Standardizing constants for ultrasonic biometry, keratometry, and intraocular lens power calculations

Jack T. Holladay, MD, MSEE

ABSTRACT

Purpose: To provide a method and values that facilitate standardization of constants for ultrasonic biometry, keratometry, and intraocular lens (IOL) power calculations.

Setting: University of Texas Medical School, Houston, Texas, USA.

Methods: Keratometry and ultrasonic biometry provide the two measured input variables for the six variable vergence equations used to calculate the appropriate IOL power for a specific patient with a cataract. A review of the literature reflecting the past 156 years of research and development reveals the appropriate index of refraction to be used with the keratometer for net optical corneal power, the location of the principal planes of the cornea, the nominal value for retinal thickness, and the appropriate velocities for ultrasonic measurement of the axial length. The relationship of the thick IOL to the thin IOL is derived along with the physical location of the thick lens. Two methods are described that provide the best IOL constant to be used by a manufacturer to minimize the prediction error for a surgeon using the lens for the first time. The formulas for phakic IOLs and secondary piggyback IOLs are also derived and applied to methods described above for standard IOLs.

Results: Using a standardized net index of refraction of 4/3 for the cornea eliminates a variability of 0.56 diopter (D) in the predicted refraction. Using a standardized 1532 m/s velocity for axial length measurements and adding a value of 0.28 mm reduces the tolerance of axial length measurements to ±0.03 mm for any length eye. The physical location of the thick IOL's secondary principal plane must be anterior to the thin lens equivalent by approximately the separation of the principal planes of the thick lens. For biconvex poly(methyl methacrylate) IOLs, the separation in the principal planes is approximately 0.10 mm. Using these relationships, the physical position of the thick lens within the eye can be used to confirm the lens constant for any IOL style.

Conclusions: Standardizing the constants for keratometry, ultrasonic biometry, and IOL power calculations can significantly improve the predictability of refractive outcomes. Back-calculating and physically measuring the position of the lens within the eye can provide surgeons with an initial lens constant known to have a standard error of the mean of ±0.05 mm (±0.10 D). Other parameters such as the cardinal points of a lens, the shape factor, the lens-haptic plane, and the center lens thickness would allow further refinement of IOL power calculations. *J Cataract Refract Surg 1997; 23:1356–1370*

Intraocular lens (IOL) power calculations have improved in accuracy over the past three decades. At present, the standard of care is to have at least 50.0% of patients within ±0.50 diopter (D) of predicted refraction, 90.0% within ±1.00 D, and 99.9% within ±2.00 D.[1,2] The increased accuracy is due to improvements in the (1) design and standardization of measurement instruments (keratometer and ultrasonic biometer), (2) surgical refinements in IOL implantation, and (3) refinement of formulas used to calculate the appropriate power of an IOL in a specific patient.

Standards for keratometers include a standardized keratometric index of refraction (1.3375).[3] Unfortunately, the diameter of the optical zone measured varies from 2.4 to 3.2 mm from one manufacturer to another, and not all manufacturers have adopted the standardized keratometric index of refraction. Although these differences cause variations of approximately ±0.25 D in normal corneas, the variations can be much larger in eyes with irregular astigmatism or eyes that have had corneal refractive surgery.

Ultrasonic biometers also have requirements for maximum error when measuring standardized poly(methyl methacrylate) (PMMA) test blocks. Several investigators[4-9] have demonstrated that the average velocity of ultrasound in the normal and cataractous eye is from 1550 to 1555 m/s. Unfortunately, no single value has been universally adopted by manufacturers, although 1553 m/s has been recommended.[10] Furthermore, using an average velocity becomes more inaccurate[11,12] the more unusual the axial length in a specific eye (deviation from 23.45 mm).[9,13] In unusual and pseudophakic eyes, it is more accurate to measure the axial length at 1532 m/s and then add or subtract a corrected axial length factor (CALF) distance caused by the different velocity in the lens than to use an average velocity.[12,14]

$$AL_U = AL_{1532} + CALF \tag{1a}$$

where AL_U is the patient's true ultrasonic axial length from the corneal vertex to the vitreoretinal interface, AL_{1532} is the distance measured at an average sound velocity of 1532 m/s, and CALF is the corrected axial length factor.[12,14] The two parameters necessary to calculate the CALF are the thickness of the lens (T_L) and the ultrasonic velocity through the lens (V_L).

$$CALF = T_L * \left(1 - \frac{1532}{V_L}\right) \tag{1b}$$

If the clinician does not have an optical instrument for measuring lens thickness, the crystalline lens thickness should be assumed to be 4.01 mm for a patient at age 1 and 4.80 mm at age 80.[15,16] Based on Bellows' data,[15] a good approximation of lens thickness for any age is to place the patient's age in years as the two digits to the right of decimal point; i.e., if the patient is age 52, the corresponding crystalline lens thickness is approximately 4.52 mm. The velocity through the cataractous crystalline lens has been shown to decrease with age, presumably from progressive formation of a cataract. In a 1975 study using 50 cataractous eyes and 4 eyes from children, Coleman and coauthors[6] found that at approximately 1 year of age the sound velocity through the crystalline lens is 1659 m/s and by 72 years the average velocity is 1629 m/s. Unfortunately, a smaller study of only 12 eyes by Jansson and Kock[4] demonstrated an average velocity of 1640.5 m/s, which is the more prevalent value. Coleman and coauthors' study also controlled for temperature, length of time between lens removal and measurement, the medium in which the lens was supported, and the integrity of the lens capsule. The decrease in velocity with age seems counterintuitive at first, since the lens would become more dense and conduct sound more rapidly with opacification and loss of accommodation. One explanation is that the water content of the lens increases with age and cataract formation.[17] Also, there is a wide range of velocities within an individual lens,[4] as well as great acoustic discontinuities.[18] The historical data would support a decrease in the sound velocity of approximately 5 m/s every decade from age 1 to age 70 or approximately 0.5 m/s per year with increasing age.[6]

Summarizing, the values for thickness and velocities at age 72 would be 0.55 mm (1640 m/s) for the cornea, 4.72 mm (1629 m/s) for the crystalline lens, 18.18 mm (1532 m/s) for the vitreous and aqueous length, and 23.45 mm for the total axial length. Using these values, the average sound speed is calculated to be

From the University of Texas Medical School, Houston, Texas, USA.

