Combining in vitro test methods for measuring light scatter in intraocular lenses

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Abstract: Intraocular lenses (IOLs) are designed for implantation for vision correction following cataract removal. The IOL typically replaces a cataractous natural lens that exhibits very high levels of light scattering. The amount of scattering is significantly reduced with an IOL, though it is rarely quantified and both the surface and the bulk of the intraocular lens may contribute to light scatter at some level, and in some cases potentially affecting patients’ post-operative quality of vision. The purpose of this paper is to describe two complementary in-vitro quantitative methods for measuring light scatter caused by IOLs. The first method directly measures light scatter from the lens in one plane for angles larger than two degrees. The second method measures light scatter in an eye model including the focal point out to three degrees in the image plane. The measured amount of light scatter from an IOL is typically lower than that found in healthy donor crystalline lenses of various ages that are used as a basis for comparison.

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1. Introduction
Knowledge of the total intensity distribution over a large angular range on the retina may predict aspects of quality of vision. The normal point spread function (PSF) of the human eye has a range spanning approximately ten orders of light intensity distributed over 180 degrees [1]. Whereas the human eye is capable of perceiving this dynamic range in light intensity and...
retinal eccentricity, currently neither single in-vivo nor single in-vitro instruments have been able to measure a comparable PSF in its entirety. This paper describes two complementary in-vitro methods to measure the angular light distribution caused by IOL light scatter.

Light scatter can be described as the deflection of light rays in random directions by irregularities in the propagation medium. Inelastic light scatter phenomena such as Raman and Brillouin light scatter are not considered here. Both, surface and bulk of the IOL may contribute to intraocular light scatter. Bulk in-homogeneities such as lens material density and/or composition fluctuations, voids, inclusions and the presence of micro vacuoles inside the lens body will deflect light rays to the retinal periphery causing them not to be refracted to the fovea. This effect could also occur when machine lines, forceps imprints, or other surface defects are present. Another source of light ray deflection is the lens design itself. Lens geometries such as diffractive patterns may cause light rays intended for foveal image formation to propagate to the retinal periphery causing glare.

In the beginning of last century Cobb [2] introduced a method to determine retinal stray light. Stray light is defined by dividing equivalent luminance by the illuminance of the glare source falling into the eye. The equivalent luminance is defined as the luminance that has the same visual effect as the glare source [1]. In vivo, a wide angular range of stray light can be determined psychophysically by assessing this equivalent luminance. Stiles [3] and Holladay [4] found that for phakic eyes this ratio approximates 10/θ^2 for angles θ of 1 to 30 degrees. With this definition stray light corresponds to the skirts of the PSF properly normalized. A stray light parameter, s, has been introduced that is defined according to Eq. (1) [5], where a and p are age and pigmentation respectively.

\[ s(θ,a,p) = \text{PSF}(θ,a,p) \times θ^2 [\text{degrees}^2 / \text{sr}] \]  \hspace{1cm} (1)

For healthy eyes of various ages and color (pigment), s is relatively constant over a large range of the angular domain. Norm values for s are presented in Vos et al [1]. The crystalline lens of young eyes contributes approximately one third to the stray light of the total eye and this is formulated in Eq. (2). The norm function for a 70yr old crystalline lens is calculated following Eq. (3).

\[ s_{\text{lens}}(0,20,1) = \frac{1}{3} \times \text{PSF}(0,20,1) \times θ^2. \]  \hspace{1cm} (2)

\[ s_{\text{lens}}(0,70,1) = 0.3 \times (\text{PSF}(0,70,1)) \times θ^2. \]  \hspace{1cm} (3)

The chosen value of p = 1 corresponds to average Caucasian eyes. In this paper these calculated s values are used as references for comparison with stray light levels found in IOLs.

2. Method 1

A method for measuring light scatter in human donor crystalline lenses has been described in depth by van den Berg [6]. The same setup with minor modifications can be used to determine the same for IOLs (Fig. 1). A wavelength controlled light beam is incident on an IOL placed in a 10mm wide fluid filled optical quality glass cell. The beam shape is defined by the field stop.

