Analysis of Wavefront Data Obtained With a Pyramidal Sensor in Pseudophakic Eyes Implanted With Diffractive Intraocular Lenses

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ABSTRACT

PURPOSE: To investigate the clinical validity of using wavefront measurements obtained with a recently available pyramidal aberrometer to assess the optical quality of eyes implanted with diffractive intraocular lenses (IOLs).

METHODS: Individual biometric data were used to create models of pseudophakic eyes implanted with two diffractive IOLs. Their synthetic wavefronts were calculated by ray-tracing with near infrared wavelength (0.85 µm). Comparisons of the through-focus visual acuity of 12 pseudophakic eyes were obtained with three different methods: clinical defocus curves; simulated defocus curves calculated from ray-tracing in the customized model eyes; and through-focus simulated defocus curves calculated from the wavefront data measured with a pyramidal aberrometer.

RESULTS: Image quality calculated from wavefront data obtained by ray-tracing with 0.85 µm wavelength, without scaling the phase to 0.55 µm, resulted in a significantly different through-focus curve compared to the reference values. Even so, after scaling of the wavefront data to 0.55 µm, the defocus curves calculated from the wavefronts measured with the pyramidal aberrometer did not match the shape and the depth of field of the clinical defocus curves or the theoretical expected values.

CONCLUSIONS: Correcting for the longitudinal chromatic aberration of the eye when measuring the wavefront of eyes implanted with diffractive IOLs under near infrared light only accounts for the best focus shift due to the longitudinal chromatic aberration, but not for the wavelength dependence of the diffractive element. The pyramidal sensor does not seem to properly sample the slopes of a wavefront measured from a pseudophakic eye implanted with a presbyopia-correcting diffractive IOL to a clinically acceptable level.

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A variety of techniques are available to clinicians that allow them to obtain a comprehensive assessment of the eye's image-forming properties, including Hartmann-Shack sensors, pyramidal sensors, laser ray-tracing, curvature sensors, interferometric techniques, and double-pass systems. Most of these devices are wavefront sensors that operate on the same principle, which is an indirect measurement of local wavefront slopes to obtain a reconstruction of the complete wavefront by integration of these slopes.[1,2]

Techniques based on Hartmann-Shack sensors are the most widely used in ophthalmology.[2] However, they are restricted to sampling the wavefront with a lateral resolution of approximately 158 µm,[3] while assuming that the incident wavefront is well approximated as locally flat on each microlens aperture. This assumption may be acceptable for continuous wavefronts but might impose some limitations when measuring wavefronts with high frequency modes and/or abrupt changes in phase. One such case, which is the subject of interest of this article, relates to the measurement of the wavefront of eyes

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implanted with diffractive intraocular lenses (IOLs), where the abrupt jumps in phase due to diffractive zone boundaries might be poorly sampled and ambiguously processed and reconstructed by the typical modal reconstruction algorithms used by Hartmann-Shack devices.[4-7]

Compared to Hartmann-Shack aberrometers, double-pass systems are more sensitive to all of the optical defects involved in retinal image degradation but are restricted to the measurement of the modulation transfer function (MTF) (phase information is lost with the double pass).[8] In addition to other possible limitations,[9] both double-pass systems and clinical aberrometers rely on near infrared light to reduce patient discomfort and pupil miosis and increase fundus reflectance. Near infrared radiation deviates by a large amount from the design wavelength commonly used in visual optics, including design of diffractive IOLs (0.55 µm), corresponding to the peak of the spectral sensitivity function of the human eye in daylight conditions. This limitation compromises the use of these instruments to obtain meaningful information on the objective image quality of eyes implanted with diffractive IOLs due to their wavelength dependence and has led to erroneous interpretations, with both aberrometers and double-pass systems.[7,9-15] Although diffractive IOLs have been available for a few decades, to the best of our knowledge, there is still no device available to clinicians that allows objective assessment of the optical quality of patients implanted with diffractive IOLs.

