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Optical Properties of Diffractive, Bifocal Intraocular Lenses

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ABSTRACT

The resolution of diffractive, bifocal intraocular lenses was studied with regard to pupil displacement and diameter size through computer simulations, bench measurements and patient vision acuity measurements. Good agreement was obtained between these three methods of investigation. In particular we find that pupil displacements of the order of 1 mm reduce the resolution considerably for these lenses.

1. INTRODUCTION

Diffractive(D), bifocal(B) intraocular lenses(IOLs) are implanted in cataract patients after removal of their diseased lenses. The advantage of BfIOLs relative to monofocal IOLs, the latter hereafter referred to as IOLs, is that they allow the patient to simultaneously have a relatively good distant as well as near vision acuity. A disadvantage is that they produce a blurred retinal image that superimposes the sharp one, which obviously reduces the image contrast. It could also be argued that the blurred background image reduces the resolution to some extent. In this work we try to estimate the resolution by weighing together the sharp and the blurred images thereby obtaining an 'effective resolution'. Further one might fear that DBfIOLs suffer from large chromatic aberrations and that their optical properties are sensitive to such parameters as pupil size and lens displacement. Note, however, that due to their designs a diffractive bifocal lens is generally less sensitive to pupil size and displacement than a two-zone refractive bifocal lens [1]. The chromatic aberration of the DBfIOL studied in this work (3M) is supposedly quite small, because the lenses are designed so that the dispersion of the lens material (polymethylmetacrylate, PMMA) partly balances the dispersion of the diffractive structure of the lens [1]. How pupil size and lens displacement affect the optical properties of these lenses has not yet been reported to our knowledge and is the topic of this study. An investigation of the optical properties of several commercial multifocal IOLs with centred, fixed diameter pupil is reported in ref. [2].

Our methods of investigating the pupil size and lens displacement sensitivity of the DBfIOLs are threefold:

1. Computer calculations of the retinal Point Spread Function (PSF) of diffraction limited DBfIOLs at a single, optimally chosen wavelength using Fresnel diffraction theory.
2. Bench measurements of the 3M DBfIOL.
3. Vision acuity measurements in patients wearing DBfIOLs (3M), the measurements being carried out 12 to 40 weeks after uncomplicated lens implantation.

2. COMPUTER SIMULATIONS

The studied DBfIOLs are manufactured so that a blazed phase zone plate structure extends all over one of the lense's two surfaces, the stepheight between neighbouring blazed zones being one-half the ideal height for 100% diffraction efficiency [1]. This latter stepheight is $h_0 = \lambda_0 / (n_2 - n_1)$, where λ_0 is the design vacuum wavelength, n_2 is the index of refraction of the lens material and n_1 is the index of refraction of the surrounding medium. With the stepheight chosen to $h_0/2$ ideally 40.5% of the light power gets diffracted into the +1 order by the blazed phase zone plate structure of the DBfIOL, provided that the light is monochromatic at wavelength λ_0 . The same amount of light power is transmitted undiffracted (zero diffraction order). Both diffraction orders are refracted by the lense's curved surfaces so that the zero and +1 diffraction order beams form

foci for distant and near vision, respectively. The remaining 19.0% of the total transmitted light power distributes itself among the other diffractive orders [1]. It is worth noting that the power fractions falling into the +1 and zero diffraction orders only balance exactly for the design wavelength λ_0 . For the studied DBIOL λ_0 is chosen to 555 nm, which is in the centre of the visible and where the light sensitivity of the human eye is highest. With that choice of design wavelength there is only a moderate imbalance between the +1 and zero diffraction orders within the visible. In the simulations we avoid all chromatic effects by always using monochromatic light at λ_0 .

