ARTICLE

Intraocular lens calculation accuracy limits in normal eyes

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PURPOSE: To quantify the current accuracy limits, analyze the residual errors, and propose the next steps for prediction accuracy improvements.

SETTING: Eye hospitals in Germany, Denmark, and Austria.

METHOD: Numerical ray tracing using manufacturer's intraocular lens (IOL) data (vertex radii, central thickness, refractive index) was used for all calculations. Postoperative lens position was predicted by a simple scaling model based on measurements in 1 patient collective. The model was compared with 2 other approaches in 2 patient collectives at 2 hospitals (1121 eyes with 13 IOL models; 936 eyes with 2 models). Axial lengths were measured optically (IOLMaster, Zeiss). No parameter adjustments or individualization of IOL types or of surgeons/localizations were done. The prediction errors and measures of systematic bias for short or long eyes were used to quantify the outcome.

RESULTS: The mean prediction errors in the 2 collectives were +0.13 diopter (D) and −0.13 D and the mean absolute errors were 0.44 D and 0.50 D without bias for long or short eyes, but depending on the IOL position model approach. The differences in the mean prediction errors for the IOL types were below the allowed manufacturing tolerances and below human recognition thresholds.

CONCLUSIONS: The need to individualize and fudge parameters decreases with better physical models of the pseudophakic eye. Further improvements are possible by individual topography to extract corneal asphericity and measured pupil size to calculate the best focus, by improved position predictions based on individual measurements of the crystalline lens and by smaller tolerances for IOL manufacturing.

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Intraocular lens (IOL) calculations based on Gaussian optics were first reported by Fyodorov[1,2] and Gernet et al.[3] Similar approaches that in part differ only in notation have been proposed by many other authors.[4–9] The implied assumption of a constant postoperative anterior chamber depth (ACD) assumed in the creation of these calculations was too inaccurate. Newer formulas, therefore, allow an individual value for this parameter.

As the nonnegligible errors from formulas based on Gaussian optics were difficult to understand and even more difficult to avoid, Retzlaff[10] and Sanders and Kraff[11] developed an "empirical" approach (SRK I formula) that was simpler and, on average, more accurate. The authors use a fit in which the corneal power (or corneal radius) and axial length enter only linearly. For each IOL type, an adjusting A constant is fitted from a large patient collective. Therefore, on average, the SRK I formula is accurate by definition because otherwise, the A constant must be modified.

Other authors combine analytical calculations in Gaussian optics and empirical fits to improve the results, particularly in longer and shorter eyes. In addition, they include adjusting factors; for example, the surgeon factor[12] or a factor for retinal thickness.[13–17] Unfortunately, the amounts of the different error contributions are not transparent in these formulas,

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particularly if the parameters are adjusted by statistical methods only, ignoring the underlying influences.

From a physical point of view, it is difficult to accept that different formulas giving different results can be used to describe the same optical problem. In fact, the formulas are mostly a mixture of exact calculations on the one hand and fudging procedures on the other hand. Unfortunately, the fudging parts (eg, the assumptions about postoperative IOL position) are often hidden. Other fudging contributions result from the fact that the description of the human eye by Gaussian optics is a poor approximation.

In contradiction to all efforts in physics and technology that have tried to standardize and calibrate methods, individualization of formula parameters has been proposed.[18] These parameters are inconsistently often called formula constants, even if they are not constant but are rather individual.

This study had 3 aims: (1) to make the different error contributions more transparent, (2) to show the current error amounts, and (3) to quantify the errors if methods calibrated for 1 patient collective are used in other collectives treated by different surgeons. The 4 principal error contributions that must be clearly separated from one another are biometric measurement errors, IOL power errors, IOL power calculation errors, and prediction errors of the postoperative IOL position (eg, ACD).

Calculation errors, intrinsic to Gaussian optics, can be avoided if formulas are replaced by numerical ray tracing. As input, this method requires true geometric data (eg, distances, local radii) of all refracting surfaces as well as the corresponding refractive indices. Derived values such as powers and principal planes are not used. Intraocular lens powers are used for labels only.

PATIENTS AND METHODS

Measurements of Axial Eye Lengths and Corneal Radii

All measurements were done with the IOLMaster (Zeiss). The length values A_i given by the instrument were transformed using the following equation: $A_t = 0.9479 \times A_i + 1.0848$. This transformation gave zero prediction error and zero steepness of the regression line fitted to the prediction error for a collective of 189 eyes at the Department of Ophthalmology, Medical University of Vienna; that is, this collective served for a calibration of the IOLMaster different from the original one.[19] Such methods have also been proposed by Norrby et al.[20] For this calibration, the IOL position was taken from the PCI measurements in these eyes (see below). Corneal vertex radii were the unchanged readouts of the IOLMaster.