Dr. Holladay is author of the Holladay Formula and provides consultation to A-scan companies that use the formula.

Reprint requests to Jack T. Holladay, MD, Houston Eye Associates Building, 2855 Gramercy, Houston, Texas 77025, USA.

1553.012 m/s, the exact value most commonly recommended by experts.[10] The most common value used on instruments today is 1555 m/s. To convert measurements taken at an average velocity of 1555 to 1532 m/s, the following ultrasonic conversion equation should be used:

$$AL_{1532} = \frac{1532}{1555} * AL_{1555} \tag{1c}$$

where AL_{1555} is the axial length for an average velocity of 1555 m/s and AL_{1532}, the axial length for average velocity of 1532 m/s.

Although this is tedious to explain, a simple table of CALFs can be created with age and crystalline lens thickness versus age and average velocity (Table 1).

Using this method makes any error in the axial length measurement independent of the patient's overall axial length. Any error is dependent on only the values used for the thickness and velocity of the crystalline lens. Using an average value for the entire eye results in errors that are much larger and dependent on axial length.

Since the average patient should move along the highlighted diagonal values in Table 1, the maximum error can be only 0.04312 mm (0.30697–0.26385). If a nominal value for CALF were 0.28 mm, the value for the average age of a cataract patient, the error at age 1 would be −0.02697 mm (0.28–0.30697) and at age 90 would be +0.01615 mm (0.28–0.26385). These errors are far below the tolerances of the measurement. It is therefore recommended that axial lengths be measured at, or converted to a distance for, 1532 m/s (AL_{1532}), to which is added a nominal value of 0.28 mm to obtain the true ultrasonic axial length (AL_U).

$$AL_U = AL_{1532} + 0.28 \tag{1d}$$

For IOLs the same process can be performed, except the exact ultrasonic velocity for each material at eye temperature is known.[12] The exact center thickness of each IOL model as a function of dioptric power is available from the manufacturer. If the dioptric power of the pseudophakic lens is unknown, a nominal value can be used for the specific material and lens style.[12]

For instruments with gates that can assign a value for each ultrasonic component of the eye (cornea, aqueous, crystalline lens or IOL, and vitreous humor), the solution is much simpler. Simply set the velocity of the lens gate (crystalline or intraocular) to the appropriate velocity for the severity of the nuclear sclerotic cataract or the IOL material. The ultrasonic velocities at eye temperature (35°C) for the cornea (1640 m/s), aqueous and vitreous (1532 m/s), PMMA (2780 m/s), silicone (980 m/s), and acrylic (2180 m/s) are well documented.[12,14] Ultrasonic biometers that use gates are more accurate, particularly in unusual eyes, than those that use an average velocity.[10,19] Adopting the velocities and methods suggested above would eliminate much of the unnecessary variability among manufacturers and would maintain a theoretical tolerance of ±0.03 mm (±0.06 D) in the normal eye and unusual phakic eye.

The second area for standardization is the surgical technique. The current preferred surgical technique for cataract surgery is a small incision, continuous tear capsulorhexis with the lens implanted in the bag.[20] This technique has little or no effect on the spheroequivalent power of the cornea or axial length. It should be adopted as the standard for implantation of most currently used IOLs. Exceptions would include anterior chamber lenses, phakic IOLs, and IOLs intended for the sulcus in eyes in which the capsular bag is not intact or able to safely support an IOL.

Table 1. Corrected axial length factor.

Age (Years)	Lens Thickness (mm)	Age (Years) / Lens Velocity (m/s)				
		1/1659	30/1649	50/1639	70/1629	90/1619
1	4.01	0.30697	0.28452	0.26179	0.23878	0.21548
31	4.31	0.32994	[illegible]	0.28137	0.25664	0.23161
51	4.51	0.34525	0.31999	0.29443	0.26855	0.24235
71	4.71	0.36056	0.33418	0.30749	0.28046	0.25310
91	4.91	0.37587	0.34837	0.32054	0.29237	0.26385

*Recommended CALF to be used for all ages

The third area for standardization is the assumptions that must be used in the first-order, or Gaussian, optics[21] which describe the relationship between spectacles, cornea, IOL, retina, and the media separating them.[22] It is this area that I would like to focus on for the remainder of the discussion. The goal is to propose a set of constants and assumptions that (1) reflect the extensive research and experience that has occurred over the past 30 years, (2) will have the least effect on existing constants, and (3) will allow the greatest flexibility for future developments.

Intraocular Lens Calculations Requiring an Axial Length Vergence Formula

Theoretical Formulas

The theoretical thin-lens formula for IOL power calculations has not changed since Gauss first invented first-order optics almost 150 years ago.[21] Credit for first applying Gaussian optics to modern-day IOLs is given to Fedorov.[23] Although several investigators have presented the theoretical formula in different forms,[24] there are only slight variations in the choice of retinal thickness, corneal index of refraction, and corneal principal planes. There are six variables in the formula: (1) optical net corneal power (K_o), (2) optical axial length (AL_o), (3) IOL effective power (IOL_e), (4) effective thin-lens position (ELP_o), (5) desired refraction (DPostRx), and (6) the vertex distance (V) of the desired refraction. An IOL power has been labeled using the effective power instead of the vertex power since the *1984* U.S. Food and Drug Administration (FDA) standardization.[25] Normally, the IOL power is the dependent variable and is solved for using the other five variables; where distances are in millimeters, refractive powers in diopters, and indices of refraction have been multiplied by 1000:

$$IOL_e = \frac{1336}{AL_o - ELP_o} - \frac{1336}{\dfrac{1336}{\dfrac{1000}{\dfrac{1000}{DPostRx} - V} + K_o} - ELP_o} \qquad (2)$$

The 1336 is 1000 times the refractive index of aqueous and vitreous and the 1000 is 1000 times the refractive index of air. The only variable that cannot be chosen or measured preoperatively is the effective thin-lens position (ELP_o). Figure 1 illustrates the physical locations of the variables. The average values for the keratometric reading and axial length of the human eye from large populations have been used.[9,13] The term effective lens position was recommended to the FDA in 1995 to describe the position of the lens in the eye,

