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Eye-tracking technology for real-time monitoring of transverse chromatic aberration

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Abstract

Objective measurements of transverse chromatic aberration (TCA) between two or more wavelengths with an adaptive optics scanning laser ophthalmoscope (AOSLO) are very accurate, but frequent measurements are impractical in many experimental settings. Here, we demonstrate a pupil-tracker that can accurately measure relative changes in TCA that are caused by small shifts in the pupil relative to the AOSLO imaging beam. Corrections for TCA caused by these shifts improve the measurement of TCA as a function of eccentricity, revealing a strong linear relationship. We propose that pupil tracking be integrated into AOSLO systems where robust and unobtrusive control of TCA is required.

It is well known that light travelling in the ocular media undergoes chromatic dispersion. A green and a red beam for example, projected in Maxwellian view through the center of the pupil along the poles of the achromatic axis, are focused in different focal planes resulting in different refractive powers, referred to as longitudinal chromatic aberration (LCA). If the eye rotates creating an angle between the beam's chief ray and the achromatic axis, or if the incident beam is laterally displaced from the achromatic axis, then an additional transverse chromatic aberration (TCA) is generated causing a lateral displacement between the image locations of the two wavelengths on the retina.

LCA has been measured in the human eye and, although some variability has been reported, (some of which is attributed to the experimental method used to measure it), there is a general consensus that LCA stays rather constant across the population, it is not affected by pupil position or eccentricity and it can be optically corrected with a fixed adjustment of the vergences of the different wavelengths [1,2,3,4,5,6].

TCA poses a more challenging problem. Its magnitude depends strongly on the position of the source in the visual field which defines the angle of the incident ray bundle and the

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centration of this bundle within the pupil [3,4,7]. The present study is concerned with the effect of pupil location.

Many studies have measured TCA using psychophysics experiments based on two-color Vernier alignment tasks or chromatic parallax techniques. Foveal TCA for example, has been reported to range between approximately 1 arcmin [3,4,7] and up to 3 arcmin [8]. There are only two reports where subjective tests have been used to estimate TCA outside of the fovea [8,9]. The change of TCA as a function of the light bundle's lateral position in the entrance pupil is governed by a linear relationship (see Figure 4 in [3]). Using a pinhole aperture in front of the eye, for example, it is possible to control the angle of incidence of the foveal chief ray and thus to quantify subjectively the magnitude of TCA. In a two-color Vernier alignment task experiment, the TCA induced by pinhole displacement showed this linear relationship (see Figure 4 in [4]). Other support for this linearity come from studies with model eyes (as reported by the analytical model in [10]) and our own data, see Figure 1). Given this linear relationship between the off-axis shift of the light bundle entering the pupil and TCA, a measure of the lateral pupil position with respect to the light beam may be sufficient to measure relative changes in the TCA objectively.

Recently, an adaptive optics scanning laser ophthalmoscope (AOSLO) [11,12] which allows the delivery of multi-chromatic beams directly to the retina and the simultaneous imaging of the retinal structure at the microscopic scale [13], has been shown to be capable of making objective TCA measurements. The first objective measures of TCA were achieved by comparing the relative frame displacement of a dual-frame video taken simultaneously at two or more wavelengths [14,15]. A second, more efficient and accurate approach, used a line-by-line interleaving technique [16]. In both cases, TCA was manifest as a retinal structure offset in the de-interleaved (or pairs of) images at different wavelengths and it could be quantified using image registration [17]. Using the latter approach, TCA was quantified with a level of accuracy within 2.6 arcsec of visual angle, less than one-tenth the diameter of the smallest cones in the fovea [16].

However, the AOSLO's image-based approach to measure TCA is typically avoided in many experimental situations, as it requires bright imaging illumination which causes photoreceptor bleaching, light adaptation and some discomfort. The imaging light alters the visual sensitivity of retina, which is often the very property of vision we are aiming to measure. A less obtrusive approach to estimating TCA is desired, one which is covert, dynamic, independent of the quality of the retinal image and robust to small changes in pupil location and fixation. In this paper we explore a pupil-tracking-based approach to estimating the TCA in order to meet these criteria.

We equipped our AOSLO with a CMOS (complementary metal-oxide semiconductor) video camera, mounted near the subject's head, oriented and focused on the subject's eye. We collected image-based TCA in three subjects at different eccentricities in the field of view while simultaneously recording pupil and retinal videos.