Prediction of Postoperative Intraocular Lens Position

Axial Length–Based Prediction Model (Model 1) In the patient collective from Vienna University Eye Hospital, postoperative IOL positions were measured with high accuracy by partial coherence interferometry (PCI),[21] thus allowing the development of model assumptions for the postoperative IOL position that are described in detail by Kriechbaum et al.[22] and Preussner et al.[23] and summarized below.

1. For an average-sized eye, the center (half the thickness) of all investigated IOL types has the same location, which is 4.6 mm behind the posterior corneal vertex. This position $P_m = 4.6$ mm was found to be independent of the IOL haptics. For the mean postoperative ACD of a particular IOL model, only half the IOL thickness must be subtracted.
2. For the individual eye, the IOL center position (P_i) is nearly linearly scaled with the individual axial length A_i, similarly as proposed by Hoffer[24]: $P_i = P_m \times (A_i/A_m)^{0.7}$, where A_m is the mean axial length (23.6 mm). The exponent of 0.7 results in little deviation from linearity. It was introduced to slightly improve the results for very long and very short eyes when comparing the predicted IOL position and the measured IOL position. The individual ACD is given by the individual center position, subtracted by half the individual IOL thickness. This ACD prediction model, derived from 189 eyes of the Vienna collective, was applied to the 2 other patient collectives (see below).

Five-Parameter Prediction Model (Model 2) The prediction model recently proposed by Olsen[25] was also applied for comparison. In this model, the ACD depends linearly on 5 parameters: axial length, spherical equivalent of the preoperative refraction, corneal radius, preoperative ACD, and thickness of the crystalline lens. This model could only be applied to the Aarhus patients because only this group had the necessary data, in particular the preoperative ACD and the thickness of the crystalline lens.

Crystalline Lens–Dependent Model (Model 3) In this model, only 2 data of the crystalline lens are used: preoperative ACD and thickness of the crystalline lens. This model, however, was derived from the data to which it should be applied (ie, retrospectively). Again, this model could be applied only to the Aarhus patients.[26] It is similar to an approach proposed by Norrby et al.[20] but does not refer to the "lens haptic plane," as proposed in this paper. Rather, it refers to the "IOL center."

Calculations

The calculations were performed by a previously described full-aperture ray-tracing procedure[27–29] using the manufacturer's original IOL data. The best-focus ray in these calculations is the ray intersecting the pupil plane at a distance of $d = 0.25\sqrt{2} \times p$ from the optical axis, where p is the pupil diameter. Calculations were performed for different pupil diameters and different corneal asphericities. For the pupil width, the true values in pupil plane were used; pupil widths measured by most pupillometers are defined with corneal magnification of $\approx 16\%$.

Patients

One patient collective comprised 1122 normal eyes of 1121 patients who had uneventful small-incision cataract extraction with continuous curvilinear capsulorhexis (CCC) and posterior chamber IOL implantation at Eye Hospital

Castrop-Rauxel, Germany. Figure 1 shows the IOL types and the number of eyes with these IOLs.

In a second patient collective, 936 normal eyes of 936 patients had uneventful small-incision cataract extraction with CCC and posterior chamber IOL implantation at University Eye Hospital Aarhus, Denmark. Figure 2 shows the IOL types and the number of eyes with these IOLs.

RESULTS

Prediction Errors of Postoperative Refraction

The results in Figures 1 and 2 were calculated under the assumption of a pupil width of 2.5 mm and a corneal numerical eccentricity of $e = 0.5$ (or an asphericity index $Q = -0.25$ with $Q = -e^2$). This assumption was not confirmed by measurements. The ACD predictions were according to Model 1.

Under these circumstances, the mean prediction errors were +0.13 diopter (D) $\pm$ 0.59 (SD) for the Castrop-Rauxel patients and -0.13 ± 0.62 D for the Aarhus patients. The mean absolute errors were 0.44 D and 0.50 D, respectively.

