Biometry and intraocular lens power calculation

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Purpose of review

Heightened patient expectations for precise postoperative refractive results have spurred the continued improvements in biometry and intraocular lens calculations. In order to meet these expectations, attention to proper patient selection, accurate keratometry and biometry, and appropriate intraocular lens power formula selection with optimized lens constants are required. The article reviews recent studies and advances in the field of biometry and intraocular lens power calculations.

Recent findings

Several noncontact optical-based devices compare favorably, if not superiorly, to older ultrasonic biometric and keratometric techniques. With additional improvements in the internal acquisition algorithm, the new IOL Master software version 5 upgrade should lessen operator variability and further enhance signal acquisition. The modern Haigis-L and Holladay 2 formulas more accurately determine the position and the shape of the intraocular lens power prediction curve.

Summary

Postoperative refractive results depend on the precision of multiple factors and measurements. The element with the highest variability and inaccuracy is, ultimately, going to determine the outcome. By understanding the advantages and limitations of the current technology, it is possible to consistently achieve highly accurate results.

Keywords

biometry, cataract surgery, intraocular lens calculation

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Introduction

The refractive power of the human eye depends on the power of the cornea and the lens, the position of the lens, and the length of the eye. Accurate assessment of these variables is essential in achieving optimal postoperative refractive results. If these biometric measurements and calculations are inaccurate, the patients may be left with a significant refractive error. Studies conducted by Olsen [1••] showed that imprecision in measurements of anterior chamber depth (ACD), axial length and corneal power contribute to 42, 36 and 22%, respectively, of the error in predicted refraction after implantation of an intraocular lens (IOL).

As a result of heightened patient expectations, there is more than ever a need to accurately predict the correct IOL power. These demands have spurred the continued improvements in technology and refinements in biometry and IOL calculations. The article reviews recent studies and advances in the field of biometry and IOL power calculations.

Axial length measurement

Variations in axial length measurement have a significant impact on the final calculated IOL power. Currently, the axial length can be obtained by using either the A-scan ultrasound or the partial coherence laser interferometer. In A-scan ultrasound biometry, a crystal oscillates to generate a high-frequency sound wave that penetrates into the eye. When the sound wave encounters a media interface, part of the sound wave is reflected back toward the probe. These echoes allow us to calculate the distance between the probe and various structures in the eye. In 1999, Carl Zeiss introduced a noncontact partial coherence laser interferometer (IOL Master; Carl Zeiss Meditec, Jena, Germany) as an alternative technique to measure the axial length of the eye. It measures the delay and intensity of infrared light reflected back from media interfaces in order to determine the distance from the cornea to the retinal pigment epithelium.

Ultrasound biometry

Two types of A-scan ultrasound biometry are currently in use. The first is contact applanation biometry. This technique requires placing an ultrasound probe on the central cornea. While this is a convenient way to determine the axial length for most normal eyes, errors in measurement almost invariably result from the probe indenting the cornea and shallowing the anterior chamber. Since the compression error is variable, it cannot be compensated for by a constant. IOL power

calculations using these measurements will lead to an overestimation of the IOL power. In shorter eyes, this effect is amplified. The second type is immersion A-scan biometry, which requires placing a saline filled scleral shell between the probe and the eye. Since the probe does not exert direct pressure on the cornea, compression of the anterior chamber is avoided. A mean shortening of 0.25–0.33 mm has been reported between applanation and immersion axial length measurements, which can translate into an error of IOL power by approximately 1 D. In general, immersion biometry has been shown to be more accurate than contact applanation biometry in several studies [2–5]. The main limitation with the A-scan ultrasound is the poor image resolution due to the use of a relatively long, low-resolution wavelength (10 MHz) to measure a relatively short distance. In addition, variations in retinal thickness surrounding the fovea contribute to inconsistency in the final measurement.

