Interaction of Monochromatic and Chromatic Aberrations in Pseudophakic Patients

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ABSTRACT

PURPOSE: To measure monochromatic aberrations at various wavelengths in eyes implanted with the Clareon monofocal aspheric intraocular lens (IOL) (Alcon Laboratories, Inc., Fort Worth, TX). The authors estimated longitudinal chromatic aberration (LCA), modulation transfer functions (MTFs), and the impact of interactions between chromatic and monochromatic aberrations on retinal image quality.

METHODS: Ten patients (age: 68.4 ± 3.21 years) were measured in two experiments: (1) Hartmann-Shack wave aberrations at five visible wavelengths (480 to 700 nm) and (2) best subjective focus at each wavelength. Objective and psychophysical LCAs were obtained from the Zernike defocus and psychophysical best focus, respectively. MTFs were calculated for the closest wavelengths to the peak sensitivity of the three cone classes (S [480 nm], M [555 nm], and L [564 nm]) using the measured aberrations and chromatic difference of focus. The degradation produced by LCA was estimated as the visual Strehl ratio for green divided by the visual Strehl ratio for blue and red.

RESULTS: The root mean square for higher order aberrations (HOAs) ranged from 0.0622 to 0.2084 µm (700 nm, 4.3-mm pupil). Monochromatic visual Strehl ratio was above 0.35 in all patients. LCA was 1.23 ± 0.05 diopters (D) (psychophysical) and 0.90 ± 0.11 D (objective). Visual Strehl ratio decreased by a factor ranging from 1.38 to 3.82 on chromatic defocus from green to blue. There was a significant correlation between native visual Strehl ratio and the degradation produced by LCA (ie, visual Strehl_{555}/visual Strehl_{480}).

CONCLUSIONS: The Clareon IOL compensates for spherical aberration, with postoperative wave aberrations dominated by astigmatism and other HOAs, being highly subject-dependent. The impact of LCA in blue is largely dependent on the magnitude of monochromatic aberrations.

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Replacement of the opacified natural crystalline lens by an artificial intraocular lens (IOL) has been standard practice in cataract surgery for decades. Although a primary objective of cataract surgery is to restore transparency in the ocular media, the procedure has long been considered a refractive correction, because typically the IOL is selected so that, in combination with the cornea, it focuses distant images on the retina. Furthermore, state-of-the-art IOLs in the market are designed with aspheric surfaces, typically inducing negative spherical aberration to compensate for the positive spherical aberration of the cornea.[1] This compensation, not found with spherical surface IOLs,[1,2] mimics the balance produced by the crystalline lens in the young human eye, a property that is known to be lost with age.[3] Numerous studies have reported low spherical aberration in eyes implanted with aspheric IOLs, such as the Acrysof IQ or the Tecnis IOL,[1,4,5] supporting theoretical predictions using computer eye models.[6] The induction of negative

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spherical aberration by the IOL has also been shown to result in a passive compensatory effect of lateral coma, found to be positive in the anterior cornea and negative in the IOL, arising from the off-axis position of the fovea.[7,8] Decreased higher order aberrations ultimately result in an improvement of the modulation transfer function (MTF) of eyes implanted with aspheric IOLs (in comparison with eyes implanted with spherical IOLs).[1] The presence of other higher order aberrations in the eye, potential residual refractive errors, and chromatic effects likely limit the full benefit of the spherical aberration compensation.[9]

Although aberrations are typically measured in monochromatic light and calculations of the MTFs of pseudophakic eyes normally refer to monochromatic image quality, it has been recognized that chromatic aberrations, in particular longitudinal chromatic aberration (LCA), also play a role in image quality degradation.[10,11] LCA arises from chromatic dispersion of the ocular media, with different wavelengths focusing at different planes.[12] There are numerous reports of the LCA in the phakic human eye, generally assessed through psychophysical techniques (average LCA = 1.50 diopters [D], in the range of 480 to 700 nm) and also through reflectometric techniques (that have been shown to underestimate this value, likely as a result of deeper layers reflecting a larger proportion of the longer wavelengths and the nerve fiber layer reflecting a larger proportion of the shorter wavelengths).[13,14]

Replacement of the crystalline lens by an IOL changes LCA, given the different dispersive properties of the IOL materials in comparison with those of the crystalline lens.[15-18] We have recently reported in vivo measurements of chromatic difference of focus and LCA in pseudophakic patients, implanted with IOLs of different materials (Tecnis, Acrysof, and hydrophobic and hydrophilic PhysIOL materials), showing differences that are consistent with the Abbe number of the IOL materials.[5,15,19] The possibility of modulating LCA is emerging with new diffractive designs, which minimize LCA for at least some distances.[20-23]

Monochromatic and chromatic aberrations cannot be regarded in isolation. It is common to evaluate the MTF of the eye assuming best focus in green (maximum sensitivity of the M cone class), and defocused by the natural LCA (for the peak sensitivity of the S cone class) to assess the relative contributions of the eye's monochromatic aberrations and LCA. It has been observed that, although the MTF in blue is largely degraded with respect to the MTF in green for a diffraction-limited eye, the difference between MTF in green and blue is much smaller for aberrated eyes.[24] This favorable interaction between chromatic and monochromatic aberration suggests that monochromatic aberrations are, in a sense, protecting the eye against chromatic blur.[24] The interactive role of chromatic and monochromatic aberrations with IOLs can be assessed computationally and on bench. However, these models are unlikely to capture the complexity of the natural eye. Therefore, individual measurements in vivo are needed to assess the optical quality of pseudophakic eyes.

In this study, we measured for the first time monochromatic aberrations of eyes implanted with a new monofocal acrylic IOL design (Clareon model CNA0T0; Alcon Laboratories, Inc., Fort Worth, TX).