A new aberrometer recently became available to clinicians (Osiris-T; CSO), which uses a pyramidal sensor to sample the incoming wavefront over a charge coupled device. Currently, the Osiris-T aberrometer offers the highest lateral resolution used by commercially available clinical aberrometers, achieving a lateral resolution of 41 µm, according to the manufacturer's specifications.[16] This instrument was recently used to compare the image quality of patients implanted with diffractive and non-diffractive IOLs, with several unexpected results.[14,15] Although the Osiris-T aberrometer brings a significant improvement in lateral resolution and does not rely on modal algorithms to reconstruct the phase, there is lack of evidence that it can properly reconstruct the phase of a wavefront formed by a pseudophakic eye implanted with a diffractive IOL and bypass the problems imposed by Hartmann-Shack aberrometers. Even if the wavefront is well reconstructed, it will not properly represent the energy distribution perceived by the pseudophakic eye under visible light, due to the wavelength dependence of the diffractive element and the near infrared light used by the instrument.[15]

To address the problems mentioned above, we first proposed and validated through optical modeling a method to scale the phase of the wavefronts obtained from patients implanted with diffractive IOLs from the near infrared to the visible light. We then investigated whether the Osiris-T aberrometer can be used to estimate the depth of field of pseudophakic eyes implanted with diffractive IOLs, by comparing the results obtained from subjective and theoretical defocus curves to simulated defocus curves, calculated from the wavefront data obtained with the Osiris-T aberrometer.

PATIENTS AND METHODS

Patients and Clinical Data

Twelve eyes were selected from a larger cohort of patients implanted with two different diffractive IOLs in a private practice (Clinsborges, Oporto, Portugal), between 2015 and 2019. This was a cross-sectional study, conducted between April and September 2022, that followed the tenets of the Declaration of Helsinki for research on human subjects. Informed consent was obtained from all participants. Institutional clinical research ethical board committee approval was obtained (SECVS 090) from the Ethics Committee for Research in Life and Health Sciences of the Ethics Council of the University of Minho. Patient ages ranged between 58 and 73 years and they had natural pupil sizes between 3 and 4 mm under photopic lighting conditions. Patients were selected based on postoperative corrected distance visual acuity of 0.1 logMAR or better and posterior capsule and remaining ocular medium transparency. After an initial evaluation of the anterior pole and fundus, their visual acuity was determined for different vergences (defocus curves) by adding trial lenses to the best distance prescription, starting from +0.50 to -3.00 diopters (D), in -0.50-D steps. During this process, the optotypes were randomized to prevent memorization. Finally, the last part of the evaluation consisted of the acquisition of individual biometric data, which included anterior corneal topography and wavefront (×3, per eye) measured with the Osiris-T, and central corneal thickness, anterior chamber depth, effective lens position, IOL thickness, and axial length measured with the optical biometer Lenstar 900 (Haag-Streit). During the wavefront acquisition, the Osiris-T acquires photographs of the pupil of the participant that were used to calculate the IOL decentration relatively to the center of the pupil. From the pixel gray level, it was also possible to assess the level of transparency of the posterior capsule (**Figure A**, available in the online version of this article).

IOLs

From the 12 eyes selected for this analysis, half had been implanted with the trifocal FineVision Micro F (PhysIOL) and the other half with the TECNIS Symfony Extended-Depth-of-Focus (Johnson & Johnson Vision). The FineVision Micro F is an acrylic diffractive IOL that splits light into three major diffractive orders (trifocal) by combining two bifocal profiles with +3.50 and +1.75 D. It has an apodized profile engraved on its anterior surface, where the step height of each diffractive zone gradually decreases from the center to the periphery of the optical zone, which has the effect of redistributing light from the near and intermediate focus to the distance focus, as the pupil size increases. It provides a spherical aberration correction of -0.11 µm for a 6-mm entrance pupil.[17,18] The TECNIS Symfony is an acrylic diffractive IOL, designed with a proprietary method for providing extended depth of focus with combined correction of spherical aberration of -0.27 µm for a 6-mm entrance pupil. Unlike the more conventional diffractive design of the FineVision Micro F, which uses 0th diffractive order for the distance focus, the TECNIS Symfony IOL uses the first diffractive order for distance, making use of the negative dispersion of the diffractive profile engraved on the posterior surface to partly correct for the chromatic aberration of the cornea at the distance focus. Its chromatic properties and design were investigated by Millán and Vega.[19] The diffractive profile of the FineVision Micro F was obtained from Loicq et al.[18] The IOLs implanted had nominal powers between +22.00 and +26.00 D.

