34 ELP Estimation

Lens Power Calculation Formulas

Thomas Olsen

ELP Estimation

The first-generation IOL power formulas were the so-called thin-lens formulas where the cornea and the IOL are regarded as single refracting planes. Examples of the thin-lens approach include the Fyorodov [1], Colenbrander [2], Binkhorst [3], Hoffer [4], Holladay [5], SRK/T [6], Haigis [7], and others. The basic formula is

$$P = \frac{n}{\mathrm{Ax} - \mathrm{ELP}} - \frac{1}{1/K - \mathrm{ELP}/n} \tag{34.1}$$

where P = IOL power of emmetropia, n = refractive index of aqueous/vitreous, Ax = axial length, K = corneal power, and ELP = estimated lens plane of the IOL. The logic of the formula is to subtract the vergence in front of the IOL (second term) from the vergence behind the IOL (first term) to give the IOL power needed for emmetropia.

Some caution should be taken about the term "ELP." The estimated lens plane (ELP) is often used to denote the value for the IOL plane to be used with the old thin-lens formulas. It is important to know that this need not be the physical position of the IOL but rather the value that predicts the observed refraction with that formula. Because the ELP in this way is a back-calculated value it becomes a virtual distance that may work to absorb any other off-set errors in the system, much like the A-constant works for the SRK formulas. To distinguish between the ELP as a virtual distance and the actual, physical position, it has been suggested to use alternate terms like the physical lens position (PLP) or the actual lens position (ALP).

Apart from questions about the *K*-reading and the axial length, the obvious unknown in Eq. (34.1) is of course the final location of the IOL in the eye after surgery. All right, we know the placement of the IOL is often in-the-bag (Fig. 34.1), but the exact location cannot be predicted on theoretical grounds. Factors like optic and haptic design [8], surgical technique, size of the capsular opening, capsular bag shrinkage, and possible change over time add uncertainty to the prediction. Remember that ±0.7 mm axial displacement of the IOL is the equivalent to a ±1 D shift in IOL power in a normal sized eye. The effect is, however, very dependent on the axial length of the eye as shown in Fig. 34.2, where the Rx error per mm change in ELP (IOL position) has been calculated in a real-world simulation dataset of 2870 eyes and plotted against the axial length. As can be seen the error amounts to about 1.4 D/mm for a 24 mm eye but doubles for an eye shorter than 20 mm and approaches zero for a long eye. The minus value in some of the very long eyes is due to the minus powered IOL. Note, however, that the accuracy of the IOL position does not matter much for a long eye because the IOL power is low.

T. Olsen (✉)
Aros Private Hospital, Aarhus N, Denmark

J. Aramberri et al. (eds.), *Intraocular Lens Calculations*, Essentials in Ophthalmology,
https://doi.org/10.1007/978-3-031-50666-6_34

