Aus dem Department für Augenheilkunde

Universitäts-Augenklinik Tübingen

Sektion Experimentelle Ophthalmochirurgie

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A Comparative Analysis of Methods for Calculating IOL Power: Combination of Three Corneal Power and Two Axial Length Measuring Techniques

Inaugural-Dissertation
zur Erlangung des Doktorgrades
der Medizin

der Medizinischen Fakultät
der Eberhard-Karls-Universität
zu Tübingen

vorgelegt von

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aus

Bishkek, Kirgisische Republik

2008

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Dedicated to my mother Kenzhe Abdyldaeva

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1. Introduction

History of IOL power calculation

The history of cataract surgery goes back to 5th century B.C. From Sanskrit manuscripts the earliest type of cataract surgery was known as couching. This techniques permitted dislocation of the mature cataract into the vitreous cavity and enabled the patient to see better. The first idea of substituting an optical device for the opaque crystalline lens belonged to Tadini in 1766 (Fechner, Fechner et al., 1979). The evolution of cataract surgery took a giant step in 1949 when Harold Ridley, developed and implanted the first intraocular lens (Ridley, 1952) and provided evidence for tolerance of a foreign body in the eye and the prospect of restoring functional vision.

At present time, cataract surgery is one of the most frequently performed and successful operations in the world. The techniques and results of cataract surgery have changed dramatically during the past three decades. The technique has moved from intracapsular cataract extraction (ICCE) to extracapsular cataract extraction (ECCE). Phacoemulsification, small incisions alone with advances in intraocular lens materials and designs, viscoelastic agents, topical anesthesia have increased safety and efficiency of cataract surgery and become the standards. These advances in technique and equipment have led to a dramatic increase in the popularity of phacoemulsification.

As cataract surgery technology and intraocular lens (IOL) technology have improved remarkable and become safe, the patients have been expecting better postoperative refractive results, which are determined by the precise intraocular lens power calculation (Hillman, 1982).

The calculation is normally based on corneal power, axial length (AL) measurements and IOL calculation formulae. These three factors are considered to be the most critical factor for accurate IOL power calculation.

Axial length is usually measured by applanation A-scan ultrasound, which is widely used technique (Binkhorst, 1981; Olsen and Nielsen, 1989; Leaming, 2001). In A-scan biometry, the sound travels at a frequency of

approximately 10 million Hz (10 MHz). This extremely high frequency allows for restricted penetration of the sound into tissues. The biometer measures axial lengths, the distance between the anterior corneal vertex and internal limiting membrane of the retina, along the optical axis with a resolution of 200 µm and precision of 150 µm (Olsen, 1989). The method requires the use of topical local anesthesia and contact of the cornea with a probe of A-scan, as ultrasound energy is emitted from the probe tip by pulsing electricity.

Studies based on ultrasound biometry demonstrated 54% of all IOL power miscalculations result from wrong AL measurements (Olsen, 1992). The measurement error in axial length of 100 µm results in postoperative refractive error of 0.25D (Binkhorst, 1981) to 0.28D (Boerrigter, Thijssen et al., 1985; Olsen, 1987(a); Drexler, Findl et al., 1998).

The IOLMaster is a noncontact partial coherence interferometry (PCI) method for AL measurement, which has recently become commercially available (Fercher, Hitzenberger et al., 1993; Drexler, Findl et al., 1998; Haigis, Lege et al., 2000). It uses infrared diode laser (λ 780 nm) of high special coherence and short coherence length (160 µm). The optical scan uses an external Michelson interferometer to split the infrared beam into coaxial dual beams allowing the technique to be intensive to longitudinal eye movement. Both components of the beam illuminate the eye and are reflected at each interface where the change in refractive index occurs. If the optical path length is within the coherence length interference signal is detected by a photodetector (Hitzenberger, 1991). The IOLMaster measures the ocular axial length between the corneal vertex and retinal pigment epithelium along the visual axis using red fixation beam, with a resolution of 12 µm and precision of 5 µm (Hitzenberger, Drexler et al., 1993; Drexler, Findl et al., 1998; Drexler, Hitzenberger et al., 1998; Findl, Drexler et al., 1998; Haigis, Lege et al., 2000; Findl, Drexler et al., 2001; Lam, Chan et al., 2001; Vogel, Dick et al., 2001; Kiss, Findl et al., 2002; Santodomingo-Rubido, Mallen et al., 2002; Nemeth, Fekete et al., 2003), (Haigis, 1999). Advantages of this technique is that there is no need for local anesthesia and pupil dilation (Drexler, Findl et al., 1998; Findl, Drexler et al., 2001), therefore method reduces the potential risk of corneal erosions or

infection (Hitzenberger, Drexler et al., 1993; Rose and Moshegov, 2003). The technique is observer-independent method for AL measurement (Drexler, Findl et al., 1998; Lam, Chan et al., 2001; Vogel, Dick et al., 2001; Santodomingo-Rubido, Mallen et al., 2002; Findl, Kriechbaum et al., 2003; Tehrani, Krummenauer et al., 2003(b)).

The measurement obtained by IOLMaster has been reported more accurate and reproducible than that by US in a normal eye (Eleftheriadis, 2003; Goyal, North et al., 2003) and in a pseudophakic eye (Haigis, 2001; Goyal, North et al., 2003). Since introducing the ultra-high precision PCI, this method has proven its accuracy in IOL power calculation using different lens formulas too (Drexler, Findl et al., 1998; Findl, Drexler et al., 1998; Vogel, Dick et al., 2001; Connors, Boseman et al., 2002; Nemeth, Fekete et al., 2003; Ueda, Taketani et al., 2007).

The incredible technique of phacoemulsification and IOL material and design provided rapid improvements in ophthalmology in recent decades and has made modern cataract surgery safe and effective. Axial eye length with an error of approximately 0.2 D is no longer the dominating error if the measurements are performed by interferometry; the same is true for corneal radii in normal eyes (Preussner, 2007). But if the total error threshold is below the error of refraction, the accuracy of the IOL power calculation formula must be improved. This important part of IOL power calculation has been growing in recent years especially in eyes that have had refractive surgery.

In the early 1970s, first commercially available ultrasound instrumentation was adopted to clinical practice. This period gave birth to the first theoretical and empirical intraocular lens power calculation formulae. The first formula for the determination of intraocular lens power was published by Fyodorov, Kolinko and Kolinko (Fyodorov SN, 1967). All original formulae by Fyodorov (Fyodorov SN, 1967), Binkhorst (Binkhorst, 1972), Colenbrander (Colenbrander, 1973), Fyodorov (Fyodorov, Galin et al., 1975), Thijssen (Thijssen, 1975), van der Heijde (van der Heijde, 1975) and Hoffer (Hoffer, 1982) are first generation theoretical formulae. They required axial length of the eye, the corneal power in diopters, corneal radius and position of the intraocular lens

along the optical axis of the pseudophakic eye or anterior chamber depth (ACD). The main feature of first-generation theoretical formulae was that position of IOL in the eye is fixed for each lens type. This assumption was not unreasonable: at that time, when cataract surgery was represented by intracapsular cataract extraction and anterior chamber intraocular lenses implantation; the anterior chamber IOL was assumed to have a defined position in relation to the anterior plane of the cornea. Although these formulas are not used in present time, they are all the basis of formulae developed or modified later.

Gills (Gills, 1980), Retzlaff (Retzlaff, 1980(a); Retzlaff, 1980(b)), Sanders and Kraff (Sanders and Kraff, 1980) and Sanders, et al. (Sanders, Retzlaff et al., 1981) developed empirically determined regression formulae. First-generation regression formulas are linear functions based on retrospective analysis of postoperative refraction and biometric data and following intraocular lens implantation of a particular lens by a particular surgeon. The most relevant of these formulae is SRK formula (Sanders, Retzlaff et al., 1981). The required measurements are axial length and corneal power. One of the variables of the SRK formula is the A-constant, a specific constant for each type of IOL, which is determined empirically on the large sample of patients underwent cataract surgery. A-constant is calculated for each lens type based on the refractive outcomes. This ensured that the A-constant lessened influence of variables like surgical technique, biometric instrumentation and measurement technique on IOL power calculation. For this reason, the SRK formula outperformed the first-generation theoretical formulae; it calculated more accurately than many of the first-generation theoretical formulae (Menezo, Chaques et al., 1984). The advantage of regression formula is that it is relatively simple to calculate.

After Kelman (Kelman, 1967) introduced the extracapsular cataract extraction by phacoemulsification, the second-generation of theoretical and regression formulae were developed. Phacoemulsification provided the opportunity to implant intraocular lenses within the capsular bag of crystalline lens. But the position of these posterior chamber intraocular lenses was difficult to predict, due to characteristics associated with individual lens capsule

shrinkage, lens haptic design and placement of the intraocular lens within the crystalline lens capsule. This variability in the position of the implanted intraocular lens was the reason for the development of the second-generation intraocular lens power formulae.

Contributors to the second-generation theoretical formulae include Holladay, Prager, Chandler et al. (Holladay, Prager et al., 1988) and Colliac (Colliac, 1990). Second-generation theoretical IOL power formulae differ from the first-generation formulae in that the position of the intraocular lens in the pseudophakic eye; is not fixed but changes as a function of two variables: axial length and corneal curvature or, corneal power, of the eye.

The second-generation regression formulae by Thompson, Maumenee and Baker (Thompson, Maumenee et al., 1984), Donzis, Kastl and Gordon (Donzis, Kastl et al., 1985), Olsen (Olsen, 1987(b)) and Sanders, Retzlaff and Kraff (Sanders, Retzlaff et al., 1988), were designed to improved accuracy through the application of non-linear regression formulae. Most prominent amongst these is the SRK II regression formula (Sanders, Retzlaff et al., 1988), a modification of the original SRK formula; it is an approximately linear function for eyes of average axial length, but exhibits nonlinearity in short and long eyes too.

Despite the advances in the precision of ocular biometry, differences in calibration, individual lens capsule shrinkage, IOL design as well as surgical variations limited the ability of any formula to predict the post-operative axial position of the intraocular lens. Hence, the modern generation formulae were developed. Most of them are modifications of original theoretical and regression formulae, through a combination of algebraic and statistical methods. Contributors to the third and fourth-generation formulas include Hoffer (Hoffer, 1993), Olsen, Corydon and Gimbel (Olsen, Corydon et al., 1995), Retzlaff, Sanders and Kraff (Retzlaff, Sanders et al., 1990), Holladay (Holladay, Gills et al., 1996) and Haigis (Haigis, 2001). Several studies have been published to compare the accuracy of the IOL power formulae available today (Sanders, Retzlaff et al., 1990; Ascaso, Castillo et al., 1991; Hoffer, 1993; Elder, 2002). Nevertheless, it is clear that the greatest challenge for the calculation of

intraocular lens power lies, not in the intraocular lens power formulae themselves, but in the accurate prediction of pseudophakic lens position.

The issue of the axial position of an intraocular lens in the pseudophakic eye is still poorly understood and misrepresented topic in intraocular lens power calculation. Different authors use in their formulae different variables like 'A-constant', 'surgeon-factor', 'anterior chamber depth' and 'effective lens position' to describe lens position in the pseudophakic eye. In this regard, the strength of the empirical approach (SRK and SRK-II regression formulae) is that it does not measure the position of the intraocular lens in the pseudophakic eye, but this value is implicit in the calculation of the A-constant for each lens type. Olsen found that for any given formula as many as 20 – 40% of all undesirable refractive outcomes following intraocular lens implantation may be related to inaccurate prediction of the pseudophakic lens position (Olsen, 1992).

In conclusion, it is now possible to significantly reduce the chance of a postoperative refractive ametropia after cataract surgery with IOL implantation. The ultra-high precision of PCI seems promising in terms of improved accuracy in IOL power calculation. Certainly the state-of-the-art corneal topographic technology is developing very fast and introducing ray tracing technology and light interference technique may bring more improvements in providing extended diagnostic data to measure the eye geometry in patients undergoing cataract or refractive surgery.

After the Food and Drug Administration first approved the excimer laser in October 1995 for correcting mild to moderate nearsightedness, refractive surgery has become no longer something just for risk takers. Refractive surgery is now a mainstream. Thus, calculation of IOL power in patients with a history of refractive surgery is becoming a crucial part in maintaining a level of visual satisfaction patients after cataract surgery. As the number of patients having refractive surgery increases, ophthalmologists must continue to improve methods of calculating IOL power for post refractive surgery eyes.

Purpose

The purpose of this study is to investigate retrospectively the effect of optimizing the A-constants for the SRK II IOL power calculation formula with respect to the refractive outcome of the patients; and to assess and compare the results of IOL power calculation with an optimized A-constant, using the combination of three corneal power and two axial length measuring devices.

2. Patients and methods

2.1. Patients

39 eyes of 35 patients (19 males and 16 females) consecutively undergoing cataract surgery with IOL implantation were included in this study. The age of the patients at the time of the cataract surgery ranged between 50 and 82 years (mean 69±8.5 yr.).

2.1.1. Inclusion criteria

The inclusion criteria were all consecutive cases of phacoemulsification. The exclusion criteria were previous ocular trauma or intraocular surgery; corneal disease or ocular infection; history of ocular disease such as glaucoma, optic atrophy, macula degeneration, retinopathy, or ocular tumor.

2.1.2. Surgery

The surgery consisted of routine phacoemulsification cataract extraction and followed IOL implantation. No surgical complications were reported. There were no breaches of lens capsule and all IOLs were placed into the capsule bag.

2.2. Methods

Before phacoemulsification surgery all patient underwent a complete ophthalmic examination, including best spectacle-corrected visual acuity (BSCVA), manifest refraction, corneal power (K-value) measurements (manual keratometry, IOLMaster keratometry, C-scan corneal topography), IOLMaster axial length (AL) and anterior chamber (ACD) measurements.

One month after phacoemulsification the following examinations were performed: uncorrected visual acuity (UCVA) and BSCVA, manifest auto refractive and manifest subjective refraction, manual keratometry, IOLMaster

keratometry, C-scan corneal topography and IOLMaster AL and ACD measurements.

Visual acuity was measured using a Snellen character projector that focused the image 5 meters in front of the patients.

Refraction was performed with automated refractometer (Refracto/Lensmeter RL-10, Canon).

Corneal power (K-value) was measured using three devices: Javal-type manual keratometer (Keratometer 10 SL/O, Karl Zeiss; and BLOCK Ophthalmometer "Rubin"), IOLMaster (Karl Zeiss) and corneal topography system (C-scan, Technomed Technology).

Axial eye length and anterior chamber depth measurements were performed with optical biometry (IOLMaster, Carl Zeiss) and ultrasound biometry (Biometer AL-2000, Tomey; and Biometer BVI AXIS) with a 10 MHz contact probe using local anesthesia.

