Title: About measuring aberrations of multifocal and extended-depth-of-focus intraocular lenses

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Studies comparing the optical quality of intraocular lenses (IOLs) *in vivo* are proliferating. This increase in popularity is largely driven by the availability of aberrometers within both the clinical and laboratory environment. However, the approach taken with interpreting the aberrometry results in pseudophakic patients has been imprudent in many cases, especially when considering multifocal intraocular lenses (MIOLs) or extended-depth-of-focus (EDOF) intraocular lenses. In the majority of published studies that compare optical quality between multiple IOLs and pupil sizes, researchers present the root mean square (RMS) as a *fait accompli*. The interpretation of results are often limited to “lens A higher-order aberrations (HOAs) RMS is greater than lens B HOAs RMS, thus optical quality of lens A is better than that of lens B”: When the lenses under study are MIOLs or EDOF IOLs, this conclusion must be regarded with caution mainly due to three reasons:

1. RMS is not the best metric to measure the optical quality of eyes[1], or eyes implanted with IOLs. For instance, an optical system having 0.4 µm of third-order Zernike coma will have the same RMS as a different system comprising 0.4 µm of fifth-order Zernike coma. However, the image formed by the second system will have lower contrast, which will influence the patient’s visual performance[2].
2. As shown by Schwiegerling et al.[3], measuring aberrations of diffractive MIOLs can lead to errors, since the rays passing through the edges of the diffractive steps will experience large deviations, thus rendering the measurement unreliable[4]. This could be partially resolved by, for example, using image processing techniques; nevertheless the majority of research about this topic do not consider that, since commercially available clinical devices tend to have restricted options.
3. Even if the wavefront error was perfectly measured, researchers tend to use Zernike polynomials for describing aberrations. These polynomials are very useful and powerful, but have limitations. Describing the wavefront error maps of MIOLs and EDOF IOLs by using a relatively small number of Zernike polynomials does not yield accurate results. Moreover, when using directly the coefficients given by clinical aberrometers, the error

committed is usually significant. To illustrate this, two refractive MIOLs designs with their corresponding wavefront maps are shown in Figures 1 and 2. Even when the wavefront is measured perfectly, if only polynomials up to 8th order are used, the result obtained is very different than the original wavefront.

The calculated point spread functions (PSF) and the image of an optotype are quite different whether the original wavefront is used or a wavefront obtained by fitting 45 Zernike polynomials, which is the usual number of coefficients given by clinical aberrometers.

In conclusion, it is insufficient to rely on the RMS values, calculated by clinical devices, when evaluating the optical quality of MIOLs or EDOF IOLs. This is demonstrated in Figures 1 and 2, where one would assume that the optical quality of these lenses is poor when considering Zernike polynomials up to 8th order. However, in this case it is the polynomial interpretation of the wavefront that is insufficient to describe the actual optical quality of the lens. These discrepancies are most apparent in the cases of diffractive MIOLs, due to discontinuities in their wavefront and in MIOLs with no rotational symmetry. Despite the numerous advantages of Zernike polynomials when describing ocular aberrations, awareness about limitations when not using a sufficient number of polynomials is essential. Using the whole wavefront error map as shown in Figures 1 and 2, avoids potential errors and misunderstanding.

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FIGURE LEGENDS

Figure 1. Top left panel shows the wavefront map of an intraocular lens design based on the Precizon Presbyopic NVA (Ophtec Inc., Netherlands) for a 6-mm pupil. Top right panel shows the wavefront of the same lens, but when described by Zernike polynomials up to 8 radial orders. Mid row shows the point spread functions (PSF) obtained with each wavefront. Yellow line has a size of 5 arcmin. Note that in order to notice the differences better, the cube root of the PSF has been plotted. Bottom row shows simulated images (as perceived by a subject) of an optotype obtained with each PSF. The DVOHC line corresponds to 0 logMAR visual acuity.

Figure 2. Same as Figure 1, but for an intraocular lens design based on the Lentis M-Plus (Oculentis GmbH, Berlin, Germany).

