Current Limitations of IOL Power Calculations

Accuracy in eyes with refractive errors can be improved only with methods that measure physical parameters, not just statistical ones.

BY PAUL-ROLF PREUSSNER, MD, PHD

Patient expectations regarding refractive outcomes after cataract surgery have continually increased over the past several years. However, routinely achieving accuracy with implanted lenses equivalent to that achieved with spectacles or contact lenses is not yet possible. Currently, the most popular approximative IOL power calculation formulas using Gaussian optics are capable of producing zero average prediction errors for a series of patients; however, these formulas partly disguise the physical background of their sources of error by relying more on statistics than on physics.

Statistics can optimize average results, but it cannot adequately address outliers caused by eyes that differ too much from the average in their physical parameters. Unfortunately, a single patient with a 3.00 D hyperopic outcome can cause a surgeon more trouble (including possible legal disputes) than an average outcome in 1,000 patients that is 0.20 D worse compared with another surgeon.

Therefore, it is worthwhile to consider in detail the current limitations and sources of error in IOL power calculation formulas from a physical point of view rather than a statistical one. Most single error sources can be clearly separated from each other. Some, however, are more intertwined with others, making the analysis more complicated. This article attempts to unravel some of the complexity.

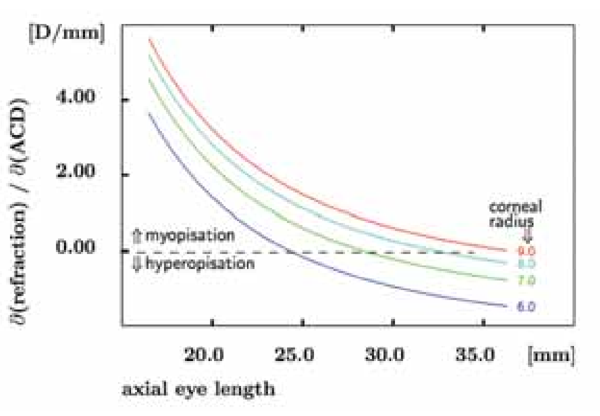
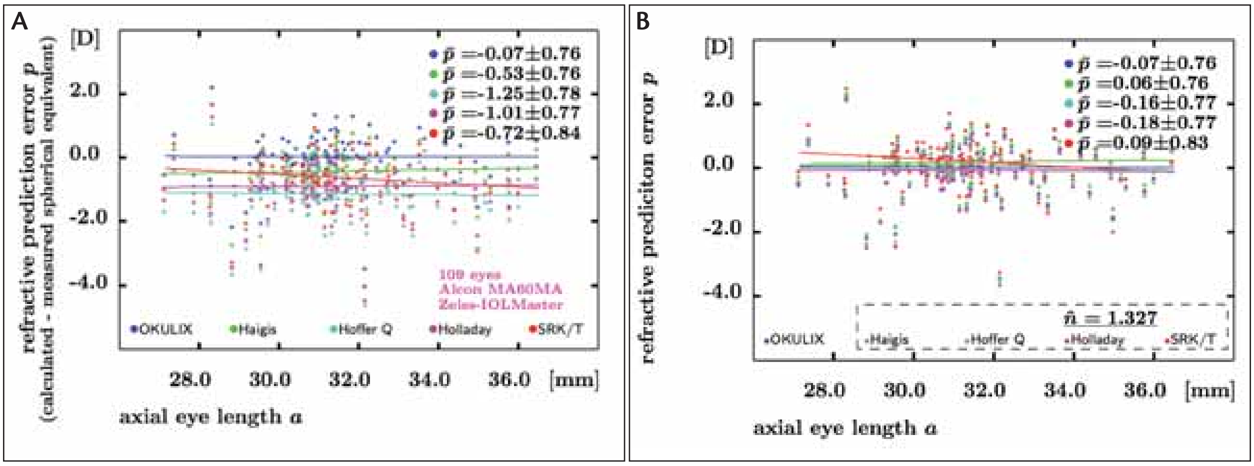


Figure 1. Impact of IOL position on refraction. For each pair of axial length and corneal radii, the IOL power to achieve emmetropia is calculated. A shift of this emmetropic IOL by 1 mm in the forward direction causes a change in refraction as indicated on the y-axis. Note that the emmetropic IOL power is positive above and negative below the dashed line. *(ACD=pseudophakic anterior chamber depth)*

SOURCES OF ERROR

Axial length. When ultrasound measurements were the standard method of biometry, axial length was the leading source of error in IOL power calculation. With state-of-the-art optical biometry, the reproducibility error of the pre- versus postoperative axial length is on the order of 0.05 mm, corresponding to 0.15 D.[1] This reproducibility error can be assumed to be close to the true average error that occurs in practice. The differences in measurement between optical instruments are a factor of 3 to 5 or smaller,[2] but they do not take into account the unknown individual refractive index of the crystalline lens. However, in eyes with dense cataracts, when ultrasound measurement is needed, typical errors of a factor of 3 to 5 or higher compared with optical methods (ie, 0.15–0.50 mm) are possible.