A-constant of implanted Alcon AcrySof SA60AT lenses recommended by the manufacture for A-scan is 118.4. The targeted postoperative refraction in most cases was emmetropia as far as possible, with preferred slight shift towards myopia.

2.2.1. Manual keratometry

Manual keratometry is the standard method on which IOL power calculation formulas were originally based. This instrument follows the variable doubling principle and it is applied as an attachment to the slit lamp. The manual keratometer measures the central 3 mm area of the cornea (2.8 mm [r=8.0 mm] to 3.2. mm [r=9.5 mm]) and evaluates four points on two orthogonal meridians separated 3 mm to 4 mm on the paracentral cornea. The measured range of instrument for corneal radii is from 4.0 to 11.2 mm, with a scale interval of 0.01 mm. The mires of the manual keratometer are rotatable 180 degrees at the optical axis. After identifying the corneal radius of one principal meridian, the mires are rotated until their images no longer appear distorted to locate and measure the other meridian at an angle of 90 degrees. The Zeiss Keratometer

does not provide direct reading of the dioptric power of the cornea at the meridian under examination. However, the scale for reading radii of curvature can be transformed to surface power according to the keratometric formula, as follows: $D = \dfrac{n-1}{R} \cdot 1000$; where D is Keratometric diopters, n is the refractive index of 1.3375 and R is radius of cornea curvature [mm]. In compare, The BLOCK Ophthalmometer can transform from sphere radius [mm] to diopters automatically. Unlike the Javal-type keratometer, the BLOCK Ophthalmometer utilizes a refractive index of 1.332 to transform the corneal radius into diopters. Hence, all keratometric values obtained with the BLOCK Ophthalmometer were recalculated using a correction index of 1.0166 to be comparable with values obtained with Javal-type keratometer. This correction factor was simply resolved form keratometric formula above: corneal power values from two instruments are $D1 = \dfrac{n1-1}{R} \cdot 1000$ and $D2 = \dfrac{n2-1}{R} \cdot 1000$ with two different refractive indexes n1 and n2 and common corneal radius R, thus the correction factor can be resolved:

$$\frac{n1 - 1}{D1} \cdot 1000 = \frac{n2 - 1}{D2} \cdot 1000 \quad \text{or} \quad D1 = \frac{n1 - 1}{n2 - 1} \cdot D2$$

Keratometric readings from BLOCK Ophthalmometer (D2) in order to be comparable to Javal-type readings (D1) should be recalculated by correction factor:

$$\frac{n1 - 1}{n2 - 1} = \frac{1.3375 - 1}{1.332 - 1} = 1.0166$$

2.2.2. Corneal topography

A Corneal Topography system is a type of computerized imaging technology that evaluates anterior corneal surface topography based on Placido reflective image analysis. The system measures corneal topography with 15 concentric rings of light, of known separation and width, reflected from air/tear interface. The reflected image is captured on charge-coupled device (CCD) camera. The separation of the 16 ring edges is measured objectively by the

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internal image analysis software of the instrument at one degree intervals over 360 degrees over the entire cornea (0-3, 3-5 and 5-7 mm) and is then calculated in reference to a known calibration file. The corneal topography instrument samples more than 1000 power points within the central 3 mm and about 10800 points over the entire cornea. Computer software analyzes the data and displays the results in a topographic map. Every map has a many color-coded scale that assigns a particular color to certain keratometric dioptric range.

Simulated keratometry (SimK) provides the power and axis of the steepest and flattest meridian in the central 3-mm area similar to values provided by the keratometer. The steep simulated K-reading is the steepest meridian of the cornea, using only the points along the central pupil area with 3-mm diameter. The flat simulated K-reading is the flattest meridian of the cornea and is by definition 90° apart. These readings define the central corneal curvature that is frequently the visually most significant. The 3-mm diameter was chosen primarily by historical reasons for the purpose of comparison with standard keratometry that is used for analysis of 4 central points, 3.2 mm apart. The keratometric diopters are derived from radius of curvature: $D = \frac{n-1}{R} \cdot 1000$; where D is Keratometric diopters, n is refractive index of 1.3375 and R is radius of cornea curvature [mm].

The system uses a short working distance, approximately 40 mm from the center of the cone to the surface of the eye. The patient's chin is placed on the chin rest and the forehead rested against the forehead strap. The patient fixates a yellow light centered in the cone and focusing of the instrument is made by maneuvering the Placido target with a joystick. The instrument employs a diode-laser focusing system. The video image of corneal topography is captured via releasing a button on a joystick and is shown on the display.

The corneal topography system provides an automatic measurement, automatic right/left detection, graphic user interface and data transfer.

2.2.3. Ultrasound biometry

The ultrasound biometer (A-scan) is an ultrasound applanation device designed for measuring the axial length. The measuring technique is based on the capability of sound to travel through a solid or liquid in a wave pattern. Ultrasound energy is emitted from the probe tip by pulsing electricity, which make vibrating of a crystal element on the probe tip at given frequency. Then, a pause of a few seconds occurs, so the returning echoes can be received by the probe tip. In this manner, the probe acts as both transmitter and receiver of ultrasound signal energy.

Measurement data can be calculated based on the time it takes the ultrasound waves to reflect back to the probe from the internal limiting membrane and preset converted velocity. The time from echo for corneal epithelium to the echo for internal limiting membrane is calculated by set conversion values for sound speed to determine the AL:

$$L = V*t / 2 \qquad \text{where,} \qquad L: AL$$

V: converted sound speed

t: measured time.

The biometer measures axial lengths ranging from 15-40 mm with an accuracy of ±0.1 mm and a resolution of 0.01 mm and transducer frequency 10 MHz±10%. Eye modes include normal, aphakic, pseudophakic and dense cataract. For maximum precision, the biometer calculates the average value of up to 10 measurements.

For proper measurement the probe tip must be directed along the visual axis.

All measurements were done with a hand-held contact probe in the automatic mode.

The biometer calculates IOL power using five of the most popular formulas: SRK II, SRK/T, Holladay, and Haigis and can also display up to three lens constants and the corresponding IOL powers.

2.2.4. IOLMaster

The IOLMaster is a combined biometry instrument for the measurement of data of the human eye needed to calculate the power of an implanted IOL. The AL measurement is based on partial coherence interferometry (PCI) principles, based on the Michelson interferometer (Vogel, Dick et al., 2001) and takes 0.4 seconds. The basic principle of PCI is depicted schematically in Fig.2.1. (Haigis, Lege et al., 2000):

Figure 2.1 Operating principal of IOLMaster
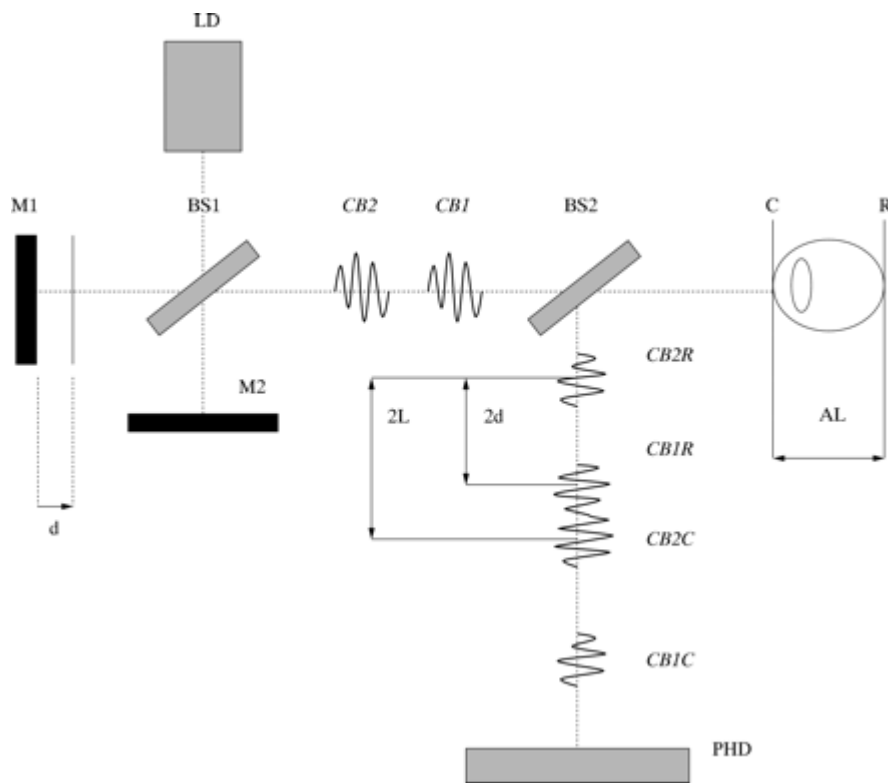
A laser diode LD emits infrared light (λ=780 µm) of a short coherence length (approximately 160 µm) that is split into two parallel and coaxial beams CB1 and CB2 of different optical path length by the beam splitting prism BS1 and reflected into the eye by two mirrors M1 and M2. Both beams are reflected by the cornea C and retina R. The light reflected by the cornea interferes with that reflected by the retina if the optical path of both beams is equal. On leaving the eye, the difference in frequency between the coaxial beams is detected by a

photodetector PHD, after passing through a second beam splitter BS2. During the measurement process, an interferometer mirror M1 is moved across the measuring range at a constant speed, scanning the eye longitudinally. The signals are amplified, filtered and recorded as a function of the position of the interferometer mirror M1 with high accuracy. From this parameter, the system determines the AL, the path difference between the corneal epithelium and the retinal pigment epithelium, in contrast with A-scan waves, which are reflected from the internal limiting membrane (ILM). Hence, in order to make the IOLMaster measurements comparable to ultrasound measurements, the IOLMaster is calibrated against the immersion ultrasound (Kiss, Findl et al., 2002; Packer, Fine et al., 2002).

Alignment of the instrument to the eye is performed via a charge-couple device (CCD) camera. If the results of measurements differ by more than $\pm$100 µm from the mean value, no mean value is displayed.

For AL measurement, the patient fixates on a fixation light and the observer focuses the light beam by looking at the reflex at the cornea. The measurement should only be taken if the patients fixate properly and the light beam is focused or the measurement results will not be reliable.

In the corneal power measurement system, six infrared diodes illuminate the cornea and six infrared points of light, arranged in a 2.3 mm diameter hexagonal pattern, are reflected from the air/tear film interface. The reflections are captured by a CCD camera. Their distances are a measure of the corneal radius. The measurements are released via a button on a joystick. The image of the reflections is shown on the display. Light-emitting diodes align the instrument to the eye. As soon as diodes appear sharply focused and centered on the display, the measurement can begin. The instrument must be focused on the six peripheral dots, not on the central dot. If the instrument is optimally focused, fine luminous circles are visible around the peripheral dots. The measurement results are distance independent within a range of $\pm$2 mm. The instrument displays the corneal radius of the two principle meridians, the corneal refraction, the axes, and the astigmatic difference. The results are the

mean values of five individual measurements. If the results of the individual measurements differ by more than 50 μm, no mean value is displayed.

Calculation of the corneal reflection is based on the measured corneal radius and the factory-set refractive index of 1.332. Although, all measurements of corneal radius made by IOLMaster in the clinic were transformed into corneal diopters using refractive index of 1.3375, set up into the IOLMaster software.

The ACD measurement is based on the optical cross-sectional image of the anterior chamber by means of a slit lamp with subsequent image analysis. The right eye is illuminated from the right and the left eye is illuminated from the left by a 0.7 mm width slit beam of light at an angle of approximately 38 degrees relative to the visual axis. The instrument camera is aligned so that the light beam forms an optical section and the internal software calculates the ACD automatically using the corneal radii that have been already performed. The instrument measures the ACD as it is usually measured in biometry. Anatomically this is the ACD plus the cornea thickness.

Therefore, the corneal power and ACD measurements are not based on the PCI principle but rather on the image-analysis principle, in which distances between light reflections on the cornea, iris and lens are measured.

The instrument measures AL, ACD and corneal radius in 1 session. The total patient examination time, including AL, ACD and corneal radius measurements and IOL calculation, is approximately 5 minutes.

The IOLMaster provides an automatic measurement, automatic right/left detection, graphic user interface including the most common IOL power calculation formulas (SRK II, SRK/T, Holladay, Haigis), and data transfer.

2.2.5. IOL power calculation

The calculation of IOL power was based on GOW70-formula (Gernet, 1970):

$$DL = \frac{n}{L - d} - \frac{n}{n/z - d}$$

with $\quad z = DC + \dfrac{ref}{1 - ref*dBC}$ $\quad$ and $\quad$ $DC = \dfrac{nC - 1}{RC}$

DL	:	refractive power of IOL
DC	:	refractive corneal power
RC	:	corneal radius
ref	:	desired refraction
d	:	optical ACD
L	:	axial length as measured by ultrasound
n	= 1.336	refractive index of aequeous and vitreous
nC	= 1.3375	(fictitious) refractive index of cornea
dBC	= 12 mm	vertex distance between cornea and glasses

First of all, an optical lens position, or optical ACD (d), was obtained using known implanted IOL power and postoperative subjective refraction. This was done with formula below simply resolved from GOW70-formula for different variables:

d = (L + n / z - ((-L - n / z) * (-L - n / z) - 4 * (L * n / z - n / DL * (n / z - L))) ^ (1 / 2)) / 2

Then, with obtained optical ACD the calculation of an emmetropic IOL for each eye was made, using the GOW70-formula itself to predict the theoretically ideal power of lens that would have delivered the desired refraction deficit to zero.

In clinical practice, for the calculation of IOL power, the SRK II formula (Sanders, Retzlaff et al., 1988) was selected as the formula of choice in the clinic:

SRK II: $\quad$ **P = A1 – 0.9*K – 2.5*L** $\quad$ with:

P : power for emmetropic IOL
K : corneal power

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L : axial length

A : A-constant

Due to adaptation of the A-constant to the different axial length SRK II formula is as followed:

A1 = A + 3   for L < 20 mm

A1 = A + 2   for 20 <= L < 21 mm

A1 = A + 1   for 21 <= L < 22 mm

A1 = A   for 22 <= L < 24.5 mm

A1 = A − 0.5   for L => 24.5 mm

Thus, with obtained emmetropic IOL power, the A-constant given by lens manufacturers was optimized for each eye to produce a mean zero prediction error by using a formula below, resolved from SRK II formula:

A = P + 0.9*K + 2.5*L + 3   for L < 20 mm

A = P + 0.9*K + 2.5*L + 2   for 20 <= L < 21 mm

A = P + 0.9*K + 2.5*L + 1   for 21 <= L < 22 mm

A = P + 0.9*K + 2.5*L   for 22 <= L < 24.5 mm

A = P + 0.9*K + 2.5*L − 0.5   for L => 24.5 mm

Then, theoretical IOL power for each eye was calculated by SRK II formula using mean optimized A-constant.