The University of California Berkeley institutional review board approved this research and all subjects signed an informed consent before their participation in this study. All

procedures involving human subjects were in accordance with the tenets of the Declaration of Helsinki. Three subjects with normal color vision participated in the experiments. Mydriasis and cycloplegia was achieved with one drop of 1% tropicamide solution administrated to the right eye approximately 30 minutes before the experiment.

The head was stabilized using a bite-bar mounted on an X-Y-Z translation stage. A small light point was used for fixation. In the first pilot measurement for subject 1, fixation was maintained at 2.5 degrees temporally (directions used throughout this paper will be in visual field space). The head (and thus the pupil centration) was shifted vertically and horizontally by small increments of approximately 0.3 mm using the fine adjustment knob connected to the chinrest. All three subjects participated in the second measurement, where pupil centration was maintained but fixation direction was shifted in each trial, in 2.5 degree steps along the nasal, temporal, inferior and superior directions. The eye video-camera was used by the operator to maintain the centration of subject's pupil relative to the beam for all viewing directions. A bulls-eye target was graphically overlaid on the video to aid this repositioning.

At each eccentricity, two co-aligned, beams of 543 nm (green) and 842nm (IR), were projected on the retina for imaging. Although the effects of TCA are cancelled on the second pass through the dispersing ocular media, the chromatic offset could be determined by registering the retinal structures in the two interleaved (green and IR) images [16] using a registration algorithm with sub-pixel precision based on discrete Fourier transform, dFT [17]. LCA was corrected by a static adjustment of the relative vergence of the infrared and green beams using a theoretical fit to approximate a human population sample [6]. The LCA was further optimized by a through-focus assessment of the images obtained from individual infrared and green illumination channels.

Live videos from the CMOS video camera were acquired during the TCA experiments and processed using the Matlab image acquisition toolbox. Any remaining tracking artifacts were detected and corrected offline. Eye tracking was implemented with a standard pupil edge detection algorithm and ellipse fitting similarly to that discussed in [18]. The center of the pupil (px, py) corresponded to the center of the fitting ellipse and was evaluated for each video frame. Other eye tracking techniques based on tracking of the bright corneal reflection (or glint) generated by the infrared beam [19] were evaluated without achieving the same accuracy and reliability of tracking. This was probably due to the size and shape of the glint which is considerably larger and more diffused (and thus more difficult to center and track) in our AOSLO system compared to the typical LEDs-generated reflections in many commercially available eye trackers.

First, we sought to determine if pupil centration was predictive of objectively measured TCA. Subject 1 was instructed to fixate at approximately 2.5 degree in the visual field temporal to the AOSLO axis; the head (and thus the pupil centration) was moved vertically and horizontally over fifteen different positions using the fine adjustment knob connected to the chinrest. Figure 1, upper left panel, shows the coordinates of pupil centers relative to the AOSLO light beam. The TCA measured for each of these 15 positions using the AOSLO is reported in Figure 1 upper right panel. The spatial distribution of the two set of points shows

a remarkable 2D linear correspondence. Data are also plotted in the same graph to show the strong horizontal and vertical linear correlation (Figure 1, lower panels). An averaged linear ratio of $\rho = 3.5$ arcmin of TCA per mm of pupil shift was calculated and used in the second experiment.

The second measurement was designed to simulate a standard experimental setting for an AOSLO measurement. Head and thus pupil centration was maintained by the operator for each trial and TCA was measured at each fixational point along the four directions. Three subjects participated. An example for Subject 1 along the nasal direction is reported (Figure 2, top). Here, TCA graphed for each fixation position shows a linear relationship with eccentricity as expected by the theory (see Introduction). The error in pupil centration corresponding to each TCA estimate and generated by the operator during the positioning of the subjects, is also reported for the same subject and the four directions (plotted with different colors, Figure 3, left).

Using the ratio ρ from the previous measurement to compensate for this error, (i.e. to correct the measured TCA by the amount of the pupil error decentration), the expected linear relationship improved significantly (Figure 2, bottom). This same level of improvement applied to all subjects and directions.

To quantify the TCA error caused by pupil shifts, we compute the residual variance around the correlation line – in the examples provided for the nasal direction and Subject 1 (Figure 2,) this error improves significantly with pupil compensation. Figure 3, right, plots the TCA error for all subjects and directions and, again, the level of improvement can be appreciated by comparing the two distributions before (blue) and after (red) the TCA compensation. A potential maximum error of more than 2 arcmin is reduced, after compensation, by more than half (reduction in SD from 0.71 to 0.30, $F= 5.3$, $p < 0.001$).