PATIENTS AND METHODS

Ten patients (mean age: 68.4 ± 3.21 years, range: 59 to 78 years) scheduled for cataract surgery at the Fundación Jiménez Díaz Hospital were implanted with the Clareon IOL and followed up in the clinic (standard clinical examinations) and at the Visual Optics and Biophotonics Lab of the Institute of Optics-CSIC (VIOBIO-CSIC) for measurements of monochromatic and chromatic aberrations using custom-developed equipment. Patients were invited to participate under the following inclusion criteria: good general and ocular health except for cataract and requiring IOL power between 18.00 and 24.00 D. Preoperative tests were performed at the Fundación Jiménez Díaz Hospital and included slit-lamp examination, autorefractometry, optical biometry and keratometry (IOLMaster; Carl Zeiss Meditec AG, Jena, Germany), and visual acuity testing (Nidek Auto-chart projector CP-670 Optotypes; Nidek, Gamagori, Japan).

The Clareon IOL is a single-piece hydrophobic IOL made of an acrylate–methacrylate copolymer, with water content of 1.5% at 35°C, refractive index of 1.55, and glass transition temperature of 9.1°C. The material incorporates an ultraviolet blocker for protection against radiation in the ultraviolet range and a blue-light filtering chromophore. The overall design of the Clareon CNA0T0 IOL is based on the platform of its predecessor, the single-piece AcrySof IOL.[25]

The IOL power was calculated with the SRK-T formula using the A-constant recommended by the manufacturer. The power expected to give the negative refraction closest to emmetropia was chosen. Lenses were loaded into a Monarch II injector (Alcon Laboratories, Inc.) and implanted through a 2.2-mm incision in the steepest meridian. All patients underwent uneventful surgeries. The postoperative treatment consisted of 3% ofloxacin drops (four times a day) for 7 days and dexamethasone drops tapered for 5 weeks.

Patients who met the postoperative inclusion criteria (no intraocular complications, residual astigmatism of less than 1.00 D, clear capsular bag, and correctable visual acuity of 0.0 logMAR or better) were finally scheduled for postoperative measurements at VIOBIO-CSIC. Patients were measured at least 4 weeks after surgery.

Patients signed informed consent forms following acquaintance with the nature of the study. The study complied with the tenets of the Declaration of Helsinki. Informed consent and study protocols were approved by the Clinical Ethical Committee of the Fundación Jiménez Díaz Hospital and the Bioethical Committee of CSIC.

Polychromatic Adaptive Optics System

Experimental measurements were performed in a custom-developed polychromatic adaptive optic system at VIOBIO-CSIC, described in detail in previous publications.[14,22] Only a subset of components and channels of this adaptive optic visual simulator were used in the current study, namely the Hartmann–Shack wavefront sensor, the psychophysical stimulus channel, and the pupil monitoring channel. The light source for all testing channels was provided by a supercontinuum laser source (SCLS; Fianium Ltd, Southampton, United Kingdom), in combination with a dual acousto-optic tunable filter module (Gooch & Housego, Ilminster, United Kingdom), operated by radio-frequency drivers, which, for the purposes of this study, allow automatic selection of visible wavelengths (480 to 700 nm, in our system configuration). This light source was used as a probe in the wavefront sensor channel (Hartmann–Shack, MIRAO52; Imagine Eyes, Orsay, France). This source also illuminated the psychophysical stimulus (a Maltese cross) generated on a Digital Micro-Mirror Device (DLP Discovery 4100 0.7 XGA; Texas Instruments, Dallas, TX) conjugate to the retina and subtending 1.62°. A Badal optometer was shared by all channels, allowing correcting of the patient's residual spherical error. The pupil monitoring system consisted of a camera (DCC1545M; Thorlabs GmbH, Bergkerchen, Germany) conjugated to the eye's pupil. Two automatized shutters allowed simultaneous illumination of the eye and the stimulus.

Experimental Protocols

Experimental protocols were similar to those described by Vinas et al.[14] using a previous version of the polychromatic adaptive optics system on phakic patients. The patients' pupils were dilated using two drops of tropicamide 1%, and aligned to the system with the aid of a bite bar and the pupil monitoring channel, while they viewed the fixating stimulus illuminated by green light. Patients were instructed to bring the stimulus in focus by moving the Badal system, controlled by the keyboard. The focus value was obtained three times and set to the average of the settings. Hartmann–Shack wavefront sensor measurements were obtained at the selected best focus, for five wavelengths, ranging from 480 to 700 nm, with at least three repeated measurements for wavelength. For psychophysical measurements, the stimulus was illuminated in random order with five wavelengths ranging from 480 to 700 nm. The focus setting was repeated for each wavelength at least three times. The luminance (approximately 20 cd/m^2) for each wavelength had been previously automatically controlled and set to appear iso-illuminant for the patients.

Objective Chromatic Difference of Focus and LCA

Wave aberrations were obtained for each wavelength and captured Hartmann–Shack wavefront sensor image using centroid detection algorithms and Zernike polynomial (7th order) reconstruction. Wave aberrations were obtained for the maximally dilated pupil (ranging from 4.3 to 5 mm). For comparison across patients, individual aberrations and root mean square (RMS) wavefront error are reported for a 4.3-mm pupil diameter, common to all patients. Full wave aberrations are reported for 700 nm.

The defocus term for the Zernike polynomial expansion was obtained for each wavelength, transformed to diopters, and reported as the average of at least three repetitions. The objective chromatic difference of focus is estimated at the difference in focus for each wavelength with respect to 555 nm. The objective LCA is calculated as the objective chromatic difference of focus between 480 and 700 nm.