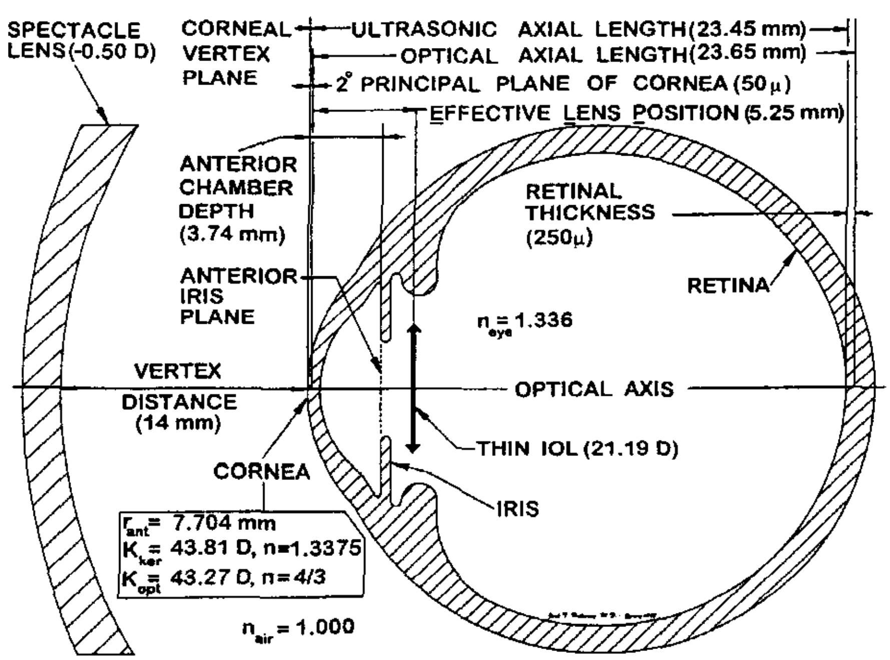

Figure 1. (Holladay) Standardized pseudophakic schematic eye (thin IOL).

since the term anterior chamber depth (ACD) is not anatomically accurate for lenses in the posterior chamber and can lead to confusion with the anatomic anterior chamber depth (AACD).

Standardized Eye Model

Optical Versus Ultrasonically Measured Axial Length

The optical axial length (AL_o) of the human eye is defined as the axial distance from the secondary principal plane of the cornea to the photoreceptors in the fovea (Figure 1). There is no significant difference among investigators in the location of the secondary principal plane of the cornea (P_{C2}). Binkhorst[26] used 0.0506 mm and Colenbrander[27] used 0.05 mm. Colenbrander's value of 0.05 mm is an appropriate choice because the additional two decimal places are far above the accuracy of the other axial distances.

The thickness of the retina (R_t), the distance between the vitreoretinal interface and the visual cell layer, was chosen as 0.50 mm by Colenbrander,[27] 0.20 mm by Oguchi and van Balen,[28] and 0.25 mm by Binkhorst.[9] Binkhorst's value has been the most commonly used since 1981.

The ultrasonically measured axial length (AL_u) would be the distance from the corneal vertex to the vitreoretinal interface and would therefore have the following relationship to the optical axial length (AL_o):

$$AL_o = AL_u - P_{C2} + R_t \tag{3a}$$

$$AL_o = AL_u - 0.05 \text{ mm} + 0.25 \text{ mm} \tag{3b}$$

$$AL_o = AL_u + 0.20 \text{ mm} \tag{3c}$$

This is the recommended standardized conversion from (AL_u) to (AL_o).

Optical Versus Keratometric Power of the Cornea

All keratometers measure the front radius of curvature of the cornea and then convert to power by dividing into the difference of two indices of refraction. The formula for converting the radius of curvature of a refractive surface bounded by two optical media is referred to as the simple spherical refractive surface formula:

$$K_k = \frac{n_2 - n_1}{r} \tag{4a}$$

The variables n_1 and n_2 are the indices of refraction of the first and second media, respectively, and r is the radius of curvature of the interface. The value for n_1 is 1.000 (index of refraction for air); the value for n_2, 1.3375 (standardized keratometric index of refraction), was chosen so an anterior radius of curvature of the cornea of 7.5 mm would yield a power of 45.0 D.[29]

$$K_k = \frac{1.3375 - 1.000}{r_a} = \frac{0.3375}{r_a} \tag{4b}$$

where r_a is the anterior radius of curvature of the cornea.

The only rationale for choosing the index of refraction of 1.3375 is that it makes the two numbers (7.5 and 45.0) agree exactly. The origin of the standardized keratometric index of refraction remains obscure, dating back to the nineteenth century. If an index of refraction of 1.336 were used, a more physiologic choice since this is the index of refraction of the tear film, the resulting power would be 44.80 D. This was the original value proposed by Javal, the inventor of the keratometer, over 100 years ago.[30] For this value to be correct, the anterior and posterior radii of the cornea must be equal. Several studies have shown that the posterior radius of the cornea is at least 1.2 mm steeper than the anterior radius, which reduces the net optical power of the cornea even more than 0.2 D.[31–35] Using the index of refraction of the corneal stroma of 1.376, a posterior corneal radius that is 1.2 mm steeper, and a corneal thickness of 0.55 mm results in the calculated net optical power of a cornea of 44.44 D. The calculated net optical power of the cornea using these conditions is approximately 0.56 D less than the keratometric power.

Using the anterior radius of 7.5 mm and a net optical power of 44.44 D, a net corneal index of refraction that would yield 1.3333 can be calculated. Recent studies have suggested that using an even lower value of 1.3315 is appropriate in IOL calculations, suggesting the posterior radius of the cornea is more than 1.2 mm steeper than the anterior radius.[36] Binkhorst chose 4/3 (1.3333...) as the optical net index of refraction for the cornea because it yielded the best results for his calculations, the same reason Olsen chose 1.3315.[36] Although Binkhorst's value yielded more accurate results with his formula, his explanation was incorrect.[37] He thought the reduced power was due to a

0.56 D flattening of the cornea after cataract surgery, which was reported by Floyd.[38] With today's modern small incision surgery, however, there is no significant change in the spheroequivalent power of the cornea. In any case, Binkhorst's use of an index of 1.333 (4/3) was more accurate than using the standardized keratometric index of refraction. The value of 4/3 for the net corneal index of refraction is an appropriate value and would have the minimum impact on current formulas.