There are currently no technical developments in the pipeline that could significantly reduce measuring errors in dense cataracts. A problem with both ultrasound and optical methods is that a gold standard to verify the absolute values of axial length does not exist. Different instruments produce different results,[3] therefore requiring a final adjustment to clinical results that should be independent of the selected IOL.[4] Adjustments by variable formula constants achieve the same result, but they



(Data from 59 eyes: courtesy of P. Hoffmann, Castrop-Rauxel, Germany; data from 50 eyes: courtesy of K. Petermeyer, Tuebingen, Germany.)

Figure 2. Hyperopic bias in very long eyes. (A) The prediction error is calculated with the corneal refractive index as defined in the formulas and (B) with the values derived from the eye model of Liou and Brennan.

cannot separate the characteristics of measuring instruments from other possible sources of error. As long as no International Organization for Standardization (ISO) standard is defined for axial length measuring instruments, different devices are calibrated relative to each other on the basis of clinical data. That is, the same eyes must be measured with these different instruments before the instruments can be used for reliable IOL calculations.

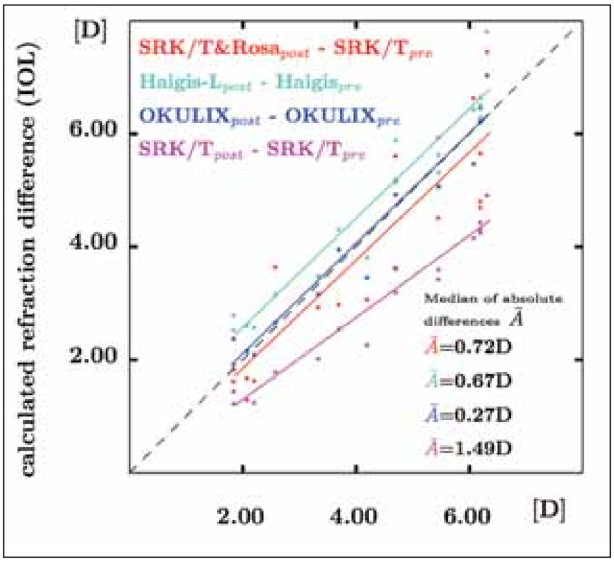
Anterior corneal radii. Anterior corneal radii are usually measured with keratometry. Standard deviations of 0.02 mm, corresponding to a 0.11 D refractive error, can be achieved under ideal conditions in normal eyes,[5] but keratometry in highly oblate aspheric post-LASIK eyes can produce a hyperopic outcome of 3.00 D or more. Keratometry must not be used in these cases. Topography and adequate consideration of asphericity are needed, together with a ray-tracing calculation, to solve this problem.[6] This is explored further below.

Posterior corneal radii. Posterior corneal radii are generally not measured but rather are assumed. The details of this assumption are hidden in the so-called corneal refractive index, which is different from the actual refractive index of corneal material. It results from a simplification when combining anterior and posterior corneal surfaces into one surface.[7] IOL calculation formulas use different values for this parameter, thereby implicitly using different eye models regarding the average ratio of anterior-to-posterior corneal radius. The resulting systematic differences are compensated for by different adjusting parameters, or formula constants. LASIK and PRK change the ratio of anterior-to-posterior corneal radius, invalidating this concept. Therefore, posterior corneal curvature or spatially resolved pachymetry must also be measured in eyes after refractive surgery. This is discussed in more detail below.

Postoperative IOL position. This is currently the dominant source of error in healthy eyes. Postoperative IOL position cannot be measured or calculated preoperatively; it can only be estimated. An improvement in accuracy is possible when the position and thickness of the crystalline lens are taken into account for the IOL position prediction algorithm.[8] The simple idea behind this approach is that position and size of the capsular bag should determine the position of the IOL, which is located inside the capsular bag. New optical devices can measure the required data with higher accuracy than older ultrasound machines. Nevertheless, the impact of individually different postoperative shrinkage of the capsular bag remains a source of uncertainty. Current investigations will determine how much the accuracy can be improved realistically in comparison with previous data. A standard deviation of 0.31 mm for the postoperative IOL position, corresponding to a 0.45 D error

TAKE-HOME MESSAGE

- Use of statistics in IOL power calculation can optimize average results; however, it does not adequately address outliers caused by eyes that differ too much from the average in their physical parameters.
- Current limitations and sources of error in IOL power calculation formulas include axial length, anterior and posterior corneal radii, postoperative IOL position, IOL mislabeling, and postoperative refraction.
- The effects of these errors differ among very short eyes, very long eyes, and eyes after corneal refractive surgery.