Optical biometry

Since its introduction, optical biometry has been gaining popularity due to the fact that it offers an easy, contact-free method to quickly and accurately assess the axial length. The axial length measured by optical biometry is not, however, directly comparable to ultrasound biometry. Ultrasound biometry measures the distance from the anterior corneal to the inner limiting membrane, while optical biometry measures from the cornea to the retinal pigment epithelium. Thus, the measured axial length obtained from ultrasound and optical biometry cannot be expected to yield the same values. Hitzenberger *et al.* [6] found that the axial lengths measured by optical biometry were 0.18 mm longer than those measured by the immersion technique and 0.47 mm longer than those measured by the applanation technique. Kiss *et al.* [7] reported a mean difference in the measured axial length obtained with optical biometry and immersion biometry of 0.22 mm (range −0.24 to +0.57 mm; $R = 0.99$, $P < 0.05$). In order to be able to continue to use the A-constants and other formula constants developed over the years with ultrasound biometry, readings taken with the IOL Master were calibrated against the immersion ultrasound biometry.

Haigis *et al.* [8] found that the postoperative refraction was predicted correctly within ±1 D in 86% and within ±2 D in 99% of all cases using the immersion biometry data. A similar result was obtained using optical biometry. Kiss *et al.* [7] also reported that the refractive outcome in cataract patients using optical biometry was comparable to that achieved with immersion biometry. Other investigators have demonstrated greater accuracy and reproducibility with the IOL Master, as infrared laser-based measurement techniques using a 780-nm wavelength has an inherent advantage over a sound-based system with a frequency of 10 MHz and a resolution of 200 μm. Olsen [9•] reported an average absolute IOL prediction error of 0.65 D with ultrasound and 0.43 D with optical biometry ($P < 0.00001$). Sixty-two percent of predictions using optical biometry were within ±0.5 D compared with 45% with ultrasound. Lam *et al.* [10] reported no significant difference in the axial length obtained using optical biometry between different operators. Vogel *et al.* [11] reported intraobserver and interobserver variability (standard deviation) of ±25.6 and ±21.5 μm, respectively, for axial length measurements using the IOL Master.

Optical biometry has several advantages over ultrasound biometry. One is that the axial length measurement is performed through the visual axis since the patient is asked to fixate into the laser spot. In highly myopic or staphylomatous eyes, this can be particularly advantageous since it can sometimes be difficult to measure the true axial length through the visual axis with an ultrasound probe. Optical biometry is also superior to ultrasound in the measurement of pseudophakic and silicone oil-filled eyes. For optical biometry, it is not as critical how the media change because the correction factor that must be applied is much smaller than in ultrasound biometry. The preoperative axial length measurement obtained with the IOL Master was shown to be 0.07 mm longer than postoperative measurements ($P < 0.001$); this difference in axial length weakly, but statistically significantly, correlated with the Lens Opacities Classification System III nuclear cataract score [11,12,13•].

Accurate measurements require that the infrared laser be able to pass through the eye and return to the interferometer. Therefore, opacities along the visual axis can block the infrared laser. Reliable measurements can be difficult to obtain in eyes with tear film abnormalities, corneal pathology, mature and posterior subcapsular cataracts, vitreous opacities, maculopathy or retinal detachment. In addition, the patient must be able to maintain fixation. Various groups have reported that 8–20% of patients cannot be measured with optical biometry due to poor fixation, dense cataract or corneal pathology [13•,14–17]. Freeman and Pesudovs [17] reported that posterior subcapsular cataract with a Lens Opacities Classification System III score of greater than 3.5 and mature cataracts accounted for 16% of measurement failures. Cortical and nuclear cataracts did not seem to affect measurements.

The new IOL Master Advanced Technology software upgrade (version 5) is designed to enhance the signal-to-noise ratio in order to improve measurement of the axial length in eyes with media opacity. The new algorithm combines the individual measurement signals

to form a composite signal. Peaks in each signal are combined, resulting in amplification of the signal, while random noise in the signal cancels each other out. The software then looks for the highest peak in this composite signal. Results of unpublished studies conducted by Warren Hill, MD, showed that the standard IOL Master is capable of measuring 50–60% of patients for all classes of cortical density. With Advanced Technology, 87% of patients with cortical densities above 3.0 can be measured, 100% of patients with a nuclear color grading up to 3 can be measured, 93% of patients with a nuclear color grading above 3 can be measured, 100% of all eyes with a posterior subcapsular density up to grade 5 can be measured and 72% of eyes with a posterior subcapsular density above grade 5 can be measured.