![Diagram](image-url)  

Fig. 1. Schematic of in vitro setup method 1.
The PSF is determined by measuring the light power $P(\theta)$ at stray light angle $\theta$ divided by the product of receptance solid angle $\Omega$ and total power $P_0$ transmitted by the IOL as formulated in Eq. (4).

$$PSF(\theta) = \frac{P(\theta)}{(\Omega \times P_0)}$$ (4)

For an aperture diameter of 3mm in front of the camera lens and a distance of 140mm from camera aperture to IOL, $\Omega = 3.61 \times 10^{-4}$ sr and the angular width is 1.2 degrees. The IOL is centered by using a laser pointer in the center of the camera wheel. The cooled scientific grade 16-bit CCD camera and the glass cell are placed in a dark room separate from the light source with filters. The camera shutter times increased from 100ms to 10s with increasing measurement angles in order to be able to detect the decreasing amount of light power. The light power $P_0$ at zero degrees is measured by inserting a density filter and enlarging the camera aperture to 22 mm resulting in $\Omega = 0.02$ sr. It is possible to determine forward as well as backward scatter for an angular range of 135 degrees for pencil illumination as well as slit illumination. The finite width of the aperture and intensity of the transmitted light beam is the limiting factor for the smallest angle to be measured and is approximately two degrees but is dependent on the IOL tested and light wavelength. The background stray light level of the fluid filled cell without IOL had to be below the level of 0.1 degrees$^2$/sr, otherwise the glass cell was rejected or cleaned and refilled with fresh immersion fluid.

Fig. 2. Typical examples of acrylic IOLs’ scatter in purified water at 30 degrees forward direction. (a) IOL with a center thickness of 0.7mm. (b) IOL with a center thickness of 0.5mm containing micro vacuoles.

The light scatter results (Fig. 2) were made with an 80µm slit illumination at 30 degrees in broadband green light and purified water as immersion fluid. In such recording bulk scatter can be seen separated from surface scatter. An interface was present between IOL and purified water resulting in a stray light level of approximately 5 degrees$^2$/sr. The picture on the right shows an example where in-homogeneities in lens material resulted in additional bulk light scatter. A saline solution reduces interface scattering significantly (below 1 degrees$^2$/sr) and is recommended as immersion fluid in testing IOLs in order to better mimic the in vivo condition.

Fig. 3. Typical straylight level for a monofocal IOL measured in saline using method 1.
The stray light level (Fig. 3) for a monofocal IOL is compared to a 20 year and 70 year old healthy crystalline donor lens according to Eqs. (2) and 3. The stray light level is lower than that of a healthy human crystalline lens, irrespective its age.

3. Method 2

In method 1, because the location of the IOL is very close to the nodal point of the eye, a cornea was deemed unnecessary. In this set-up the scatter induced angular intensity distribution remains similar for an IOL in a liquid cell with or without a cornea in front. Subsequently, vergence and refraction will not affect the measured intensity distribution. In method 2, the measurements take place for angles close to on-axis or paraxial angles. An artificial cornea in front of the cell is introduced to emulate the refraction contribution to the image intensity distribution of the PSF. The setup (Fig. 4) is identical to the eye model prescribed in the ISO guidelines for IOL quality testing [7].

![Fig. 4. Schematic setup for method 2.](image)

The artificial cornea, an equi-biconvex BK7 lens, provides the same power as the ISO cornea, and it reproduces the average corneal spherical aberration with Zernicke coefficient $c(4,0) = 0.27\mu$m for 6mm cornea aperture and the average longitudinal chromatic aberration of 0.4mm between hydrogen F and C lines [8].