The recent development of the AOSLO system, allowing microscopic imaging of the photoreceptor mosaic of the retina and the simultaneous delivery of multi-chromatic light stimuli at a cone-size scale, has opened exciting new frontiers in the study of human retinal cells. The limitations of the human eye optics are a critical parameter since imaging and light stimulation are often defined in different wavelength domains and chromatic aberration is consequently generated. In a condition similar to the one considered in this study for example, if imaging is infrared at 842nm and stimulus is green at 543nm, assuming that the longitudinal component LCA can be corrected by adjusting the stimuli vergence, TCA can generate a critical error in the range of several arcmin (see for example the distribution of the TCA error, Figure 3, right). This could be very problematic considering that the size of a cone ranges approximately from 0.5 arcmin (corresponding to a cone separation of about 0.5 arcmin) in the fovea up to 2 arcmin (3 arcmin cone separation) at 10 degree eccentricity in the periphery [20, 21]. This constitutes the static TCA, which can be corrected before each experimental run. The dynamic TCA, i.e. that arising from the small pupil displacements while fixation, can be a few hundreds of microns, corresponding to approximately 0.9 arcmin of TCA.

It is important to note here that the dependence of TCA on pupil position applies only to systems where the beam diameter is smaller than the pupil. If the incident beam is bigger than the pupil, then as the head moves the TCA would be unchanged. However, using the subject's maximum pupil size for an adaptive optics ophthalmoscope is not optimal or practical in many imaging situations and so most AO systems use a fixed pupil size, restricted by some other artificial aperture in the system, and the eye's pupil is generally underfilled.

The AOSLO system is, to date, the only objective method demonstrated to quantify TCA. It is also very accurate, as microscopic imaging of the same area in the retina simultaneously with two different interleaved wavelengths allows measurement of aberration directly at the retina [16].

During a typical experiment with the AOSLO, TCA (of the eye plus the AOSLO system) is calculated at the beginning, for the specific eccentricity required, and used as a static reference for controlling stimulus delivery [22]. It is unrealistic to repeat TCA evaluation during the experiment. However, even with the most trained and cooperative subjects, pupil centration and fixation stability cannot be guaranteed and thus the original reference value of the TCA cannot hold precisely through the entire duration of the experiment.

A video-based eye/pupil gaze tracking system can serve the purpose to monitor and control TCA. Our data showed that by applying a linear TCA correction, compensating by a ratio ρ the experimental pupil centration error (or shift during the measurement), we can achieve an important error reduction of TCA. Although we don't expect this ratio ρ to be significantly different, it could be easily recalibrated for each specific subject and for each retinal location prior to an AOSLO experiment.

It is important to note that all TCA values in this report include the AOSLO system's inherent TCA due to the offset and misalignment between the two wavelength channels in the system; in this sense, our data does not report the true TCA of the eye. A separate study of the actual TCA in the human eye as a function of eccentricity is in progress.

In summary, the nature and behavior of the pupil center compensation agrees with the general rationale reported in previous TCA studies. Based on the results in this manuscript, we propose that the eye/pupil tracking technology described here can be easily integrated in the experimental control loop for AOSLO or any ophthalmic system that employs multiple wavelength channels. The system can be used to monitor and eventually correct the subject's pupil position relative to the axis of the ophthalmoscope. With the continuous improvement of CMOS technology, there are now commercially available high quality sensors with over several Gpixel/sec or thousand fps. It is thus plausible to track the eye at a very high speed and reach a time resolution that could be fed back and coupled with the stabilized video imagery mechanism of the AOSLO to obtain real-time TCA compensation. But even a lower frame rate, TCA correction can be useful. For example, in experiments involving multiple brief presentations of a visible light flash, the TCA could be adjusted for pupil shifts prior to each delivery.

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Roorda has two patents on technology related to the Adaptive Optics Scanning Laser Ophthalmoscope USPTO #7,118,216 #6,890,076. These patents are assigned to both the University of Rochester and the University of Houston. The patents are currently licensed to Canon, Inc. Japan. Both Roorda and the company may benefit financially from the publication of this research.