Eye Models

The individual biometric data obtained from each patient were used to implement a set of individual eye models in Zemax Optic Studio software (Ansys). The anterior corneal surface was modeled with Optic Studio Zernike Standard Sag surface, computed from the topographic data of the participant. Further details of this modeling process are described elsewhere.[20] The refractive indexes and chromatic properties of the IOLs were obtained from the manufacturer's specifications. Distances between the different ocular surfaces were obtained from the biometric data collected with the Lenstar optical biometer (Haag-Streit). Coordinate breaks were introduced before and after the cornea and the IOL to apply the proper decentrations between the corneal apex, the IOL, and the pupil. At this point in the modeling process, some unknowns had to be assumed. The first was the posterior corneal topography. For simplicity it was assumed to be spherical, with an apical radius equal to $0.84 \times R_{anterior}$. Possible small changes in corneal astigmatism due to the non-inclusion of internal corneal topography were accounted for in the residual refraction. The position of the iris was assumed to be 0.90 mm before the IOL apex. The third unknown is related to the IOL's shape factor for different powers. All powers were designed with a Coddington shape factor

$$\left(X = \frac{R_{posterior} + R_{anterior}}{R_{posterior} - R_{anterior}}\right)$$

of a correspondent +20.00 D power. The asphericity of the anterior (TECNIS Symfony) or the posterior (FineVision Micro F) surfaces was calculated to provide the level of spherical aberration correction specified by the manufacturer. Both approximations were tested to evaluate the optical impact caused by lens decentration due to possible deviations from the real geometry. Results showed negligible differences. Another assumption is related to IOL tilt. Because it was not possible to determine the amount of tilt of each IOL, the eye models assumed 0° tilt.

Objective Through-Focus Image Quality

Objective image quality was defined as the area under the radially averaged MTF (MTFa), integrated up to 50 cycles/mm.[21] After correcting each model with its best spherical-cylindrical correction, the through-focus MTFa was calculated by adding defocus wavefronts to the distance-focused wavefront, using common Fourier techniques based on far-field scalar diffraction. A detailed description of the procedure can be found elsewhere.[22]

Data Analysis

The analysis of the data was divided in two sections. In the first section, wavefront data were calculated for 0.55 µm and for near infrared light (0.85 µm) by ray-tracing in two customized model eyes, implanted with the FineVision Micro F and the TECNIS Symfony IOLs. Results for the two wavelengths were compared using two different approaches: the first approach, used by commercial aberrometers, only considers a defocus correction from near infrared to visible light, due to the longitudinal chromatic aberration (LCA) of the eye.[23] In the second approach, the wavefront data calculated for 0.85 µm were fitted with Zernike polynomials up to the 2nd order to subtract their contribution from the total wavefront. The residual wavefront represents the optical path difference (OPD) added by the diffractive element plus the higher order aberrations of the cornea and the IOL carrier. Next, a correction factor was applied to the residual wavefront to account for the wavelength dependence of the diffractive element:

$$\boldsymbol{OPD_{0.55}} = \boldsymbol{OPD_{0.85}} \times \frac{0.85}{0.55} \times \frac{\Delta n_{0.55}}{\Delta n_{0.85}}, \text{ were } \boldsymbol{\Delta n} = \boldsymbol{n_{IOL}} - \boldsymbol{n_{Aqueous}}$$

The total OPD converted from infrared to visible light was obtained by adding back the two parts of the wavefront plus a defocus correction to compensate for the LCA.

The second section evaluated whether the Osiris-T aberrometer can reconstruct these complex wavefronts with high enough fidelity to be used to predict through-focus visual acuity. This was achieved by direct com-