The lens power, estimated in this way was used to predict the refractive outcome by GOW70_ref-formula simply resolved from GOW70-formula for different variables:

$$\text{Ref} = \frac{DC - z}{DC*dBC - z*dBC - 1}$$

$$z = \frac{n}{\dfrac{n}{\dfrac{n}{L - d} - DL} + d}$$

To assess the predicted performance, the refractive outcome was determined as mean numerical error (MNE) and mean absolute error (MAE). MNE and MAE were estimated for six combinations of devices for measuring axial length (A-scan and IOLMaster) and corneal power (manual keratometry, IOLMaster and C-scan). The combined techniques respectively are:

1. A-Scan AL + Keratometer K-value (combination A_1K_1)
2. IOLMaster AL + Keratometer K-value (combination A_2K_1)
3. A-Scan AL + IOLMaster K-value (combination A_1K_2)
4. IOLMaster AL + IOLMaster K-value (combination A_2K_2)
5. A-Scan AL + C-Scan K-value (combination A_1K_3)
6. IOLMaster AL + C-Scan K-value (combination A_2K_3)

2.3. Statistics

The data were processed using a personal computer and statistically analyzed using Microsoft Excel for Windows XP. Measurement values of variables were described with mean, standard deviation, minimum and maximum values. For statistical analysis of the difference and the correlation between pare of six methods the Pearson test and paired *t*-test was applied. A *P* value less than or equal to 0.05 was considered to show a statistically significant difference.

3. Results

3.1. Visual acuity

Visual acuity of 39 eyes was compared preoperatively and postoperatively. Preoperatively mean BSCVA was 0.41 ± 0.24(SD) of Snellen lines (range 0.02 to 1.0) and 0.91 ± 0.26(SD) postoperatively (range 0.3 to 1.4). The preoperative BSCVA was significantly different from postoperative (difference 0.49 ± 0.36(SD) of Snellen lines; $P<.001$) (Figure 3.1). Figure 3.2 shows distribution of BSCVA before and after cataract surgery.

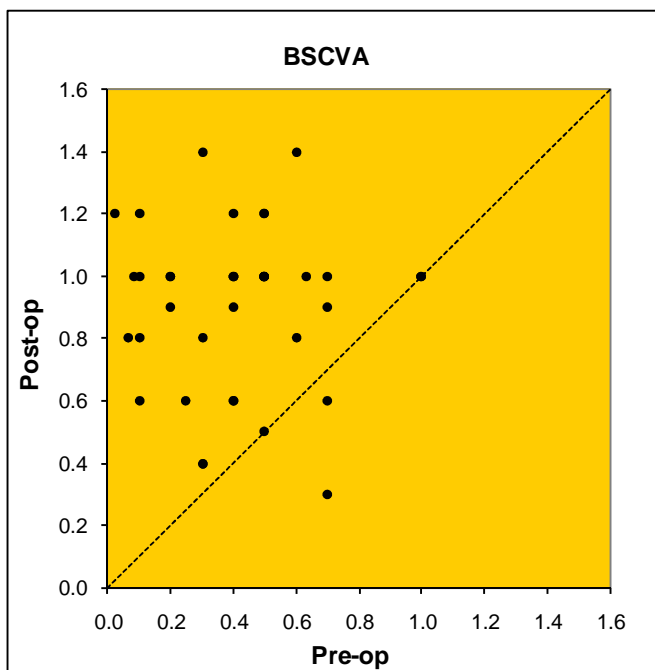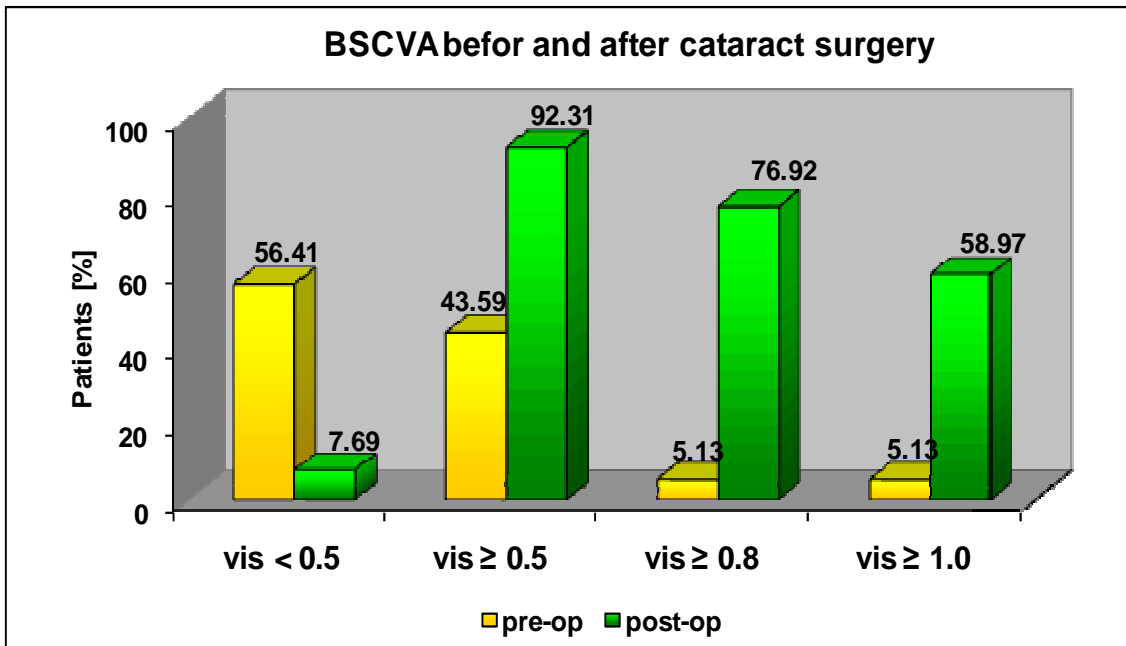**Figure 3.1** BSCVA before and after cataract surgery

Figure 3.2 Distribution of BSCVA before and after cataract surgery

Postoperatively mean UCVA was 0.62±0.28(SD) of Snellen lines (range 0.1 to 1.0). The difference between postop mean BSCVA and UCVA was significant (difference -0.28±0.25(SD) of Snellen lines; *P*<.001). In Figure 3.3 distribution of BSCVA and UCVA after cataract surgery is shown.

Figure 3.3 Distribution of UCVA and BSCVA after cataract surgery
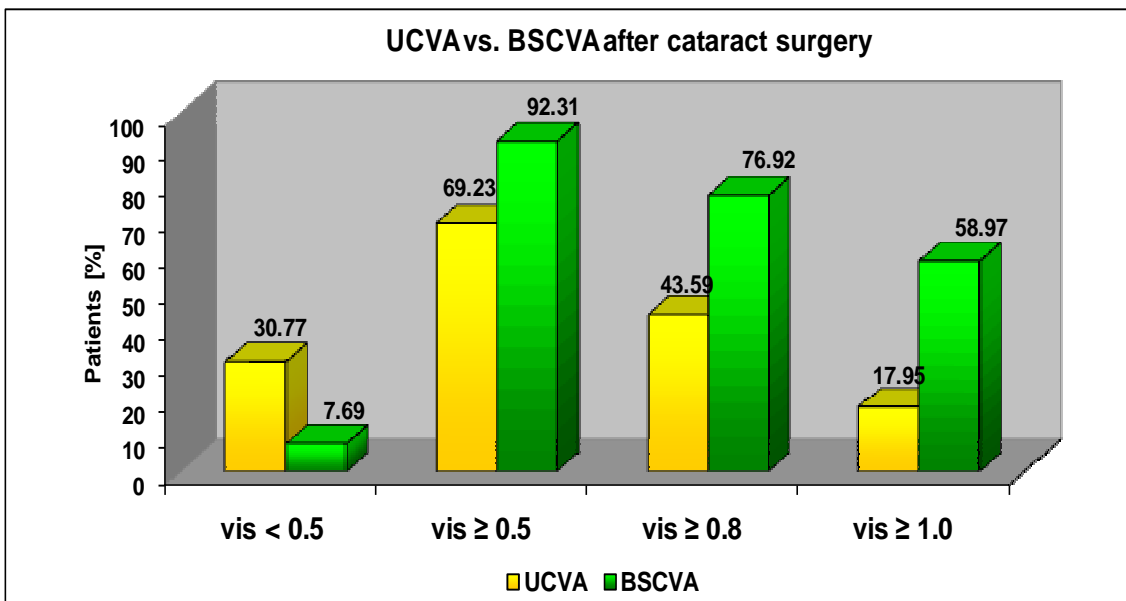
3.2. Refraction

To compare preoperative and postoperative manifest refraction, 37 eyes were available; in 2 eyes refraction could not be measured preoperatively due to dense cataract. Preoperative mean spherical equivalent (SE) was -0.57±3.68(SD) D (range -9.38 to 5.63D) and postoperative mean SE was -0.36±0.99(SD) D (range -2.88 to 2.75D). Figure 3.4 shows the distribution of SE before and after cataract surgery.

Figure 3.4 Distribution of SE before and after cataract surgery (subjective refraction)
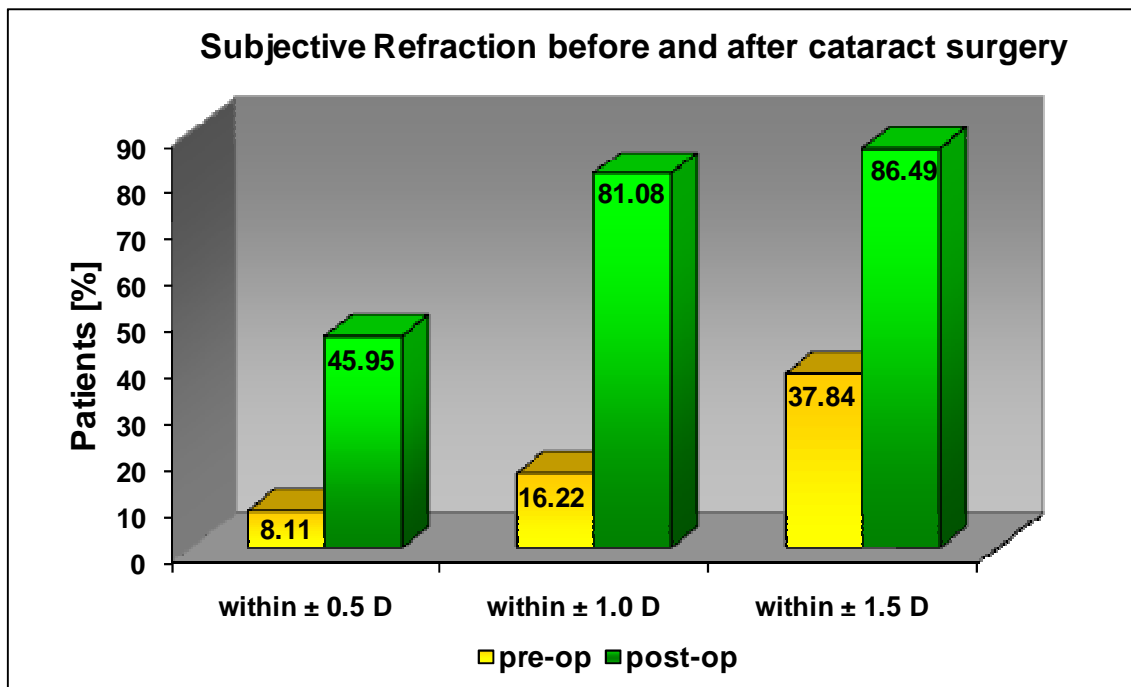
3.3. Corneal power

Table 3.1 summarizes the corneal power values (K-readings) of conventional keratometer, IOLMaster and corneal topography system (C-scan).

Table 3.1 K-readings of keratometer, IOLMaster and C-Scan

	K-reading			
	Preop		Postop	
	Mean±SD (D)	Range (D)	Mean±SD (D)	Range (D)
Keratometer	43.46±1.25	40.60 – 47.47	43.54±1.52	39.67 – 47.02
IOLMaster	43.74±1.41	40.31 – 47.51	43.80±1.52	39.92 – 47.86
C-scan	43.43±1.39	40.25 – 47.26	43.52±1.40	40.27 – 47.24

The K-readings obtained post- and preoperatively with manual keratometer were highly correlated and the difference was found not to be statistically significant, difference -0.08±0.80(SD) D; r=0.85; P=0.26 (Figure 3.5). The negative sign in difference indicates that postoperative Keratometer measures K-readings were smaller than measured preoperatively. The differences between postop and preop K-readings for IOLMaster and for C-scan were found to be statistically not significant, 0.06±0.43(SD) D, r=0.96, P=0.18 for IOLMaster; and 0.09±0.55(SD) D, r=0.92, P=0.16 for C-scan (Figure 3.6 and 3.7).