Figure 2 also shows the IOL power tolerances according to International Organization of Standardization ISO 11979 (Available at http://www.iso.org/iso/iso_catalogue/catalogue_ics/catalogue_detail_ics.htm?csnumber=34064&ICS1=11&ICS2=040&ICS3=70. Accessed February 321, 2008). The figure contains 2 simplifications that do not, however, change the message of the graph or the order of magnitude of the error amount.

1. ISO 11979 refers to IOL powers rather than to axial lengths. The locations of the power steps (15.0 D, 25.0 D, and 30.0 D) are for an average-sized eye in which these IOL powers result in emmetropia with the corresponding axial length. For the real data, confusing scattering would occur instead of steps.
2. The ratio between the IOL power error inside the eye and the corresponding refraction error in the corneal plane has been assumed to be 3:2. In fact, it individually depends on IOL position, once again resulting in scattering instead of straight lines.

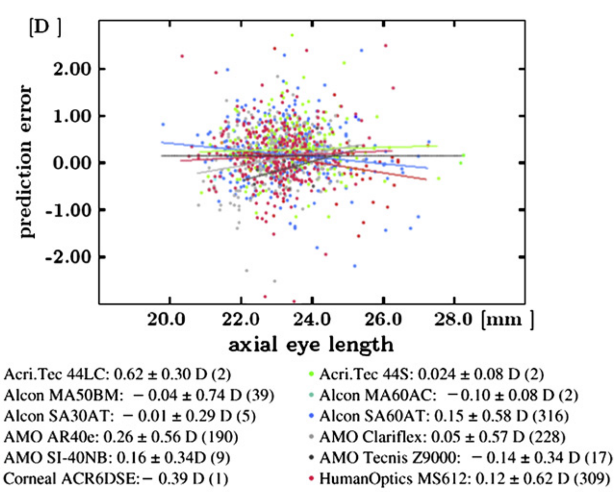

Figure 1. Prediction error of the 1121 Castrop-Rauxel Eyes. The black line is the overall regression line (steepness −0.00074 D/mm). The colored lines are the regression lines for subsets with more than 25 eyes. The mean prediction errors and numbers of eyes in the subsets (in parentheses) are shown below the figure.

Dependence on Pupil Width and Corneal Asphericity To show the influence of pupil width and corneal asphericity, these values were varied for some subsets of data given above. All other parameters are the same as in Figures 1 and 2, see Table 1. The different outcomes resulted from different best-focus positions depending on spherical aberration. The results depended on many parameters and their distribution in the subcollective. The dependence of the focus shift on pupil width was higher in shorter eyes and for IOL models in which the posterior radius was relatively steeper than the anterior radius. Only aspherical IOLs in which the asphericity is fitted to the eye in such a way that the resulting spherical aberration is zero did not show a pupil width–dependent focus shift.

Intraocular Lens Power Errors Intraocular lens power errors (ie, differences between the actual power and the power on the label) are often overlooked. In

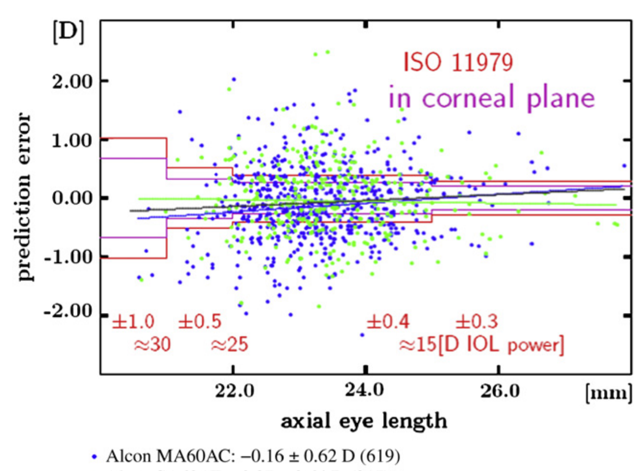

Figure 2. Prediction error of the 936 Aarhus Eyes (ACD Model 1). The black line is the overall regression line (steepness 0.049 D/mm). The green line (steepness −0.015 D/mm), and blue line (steepness 0.075 D/mm) are the regression lines for the 2 IOL types. The mean prediction errors and numbers of eyes (in parentheses) are shown below the figure. The red line is the allowed IOL power tolerance according to ISO 11979, depending on IOL power P, as follows: $\pm$0.3 D for P$\leq$15 D; $\pm$0.4 D for 15.0 D$<$P$\leq$25 D; $\pm$0.5 D for 25.0 D$<$P$\leq$30.0 D; $\pm$1.0 D for P$>$30 D. The magenta line shows the corresponding refraction errors in the corneal plane.

