Visual performance in emmetropia and low myopia after correction of high-order aberrations

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Myopic observers may not benefit to the same extent as emmetropes from adaptive optics (AO) correction in a visual acuity (VA) task. To investigate this, we measured AO-corrected VA in 10 low myopes and 9 emmetropes. Subjects were grouped by refractive error. Mean spherical equivalent refractive error was -2.73 D ($SEM = 0.35$) for the myopes and 0.04 D ($SEM = 0.1$) for the emmetropes. All subjects had best corrected VA of 20/20 or better. The AO scanning laser ophthalmoscope was used to project ultrasharp stimuli onto the retina of each observer. High-contrast photopic acuity was measured using a tumbling E test with and without AO correction. AO-corrected minimum angle of resolution was 0.61' ($SEM = 0.02'$) for the myopes and 0.49' ($SEM = 0.03'$) for the emmetropes. The difference between groups is significant ($p = .0017$). This effect is even greater ($p = .00013$) when accounting for spectacle magnification and axial length, with myopes and emmetropes able to resolve critical features on the retina with a mean size of 2.87 μm ($SEM = 0.07$) and 2.25 μm ($SEM = 0.1$), respectively. Emmetropes and low myopes will both benefit from AO correction in a VA task but not to the same extent. Optical aberrations do not limit VA in low myopia after AO correction. There is no difference in the high-order aberrations of emmetropes and low myopes. Retinal and/or cortical factors limit VA in low myopes after AO correction.

Introduction

The relationship between the optical image on the retina and human visual perception continues to be a primary focus of vision research (for an excellent review, see Westheimer, 2006). Recent technical innovations such as wavefront sensing and adaptive optics (AO) have allowed researchers for the first time to precisely quantify the complex optical system that is the human eye and attempt to control the retinal image with unprecedented sophistication. Prior to the development of AO, the only way to nullify the effects of the eye's optical aberrations was to bypass them by creating interference fringes directly on the retina (Campbell & Green, 1965; Westheimer, 1960). This technique has been very successful; among numerous other applications, it has been applied to estimate the contrast transfer function of the human eye (Campbell & Green, 1965), to determine the absolute contrast sensitivity (CS) of the neural visual system (Williams, 1985), and to estimate the topography of the foveal cone mosaic

(Williams, 1988). A limitation of the technique is that it does not allow for the presentation of complex stimuli, such as those found in natural scenes or those used in standard measures of visual acuity (VA).

Since the invention of the scanning laser ophthalmoscope (SLO), modulation of the scanning laser beam to project complex patterns onto the retina and simultaneously record retinal images has found many applications. It has been applied for locating scotomas with microperimetry, finding the preferred retinal locus in eyes with central scotomas, measuring VA and visual function, and examining fixational dynamics during reading and in patients with retinal pathology (Guez et al., 1998; Le Gargasson, Rigaudiere, Guez, Schmitt, & Grall, 1992; Mainster, Timberlake, Webb, & Hughes, 1982; Timberlake et al., 1986; Timberlake, Mainster, Webb, Hughes, & Trempe, 1982; Webb, Hughes, & Pomerantzeff, 1980). Technical limitations of the traditional SLO, including its wide scan field and low resolution, have limited its usefulness for certain applications, such as probing the fine structure of the retina and delivering photoreceptor

scale stimuli to the retina. The AOSLO overcomes many of these limitations (Roorda et al., 2002).

AO correction of the eye's optical aberrations in the AOSLO allows not only for much higher resolution imaging of the retina than conventional SLO but also for the delivery of ultrasharp photoreceptor scale complex stimuli (Poonja, Patel, Henry, & Roorda, 2005; Roorda et al., 2002). Visual stimulus delivery in the AOSLO allows for the display of static images as well as for presentation of dynamic imagery, such as videos or animations. With recent software and hardware innovations, this has become almost as simple as presenting them on a CRT (Poonja et al., 2005). The greatest advantage that AOSLO brings to psychophysics is its ability to present complex stimuli to the retina of higher optical quality than the visual system has ever experienced. The AOSLO is, therefore, a powerful tool to examine the performance of the retinal and cortical visual system nearly independent of the eye's optics.

With the development of AO systems, researchers have examined the improvement in CS and VA afforded by directly measuring the eye's high-order aberrations with a wavefront sensor and correcting them with either a deformable mirror (DM), (Liang, Williams, & Miller, 1997; Poonja et al., 2005; Williams et al., 2000; Yoon & Williams, 2002) or a phase plate (Yoon, Jeong, Cox, & Williams, 2004). The CS improvement for normal healthy eyes has been shown to be substantial, affording a sixfold increase in sensitivity at 27.5 cycles per degree (cpd) and allowing for the detection of very high spatial frequency gratings at 55 cpd, which were imperceptible while viewing through the eye's normal optics (Liang et al., 1997).

Improvements have also been seen in VA, but they have not been shown to be as great as the improvements in CS (Yoon et al., 2004; Yoon & Williams, 2002). Yoon and Williams reported a 1.6-fold improvement in VA (a 37.5% reduction in the minimum angle of resolution [MAR]) after correcting for both monochromatic and chromatic aberrations in seven eyes (Yoon & Williams, 2002). Poonja et al. (2005) showed an average reduction in MAR of 33% for six eyes after AO correction in the AOSLO used in this study.

