Testing the effect of ocular aberrations in the perceived transverse chromatic aberration

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Abstract: We have measured the ocular transverse chromatic aberration (TCA) in 11 subjects using 2D-two-color Vernier alignment, for two pupil diameters, in a polychromatic adaptive optics (AO) system. TCA measurements were performed for two pupil diameters: for a small pupil (2-mm), referred to as ‘optical TCA’ (oTCA), and for a large pupil (6-mm), referred to ‘perceived TCA’ (pTCA). Also, the TCA was measured through both natural aberrations (HOAs) and AO-corrected aberrations. Computer simulations of pTCA incorporated longitudinal chromatic aberration (LCA), the patient’s HOAs measured with Hartmann-Shack, and the Stiles-Crawford effect (SCE), measured objectively by laser ray tracing. The oTCA and the simulated pTCA (no aberrations) were shifted nasally 1.20 arcmin and 1.40 arcmin respectively. The experimental pTCA (-0.27 arcmin horizontally and -0.62 vertically) was well predicted (81%) by simulations when both the individual HOAs and SCE were considered. Both HOAs and SCE interact with oTCA, reducing it in magnitude and changing its orientation. The results indicate that estimations of polychromatic image quality should incorporate patient’s specific data of HOAs, LCA, TCA & SCE.

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1. Introduction

In polychromatic light, the retinal image is affected by both monochromatic and chromatic aberrations of the eye and their interactions [1]. Chromatic aberrations include the Longitudinal Chromatic Aberration (LCA) and the Transverse Chromatic Aberration (TCA) [2].

LCA arises from the dependence of the refractive power of the eye with wavelength and its effect on image contrast is given by defocus [3]. There are numerous literature reports of ocular LCA, measured either with objective (reflectometric) methods [4,5] or, more extensively, through psychophysical methods [6–8]. For the adult human eye, it is well accepted that the magnitude of the LCA is ~2 D across the full visible spectrum (400 to 700 nm) [9], ~1.8 D across 450-700 nm and ~1.5 D across 480-700 nm [10].

TCA is the variation of image location and magnification as a function of wavelength, since the chief ray passing through the center of the pupil strikes on the retinal surface at different positions depending on wavelength. While the TCA is generally attributed to the lack of centration of the optical components of the eye, including the off-axis position of the fovea, other potential sources contributing to TCA have also been invoked [11]. Besides, significant differences are found in the TCA measured with centered small pupils or with dilated pupils. As double-pass techniques cancel out the TCA, TCA is generally measured psychophysically [2,12–14], except for objective reports using an adaptive optics scanning laser ophthalmoscopy [15–17]. The psychophysical measurement of TCA generally relies on alignment techniques, including different forms of a Vernier alignment task [2,13,14,18]. Simonet and Campbell [14] coined the term optical TCA (oTCA) to refer to the TCA that is measured with a small centered pupil and, therefore, not affected by the Higher Order Monochromatic Aberrations (HOAs) of the eye and by the Stiles-Crawford effect. Conversely, they used the term perceived TCA (pTCA) to refer to the
effective TCA with large pupils. It is thought that the HOAs alter the effective centroid of the retinal image, producing shifts in the Vernier alignment, while the Stiles-Crawford effect further affects the centroiding as a result of differential weights in the effective luminance efficiency across the pupil [19]. Marcos et al. [20] used a spatially resolved refractometer and a point stimulus that the patients viewed through a magenta filter (superimposed to a cross-target viewed with a small centered pupil) as a small entry pupil moved across the dilated natural pupil. The relative lateral separation between the red and blue spots with respect to the reference was taken as the TCA, for different entry pupils. The oTCA was estimated as the lateral separation between the red and blue spots for a centered pupil. The full spot diagrams for blue and red were used to calculate the pTCA (affected by the ocular aberrations, and following a weighting function, the Stiles-Crawford effect).

Knowledge of TCA is important to estimate polychromatic image quality [11,20]. TCA has also been reported [14] to play a major role on monocular chromatic diplopia and chromostereopsis [21]. Furthermore, it is well known that monochromatic and chromatic aberrations interact favorably, so that the presence of monochromatic aberrations mitigate the impact of the LCA [1,22]. Attempts to correct LCA (by means of achromatizing lenses [23], and more recently using Badal systems [24] and Spatial Light Modulators [25]) have shown little benefit in improving vision, presumably because of the residual monochromatic aberrations, neural adaptation to chromatic blur and the presence of TCA. Measuring and understanding the TCA is also important in assessing potential benefits of new released intraocular lenses aiming at correcting the LCA, while likely leaving the TCA uncorrected [26,27].

The role of HOAs in the magnitude and direction of the TCA has been predicted on the basis of the differences of measured TCA with small and large pupils [14,20]. However, to our knowledge, the impact of the ocular aberrations on pTCA has not been directly measured. Likewise, while the Stiles-Crawford effect (differences in light efficiency across the pupil arising from cone directionality) has been invoked as a contributor to shifts in the pTCA with respect to the oTCA, the relative contributions of aberrations and SCE have not been directly evaluated.