Anterior chamber depth measurement

ACD measurements are required by several newer theoretical IOL power formulas to fine tune the IOL power prediction curve; thus, accurate measurements are essential to minimize the risk of unwanted refractive outcomes. Techniques currently available to measure the ACD include the A-scan ultrasonography, partial coherence interferometry, slit-scanning videokeratography, Scheimpflug imaging and anterior segment optical coherence tomography (OCT).

Corneal indentation, risk of corneal abrasions and infection, and off-axis measurements are some of the disadvantages of A-scan ultrasound [18,19]. Noncontact methods have become more popular because they avoid many of the pitfalls found in A-scan ultrasound. These include the IOL Master, the ACMaster (Carl Zeiss Meditec), the Orbscan II (Bausch & Lomb, Rochester, New York, USA), anterior segment optical coherence tomography (Visante OCT; Carl Zeiss Meditec, Dublin, California, USA) and the Pentacam (Oculus, Lynnwood, Washington, USA). The IOL Master uses a slit-beam photographic technique to acquire the ACD, while the ACMaster uses partial coherence interferometry. The Orbscan II is a three-dimensional scanning slit-beam topography system. The Pentacam uses a Scheimpflug camera to create a three-dimensional scan of the anterior segment of the eye. The anterior segment OCT provides high-resolution cross-sectional images of the anterior segment using a 1.3-mm infrared light.

Nemeth *et al.* [20] reported good reproducibility of ACD measurements using both the IOL Master and immersion ultrasound (coefficient of variation 0.13 mm and 2.20%, respectively). The ACD was significantly longer with the IOL Master ($P = 0 < .001$) compared with ultrasound, with no correlation between the ACD measurements of these two techniques ($R = 0.079$; $P = 0.397$). Vetrugno *et al.* [21] reported the Orbscan underestimated the ACD by a mean difference of 0.17 mm (4.68%) compared with applanation ultrasound. The reliability of ACD measurements was, however, higher for the Orbscan than for ultrasound ($P < 0.001$). Auffarth *et al.* [22] also reported a mean ACD difference of 0.04 ± 0.15 mm (1.2%) between immersion ultrasound and Orbscan. The correlation coefficient was 0.96. ($P < 0.00001$). Reddy *et al.* [23] found that contact ultrasound measured ACD 13% shorter, while the Orbscan and IOL Master showed good correlation. The Orbscan and the conventional nonrotating Scheimpflug camera were also reported to have excellent correlation [24]. Nemeth *et al.* [25•] showed that the ACD measurements were similar between the Pentacam and applanation A-scan ultrasound ($P = 0.84$); however, the measurements with Orbscan were on average 0.046 mm longer than the Pentacam ($P < 0.0005$) [26]. Although Sacu *et al.* [27] reported that the ACMaster provides high-precision anterior segment measurement in a user-friendly fashion, further studies are needed to assess its validity compared to the other techniques. Lavanya *et al.* [28•] reported mean ACDs of 3.08 and 3.14 mm with the IOL Master and anterior segment OCT, respectively ($P < 0.0001$). The repeatability and reliability coefficient were better with the anterior segment OCT, while the reproducibility of measurements was equal between anterior segment OCT and immersion A-scan ultrasound [29•].

Corneal refractive power

Obtaining an accurate corneal power can be challenging, since no keratometer directly measures the corneal power. With conventional manual and automated keratometry, the corneal power is calculated from the measured radius of curvature of the reflected corneal surface. To simplify the calculation, the cornea is assumed to be spherocylinder and a thin lens with a fixed anterior to posterior corneal curvature ratio. In most normal eyes with regular astigmatism, the calculated corneal power is easy to obtain and fairly accurate. The corneal powers in eyes that have undergone prior myopic photorefractive keratectomy or laser-assisted in-situ keratomileusis are, however, underestimated due to flattening in the anterior central corneal surface [30–33]. In these eyes, the relationship between the anterior to posterior curvature of the cornea is surgically altered, such that the standardized corneal index of refraction of 1.3375 is no longer correct. Thus, inaccurate assumptions by keratometry and topography may lead to overestimation of corneal power and a consequent hyperopic surprise.