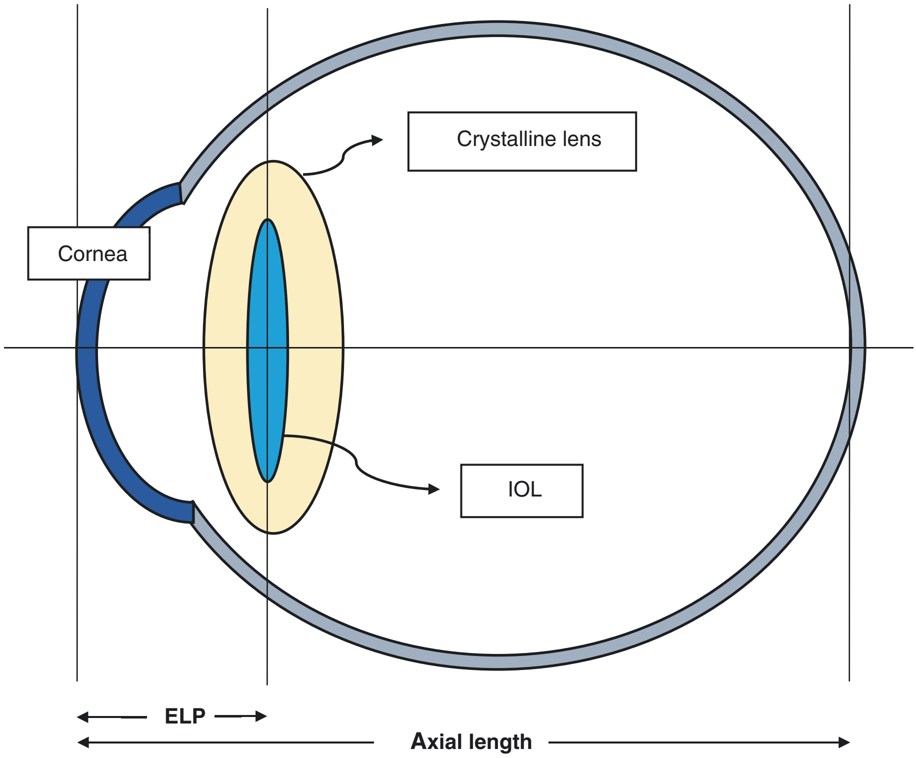
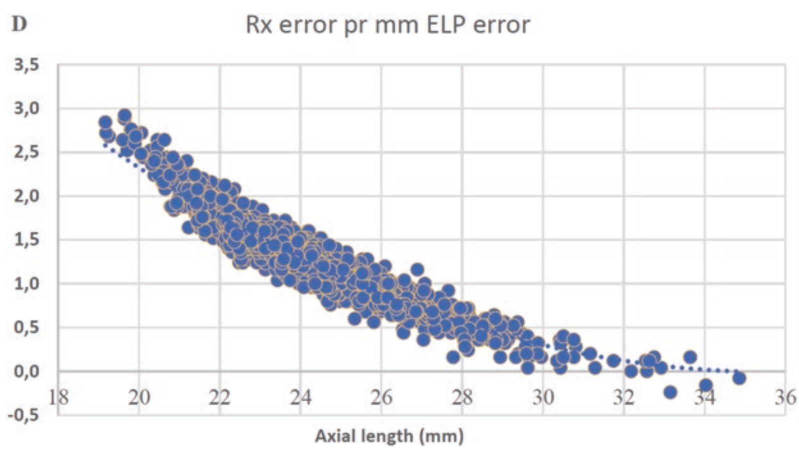


Fig. 34.1 The estimated lens plane (ELP) refers to the plane of the IOL after surgery

Fig. 34.2 The Rx error per mm change in ELP (IOL position) calculated in a clinical dataset of 2870 eyes



Methods to Estimate the ELP

The first IOL power formula in the world was described by S Fyodorov in 1967 [8] in a Russian paper and 1975 republished in Invest Ophthalmol [9]. To estimate the ELP, he used the height of the corneal dome from the iris plane based on Pythagoras theorem:

$$\mathrm{ELP} = r - \sqrt{r^2 - d^2/4} \tag{34.2}$$

where r = corneal radius, d = corneal diameter (taken as the corneal diameter plus 10%). This method was developed for iris-clip lenses after intracapsular extraction which was popular at that time. The idea of using the *K*-reading and corneal diameter has later been taken up by several authors as one of the predictors for the ELP for modern posterior chamber IOLs. Now, more than 50 years since the paper by Fyodorov, you can still find this ELP concept inside the SRK/T and the Holladay formulas.

A common procedure of many formulas has been to back-calculate for the ELP based on the actual outcome: In each case, the ELP is solved that gives the actual outcome, and statistical analysis is applied to find the covariation with possible predictors in a representative sample. The statistical ELP dependence—typically a regression equation—is then incorporated into the formula. In this way, the formula can be made to work even if the optical model of the formula may not be correct! For example, what happens if the corneal power is input as the *K*-reading (and we know this may be a falsely high value)? The formula would need to move the ELP a little further back to work. This underlines the fact that the ELP calculated in this way is not a physical distance but rather a virtual distance which cannot be verified by direct measurement of the IOL position.

It has been common practice for IOL manufacturers to state the ELP on the IOL label along with the A-constant. As far as the author knows, this ELP refers to the old Binkhorst formula. The reader may have noticed that the labeled "ELP" value typically reads more than 5 mm, which is higher than the real position found after surgery from actual measurements. The explanation is the *K*-reading issue as mentioned above (Binkhorst uses keratometer index 1.3333 rather than 1.3375 originally advocated to account for some flattening of the cornea after surgery). Some of this confusion may be avoided if the formula does not take the *K*-value directly from standard keratometry but takes the corneal radius as a parameter. Still, the radius needs to be converted to a corneal power inside the formula.

