Design of a Test Bench for Intraocular Lens Optical Characterization.

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Abstract: The crystalline lens is the responsible for focusing at different distances (accommodation) in the human eye. This organ grows throughout life increasing in size and rigidity. Moreover, due this growth it loses transparency through life, and becomes gradually opacified causing what is known as cataracts. Cataract is the most common cause of visual loss in the world. At present, this visual loss is recoverable by surgery in which the opacified lens is destroyed (phacoemulsification) and replaced by the implantation of an intraocular lens (IOL). If the IOL implanted is mono-focal the patient loses its natural capacity of accommodation, and as a consequence they would depend on an external optic correction to focus at different distances. In order to avoid this dependency, multifocal IOLs designs have been developed. The multi-focality can be achieved by using either, a refractive surface with different radii of curvature (refractive IOLs) or incorporating a diffractive surface (diffractive IOLs). To analyze the optical quality of IOLs it is necessary to test them in an optical bench that agrees with the ISO11967-2 1999 standard (Ophthalmic implants. Intraocular lenses. Part 2. Optical Properties and Test Methods). In addition to analyze the IOLs according to the ISO standard, we have designed an optical bench that allows us to simulate the conditions of a real human eye. To do that, we will use artificial corneas with different amounts of optical aberrations and several illumination sources with different spectral distributions. Moreover, the design of the test bench includes the possibility of testing the IOLs under off-axis conditions as well as in the presence of decentration and/or tilt. Finally, the optical imaging quality of the IOLs is assessed by using common metrics like the Modulation Transfer Function (MTF), the Point Spread Function (PSF) and/or the Strehl ratio (SR), or via registration of the IOL’s wavefront with a Hartmann-Shack sensor and its analysis through expansion in Zernike polynomials.

Keywords: Intraocular lens, Cataracts, MTF, aberration & wavefront analysis.

1. Introduction
In the eye, the crystalline lens provides the ability of focusing objects placed at different distances in a process referred as accommodation. The crystalline grows throughout life increasing in size and rigidity. Due to this growth the accommodation capability decreases with age (presbyopia) and an external optical correction is often needed to focus at objects located at near distances. At the same time, the lens becomes gradually opacified causing what is known as cataracts. Cataract is the most common cause of visual loss in the world [1]. To recover this visual loss the crystalline has to be extracted and an intraocular lens (IOL) implanted.
Harold Ridley made the first IOL implantation in 1949. He used a Poly-Methyl- Metacrilate IOL of +18D that was the refractive power attributed to the crystalline lens in those days. Because not all the eyes have
the same characteristics, many of the implanted ones turned out with high postoperative refractive errors. Nowadays, the development of the IOL manufacturing, the measuring of ocular parameters techniques and the surgical procedure enables a high precision on calculating the correct power of the IOL. Moreover, in some cases of high ametropia, the surgery involves removing the healthy crystalline (i.e., without opacification) to eliminate the spectacle dependency.

The actual surgery technique consists of a corneal peripheral micro-incision (2-3mm) through which an ultrasound probe is inserted. Then the crystalline is destroyed (phacoemulsification) and an IOL is implanted instead.

The aberrations of the human eye depend on the sign of the corneal and crystalline individual aberrations, which add up to give the whole eye aberration value. The mean human cornea have a positive spherical aberration [2], [3] (SA) which seems to be partially compensated in young eyes, by the negative SA of the crystalline. A spherical IOL has positive spherical aberration (that would add to the corneal aberration), and thus, the implantation of a spherical IOL may result in an emetropic eye but with reduced optical quality. For this reason in recent years aspheric IOLs have been designed to either, compensate (totally or partially) the SA of the cornea, or induce no additional spherical aberration to the eye [4],[5].

After surgery the eye loses its natural capacity of accommodation. If the IOL implanted is mono-focal, the target is that to allow patient to focus at far distance. However, the patient will depend on an external optical correction to focus at a near distance. To remove this dependency, multifocal, bifocal or accommodating IOL have been recently developed. This multifocality can be achieved using different radii of curvature (refractive IOLs), incorporating a diffractive surface (diffractive IOLs) or changing the position of the optics (accommodative IOLs).