Table 1. Mean prediction errors of the refraction for different pupil widths (2.0 mm and 3.0 mm as wells as paraxial for comparison) and for different corneal asphericities (numerical eccentricities $e = 0$ and $e = 0.5$).

		Mean Predicton Error (D)				
IOL Model	n	Paraxial	$2.0/e = 0$	$3.0/e = 0$	$2.0/e = 0.5$	$3.0/e = 0.5$
Corneal ACR6DSE	1	−0.13	−0.36	−0.65	−0.29	−0.51
AMO Tecnis	17	−0.17 ± 0.34	−0.21 ± 0.34	−0.27 ± 0.34	−0.15 ± 0.34	−0.13 ± 0.34
Alcon MA60AC	619	+0.03 ± 0.61	−0.16 ± 0.62	−0.39 ± 0.63	−0.09 ± 0.62	−0.25 ± 0.62

IOL = intraocular lens; n = number of eyes in the subcollective

particular, it is not obvious to ophthalmologists to what extent the allowed tolerances contribute to the overall error. These tolerances were defined by the International Organization of Standardization following a measurement of IOL powers in different institutions (companies and U.S. Food and Drug Administration; see Discussion). Figure 2 shows the allowed amount of IOL manufacturing tolerances. Even if some manufacturers claim to produce IOLs with significantly lower power tolerances, surgeons are not given information about the actual data.

Reproducibility of Axial Length Measurements

The reproducibility of axial length measurements could be evaluated in the Aarhus patients because the measurements were repeated postoperatively. The IOLMaster readouts had to be corrected for the IOL instead of the crystalline lens inside the eye. Unfortunately, group refractive indices at the measuring wavelength of the IOLMaster were not known for the IOL; therefore, the known values for 540 nm were used. This may cause a small, but negligible, bias. Figure 3 shows the results.

Variation of the Anterior Chamber Depth Prediction Model

In the calculations presented thus far, ACD was predicted according to Model 1. Figure 4 shows the outcome for Model 2. For ACD prediction according to Model 3, some preparations are necessary. First, the "true position" of the IOL relative to the anterior and posterior capsular bag of the crystalline lens must be determined. The postoperative IOL position relative to the posterior corneal vertex can be recalculated from manifest refraction by ray tracing. Figure 5 shows this position, here parameterized by the postoperative ACD, together with the preoperative anterior and posterior surface position of the crystalline lens. Instead of the postoperative ACD, the IOL center position can be calculated as all individual IOL thicknesses are known. Based on the results in Figure 5, IOL position (ACD as well as center) can now also be calculated as a function of the thickness of the crystalline lens relative to the preoperative ACD; that is, altogether as a function of the geometry of the crystalline lens alone (Figure 6). The mean prediction error of the refraction with Model 3 is 0.00 ± 0.65 D. A value of zero for the mean confirms the calculation is consistent; that is, without internal errors, the result is the standard deviation of 0.65 D.

DISCUSSION

The most important results deduced from the evaluation of the patient data are the following:

1. The mean prediction error as well as the means of the differences in prediction error between the

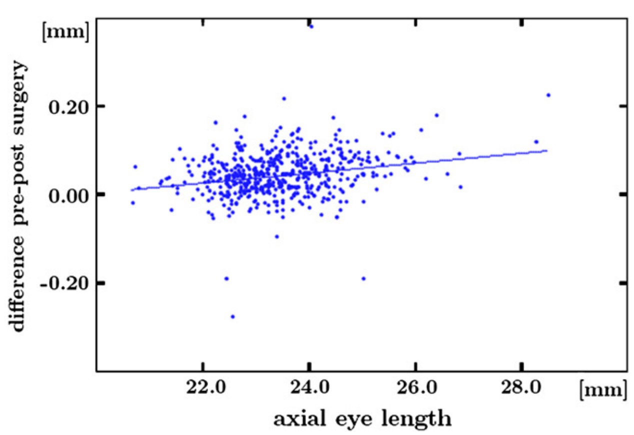

Figure 3. Axial length measurement presurgery and postsurgery. The difference is $d = A_a - A_p$ (mm), where A_a is the preoperative axial length and A_p is the corrected postoperative axial length are shown as function of A_a (mm). The mean difference was 0.041 ± 0.051 mm, decreasing for short and increasing for long eyes with a steepness of 0.011 mm/mm. The performed correction for the IOL is as follow: $A_p = A_m + t \times (n_L - \overline{n})/\overline{n}$, where A_m is the measured value in the pseudophakic eye, t is the thickness of the implanted IOL, n_L is the refractive index of the IOL, and $\overline{n}$ is the weighted mean refractive index of the ocular tissues (1.3527). This weighted mean resulted when the total optical path length (ie, the sum of the products of refractive indices and the geometrical lengths) was calculated for an average-sized eye and then divided by the total geometrical length. However, the resulting axial length corrections very weakly depended on variations of this parameter because the IOL thickness was small compared to the axial eye length.