As mentioned PSFs are calculated in the simulations. In so doing the phase distribution of a diffracting plane is built up in two steps: First we generate the phase of a spherical wavefront, which one would have in a plane immediately behind a diffraction limited lens, after a plane, monochromatic wave had passed through the lens. That simulates the action of the refractive part of a diffraction limited DBIOL, its focusing power being 17 diopters in water. Next we add the phase that would result after the wave had passed through a 3.5 diopters (water) blazed phase zone plate having the stepheight $\lambda_0/(2 \cdot (n_2 - n_1))$, where $\lambda_0 = 555$ nm, $n_2 = 1.49$ (PMMA) and $n_1 = 1.33$ (water). To simulate the eye pupil the field amplitude is set constant within a circular area, whose diameter, $\emptyset$, and displacement, Δ , relative to the optic axis can be chosen freely, and zero outside the circular area. We obtain the PSF in any desired plane in the neighbourhood of the two focal planes by numerically calculating the diffracted field with Fresnel diffraction theory utilizing a Fast Fourier Transform algorithm. The matrix size is 512x512 and the pixel width in the diffracting plane 12.5 μm . (The f-numbers in the simulations are larger than 20, which means that Fresnel diffraction theory is accurate.) The PSF intensity is conveniently integrated in one direction so that one obtains what can be called the line spread function (LSF) intensity and which we believe relates more directly to seeing than does the PSF intensity.

Figure 1. Simulation of images from 17+3.5 diopters diffraction limited diffractive bifocal intraocular lens (DBIOL). Object: monochromatic line source at infinity. Pupil diameter $\emptyset = 1.5$ mm.
a) Near focus. Standard deviation of intensity, $\sigma_1 = 2.55$ mrad. Standard deviation of intensity squared, $\sigma_2 = 0.43$ mrad.
b) Half-way between foci. $\sigma_1 = 2.47$ mrad, $\sigma_2 = 0.55$ mrad.
c) Distant focus. $\sigma_1 = 2.55$ mrad, $\sigma_2 = 0.43$ mrad.

Figure 2. Simulation of images from 17+3.5 diopters diffraction limited DBIOL. Object: monochromatic line source at infinity. Pupil diameter $\emptyset = 3.0$ mm.
a) Near focus. $\sigma_1 = 3.51$ mrad, $\sigma_2 = 0.51$ mrad.
b) Half-way between foci. $\sigma_1 = 3.29$ mrad, $\sigma_2 = 1.24$ mrad.
c) Distant focus. $\sigma_1 = 3.51$ mrad, $\sigma_2 = 0.51$ mrad.
For comparison a diffraction limited monofocal IOL yields $\sigma_1 = 0.45$ mrad, $\sigma_2 = 0.043$ mrad.

Figures 1 and 2 show LSF intensities in the two focal planes and half-way between them, calculated with centred pupils, the pupil diameter being 1.5 and 3.0 mm, respectively. As expected Figs. 1a, 1c, 2a and 2c each display a sharp diffraction peak superimposed on a pedestal, whose width is determined by the pupil diameter. The intensity intermediate between the foci (Figs. 1b and 2b) is due to the interference between the unfocused diffractive and refractive beams. Note in Fig.1 that the diffraction peak and the pedestal more or less merge, together forming a single peak, so that in this case the effective resolution should definitely be determined by the *whole* LSF intensity profile and not only by the sharp diffraction peak. Analogously we assume in this work that the full LSF intensity profile always determines the effective resolution. By merely inspecting Figs.1 and 2 one is prone to say that with 1.5 mm pupil diameter the effective resolution is almost as good half-way between the foci as it is in the focal planes, whereas in the 3.0 mm pupil diameter case the effective resolution appears to be much better in the focal planes than it is between them. However, the calculated standard deviation of the intensity profiles, σ_1 , of either Fig. 1 or 2 is essentially determined by the pupil diameter only. Specifically, with 3.0 mm pupil diameter the calculated width σ_1 displays no distinct foci. This is in conflict with intuition and also in disagreement with what patients wearing DBIOLs experience (c.f. IV. Vision acuity measurements, Fig. 8). Therefore, to pronounce the importance of a sharp intensity peak in the focal planes for seeing, we square the intensity before calculating the LSF intensity width. This new line width measure is denoted σ_2 and is defined as the standard deviation of the squared LSF intensity profile. As is shown in Section IV this manipulation yields quite good agreement between width calculations and vision acuity observations in patients. σ_2 is our measure of the effective resolution used in this paper.