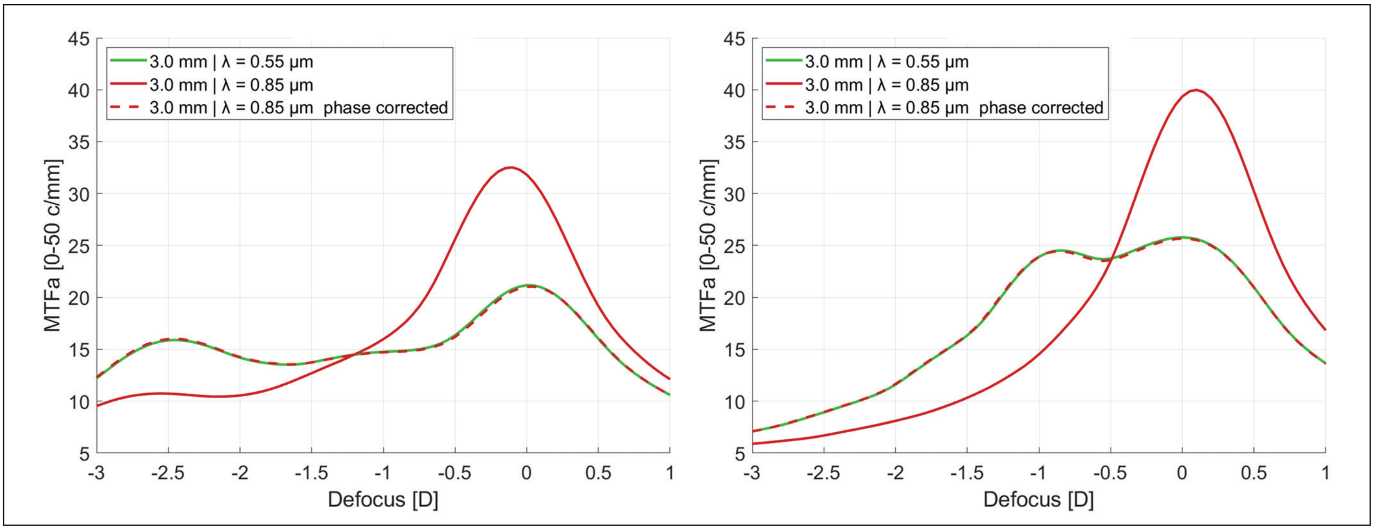

Figure 1. Area under the through-focus modulation transfer function (MTFa), calculated with different methods, of eyes implanted with the FineVision Micro F (PhysIOL) (left) and the TECNIS Symfony (Johnson & Johnson Vision) (right). MTFa calculated from the wavefront data obtained at 0.55 µm (green line); MTFa calculated from wavefront data obtained at 0.85 µm corrected only for longitudinal chromatic aberration (continuous red line); MTFa calculated with 0.85 µm after applying the scaling of the phase using equation 1 (dashed red line).

parison between three through-focus visual acuity curves: clinical visual acuity curves, simulated visual acuity calculated from the theoretical wavefronts, and simulated visual acuity calculated from the wavefronts obtained with the Osiris-T aberrometer, all matched for the patient's natural pupil size. **Figure B** (available in the online version of this article) shows an example of a real and a synthetic wavefront of an eye implanted with the trifocal FineVision Micro F IOL.

RESULTS

Spectral Scaling of the Wavefront

Image quality calculated from wavefront data obtained by ray-tracing with near infrared light, without scaling the OPD to 0.55 µm, resulted in a significantly different MTFa through-focus, compared to the reference values obtained with the design wavelength, with most of the constructive interference occurring at the distance focus. When accounting for the scaling of the OPD by applying Equation 1, the MTFa calculated from the data obtained with near infrared light showed only negligible deviations from the reference data obtained with 0.55 µm. Results are depicted in **Figure 1**.

Examples in **Figure 1** show that, after scaling the phase using the proposed method, the wavefront calculated with near infrared light perfectly matches the results obtained by ray-tracing in the customized model eyes with 0.55 µm wavelength. Instead, the usual correction of the LCA applied to the wavefront data obtained at 0.85 µm results in a through-focus curve that departs significantly from the expected through-focus performance, which in these examples provides an almost monofocal behavior rather than the expected extended depth of field.

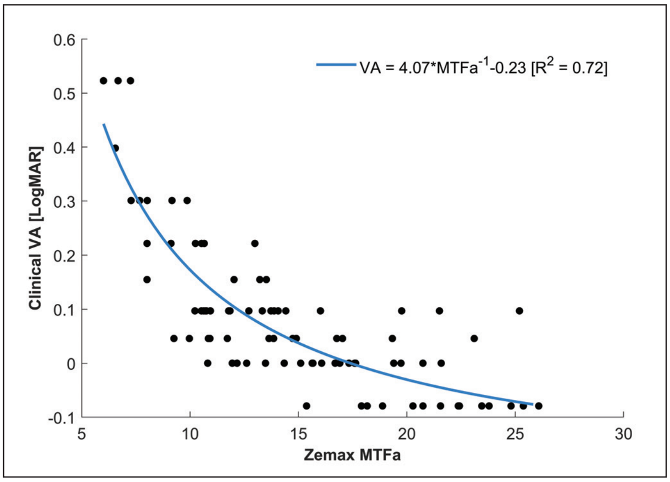

Figure 2. Prediction model of clinical visual acuity as a function of the modulation transfer function area (MTFa) calculated from the patients' customized eye models.