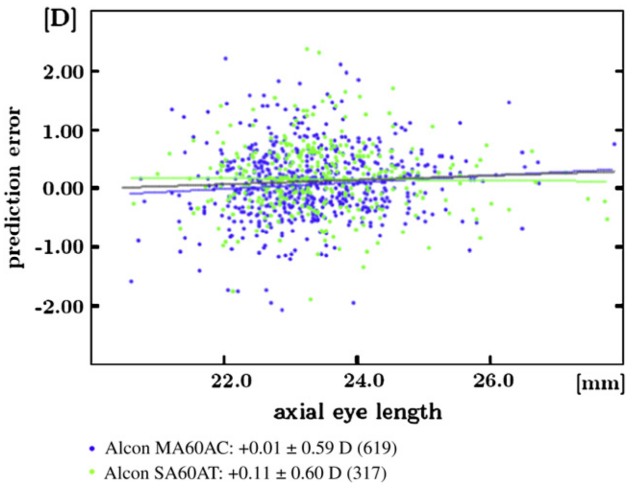

Figure 4. Prediction error of the 936 Aarhus Eyes (ACD Model 2). Except for the ACD model, all other parameters are the same as in Figure 2. The overall mean prediction error was $+0.05 \pm 0.59$ D and the mean absolute error, 0.45 D. The black line is the overall regression line (steepness 0.038 D/mm). The green line (steepness -0.0085 D/mm) and blue line (steepness 0.056 D/mm) are the regression lines for the 2 IOL types. The mean prediction errors and numbers of eyes in the subsets (in parentheses) are shown below the figure.

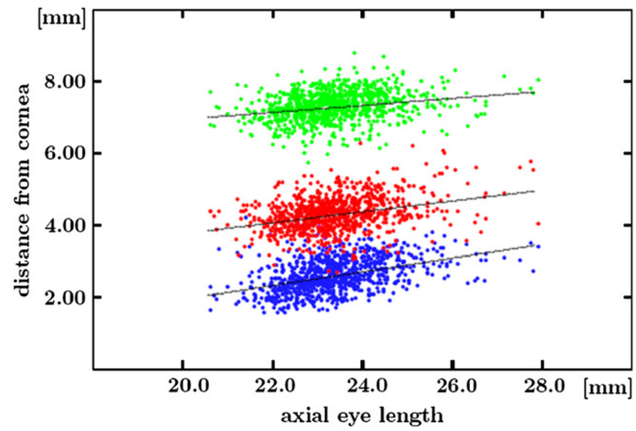

Figure 5. Position of crystalline lens and IOL The distances y (mm) between the posterior corneal vertex and the anterior (*blue*) and posterior (*green*) surfaces of the crystalline lens and of the anterior IOL surface (*red*) are shown as functions of the axial eye length A (mm) for the eyes of the Aarhus collective. The IOL positions are recalculated from the postoperative refraction. The regression lines are as follows: *Blue*, $y = 0.191A - 1.90$; *Red*, $y = 0.151A + 0.741$; *Green*, $y = 0.0961A + 5.02$. Similar results have been obtained using optical pachymetry.[24]

investigated patient collectives are below human recognition threshold, as are the differences between the IOL models (Figures 1 and 2). Variations of pupil widths or corneal asphericities cause differences that are at least as high or higher (Table 1).

2. The standard deviations of the prediction error are about 0.6 D for both patient collectives, without relevant differences for the different ACD models investigated (eg, compare Figures 2 and 4).