Figure 3.5 Correlation between Keratometer pre- and postop mean K-values
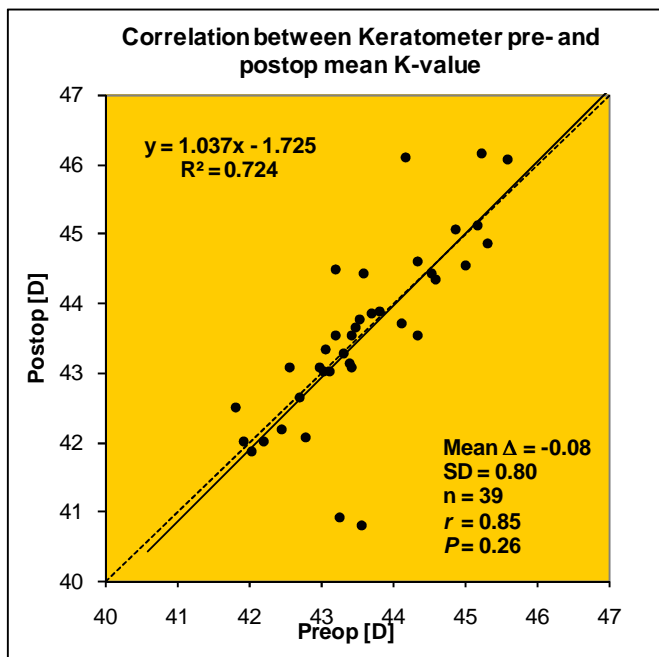
Figure 3.6 Correlation between IOLMaster pre- and postop mean K-values

Correlation between IOLMaster pre- and postop mean K-value

y = 1.039x - 1.672
R² = 0.922

Mean Δ = 0.06
SD = 0.43
n = 39
r = 0.96
P = 0.18

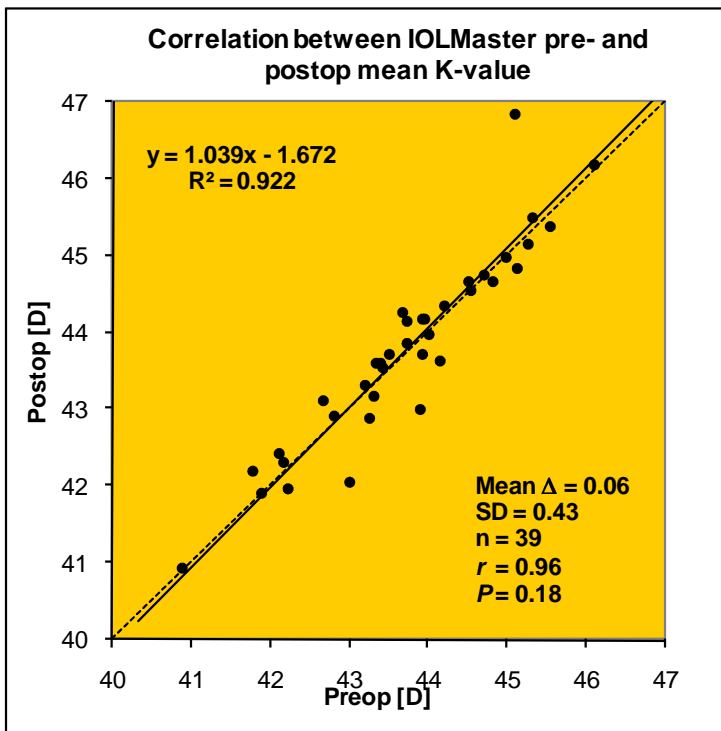**Figure 3.7** Correlation between C-scan pre- and postop mean K-values

Correlation between C-scan postop and preop mean K-value

y = 0.933x + 2.982
R² = 0.850

Mean Δ = 0.09
SD = 0.55
n = 39
r = 0.92
P = 0.16
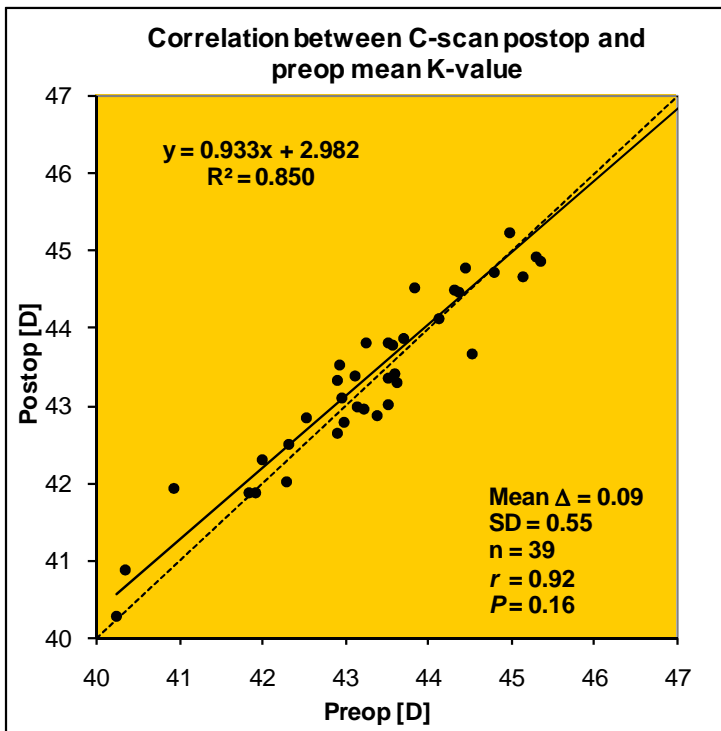
For comparison of the corneal power values among keratometer, IOL-Master and C-scan 39 eyes were available preoperatively for the same set of patients. Preoperatively the difference between keratometric values given by the IOLMaster and keratometer was 0.11 ± 0.56(SD) D ($P=0.11$) and the values between two sets of measurements were highly correlated ($r=0.92$) (Figure 3.8).

Figure 3.8 Correlation between Keratometer and IOLMaster preop mean K-values
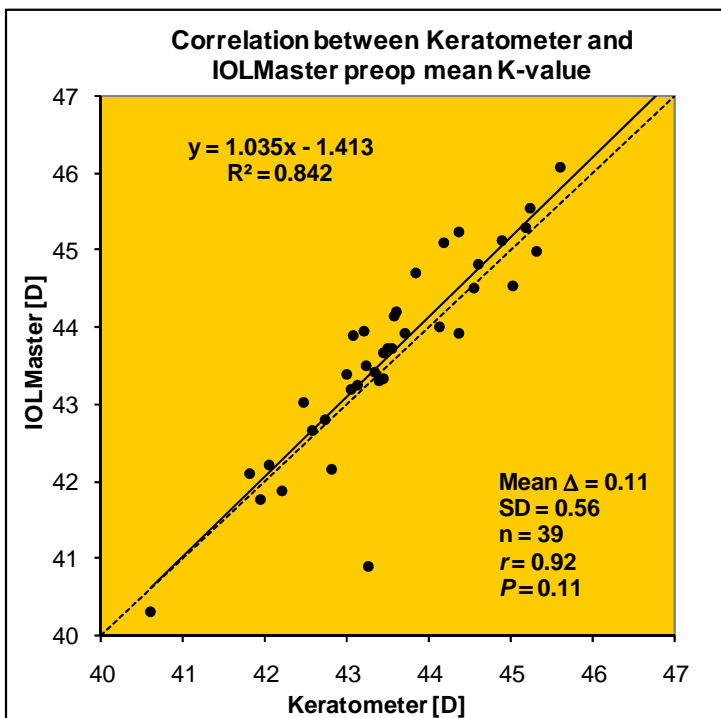
The K-readings given by the C-scan and keratometer were also highly correlated ($r=0.89$), with a mean difference of values -0.20 ± 0.63(SD) D, but statistically significant, $P<0.05$ (Figure 3.9). The negative sign in difference indicates that C-scan measures K-readings smaller than keratometer.

Figure 3.9 Correlation between Keratometer and C-scan preop mean K-values
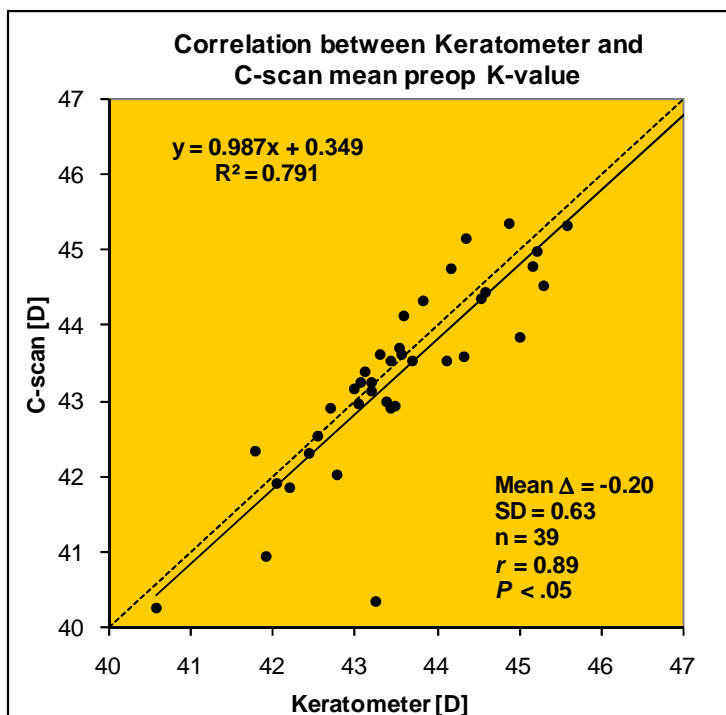
K-values given by the C-scan and IOLMaster were also highly correlated ($r = 0.98$), with a mean difference of values -0.31±0.29(SD) D and $P<.001$, (Figure 3.10). The negative sign in difference indicates that C-scan measures K-readings smaller than IOLMaster.

Figure 3.10 Correlation between IOLMaster and C-scan preop mean K-values

Correlation between IOLMaster and C-scan preop mean K-value

$y = 0.963x + 1.299$
$R^2 = 0.957$

Mean Δ = -0.31
SD = 0.29
n = 39
r = 0.98
P < .001

C-scan [D] (vertical axis)
IOLMaster [D] (horizontal axis)
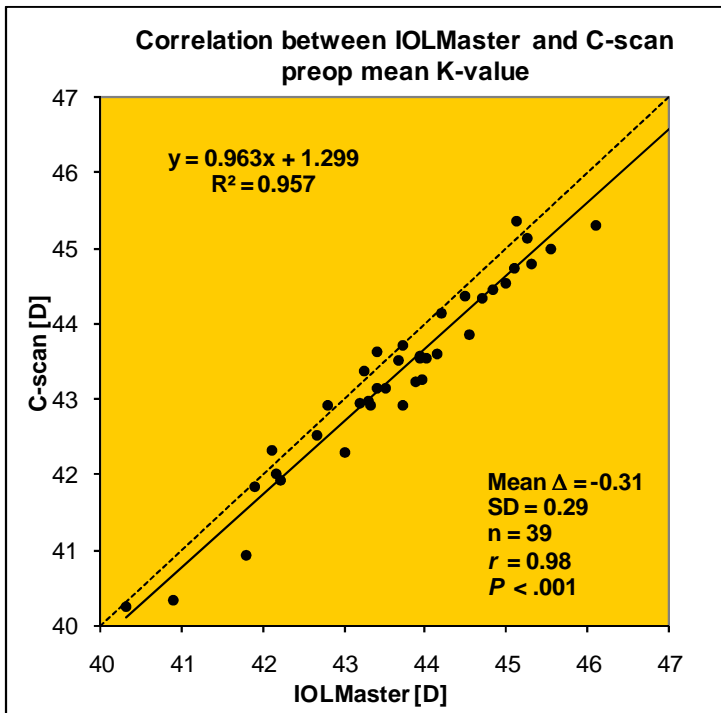
3.4. Axial length

Preoperative axial length measurement values in 39 eyes to compare optical and ultrasound biometry are shown in Table 2.

Table 3.2 Comparison of ultrasound and IOLMaster preoperative AL values

	A-scan		IOLMaster		Correlation (Pearson)	Difference (paired *t* Test)	
	Mean±SD	Range	Mean±SD	Range	*r*	Mean±SD	*P*
AL (mm)	23.14±1.30	20.44 – 26.16	23.45±1.24	21.00 – 26.46	0.97	0.31±0.32	<.001

Preoperatively the IOLMaster produces larger mean AL value 23.45±1.24(SD) mm than A-scan 23.14±1.30(SD) mm with statistically

significant mean difference of 0.31±0.32(SD) mm between IOLMaster and A-scan measurements (*P*<.001). The correlation between devices was 0.97.

In Figure 3.11 the preoperative results of the two instruments are plotted against each other. The IOLMaster measured AL systemically longer than A-scan (by 0.31 mm).

Figure 3.11 Correlation between A-scan and IOLMaster preop axial length
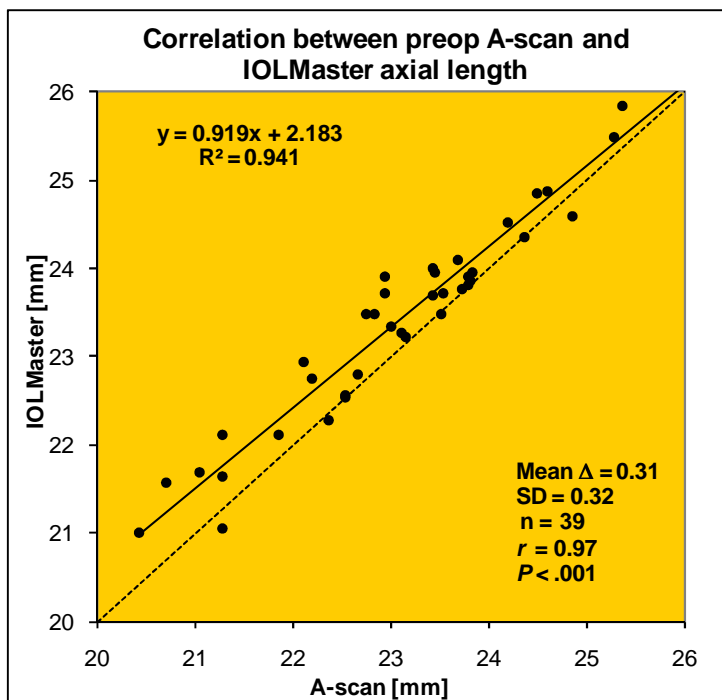
To compare AL values obtained preoperatively and postoperatively with IOLMaster, 39 eyes were available for the same set of patients. Preoperative and postoperative mean IOLMaster axial length was 23.45±1.24(SD) mm (range 21.00 – 26.46 mm) and 23.36±1.23(SD) mm (range 20.95 – 26.36 mm) respectively.

There was found a statistically significant difference of -0.09±0.11(SD) mm (*P*<.001) between preoperative and postoperative IOLMaster axial length values. The negative sign in difference indicates that postoperatively IOLMaster measured AL shorter than preoperatively. The AL values were highly correlated preoperatively and postoperatively, *r* = 0.996 (Figure 3.12).

Figure 3.12 Correlation between IOLMaster preop and postop axial length
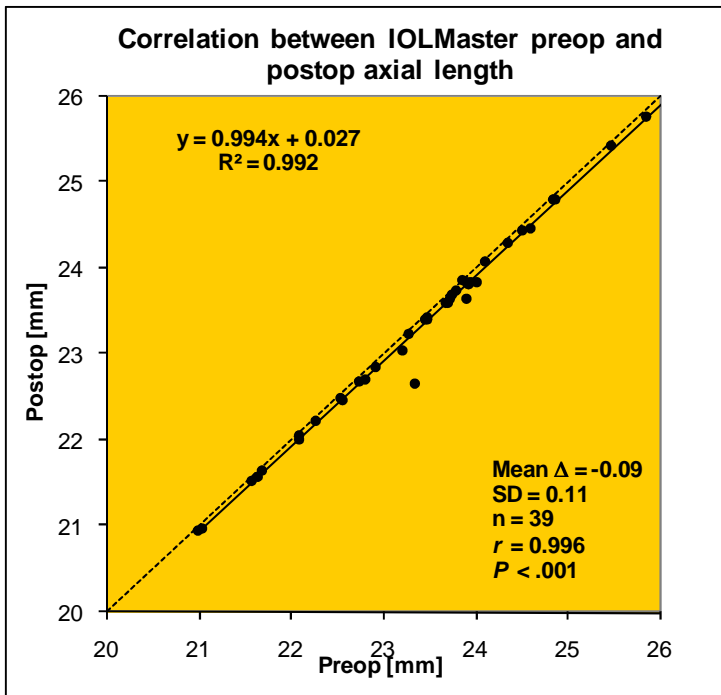
3.5. Anterior chamber depth

Preoperative anterior chamber depth measurement values in 39 eyes to compare optical and ultrasound biometry are shown in Table 3.3:

**Table 3.3** Comparison of ultrasound and IOLMaster preoperative ACD values

	A-scan		IOLMaster		Correlation (Pearson)	Difference (paired *t* Test)	
	Mean±SD	Range	Mean±SD	Range	*r*	Mean±SD	*P*
ACD (mm)	3.06±0.49	1.69 – 4.35	3.23±0.46	2.09 – 4.32	0.73	0.18±0.36	< .01

The ACD values with the IOLMaster were significantly higher (mean difference 0.18±0.36(SD) mm) than the ultrasound values (*P*<.01) and the correlation between the 2 sets of values was 0.73 (Figure 3.13).