Adaptive Optics Visual Simulators allow manipulating the aberrations of the eye, in particular the correction of its high order aberrations [28–30]. In a prior study, we used the VIOBIO polychromatic AO system [28,31] to evaluate the impact of the HOAs on the LCA [10]. In this study, we implemented a new psychophysical channel in this polychromatic Adaptive Optics (AO) visual simulator to measure the TCA (both oTCA and pTCA) in normal subjects, under natural and AO-corrected aberrations. In addition, we measured the reflectometric SCE using a Laser Ray Tracing [32] which allowed evaluating the contribution of the SCE on pTCA.

2. Methods

A polychromatic Adaptive Optics system was used to measure TCA (Vernier alignment method) on 11 subjects, with small (2 mm, oTCA) and large (6 mm, pTCA) pupil diameters, and with natural (HOAs) and corrected aberrations (AO-correction). TCA measurements were performed with uncorrected LCA. Measurements of monochromatic aberrations and reflectometric Stiles-Crawford effect on these subjects were performed. Also, LCA was obtained from reflectometric/psychophysical measurements of best focus at different wavelengths on 5 of the 11 subjects. These measurements were used in computer simulations to evaluate the relative contribution of LCA, HOAs and SCE on pTCA.

2.1. Subjects

Eleven subjects (26.8 ± 3.24 years) participated in the experiment. The TCA was measured in all 11 subjects, LCA was measured in the first 5 subjects (showing values that were confirmatory and consistent with previous literature [10]), and SCE was measured in 9 of the subjects (who were available for a second session). The measurements were performed monocularly, in the
right eye of each subject. Spherical errors ranged between 0 and −6.75 D (−1.91 ± 1.30 D), and astigmatism was ≤ −1.25 D in all cases. All participants were acquainted with the nature and possible consequences of the study and provided written informed consent. All protocols met the tenets of the Declaration of Helsinki and had been previously approved by the Spanish National Research Council (CSIC) Ethical Committee.

2.2. Polychromatic adaptive optics system

Measurements of monochromatic aberrations at different wavelengths and TCA were performed in a custom-developed polychromatic AO system at the Visual Optics and Biophotonics Lab (IO-CSIC, Madrid, Spain), described partially in previous publications [10,28,31–36]. Optical aberrations were measured using a Hartmann-Shack wavefront sensor and corrected using a Deformable Mirror. The following channels of the system are used in the current experiment:

1. A illumination channel, with light coming from a supercontinuum laser source (SCLS, SC400 femtopower 1060 supercontinuum laser, Fianium Ltd, United Kingdom) [37], spanning from below 450 nm to beyond 1100 nm and in combination with a dual acousto-optics tunable filters (AOTF) module (Gooch & Housego, United Kingdom), operated by RF drivers, which is used to automatically select visible or near infrared wavelengths. In this study different wavelengths are used in the psychophysical measurements (450, 490, 532, 555, 632, 680 nm) and Hartmann-Shack measurements (490, 555, 680, 880 nm). In all cases, the maximum power reaching the eye was at least one order of magnitude below the safety limits prescribed by the American National Standard Institute for all wavelengths used in the experiments [38].

2. Hartmann-Shack wavefront sensor (microlens array 40 × 32, 3.6-mm effective diameter; HASO Imagine Eyes, France). Aberrations are fit by Zernike coefficients up to 35 terms, with notation following the recommendations of the OSA Standard Committee [39].

3. Electromagnetic deformable mirror (52 actuators, 15-mm effective diameter, 50-µm stroke; MIRAO Imagine Eyes, France), which for the purposes of this study was used to correct the aberrations of the optical system and the subjects’ HOAs. The state of the mirror that corrected for the aberrations of the system at 880 nm (IR) resulted in 0.022 µm Root Mean Square Wavefront Error (RMS) for residual aberrations.

4. Psychophysical channel, with a Digital Micro-Mirror device (DMD) (DLP Discovery 4100 0.7 XGA Texas Instruments, USA), placed in a conjugate retinal plane used to display fixation targets and psychophysical stimuli (in particular the Vernier visual stimuli to measure TCA, Section 2.3). The visual target subtends 1.62°. A holographic diffuser (HD) placed in the beam path breaks the coherence of the laser, providing a uniform illumination of the stimulus. The DMD was simultaneously illuminated (490-680 nm) with light coming from the SCLS.

5. Pupil monitoring channel, consisting of a camera (DCC1545M, High Resolution USB2.0 CMOS Camera, Thorlabs GmbH, Germany) acquiring images of the pupil. The patient’s eye was aligned to the system (using an x-y stage) while visualizing the natural pupil in a monitor, using the line of sight as a reference.

6. The Badal optometer channel corrects for defocus in AO, can be controlled automatically or with a keyboard operated by the subject to adjust her/his subjective best focus.

7. A new Transverse Chromatic aberration measurement channel, described in 2.3
All optoelectronic elements of the system (SCLS main source, Badal system, pupil camera, Hartmann-Shack and deformable mirror) are automatically controlled and synchronized using custom-built software programmed in Visual C++ and C# (Microsoft, United States).

2.3. Transverse chromatic aberration measurement channel

A new channel was developed specifically for this study to measure foveal TCA in the polychromatic AO system. The implemented method is based on two color two-dimensional Vernier alignment technique developed by Thibos et al [2].