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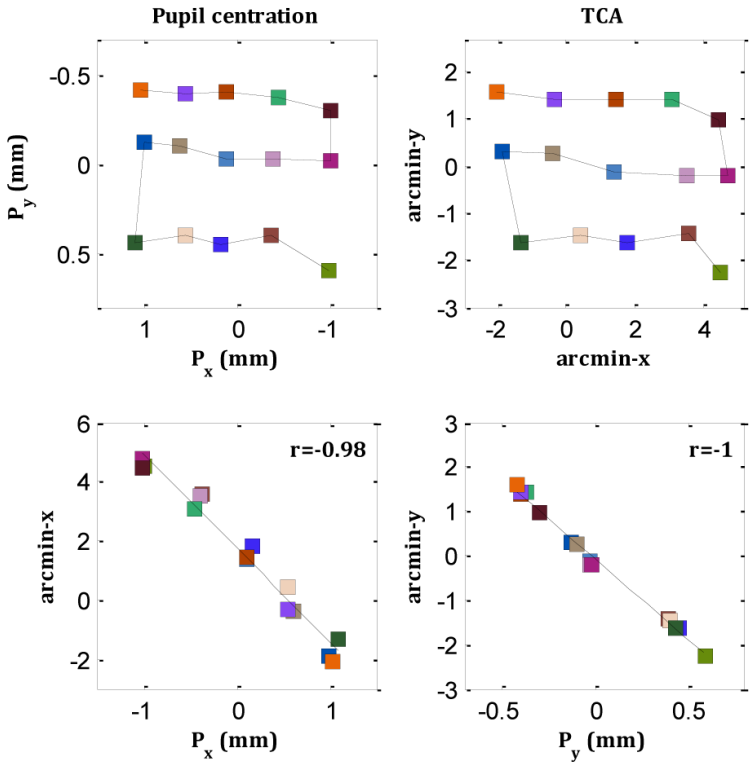

Figure 1.

Subject 1's pupil position was moved into fifteen different positions (top left). TCA, measured with the AOSLO (top right), shows a remarkable linear correspondence and an averaged linear ratio of ρ = 3.5 arcmin of TCA per mm of pupil shift. Vertical and horizontal regressions are reported below (each color represents the same measurement position in all four plots).

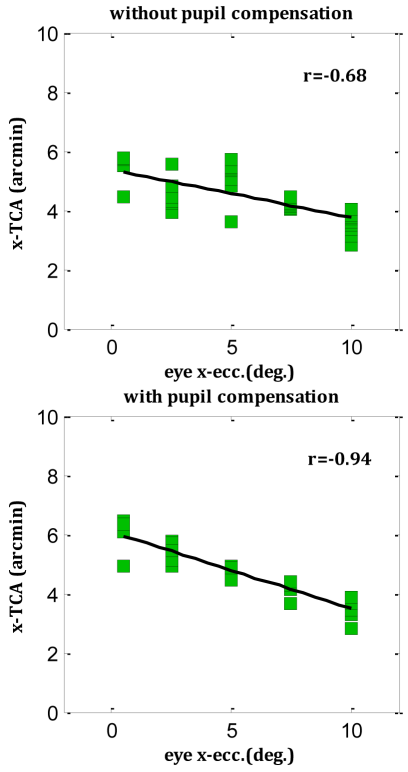

Figure 2.

The TCAs for Subject 1, measured along the nasal direction at different angles of eccentricity from the fovea is plotted before (top) and after (bottom) pupil position correction. The TCA linearity and variability both improve.

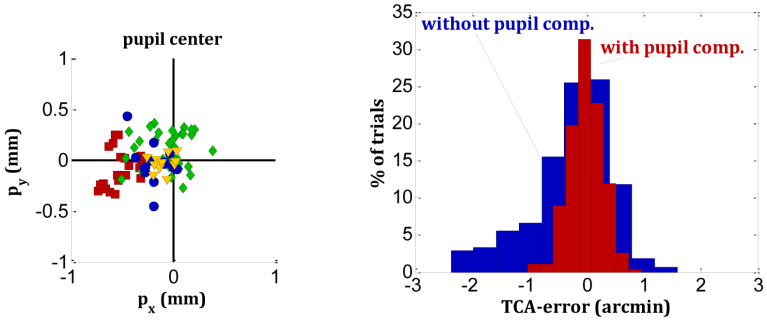

Figure 3.

The mean pupil center for one subject and all trials (left, with the four colors corresponding to the four eccentricity directions) shows the magnitude of experimental variability, i.e. how well the pupil was centered. All subjects' TCAs were collected in one single histogram to show variability before and after pupil correction (we referred to this as TCA error, right).