The effect of a 1.0 mm pupil decentration on the LSF intensity is demonstrated in Fig. 3 for 1.5 pupil diameter, the PSF intensity being integrated perpendicularly to the decentration.

Figure 3. Simulation of images from 17+3.5 diopters diffraction limited DBIOL. Object: monochromatic line source at infinity. Pupil diameter $\phi=1.5$ mm. Pupil displacement $\Delta=1.0$ mm.
a) Near focus. $\sigma_1=4.20$ mrad, $\sigma_2=1.24$ mrad.
b) Half-way between foci. $\sigma_1=4.12$ mrad, $\sigma_2=1.97$ mrad.
c) Distant focus. $\sigma_1=4.19$ mrad, $\sigma_2=1.24$ mrad.

Figure 4. Calculated image line width σ_2 for a 17 + 3.5 diopters diffraction limited DBIOL vs. pupil displacement Δ . Object: monochromatic line source at infinity.

As can be seen in this figure the pupil decentration displaces the refractive and diffractive beams relative to one another, which increases σ_2 (and σ_1). This width increase is demonstrated in Fig.4, which shows σ_2 calculated at various pupil displacements.

For ease of interpreting Fig.4 this figure only shows data for the near focus, since the distant and near foci are almost identical. From Figs.1-4 we find that;

- The effective resolution of the DBIOL is always worse than the resolution of a diffraction limited IOL;
- with centred pupils the effective resolution is slightly better at 1.5 mm pupil diameter than it is at 3.0 mm;
- with centred pupils the effective resolution remains essentially constant between the two foci in the 1.5 mm pupil diameter case contrary to the 3.0 mm pupil diameter case;
- The effective resolution in the focal planes remains almost unaffected by pupil displacements up to 0.5 mm, but a 1.0 mm pupil displacement lowers the effective resolution considerably in the focal planes and more severely so at 1.5 mm than at 3.0 mm pupil diameter.

3. OPTICAL BENCH MEASUREMENTS

In this section we report on optical bench measurements of a DBIOL (3M), specified to have a refractive strength of 17 diopters (water) and with 3.5 diopters added diffractively (water). To imitate normal lens using conditions white light was utilized. So, although the chromatic aberrations of the lens are stated to be small the measured lens resolution is still expected to be worse than the calculated one (remembering that the simulations were carried out with an ideal DBIOL at the nominal wavelength).

In the measurements a 100 μm wide, diffusely illuminated, vertical slit was imaged by the DBIOL, which was mounted in a water filled quvette. A Tungsten lamp (3000 K), equipped with an IR blocking filter, served as light source. The distance between the slit and the lens was 105 mm, the slit being positioned on the optical axis of the lens. A 2.5 mm diameter circular aperture, which could be moved sideways, was positioned immediately in front of the 10 mm wide quvette. The intensity of the slit image was measured by horizontally scanning a vertical, narrow slit in 50 μm steps across the beam, the measuring slit mounted in front of a silicon photodiode (slit dimensions: 50 μm by 2.0 mm). The photodiode current data were stored in a desktop computer for analysis. The experimental setup allowed beam cross sections to be measured at different locations along the beam in the neighbourhood of the two foci, positioned ~ 60 mm and ~ 80 mm behind the lens. The two slits of the measurement setup can be imagined to be replaced by an effective source slit, whose width can be estimated at ~ 124 μm . This is equivalent to ~ 4.4 times the Rayleigh angular width, which is 0.204 milliradians. (The presence of the pedestal in the LSF makes it less important that the measurement is diffraction limited [4].)