Many methods have been suggested to model the ELP prediction and each formula has its own. In the early days of IOL power formulas, the ELP was expressed as a function of the *K*-reading (Fyodorov) and the axial length (Binkhorst) with various mathematical representation. As more clinical data became available in larger series, other parameters like corneal diameter, anterior chamber depth, lens thickness, and other factors like age, sex, and refraction have been tried. Table 34.1 is a summary of some of the suggested predictors of the ELP in the various formulas.

Table 34.1 ELP predictors used by different authors of some optical formulas

Formula	Axial length	*K*-reading	ACD	Lens thickness	Other
Fyodorov	–	X	–	–	–
Binkhorst	X	–	–	–	
SRK/T	X	X	–	–	–
Hoffer Q	X	X	–	–	–
Holladay I	X	X	–	–	
Holladay II	X	X	X	X	CD [a], Rx [a], age [a]
Haigis	X	–	X	–	–
Olsen	(x)	(x)	X	X	–
Preussner	X	X	X	X	
Barrett II	X	X	X	X [a]	CD [a], Rx
Kane	X	X	X	X [a]	Gender, CCT [a]

ACD preoperative anterior chamber depth, *CD* corneal diameter, *Rx* preoperative refraction, *CCT* central corneal thickness

[a] Optional

Beware the Unusual Eyes!

As mentioned, for optimization purposes, the ELP is often back-calculated as the value that will “predict” the outcome with a given formula. When this virtual distance is correlated with all available parameters like axial length, *K*-reading, ACD, lens thickness, corneal diameter distance, corneal thickness, refraction, gender, age, shoe size (sorry, not published), and subjected to a data cruncher, it often happens that small correlations are found that will tend to improve the refractive predictions with a small statistical significance. However, as is the case with statistical analysis, the correlations are strictly speaking only valid for the dataset on which the analysis was performed, and care has to be taken when we move outside the normality.

A classic example is the post-LASIK cases where the anatomy of the cornea has changed so that the *K*-reading is not representative of the true corneal power in the first place but also cannot be used as a predictor for the ELP in the second place as the Fyodorov “height” formula (used by the SRK/T and the Holladay formulas) is based on a normal anterior segment. For such cases, it has been suggested to use the so-called double K method principle [10] where the ELP dependence is replaced by the pre-LASIK value or a standard value. These considerations also apply to keratoconus, megalocornea, keratoplasties, and other abnormal cornea with a disrupted anterior segment.

Another example is the use of the preoperative refraction for the prediction of the ELP. This variable may be shown to have a small influence in a large sample. However, what happens in case of lenticular myopia? This is outside the normal covariation between the refractive components of the eye and can lead to a gross error if included as a predictor.

So, each formula has its limitations, often to be found in the “engine room” of the formula, i.e., the ELP method. Especially methods that use multiple predictors have a risk of being misguided if one of the predictors is out-of-range. Eventually, it is up to the user to identify those outliers and maybe switch to another formula if an error is anticipated. Therefore, careful screening of patients scheduled for lens surgery is highly recommended.

The C-Constant

Optical biometry (Zeiss IOLMaster 500) was originally introduced for the measurement of axial length by partial coherence interferometry (PCI). However, the measurement of the ACD with the IOLMaster was still based on a slit-lamp technique. A decade later, Haag-Streit introduced another optical biometer called the Lenstar LS 900. The working principle of the Lenstar was optical low coherence reflectometry (OLCR) which has some advantages over PCI in the extended range of measurement, covering all the intraocular distances including the corneal thickness, the ACD, and the lens thickness.

For the prediction of the ELP, many previous studies (see section above) had shown a significant role of both the preoperative ACD and lens thickness, but those studies were mainly based on ultrasound biometry. The question that may be asked is this: given the new accuracy of the laser biometer for all intraocular distances, do we have better options for the prediction of the IOL position?

Studies were undertaken by the author to measure the actual IOL position routinely after surgery in a series of cataract cases and to establish the possible predictive value of all available predictors: *K*-reading, axial length, anterior chamber depth, lens thickness, Corneal Diameter distances all of which were measured by the Lenstar biometer (Fig. 34.3). For the present chapter, a reanalysis was made on the database collected over the years while working at the University department. It included the original dataset from 2014 [11] and additional 200 cases, making a total 1622 cases.