Through-Focus Visual Acuity

To establish a direct comparison between clinical visual acuity, the through-focus MTFa curves were converted to simulated visual acuity according to the nonlinear relation simulated visual acuity = a × MTFa^b + c described by Alarcon et al,[21] with a = 4.07, b = -1, and c = -0.23, obtained from fitting clinical visual acuity to the theoretical MTF area calculated from the eye models (**Figure 2**).[24] Results plotted in **Figures 3-4** compare the visual acuity measured with trial lenses with the simulated visual acuity calculated from the synthetic and measured wavefronts, for 12 eyes implanted with the trifocal FineVision Micro F and the extended depth-of-focus TECNIS Symfony IOLs. All curves were normalized

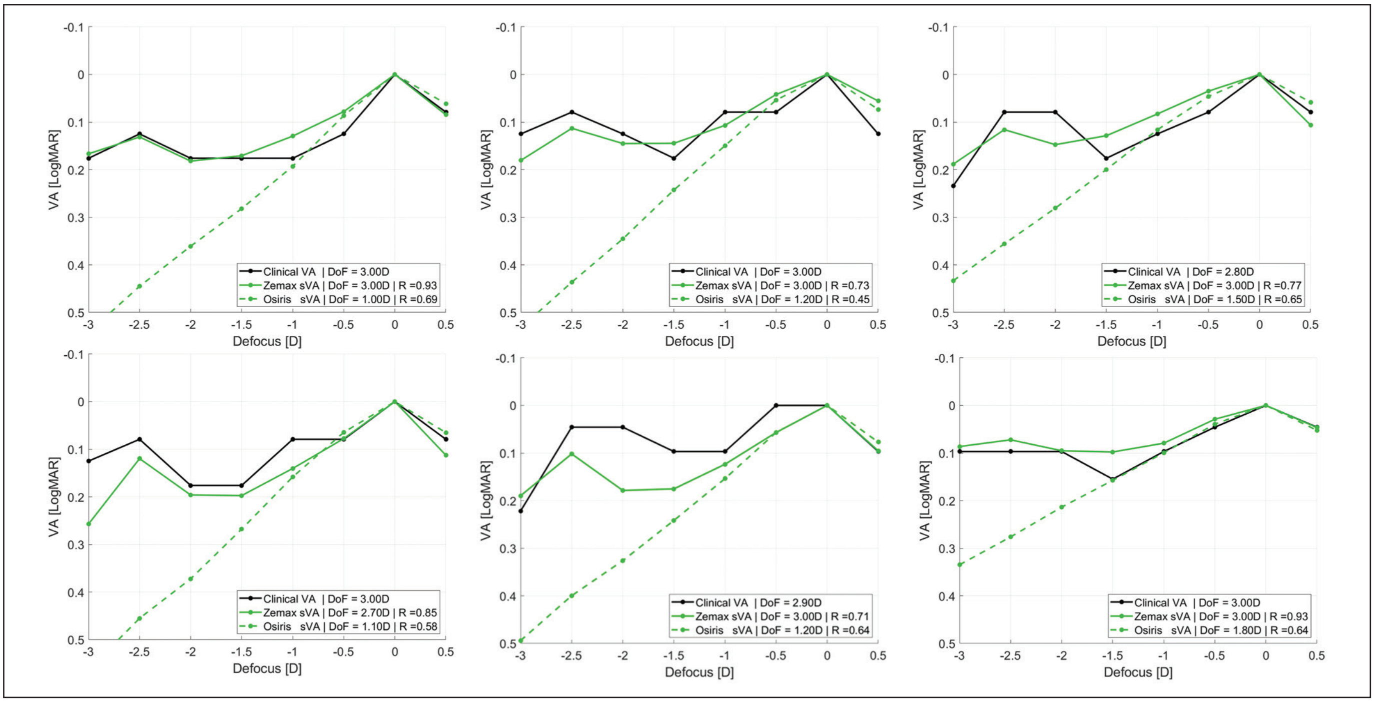

Figure 3. Direct comparison of the through-focus clinical visual acuity (VA) (black line) and the simulated visual acuity (sVA) calculated from either the synthetic wavefront data (green line) or the wavefront data measured by the Osiris-T aberrometer (CSO) (green dashed line), after applying the scaling of the optical path difference from 0.85 to 0.55 µm described in Equation 1, for the set of eyes implanted with the trifocal FineVision Micro F intraocular lens (PhysIOL). Depth of field is defined as the maximum defocus value with a visual acuity of 0.2 logMAR or better; R is the Pearson coefficient of the clinical versus simulated VA curves.