The configuration of method 2 has good infinite object testing capability [9] and is used for image quality assessment of IOLs by determining the modulation transfer function MTF from the measured line spread function LSF. The stray light aspect can be approached from the tails of the LSF. The basic setup has a limited angular range of $\pm 0.8$ degrees and is extended by stepwise lateral displacement of the CCD camera by a total of approximately 0.5mm with for each step a new shutter time ranging from 0.1ms to 4s. An 8 bits or cooled scientific grade 12 bits CCD camera is used. The light intensities to be measured for method 2 are higher than for method 1 and therefore a less stringent dark condition is necessary. The glass cell and the light path to the CCD camera are shielded from the very dim environment light. The recorded intensity distributions are stitched together. The distance dp from pixel location to peak intensity is used to calculate the angle $\theta$ at the pixel location with $ATAN(dp/M*ND)$ where ND = 21mm is the nodal distance for a medium powered IOL. M is the calibrated magnification (9.1X, NA0.45) of the used APO microscope objective lens. The angular intensity is resolved to 0.0019 degrees for a CCD pixel size of 6.45µm.

Figure 5 shows an example where an LSF of a monofocal lens was stitched together in five steps leading to an intensity range of six decades and an angular range of $\pm 2.5$ degrees. Since the LSF is the integral of the PSF along the imaged slit the reliable angle from which the stray light could be determined is governed by the object angle of the slit target length and collimator focal length. This is demonstrated and indicated in Fig. 5 by the 0.43 degrees of angular width at the location of the shoulder points corresponding to the ratio of 3mm slit length and 400mm collimator length. In this case, accurate stray light results could only be obtained for radial angles much larger than 0.215 degrees. When a pinhole target is placed in the focal point of the collimating lens a larger part of the central PSF can be recorded. The maximum angle measured could extend to three degrees dependent on test target, wavelength and IOL tested. For angles larger than 0.1 degrees the background s value of cornea and fluid filled cell without IOL had to be 0.5 log unit below the level of a healthy 20yr old crystalline lens according to Eq. (2).
For rotationally symmetric systems it is adequate to record the radial PSF. The PSF is again determined according Eq. (4). The solid angle is determined by the imaging optics and the magnifying objective lens projecting the image onto the CCD. Each pixel subtends the square of the pixel size $ps$ and $\Omega$ is calculated as $ps^2/(M*ND)^2$. For a pinhole target the transmitted power is the volume under the stitched PSF curve. For the case of a slit target the transmitted power is the area under the stitched LSF curve multiplied by the length of the imaged slit in number of pixels.

4. Discussion

Two quantitative in-vitro methods are described and demonstrated to be complementary in the angular domain. The combined methods have the capability to record ten decades of light intensity variation from focal image to forward scatter position of 30 degrees, with the possibility of extending to 135 degrees. There is a blind angular range of approximately 25 degrees centered around 90 degrees. Edge glare effects could be considered as a separate source of light scatter [10]. For angles larger than 0.5 degrees, a monofocal IOL essentially free of bulk in-homogeneities has lower stray light levels than that of a healthy human crystalline lens, irrespective of its age (Fig. 6). Positive effects of cataract surgery on stray light levels are discussed in literature. In vivo straylight measurements with a glare source at seven degrees were done before and after surgery [11,12].

In vitro light scatter measurements in intraocular lenses are useful because they reveal stray light levels separately from other in vivo contributing stray light factors such as that from the cornea, vitreous or retina. Stray light may not necessarily impact visual acuity but it could seriously hinder good quality vision. Forward light scatter will be incident on the retina and therefore reduces patients' quality of vision. Forward scatter and backward scatter can be...
determined in vitro in one setup and both scatter directions can be compared. If there is a difference in forward and backward scatter levels, this may then serve as an explanation for differences found in clinical observations which are dominated by backward scatter versus patients’ perceptions which depend on forward scatter.

The main purpose of this paper is to describe an overview of laboratory methods for objectively measuring scatter from IOLs. Measurement results are given to illustrate the method, and not to evaluate different IOLs in detail. Methods defined in this paper may be used in the future to examine the clinical phenomena of scatter in several monofocal and diffractive multifocal IOLs and to determine whether scatter introduced by these lens designs may contribute significantly to disability glare.

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