The stimulus consisted of a red square (40 arcmin) surrounded by a blue background (120 arcmin), with black reference lines (2 arcmin) placed at fixed locations on the blue field. A movable black crossed target inside the red square was mapped on the DMD projector (Fig. 1) [18]. The DMD was simultaneously illuminated with two wavelengths (490 and 680 nm) coming from the SCLS. To split the two wavelengths and generate a stimulus with two different zone colors, we inserted a filter slide in a conjugate retinal plane. The filter was a photographic slide with two concentric squares in two colors (blue in the background and red in the central square), such that, considering the magnification of the system, it superimposed to the stimulus projected on the DMD.

Extinction tests were carried out to verify that only the corresponding wavelengths passed through each zone, and that there was no spectral leakage. The effective spectral content (combination of the SCLS and filter) was 490 ±10 nm in blue and 680 ±15 nm in red. The measured target luminance was approximately 25 cd/m² in blue and red. The luminance of the red field was adjusted by a control group of four subjects to perceptually match the luminance of the blue field, through a computational increase of the gray level of the pixels in the DMD corresponding to the region illuminated in red. The effective reduction in the luminance of the red field was by a factor of 0.02.

The vertical position of the adjustable horizontal line and the horizontal position of the adjustable vertical line in the DMD were controlled by the subject by means of the computer’s keyboard. The subject’s task was to align the black lines in both the blue and the red regions of the target, both horizontally and vertically. In the initial configuration, the vertical/horizontal black lines of the stimulus were moved with the keyboard in steps of 5 pixels. To refine the alignment, subjects had the possibility to press the central key of the keyboard to change the magnitude of the step to 1 pixel. The precision of the lateral displacements was assessed using an artificial eye (an achromatic lens with a CCD detector at its focal plane) in place of the subject’s eye. The offset was found to be 1 pixel (0.1962 arcmin) in both the horizontal and the vertical axes, which was taken as the system’s TCA, and subtracted from the values measured in the eye.
2.4. Reflectometric Stiles-Crawford effect

A Laser Ray Tracing (LRT) system, described in earlier publications [5,40] was used to measure the intensity of the light reflected back from the retina as a function of entry pupil. The LRT is normally used to measure the aberrations of the eye from measurements of the displacement of the retinal image as an entry beam moves across the pupil. An earlier study demonstrated that the intensity of the retinal images (with green light illumination) across the pupil followed a Gaussian distribution, with a peak that closely matched that obtained from dedicated photoreceptor alignment techniques [41] and the Stiles-Crawford effect [32].

A laser beam (543 nm) was scanned across the pupil with the aid of dual galvanometric scanning mirrors. A total of 37 ray entry locations (at 1-mm steps) sampled a 6-mm pupillary region in a hexagonal sampling configuration, using the pupil center as a reference. Subjects fixated a cross-target placed in a conjugate retinal plane, and the pupil was monitored on a CCD camera in conjugate pupil plane to guarantee centration. A Badal optometer corrected for the eye’s spherical refractive errors. Retinal aerial images of a spot were simultaneously captured using a cooled CCD retinal camera. The power incident on the cornea was 6.9 µW. A LRT measurement lasted approximately 1.5 s. Each measurement consisted of five consecutive runs, with a set of 37 aerial images of the retina for each run. A previous study, and additional control tests in the current study, demonstrated cone photobleaching after the first acquisition [32,42].

Dedicated routines in Matlab (Mathworks, Natick, MA) were used to estimate the intensity of the aerial images (within a 20 pixels radius circle around the centroid) and to fit the resulting intensity estimates to Gaussian distributions as a function of pupil entry position, for average intensity data across four consecutive measurements (Fig. 2). The following function was used to fit the intensity as a function of entry pupil:

\[
I(x, y) = B + I_{\text{max}} 10^{-\rho[(x-x_0)^2+(y-y_0)^2]} \tag{1}
\]

where B is a background intensity, \(I_{\text{max}}\) is the intensity at the peak, \(\rho\) is a shape factor and \(x_0\) and \(y_0\) are the pupillary coordinates of the peak.

2.5. Experiments

Measurements were obtained in two experimental sessions: Polychromatic AO system including Hartmann-Shack measurements wave aberrations at various wavelengths, TCA measurements (Session 1, 2 hours duration, performed on 11 subjects) and LRT measurements (Session 2, 20 minutes, performed on 9 subjects).

In all measurements, subjects were stabilized using a dental impression mounted on an x-y stage. The eye's pupil was aligned to the optical axis of the instruments using the line of sight as a reference, while the natural pupil is viewed on the monitor. Pupil centration was verified before each trial and between trials. Also, all measurements were performed monocularly (right eye), in a darkened room, under cycloplegia (by instillation of Tropicamide 1%, 2 drops 10 minutes prior to the beginning of the study, and 1 drop every 1 hour), and best correction of subjective defocus using Badal systems.

**Monochromatic aberrations.** For all measurements performed with the Polychromatic AO system, the best subjective focus was initially searched by the subject while viewing a Maltese cross projected on the DMD at reference wavelength of 555 nm. Ocular HOAs were measured with the Hartmann-Shack wavefront sensor (and corrected in a closed loop) at 880 nm, and at various visible wavelengths (490, 555, 680 nm) for selected subjects. All wave aberrations were obtained for 6-mm pupil diameters. Each measurement was repeated at least 5 times per condition.