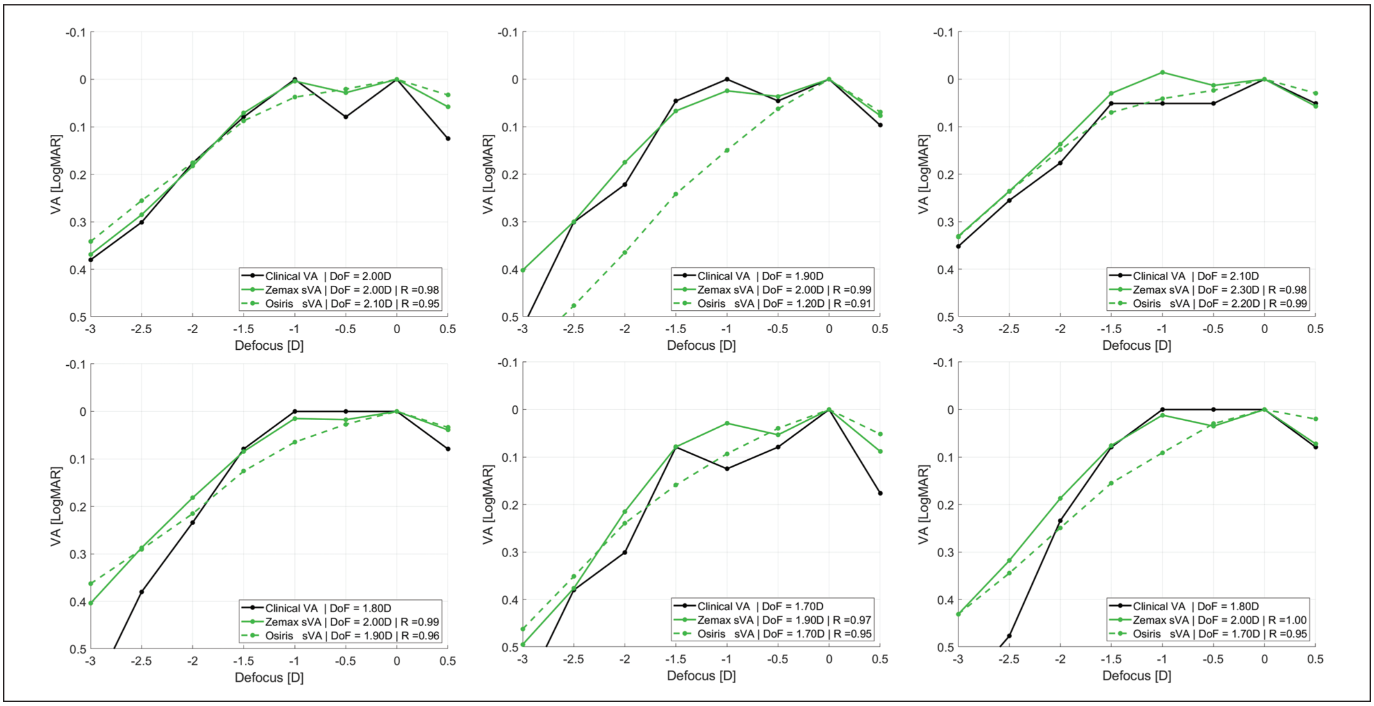

Figure 4. Direct comparison of the through-focus clinical visual acuity (VA) (black line) and the simulated visual acuity (sVA) calculated from either the synthetic wavefront data (green line) or the wavefront data measured by the Osiris-T aberrometer (CSO) (green dashed line), after applying the scaling of the optical path difference from 0.85 to 0.55 µm described in Equation 1, for the set of eyes implanted with the extended depth of focus TECNIS Symfony intraocular lens (Johnson & Johnson Vision Surgical). Depth of field is defined as the maximum defocus value with a visual acuity of 0.2 logMAR or better; *R* is the Pearson coefficient of the clinical versus simulated VA curves.

to obtain a 0.0 logMAR visual acuity at distance (0D) to facilitate comparisons and estimate the respective depth of field of each curve, defined as the maximum defocus value with a visual acuity of 0.2 logMAR or greater, estimated in 0.10-D defocus steps linearly interpolated.

Visual inspection of the curves plotted in **Figure 3**, obtained from the eyes implanted with the FineVision Micro F IOL, reveals that the through-focus simulated visual acuity calculated from the wavefronts measured with the Osiris-T have a distinct shape from either the clinical visual acuity curve or the simulated visual acuity curve calculated from the synthetic wavefronts, which show a close agreement in shape and depth of field with the clinical data. The value of the Pearson coefficient reveals only moderated correlations between the simulated visual acuity curves obtained from the Osiris-T wavefronts and the clinical visual acuity curves, opposite to the high and very high correlations obtained between the synthetic simulated visual acuity curves and the clinical visual acuity curves. The through-focus simulated visual acuity curves obtained from the wavefronts measured with the Osiris-T do not show the expected depth of field, with differences that vary between 2.00 and 1.80 D compared to the clinical measured values, revealing that the wavefront captured by the instrument's charge coupled device does not correspond to the real wavefront of those eyes.