Psychophysical Chromatic Difference of Focus and LCA

The subjective focus setting was obtained for each wavelength and reported as the average of at least three repeated settings. The psychophysical chromatic difference of focus is estimated as the difference in subjective focus for each wavelength with respect to the setting at 555 nm (taken as the reference). The psychophysical LCA is given as the psychophysical chromatic difference of focus between 480 and 700 nm.

Calculations of MTF in Green and Blue Light

The diagram in **Figure A** (available in the online version of this article) illustrates the calculations performed using the measured wave aberrations and psychophysical LCA. The MTFs for 3rd and higher order aberrations (ie, no astigmatism) were calculated using Fourier optics, assuming a constant amplitude (ie, no Stiles-Crawford effect) and a 4.3-mm pupil diameter, and the wave aberration as the phase of the pupil function. Calculations were done for the wave-

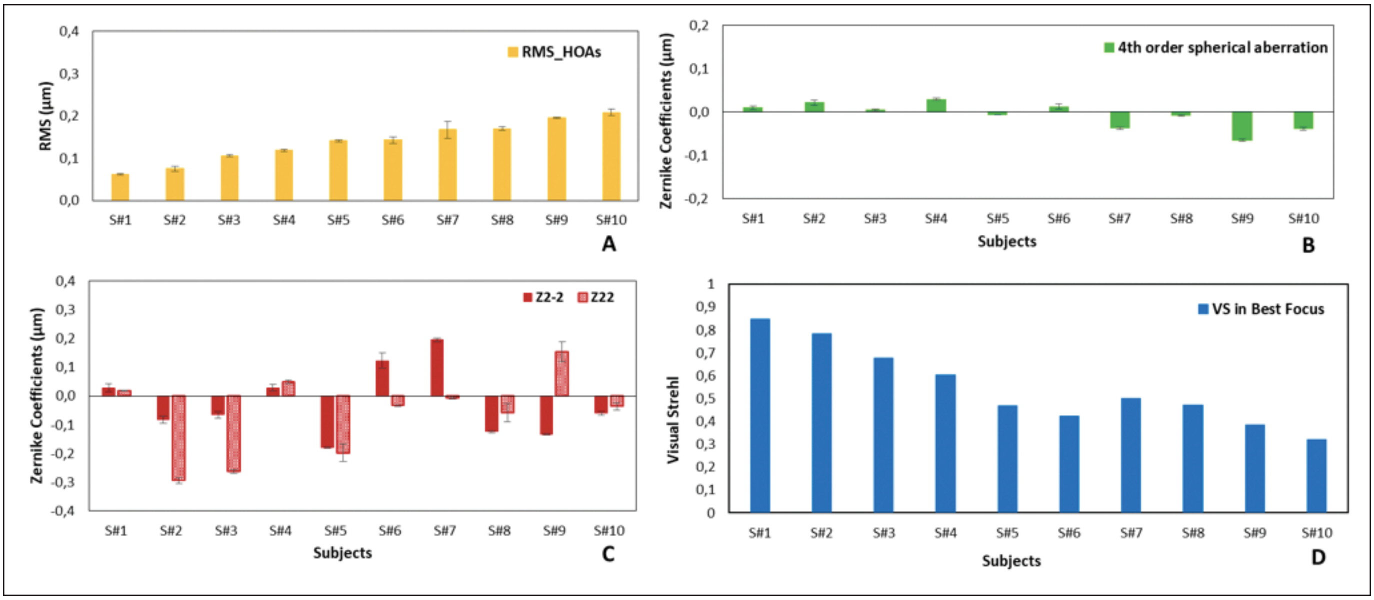

Figure 1. (A) Root mean square (RMS) for 3rd and higher order aberrations (HOAs). (B) 4th order spherical aberration Zernike term. (C) Astigmatism Zernike coefficient terms. (D) Visual Strehl (VS) ratio. All values are for a 4.30-mm pupil diameter.

lengths closest to the peak sensitivity of the three cone classes (S, M, and L at 480, 555, and 564 nm, respectively), using the Zernike coefficients obtained for each wavelength and patient (average across three repeated measurements). The defocus term was set to 0 at 555 nm. The corresponding psychophysical chromatic difference of focus (in microns) was used as the defocus terms for 480 and 700 nm for each patient. The psychophysical value was used instead of the Zernike value obtained from the Hartmann–Shack wavefront sensor measurements, because it is known that the objective chromatic difference of focus obtained from reflectometric measurements (Hartmann–Shack wavefront sensor or double-pass) underestimates the actual value.[14] The MTFs are represented as radial profiles (averaged across all orientations, cut-off at 60 c/deg.

Data Analysis

The optical quality at 480, 555, and 564 nm was calculated for each patient and wavelength in terms of visual Strehl ratio (volume under the MTF modulated by the neural contrast sensitivity function). The degradation produced by the LCA was estimated at the ratio of the visual Strehl_{480}/visual Strehl_{555} and visual Strehl_{564}/visual Strehl_{555} (**Figure 1**). Given the larger impact of the degradation in blue light, further comparative analysis was performed for 480 nm. For comparison, calculations were also performed assuming the same lens (and measured monochromatic performance) but with Acrysof and Tecnis materials.

RESULTS

Table 1 summarizes the preoperative and postoperative data of the participants.

Monochromatic Wave Aberrations

Figure B (available in the online version of this article) shows wave aberration maps for patients at their maximum pupil diameter (ranging from 4.3 to 5 mm). The patients (**Table 1**) are ordered by increasing RMS wavefront error (for 4.3 mm), from least aberrated (S#1) to most aberrated (S#10).