**LCA measurements.** LCA was obtained from Hartmann-Shack and psychophysical measurements. In objective measurements, the best focus was obtained from the defocus term in the
Fig. 2. (A) Laser Ray Tracing retinal aerial images at entry position locations (-2,0) and (+2,0) and calculations of the aerial image intensity (integrated within the area marked by the red circle, of 20-pixel radius, centered at the position of the image maximum intensity). (B) Arial Image intensity as function of entry pupil, for an LRT series of 37 images. Each square represents average intensity across four repeated measurements. (C) Gaussian fitting of the LRT aerial image intensities (according to Eq. (1)). The asterisk marks the coordinates of the Stiles-Crawford peak. Positive horizontal coordinates stand for nasal displacements and vertical coordinates stand for superior displacements from the pupil center.

Zernike expansion at each wavelength. In psychophysical measurements, patients selected their best subjective focus using the Badal system while viewing the stimulus illuminated with three different wavelengths in visible light (490, 555, 680 nm). Each setting was repeated 5 times

**TCA measurements.** TCA was measured under natural aberrations and under AO-correction, with centered small and large pupils, in the following order: (1) Perceived TCA under natural aberrations (pTCA-HOA). TCA measurements were performed for 6-mm pupil diameters, with the best focus set a 555 nm, and with natural optics; (2) Perceived TCA AO-correction (pTCA-AO). TCA measurements were performed for 6-mm pupil diameters, with the best focus set a 555 nm, under AO-correction (closed-loop at 880 nm); (3) Optical TCA under natural aberrations (oTCA-HOA). TCA measurements were performed for 2-mm pupil diameters, with the best focus set a 555 nm, and with natural optics. (4) Optical TCA (oTCA-AO). TCA measurements were performed for 2-mm pupil diameters, with the best focus set a 555 nm, under AO-correction (closed-loop at 880 nm).

Subjects were instructed on the Vernier Alignment task and on the use of the keyboard to shift the horizontal and vertical bars of the cross-target. After an initial trial run, subjects repeated the Vernier alignment 5 times per condition. Aberrations were monitored throughout the experiment every two trials to ensure that each trial was performed under the desired state of aberrations correction.

The small and large pupil diameters (2 and 6 mm) were achieved with an artificial iris (3D printed) placed in a conjugated pupil plane, centered with the help of the pupil monitoring camera.

**SCE measurements.** Five consecutive series of 37 retinal aerial images were captured scanning the subject’s pupil, with foveal fixation. Background images (replacing the subject’s eye by a black target) were obtained and subtracted from the test images.
2.6. Computational analysis of perceived transverse chromatic aberration

Point Spread Functions (PSF) in blue (490 nm) and red (680 nm) were calculated from the measured HOA aberrations (at the specific measured wavelengths—in the patients in which they were available—or at 880 nm), and the corresponding chromatic difference of focus between 555 nm (for which the best focus was found) and 490 nm (0.55 D) and between 555 nm and 680 nm (0.90 D) [10]. The same LCA values were used for all subjects. PSFs were calculated for each subject for the measured natural HOAs, or the residual aberrations after AO-correction, both for 2-mm and 6-mm, using Fourier Optics. The modulus of the pupil function was taken either as constant or with a weighted efficiency by the 2D-Gaussian function (normalized Eq. (1)) representing the SCE. For simplicity, a constant ρ of 0.1 mm$^{-2}$ was used in Eq. (1) for all subjects [43].

To simulate a Vernier alignment test, Line Spread Functions (LSF) were calculated for horizontal and vertical directions, by integrating the energy in the PSF in both directions. The process is outlined in Fig. 3. The shift between the peak of the LSF for blue and for red in the horizontal or vertical directions accounts for the shift in the coordinates of the TCA arising from the presence of aberrations only, or aberrations and SCE [44]. The offset coordinates were vectorially added to the $\delta$TCA-AO (2-mm pupil, with corrected aberrations), to generate the computational pTCA-HOA (6 mm, effect of aberrations), and pTCA-HOA-SCE (6 mm, effect of aberrations and Stiles Crawford), respectively.

Fig. 3. Illustration of the calculations of the computational pTCA-HOA obtained from estimated Point Spread Functions (PSFs) and Line Spread Functions (LSFx,LSFy) in red and blue for 6-mm pupils, vectorially added to the measured $\delta$TCA-AO (2 mm, AO-correction). Calculations of computational pTCA-HOA-SCE and pTCA-AO-SCE are performed similarly, including the SCE (simulated applying a Gaussian pupil mask), with or without aberrations respectively, to calculate the corresponding LSFx,LSFy to add to the $\delta$TCA-AO.

A similar approach was followed, to isolate the effect of SCE, calculating the PSF (and corresponding LSF) for the AO-corrected condition (residual ocular aberrations, 6 mm) and the SCE Gaussian pupil mask. LSFs and offset coordinates were calculated and vectorially added to the $\delta$TCA-AO (2-mm, with corrected aberrations), to generate the computational pTCA-AO-SCE (6 mm, SCE).
2.7. Data analysis

Statistical analysis was performed with SPSS software Statistics 24.0 (IBM, United States). We tested the dependency of HOAs on wavelength (one-way ANOVA) and evaluated the differences between pTCA and oTCA, and AO and HOAs conditions (paired-samples t-test).