To analyze and compare the optical quality of these lenses it is necessary to test them in an optical bench that meets the ISO119679-2 1999 [6] standard. However, with the new IOL designs it is important to study their features under conditions close to a real human eye. To do this, we have designed a test bench that allows us to use artificial corneas with different amount of SA. Several illuminating sources with different spectral distributions can also be used. Moreover, our test bench permits to study the optical performance of the IOLs off-axis, or under controlled decentration and/or tilt. Finally, it is worth mentioning that the IOL assessment is made from image quality analysis and by using conventional metrics such as the Modulation Transfer Function (MTF), Strehl ratio (SR) and/or the Point Spread Function (PSF), or via registration of the IOL’s wavefront with a Hartmann-Shack sensor and wavefront analysis through expansion in Zernike polynomials.

Figure 1: Sketch of the optical bench.
2. Test bench design

A sketch of the test bench is shown in Fig. 1. The main parts are: a collimator with the illumination source and test object, the model eye with the artificial cornea and the wet cell where the IOL is immersed, and finally the image and wavefront analysis part with either, a 10x microscope, or a Hartmann-Shack sensor. The latter devices may be eventually replaced by a spectrophotometer for spectral analysis of the IOLs. Lets us give further details of these components.

The Collimator.
The specific features of the collimator are:
- Illumination sources: The ISO standard considers an illuminant of wavelength 546±10nm. Our optical bench uses four types of LEDs, three quasi-monochromatic LEDs with emission bandwidth centered at $\lambda_r=637\text{nm}$, $\lambda_g=521\text{nm}$ and $\lambda_b=459\text{nm}$ and a white LED (see Fig. 2). It is then possible to address issues such as the chromatic aberration of the IOLs, which is especially relevant in multifocal diffractive lenses, where the chromatic dispersion is of opposite sign and much higher than in purely refractive IOLs.
- Test objects: Pinholes with different diameters, USAF target, grid test, etc. to check the optical performance of the IOLs from the analysis based on image quality (MTF from an edge, SR and PSF from a pinhole and contrast measurements). The pinhole diameter turned out to be a critical parameter. On the one hand, it has to be large enough to allow the pass of a certain amount of light, but on the other hand, it must be kept lower than 300 $\mu$m in order to obtain a flat wavefront to illuminate the artificial eye (Fig. 3).

![Figure 2: Power spectral distribution of the four LEDs used as light sources.](image)

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![Figure 3: Colored display of the wavefronts and their root mean square (RMS) error measured in collimated beams obtained using pinholes (Ph) of different diameters, from 50$\mu$m up to 500$\mu$m.](image)
Collimation: To test the IOLs according to the ISO standard the object has to be at infinity. To this end, the object is placed at the focal point of a collimator of 200 mm focal length. Moreover, we can also place the object at a finite distance to evaluate the performance of IOLs in near vision. Finally, to evaluate the IOL performance off axis the whole collimator is mounted in a pivoting arm branch that can rotate (Fig. 1).

Model Eye
The ISO standard considers an aberration-free artificial cornea. Actually, eye corneas are not usually aberration-free [2],[3] and some IOLs are designed to compensate partially or completely the natural corneal aberrations. A procedure to design artificial corneas with controlled amount of SA can be found in Ref. 7. We have used two of these corneas; one of them is virtually aberration-free (referred hereafter as LAO-cornea, with a measured focal length of 36mm) and the other one (referred hereafter as DCX-cornea, with a measured focal length of 40mm) that produces a wavefront with positive SA. The wavefronts as well as the measured PSFs of the corneas are shown in Fig. 4. It is worth noticing the differences in the Strehl Ratios, 0.903 for the LAO cornea, i.e. nearly diffraction limited, versus 0.598 for the DCX cornea. The analysis of the wavefronts of the two corneas through expansion in Zernike polynomials (see Table 1) shows that all the Zernike coefficients, except the 3rd order SA coefficient (boldfaced and underlined in Table 1), are very small and so are their contributions to the wavefront error. Further confirmation of the different behavior of the two corneas can be found in Fig. 5 where their MTFs are plotted. The measured MTF of the LAO cornea (solid blue line) is practically diffraction limit (dashed blue line) while the performance of the DCX (measured MTF in solid red line) is relatively far from the diffraction limit (dashed red line) because of the SA mentioned above.

Table 1: Analysis of the wavefront through expansion in Zernike polynomials of the LAO and DCX corneas. The Entrance Pupil was set to 6 mm.

| Zernike coeff. | LAO (µm) | DCX (µm) |
|---------------|----------|----------|
| Astigmatism 0º | +0.014 | +0.017 |
| (Z4) | | |
| Astigmatism 45º | +0.003 | -0.009 |
| (Z5) | | |
| Coma 0º | -0.020 | -0.003 |
| (Z6) | | |
| Coma 90º | +0.013 | +0.011 |
| (Z7) | | |
| 3rd order SA | +0.020 | +0.122 |
| (Z8) | | |
| Trefoil 0º | +0.003 | +0.010 |
| (Z9) | | |
| Trefoil 90º | +0.012 | +0.001 |
| (Z10) | | |
| 5º order SA | +0.009 | +0.015 |
| (Z15) | | |

Figure 4: Left: Wavefronts and 3rd order spherical aberration (SA) measured behind the LAO and DCX artificial corneas. Right: Point Spread Function and Strehl Ratio (SR).