Figure 3.13 Correlation between IOLMaster and A-scan preop ACD

Correlation between preop A-scan and IOLMaster anterior chamber depth

$y = 0.681x + 1.149$
$R^2 = 0.526$

IOLMaster [mm]

Mean Δ = 0.18
SD = 0.36
n = 39
r = 0.73
P < .01

A-scan [mm]
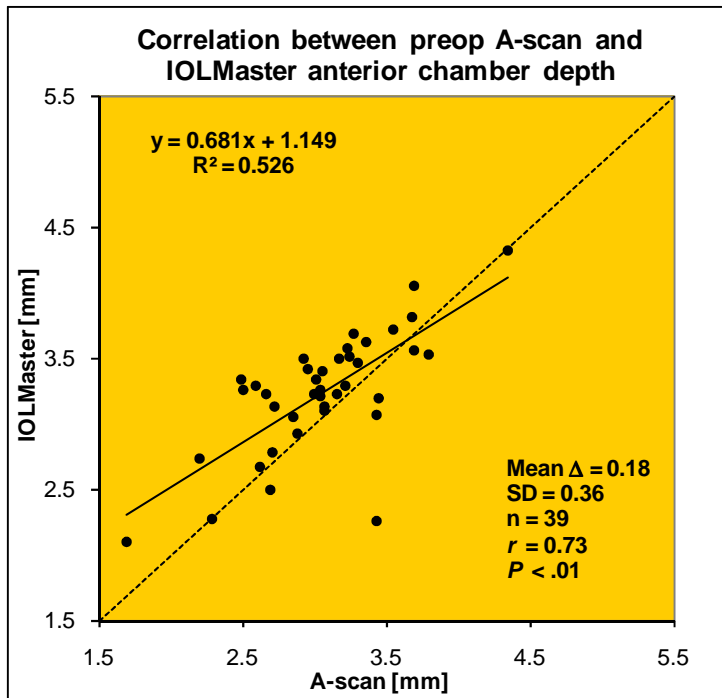
3.6. IOL A-constant

For calculation of an optimized A-constant, 33 patients were available as 6 patients were eliminated due to different types of IOL implanted than Acrysof SA60AT with recommended by lens manufacture A-constant 118.4 for A-scan. The results are given in picture 3.14.

Figure 3.14 Optimized A-constant among different IOL power calculating methods

Optimized A-constant among different IOL power calculating methods

- ■ Keratometer K-value + A-scan axial length
- ■ Keratometer K-value + IOLMaster axial lenght
- ■ IOLMaster K-value + A-scan axial length
- ■ IOLMaster K-value + IOLMaster axial lenght
- ■ C-scan K-value + A-scan axial lenght
- ■ C-scan K-value + IOLMaster axial length
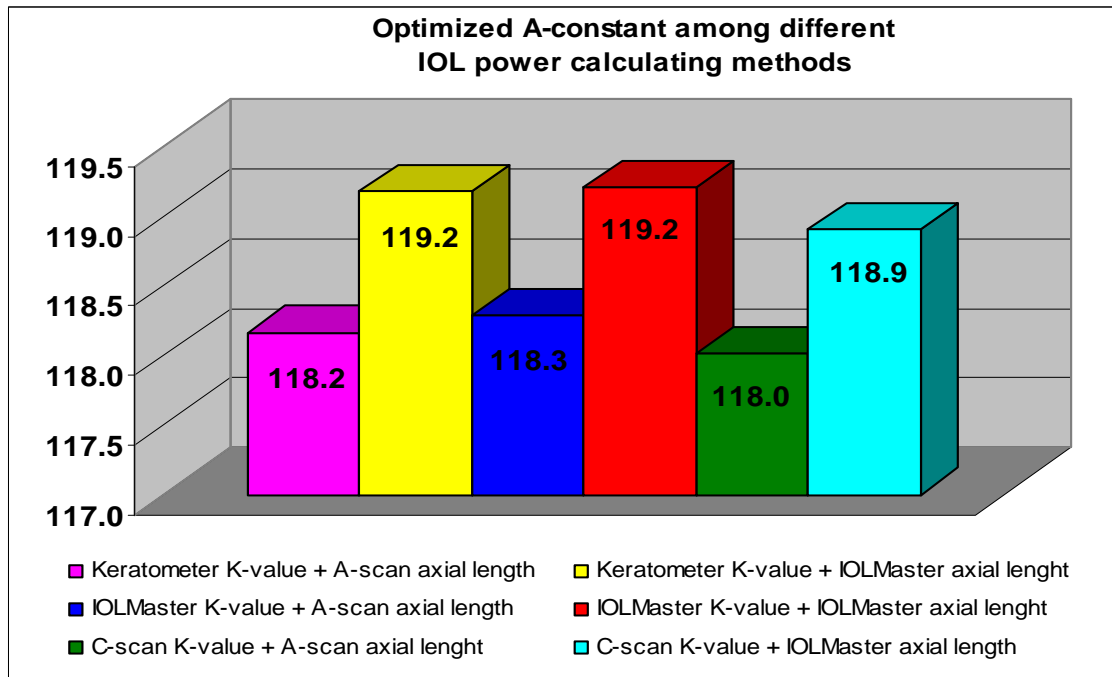
3.7. Refractive error calculation

Comparison analyses were made between the mean numerical error (MNE) and the mean absolute error (MAE) for each IOL power calculating methods. The numerical error suffers from the disadvantages of averaging both positive and negative errors. Thus, the absolute error is the more useful measure of the true size of error. The estimates of MNE and MAE using six refractive error calculating methods are presented in Figure 3.15 and 3.16.

Figure 3.15 Predicted MNE among different IOL power calculating methods

Mean numerical refractive error among different IOL power calculating methods

- ■ Keratometer K-value + A-scan axial length
- ■ Keratometer K-value + IOLMaster axial length
- ■ IOLMaster K-Value + A-scan axial length
- ■ IOLMaster K-value + IOLMaster axial length
- ■ C-scan K-value + A-scan axial length
- ■ C-scan K-value + IOLMaster axial length
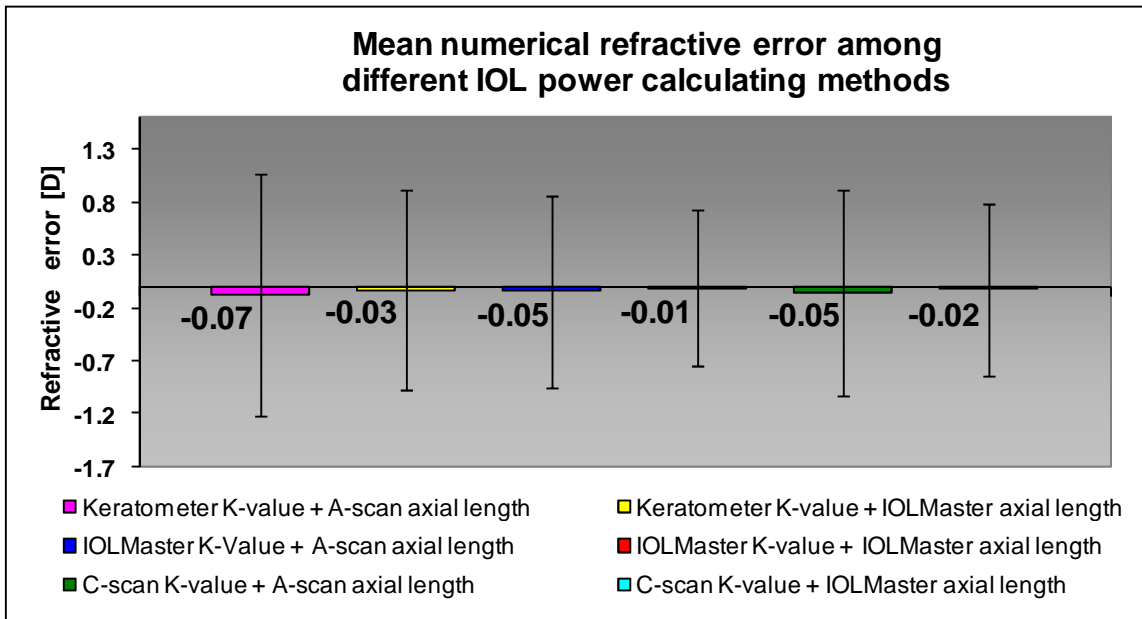
Figure 3.16 Predicted MAE among different IOL power calculating methods

Mean absolute refractive error among different IOL power calculating methods

- ■ Keratometric K-value + A-scan axial length
- ■ Keratometric K-value + IOLMaster axial length
- ■ IOLMaster K-value + A-scan axial length
- ■ IOLMaster K-value +IOLMaster axial length
- ■ C-scan K-value + A-scan axial length
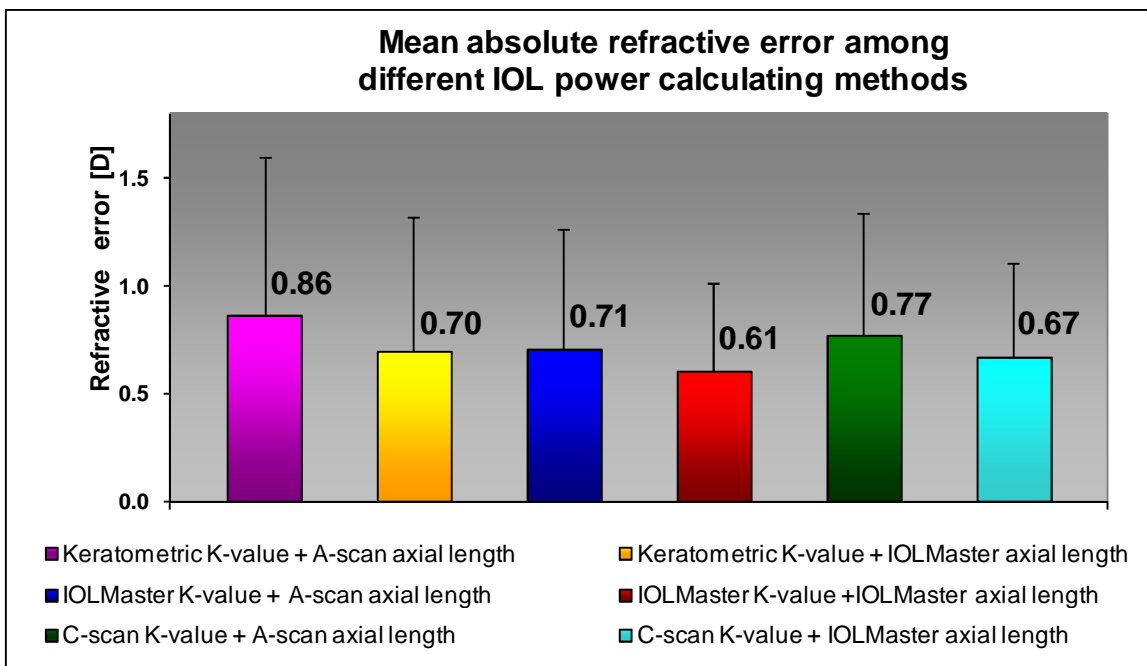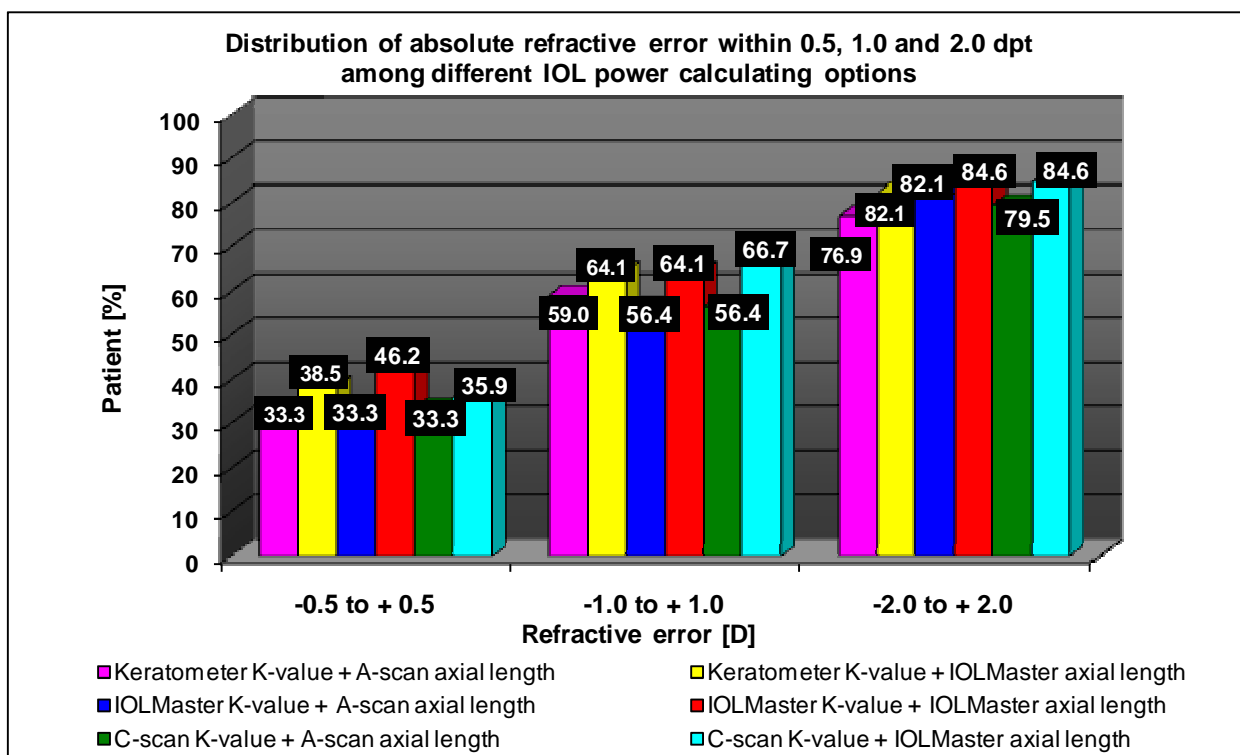- ■ C-scan K-value + IOLMaster axial length

Figure 3.17 shows distribution of patients with predicted mean absolute refractive error within $\pm$0.5, $\pm$1.0 and $\pm$2.0D.

