Ultrasonographic assessment of the implant lens required to produce emmetropia after implantation

George A. Leary

Present-day implant lenses are essentially thick lens systems. The usual methods by which the required power is assessed involve treating them as thin lens systems, thus giving rise to error. It is possible to avoid such an error by changing the approach to the problem. No attempt need be made to calculate the equivalent power of the implant lens required to produce emmetropia after implantation. The alternative, presented in this report, is to ascribe arbitrarily chosen parameters to the implant lens, thus reducing the calculation to the assessment of the back surface power required to focus light onto the retina.

Key words: implant lenses, lens power, lens constants, surface computation, ultrasonography.

The term "lens implant" is defined here as an acrylic lens surgically replacing the crystalline lens of the eye and fitting in the anterior chamber. One of the problems involved in any lens implant technique is the assessment of the power required to achieve emmetropia after implantation. The method presented in this report makes use of a technique whereby the implant lens front surface power together with the thickness and position are arbitrarily chosen according to the requirements of the implant lens technique. It is to be noted that a knowledge of the refractive error either before or after surgery is not required.

Method

The essence of the method depends upon determining the power of the cornea with one of the usual clinical keratometric techniques plus an axial length assessment by ultrasonography. Although it is not possible to give a review of the ultrasonographic method in this article, it is pertinent to summarize the basic technique. A short list of investigators using particular methods in this rapidly expanding field is given in the references at the end of this article.1-10

The type of ultrasonography under consideration is termed "time-amplitude ultrasonography" (TAU); a transducer incorporating a ceramic "crystal" transmits high-frequency sound waves by means of an induced piezoelectric effect. Ultrasound produced during the transmission phase of the transducer is passed into the eye via a coupling medium to be reflected from the internal surfaces back to the transducer. Here, during the receiving phase, the transducer converts the wave energy into electrical energy and passes it to an oscilloscope where a "spot" moving rapidly across the screen forms a horizontal line.
Fig. 1.

termed the time base. At the receipt of each echo, the time base is deflected vertically to form a peak—each represents the acoustic position of the structure from which the echo originated within the eye. A trace therefore consists of the peak for the cornea (with high resolution both surfaces are resolved), a peak for each of the lens surfaces, and a retinal peak with a complexity of peaks from the rear walls of the eye.

With calibration of the instrument, a knowledge of the velocity of ultrasound in the ocular media, the depth of the anterior chamber, lens thickness, and vitreous chamber depth may be assessed from the interpeak distances and the axial length deduced.

Ultrasonography is often hailed as a simple method; the simplicity of application together with the ease of production of visible peaks cannot be denied. Here, however, the simplicity ceases, since there are many decisions to be made as to quality, reliability, and possible artifacts of the trace obtained. The methods used by most investigators show good agreement; each investigator has, however, tailored the method to suit his own investigational needs.

From a knowledge of the corneal power and length of the eye, i.e., the axial length, it is possible to compute the back-surface power of the implant lens of chosen front-surface power and thickness.

Data required. Where two widely differing power meridians exist on the cornea, choice of the meridian to enter the calculation must be made on a power magnitude and axis orientation. Here, obviously, choice will be guided by the necessity to correct the residual astigmatic error by normal prescription methods. Until such time as it is convenient to "work" cylindrical curvatures on the implant lens, only one meridian can be considered.

The method of ultrasonography is employed to determine the axial length. It is pertinent to note, however, that in the presence of cataract the velocity of the ultrasound in the crystalline lens will differ from that present in the "normal lens." Since the velocity of the ultrasound has been established merely within "normal" ocular tissues, the determined thickness of the lens (and hence the total axial length) will suffer error. The degree of error is, possibly, small relative to the distance measured; to be prophetic the direction of the error will be toward a thinner lens than is actually present in the eye.

In the light of the magnitudes of error resulting from the difficulty in exact positioning of the implant lens, the error due to the presence of cataract is not paramount. Application of the ultrasonography after surgery on the aphakic eye would remove this problem but at the same time demands two surgical procedures, extraction and then implantation would be required.

It is also necessary that the refractive index of the material from which the implant lens is made be known.

Data arbitrarily chosen. The method for the calculation has been devised such that the surgeon may arbitrarily choose: (1) the thickness of the implant lens, (2) the power of the front surface, and (3) the position that the front surface is to occupy behind the cornea.

Calculation of required back-surface power of the implant lens. In the following equations all distances are, for convenience, expressed in meters and designated by lower case. Dioptric values are designated by upper case. The equations follow the normal conjugate focal relationships and employ reduced distances.

It should be noted that at no time throughout the calculation is the implant lens considered either as thin or as an equivalent lens. The concern is not with "what power of lens is required to make the eye emmetropia"; the unknown to be calculated is the back-surface power for an existing implant lens to produce emmetropia in an eye of known axial length.