In Fig. 34.4, the position of the IOL (measured by OLCR optical biometry) has been plotted against the axial length as well as the preoperative position of the anterior and posterior capsule of

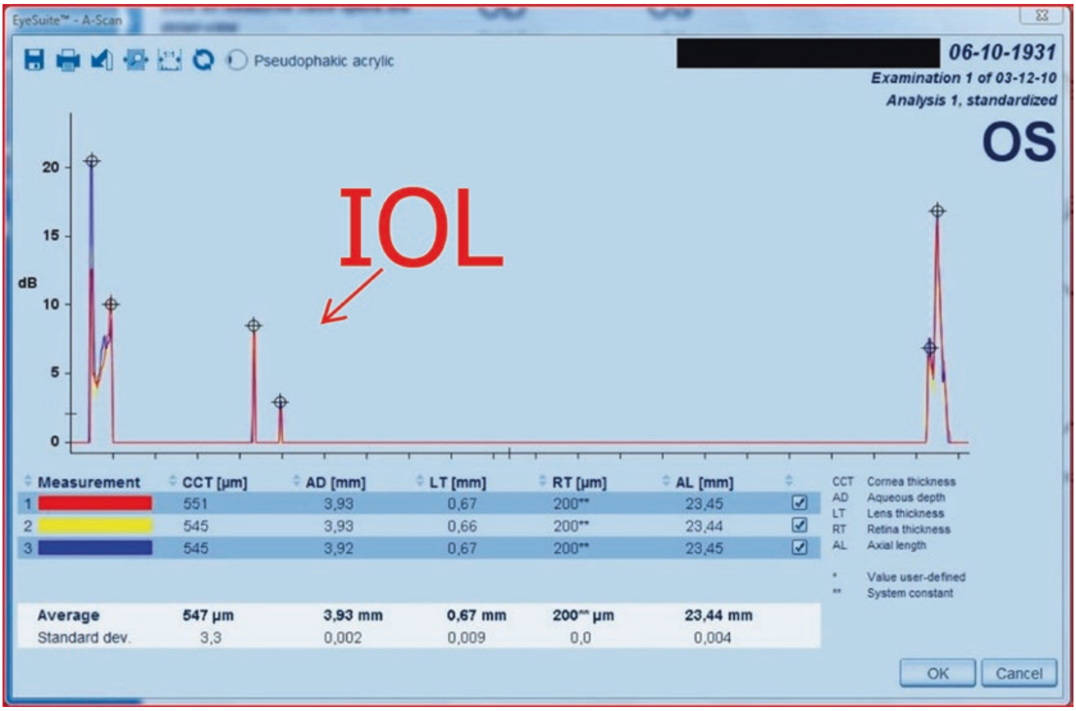


Fig. 34.3 IOL position measured by laser biometry

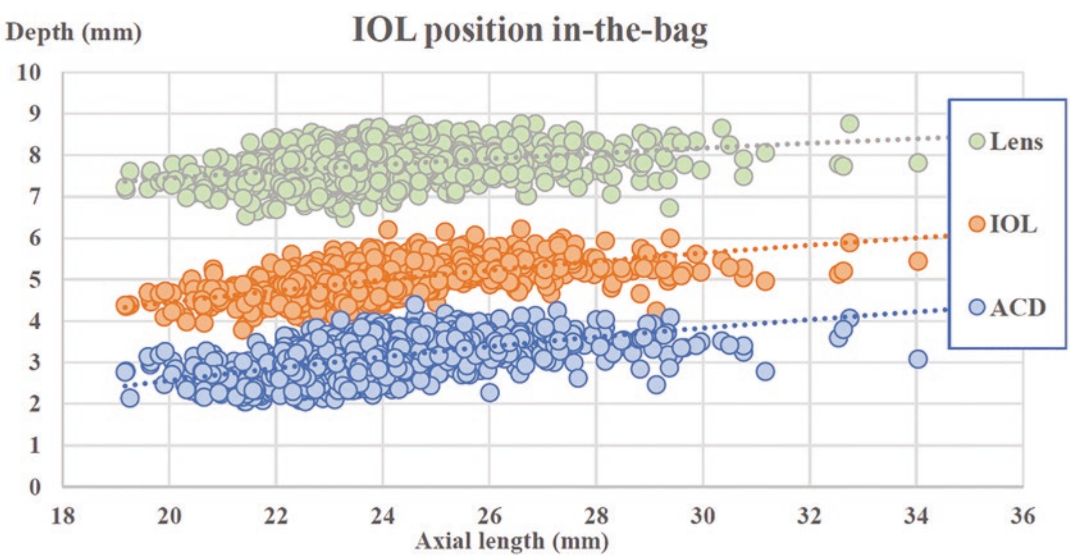


Fig. 34.4 IOL position vs axial length and located of anterior and posterior capsule of the lens