3. Results

3.1. Wave aberration at different wavelengths and AO-correction

Ocular aberrations were measured in three wavelengths in the visible (490, 555, 680 nm) and in the IR (880 nm). We checked that, except for defocus, there was a negligible dependency of the HOAs on wavelength. In particular, we ensured that the AO-correction (which was obtained in a closed-loop in IR), was also efficient at lower wavelengths. Figure 4(A) shows an example of wave aberrations for HOA, at all measured wavelengths, for natural aberrations (upper row) and AO-correction (lower row), for one subject (S#3). Wavefront maps are similar, with no systemic change, across wavelengths. Figure 4(B) shows the average RMS at each wavelength, for natural aberrations and AO-correction, across all measured subjects. On average, the RMS standard deviation across wavelengths was 0.04 µm for natural aberrations and 0.01 µm for AO-correction. There was no systematic variation of the RMS with wavelength (neither in natural nor in AO-corrected conditions) (one-way ANOVA, p = 0.484 & p = 0.108).

From the defocus Zernike term (not shown in Fig. 3) we could also calculate the chromatic difference of focus between 490 and 680 nm (for objective/reflectometric LCA), which on average was 0.74 ± 0.01 D (with natural aberrations) and 0.74 ± 0.01 D (under AO-correction).

3.2. Psychophysical LCA

Figure 5 shows the psychophysical LCA for all measured subjects, obtained from subjective focus settings between 490-and 680 nm. The LCA was obtained from a linear fitting to the chromatic defocus as a function of wavelength. LCA was 1.41 ± 0.08 D with natural aberrations and 1.39 ± 0.12 D under AO-correction.

3.3. Experimental Optical and Perceived TCA: Impact of AO-correction of HOAs

Figure 6 shows the horizontal and vertical coordinates of the optical TCA (oTCA, 2 mm, Fig. 6(A)) and perceived (pTCA, 6 mm, Fig. 6(B)) measured experimentally in the 11 subjects of the study. Positive horizontal coordinates stand for nasal displacement in the retina and positive vertical
coordinates stand for superior displacement in the retina. The average oTCA (2 mm) horizontal coordinates were $1.29 \pm 0.10$ arcmin (HOA) and $1.13 \pm 0.14$ arcmin (AO-correction), and the average oTCA vertical coordinates were $-0.11 \pm 0.08$ arcmin (HOA) and $-0.16 \pm 0.12$ arcmin (AO-correction), coordinates that are represented by small circles in Fig. 6(C) for measurements obtained under natural HOA (red) and AO-correction (green). The average pTCA horizontal coordinates were $0.11 \pm 0.07$ arcmin (HOA) and $-0.28 \pm 0.08$ arcmin (AO correction), and the average pTCA vertical coordinates were $-0.65 \pm 0.11$ arcmin (HOA) and $-1.01 \pm 0.08$ arcmin (AO correction), represented by the large circles in Fig. 6(C)). Figure 6(D) shows the average absolute shift in TCA when changing the pupil diameter from 2 to 6 mm (pTCA-oTCA), for natural aberrations ($0.72 \pm 0.11$ arcmin, red bar) and AO-correction ($1.24 \pm 0.08$ arcmin, green bar). Figure 6(D) also shows the average absolute shift when correcting the HOA aberrations for pTCA (i.e 6 mm pupil, $1.09 \pm 0.11$ arcmin, solid blue bar) and for oTCA (i.e. 2 mm, $0.23 \pm 0.14$ arcmin, patterned blue bar). The differences between oTCA_HOA and oTCA_AO ($0.23$ arcmin) are within the order of the resolution of the technique ($0.1962$ arcmin), but the differences between pTCA_HOA and pTCA_AO are statistically significant (p-value ranging between $0.001$ and $0.03$, except for subjects S#1 and S#7). Also, increasing pupil diameter from 2 to 6 mm (oTCA vs pTCA) produces a significant shift in the TCA, both for natural aberrations (p = 0.046) and AO-correction (p = 0.031).

### 3.4. Reflectometric SCE

Figure 7(A) shows the reflectometric SCE maps (2D-Gaussian fittings) in all measured subjects. The SCE peak positions are marked with black asterisks. Figure 7(B) shows the coordinates of the SCE peaks for all subjects. On average, the SCE peak location lies at $0.89 \pm 0.45$ mm nasally and $-0.20 \pm 0.36$ mm inferiorly from the geometric center of the pupil.

### 3.5. Computational perceived TCA

Figure 8 shows an example of the computation of perceived TCA (for various conditions) for one subject (S#3). Figure 8(A) shows the estimated shifts of the peak of the blue and red LSFs by the effect of HOAs and SCE (taking the experimental value of oTCA-AO as a reference). The oTCA-AO is represented by the distance between the small red circle (arbitrarily placed at 0.0) and the small blue circle. The shifts in the peak of the LSFs for red and blue are then added to those reference values for blue and red, considering the effect of HOAs (represented by “x”), considering both HOAs and SCE (represented by “+”), and considering the effect of SCE only (represented by “o”). Figure 8(B) shows the computational perceived TCA for S#3, estimated from the distances between the estimated blue and red peaks: pTCA-HOA being the distance between the blue and red “x” in Fig. 8(A), pTCA-HOA-SCE being the distance between the blue and red “+”, and pTCA-SCE being the distance between the blue and red “o”. The
Fig. 6. (A) Optical TCA (oTCA, 2 mm), with natural aberrations (oTCA-HOA, small red circles) and AO-correction (oTCA-AO, small green circles), for all subjects. Error bars stand for standard deviation of repeated measurements. (B) Perceived TCA (pTCA, 6 mm), with natural aberrations (pTCA-HOA, large red circles) and AO-correction (pTCA-AO, large green circles), for all subjects. Error bars stand for standard deviation of repeated measurements. Note: One subject not shown because of falling outside the axis range (-4 to 4 arc min). (C) Average (across subjects) coordinates for oTCA (small circles) and pTCA (large circles), for natural aberrations (red symbols) and AO-correction (green symbols). Error bars stand for standard deviations across subjects. (D) Absolute TCA magnitude differences, for different conditions: changing the pupil size (pTCA-oTCA) under natural aberration (red solid bar), AO-correction (green solid bar); correcting aberrations, for a given pupil diameter (6 mm, pTCA-HOA – pTCA-AO, blue solid bar and 2 mm, oTCA-HOA – oTCA-AO, blue patterned bar). Error bars stand for standard deviations across subjects.