Figure 3.17 Distribution of predicted mean absolute refractive error within +0.5,
+1.0 and +2.0D among different IOL power calculating methods

**Distribution of absolute refractive error within 0.5, 1.0 and 2.0 dpt
among different IOL power calculating options**

■ Keratometer K-value + A-scan axial length
□ Keratometer K-value + IOLMaster axial length
■ IOLMaster K-value + A-scan axial length
■ IOLMaster K-value + IOLMaster axial length
■ C-scan K-value + A-scan axial length
□ C-scan K-value + IOLMaster axial length

4. Discussion

4.1. Keratometric readings

Preoperatively and postoperatively K-readings for three devices (Keratometer, IOLMaster and C-scan) were all closely correlated and not significantly different with a mean difference -0.08 ± 0.80D for Keratometer (r=0.85, P=0.26), a mean difference 0.06 ± 0.43D for IOLMaster (r=0.96, P=0.18) and a mean difference 0.09 ± 0.55D for C-scan (r=0.92, P=0.16).

The keratometric values measured preoperatively with the IOLMaster and Javal-type keratometer were closely correlated (r=0.92) with a mean difference in values of 0.11 ± 0.56D, which was not significant (P=0.11). These results agree with those published by others. In a study by Nemeth, et al. (2003) (Nemeth, Fekete et al., 2003). The corneal powers measured by the IOLMaster and Javal-type keratometer were closely correlated (r=0.995, P<.001) and gave a mean difference of 0.17 ± 0.48D. Rose and Moshegov (2003) (Rose and Moshegov, 2003) found K values of IOLMaster and manual keratometer not to be significantly different (P=.61). In the study by Gantenbein (Gantenbein, Lang et al., 2003) a comparison of eye keratometric measurements showed a good correspondence between the obtained measurements by both methods, and Javal-type, yielding a significantly (P<.001) higher mean corneal refraction power than the IOLMaster.

There was found closely correlated and significantly different keratometer and C-scan preoperative K-readings with a mean difference -0.20 ± 0.63D (r=0.89, P<.05); and highly correlated and significantly different IOLMaster and C-scan preoperative K-readings with mean difference -0.31 ± 29D (r=0.98, P<.001). The negative sign in both differences indicates that C-scan measures preoperatively K values smaller than keratometer and IOLMaster. Literature review suggests that similar results were reported by other investigeators. Uçakhan, et al. (Uçakhan, 2000) compared the keratometric readings, obtained from 45 healthy eyes by Intraoperative PAR Corneal Topography System to those produced by manual keratometer, autokeratometer, corneal topography

and slit lamp PAR CTSF and estimated average differences between the measurements taken from pairs of instruments with corresponding 95% confidence intervals. The observed differences were within the agreement range and varied from 0.33 to 0.82D. Giráldez, at el. (Giráldez, 2000) compared measurements obtained from 100 normal eyes using Javal ophthalmometer, and Nidek autokeratometer, and Corneal Analysis System (EyeSys) and found that 95% confidence limits showed a lack of agreement between instruments.

The reasons of pure agreement may be that different keratometry devices may give different readings due to internal differences in calibration or different refractive index used to transform corneal radius to diopters. Another source may be the fact that a manual keratometer requires the user to align the keratometer mires along the principal meridians and corneal radius; hence the obtained values depend on subjective alignment of the mires. Potvin, et al. (Potvin R, 1996) investigated in vivo performance of corneal topography systems and compared it with manual and automated keratometry. Overall results suggested that manual keratometry is highly variable between operators and is a poor comparator for topography repeatability.

Keratometric measurements are usually presented in diopters; however all instruments measure radii of curvature which are then transformed into diopters by keratometric formula based on spherical geometry. But the corneal optics is assumed to be spherocylindrical, thus asphericity or asymmetry of corneal shape cannot be measured with all methods fairly, as they utilize formula based on spherical geometry. In this regard, manual keratometry, IOLMaster keratometry and corneal topography are reasonably accurate and reliable methods for measuring corneal contours when the surface is spherical. For aspheric corneas, corneal topography – also known as videokeratography or corneal mapping – represents a significant advance in the measurement of corneal curvature over keratometry; it provides both qualitative and quantitative information about the corneal surface with micron resolution. Unlike manual keratometry, which evaluates only four points on two orthogonal meridians separated 3 mm to 4 mm on the paracentral cornea and does not provide data from the central or peripheral cornea, the corneal topography instrument

samples 8,000 to 1,000 points within the central 3 mm and 5000 points over the entire cornea and provides greater accuracy in determining the corneal power with irregular astigmatism compared with manual keratometer. Although while performing manual keratometry examiners can see the reflected mires and the amount of given irregularity; however, seeing the mires does not help to get better measurements, but allows observers to discount the measurements as unreliable (Seitz and Langenbucher, 2000).

Overall, it can be assumed that different instruments to measure corneal power cannot be used interchangeably to obtain keratometric values for intraocular lens power calculation. Although keratometry and corneal topography have comparable accuracy in the paracentral region of the cornea, keratometry gives no information about the peripheral cornea or about asymmetry of the cornea. In this regard, corneal topography is becoming increasingly important in the determination of intraocular lens power in difficult cases such as patients undergoing combined cataract extraction and penetrating keratoplasty or post refractive cataract surgery..

After the Food and Drug Administration first approved the excimer laser in October 1995 for correcting mild to moderate nearsightedness, refractive surgery has become no longer something just for risk takers. Refractive surgery is now a mainstream. The number of patients who have had keratorefractive surgery is increasing every year. These refractive surgery patients expect similar results after cataract surgery. Thus, calculation of IOL power in patients with a history of refractive surgery is becoming a crucial part in maintaining the level of visual satisfaction in these patients.

Corneal topography together with other methods has become now a subject of close investigation in IOL power calculation methods. The recent literature shows no clear consensus about what method of IOL calculation is best after refractive surgery.

Several authors reported cases when corneal topography was not adequate to determine corneal power in patients with previous photorefractive keratectomy or penetrating keratoplasty. Weindler, et al. (Weindler, Spang et al., 1996) compared the standard keratometry with the computer assisted

corneal topography performed on 43 eyes with irregular postoperative astigmatism following penetrating keratoplasty. Based on the results of this study, the authors considered standard keratometry more reliable to identify patients with high postoperative astigmatism following penetrating keratoplasty. In a non-randomized, prospective, cross-sectional, clinical study (n=31) Seitz, et al. (Seitz, Langenbucher et al., 1999) assessed the validity of corneal power measurement (subjective refractometry, standard keratometry, corneal topography, and pachymetry) and standard intraocular lens power calculation after photorefractive keratectomy (PRK). They found that direct power measurements underestimated corneal flattening after PRK by 24% on average. Corneal topography analysis seemed to increase the risk of error. However, because the study was retrospective and theoretical, the authors emphazied the need for a large prospective investigation to validate the findings. Ladas (Ladas, Boxer Wachler et al., 2001) reported two cases when corneal topography was used to determine corneal power to calculate intraocular lens power in two eyes with previous photorefractive keratectomy, who subsequently underwent cataract extraction years later. Intraocular lens calculations after photorefractive keratectomy resulted in a hyperopic postoperative refractive error requiring implantation of a piggyback intraocular lens. The authors found that corneal topography (with their device used) was a poor method to measure central corneal power and concluded that the clinical history method was the best. Randleman, et al. (Randleman, Loupe et al., 2002) retrospectively reviewed 10 eyes to compare the accuracy of several techniques for calculating IOL power after laser in situ keratomileusis. Corneal power was measured by manual keratometry, refractive history, contact lens overrefraction, videokeratography, and an average of the refractive history and contact lens methods. Authors found corneal topography and K readings were poor methods; clinical history and contact lens overrefraction were better methods, but an average of these last two was best. Kim, et al. (Kim, Lee et al., 2002) determined that the clinical history was the best method followed by the contact lens overrefraction as the second best method. Stakheev and Balashevich (Stakheev and Balashevich, 2003) found no methods very good

and suggested using multiple methods and selecting the lowest corneal power as determined by these methods in order to decrease the chance of postoperative hyperopia. Argento, et al. (Argento, Cosentino et al., 2003) found both contact lens overrefraction was a poor method to evaluate corneal curvature; clinical history method was the best, while corneal topography as second best.

On the other hand, positive results by using corneal topography for IOL power calculation for patients having previously undergone refractive surgery were published. Celikkol, et al. (Celikkol, Pavlopoulos et al., 1995) used a computerized videokeratography-derived corneal curvature value for intraocular lens calculations to compare with keratometric value standard keratometry, contact lens overrefraction, and refractions before and after radial keratotomy. Results suggested that, using the keratometric values, derived from computerized videokeratography after radial keratotomy for intraocular lens calculations was more accurate than using keratometric values measured by routine methods. Cua, et al. (Cua, Qazi et al., 2003) reported two cases when patients with irregular corneal astigmatism had an IOL exchange after a "surprise" post-cataract-surgery refraction. The central corneal power before IOL exchange was assessed using manual keratometry, computerized videokeratography maps, and contact lens overrefraction. The computerized videokeratography and contact lens overrefraction method provided the most accurate estimates of central corneal power in these 2 patients. Authors concluded that this type of analysis might improve the accuracy of IOL calculation in patients with corneal pathology and irregular astigmatism. In a recent study by Preussner (Preussner, 2007), analysed the reasons for single errors, like axial length and corneal radii, pupil width, asphericity of cornea and IOL and IOL geometry, calculation methods, estimation of postoperative IOL position, IOL manufacturing errors, which can contribute to the overall refractive error- The author reported that an error of 0.2D can cause the dominant error in eyes after corneal refractive surgery (approximately 1.5D) if measured only by keratometry. This error could be avoided if a topographic measurement is included into the raytracing.

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A very interstenig study was conducted separately by Gelender (Gelender, 2006) and Qazi, et al. (Qazi, Cua et al., 2007), showing that keratometric values derived from corneal topography mean power maps at a specific measurement zone accurately determined the power of an IOL for planned cataract surgery in patients who have undergone prior refractive surgery. Gelender (Gelender, 2006) compared change in corneal topography mean power maps at five central zones (1.0, 1.5, 2.0, 2.5, and 3.0 mm) with the refractive change from LASIK (n=59) to determine the optimum corneal topography correlation zone. Then, the power of the LASIK-altered cornea was measured by corneal topography and applied to IOL calculations for 17 eyes undergoing cataract surgery. The results of this study showed that the 1.5-mm corneal topography zone measurements of effective power of the LASIK-altered cornea, applied to an IOL calculation formula, accurately predicted the IOL power for planned cataract surgery. Qazi, et al. (Qazi, Cua et al., 2007) in their study concluded that the corneal topography zone of 5.0 mm total axial power and 4.0 mm total optical power can be used to more accurately predict true corneal power than the history-based method and may be particularly useful whenever pre-LASIK data are unavailable.

Certainly the state-of-the-art imaging techniques of the cornea are developing rapidly and mainly because of recent advances in refractive surgery. Most valuable for the detection of postoperative astigmatism, the planning of removal of sutures, the postoperative fitting of contact lenses, the evaluation of irregular astigmatism especially after penetrating keratoplasty, corneal topography is becoming *the* essential preoperative diagnostic procedure in patients undergoing cataract surgery; this applies in particular with regard to previous refractive surgery ,even more so after several laser ablations, unavoidingly leading to multifocal corneal profiles.

Refractive surgery patients nowadays have very high expectations. They have enjoyed years of spectacle independence and expect similar results after cataract surgery. While there are many factors affecting the accuracy of IOL calculations after keratorefractive surgery, the primary problem is that current methods to measure the central corneal curvature (keratometry and

topography) after keratorefractive surgery are inaccurate. The solution to this problem is foreseen in a combination of mathematical calculations for the optimal curvature and new methods or technology to directly measure the existing individual corneal curvature. Introducing 3D topography, slit-scan imaging, ray tracing, very high frequency ultrasonography or light interference technologies together with improvements and innovations in IOL technology may bring further diagnostic improvements for patients preparing for cataract surgery.

4.2. Axial length and anterior chamber depth

Axial length

The results of the current study revealed that preoperative optical biometry produces statistically significant larger mean axial length measurements compared to applanation ultrasound, represented by a difference 0.31 ± 0.32 mm; the measurements of two biometry method were closely correlated (r=0.97) and significantly different (P<.001).

The difference in axial length between two techniques was also found in other studies (Drexler, Findl et al., 1998; Findl, Drexler et al., 2001; Eleftheriadis, 2003; Gantenbein, Lang et al., 2003; Goyal, North et al., 2003; Nemeth, Fekete et al., 2003; Rose and Moshegov, 2003; Tehrani, Krummenauer et al., 2003(a); Olsen, 2007; Ueda, Taketani et al., 2007). Olsen, et al. (Olsen, 2007) reported axial length measured by US and IOLMaster as 23.45 and 23.07 mm respectively. Ueda, et al. (Ueda, Taketani et al., 2007) showed statistically different AL values of 23.33 and 23.12 mm (P< .00001)

obtained by two techniques. In a study by Sheng, et al. (Sheng, Bottjer et al., 2004) the two instruments showed modest agreement with each other with mean difference of +0.12 mm; 95% LoA, -0.39 to +0.64 mm; P>.0125. In a study by Nemeth, et al. (Nemeth, Fekete et al., 2003) the PCI values were significantly higher than those of the ultrasound A-scan (mean difference -0.39$\pm$0.36 mm, r=0.985, P<.001). Thehrani, *et al.* (Tehrani, Krummenauer et al., 2003(a)) reported a statistically significant (P<.001) mean difference of

0.16±0.27 mm and 0.15±0.35 mm in two groups of patients. Rose and Moshegov (Rose and Moshegov, 2003) reported a difference of 0.15 mm (*P*=0.011) and Findl, et al. (Findl, Kriechbaum et al., 2003) reported (n=696) the fifference of 0.15 mm versus 0.22 mm between experienced and less experienced operatore (*P* < .01). In the separate studies by Goyal, et al. (Goyal, North et al., 2003) and Verhulst and Vrijghem (Verhulst and Vrijghem, 2001) the difference of 0.2 mm between A-scan ultrasound and IOLMaster was found. Eleftheriadis (Eleftheriadis, 2003) in the study of 100 eyes estimated that the optical axial length obtained by the IOLMaster was significantly longer (*P*<.001) than the axial length by applanation ultrasound, 23.36±0.85 mm vs. 22.89±0.83 mm. Drexler, et al. (Drexler, Findl et al., 1998)found axial length measured with two techniques differed by a mean of 0.46 mm.