For comparison computer simulations of these measurements were also conducted, in which we took account of the presence of the two, finite-width slits in the measuring setup by convolving the LSF intensity with a rectangle function twice in sequence. In the convolution the light source slit width corresponds to 11 focal plane pixel widths (each 0.0652 milliradians wide) and the measuring slit width to 10 and 7 focal plane pixel widths in the near and distant focal planes, respectively. Figure 5 summarizes the measurements, showing the line width σ_2 at a number of locations in the neighbourhood of the two foci at 0.0, 0.5 and 1.0 mm pupil displacement. The corresponding simulated values are shown in Fig.6. As can be seen in these figures there is good agreement between the measured and the simulated values in the centred pupil case. However, with increasing pupil displacement there is an increasing discrepancy between the measurements and the simulations, which not unexpectedly indicates that the lens is aberrating. More precisely, in terms of the σ_2 measure the DBIOL gradually loses its bifocality with increasing pupil displacement, in contrast with the simulations. It is worth noting that the effective resolution of the refractive (distant) focus of the DBIOL is actually better than the corresponding simulated effective resolution, whereas the diffractive (near) focus of the DBIOL is considerably worse than the simulated one [5]. Although the σ_2 measure seems to indicate that the diffractive focus more or less disappears at 1.0 mm pupil displacement (c.f. Fig.5), this is in fact not the case. What actually happens is that the focused diffractive peak gets substantially displaced relative to the refractive peak and that a double peak is formed. This can readily be observed in Fig.7, which shows the measured LSF intensity for the diffractive focus at 1.0 mm pupil displacement.

Figure 5. Measured image line width σ_2 for a 17 + 3.5 diopters DBIOL(3M) vs. position along optic axis. Object: Diffuse white light slit. Effective slit width ~ 0.9 mrad. Pupil diameter $\Phi = 2.5$ mm. The line profile a is shown in Fig. 7.

Figure 6. Calculated image line width σ_2 for a 17 + 3.5 diopters diffraction limited DBIOL vs. position along optic axis. Object: Diffuse monochromatic slit. Effective slit width ~ 0.9 mrad. Pupil $\Phi = 2.5$ mm. Dotted line shows behaviour of a diffraction limited monofocal lens.

Figure 7. Measured image line for a 17 + 3.5 diopters DBIOL(3M). Object: Diffuse white light slit. Effective slit width ~ 0.9 mrad. Pupil $\Phi = 2.5$ mm, position: 20.0 diopters; $\Delta = 1.0$ mm.