As shown in **Figure 1A**, the RMS ranged from 0.0622 to 0.2084 µm. The fourth spherical aberration (**Figure 1B**) is close to zero in all patients (-0.0077 ± 0.0029 µm, on average). The most relevant aberration in several patients is astigmatism. The intersubject variability in postoperative astigmatism magnitude (0.25 D standard deviation) and angle is likely due to the variations in preoperative astigmatism and the correction/induction of astigmatism created by the incision in combination with preoperative values. Monochromatic visual Strehl ratio was above 0.35 in all patients, reaching almost diffraction-limited values (0.8 and above) in S#1. Although there is some correspondence between low RMS and high visual Strehl ratio, the moderate correlation is due to interactions between aberrations influencing retinal image quality metrics, visual Strehl ratio in particular.

Objective and Psychophysical Chromatic Difference of Focus and LCA

Chromatic difference of focus plots (**Figure C**, available in the online version of this article) were fitted by

TABLE 1

Preoperative and Postoperative Data

					Preop Refraction					Postop Refraction			
ID	Age (y)	Eye	AL (mm)	Ks × Kf (D)	Sph (D)	Cyl (D)	Axis (Degrees)	Preop VA (Decimal)	IOL Power (D)	Sph (D)	Cyl (D)	Axis (Degrees)	Postop VA (Decimal)
S#1	73	OS	23.48	43.95 × 43.60	+1.25	-1.00	80	0.4	21.00	0.00	-0.50	70	1.0
S#2	70	OD	22.28	43.95 × 43.55	-0.75	0.00	0	0.6	23.00	0.00	-1.00	100	1.0
S#3	78	OD	24.61	40.76 × 40.13	+3.00	-1.75	110	0.6	20.50	+0.25	-0.50	90	1.0
S#4	74	OD	23.42	44.41 × 43.38	+0.25	-1.25	80	0.4	21.00	0.00	0.00	0	1.0
S#5	70	OD	23.58	45.24 × 45.24	0.00	-1.50	110	0.5	19.00	0.00	-0.75	60	1.0
S#6	70	OS	24.17	41.26 × 40.91	+2.00	-0.50	15	0.8	22.00	-0.25	-0.50	50	1.0
S#7	59	OS	23.21	45.06 × 44.12	+2.00	0.00	0	0.4	21.00	0.00	-0.50	20	1.0
S#8	62	OS	22.54	45.98 × 45.24	-1.25	0.00	0	0.4	21.50	+0.50	-0.50	125	1.0
S#9	59	OD	23.20	45.30 × 43.95	+1.00	-1.25	170	0.5	21.00	-0.25	-0.75	150	1.0
S#10	69	OS	22.60	44.76 × 43.38	-0.25	-0.50	75	0.5	23.00	+0.25	-0.50	85	0.8

AL = axial length; Ks = anterior corneal curvature at the steepest meridian; Kf = anterior corneal curvature at the flattest meridian; D = diopters; sph = sphere; cyl = cylinder; preop = preoperative; VA = visual acuity; IOL = intraocular lens; postop = postoperative; OS = left eye; OD = right eye

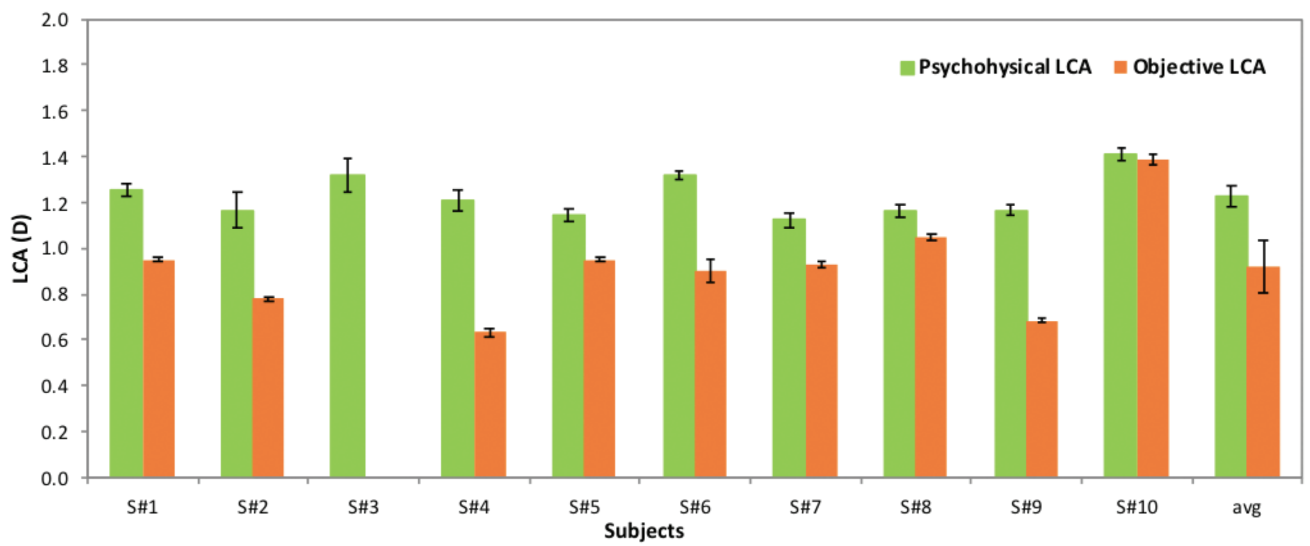

Figure 2. Longitudinal chromatic aberration (LCA) (480 to 700 nm), from objective (orange bars) and psychophysical (green bars) measurements, for all patients and average across patients. D = diopters

linear functions in the 480 to 700 nm range, and showed lower slopes for the objective functions (0.0042 D/nm, on average, **Figure CA**) than for the psychophysical functions (0.0056 D/nm, on average, **Figure CB**). Intersubject variability was low (0.0005 and 0.0002 D/nm standard deviations of the slopes for objective and psychophysical functions, respectively). The estimated LCA showed a consistent shift between objective and psychophysical LCA in all patients, being on average 0.92 ± 0.11 and 1.23 ± 0.05 D, respectively (**Figure 2**).