Using the value of 4/3 as the net corneal index of refraction, the net optical corneal power can be related to the keratometric power by the following equation:

$$K_o = K_k * \frac{4/3 - 1}{1.3375 - 1}$$
$$= K_k * \frac{1/3}{0.3375} = 0.98765431 * K_k \qquad (5)$$

This is the recommended "standard" method of converting the keratometric power (K_k) to the net optical power of the cornea (K_o). Since a few keratometers use a value other than the standardized keratometric index, clinicians should confirm the value used on their instrument. If the value is not 1.3375, the actual value used (e.g., 1.336) should be substituted for 1.3375 in equation 5.

Determining the Optimal ELP_o for a Surgeon and Manufacturer Using the Axial Length Vergence Formula

In 1988, we first published the quadratic solution to the axial length vergence formula for the ELP_o.[2] Equations 6a through 6e are the reverse solution of the axial length vergence formula for the ELP_o given the stabilized actual postoperative refraction (APostRx) and the actual power of the implanted IOL (IOL_e).

$$X = \frac{1336}{\dfrac{1000}{\dfrac{1000}{APostRx} - V} + K_o} \qquad (6a)$$

$$A = IOL_e \qquad (6b)$$

$$B = -IOL_e * (AL_o + X) \qquad (6c)$$

$$C = 1336(AL_o - X) + IOL_e * X * AL_o \qquad (6d)$$

$$ELP_o = \frac{-B \pm \sqrt{B^2 - 4A * C}}{2A} \qquad (6e)$$

In equation 6e, there is a ± sign; the plus sign is used for negative IOL powers and the minus sign is for positive IOL powers.

Equations 6a through 6e allow surgeons to determine their optimal or "personalized" AVG_s ELP_o based on their experience with any style lens by calculating the average back-calculated ELP_o from their experience in 20 to 30 cases with a lens style. Personalizing the lens constant reduces prediction error to a minimum.[1,2] The actual number of cases for a single surgeon should be sufficient to have a standard error of the mean (SEM) of less than ±0.125 mm, which is approximately ±0.250 D. A manufacturer would need approximately 10 or more surgeons to produce a nominal or manufacturer's recommended initial AVG_m ELP_o with an SEM less than ±0.05 mm (approximately ±0.10 D). Since approximately 90% of cases done today use phacoemulsification with continuous tear capsulorhexis and the lens is placed in the bag, these conditions should be the inclusion criteria for the cases used by the manufacturer. A postoperative visual acuity of better than 20/50 should also be required to avoid inaccuracies in the measurement of the actual postoperative refraction (APostRx).[1,2]

Conversion of Existing Lens Constants to ELP_0

As mentioned previously, using ELP_0 rather than anterior chamber depth (ACD) to represent the position of the equivalent thin lens vis a vis the secondary principal plane of the cornea (P_{C2}) avoids confusion with the anatomic anterior chamber depth (AACD). In the mid 1960s, when IOLs were being more widely used clinically, this distinction was not important because convex-plano IOLs that were iris-supported were very near the AACD. As the preferred implantation location moved to the posterior chamber, first in the sulcus and then in the bag, the term ACD became more inaccurate and confusing.

The lens constant values that currently exist on labels, package inserts, and promotional material need not be changed, but the term ACD should gradually be changed to the term ELP_0 as new informational material is printed. New lenses should adopt ELP_0 from the outset. As mentioned above, the value for ELP_0 for specific new lens model should ideally represent the average effective thin lens position (AVG_m ELP_o) from a statistically significant number of surgeons with a final SEM of less than ±0.05 mm.

Empirically determining the lens constant is problematic for the manufacturer at present because the lens constant must be in the labeling before the lens can be approved for implantation. It is recommended that the manufacturer be allowed to have a preliminary lens constant derived from theory, parent lenses, etc., until the manufacturer has sufficient data to provide an empirical lens constant based on actual surgical experience.

Other prevalent historical lens constants besides ACD include the A-constant (A-const) and the surgeon factor (SF). The A-const originated in 1980[39–41] with linear regression formulas. The linear regression formulas allowed simple calculations for IOL powers and simple calculations of personalized constants before computers and programmable calculators were available. Although the regression formulas yielded good results in normal eyes, they became very inaccurate the more unusual the eye and the higher the degree of targeted postoperative refractive error. The authors who developed the A-const recognized this problem and adopted the theoretical formula in 1990 to reduce the errors in unusual eyes.[42]

The A-const is in units of diopters. This unit was forced because the linear regression formula had the dependent variable as the required IOL power, and the constant in a linear regression formula must be in the same units as the dependent variable. By using a constant that was in diopters, two different lens styles with the same A-const could be interchanged with the same refractive result. Unfortunately, this interchangeability is only true when the lens power is near the mean value of the IOL; i.e., near 21.0 D. As the required lens power for a specific patient becomes more unusual, the lower the probability that two different lens models with the same A-const are interchangeable. Additional parameters such as central lens thickness, shape factor, and lens-haptic plane distance would have to be equal over the power range of the two lens models for them to be equivalent (see "Other Intraocular Lens Parameters").

Although the A-const is in diopters, it must have a linear transformation to ELP_o, i.e., in a sufficiently large population statistically, the mean IOL power, mean K-reading, and mean axial length must yield a mean A-const that is comparable to the mean ELP_o. This point of coincidence is where they each achieve a minimum prediction error of the population.[2]

We performed this calculation for the point of coincidence in 2000 eyes from 12 surgeons in 1988.[2] The results are given below:

$$ELP_o = \frac{(\text{Aconst} * 0.5663) - 65.600 + 3.595}{0.9704} \tag{7a}$$

For example, if the A-const were 118.50 D, then

$$ELP_o = \frac{(118.50 * 0.5663) - 65.600 + 3.595}{0.9704} = 5.26 \text{ mm}$$

Re-solving for the A-const, we have

$$\text{Aconst} = \frac{(ELP_o * 0.9704) + 65.600 - 3.595}{0.5663} \tag{7b}$$

In 1988,[2] we described the Holladay 1 formula and used a lens constant named the surgeon factor (SF). The SF was chosen because we believed that the distance from the pseudophakic anterior iris plane to the principal plane of the thin IOL (ELP_o) was the most consistent parameter for a lens model for any size eye. Although this concept may still be true, the SF should also be converted to ELP_o for standardization. The relationship of the SF to ELP_o was determined with the same 2000 eyes and 12 surgeons, yielding the following relationship:

$$ELP_o = \frac{SF + 3.595}{0.9704} \tag{8a}$$

For example, if the SF were 1.51 mm, then

$$ELP_o = \frac{1.51 + 3.595}{0.9704} = 5.26 \text{ mm}$$

Re-solving for the SF, we have

$$SF = (ELP_o * 0.9704) - 3.595 \tag{8b}$$

During the past 9 years, several manufacturers, researchers, and clinicians have used equations 7 and 8 to convert from one lens constant to another. Adopting these conversion equations will eliminate future inconsistencies as the older lens constants are phased out.

