IOL Master Optical Biometry versus Conventional Ultrasonic Biometry in Intraocular Lens Power Calculations in Highly Myopic versus Emmetropic Eyes

Thesis submitted for partial fulfillment of master degree in Ophthalmology

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<th>Description</th>
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<tbody>
<tr>
<td>AL</td>
<td>Axial Length</td>
</tr>
<tr>
<td>ACD</td>
<td>Anterior Chamber Depth</td>
</tr>
<tr>
<td>AUS</td>
<td>Applanation Ultrasound</td>
</tr>
<tr>
<td>A-S</td>
<td>A – Scan</td>
</tr>
<tr>
<td>AXL</td>
<td>Axial Length</td>
</tr>
<tr>
<td>BCVA</td>
<td>Best Corrected Visual Acuity</td>
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<tr>
<td>CCT</td>
<td>Central corneal thickness</td>
</tr>
<tr>
<td>ELP</td>
<td>Estimated Lens Position</td>
</tr>
<tr>
<td>ILM</td>
<td>Inner Limiting Membrane</td>
</tr>
<tr>
<td>IOL</td>
<td>Intraocular Lens</td>
</tr>
<tr>
<td>IOL M</td>
<td>IOL Master</td>
</tr>
<tr>
<td>IOP</td>
<td>Intraocular pressure</td>
</tr>
<tr>
<td>Kav</td>
<td>Average K reading</td>
</tr>
<tr>
<td>LT</td>
<td>Lens thickness</td>
</tr>
<tr>
<td>MAE</td>
<td>Mean Absolute Error</td>
</tr>
<tr>
<td>MNE</td>
<td>Mean numerical error</td>
</tr>
<tr>
<td>NE</td>
<td>Numerical Error</td>
</tr>
<tr>
<td>PCI</td>
<td>Partial Coherence Interferometry</td>
</tr>
<tr>
<td>PSC</td>
<td>Posterior Subcapsular Cataract</td>
</tr>
<tr>
<td>RPE</td>
<td>Retinal pigment epithelium</td>
</tr>
<tr>
<td>SE</td>
<td>Spherical Equivalent</td>
</tr>
<tr>
<td>SS-OCT</td>
<td>Swept source OCT(optical coherence tomography)</td>
</tr>
<tr>
<td>WTW</td>
<td>White to white</td>
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Aim of the work

The aim of this work is to determine whether intraocular lens (IOL) power calculations for cataract surgery - as measured by postoperative refractive error - using IOL master are more accurate in improving postoperative outcomes than applanation ultrasonic biometry (AUS) in highly myopic versus emmetropic eyes.
Introduction

Cataract is the leading cause of preventable blindness worldwide. Cataract extraction with implantation of an intraocular lens (IOL) is the most frequently performed ophthalmic surgical procedure worldwide, and perhaps the most effective surgical procedure in all of medicine. However, accurate calculation of the IOL power for attaining the desired postoperative refraction remains a research issue. (Raymond et all, 2009).

Several factors affect the refractive outcome after cataract surgery, including axial length, keratometry, and lens formulas. Of these factors the preoperative axial length measurement is a key determinant in the choice of intra-ocular lens (IOL) power when performing cataract surgery (Raymond et all, 2009).

Preoperative error in axial length measurement is the most significant factor in IOL power miscalculation. A 1 mm error in AL measurement results in a refractive error of 2.35 D in a 23.5 mm eye. This refractive error declines to 1.75 D/mm in a 30mm eye but rises to 3.75 D/mm in a 20mm eye. This means that accuracy of AL measurement is more important in a short eye than in a long one (Qing-Hua Yang et all, 2014).