Results of this study agree with others in so far as ultrasound applanation can underestimate the AL. The most common error in the contact technique is corneal compression. This inevitably occurs because the eye is soft, thus mechanically compressible as the cornea is indented by even minimal pressure from the probe tip. The lower the intraocular pressure, the softer the eye and the more significant the corneal compression. Therefore, the amount of compression can vary not only with operator's experience but also even with the same operator. During contact US measurements the probe can applanate the cornea and shorten the AL by an average of 0.14 to 0.36 mm (Olsen and Nielsen, 1989).

The second most common error is misalignment. In optical biometry, measurements are made parallel to the visual axis because the patient fixates on a beam within the instrument. In contrast, in ultrasound biometry measurements are made along the anatomic or optical axis.

Another possible explanation is light reflection. In US biometry, the sound is reflected at the internal limiting membrane; in optical biometry light is reflected at the pigment epithelial layer. The resulting difference is about 130 μm and may increase if the sound does not directly spot the fovea (Screcker, 1966).

Extremely dense cataracts can be a challenge because of absorption of the sound beam as it passes through the lens. A dense cataract produces multiple spikes within the lens. The posterior lens gate may be erroneously aligned along one of the echoes within the lens nucleus, resulting in an erroneously thin lens thickness and erroneously long vitreous length; this in turn may result in an error of the total length of the eye.

All together, this may result in erroneous measurements. Typically, US biometer is accurate to 0.1 to 0.15 mm (Bamber, 1988; Olsen, 1989). A 0.1 mm error can result in 0.25D (Binkhorst, 1981; Boerrigter, Thijssen et al., 1985) to 0.28D postoperative refractive error (Drexler, Findl et al., 1998) (Olsen, 1987(a)). Therefore, an error of 0.5 mm will result in 1.25 to 1.4D refractive error, and an error of 1.0 mm will result in 2.5 to 3.0D postoperative refractive error that shifts the post-op refraction towards the myopic direction. For example, in an eye with a staphyloma, a measurement taken along the anatomic axis can result in an error of 3.0 mm, which can lead to a refractive error of up to 8.00D (Holladay, Prager et al., 1986; Haigis, Lege et al., 2000). In a study by Thehrani, et al. (Tehrani, Krummenauer et al., 2003(b)) the median difference of 0.14 mm could result in a refractive error calculation of 0.36D.

The accuracy of the IOLMaster has been reported of 0.005 to 0.03 mm (Drexler, Findl et al., 1998; Findl, Drexler et al., 1998; Vogel, Dick et al., 2001; Nemeth, Fekete et al., 2003). Thus, refractive errors stemming from AL measurements with optical biometry are limited to 0.05D, which are 5 times more accurate than by applanation US.

Using the IOLMaster, the axial length was successfully measured in 85.2% of eyes in this study. Thus, the occurrence of an unsuccessful optical AL measurement was 14.8%. Haigis and Lege, (Haigis, Lege et al., 2000) reported successful measurements in 91% of eyes but healthy subjects were included in addition to cataract patients. Tehrani, et al. (Tehrani, Krummenauer et al., 2003(a)) reported a failed measurement rate of 17% using optical biometry. In study by Siahmed K, et al, (Siahmed, Muraine et al., 2001) there were 10% of failures for axial length measurement by optic biometry due to dense cataract.

Because the measurement of axial length by ultrasound biometry has traditionally been considered the most crucial step in intraocular lens (IOL) power calculation, accounting for 54% of total prediction error (Olsen, 1992), the ultra-high precision of PCI seemed promising in terms of improved accuracy in IOL power calculation. Thus, several investigations have been conducted to compare IOL power prediction using PCI and ultrasound and have shown better results from PCI than from ultrasound (Drexler, Findl et al., 1998; Findl, Drexler et al., 1998; Findl, Drexler et al., 2001; Vogel, Dick et al., 2001; Connors, Boseman et al., 2002; Eleftheriadis, 2003; Nemeth, Fekete et al., 2003; Madge, Khong et al., 2005; Ueda, Taketani et al., 2007). These studies showed that nowadays it is possible to significantly reduce the chance of a postoperative ametropia after cataract surgery with IOL implantation. Norrby (Norrby, 2008) in his study identified that preoperative estimation of axial length measured by PCI contributed only 17 % of total source of refractive error postoperatively, in compare the measurement of postoperative intraocular lens position was 35%.

Anterior chamber depth

The statistically significant difference in ACD values measured preoperatively with the IOLMaster and A-scan ultrasound in our study was found 0.18 ± 36 mm ($P<.01$), with the higher values of the IOLMaster. But there was a weak correlation between the two biometry methods ($r=0.73$).

This difference was also found in other studies. Hashemi H, et al. (Hashemi, Yazdani et al., 2005) conducted a comparison of ACD measurement by 3 devices of EchoScan, Orbscan II, and IOLMaster (n=88). There was a statistically significant difference between measurements made with the three devices ($P<.001$). The mean difference between IOLMaster and Echoscan measurements was $+0.09\pm-0.14$ mm with the 95% LoA from -0.18 to +0.36 mm. On average, IOLMaster readings were higher than Echoscan readings. Both Orbscan II and IOLMaster agreed with Echoscan in measuring ACD. In the prospective study (n=81) by Reddy, et al. (Reddy, Pande et al., 2004) ACD estimation was done by 3 methods – scanning slit topography (Orbscan II), partial coherence interferometry (IOLMaster), and contact ultrasound A-scan. There was a statistically significant difference 0.43 mm between measurements

recorded by contact A-scan and IOLMaster (P<.01), with lower value by A-scan. In a study by Sheng, et al. (Sheng, Bottjer et al., 2004) IOLMaster gave significantly longer anterior chamber depths than ultrasound (mean, +0.18 mm; 95% LoA, -0.02 to +0.37 mm; P<.0125). Nemeth, et al. (Nemeth, Fekete et al., 2003) found the significant difference of -0.28$\pm$0.68 mm (P<.001) with no correlation (r=0.079). In study by Kriechbaum, et al. (Kriechbaum, Findl et al., 2003) statistically significant (P<.01) mean difference of 0.28$\pm$0.20 mm between the IOLMaster and US was found. Findl O, et al. (Findl, Kriechbaum et al., 2003) found applanation US measured ACD shorter than the IOLMaster (n=462); mean numerical difference was 0.19 mm and 0.29 mm between experienced and less experienced operator (P<.05). Lam AC, et al. (Lam, Chan et al., 2001) reported the mean difference in anterior chamber depth between the IOLMaster and ultrasound biometry was 0.15, with 95% limits of agreement between 0.34 and − 0.03.

Compared to ACD measured with US biometry in contact technique, the IOLMaster ACD is likely to be slightly longer because it is not affected by possible globe indentation, as might be in the case with contact US.

Another reason for the difference in the ACD measurements between the IOLMaster and ultrasound A-scan may be related to the lack of pupil dilation. In this case the distance between the anterior corneal surface and the iris may be erroneously measured as the ACD, which might cause the ultrasound measurement value to be smaller than the true value (Nemeth, Fekete et al., 2003).

Ultrasound A-scans of axial length and anterior chamber depth with an implanted IOL are more difficult to interpret and show low reliability because the artificial IOL generates a diversified echo pattern as well as a multitude of measurements artifacts at the posterior lens surface and in the vitreous (Artaria and Freudiger, 1984; Naeser, Naeser et al., 1989; Vetrugno, Cardascia et al., 2000; Kriechbaum, Findl et al., 2003). Thus, no biometry measurements were done post-op on the pseudophakic eyes.

Immersion ultrasound

The immersion technique of biometry is accomplished by placing a small scleral shell between the patient's lids, filling it with saline, and immersing the probe into the fluid. In compare with applanation ultrasound immersion technicue has better reproducibility, as with the immersion technique the probe tip does not come into contact with the cornea, hense there is the lack of corneal compression. Packer, et al. (Packer, Fine et al., 2002) compared immersion ultrasonography and partial coherence interferometry and found that immersion ultrasonography and PCI correlated in a highly positive manner (r=0.996) and 92% of eyes were within $\pm$0.5D of emmetropia based on immersion axial length measurements. Authors concluded that immersion ultrasonography provided highly accurate axial length measurements and permitted highly accurate IOL power calculations. Kiss, at al. (Kiss, Findl et al., 2002). evaluated the refractive outcome of cataract patents (n=45) with the two techniques. PCI and immersion US did not differ significantly (P =.28). The mean absolute error was 0.48D and 0.46D for IOLMaster and immersion US, respectively. Haigis, et al. (Haigis, Lege et al., 2000) measured AL with immersion Us 108 patients for planning of cataract surgery; postoperative refraction was predicted correctly within ±1 D in 85.7% and within ±2 D in 99% of all cases. Same result was achieved with axial length data measured with PCI after suitable transformation of optical path lengths into geometrical distances.

Partial coherence interferometry is a noncontact, user- and patient-friendly method for axial length determination and IOL planning with an accuracy comparable to that of high-precision immersion ultrasound. However, all forms of ultrasound based biometry have two basic limitations. First, they use a large 10-MHz sound wave to measure a relatively small distance. Second, the area around the center of the macula is not flat, but thinnest at the fovea, with thicker shoulders. Thus, the ultrasound beam should be properly aligned with the center of the macula to obtain true axial length.

Summary

In summary, partial coherence interferometry allows to measure axial length more accurately than ultrasound biometry with lower variability. It provides a true measure from cornea to fovea (Connors, Boseman et al., 2002). It is less operator-dependent than conventional A-scan biometry and time saving technique, showing high intrasession, intersession and interobserver precision (Drexler, Findl et al., 1998; Lam, Chan et al., 2001; Vogel, Dick et al., 2001; Santodomingo-Rubido, Mallen et al., 2002; Findl, Kriechbaum et al., 2003; Tehrani, Krummenauer et al., 2003(b); Sheng, Bottjer et al., 2004).

Partial coherence interferometry showed advantages in patients with asymmetrically shaped globes, eccentric fixation, silicone oil-filled eyes and a fearful/nervous disposition (Lege and Haigis, 2004). Disadvantages of the system were revealed in cases of dense cataract,retinal detachment, severe opacities along the visual axis and poor patient cooperation (Lege and Haigis, 2004). Immersion ultrasound will be necessary for patients who cannot be measured by optical coherence to ensure the same high level of accuracy.

The noncontact optical method, which is essentially operator independent, gives a significantly more reliable biometry before cataract surgery, especially in the case of less experienced operators. The high accuracy, high repeatability, low variability of the IOLMaster suggests that this observer independent technique should become the standard for axial length measurement. Axial eye length with an error of approximately 0.2D is no longer the dominating error if the measurements are performed by interferometry; the same is true for corneal radii in normal eyes (Preussner, 2007). But if the total error threshold is below the error of refraction, the prediction accuracy of IOL power calculation formula must be improved.

4.3 Refractive error prediction

In this study mean numerical error was -0.07$\pm$1.14, 0.03$\pm$0.94, -0.05$\pm$0.91, -0.01$\pm$0.74, -0.05$\pm$0.97 and -0.02$\pm$0.81(SD) D for six device combinations:

1. A-Scan AL + Keratometer K-value (combination A_1K_1)
2. IOLMaster AL + Keratometer K-value (combination A_2K_1)
3. A-Scan AL + IOLMaster K-value (combination A_1K_2)
4. IOLMaster AL + IOLMaster K-value (combination A_2K_2)
5. A-Scan AL + C-Scan K-value (combination A_1K_3)
6. IOLMaster AL + C-Scan K-value (combination A_2K_3)

But the numerical error suffers from the disadvantages of averaging both positive and negative errors. Thus, the absolute error is the more useful measure of the true value of the error. The mean absolute error in this study was 0.86$\pm$0.74, 0.70$\pm$0.62, 0.71$\pm$0.56, 0.61$\pm$0.41, 0.77$\pm$0.57 and 0.67$\pm$0.44(SD) D for six abovementioned methods A_1K_1, A_2K_1, A_1K_2, A_2K_2, A_1K_3 and A_2K_3 respectively. In this study the smallest error is predicted by method A_2K_2 (IOLMaster), followed by the method A_2K_3 (combination of C-scan K-value and IOLMaster axial length). The numerical and absolute errors in this study are slightly higher than those published by other investigators. In the recent study by Olsen (Olsen, 2007) the average absolute IOL prediction error (observed minus expected refraction) in 461 consecutive cataract operations was 0.65D with ultrasound and 0.43D with PCI using the Olsen formula, which uses 5-variable ACD prediction method (*P*<.00001). The 2-variable ACD method (Haigis formula) resulted in an average error in PCI predictions of 0.46D, which was significantly higher than the error in the 5-variable method (*P*<.001). The number of predictions within ±0.5D, ±1.0D and ±2.0D of the expected outcome was 62.5%, 92.4% and 99.9% with PCI, compared with 45.5%, 77.3% and 98.4% with ultrasound. Ueda, et al. (Ueda, Taketani et al., 2007) found significant difference between IOLMaster and US in the mean predictive absolute refractive error. The mean absolute predictive error was 0.57$\pm$0.26(SD) D with the IOLMaster and 0.79$\pm$0.53(SD) D with US (*P*<.0001). Connors, et al. (Connors, Boseman et al., 2002) also found that IOLMaster was significantly better in the mean absolute error (0.533D$\pm$0.589(SD) versus 0.757$\pm$0.723(SD); *P*=.012) and in the percentage of eyes within $\pm$0.5D (61.2%

versus 42.3%; *P*=.003) and $\pm$1.0D (87.4% versus 77.5%; *P*=.05) of the predicted refraction. Drexler, et al. (Drexler, Findl et al., 1998) reported a 27% of improvement with partial coherent interferometry using SRK II formula. The study found $\pm$1D error with ultrasound of 72.9% was improved to 85% with IOLMaster and $\pm$2D error with US of 96.4% was improved to 100% with IOLMaster.

Overall, the percentage of MAE reported in different studies was 77.5-86.7% with ultrasound and 84.7-92.4% for IOLMaster within $\pm$1.0D; and 96.4-99% with US and 99.0-100% with IOLMaster within $\pm$2.0D of predicted error (Drexler, Findl et al., 1998; Haigis, Lege et al., 2000; Connors, Boseman et al., 2002; Rose and Moshegov, 2003; Olsen, 2007). The expected error within $\pm$1.0D in this study was only 59% using conventional ultrasound and 64.1% with IOLMaster; and 76.9% and 84.6% cases came within $\pm$2.0D with US and IOLMaster respectively. However, in this study multiple device combination was compared among each other: A_1K_1, A_2K_1, A_1K_2, A_2K_2, A_1K_3 and A_2K_3.