First stage of calculation. The convergence of
the light passing through the cornea is obtained from Equation 1, where \( F_i \) is the corneal power, \( n_2 \) is the refractive index of the aqueous, and \( f_i \) is the focal distance after refraction by the cornea (Fig. 1).

\[
f_i = \frac{n_2}{F_i}
\]  

**(Second stage of calculation.** The vergence of the light (AB., Fig. 1) incident at the front surface of the implant lens is computed from Equation 2, where \( d_1 \) is the arbitrarily chosen anterior chamber depth, \( L_1 \) is the vergence of the light at the anterior surface of the implant lens, and \( n_2 \) is the refraction index of the aqueous.

\[
L_1 = \frac{n_2}{f_i - d_1}
\]  

**(Third stage of calculation.** It is now necessary to compute the vergence of the light incident at the unknown back surface of the implant lens, i.e., the effective vergence \( L_2 \). This is done in Equation 4, where \( n_3 \) is the refractive index of the implant lens and \( d_2 \) is the chosen thickness of the implant lens.

\[
L_2 = \frac{n_3}{L_1 - d_2}
\]  

**(Fourth stage of calculation.** At this point of the calculation, the amount of convergence of the light incident on the back surface of the implant lens has been calculated. The progress of the computation resolves itself into determining the power required on this back surface to focus the light after refraction onto the retina. Through the distance \( d_3 \) (Fig. 1), i.e., the vitreous chamber, Equation 5 can be computed, where \( n_4 \) is the refractive indices of the implant lens and \( d_3 \) is the axial length and \( d_2 \) is the depth of anterior chamber and implant lens thickness, respectively.

\[
d_3 = d_3 - (d_1 + d_2)
\]  

It follows at this point that to produce emmetropia the vergence of the light \( L_3 \) (previously determined), after modification by the unknown back surface of the implant lens, must focus at a distance \( d_3 \) onto the retina. Hence, the vergence represented by the distance \( d_3 \) must be the convergence of the light required after refraction by the back surface. Then we have Equation 6, where \( L_3 \) is the vergence to the retina and \( n_4 \) is the refractive index of the vitreous.

\[
L_3 = \frac{n_4}{d_3}
\]  

**(Fifth stage of calculation.** The power \( F_3 \) required on the back surface of the implant lens is now deduced from the vergence of the light \( L_1 \) from the retina to the lens and the effective vergence \( L_2 \) previously defined (Equation 7).

\[
F_3 = L_1 - L_2
\]  

The radius \( (r_3) \) of the back surface of the implant lens is shown by Equation 8.

\[
r_3 = \frac{n_4 - n_3}{F_3}
\]  

**A compound equation.** In demonstrating the method for calculation, the equations have been shown as individual steps. It follows that from this so called "step-along" method of treating the lens surfaces, the equations can be compounded. By doing so, the calculation is simplified for use on any electronic calculator with a storage position.

The compounded equation is shown in Equation 9.

\[
L_3 = \left( \frac{n_3}{L_1 - d_2} \right) - d_3
\]  

This provided the reduced vergence of the light onto the back surface of the implant lens after refraction by the cornea and front surface of the implant lens.

The vergence from the retina to the back surface of the implant lens is obtained from Equation 10.

\[
L_1 = \frac{n_3}{L_1 - d_2} + F_3
\]  

Hence, the power of the back surface of the implant lens is shown by Equation 11.

\[
F_3 = L_1 - L_3
\]  

**Discussion**

Few methods are a panacea for all conditions. In the method given here the
accuracy can be assessed only in terms of the surgeon's ability for correct positioning of the implant lens and adeptness in the ultrasonographic technique. The nature of the article was to suggest a method for those skilled in implant surgery.

The problem of the high anisometrope has not been discussed. Present-day ophthalmologic biometric techniques allow the image size of the nonaffected eye to be assessed and the implant lens for the affected eye to be computed to give equal or near equal image size; obviously the resultant ametropia requires normal spectacle correction along with the original ametropic eye. Such a technique involves ophthalmophakometry combined with ultrasonography as used by the author in this laboratory for the study of ametropia. The equations and required computerization may be obtained from the author. It must be stated that the process is a lengthy one and in the author's opinion not suited for clinical practice.

Conclusion

It has been demonstrated here that a simple clinical keratometry combined with ultrasonography will enable the back-surface power of an implant lens to be assessed by calculation. The front-surface power, lens thickness, and position of the lens behind the cornea were parameters arbitrarily fixed by the surgeon. At no time was it necessary to obtain the refractive error.

Although in this report the above parameters were chosen, it need not be so. Once the corneal power and axial length have been determined, any form of implant lens, singular or in combination, may be computed from simple optical principles.

The optical progress of the calculation was illustrated, and then the equations were compounded to facilitate use on one of the electronic calculators equipped with a storage cell.

Use of the method given in this report would allow "blank" implant lenses of say 10.0, as the arbitrarily chosen front-surface power to be stocked. The final back-surface power could be completed for each patient.

REFERENCES