Results plotted in **Figure 4** show the same analysis for the eyes implanted with the TECNIS Symfony IOL. There was a high agreement ($R > 0.98$) between the clinical results and the simulated visual acuity obtained from ray-tracing in the synthetic eyes, with both sets of curves following a close shape and with only minor differences between measured and estimated depth of field. There was also better agreement between the clinical and the simulated visual acuity curves calculated from the data obtained with the Osiris-T, compared to the FineVision Micro F IOL results. With the exception of one eye (top middle plot), the depth of field estimated from the measured wavefronts showed only minor deviations from the clinical values (≤ 0.10 D).

DISCUSSION

Objective evaluation of image quality in pseudophakic patients implanted with diffractive IOLs has been a subject of debate in recent years, because no instrument has yet proved capable of doing it without spurious results, which led to several misinterpretations with both aberrometers and double-pass systems.[7,9,11-15] One of the problems directly related to these erroneous interpretations is related to the near infrared light used by these clinical instruments.

Near infrared light has several advantages for clinical testing, but, in the special case of diffractive elements, introduces two major important modifications to the wavefront captured by the instrument's charge coupled device. First, the diffractive effective addition power will be higher for the longer wavelength because diffractive power is linearly dependent on wavelength. Second, because the step heights of the diffractive profile are normally optimized for the maximum spectral sensitivity of the human eye under daylight conditions (0.55 µm), light distribution will be unbalanced for near infrared radiation with most of the constructive interference occurring at the distance focus. Consequently, even if the aberrometer is capable of properly reconstructing the diffractive wavefront, it will not reproduce the light distribution perceived by that eye under visible light. Correcting for the LCA of the eye when measuring the wavefront of eyes implanted with diffractive IOLs under near infrared light only accounts for the best focus shift due to the LCA, but not for the wavelength dependence of the diffractive element, which greatly affects the effective add power and light distribution.[5-7] The method described in the first section of this study needs to be applied to properly scale the OPD from near infrared to visible light.

The second part of this study investigated whether the pyramidal wavefront sensor can overcome the limitations imposed by the more common Hartmann-Shack sensor and reconstruct the wavefront formed by a diffractive IOL with high enough fidelity to be used to predict through-focus visual acuity. The Osiris-T aberrometer provides a much higher lateral resolution than the typical Hartmann-Shack sensor (in the order of 4× times higher),[3,16] which allows better sampling of the slopes and the use of direct integration methods instead of the typical modal reconstruction. Our results show that the wavefronts measured in patients implanted with the trifocal FineVision Micro F did not carry the correct information of the real wavefronts. However, we observed an improvement in the difference between the simulated visual acuity calculated from the Osiris-T wavefronts and the synthetic and clinical visual acuity curves in the eyes implanted with the TECNIS Symfony IOL. This can be justified by two characteristics of the TECNIS Symfony design that might contribute to improve the measurement. The first is related to the larger diffractive zones of the TECNIS Symfony IOL (approximately 1.41 times larger than the trifocal FineVision Micro F), which decreases the odds of each pixel to integrate light from more than one zone (larger zones equal fewer phase jumps). The second might be related to the height of the diffractive steps of the TECNIS Symfony IOL, that for the wavelength used by the Osiris-T aberrometer induces a phase shift close to that of a monofocal IOL,[19] which might contribute to improve the measurement.[25]

Even after correcting for the phase difference due to the acquisition with infrared light, the wavefront measured by the Osiris-T aberrometer does not seem to properly reproduce the real wavefront from the pseudophakic eye implanted with a presbyopia-correcting diffractive IOL, with good enough fidelity to be used to predict their through-focus visual acuity, especially for lenses with higher add powers. Clinical scientists should be aware that there is still no objective way to estimate the visual quality of patients implanted with diffractive IOLs; therefore, it is not possible to obtain reliable conclusions with these devices.

AUTHOR CONTRIBUTIONS

Study concept and design (MF-R); data collection (MIPF, AFM-B, JS-B); analysis and interpretation of data (MF-R, JMG-M); writing the manuscript (MF-R); critical revision of the manuscript (MF-R, JMG-M, MIPF, AFM-B, JS-B); administrative, technical, or material support (MIPF, AFM-B, JS-B); supervision (MF-R)