4. VISION ACUITY MEASUREMENTS

Eight cataract patients having DBIOLs (3M) and without other obvious ocular diseases were tested for vision acuity. The test was carried out 12 to 40 weeks after uncomplicated cataract extraction and lens implantation. All the subjects had a letter acuity of between 20/30 and 20/15 with a mean value 20/23 [6] which is slightly below normal acuity in age matched groups and in good coherence with other similar studies [7,8]. Letter acuity tests are not ideally suited for resolution measurements, however [9]. (It seems that many times a letter can be correctly identified before quite resolved. Furthermore, in repeated tests the subjects tend to memorize letter sequences on the test chart.) Instead, for our study we chose the Landolt C test - the standard vision resolution test -, in which the subject is asked to identify the position of an opening in a circular ring presented to him/her on a CRT screen[10]. The patients were tested with each of a number of different glasses mounted in front of their eyes. The glasses had strengths from +3.5 diopters to -5.5 diopters in steps of 0.5 diopters, which covers the two foci of the DBIOL. Before starting a measurement the patients were given glasses for best spherical and astigmatic correction for distant vision. Through the test we obtained a measure of the minimal angular resolution (MAR) as a function of added lens power. An example of such a measurement for one of the patients, who had a centred 3.0 mm pupil diameter, is presented in Fig.8. The normal limits of MAR for the age group of the tested patients are between 0.21 and 0.34 mrad [11]. The presence of the two foci can clearly be seen in figure 8 and could also be observed in the data from all the other patients. The average MAR values among the patients in the near focus, half-way between the foci and in the distant focus were 0.57, 1.15 and 0.53 milliradians, respectively. Hence, on the average the resolution half-way between the foci is approximately half of what it is in the focal planes. This is in good agreement with the corresponding σ_2 -values shown in Figs. 2,4 and 5. The comparison is meaningful to make since the average pupil displacement among the patients was ~ 0.6 mm and the average pupil diameter was ~ 3.0 mm. How a small pupil diameter affects the resolution was tested in the following way. The same patient, whose MAR vs. lens power diagram at $\phi=3.0$ mm is shown in Fig.8, was given Pilocarpin eye drops, which made her pupils contract to $\phi=1.5$ mm. Her MAR vs. lens power diagram measured with this smaller pupil diameter is also shown in Fig. 8. Since her pupil was well centred comparison can readily be made with the computer simulation data presented in Fig. 1. Indeed, both patient data and simulation data show that the resolution remains essentially constant between the foci at the small pupil diameter 1.5 mm.

Figure 8. Measured minimal angular resolution for a DBIOL patient vs. added lens power.
Method of measurement: Landolt C test. Centred pupil with diameters: 3.0 mm(x) and 1.5 mm(o). Normal limits of MAR for people in the same age group as the tested patients are 0.21 - 0.34 mrad.

Figure 9. Measured minimal angular resolution for a DBIOL patient vs. added lens power.
Method of measurement: Landolt C test. Pupil diameters: 3 mm(x) and 1.5 mm(o). Aperture decentration: 0.9 mm.
Normal limits of MAR for people in the same age group as the tested patients are 0.21 - 0.34 mrad.

In sections II and III pupil decentration of the order of 1 mm were found to significantly reduce the effective resolution, particularly at small pupil diameters. A DBIOL patient, who had a measured aperture decentration of 0.9 mm, went through the Landolt C test described above with normal and contracted pupil. The result is shown in Fig. 9, which supports the findings of the previous sections. It should be mentioned that this patient is the only one tested so far who had a significantly decentred aperture.

5. DISCUSSION

In this work LSF intensities of DBIOLs at various pupil diameters $\emptyset$ and displacements Δ were calculated and measured (Sections II and III, respectively). Visual acuity measurements of DBIOL patients as a function of $\emptyset$ and Δ were also performed (Section IV). In trying to interrelate these results the central and non-trivial question arises: how should LSF intensities be evaluated to yield a good quantitative measure of resolution? Attempting to find an answer we assumed that the effective resolution of vision is not only determined by the sharpest feature of the LSF intensity, which is the diffraction peak of the focused beam, but at least to some extent depends on the unfocused background as well. However, the standard deviation width of the LSF intensity, which one might at first expect would be proportional to the resolution, turned out not to (see figure legends of Figs. 1 and 2). To get agreement with resolution data on patients on trial we squared the LSF intensity before calculating the standard deviation. This manipulation puts heavier weight on the high intensity parts of the LSF than on the low intensity parts. Defining our estimated effective resolution as the standard deviation of the squared LSF intensity we get good agreement with the MAR values found in patients. The observed agreement indicates that the human brain performs a non-linear weighing of the intensity in the process of seeing and that our assumptions are reasonable. (The complex matter of signal processing of vision is discussed in ref. [12].) We like to point out that intensity squaring should not be considered in resolution estimates when there is no unfocused background, since that would yield resolution estimates better than given by the Rayleigh criterion. (The above statement is not in contradiction with Fig. 6 where the σ_2 - measure was used for the monofocal lens. The purpose of that simulation was not to estimate the resolution of the monofocal lens; the slit was too wide for that.)

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