Estimated Impact of the LCA

Figure 3 shows the MTFs for green, red, and blue cones, respectively, in eyes implanted with the Clareon IOL. Although there was little difference between the MTF for green and red cones given the proximity of their spectral sensitivity (visual Strehl ratio), there was a significant drop in optical quality for blue, particularly in the least aberrated eyes (S#1 to S#2), but practically no difference in the most aberrated eyes, as shown in **Figure 4A** (solid green bars). The average visual Strehl$_{555}$/visual Strehl$_{480}$ ratio for the Clareon IOL was 2.36. Individual visual Strehl$_{555}$/visual Strehl$_{480}$ ratio ranged from 1.38 to 3.82. As expected, the visual Strehl$_{555}$/visual Strehl$_{480}$ ratio decreased for larger pupils (1.21 for S#10 and 1.42 for S#5, with pupil diameters of 5.15 and 5.04 mm, respectively). To evaluate the impact of the material

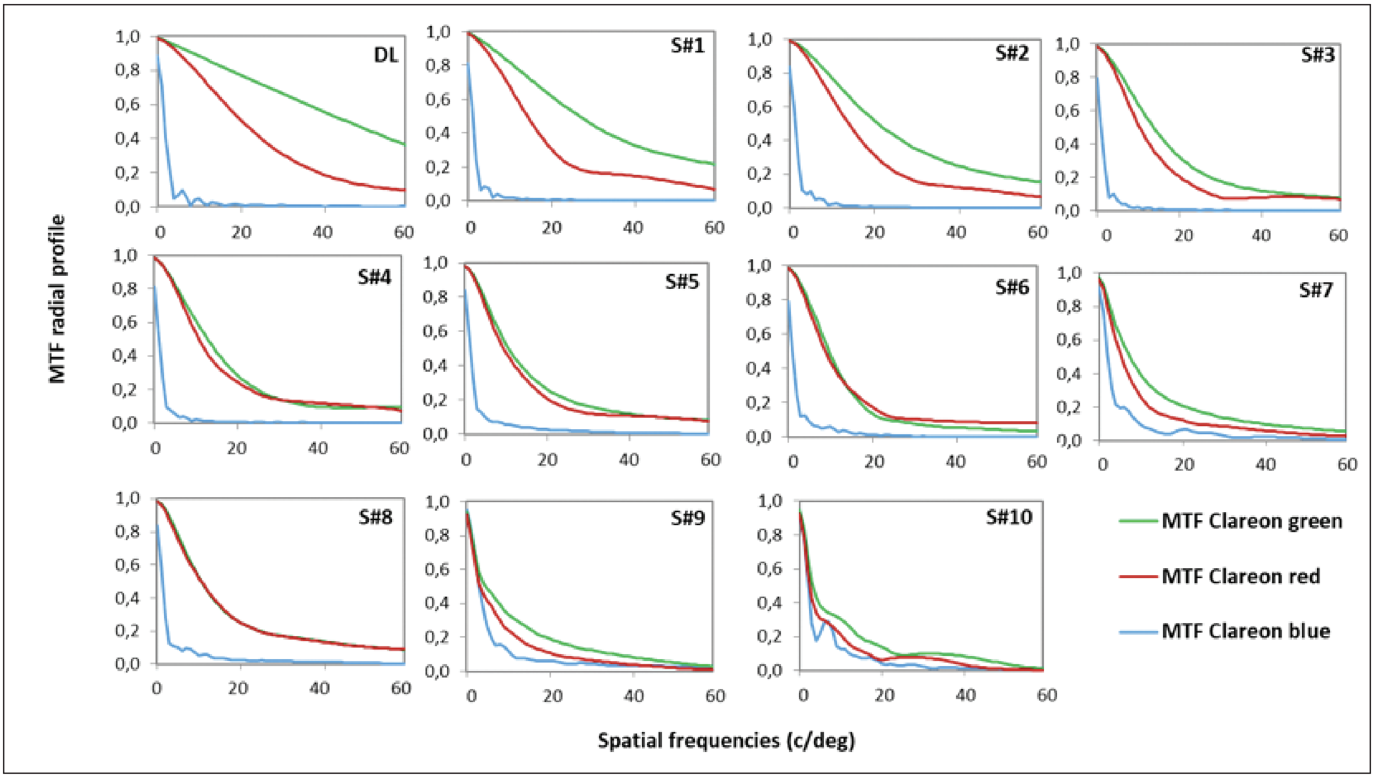

Figure 3. Estimated modulation transfer functions (MTFs) in green (555 nm) best focus, blue (480 nm), and red (564 nm) defocused by the measured chromatic defocus. Data for all patients ranked by increasing root mean square.