the crystalline lens. As can be seen, the postoperative IOL position was tightly connected to both the preoperative ACD and the lens thickness in a way that clearly depicted the in-the-bag placement of the IOL. The IOL appeared to locate itself at a constant fraction (around 40%) of the space between the anterior and the posterior capsule (= lens thickness), irrespective of the axial length.

Thus, the IOL position could be described as

$$\text{IOLpost} = \text{ACDpre} + C \times \text{LensT} \quad (34.3)$$

where IOLpost is the postoperative position of the IOL center, ACDpre is the preoperative anterior chamber depth, LensT is the preoperative lens thickness, and C is the constant predicting the axial position of the IOL center (Fig. 34.5). The postoperative ACD can be found by subtracting half of the IOL thickness from the IOLpost.

Despite its simple form, statistical analysis showed the method to be highly effective. One of the advantages is that it works without the indirect predictors like *K*-reading and axial length and the principle is less prone to abnormal K's (post-LASIK) or conditions causing a disproportionate relationship between the anterior segment and length of the eye. What matters is the position and dimension of the crystalline lens which is the target of the surgery.

Of course, there must be different C-constants for different lens types, depending on the haptics, the shape of the optic, and the behavior of the IOL inside the bag after surgery as a result of capsular contraction. Much like the A-constant summarizes the refractive effect of a given lens type, the C-constant describes the IOL-specific anatomic relationship with the capsular bag. However, whereas the A-constant includes the optical properties of the IOL, the C-constant only describes the physical location of the IOL. The optical properties like optic configuration and wavefront correction of spherical aberration must be accounted for separately. With the Olsen formula, these optical properties are included in the IOL settings for the given IOL. This means for each IOL type, the refractive index, the (average) curvature of front and back surface of the IOL, the thickness and the wavefront correction of spherical aberration, if any, must be stated. The reader might argue that the curvature of the IOL surfaces varies according to the power and this is true. However, according to the ANSI standard, an IOL power is labeled as the paraxial power, and it is therefore possible to calculate the curvatures for a given IOL power as long as the overall shape of the optic configuration is known (biconvex 1:2, biconvex 1:1, biconvex 2:1, etc.). This is done internally by the Olsen formula from the average IOL definition. As a result, it is possible to model the exact physical properties of the IOL eye, which can be used for ray tracing and further optical analysis.

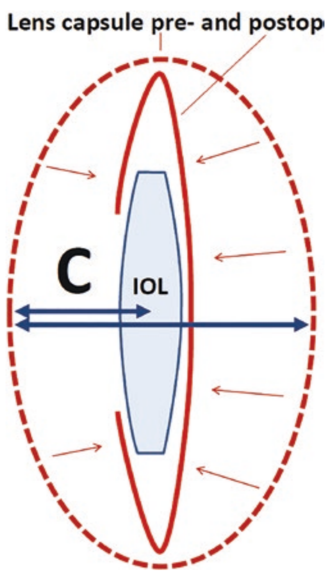


Fig. 34.5 The C-constant predicts the location of the IOL as a fraction of crystalline lens thickness

Error Propagation Model

No matter how good the biometry or the formula is, a statistical error will always be associated with the refractive predictions. You may divide this residual error into measurement errors and formula errors.

One important source of error to be considered is the measurement error of the axial length. In the old days of ultrasound biometry, this was a major source of error. What is measured is the transit time of ultrasound traveling from the corneal surface to the vitreoretinal interface. The time is translated into distance assuming a certain velocity of sound through the ocular media. Many uncertainties exist by this technique: possible indentation of the cornea, alignment issues, velocity settings, impact of the cataractous lens, retinal thickness, and the fact that there is a limit to the resolution given by wavelength of ultrasound. According to the author's experience, the reproducibility of good ultrasound readings is typically within ±0.2 mm. Recalling that 1 mm of error in axial length amounts to 2.5 D error in the spectacle plane, the ultrasound reproducibility of 0.2 mm is the equivalent of 0.5 D error in the spectacle plane.