The graph also shows the experimental TCA measurements (pTCA-HOA, pTCA-AO, oTCA-HOA and oTCA-AO) as solid circles. The computational pTCA-HOA and pTCA-HOA-SCE are to be compared with the experimental pTCA-HOA. With natural aberrations, it is expected that both HOA and SCE contribute to the difference between pTCA and oTCA. The computational pTCA-AO-SCE is to be compared with pTCA-AO. Under corrected aberrations, it is expected that residual HOA and SCE contributes to the difference between pTCA-AO and oTCA-AO.

Figure 9 compares the experimental and computational pTCA in all subjects: the experimental pTCA-HOA and the computational pTCA considering HOAs only (Computational pTCA-HOA), shown in Fig. 9(A), and the experimental pTCA-HOA and the computational pTCA considering both HOA and SCE (Computational pTCA-HOA-SCE), shown in Fig. 9(B). The computational pTCA-HOA presents more spread values than the experimental measurement (average difference computational-experimental 2.04 arcmin). However, the computational pTCA-HOA-SCE captures both the orientation and magnitude of the experimental measurements (average difference computational-experimental 0.37 arcmin, close to the resolution of the technique). Figure 9(C) shows the experimental pTCA-AO (i.e. with HOA aberrations corrected) and the computational pTCA-SCE, with AO-correction but considering the SCE (average
Fig. 7. A. 2D-Gaussian fitting of the LRT aerial image intensities (according to Eq.1) for all subjects. The asterisk marks the coordinates of the Stiles-Crawford peak. Positive horizontal coordinates stand for nasal displacements and vertical coordinates stand for superior displacements from the pupil center. The x- and y- axis ranges (pupil coordinates) are shown for S#1, and are similar for all subjects. B. SCE peak positions in nine eyes of the study positive horizontal coordinates stand for nasal displacement, and positive vertical coordinates for superior displacement. Each eye is depicted by different symbol shape. Error bars (standard deviations of estimations from five repeated LRT series) are smaller than the symbols (0.04 ± 0.01 mm).

Fig. 8. Illustration for one subject (S#3) of the estimated effects of HOA and SCE on the perceived TCA. (A) Estimated shifts in the peak of the LSF in the horizontal and vertical coordinates for blue and red when considering the HOA and SCE, with respect to the LSF peaks for blue and red in absence of HOA and SCE, taken at the 0,0 coordinate for red and shifted by the experimental value of oTCA-AO for blue. The coordinate shift by HOA is represented by the symbol “×”, the coordinate shift by HOA + SCE is represented by “+”, and the coordinate shift by SCE is represented by “o”. (B) Computational pTCA, calculated from the vectorial difference of the blue and red peak coordinates: pTCA-HOA from the difference of the coordinates represented by “×”, pTCA-HOA-SCE from the difference of the coordinates represented by “+”, pTCA-AO-SCE from the difference of the coordinates represented by “o”. The experimental oTCA-HOA, oTCA-AO, pTCA-HOA and pTCA-AO are also shown.
difference computational-experimental 0.58 arcmin). Figure 9(D) shows the experimental pTCA-HOA, pTCA-AO and oTCA-AO, and computational pTCA-HOA, pTCA-HOA-SCE, pTCA-SCE and oTCA-AO, averaged across subjects.

**Fig. 9.** (A) Experimental pTCA-HOA (large red circles) and computational pTCA-HOA (HOAs only, represented by red “×”). (B) Experimental pTCA-HOA (large red circles) and computational pTCA-HOA-SCE (HOAs and SCE, represented by red “+”). (C) Experimental pTCA-AO (large green circle) and computational pTCA-SCE (no HOAs, SCE only, represented by green ◦). The gray lines link the symbols of experimental and computational data for each subject (in A, B, C). (D) Experimental pTCA-HOA, pTCA-AO and oTCA-AO and computational pTCA-HOA, pTCA-HOA-SCE, pTCA-AO and pTCA-AO-SCE, averaged across subjects. In all the graphs, the red symbols correspond to the HOAs condition, and the green symbols correspond to the AO-correction. Error bars stand for standard Errors across subjects (for horizontal and vertical coordinates).

### 4. Discussion

Using a Polychromatic Adaptive Optics Visual Simulator, we measured monochromatic aberrations, longitudinal chromatic aberrations and Transverse Chromatic Aberrations in normal human subjects, under natural and corrected monochromatic aberrations. We found that the TCA was systematically shifted nasally for small pupils and that the presence of monochromatic aberrations produced an intersubject spread of the perceived TCA. Correcting HOAs in fact increased the perceived TCA in most subjects. Measurements of the Stiles-Crawford effect in
addition to monochromatic aberrations and LCA allowed predictions of the relative impact of SCE and aberrations on the perceived TCA. We found that incorporating an apodized pupil given by the individual SCE produced accurate predictions of the experimental perceived TCA.