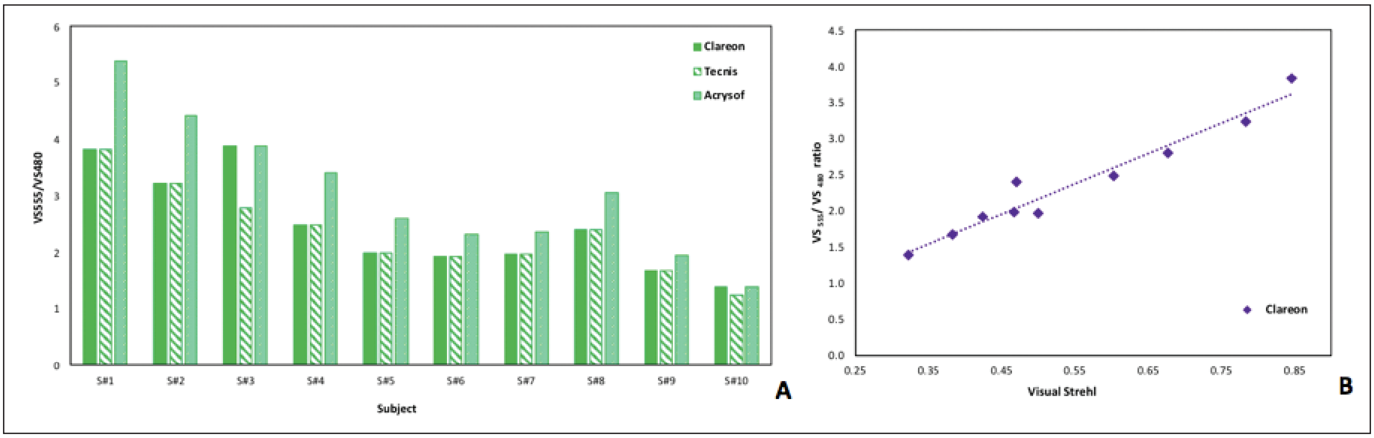

Figure 4. (A) Visual Strehl$_{555}$/visual Strehl$_{480}$ ratios for Clareon (Alcon Laboratories, Inc.), Acrysof (Alcon Laboratories, Inc.), and Tecnis (Johnson & Johnson) materials. (B) Visual Strehl$_{555}$/visual Strehl$_{480}$ ratios as a function of optical quality at best focus and corresponding linear regression (r = 0.97, P < .0001).

alone, we repeated the calculations for theoretically identical IOLs, but with different materials, and therefore different chromatic properties (Tecnis and Acrysof). The corresponding visual Strehl$_{555}$/visual Strehl$_{480}$ ratio is also shown in **Figure 4A**. We will discuss these data in a wider context in the discussion. Also, we found a statistically significant negative correlation between visual Strehl ratio in focus and visual Strehl$_{555}$/visual Strehl$_{480}$ ratio, indicating a larger impact of LCA on the least aberrated eyes (**Figure 4B**).

DISCUSSION

We measured for the first time monochromatic aberrations in patients implanted with a the new monofocal aspheric Clareon IOL. We found that the IOL (with a nominal spherical aberration of -0.20 µm) corrects, on average, spherical aberration in patients, likely as the result of balance of positive spherical aberration of the cornea with negative spherical aberration induced by the IOL. Spherical aberration for 4.3-mm pupils was practically zero (< 0.01 µm) in magnitude in 2 patients, slightly positive for 3 patients (0.0129 to 0.0299 µm), and slightly negative in 5 patients (-0.0388 to -0.0066 µm). Differences are expected due to variability in corneal spherical aberration in the population. In any case, the postoperative spherical aberration is negligible and comparable to the spherical aberration reported in patients implanted with state-of-the-art aspheric IOLs (ie, Acrysof IQ 0.041 ± 0.06 µm and Tecnis 0.035 ± 0.07 µm, for 5-mm pupils, scaled down to 0.18 and -0.0067 for 4.3-mm pupils.[8] Other authors have reported spherical aberrations of 0.075 and 0.02 µm, for 4-mm pupil diameters, for the Alcon SN60AT IOL and Hoya XY-1 IOLs.[20]

In this study, we also characterized, for the first time, the LCAs in patients implanted with IOLs of a new material, the Clareon. It is well known that material properties, defined by the Abbe number,[26] determine the chromatic aberration of the IOL and therefore impact the chromatic aberration of the eye in which it is implanted. In previous studies, we measured LCA in phakic patients and in patients implanted with monofocal and multifocal IOLs with hydrophilic, hydrophobic, and acrylic materials.[15,19,22] In agreement with all previous studies, we found that the LCA obtained from wavefront sensing is consistently lower than that obtained psychophysically, with an offset value of 0.31 D in this study, slightly lower than that found between in phakic and pseudophakic patients with hydrophilic and hydrophobic IOLs (approximately 0.50 D).[14,15,22] This difference likely results from the wavelength-dependency of the retinal reflection (longer wavelengths are reflected deeper in the retina and shorter wavelengths more superficially), partially compensating for LCA when using reflectometric methods, such as wavefront sensing.[13,14] Both intrasubject variability (0.04 and 0.02 D for psychophysical and reflectometric LCA, respectively) and intersubject variability (0.05 and 0.11 D for psychophysical and reflectometric LCA, respectively) are low and without clinical relevance.

The estimated psychophysical LCA (1.23 D) within the psychophysically LCA reported for other materials: 1.37 D (PhysIOL hydrophobic); 1.21 D (PhysIOL hydrophilic) ranging from 480 to 700 nm[15]; 0.75 D in polymethylmethacrylate IOLs and 1.20 D in Acrysof IOLs ranging from 500 to 640 nm[27]; 0.96 D in SN60WF (Alcon); and 0.80 D in NY-60 (HOYA), 1.01 D in XY-1 (HOYA), and 0.66 D in ZCB00V (AMO) ranging from 561 to 840 nm.[20] Pérez-Merino et al.[19] reported objective LCA for Acrysof and Tecnis ranging from 532 to 785 nm. Considering the offset between objective and psychophysical LCA and extrapolating the values to the 480 to 700 nm range, a psychophysical LCA of 1.49 and 1.19 D is predicted for the Acrysof and Tecnis IOLs, respectively.[19] The Clareon material's LCA (1.23 D) lies between those values and it is also lower than the LCA in phakic eyes (1.52 D).[14]