Preoperative Prediction of Effective Lens Position in a Specific Case from the Thin IOL

Improvements in IOL power calculations using the axial length vergence formula over the past three decades are a result of improving the predictability of the variable ELP_X for a specific patient from the AVG_m ELP_0 from the manufacturer. Before 1980, the ELP_X for every iris clip IOL was 4.0 mm (first-generation theoretical formula) in every patient. This value worked well in most patients because most lenses implanted had iris-clip fixation, in which the principal plane averages approximately 4.0 mm posterior to the corneal vertex.

In 1981, Binkhorst improved the prediction of ELP_X for a specific patient by using a single variable predictor, the axial length, as a scaling factor for ELP_X (second-generation theoretical formula).[43] If the patient's axial length was 10% greater than normal (23.45 mm), Binkhorst would increase the ELP_X by 10%. The average value (AVG_m ELP_0) was increased to 4.5 mm in the early 1980s because the preferred location of an IOL was in the ciliary sulcus, approximately 0.5 mm deeper than the iris plane. Also, most lenses were convex-plano, similar to the shape of iris-supported lenses.

The average AVG_m ELP_0 in 1997 has increased to 5.25 mm. This increased distance has occurred for two primary reasons. First, the majority of implanted IOLs are biconvex, moving the principal plane of the lens even deeper into the eye, and second, the desired location for the lens is within the capsular bag, which is 0.25 mm deeper than the ciliary sulcus.

In 1988,[2] we proved that a two-variable predictor using axial length and keratometry could significantly improve the prediction of ELP_X for a specific patient, particularly in unusual eyes (third-generation theoretical formula). The prediction formula for ELP_X in the original Holladay 1 formula was based on the geometric relationships of the anterior segment. Although several investigators have modified the two-variable predictor from the original Holladay 1 formula, no comprehensive studies involving several surgeons and thousands of cases have shown any significant improvement using only these two variables.

In 1995, Olsen and coauthors[44] published a four-variable predictor that used axial length, keratometry, preoperative phakic anterior chamber depth, and phakic lens thickness. The results showed improvement over the two-variable prediction formulas, including the Holladay 1, for a simple reason. The more information we have about the anterior segment of a specific patient, the better we can predict the ELP_X. This explanation is a well-known theorem in prediction theory in which the more variables that can be measured describing an event, the more precisely one can predict the outcome.

In a recent study,[45] we discovered that the anterior and posterior segments of the human eye are often not proportional in size, causing significant error in the prediction of the ELP_X in extremely short eyes (<20.0 mm). We found that in eyes shorter than 20.0 mm, the anterior segment was completely normal in most cases. Because the axial lengths were so short, the two-variable prediction formulas severely underestimated the ELP_X, explaining part of the large hyperopic prediction errors with two-variable prediction formulas. After recognizing this problem, we began to take additional measurements on extremely short and extremely long eyes to determine whether the prediction of ELP_X could be improved by knowing more about the anterior segment. Table 2 shows the clinical conditions that illustrate the independence of the anterior segment and the axial length.

For the past 3 years, we have been gathering data from 35 investigators worldwide. Several additional measurements of the eye have been taken, but only seven preoperative variables (axial length, corneal power, horizontal corneal diameter, phakic anterior chamber depth, phakic lens thickness, preoperative refraction, and age) have been useful in significantly improving the prediction of ELP_X in eyes that range from 15.0 to 35.0 mm in axial length.

The improved prediction of ELP_X is not totally due to the use of seven preoperative measurements but is also a function of the improved technical skills of the surgeons who are consistently performing capsulorhexis

Table 2. Clinical conditions demonstrating the independence of the anterior segment and axial length.

Anterior Segment Size	Axial Length: Short	Normal	Long
Small	Small eye Nanophthalmos	Microcornea	Microcornea + axial myopia
Normal	Axial hyperopia	Normal	Axial myopia
Large	Megalocornea Axial hyperopia	Megalocornea	Large eye Buphthalmos + axial myopia

and placing the lens in the capsular bag. If the lens is not placed in the bag, the axial misplacement of the lens causes a refractive surprise that varies with the power of the lens and the amount of displacement. A 20.0 D IOL that is 0.5 mm axially displaced from the predicted ELP_X will result in approximately a 1.0 D error in the stabilized postoperative refraction. A 40.0 D lens axially misplaced by the same amount would cause a 2.0 D error. Because of this direct relationship to the lens power, the problem is much less evident in extremely long eyes; the implanted IOL is either low plus or minus to achieve emmetropia following cataract extraction.

Thick Lens Versus Thin Lens Paradox

Figure 2 illustrates the relationship of a thin lens with principal point ELP_o and a thick lens with principal points ELP_1 and ELP_2 in aqueous, each having an equivalent power of IOL_e. For distant objects (collimated light), both thin and thick lenses have the same effective focal length and will bring rays into the same focal point (f_e) when ELP_o is coincident with ELP_2.

Unfortunately, this relationship is only true when collimated light is incident on the IOL. When the rays incident on the IOL are converging, such as those from the cornea, this relationship is not true. Binkhorst[37] and Jalie[46] overlooked this when generalizing their formulas for the thick-lens equivalent. For plus lenses, the thick lens must be placed anteriorly to the thin lens, as shown in Figure 2, to bring the final rays into the same point of focus on the fovea. The amount of anterior displacement ($L_{2,0}$) of the thick lens (ELP_2) with respect to the thin lens (ELP_o) is nonlinear and depends on corneal power (K_o), IOL power (IOL_e), the separation of the thick lens principal planes ($T_{1,2}$), and the position of the thin lens (ELP_o). Because the relationship is nonlinear, the simplest method to determine $L_{2,0}$ is by successive iteration. The first approximation of $L_{2,0}$ is to set it equal to $T_{1,2}$.