Historically, the most common technique used among biometrists for AL measurement is applanation ultrasonic biometry (AUS). Recently, partial coherence laser interferometry (PCLI) for AL measurements is preferred for IOL calculation (Raymond et all, 2009).
Ultrasonic biometry requires contact of a transducer with the eye either by direct contact in the applanation technique or indirect contact in the immersion one. Differences in AL between immersion and applanation reach up to 0.36 mm due to various amounts of pressure exerted on the eye during applanation, however applanation is widely used for biometry. Studies show that errors from AL measurement with ultrasonic biometry are responsible for 54% of the refractive errors after cataract surgery and IOL implantation. (Eleftheriadis, 2003)

Applanation ultrasonic technique can result in corneal epithelial injury, infection, patient discomfort and errors due to corneal indentation. It’s also confounded by certain clinical problems such as globe deformities, myopic staphyloma, and silicone oil tamponade. (Roessler et al, 2014)

Partial coherence laser interferometry (PCLI), which is based on the principle similar to that of the optical coherence tomography has been developed to overcome ultrasound limitation. This method doesn’t need contact with the patient so it has the advantage of giving the patient less discomfort and has a low observer error. (Moon et al, 2014)

When staphylomas are present in highly myopic eye, PCLI could be an advantage as it allows evaluation of the axial length along the visual axis. (Roessler et al, 2014)

In eyes with a 4+ nuclear sclerotic cataract or a white cataract, PCLI cannot get a reading. The patient has to be able to achieve fixation, and the light from the instrument has to be able to reach the fovea and return to the detector. The technology will not work
with cataracts that reduce the patient’s visual acuity to 3/60 to finger counting, i.e. about 5% to 10% of patients in a typical practice. In those patients, we must use the available back-up technology, which is ultrasound. (Holladay, 2009)

Since partial coherence laser interferometry (PCLI) relies on adequate foveal fixation, eyes with corneal scarring, posterior capsule plaques, eccentric fixation, and macular degenerations fail to obtain reliable results (Rajan et al., 2002).
The A-scan ultrasound

Sound is a vibratory disturbance within a liquid or solid that travels in a wave pattern. Sound is audible to the human ear when the sound frequency is between 20 hertz (Hz) and 20,000 Hz. To be considered ultrasound, sound waves must have a frequency more than 20,000 Hz (20 KHz), rendering them too high in frequency to be audible to the human ear. (Byrne & Green, 2002)

The principles of ocular ultrasound are the same as other applications of this technology. The sound waves that have a frequency greater than 20 kHz travel along the tissues and are reflected back to the transducer. When the sound waves return, a piezoelectric crystal in the transducer vibrates, resulting in electrical impulses that are translated into an image or other data. (Shlensky & Alexander, 2015)

Higher frequency waves have a lower tissue penetration capability than low frequency waves, but have better resolution. Most A-scan and B-scan ultrasound probes use extremely high frequency of approximately 10 million Hz (10 MHz) which is predesigned by the manufacturer. These high frequent waves allow for excellent resolution and restricted depth of penetration. (Byrne, 1995)

All waves including ultrasonic, have predictive behaviors based on the properties of the medium they travel through. Sound waves have a higher velocity when traveling through solids than through liquids. This is an important principle to understand because
the eye is composed of both. In A-scan biometry, the ultrasonic waves travel through the solid cornea, the liquid aqueous, the solid lens, the liquid vitreous, the solid retina, choroid, sclera, and then orbital tissue; therefore, it continually changes velocity. (Shlensky & Alexander, 2015)

The sound velocity through the cornea and the lens (average lens velocity for the cataract age group, ie, approximately 50-65 y) is 1641 meters/second (m/s), and the sound velocity through the aqueous and vitreous is 1532 m/s. The average velocity through the phakic eye is 1555 m/s, while it is 1532 m/s through the aphakic eye. The velocity through the pseudophakic eye is 1532 m/s plus the correction factor for the intraocular lens (IOL) material. (Hoffer et al, 1994)

Ultrasonic examinations of soft tissues use reflective systems analogous to those used in radar and sonar. This approach allows examination within a thin “slice” through tissue structures. In A-scan biometry, one thin parallel sound beam is emitted from the probe tip at a frequency of approximately 10 MHz, then reflected back to the probe tip as the sound beam strikes each interface. An interface is the junction between two media of different velocities and densities, which in the eye, includes the anterior corneal surface, the aqueous-anterior lens surface, the posterior lens capsule-anterior vitreous, the posterior vitreous-retinal surface, and the choroid-anterior scleral surface. (Coleman, 2005)

The echoes that are received back into the probe from each interface are converted by the biometer to spikes arising from a baseline. If the difference in the two media at each interface is great, the echo is strong and the displayed spike is high. If that difference is small, the echo is weak and the spike is short (eg, vitreous floaters, posterior vitreous