To understand the impact of the measured LCA on optical quality in patients implanted with the Clareon IOL, we estimated the MTFs for the green, red, and blue cones. We found that the impact of LCA on optical quality in blue was highly dependent on the magnitude of higher order aberrations. LCA produced a higher drop in the MTF in eyes with a lower amount of aberrations, and a lower drop in more aberrated eyes, which are almost insensitive to LCA, The visual Strehl_{555}/visual Strehl_{480} ratio ranged from 1.38 to 3.82. This is consistent with observations in normal phakic eyes, which led to the conclusion that monochromatic aberrations are the eye's protection against chromatic blur.[24] In fact, a shift in focus by 0.50 D further improves the MTF in blue and degrades it in green, shifting the visual Strehl_{555}/visual Strehl_{480} to 0.38 on average. To evaluate differences in the interactions of monochromatic aberrations and LCA for different IOL materials, we repeated the calculations assuming similar monochromatic aberrations but different chromatic difference of focus (555/480 nm), for the Acrysof (0.52 D) and Tecnis (0.40 D) IOLs. The visual Strehl_{555}/visual Strehl_{480} ratios at 480 nm for all three materials are shown in **Figure 4**. As expected, the estimated effect of LCA on optical quality is highest for the Acrysof material (1.2 times higher than in the Clareon, on average), whereas the Clareon and Tecnis materials behave more similarly.

It should be noted that the contributions of the LCA in the polychromatic MTF would be further attenuated by the retinal spectral sensitivity (and also by the emission spectra of daylight or standard lamps), which reduces the effect of the red and blue ends of the spectrum. We did not attempt to generate realistic polychromatic MTFs because the transverse chromatic aberration[26] was not measured (and it is expected to vary across eyes). Knowledge of LCA, transverse chromatic aberration, and monochromatic aberrations makes it possible to estimate the polychromatic MTF and predict the optical benefit of correcting monochromatic aberrations (while leaving transverse chromatic aberration and LCA), or correcting LCA while leaving monochromatic aberrations and transverse chromatic aberration.[12,28-30] Polychromatic MTFs need to consider the spectral content of the light source, the spectral sensitivity of the cones (V[l]) and the yellow chromophore filtering of the IOLs (although the impact of the latter is expected to be negligible on optical quality[24]). Although the presented calculations are prediction exercises, they may represent realistic situations because emerging lens designs are custom corrected and chromatic aberration starts to be modulated with new diffractive designs. Furthermore, we may not expect a direct correspondence between optical and perceived quality in polychromatic light, because the eye appears to be adapted to its native monochromatic aberrations[31,32] and likely also to its native LCA.

Overall, the Clareon IOL adequately fulfils the goal of correcting corneal spherical aberration, whereas the wave aberration appears dominated by other higher order aberrations, characteristic to the patient's cornea/eye. The LCA with the Clareon IOL is similar to that of other lenses, and its optical effect largely depends on the interaction of monochromatic and chromatic aberrations, in a similar way to how to it occurs in virgin eyes. Further studies should investigate the potential visual impact of these variations in chromatic and monochromatic aberrations.

AUTHOR CONTRIBUTIONS

Study concept and design (SM, MV, NA, IJ-A); data collection (SM, MR, CB-G, AG-R); analysis and interpretation of data (SM, MR, CB-G, AGR, MV); writing the manuscript (SM, MR); critical revision of the manuscript (SM, CB-G, AG-R, MV, NA, IJ-A); statistical expertise (SM, MR); administrative, technical, or material support (SM, MR, CB-G, AG-R, MV, NA, IJ-A); supervision (SM, MV, NA, IJ-A)

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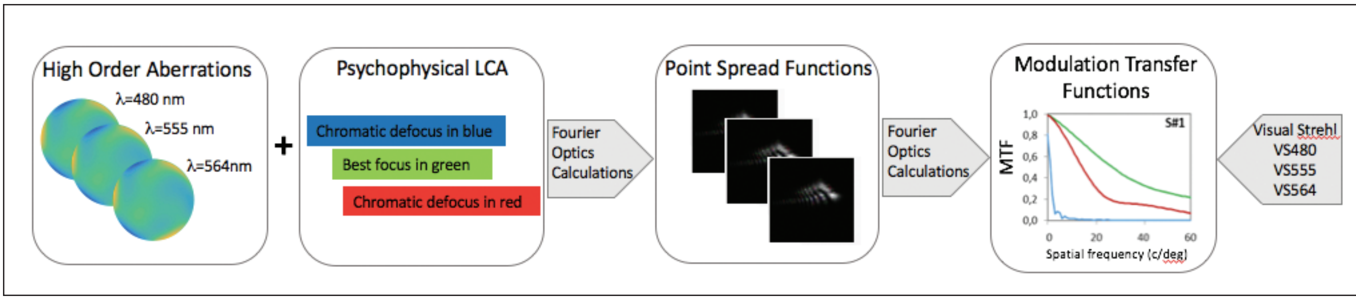

Figure A. Illustration of the calculations performed to obtain the modulation transfer function (MTF) and visual Strehl ratio (VS) for the S, L, M cone classes peak wavelengths. LCA = longitudinal chromatic aberration