For example, using the thin lens in Figure 1, we have an ELP_o of 5.25 mm, IOL_e of 21.19 D, K_o of 43.27 D, and APostRx of −0.50 D at vertex (V) of 14.0 mm. To find the equivalent position of a thick lens with principal planes separated by 0.1000 mm ($T_{1,2}$) and the same effective IOL power (IOL_e) of 21.19 D, successive iterations indicate that $L_{2,0}$ must be 0.1001 mm. Therefore, the thick-lens ELP_2 must be 5.15 mm (5.25–0.10) behind the secondary principal plane of the cornea (P_{C2}). Note that in this example, $L_{2,0}$ and $T_{1,2}$ were virtually the same. This relationship is not always as close and must be determined by successive iterations.

Once the location of the secondary principal point ELP_2 of the thick lens is known, it is easy to determine the physical location of the anterior vertex (AV) of the

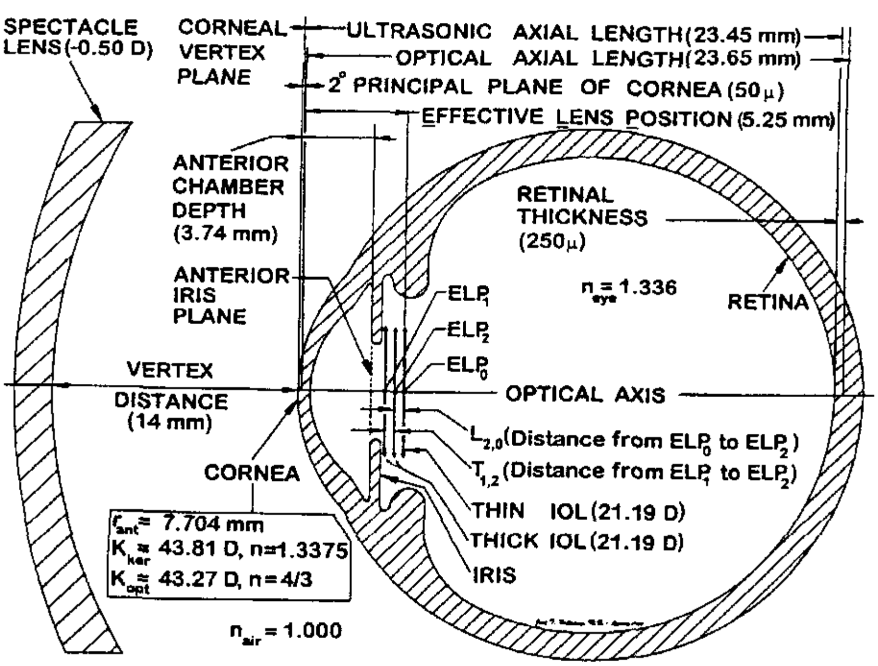

Figure 2. (Holladay) Standardized pseudophakic schematic eye (thick IOL).

thick lens in relation to the actual vertex of the cornea (P_{C1}) as shown in Figure 2. If the lens in the above example were an equiconvex 21.19 D IOL with a 1.0 mm thickness (T_L) and a separation in the principal planes of 0.10 mm ($T_{1,2}$), the first and second principal planes (H_1 and H_2) must be 0.45 mm from the front and back vertex of the IOL (e_1 and e_2), respectively. In this example, the anterior vertex of the thick lens must be 0.65 mm anterior to ELP_0 or 4.60 mm (5.25–0.65) posterior to the secondary principal plane of the cornea, P_{C2}. Since P_{C2} is 0.050 mm posterior to the anterior corneal vertex, the distance from the anterior vertex of the cornea to the anterior vertex of the thick IOL (AVLP) is 4.65 mm (4.60 + 0.05). These relationships can be expressed in equation form:

$$ELP_2 = ELP_0 - L_{2,0} = 5.25 - 0.10 = 5.15 \text{ mm} \quad (9a)$$

$$AV_{PC2} = ELP_2 - T_{1,2} - e_1 = 5.15 - 0.10 - 0.45 = 4.60 \text{ mm} \quad (9b)$$

$$AV_{PC1} = AV_{PC2} + P_{C2} = 4.60 + 0.05 = 4.65 \text{ mm} \quad (9c)$$

The distance from the corneal vertex to the anterior vertex of the IOL (AV_{PC1}) can be measured clinically using optical or ultrasonic methods. The physical measurement can be used to validate the calculated value described above. Reversing the equations to calculate ELP_0 from AV_{PC1},

$$AV_{PC2} = AV_{PC1} - P_{C2} = 4.65 - 0.05 = 4.60 \text{ mm} \quad (10a)$$

$$ELP_2 = AV_{PC2} + T_{1,2} + e_1 = 4.60 + 0.10 + 0.45 = 5.15 \text{ mm} \quad (10b)$$

$$ELP_0 = ELP_2 + L_{2,0} = 5.15 + 0.10 = 5.25 \text{ mm} \quad (10c)$$

The manufacturer has two independent methods for determining ELP_0, one using the actual postoperative refraction (ApostRx: equations 6a to 6e) and the other, the measurement from the corneal vertex to the anterior vertex of the lens (AV_{PC1}: equations 10a to 10c).

Other Intraocular Lens Parameters

Cardinal Points

Every lens can be characterized optically by three pairs of cardinal points: two equivalent focal points (f_e and f'_e), two principal points (P_1 and P_2), and two nodal points (N_1 and N_2).[47] For IOLs, the nodal points coincide with the principal points because the refracting medium (aqueous) is the same on both sides of the lens. The distances from the principal points to the respective anterior and posterior vertex of the lens (e_1 and e_2) provide all the necessary information to relate the theoretical thin lens to the physical thick lens (e.g., f_{AV}, anterior vertex focal length and f_{PV}, posterior vertex focal length). The cardinal points for a biconvex IOL are shown in Figure 3.

Lens Shape Factor

The shape factor for a lens model allows the user to determine the relationship of the principal planes to the physical dimensions of the lens. The formula for the shape factor is

$$\text{Shape Factor} = \frac{C_1 + C_2}{C_1 - C_2} \quad (11)$$

where C_1 is the curvature of the anterior surface, C_2 is the curvature of the posterior surface, and convex radii toward the cornea are positive while concave radii toward the cornea are negative.[48] For positive IOLs, the relationship between shape factor and the form of the lens is shown in Table 3.