These results are satisfying in the sense that so-called individualizations of collectives, hospitals, and surgeons seem dispensable as long as the same surgical procedures and the same measuring equipment are applied. At least the 3 IOLMaster devices used for the 3 collectives seemed to have been calibrated relative to each other with sufficient accuracy. However, the results are not satisfying with respect to the mean individual error of the single eye; for example, that given by the standard deviation of the prediction error. Therefore, if better predictability is required, particularly for spectacle-free car driving, the following steps for improvement are proposed:

1. The assumed pupil width for the driving condition should at least be estimated. The pupil-dependent focus shift is the same phenomenon that is normally known as night myopia.
2. Corneal asphericity should be determined for the same reason. This gives the opportunity to calculate the best focus for the full aperture of the optical system. The easiest and most accurate approach is a ray tracing based on measured corneal topography. This method can be applied to normal eyes as well as to eyes after corneal refractive surgery without any difference.[29]
3. Estimation of postoperative IOL position should be based on the parameters by which it is physically/geometrically mostly determined. For a state-of-the-art cataract operation, this is the geometry of the

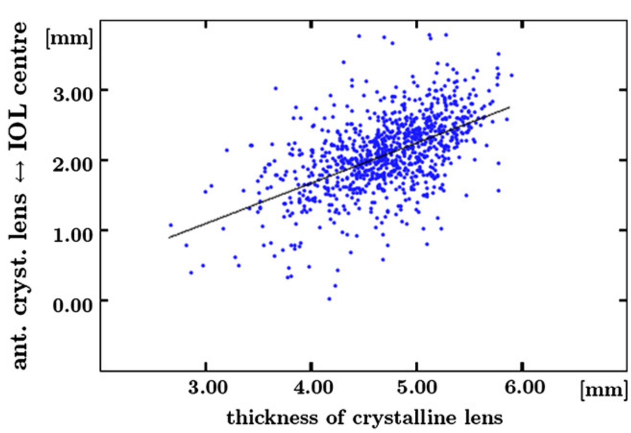

Figure 6. The IOL position relative to the capsular bag. The distance y (mm) between the anterior surface of the crystalline lens and the center of the IOL is shown as function of the thickness d (mm) of the crystalline lens. The IOL center position relative to the posterior corneal vertex is recalculated from refraction. The regression line is $y = 0.574d - 0.632$. Thus, the individual ACD according to Model 3 is given by $ACD = ACD_p + y - t/2$, where ACD_p is the preoperative ACD and t is the thickness of the IOL.

capsular bag. In first approximation, preoperative ACD and thickness of the crystalline lens can be tried. Even if in this paper the outcome with this approach (our Model 3) is slightly worse than with the other 2 methods, this is not a general counterargument. In our investigation, the measurements were performed by ultrasound, which has a much higher intrinsic measurement error than optical methods, which unfortunately were not available. Furthermore, the recalculated IOL positions used for Model 3 in this paper have additional biases resulting from the uncertainties of other parameters (IOL power errors, unknown pupil size, unknown corneal asphericity). Statistical approaches such as our Model 1 and Model 2 have the disadvantage that in eyes far from the statistical mean, results may be unpredictable and may thus cause outliers because these models depend on the properties of collectives rather than on individuals and use statistical correlations instead of functional physical connections.

4. Intraocular lens power errors[30] may contribute significantly to the total error, particularly for the high-power IOL (Figure 2). Tolerances of ± 0.33 D in the corneal plane for an IOL above 25.00 D or even ± 0.66 D for an IOL above 30.00 D are not acceptable and should be reduced. However, the importance of the IOL power error cannot finally be evaluated here because some manufacturers claim to produce IOLs with much smaller power variations than is allowed in ISO 11979.

The error in performing subjective or objective refraction also contributes to the total prediction error shown in Figures 1, 2, and 4. However, this error depends partly on variables (eg, tear film stability, unknown pupil width) that also influence the parameters used for IOL calculation. Therefore, we do not discuss it as an independent error source even if refraction errors on the order of 0.4 D can be assumed. Compared to the issues that require an improvement in accuracy and that principally can be improved, axial length measurement accuracy seems to have reached its preliminary "end point": The standard deviation of the mean reproducibility error (Figure 3) of 0.051 mm, expressed in terms of refraction in the corneal plane, is below ≈ 0.15 D. The systematic bias seen in Figure 3 is not relevant because it vanishes in the calibration process of the instrument by a patient collective.[31] Furthermore, the physical accuracy limit of the PCI as well as that of an ultrasound device is given by the unknown parameters of the crystalline lens. The refractive index for PCI and sound velocity for the ultrasound device cannot be measured in the individual crystalline lens altered by cataract, and this can also not be expected from future developments.

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