The study has shown an improvement in IOL power calculation using partial coherence interferometry compared to ultrasound in three combination pairs. The mean absolute error in theoretical refractive outcome decreased from 0.86D with combination A_1K_1 (ultrasound AL and keratometer K-readings) to 0.70D with combination A_2K_1 (IOLMaster AL and keratometer K-readings); from 0.71 with combination A_1K_2 (ultrasound AL and IOLMaster K-readings) to 0.61D with combination A_2K_2 (both AL and K-readings with IOLMaster); and from 0.77 with combination A_1K_3 (ultrasound AL and corneal topography keratometer K-readings) to 0.67D with combination A_2K_3 (IOLMaster AL and corneal topography K-readings). In the same manner, results have shown that corneal power measuring devices produce different values in predicting the theoretical refractive outcome. The MAEs were 0.86, 0.77 and 0.71D for combination A_1K_1, A_1K_2 and A_1K_3 (ultrasound AL with K-readings of keratometer, corneal topography and IOLMaster respectively); and 0.70, 0.67 and 0.61D for combination A_2K_1, A_2K_2 and A_2K_3 (IOLMaster axial length with K-readings of keratometer, corneal topography and IOLMaster respectively).

The distribution of predicted errors revealed that 59%, 56.4% and 56.4% of patients would have come within ±1.0D if device combinations A_1K_1, A_1K_2 and A_1K_3 respectively were used (ultrasound AL with K-readings of keratometer, corneal topography and IOLMaster respectively); while 67.7%, 64.1% and 6.1% of patients would have come within ±1.0D if device combinations A_2K_1, A_2K_2 and A_2K_3 respectively were used (IOLMaster AL with K-readings of keratometer, corneal topography and IOLMaster respectively). The distribution of predicted errors within ±2.0D among combinations A_1K_1, A_1K_2 and A_1K_3 (ultrasound AL with K-readings of keratometer, corneal topography and IOLMaster respectively) was 76.9%, 82.1% and 79.5% respectively; and 82.1%, 84.6% and 84.6% within ±2.0D among combinations A_2K_1, A_2K_2 and A_2K_3 respectively. These results have shown clearly that the expected outcome was higher when using PCI technique compared to ultrasound. This represents 8%, 12% and 16.7% improvement in accuracy of prediction error within $\pm$1.0D of emmetropia among A_1K_1, A_1K_2, and A1K3 vs. A_2K_1, A_2K_2, and A_2K_3.

Best predicted refractive outcome of 0.61$\pm$0.41(SD) D for cataract patients was achieved by the combination A_2K_2 (both AL and K-readings with IOLMaster), followed by the method A_2K_3 (IOLMaster axial length and corneal topography K-values) 0.67$\pm$0.44(SD) D; the difference between two methods is minimal and practically irrelevant clinically. Interestingly, the distribution of MAEs with A_2K_3 was slightly higher than with A_2K_2 (64.1% vs. 66.7%) within $\pm$1.0D and identical within $\pm$2.0 (84.6% vs. 84.6%). The least accurate result of 0.86$\pm$0.74D was yielded by method A_1K_1 (US axial length and keratometer K-values). The actual posoperative refrcative outcome patients in this study was 0.81$\pm$0.67D, which is very close to predicted by combination A_1K_1. It has to be mentioned that combination A_1K_1, or conventional technique to calculate IOL power, at the time the preoprative data were collected was preferred method among sergeons in the clinic. Overall, methos (A_2K_1), (A_2K_2), and (A_2K_3) with axial length measured by IOLMaster showed better results in prediction of MAE than methods (A_1K_1), (A_1K_2) and (A_1K_3) with axial length measured by ultrasound.

However, the results of the present study are worse than published by other authors in regard US vs. PCI techniques. The possible explanation for higher values of MAE and lower accuracy ± 1.0 and ± 2.0D of predicted refractive error in present study is a relatively small population group. The second reason was predetermined by fact that the present study was based on routine checkup made by various members of staff in a busy university clinic; accordingly predicted refractive outcome in IOL power calculation was influenced by several variables like cataract performance by several surgeons and preoperative data obtained by several examiners for IOL power calculation. Thus, to find out how accurate IOL power formula would be for different IOL power calculation combinations (A_1K_1, A_2K_1, A_1K_2, A_2K_2, A_1K_3 and A_2K_3) in a given clinical setting, the A-constant was optimized in each of six combinations. The issue of accuracy of any particular IOL power calculation formula was not addressed in this study; the present study was designed to compare accuracy among different devices combinations for IOL power calculation. Hence, SRK II formula was chosen as formula of choice in the university clinic; and A-constants can be easily modified to make it surgeon-specific; and it is a simple formula to calculate. The principle of this formula is based on regression analysis of empirical data taken postoperatively; and the calculation of predicted refractive error in this study is described in methods chapter. By optimizing in this study process different A-constants were calculated for different combinations; they are 118.2, 119.2, 118.3, 119.2, 118.0 and 118.9 for (A_1K_1), (A_2K_1), (A_1K_2), (A_2K_2), (A_1K_3) and (A_2K_3) s respectively. In this study all patients were implanted with Alcon AcrySof SA60AT lenses with A-constant recommended by manufacture for A-scan of 118.4. These optimized A-constants were used in SRK II formula to predict the theoretical refractive error.

Although most of these calculated A-constants are hard to compare, as these combinations are only hypothetical (A_2K_1, A_1K_2, A_1K_3 and A_2K_3), however the A-constant calculated for combination A_2K_2 (or IOLMaster) was found to be very close to the one published by User Group for Laser Interference Biometry for this particular IOL (http://www. augenklinik. uni-wuerzburg. de/eulib/ const.htm); and A-constant calculated for combination A_1K_1 (US axial length and

keratometer K-readings) is also very close to the one ,recommended by the manufacturer. This allows the assumption that the A-constant optimizing process in this study was fairly correct, and calculated MAEs for six combinations can be expected in reality.

Results of this study along with results published by others suggest the total error in IOL power calculation decreases significantly as a result of decrease in the variability of axial length values with optical axial l ength measurement. Thus, reported in the end of XX century 54% errors originated from inaccurate axial length measurement (Olsen, 1992) are today accounted for only 17% (Norrby, 2008) of total source of refractive error postoperatively. Therefore, with PCI, the largest source of error in IOL power calculation is no longer axial length measurement but the method used to predict the postoperative IOL position in pseudophakic eye followed by the keratometry. In the study by Norrby (Norrby, 2008) preoperative estimation of IOL position was reported to be largest source of refractive error, accounting for 35%. Olsen found that for any given formula as many as 20–40% of all undesirable refractive outcomes following intraocular lens implantation may be related to inaccurate prediction of pseudophakic lens position (Olsen, 1992).

The issue of the axial position of intraocular lens in the pseudophakic eye is still poorly understood and misrepresented topic in intraocular lens power calculation. Different authors use in their formulas different variables like 'A-constant' (Sanders, Retzlaff et al., 1981; Sanders, Retzlaff et al., 1988; Sanders, Retzlaff et al., 1990), 'surgeon-factor' (Holladay, Prager et al., 1988), 'anterior chamber depth' (Hoffer, 1993) to describe lens position in the pseudophakic eye. Literature review shows that so far the position of IOL remains an empirical part in each IOL power calculation formula. In order to improve IOL calculation formulae, scientists have introduced multiple additional variables like effective lens position, index of refraction, and different adjustments for myopic and hyperopic refractive surgery in order to improve the prediction accuracy of intraocular lens position within the pseudophakic eye (Haigis, 1991). Multivariable approach in modern formulas makes them more accurate than original theoretical and regression formulae, but at the same time rather

complex. In this regard, the strength of the empirical approach (SRK and SRK-II regression formulas) is that it does not measure the position of the intraocular lens in the pseudophakic eye, but this value alone with other unknown variables in the system is implicit in the calculation of the A-constant for each lens type. It is a simple formula; and A-constants used can be easily modified to make it surgeon-specific. But, to correctly optimize any of the formulas, no complicated cases should be included. Ideally, cases with irregular or aspheric cornea should be left out, so this basic optimization can be applied with excellent results to the majority of *normal* eyes.

But, the problem is still not resolved for *abnormal* corneas (after corneal refractive surgery or keratoplasty). Nowadays refractive surgery is a mainstream. Eventually many of these patients will need cataract surgery after some time. Thus, recent literature shows increasing need for methods for IOL power calculation in patient who have had refractive surgery. Corneal refractive procedures deliberately modify the anterior surface of the cornea and its thickness to correct a refractive error. The normal prolate (convexity steeper in the center) anterior surface is converted to an oblate (convexity flatter in the center) surface. Most IOL calculation formulas may not be applied to surgically modified corneas, as their conventional variables assumed for normal spherical corneas. At present time, there is no clear consensus about which technique ought to be used to measure corneal power and what method of IOL calculation is best for patients who underwent corneal reshaping or transplantation (Seitz, Langenbucher et al., 1999; Ishikawa, Hirano et al., 2000; Ladas, Boxer Wachler et al., 2001; Kim, Lee et al., 2002; Randleman, Loupe et al., 2002; Aramberri, 2003; Argento, Cosentino et al., 2003; Cua, Qazi et al., 2003; Stakheev and Balashevich, 2003; Wang, Booth et al., 2004; Latkany, Chokshi et al., 2005; Camellin and Calossi, 2006; Gelender, 2006; Jin, Crandall et al., 2006; Mackool, Ko et al., 2006; Savini, Barboni et al., 2006; Walter, Gagnon et al., 2006; Awwad, Dwarakanathan et al., 2007; Chokshi, Latkany et al., 2007; MacLaren, Natkunarajah et al., 2007; Qazi, Cua et al., 2007; Rabsilber, Reuland et al., 2007; Shammas and Shammas, 2007; Fam and Lim, 2008; Khalil, Chokshi et al., 2008).

Refractive surgery patients nowadays have very high expectations. They have enjoyed years of spectacle independence and expect similar results after cataract surgery. While there are many factors affecting the accuracy of IOL calculations after keratorefractive surgery, the primary problem is that current methods to measure the central corneal curvature (keratometry and topography) after keratorefractive surgery are inaccurate. The solution to this problem is foreseen in combination of mathematical calculation the correct curvature and new methods or technology to directly measure complex corneal curvatures. Introdicing 3D topography, slit-scan imaging, ray tracing, very high frequency ultrasonography or light interference technologies together with improvements and innovations in IOL technolgy may bring further improvemnets in modern IOL power calculation methods.

Various A constants resulting from different measurement methods and devices, as shown here, are probably a last option to improve the present regime of approximative formulae. This approach not even includes multifocal corneal shapes (e.g. after corneal refractive surgery). A new step towards a more comprehensive approach appears necessary: highly reproducible coaxial measurements of biometric and corneal data together with their computation in a ray tracing system, as proposed recently (Preussner and Wahl, 2000; Norrby, 2004; Einighammer, Oltrup et al., 2007), could offer an elegant escape from those problems, yet unresolved.

5. Conclusion

At present, cataract surgery is one of the most frequently performed and successful operations in the world. As cataract surgery technology and intraocular lens technology have improved remarkable and become safe; the patients have been expecting better postoperative refractive results, which are determined by the precise intraocular lens power calculation. There are multiple techniques and methods to measure corneal power and axial length necessary for different IOL calculation formulae existing at present time. This study compared different devices for measuring corneal power and axial length as well as investigated retrospectively the effect of optimizing the A-constants for the SRK II IOL power calculation formula with respect to the refractive outcome of the patients. The results of IOL power calculation with optimized A-constant using the combination of three corneal power and two axial length measuring devices were assessed and compared to each other. Best predicted refractive outcome for cataract patients (with no previous corneal surgery) was achieved by combination A_2K_2 – meaning both axial length and corneal power measured with IOLMaster, closely followed by the combination A_2K_3 - combination of axial length measured with IOL Master and corneal power measured with corneal topography; the difference between them is minimal and practically irrelevant clinically. The least accurate combination is A_1K_1 – combination of axial length measured with ultrasound and corneal power measured with manual keratometry. However, it confirms our understanding that the axial length measurement is critical for the precision of the IOL formula, being best with optical axial length measurements. Retrospective analysis of results showed that corneal power and axial length values for IOL power calculation, obtained by different devices, generated different refractive outcome in terms of mean absolute error even by optimizing of A-constant.

The A-constant, integrating all occurring approximations, thus is not only *specific for an IOL type;* it is also specific for the *combination of measurement biometry/keratometry methods and devices* used.

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Acknowledgments:

It is a pleasure to express my gratitude to those who gave me the possibility to complete this thesis, which was born during my scholarship at the University Eye Hospital, Tübingen.

First of all, I would like to thank the Division "Experimental Ophthalmic Surgery" of the University Eye Hospital, Tübingen. I am deeply indebted to my supervisor Prof. Dr. Benedikt Jean, whose invaluable assistance, support and guidance, stimulating suggestions, valuable advice in science discussion and encouragement helped me through all the time of research and writing of this thesis. I was extraordinarily fortunate in having Prof. Dr. Benedikt Jean not only as my supervisor, but also as my excellent mentor and outstanding advisor. I gratefully acknowledge Dr. Jens Einighammer for his advice and sharing the literature, for reviewing my thesis and providing valuable feedback, for theoretical calculations, which became crucial contribution to this thesis. I want to thank Prof. Dr. Thomas Bende and Dipl.-Ing. Theo Oltrup for all their help, support, interest and valuable hints. I convey special acknowledgement to Gudrun Stiegler for her help dealing with administration and bureaucratic matters during my stay and my commute between Kyrgyzstan, Philippines and Germany.

I am very thankful to German Academic Exchange Service (Deutsche Akademische Austaushdeinst - DAAD) and Division "Experimental Ophthalmic Surgery" of the University Eye Hospital, Tübingen, for giving me opportunity to do my research in Germany and for providing financial support during my stay abroad.

My special thanks go to Prof. Dr. Karl-Ulrich Bartz-Schmidt, the Director of the University Eye Hospital, Tübingen, who accepted me as DAAD grantee to do my training and necessary research work. Many thanks go in particular to PD Dr. Peter Szurman who helped me organize collecting clinical data in Cataract Department. To my colleges in the clinic at the time I was collecting clinical data, to Dr. Khakima Karim-Zade, I would like to thank her for such a great friendship throughout my stay in Tübingen; to Waleska Vega, it was a pleasure to collaborate with her; to Dr. Rudolf Berret, Dr. Alexander Hermann

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