Although a few meniscus lenses are still manufactured, most lenses implanted today have shape factors between −1 and +1 because they are convex-plano, biconvex, or plano-convex. If the shape factor changes as a function of power, the shape factor for each

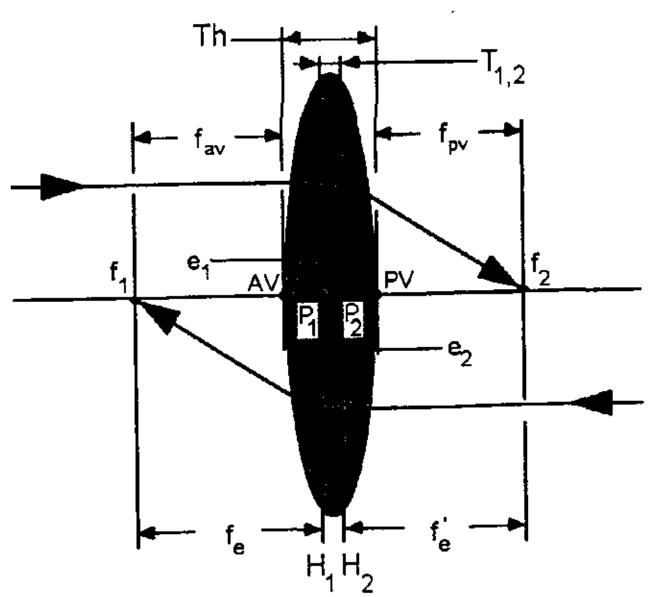

Figure 3. (Holladay) Primary and secondary principal planes and focal points.

Table 3. Relationship between shape factor and form of the lens.

Shape Factor	Form of Lens
> +1	Plus meniscus; more curved surface toward cornea
= +1	Convex-plano; curved surface toward cornea
< +1 but > 0	Biconvex; more curved surface toward cornea
= 0	Equi-convex
< 0 but > −1	Biconvex, more curved surface toward retina
= −1	Plano-convex; curved surface toward retina
< −1	Plus meniscus; more curved surface toward retina

dioptric power should be provided in a table or formula. These values would help investigators trying to reduce prediction errors in unusual eyes.

Lens Center Thickness

The center thickness of a lens is necessary, along with the shape factor, to determine the physical location of the lens vis à vis its principal planes. For positive lenses, the center thickness will increase with increasing lens power. Providing this information in tabular form or an equation should be easy. These values would also help investigators to reduce prediction errors in unusual eyes.

Lens-Haptic Plane Distance

The lens-haptic plane (LHP) distance can be calculated in the uncompressed state from currently available parameters such as the loop-to-loop diameter, haptic angulation, and optic diameter. The LHP in the compressed state within the bag is another matter. The vaulting characteristics of an IOL within the bag depend on many variables such as amount of bag contraction, vitreous pressure, haptic angle, and loop compressibility. Recent work has shown that the LHP may also help improve the predictability of the IOL's position within the eye.[49] Providing the value for LHP in a compressed state, e.g., 10.5 mm, is currently under consideration by the FDA.

Intraocular Lens Calculations Using Refractive Vergence Formula

Formula and Rationale for Using Preoperative Refraction Versus Axial Length

In a standard cataract removal with IOL implantation, the preoperative refraction is not useful in calculating the IOL power because the crystalline lens will be removed; thus, dioptric power is being removed and then replaced. In cases in which no power is being removed from the eye, such as (1) secondary IOLs for aphakia, (2) secondary piggyback IOLs for pseudophakia, and (3) a phakic minus or plus IOL for high myopia or hyperopia, respectively, the necessary power (IOL_e) for a desired postoperative refraction can be calculated from the corneal power and preoperative refraction—the axial length is not helpful. Since these lenses and procedures are becoming more common, a discussion of the standardization of these conditions, the relevant equations, and specific examples is appropriate.

The formula for calculating the necessary IOL power from the preoperative refraction is given is as follows:[50]

$$IOL_e = \frac{1336}{\dfrac{1336}{\dfrac{1000}{\dfrac{1000}{PreRx} - V} + K_o} - ELP_o} - \frac{1336}{\dfrac{1336}{\dfrac{1000}{\dfrac{1000}{DPostRx} - V} + K_o} - ELP_o} \qquad (12)$$

where ELP_o = expected thin-lens position in millimeters (distance from the secondary principal plane of the cornea, P_{C2}, to principal plane of the thin-IOL equivalent, ELP_o), IOL_e = IOL power in diopters, K_o = net corneal power in diopters, PreRx = preoperative refraction in diopters, DPostRx = desired postoperative refraction in diopters, and V = vertex distance in millimeters of refractions.

Intraocular Lens Calculation from Preoperative Refraction

As mentioned, there are three appropriate cases for using the preoperative refraction and corneal power to determine IOL power. In each of these cases, no dioptric power is being removed from the eye; the problem is to place the IOL at a given distance behind the cornea ELP_0 that is equivalent to the spectacle lens at a given vertex distance in front of the cornea. If emmetropia is not desired, the desired postoperative refraction (DPostRx) must be determined.[50]

Example: Secondary IOL for aphakia. The patient is 72 years old and is aphakic in the right eye and pseudophakic in the left. The right eye can no longer tolerate an aphakic contact lens. The capsule in the right eye is intact and a posterior chamber IOL is desired. The patient is −0.50 D in the left eye and would like to be the same in the right eye.

$$\begin{aligned} K_K &= 45.00\ \text{D} \\ \text{PreRx} &= +12.00\ \text{sphere @ vertex of } 14.0\ \text{mm} \\ \text{ELP}_0 &= 5.00\ \text{mm} \\ \text{DPostRx} &= -0.50\ \text{D} \end{aligned}$$

Using these input values and converting K_K to K_0 using equation 5, IOL_e is +22.90 D.

Example: Secondary piggyback IOL for pseudophakia. In patients with a significant residual refractive error following primary IOL implantation, it is often easier surgically and more predictable optically to leave the primary IOL in place and calculate the secondary piggyback IOL power to achieve the desired refraction. This method does not require knowledge of the power of the primary IOL or the axial length so it is particularly useful in cases in which the primary IOL may be mislabeled. The formula works for plus or minus lenses, but negative lenses are just becoming available.

