at the IOL is given at the right.

4. DISCUSSION

Calculations using the ISO eye model (either ISO1 or ISO2) show the effect that different IOL styles have on defocus, and this is the same type of information that is generated when a clinical A-constant is calculated. This evaluation could be performed before any IOLs are implanted, as part of a preclinical evaluation of a new lens style. The method is described here with theoretical calculations, but the most valuable use of the method would be with actual measured data for IOLs that have been fully hydrated in saline, with the measurements made at 35°C, in order to also address any unexpected variables. The defocus results should then be as similar as possible to the clinical values, with the main limitation being the estimate for the IOL location. It is envisaged here that the capsule of the original crystalline lens shrinks down around the IOL while also stretching it flat, and this concept has to be applied to the designs of both the reference lens and the new lens to estimate the axial locations. This estimation could be improved by comparing postoperative measurements of physical IOL optic locations for different existing lens styles, to obtain a fuller picture of how IOL styles with different designs interact with the capsular bag. The final mechanical location of the IOL optic can be evaluated independently from the refractive result.

Measurements for very few physical IOLs would be needed, because manufacturing variations would be distributed across the different powers, and it is the trend across powers that is used. An average eye is envisaged, and although the ISO eye models capture the general features, the parameters are not perfectly known. The eye model with SA = 0 microns has been used as the main example, because it is probably the one that is most widely available. The axial IOL location used here is one that is listed in the ISO standard, but it is probably slightly more anterior than the corresponding location in the actual eye (though this is expected to have a modest effect because of the relative nature of the evaluation).

There are several reasons to prefer measurements of real IOLs, rather than to use only calculations. One is simply that the design of a reference IOL may not be known, since this type of information is rarely published (though the clinical lens constant should be available). There are also many factors that might affect the image distance of an actual lens, particularly ones relating to characteristics of the manufacturing and testing methods that are used. Manufacturers label lens powers with consistent values, but there may be details that are not perfectly known, and the lens constant compensates for these. It should also be possible to combine information from different lens styles to provide the reference for a new lens style, because the method is relating labeled power, a clinical A-constant that is matched to the labeled power, and the defocus of the image when the IOL is positioned appropriately. The evaluation here also demonstrates that the IOL power calculation methods are tolerant to many variables through the use of a clinical A-constant that adjusts for all unknown factors. The labeled powers of all the lenses could actually be systematically shifted by a small amount, yet with an adjustment to the lens constant a similar refractive result could be achieved.

The calculations also show the sensitivity of IOL power calculation to the shape factor and asphericity of the lens designs, with the effective power changing over a range of 2 D. The shape factor effect is due to IOLs being labeled with their power in collimated light, rather than with the "back vertex power" that is used for spectacles and contact lenses. This moves the principal plane in comparison to the physical lens (though the principal plane is not actually used for the traditional IOL power calculation methods, and an ACD parameter, or "effective lens position", is actually a complicated back-calculation for all the effects that influence the final refraction). The asphericity data here also give an indication of how the refractive error might change for different actual corneas, with differences corresponding to about 0.5 D for a 3mm actual pupil diameter. With larger pupil diameters the focal shift due to aberrations may increase. The 3mm value is the illuminated diameter at the IOL, and the entrance pupil is about 15 % larger than this, so the apparent pupil measured when looking at the patient would be about 3.5 mm diameter (though this distinction is not always clear in publications). The power that is discussed here is also the actual power effect at the IOL, and the corresponding refractive error at the cornea would be only about 0.7 times this value, because the IOL is not located at the cornea.

The calculations are consistent with published A-constant values for two lens styles that are said to only differ by the addition of asphericity [17]. The A-constant increase of 0.7 in that publication is comparable to the power change in Fig. 3(b) for a lower index lens when a change is made from a spherical design to an aspheric design. The asphericity weakens the focusing effect, and a higher labeled power is needed.

Discussions of IOL power calculation still often refer to a need for "personalization", which is always useful, and "optimization" is needed anyway in order to determine the optimal lens constants for a set of clinical data [3,6]. In practice, equipment and methods are now broadly similar from place to place, and lens constants that characterize the general clinical experience have the most value [18]. One characteristic that can have a large effect on the lens constant is the definition for corneal power, where K values can vary by 1 D or more for the same eye because of different keratometer index values [5,19]. This is handled automatically when calculations are performed on the measurement instrument, but care must be taken when sharing K values between different devices. Another characteristic is the test distance that is used for the postoperative refraction, and refractive errors should normally be converted to a value at 6 meters for a standardized lens constant [18,20,21]. This topic has been confused somewhat by formal clinical studies where specialized testing is sometimes performed at 4 meters, with some refraction values then being adjusted to true infinity. There can be uncertainties of 0.4 D if care is not taken to convert values to the 6 meter reference distance for IOL power calculations.

The modeling here is for a circularly symmetric eye, even though the real eye has tilted and decentered components, but that is also the case for the IOL power calculation methods. For both the calculation and laboratory methods the evaluation is relative, comparing one lens to another, rather than making an absolute measurement. This is similar to the use of the A-constant anyway, where not all the details that affect the clinical results are known. Despite this, the A-constant is very sensitive to the average refractive result, and although a single patient may only be refracted to 0.25 D, the A-constant is back-calculated as an average for many patients, and it can be accurate to a small fraction of a diopter.

Overall, an accurate lens constant is as important as consistent IOL power labeling in order to achieve the best refractive result, and the lens constant adjusts for many small factors that affect the final refraction. The methods described here could be used to provide a provisional value for the lens constant of a new IOL style, with the main limitation being the estimation of the relative physical axial location for the new lens in comparison to the reference lens. A cross-check on the method itself would be to compare the results for known lenses to their clinical A-constants.

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