The polychromatic AO system allows the correction of aberrations while performing psychophysical measurements of LCA and TCA. Even if the closed loop is performed while aberrations are measured and corrected at 880 nm, the level of monochromatic aberration correction is as high in the visible as in the IR wavelengths (corrected wave aberration RMS = 0.01 µm, for 6-mm). As previously found in the literature, we found that the reflectometric LCA obtained from the defocus term of the Zernike expansion to the Hartmann-Shack measured wave aberrations (+0.74 ± 0.01 D) was consistently lower than the psychophysical LCA (+1.39 ± 0.10 D) in a range of 490-680 nm. In all cases, the magnitude of the measured LCA magnitude is independent of the presence or correction of natural aberrations. These results are in line with reports from Vinas et al. of reflectometric/psychophysical LCA with HOAs and AO-correction [10] and also consistent with known values for LCA [2,4].

The measured TCA values (490-680 nm spectral range) were of the same order of magnitude as other reported TCA, psychophysically measured through different versions of Vernier alignment tasks. The oTCA (small pupil) measured in this study ranged from -0.3 arcmin temporal to +2.24 arcmin nasal (horizontally) and -1.10 arcmin inferior to +1.33 arcmin superior (vertically) and pTCA (large pupil) ranged from -1.40 arcmin temporal to +1.70 arcmin nasal (horizontally) and -2.20 arcmin inferior to +0.47 arcmin superior (vertically), for natural aberrations (HOAs). As in other studies, TCA was measured while leaving LCA uncorrected. In our study, that also applied to measurements under AO-correction. We have compared the standard deviations values in the Vernier alignment for both conditions (HOA and AO-correction), the value is low for both conditions 0.09 ± 0.01 arc min, so for the subjects the task was performed with the same challenge, and any observed difference can be attributed to the TCA magnitude alone, and its dependence on the actual measurement condition.

The reported magnitudes of TCA varied across studies, likely due in part to specific measurement conditions. For example, Ogboso and Bedell [13] reported TCA between 0.9 to 3.0 arcmin nasal (435-572 nm) and Rynders et al 0.05 to 2.67 arcmin (497-605 nm) under natural conditions; Simonet and Campbell [14] reported 0.25 arcmin temporal to 1.56 arcmin nasal (486-656 nm) under Maxwellian view [18,45]. Thibos et al. [2], using a method that involved displacing a pupil, estimated foveal TCAs (433-622 nm) of 0.36 arcmin temporal to 1.67 arcmin nasal for natural pupils. Marcos et al. [20], using a spatially resolved refractometer in combination with a magenta filter (473-601 nm), reported TCA ranging from -3.77 to +0.20 arcmin (horizontal) and -5.03 to +0.11 arcmin (vertical) for small centered pupils and estimated TCA ranging from -3.29 to 0.22 arcmin (horizontal) and +3.74 to +1.99 arcmin (vertical) for large pupils. Reported objective foveal TCA measurements using adaptive optics scanning laser ophthalmoscope (AOSLO) [15] were 2.1 arcmin on average. Despite changes in the spectral range, technique used and pupil diameter, a common conclusion of all the studies is the large intersubject variability of TCA within each sample.

Our study found TCA included 4 different experimental conditions (large and small pupils, HOA and AO-correction, in all subjects) and the values were within the range reported in the literature. However, our intersubject variability of oTCA (0.24 arcmin for oTCA-HOA and 0.28 arcmin for oTCA-AO) is much lower in our sample than previously reported. The higher magnitude in the horizontal (+1.36 ± 0.10 arcmin nasal) than in the vertical (-0.16 ± 0.08 arcmin inferior) oTCA coordinates that we found, is consistent with previous findings, and may arise from the mostly horizontal off-axis position of the fovea [46,47]. We did not attempt to measure angle kappa in these subjects, but we could hypothesize that low intersubject differences in angle kappa may be the cause of the similarity of oTCA across subjects.
Previous studies suggested a potential contribution of high order aberrations and Stiles-Crawford effect to the magnitude, orientation and intersubject variability of the perceived Transverse Chromatic Aberration [21,48,49]. These factors have been attributed to play a major role in the differences in chromosteropsis (depth perception of 2-dimensional blue-red images) across subjects and with pupil size. An earlier publication using a spatially resolved refractometer [20] attempted prediction of the change in TCA magnitude and orientation when the pupil diameter increases. The calculations were performed on two subjects, shifting the lateral positions of blue and red from those corresponding to the centered pupil to the centroid of the respective spot diagrams in red and blue. In a second calculation, the spots were weighted by the pupil luminous efficiency (SCE). Changes in sign TCA were observed both horizontally and vertically.

To our knowledge, the current study represents the first attempt to test directly the impact of aberrations on TCA, performing measurements with natural aberrations and corrected aberrations (with adaptive optics) both for small and large pupils. It also presents the first direct comparison of experimental measurements of perceived TCA with computational predictions of perceived TCA, taking into account the high order aberrations of the patient (measured in the same adaptive optics set-up) and their individual Stiles-Crawford (measured through a reflectometric technique, LRT).