Computerized videokeratography may better assess corneal power in postrefractive surgery eyes, since it usually takes more central corneal readings. Seitz and Langenbucher [34] and Qazi *et al.* [35••], however, showed that placido-based videokeratography still overestimated

the true corneal power in patients with previous photorefractive surgery. In cases of irregular corneal astigmatism (such as nodules or scars), the Orbscan – a scanning-slit videokeratography technique – provides a good estimation of the corneal power [36].

Alternative methods to estimate the corneal power after refractive surgery include the clinical history method [37,38], the contact lens over-refraction method [39,40] and the double-*K* adjustment method [41]. While these methods provide improved accuracy in eyes after refractive surgery, knowledge of pre- and postoperative refractive data is required or a refraction is needed before cataract surgery, which can be altered or hindered by the presence of the cataract.

Borasio *et al.* [42••] reported on the BESSt formula, which uses the anterior and posterior corneal curvature measurement from the Pentacam in the Gaussian optics formula to calculate corneal power. Performing the calculations on 13 eyes, they found that the BESSt formula was statistically significantly more accurate than the clinical history method, the clinical history method with double-*K* adjustments, Holladay *2* with *K* values estimated with the contact lens over-refraction method and the Holladay *2* with *K* values from a topographer. Qazi *et al.* [35••] reported that the Orbscan II 5.0 mm total axis power and the 4.0 mm total optical power can be used to accurately predict the true corneal power. Gelender [43••] also found that the corneal power derived from the Orbscan II could accurately determine the power of an IOL. The main advantage of these two methods is that knowledge of preoperative refractive data is not required.

At this time, comparison of the results obtained using several different methods in determining corneal power is advisable rather than relying simply on any one method alone.

Intraocular lens power calculation

The Hoffer Q, Holladay 1 and SRK/T are third-generation theoretical IOL power calculation formulas. They are two-variable formulas that mainly differ in the way they calculate the final position of the IOL. Their main limitations include making assumptions based upon normal schematic eye parameters that may not apply to all eyes, and predicting the final position of the IOL based solely on axial length and central corneal power. The Haigis-L formula represents a significant improvement over other two-variable formulas. It uses three IOL and surgeon-specific constants ($a0$, $a1$ and $a2$), and a measured ACD to alter and more accurately determine the position and the shape of the IOL power prediction curve. Corneal power measurements are not required in the calculation; thus, errors in measurement of the anterior corneal radius and the prediction of postoperative effective lens position are avoided. The main limitation of the Haigis formula is that the three a constants must be derived by regression analysis based on surgeon-specific data of a large number of cases ($n > 50$) containing a wide range of axial lengths. The Haigis-L formula is included as part of IOL Master's standard software package. The Holladay *2* formula, available since 1998, is one of the more accurate theoretic formulas currently available. The formula is easy to optimize and works well across a wide range of axial lengths; however, it requires input of seven variables to predict the effective lens position, including the axial length, average *K*, lens thickness, horizontal white-to-white corneal diameter, ACD, preoperative refraction and age of the patient. The lens thickness is currently not measured on the IOL Master, but can be obtained using the immersion A-scan.

Conclusion

With increasing patient expectations, the first step to obtain an accurate IOL power calculation is to be able to identify the patient's visual goals, especially if they have specific vocational or avocational needs. Using today's technology, it is possible to consistently have postoperative refractive outcomes within ± 0.25 D of the targeted refraction. In order to achieve these results, attention to proper patient selection, accurate keratometry and biometry, appropriate IOL power formula selection with optimized lens constant, and proper configuration of the capsulorhexis are required. Ultimately, the part with the highest variability and inaccuracy is going to determine the outcome. The accuracy of IOL biometry can be improved by implementing the following: minimizing variability and improving consistency by assigning a single properly calibrated instrument and experienced technician for the work-up, repeating and verifying measurements by a second person when necessary, using the IOL Master or immersion biometry rather than an applanation technique, using one of the newer IOL power calculation formulas and personalizing the lens constants for each formula, tracking your refractive outcomes, and optimizing your surgical technique by making the capsulorhexis round, centered and slightly smaller than the lens optic can all help to optimize your postoperative outcomes. By understanding the advantages and limitations of the current technology and following these guidelines, it is possible to consistently achieve highly accurate results.

Acknowledgements

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Additional references related to this topic can also be found in the Current World Literature section in this issue (p. 73).

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