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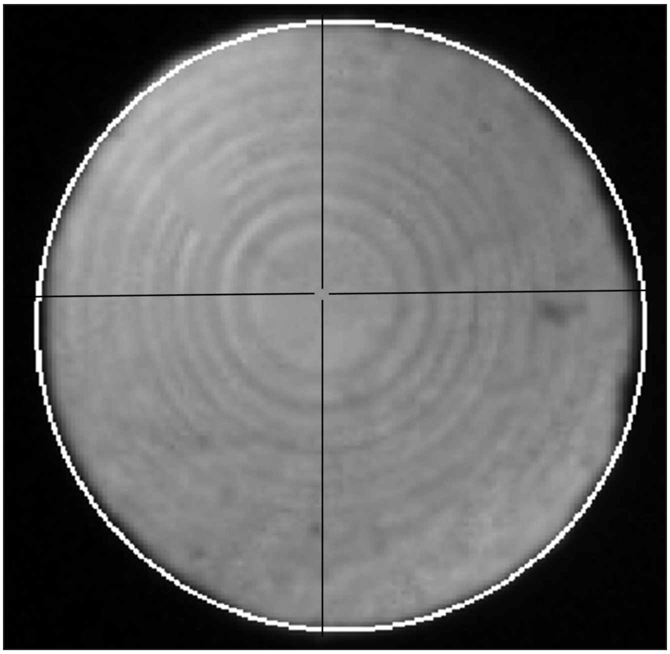

Figure A. Example of the image captured by the Osiris-T aberrometer (CSO), used to obtain the center of the intraocular lens relatively to the center of the pupil. D = diopters

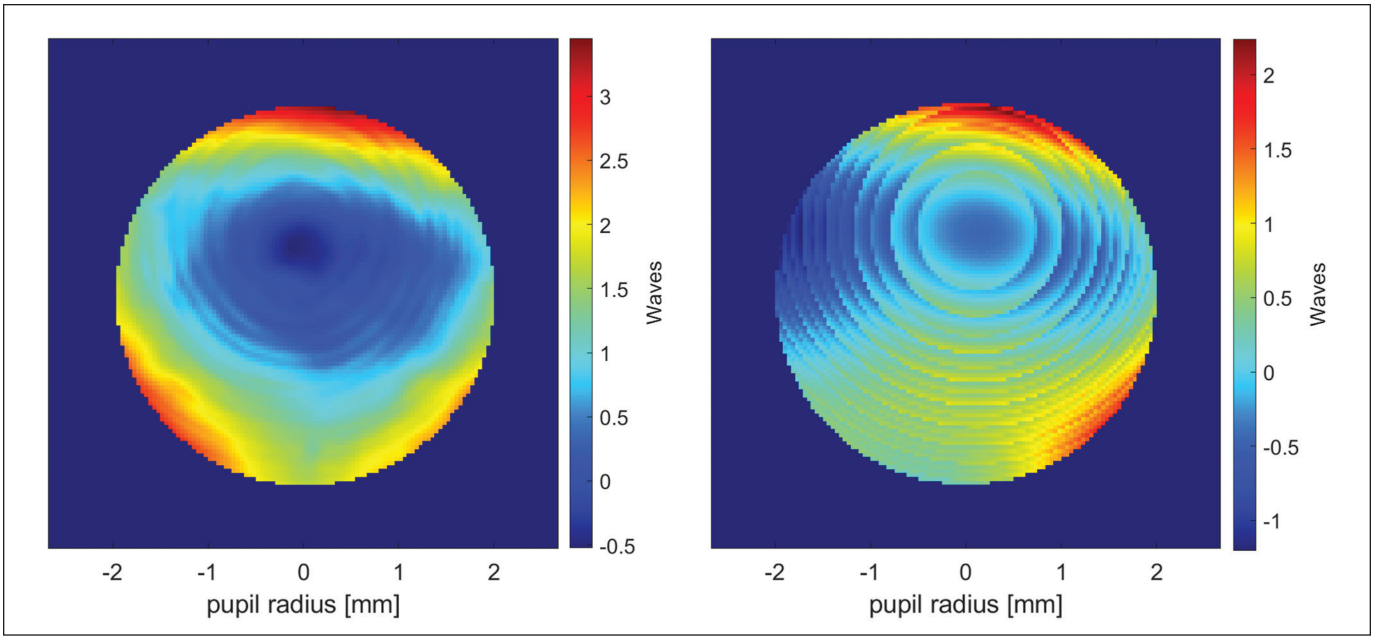

Figure B. (Left) An example of a real wavefront obtained with the Osiris-T aberrometer (CSO) in an eye implanted with the trifocal FineVision Micro F intraocular lens (PhysIOL), calculated from the average of three measurements, after scaling the phase from 0.85 to 0.55 µm and correcting the residual distance refraction. (Right) The synthetic wavefront error map of the equivalent synthetic eye, obtained by ray-tracing in Zemax Optic Studio software (Ansys).