The introduction of optical biometry more than 20 years ago [12] was a quantum leap in the era of IOL power calculation. First, the wavelength of light is so much shorter than that of ultrasound giving an ultrahigh tissue resolution. (A laser wavelength of 1060 nm corresponds to about 800 nm in ocular tissue and 10 MHz ultrasound with velocity of 1550 m/s in the eye has a wavelength of about 0.16 mm.) Second, the measurements are performed contact free in the line of sight and the end point is the pigment epithelium. So, the measurements are less prone to alignment issues, and the off-set issues of ultrasound like deformation and the question of retinal thickness do not exist. It should be remembered, however, that the laser does not measure the geometrical distance directly. What is measured is the time—or optical path—for light to travel from the corneal to the retinal reflection. Akin to the velocity issue of ultrasound we need to assume a refractive index of the ocular media in order to translate the optical path into geometrical distance. The group refractive index used by the IOLMaster was calculated by Haigis [13] who calibrated the laser readings against the results of immersion ultrasound, assuming this was the true distance measurement. By doing this, the output reading of the IOLMaster was in reality similar to that measured by ultrasound. The advantage of this calibration was no need to change existing IOL constants based on numerous ultrasound measurements.

It has been questioned by the author whether the Haigis group refractive index of the phakic eye was indeed the most accurate. The question arose from the observation that there is a systematic difference between pre- and postoperative readings with the IOLMaster. The difference amounted to 0.08 mm shorter readings of the IOL eye as compared to the preoperative phakic eye. There is no reason to believe that the eye shortens by the surgery so the explanation must be found in the assumed refractive indices of the ocular components, in particular the crystalline lens which is hard to examine. The author has shown that if the index of the crystalline lens is changed from the Haigis assumed value of 1.407 to 1.429, there will be consistency between the preoperative and the postoperative measurements [14]. With the Olsen calibration, the overall group refractive index of the phakic eye changes from 1.3574 to 1.3616. The difference is slight in the normal range but becomes larger in the longer eyes.

Whatever calibration of the optical biometer, the reproducibility of measurements with optical biometry is impressive and readings often fall within 0.02 mm. So, if optical biometry was the only source of error in the system, the refractive predictions would be within 0.05 D error only (!). However, as everyone knows this accuracy is not achieved in clinical work and therefore other errors must be at work.

Keratometry must also be considered as a significant source of error. Generally, autokeratometry tends to give good readings if one pays attention to the quality of the tear film, focus, alignment issues, lid pressure, contact lens wear, and other confounders. Beware the post-LASIK cases, keratoconus, high astigmatism and other odd cases. However, even "perfect" readings do have a variation and it may sometimes be wise to repeat the measurement with days apart to have consistent readings. It is not just about the spherical equivalent but also about the astigmatism that need to be assessed accurately. It is the experience of the author that the error of good, consistent *K*-readings should be well below 0.1 D (spherical equivalent) or better.

The most critical formula error is, however, the error associated with the prediction of the IOL position (ELP). If we were able to predict the ELP with 100% accuracy, the only source of error would be the measurement error associated with the *K*-reading and the axial length. It may be difficult to assess the error of ELP prediction. First of all the ELP in many formulas is not a physical distance but rather a virtual distance calculated in retrospect and therefore not directly measurable. One exception to this is the Olsen formula which was designed to use the physical dimensions all through the calculations. This includes the shape of the IOL as well as the real pseudophakic ACD.

An error propagation model of the total error associated with IOL power calculation was first published by Olsen in 1992 [15] and by Norrby in 2008 [16]. The assumption is that the total error is the sum of individual and independent components. The individual sources of error mainly consist of measurement errors from keratometry and axial length measurements. For completeness we also need to consider the process of taking the refraction itself as recommended by Norrby and probably other factors like pupil size, variations in Gullstrand ratio of the cornea, IOL tilt and IOL power tolerance. However, the most important source of error—and we shall see how important—is the error associated with the prediction of the ELP.