The patient is 55 years old and had a refractive surprise after the primary cataract surgery. He was left with a +5.00 D spherical refraction in the right eye. There is no cataract in the left eye and he is plano. Both surgeon and patient desire him to be −0.50 D, which was the target for the primary IOL. The refractive surprise is believed to be from a mislabeled IOL that is centered in the bag and would be difficult to remove. The secondary piggyback IOL will be placed in the sulcus. This is important, since trying to place the second lens in the bag several weeks after the primary surgery is difficult. More important, it may displace the primary lens posteriorly, reducing its effective power and leaving the patient with a hyperopic error. Placing the lens in the sulcus minimizes this posterior displacement.

$$\begin{aligned} K_K &= 45.00\ \text{D} \\ \text{PreRx} &= +5.00\ \text{sphere @ vertex of } 14.0\ \text{mm} \\ \text{ELP}_0 &= 5.00\ \text{mm} \\ \text{DPostRx} &= -0.50\ \text{D} \end{aligned}$$

Using these input values and converting K_K to K_0 using equation 5, IOL_e is +8.64 D.

Example: Primary minus anterior chamber IOL in a highly myopic phakic patient. The calculation of a minus IOL in the anterior chamber is similar to the aphakic calculation of an anterior chamber lens except that the power of the lens is negative. In the past, these lenses have been reserved for high myopia that could not be corrected by radial keratotomy, photorefractive keratectomy, or laser in situ keratomileusis. Since these lenses fixate to the iris in the anterior chamber angle (ACL) or in the posterior chamber (intraocular contact lens), iritis and glaucoma are concerns. Nevertheless, several successful cases have been performed with good refractive results. Interestingly, the power of the negative IOLs is close to the spectacle refraction for normal vertex distances, whereas the plus lenses are approximately 1.5 times the spectacle refraction as seen in the previous two examples.

$$\begin{aligned} K_K &= 45.00\ \text{D} \\ \text{PreRx} &= -20.00\ \text{sphere @ vertex of } 14.0\ \text{mm} \\ \text{ELP}_0 &= 3.50\ \text{mm} \\ \text{DPostRx} &= -0.50\ \text{D} \end{aligned}$$

Using these input values and converting K_K to K_0 using equation 5, IOL_e is −18.49 D.

Determining the optimal ELP_o for a surgeon and manufacturer using the refraction vergence formula. Equations 13a through 13f are the quadratic solution of the refraction vergence formula for the ELP_o given the stabilized actual postoperative refraction (APostRx) and the actual power of the implanted IOL.[50]

$$X = \frac{1336}{\dfrac{1000}{\dfrac{1000}{\text{PreRx}} - V} + K_o} \tag{13a}$$

$$Y = \frac{1336}{\dfrac{1000}{\dfrac{1000}{\text{APostRx}} - V} + K_o} \tag{13b}$$

$$A = IOL_e \tag{13c}$$

$$B = -IOL_e * (X + Y) \tag{13d}$$

$$C = 1336(X - Y) + IOL_e * X * Y \tag{13e}$$

$$ELP_o = \frac{-B \pm \sqrt{B^2 - 4A * C}}{2A} \tag{13f}$$

where (+) is used for negative IOL and (−) for positive IOLs for the ± in equation 13f, ELP_o = expected thin lens position in millimeters (distance from the secondary principal plane of the cornea, P_{C2}, to principal plane of the thin IOL equivalent, ELP_o), IOL_e = IOL power in diopters, K_o = net corneal power in diopters, PreRx = preoperative refraction in diopters, APostRx = actual postoperative refraction in diopters, and V = vertex distance in millimeters of refractions.

Equations 13a to 13f allow surgeons to determine their optimal or personalized AVG_s ELP_o based on 20 to 30 cases with any style lens by taking the average of the back-calculated ELP_o for each case, similar to the vergence formula using axial length. The surgery and placement of the lens must be standardized as with the axial length vergence formula. A manufacturer would need a sampling of approximately 10 surgeons to produce a nominal or manufacturer's recommended *initial* AVG_m ELP_o *with a SEM less than* ±0.05 *mm.*

Table 4. Recommended values for ultrasonic biometry, keratometry, and IOL power calculation constants.

Eq. 1a $$AL_{1532} = \frac{1532}{1555} * AL_{1555}$$

AL_{1555} = axial length for 1555 m/s; AL_{1532} = *axial length for 1532 m/s*

Eq. 1c $$AL_u = AL_{1532} + 0.28$$

AL_u = true ultrasonic axial length

Eq. 3c $$AL_o = AL_u + 0.20 \text{ mm}$$

AL_o = optical axial length

Eq. 4b $$K_k = \frac{1.3375 - 1.000}{r_a} = \frac{0.3375}{r_a}$$

r_a = anterior radius of the cornea, K_K = keratometric corneal power

Eq. 5 $$K_o = K_k * \frac{4/3 - 1}{1.3375 - 1} = K_k * \frac{1/3}{0.3375}$$

K_o = optical net corneal power

Eq. 7a $$ELP_o = \frac{(Aconst * 0.5663) - 65.600 + 3.595}{0.9704}$$

Aconst = SRK A-constant; ELP_0 = effective lens position of thin IOL

Eq. 8a $$ELP_o = \frac{SF + 3.595}{0.9704}$$

SF = Holladay surgeon factor

Eq. 9a $$ELP_2 = ELP_0 - L_{2,0}$$

$L_{2,0}$ = *distance from* ELP_0 *of thin IOL to* ELP_2; ELP_2 = *distance to secondary principal plane of equivalent thick IOL*

Eq. 9b $$AV_{PC2} = ELP_2 - T_{1,2} - e_1$$

e_1 = distance *from front anterior vertex of thick IOL* to first principal plane of thick IOL; $T_{1,2}$5 distance from primary to secondary principal plane of thick IOL; AV_{PC2} = distance from anterior vertex of thick IOL to secondary principal plane of cornea

Eq. 9c $$AV_{PC1} = AV_{PC2} + P_{C2}$$

P_{C2} = distance from first to second principal plane of cornea; AV_{PC2} = distance from anterior vertex of thick IOL to primary principal plane of cornea (*anterior vertex of cornea*)

Eq. 11 $$\text{Shape factor} = \frac{C_1 + C_2}{C_1 - C_2}$$

C_1 = curvature of the anterior surface of thick IOL; C_2 = curvature of the posterior surface of thick IOL

Summary

The equations in Table 4 summarize the recommended values to be used to standardize ultrasonic biometry, keratometry, and IOL power calculation constants.

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