Data courtesy of T. Hofmann, Binningen, Switzerland)

Figure 3. IOL calculation errors in eyes undergoing LASIK.

in a mean-sized eye, has been reported.[5] Such averages can be misunderstood, however, because the effect of IOL position errors depends highly on other optical characteristics of the individual eye, particularly axial length and IOL power.

Figure 1 illustrates that the impact of IOL position errors on refraction increases greatly toward shorter axial lengths and decreases for longer eyes, becoming zero for 0.00 D IOL power and changing the sign in even larger eyes.

This behavior has led to another overlooked problem: Classic IOL calculation formulas such as the Haigis, Hoffer-Q, Holladay, and SRK/T formula use an assumed effective lens position (ELP) for the adjustment.[9-12] The ELP is different in each formula because the assumed corneal refractive indices are different, as noted above. On average, ELP can be adjusted so that the differences caused by these corneal refractive indices are compensated for and zero average prediction errors can be achieved for series of patients—a process called *individualization* or *optimization of formula constants*.

Unfortunately, this method cannot work when the effect of IOL position on refraction is close to zero in very long eyes.[13] In fact, all of these formulas end up with hyperopia of 1.00 to 2.00 D in very long eyes, which is ultimately a result of wrong assumptions about the ratio of the anterior-to-posterior corneal radius. This problem can be solved with these formulas[14] when the ratio of anterior to posterior corneal radius is taken from a more realistic eye model (Figure 2).[15]

IOL mislabeling. Errors in IOL labeling are the difference between the true IOL power and the power on the label. According to ISO Standard No. 11979, which specifies requirements for mechanical properties of IOLs, the allowed labeling error tolerances are 0.30 D for IOL powers below 15.00 D, 0.40 D for powers between 15.00 and 25.00 D, 0.50 D for powers between 25.00 and 30.00 D, and 1.00 D for powers above 30.00 D. Adjusting parameters such as the A-constant can compensate for systematic mislabeling errors; however, stochastic mislabeling errors remain undiscovered.

Postoperative refraction. Postoperative refraction contributes significant error to the total error when the results of an IOL calculation method are investigated, even if it is not an error of the IOL calculation method per se. In his analysis, Norrby[5] assumes a standard deviation of 0.39 D. However, refraction errors occur in the same way in spectacle or contact lens prescriptions.

PATIENT GROUPS

The effects of these errors differ among patient groups. Three types of eyes are of particular interest: very short eyes, very long eyes, and eyes after corneal refractive surgery.

Very short eyes. The dominant error in these eyes is the prediction error for postoperative IOL position. Because this error greatly affects the final refraction, the outcome will always be less accurate in these eyes compared with healthy ones (Figure 1).

Very long eyes. These eyes should theoretically achieve at least the same accuracy as healthy eyes, and, in fact, this can be achieved with ray tracing. The hyperopic bias of classic formulas can be avoided with a more realistic eye model, even with these formulas (Figure 2).

Eyes that have undergone corneal refractive surgery. Two-dimensional measurements of anterior and posterior corneal surfaces can be obtained with topography, Scheimpflug images, and optical coherence tomography (OCT). This data can be used as input for ray-tracing calculations.

Results of a pilot study[16] investigating the accuracy of such an approach are shown in Figure 3. In this study, topography and spatially resolved pachymetry were measured with the TMS5 (Tomey, Nagoya, Japan), and axial length and corneal radii were measured with the IOLMaster (Carl Zeiss Meditec, Jena, Germany) in 17 eyes before and after LASIK for myopic correction. Preoperative data were used to calculate the IOL power needed to achieve emmetropia with an Ar40e IOL (Abbott Medical Optics Inc., Santa Ana, California). Residual refraction was calculated with Okulix (Tedics, Dortmund, Germany) ray tracing based on the corneal data before and after LASIK.

The difference between these two calculations should theoretically be identical to the change in manifest refraction after LASIK (always in spherical equivalent); the deviation between measurement and calculation is a measure of the accuracy of the approach. For comparison, the same procedure was performed with the SRK/T formula using the correction proposed by Rosa et al[17] for post-LASIK data, the Haigis/Haigis-L formula,[18] and the uncorrected SRK/T formula (Figure 3). Note that the results of the formulas are independent of the choice of the adjusting parameters (ie, the A-constant) because their impact disappears in the calculation of differences.

CONCLUSION

Statistical-based methods are principally unable to avoid outliers. IOL calculation accuracy in nonnormal eyes can be improved only by the application of more physically based methods. This comprises measured data of anterior and posterior corneal surfaces and of the crystalline lens as well as a more elaborate ray-tracing calculation. However, all of these methods are available and already proven in initial clinical applications. ■

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