According to the error propagation model, if we know the error of each component, we can calculate the total error by adding the variances of each component and take the square root of the sum. In our case, we have

$$\delta(\text{Total}) = \sqrt{\delta^2(\text{Ax}) + \delta^2(K) + \delta^2(\text{ELP}) + \delta^2(\text{Rx}+)} \tag{34.4}$$

where δ(Total) = total error of the IOL power prediction as standard deviation, δ(Ax) = error of axial length, $\delta(K)$ = error of keratometry, δ(ELP) = error of ELP prediction, and δ(Rx+) = error of taking the refraction and other errors.

How do we assess the error of each component? One method would be simply to take a number of measurements and calculate the error between repeated measurements. In this way, we get the intra-session error, but this need not be the real variability because of day-to-day variation in tear film, intraocular pressure, pupil size, observer dependent bias, etc. In the attempt to estimate the total error, Olsen in his 1992 publication estimated the variation between pre- and postoperative measurements, thereby including the surgical influence. However, at that time ultrasound was used for biometry and other instrumentation like keratometry may not be representative of modern technique with optical biometry, accurate keratometry with confirmation from several devices, standardized small-incision surgery with capsulorrhexis, and in-the-bag placement of the IOL and improved ELP prediction.

As mentioned above, the difference between repeated optical biometry readings is often within 0.02 mm. So, a conservative estimate of the standard deviation might be in the region of 0.03 mm. This is the equivalent of 0.075 D in the spectacle plane. For keratometry there is one study comparing the inter-session variability of different keratometry devices [17] showing standard deviations from 0.12 D (Nidek TonoRef II) to 0.17 D (IOLMaster 500). The author has a preference of using autokeratometry and therefore a reasonable estimate might be 0.15 D for the standard deviation of keratometry.

The error predicting the IOL position can be assessed by measuring the postoperative anterior chamber depth and comparing with the predicted value. As mentioned above, this is not possible with the standard thin-lens formulas because the ELP is a virtual distance. However, with the Olsen formula, this is possible because the formula was developed to accept the physical (measurable) dimensions all through the calculations. In the paper describing the C-constant for prediction of the IOL position [18], the mean difference between the expected and the observed IOL position as measured by laser biometry (Lenstar) was 0.0 ± 0.17 mm (SD). This corresponds to 85.9% of the cases within ±0.25 mm difference. The observed error may of course include some measurement error but for now a reasonable estimate might be to use the value 0.17 mm, which corresponds to 0.28 D error in the spectacle plane.

Finally, some error will arise from taking the refraction itself and other sources. Norrby [16] cites a study on 80 patients aged 11–60 years by Bullimore [18] who found the 95% limits of agreement between automated and manual refraction ranged from −0.90 to +0.65 D with an SD of 0.39 D. To the author this seems to be a huge variability and difficult to extrapolate to a clinical setting with premium implants where patients may be intolerant to variations in the refraction of a quarter of a diopter.

The reproducibility of manifest refraction was recently reported by Taneri et al. [19], who studied the latest 2 manifest refractions of 1000 eyes obtained at 2 separate visits. The study population was mostly myopic with a median age of 35 years. They found a standard deviation of the pairwise difference of 0.19 D. One might argue that accurate refractions are more difficult in young, phakic patients as compared to pseudophakic patients. For the present study and considering the difference between phakic and pseudophakia patients, the author believes a reasonable estimate for the error to be 0.20 D (standard deviation).

Having defined the error of these four individual components, the calculation of the total error is straightforward as shown in Table 34.2. The variances in D units are calculated for each component and summed to give the total variance of the model. The total standard deviation is then found as the square root of the total variance. In the numerical example, an SD of 0.385 D was found. This corresponds to a mean absolute error (MAE) of 0.308 D with 81% of the cases within 0.5 D prediction error. This is not far from reality in the author's own clinical experience.

The relative contribution of the different components of error is shown in Fig. 34.6. Note the small contribution of the axial length and the dominant contribution of the ELP prediction accounting for more than 50% of the total error. To improve the accuracy further, we need to improve the prediction of the ELP.

The reader is asked to copy the scheme of Table 34.2 into a spreadsheet and see what impact a change in error of each of the four components will have on the total error. In this way, we can predict the limits of accuracy based on the error of each component. There is no magic.