Figure 5: MTFs of DCX (red lines) and LAO (blue lines) artificial corneas.
The IOL must be immersed in water or saline solution that is contained in a wet cell. According to the ISO standard, the convergent beam refracted by the artificial cornea exposes the central 3.0±0.1mm of the IOL. In our optical bench, an iris diaphragm placed in front of the cornea regulates the effective diameter of the illuminated area of the IOL. In Figure 6 we have plotted this diameter as a function of the iris diaphragm diameter. The results show a good linear relationship and then it is easy to control the size of the illuminated area of the IOL.

Finally, it is worth mentioning that precision X,Y,Z position and X and Y tilt controllers allow us to study the optical performance of the IOLs upon misalignment and/or tilt (Fig. 1).

**Image & Wavefront Analysis**

The image formed by the model eye with the IOL under test is captured by an 8-bit CCD camera through a 10X microscope. By using a pinhole or a USAF object, the optical quality of the IOL can be assessed from the MTF and PSF. Moreover, we can determine the focal length of the model eye, measure the image contrast at different cycles per millimeter as well as the resolution efficiency.

Analysis of the wavefront through expansion in Zernike polynomials and evaluation of the related aberrations induced by the IOL can be done using a Hartmann-Shack (H-S) wavefront sensor.

![Figure 7: Images of a pinhole and a USAF test formed by the model eye with two different IOLs. Left: SN60WF IOL with spherical design. Right: SN60AT IOL with aspheric design. Contrast measured in the G2T1 group of the USAF test is also indicated.](image)

**Figure 7: Images of a pinhole and a USAF test formed by the model eye with two different IOLs. Left: SN60WF IOL with spherical design. Right: SN60AT IOL with aspheric design. Contrast measured in the G2T1 group of the USAF test is also indicated.**

![Figure 8: MTFs of two IOLs computed from the images in figure 7.](image)

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3. Preliminary results
Although an aberration free cornea, such as the LAO one, is recommended by the ISO standard, it is not suitable for testing IOLs with aspheric design [8], which are intended to reduce or compensate for the natural SA of the human cornea. Since our test bench is mainly conceived to simulate the conditions of real human eyes, we will present results obtained after testing the IOLs with the DCX cornea. Figures 7 and 8 allow us to compare the optical performance of two different IOLs of the same dioptric power (20D). One of the IOLs (SN60AT) is equiconvex with spherical surfaces, while the other (SN60WF) is an equiconvex IOL with an aspheric design to partially compensate for the SA of a human cornea. The measured focal length of the model eye was 29.69 mm with the SN60AT and 27.91 mm with the SN60WF respectively. From these images the improvement that represents the aspheric design versus the spherical one in terms of image contrast and resolution is clear. Further support to this conclusion is provided by Fig. 8, where it is shown that the MTF obtained with the aspheric IOL is far better than with the spherical IOL.

The vast majority of commercial IOLs are transparent, but there is a brand that manufactures yellow-tinted IOLs. It has been claimed that the UV radiation blocking effect of these yellow-tinted IOLs may help to prevent age related retinal pathologies. Figure 9 shows the spectral transmittance of three similar yellow-tinted IOLs that only differ in their optical power. The results show that there is a reduction of the transmittance (in the 400 nm to 500 nm wavelength range) as the optical power of the IOL increases, more likely due to the fact that the thickness of the IOL is higher as their optical power increases. The spectral transmittance of a non-tinted IOL is included in Fig. 9 for the sake of comparison.

4. Conclusions
We have designed an optical bench for IOL testing in conditions close to a real human eye. One remarkable feature is the use of an artificial cornea with values of SA aberration similar to that of human cornea. The analysis of the IOLs can be done through different metrics, such as PSF, MTF, SR or contrast measurements, which are conventional in imaging quality testing and wavefront measurements. Results obtained with an aspheric IOL show a significant improvement in terms of image contrast and resolution when it is compared with an IOL with similar power but of spherical design.
5. Acknowledgements
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6. References
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