The small difference (0.23 ± 0.15 arcmin) between the TCA measured for small pupils between the natural aberrations (OTCA-HOA) and AO-corrected aberrations (OTCA-AO) indicates a small contribution of the high order aberrations in OTCA measured with 2-mm pupils. This small difference in the RMS values (RMS_HOA = 0.08 ± 0.03 µm and RMS_AO = 0.05 ± 0.01 µm) justifies the small difference in the modulus of the OTCA between both conditions (HOA / AO). When the pupil is increased from 2 to 6 mm, there is a slight decrease in the TCA magnitude (particularly for pTCA-HOA) in both meridians, but most remarkably (both for pTCA-HOA and pTCA-AO) there is large change in TCA orientation, shifting from nasal to an arbitrary pattern that changes horizontally (nasal-temporal) and vertically (superior-inferior). A potential shift in the pupil center with pupil dilation, often invoked as a factor playing a role in the differences between OTCA and pTCA [14], can be applied here, because the pupil was concentrically changed using an artificial iris, but the physical center of the pupil may change with dilation. Also, the relatively larger magnitude of pTCA-AO than pTCA-HOA indicates that correcting monochromatic aberrations does not necessarily imply a reduction on pTCA, and in fact it may accentuate its effect. The current findings therefore suggests that we may extend the observation that the presence of monochromatic optical aberrations protects vision against longitudinal chromatic defocus [1] to a protection against the transverse chromatic aberration as well. While a decrease visual benefit of correcting the eye’s optics is not surprising given the neural adaptation to the subjects native aberrations [50], the fact that these effects can be predicted by optical simulations suggests that these effects rely, at least in part, on optical grounds.

The realistic computational simulations of pTCA allow understanding the factors contributing to pTCA. A calculation of the pTCA taking into account high order aberrations predicts a spread of pTCA values and account for 31% of the variance; r²=0.31 (in the experimental pTCA-HOA vs computational pTCA-HOA correlation). However, when the SCE was incorporated, the computational pTCA-HOA-SCE values match highly the experimental values (i.e. HOA and SCE account for 81% of the variance; r²=0.81 in the experimental pTCA-HOA vs computational pTCA-HOA-SCE correlation, and an offset of only 0.37 ± 0.2 arcmin). Our experimental values of SCE peaks (0.89 ± 0.45 mm horizontal and -0.20 ± 0.36 mm vertical, on average) are in good agreement with previous reports in the literature, which show that SCE peaks are in general in the nasal-inferior pupil [42,51,52], although intersubject –and even intrasubject- variability occurs [53]. In fact, we found that the average SCE peak that minimizes the experimental/computational difference (0.94 ± 0.59 mm nasal and -0.59 ± 0.49 mm inferior) is very close to our SCE peak
experimental value. Further adjustments to the impact of SCE can be made by modulating the shape factor of the effect (we assumed \( \rho = 0.1 \text{ mm}^{-2} \), a value generally reported for reflectometric data, but this may differ from psychophysical values and may vary across individuals [42]). It appears that the nasal coordinate of the oTCA is offset by the increase in the luminous efficiency of the pupil towards the nasal side (where the achromatic axis must lie), resulting in a horizontal shift towards the center of the pTCA horizontal coordinate. The vertical inferior shift of the pTCA vertical coordinate may also be associated to the inferior shift of the SCE peak. The bias by the aberrations may also contribute in this direction, at least in some subjects. The relevance of the SCE effect on pTCA is further supported by the experimental pTCA-AO results. Despite the minimization of HOAs, pTCA values are spread, and shifted from the oTCA (a condition that given the smaller pupil, also minimizes the contribution of the HOAs). However, when SCE is uniquely considered (computational pTCA-SCE), the pTCA-AO is better predicted (40% of the variance; \( r^2 = 0.40 \) in the experimental pTCA-AO vs computational pTCA-SCE correlation).

Experimental measurements of monochromatic aberrations, longitudinal chromatic aberration and transverse chromatic aberration allow full estimation of the polychromatic image quality. These measurement channels are all available in a Polychromatic Adaptive Optics system, which allows correcting the monochromatic aberrations while performing psychophysical measurements, in particular to assess the impact of chromatic aberrations on vision. Previous studies have shown that correcting high order aberrations in fact eliminates the protection of the eye against chromatic blur [1,22]. Here we show that correcting high order aberrations does not reduce perceived chromatic aberration (with large pupils). Furthermore, simulations incorporating the SCE indicate that SCE pays an important role in determining the perceived TCA. It appears that a critical combination of ocular misalignment (likely driving oTCA to a large extent), optical aberrations and pupil luminous efficiency (Stiles-Crawford peak) optimize the perceived TCA. We may conclude that, while SCE may have a secondary effect in the monochromatic Modulation Transfer Function (MTF), it may be critical to predict polychromatic image quality with large pupils, both in calculations that use wavefront, LCA and oTCA, or ray tracing on fully anatomically-based eye models [54,55]. An open question is the extent to which subjects may be adapted to their own transverse chromatic blur, which may be altered by changes in the optics by disease, treatment or optical correction. Furthermore, the consideration of TCA becomes relevant with the release of new corrections (i.e. in the form of diffractive IOLs) that aim at modulating Longitudinal Chromatic Aberration, at least at some distances [56–58].

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