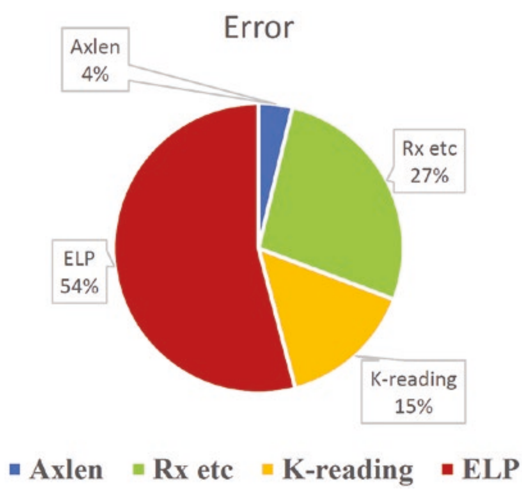


Fig. 34.6 The components of error in IOL power calculation

Table 34.2 Error propagation model of total IOL prediction error

Source of error	Error (SD)	Rx (SD)	Variance (SD^2)	Per cent
ELP, mm	0.17	0.28	0.0803	54.1
Rx, other, D	0.20	0.20	0.0225	15.2
Keratometry, D	0.15	0.15	0.0144	13.1
Axial length, mm	0.03	0.075	0,0056	3.8
Total, D	0.38	<<<	0.1484	100

References

1. Fedorov SN, Kolinko AI, Kolinko AI. Estimation of optical power of the intraocular lens [Russian]. Vestn oftalmol. 1967;80(4):27–31.
2. Colenbrander MC. Calculation of the power of an iris clip lens for distant vision. Br J Ophthalmol. 1973;57:735–40.
3. Binkhorst CD. Dioptrienzahl Künstlicher Augenlinsen. Klin Monatsbl Augenheilkd. 1973;162:354–61.
4. Hoffer KJ, Allen DR. A simple lens power calculation program for the HP-67 and HP-97 calculators. J Cataract Refract Surg. 1978;4:197–9.
5. Holladay JT, et al. A three-part system for refining intraocular lens power calculations. J Cataract Refract Surg. 1988;14:17–24.
6. Retzlaff JA, Sanders DR, Kraff MC. Development of the SRK/T intraocular lens implant power calculation formula. J Cataract Refract Surg. 1990;16:333–40. Errata: 1990;16:528 and 1993;19(5):444–446.
7. Haigis W, et al. Comparison of immersion ultrasound biometry and partial coherence interferometry for intraocular lens calculation according to Haigis. Graefes Arch Clin Exp Ophthalmol. 2000;238:765–73.
8. Norrby S. Using the lens haptic plane concept and thick-lens ray tracing to calculate intraocular lens power. J Cataract Refract Surg. 2004;30(5):1000–5.
9. Fyodorov SN, Galin MA, Linksz A. Calculation of the optical power of intraocular lenses. Investig Ophthalmol. 1975;14:625–8.
10. Aramberri J. Intraocular lens power calculation after corneal refractive surgery: Double-K method. J Cataract Refract Surg. 2003;29:2063–8.
11. Olsen T, Hoffmann P. C constant: new concept for ray tracing–assisted. Intraocular lens power calculation. J Cataract Refract Surg. 2014;40:764–73.
12. Drexler W, Findl O, Menapace R, et al. Partial coherence interferometry: a novel approach to biometry in cataract surgery. Am J Ophthalmol. 1998;126:524–34.
13. Haigis W. Pseudophakic correction factors for optical biometry. Graefes Arch Clin Exp Ophthalmol. 2005;239:589–98.
14. Olsen T, Thorwest M. Calibration of axial length measurements with the Zeiss IOLMaster. J Cataract Refract Surg. 2005;31:1345–50.
15. Olsen T. Sources of error in intraocular lens power calculation. J Cataract Refract Surg. 1992;16:125–9.
16. Norrby S. Sources of error in intraocular lens power calculation. J Cataract Refract Surg. 2008;34:368–76.
17. Laursen JVN, Jeppesen P, Olsen T. Precision of 5 different keratometry devices. Int Ophthalmol. 2016;36:17–20.
18. Bullimore MA, Fusaro RE, Adams CW. The repeatability of automated and clinician refraction. Optom Vis Sci. 1998;75:617–22.
19. Taneri S, Arba-Mosquera S, Rost A, Kießler S, Dick HB. Repeatability and reproducibility of manifest refraction. J Cataract Refract Surg. 2020;46:1659–66.