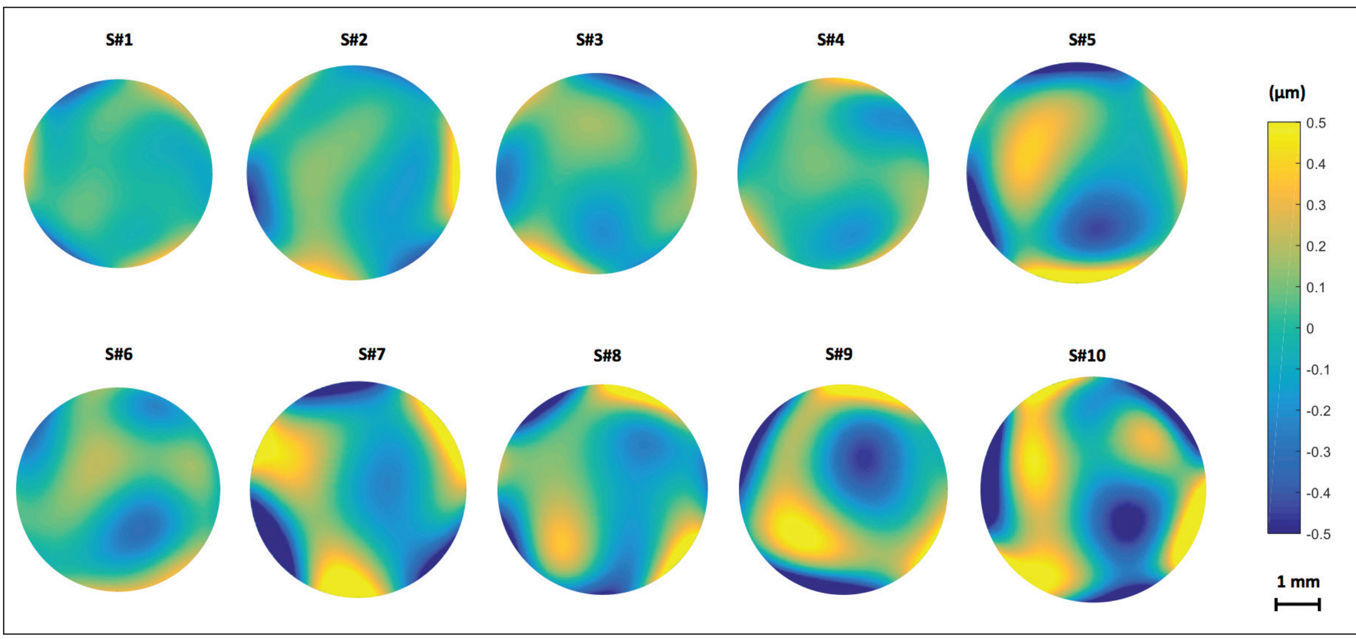

Figure B. Wave aberrations for all patients of the study (pupil diameters ranging from 4.3 to 5 mm), measured at 700 nm.

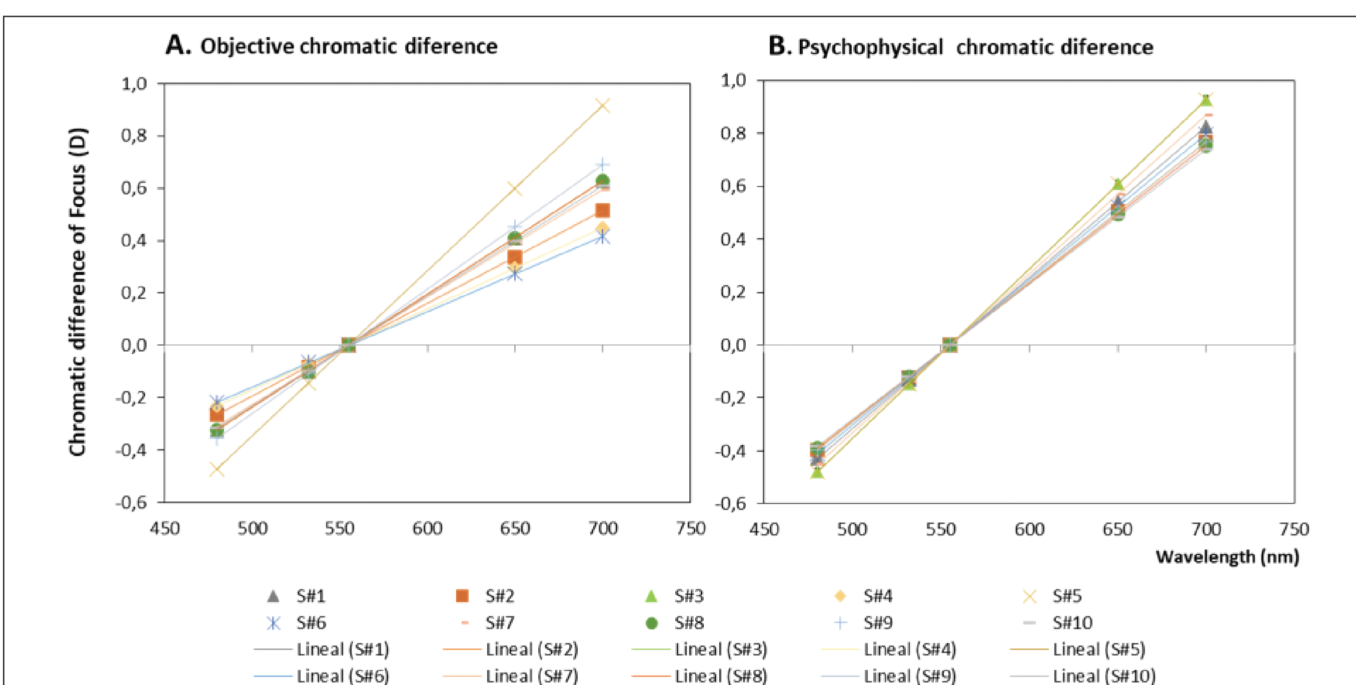

Figure C. (A) Chromatic difference of focus for all patients from Hartman–Shack (objective). (B) Chromatic difference of focus for all patients from psychophysical focus settings (psychophysical). Defocus was set to zero at 555 nm in both cases. D = diopters