Figure 1
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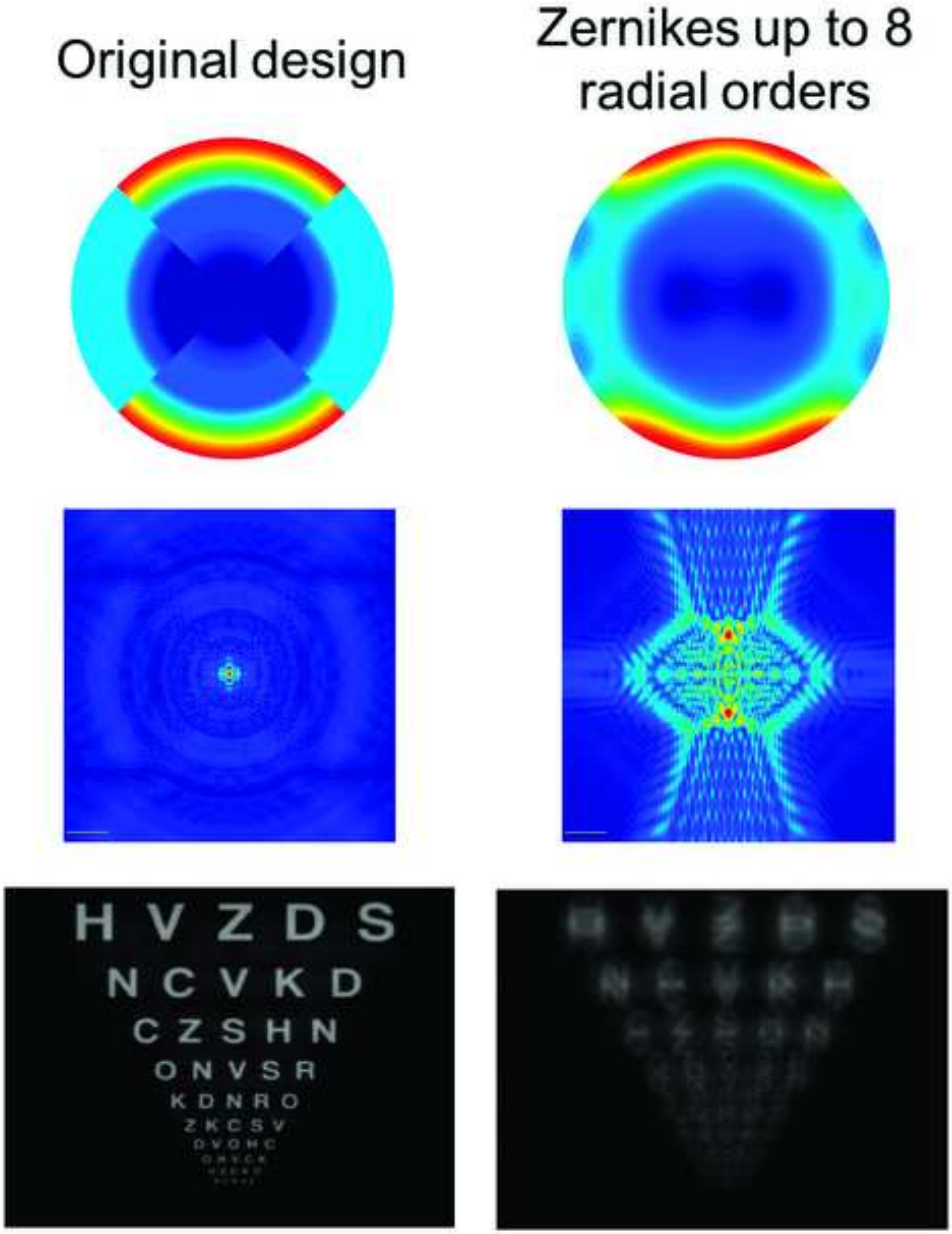

Figure 2
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