VA and CS are, of course, not directly comparable, as benefits at a particular low spatial frequency may not result in improvements in a VA task; this would depend upon whether or not observers used information at that spatial frequency in making their decision in the VA task. The high spatial frequency cutoff of the CS function (usually only detectable at 100% contrast) can be considered the grating acuity limit (expressed in MAR) (Thorn & Schwartz, 1990). Unfortunately, previous studies that looked at CS after AO correction did not measure CS out to the high spatial frequency cutoff. Liang et al. (1997) approached it by measuring out to 55 cpd, but Yoon and Williams (2002) only tested out to 32 cpd. If they had, it is likely that they would have found a higher

spatial frequency cutoff after AO correction than before (as Liang et al., 1997, did). The modest improvements found in VA may be due to the fact that letter acuity is more sensitive than grating acuity to changes in resolution resulting from blur (Green & Campbell, 1965; Hirsch, 1945; Radhakrishnan, Pardhan, Calver, & O'Leary, 2004a; Strang, Winn, & Bradley, 1998; Thorn & Schwartz, 1990); hence, any residual aberrations in the eye or instrument that caused retinal image blur would likely affect the VA measures more than the CS measures. This is especially likely, considering that VA is generally tested clinically with letter stimuli that have broad spatial frequency content and that blur is more likely to affect higher spatial frequencies to a greater extent than lower spatial frequencies. In addition, any residual aberrations causing phase distortions would disrupt the phase relationships between the individual spatial frequency components of a complex stimulus, whereas a grating of a single spatial frequency would only be displaced.

It is clear that high-order aberrations present in all eyes limit visual performance, but it is not clear if all eyes will benefit to the same extent from AO correction. The amount of high-order aberrations that can be corrected varies as a function of pupil size, with the biggest pupils having the greatest potential for large improvements in optical quality. This is because the magnitude of high-order aberrations increases and the blur from diffraction decreases with increasing pupil size. When the pupil is small (~3 mm), diffraction dominates and monochromatic aberrations beyond defocus and astigmatism are less of a factor (Liang et al., 1997). Additionally, eyes that naturally have more high-order aberrations should theoretically benefit the most from AO correction (Liang et al., 1997).

The two previous studies with regard to the benefit of AO correction on VA have had a small number of observers (seven or fewer) and have not been explicit as to the magnitude of ocular aberrations present in these eyes or their refractive error. Yoon and Williams (2002) reported the refractive error of only two of the six observers whose acuities were measured (both slightly myopic), and Poonja et al. (2005) made no reference to the refractive error of the subjects they tested. Refractive error is an important factor to consider when examining the improvements afforded by AO, as there is a large amount of evidence for reduced VA and high-frequency CS in myopia (Atchison, Schmid, & Pritchard, 2006; Coletta & Watson, 2006; Collins & Carney, 1990; Fiorentini & Maffei, 1976; Jaworski, Gentle, Zele, Vingrys, & McBrien, 2006; Liou & Chiu, 2001; Radhakrishnan, Pardhan, Calver, & O'Leary, 2004b; Strang et al., 1998; Subbaram & Bullimore, 2002; Thorn, Corwin, & Comerford, 1986).

This reduction in visual performance can be attributed to optical, retinal, or cortical factors or to some combination of the three. Some studies have shown that the optical quality of myopic eyes is worse than that of emmetropic

eyes, in that they exhibit increased monochromatic aberrations (Applegate, 1991; Buehren, Collins, & Carney, 2005; Collins, Wildsoet, & Atchison, 1995; He et al., 2002; Marcos, Barbero, & Llorente, 2002; Paquin, Hamam, & Simonet, 2002), whereas others have not found a correlation between refractive error and monochromatic aberrations (Artal, Benito, & Tabernero, 2006; Carkeet, Luo, Tong, Saw, & Tan, 2002; Cheng, Bradley, Hong, & Thibos, 2003; Netto, Ambrosio, Shen, & Wilson, 2005; Porter, Guirao, Cox, & Williams, 2001; Radhakrishnan et al., 2004b; Zadok et al., 2005). For a thorough review of the literature concerning aberrations and myopia (up to 2005), see Charman (2005).

Retinal changes due to increased axial length may also affect visual performance and retinal functioning in myopia. Retinal stretching in myopia decreases the neural sampling density of the myopic retina (Coletta & Watson, 2006; Troilo, Xiong, Crowley, & Finlay, 1996). Peripheral acuity has been shown to be limited by retinal stretching in myopia, and in some cases, it has been shown to limit foveal acuity in myopes (Chui, Yap, Chan, & Thibos, 2005). Jaworski et al. (2006) found reduced visual sensitivity in a spatial summation task in high myopia but normal CS at low spatial frequencies, implying normal photoreceptor function but dysfunction of postreceptoral elements. This would suggest that the photoreceptors maintain their sensitivity but are stretched over a larger area on the retina. Further evidence for altered retinal functioning in myopia comes from multifocal electroretinograms of myopic retinas, which show reduced and/or delayed responses, even in low myopia (Chen, Brown, & Schmid, 2006; Kawabata & Adachi-Usami, 1997).

Differences in cortical sensitivity in myopia may also play a role in limiting the ability of myopes to resolve high spatial frequencies. This impairment may be due to the fact that during the critical early stages of development of the visual system, the neurons responsible for processing fine detail are stimulated less frequently. According to the Fourier theory of vision, there exist within the visual system quasi-linearly operating independent mechanisms (channels) selectively sensitive to limited ranges of spatial frequencies (Campbell & Robson, 1968). There is overwhelming evidence that the constituent elements of these channels are neurons located in V1, which are tuned to a specific band of spatial frequencies (DeValois & DeValois, 1988). It is likely that these channels would require input at the proper spatial frequency range for normal development. The highest spatial frequency channels could be impaired in myopia due to the fact that uncorrected young myopes will only experience clear vision of fine details when objects are very near to them (Fiorentini & Maffei, 1976). Even the best corrected adult myope may typically experience more chronic blur. This could be a result of their not constantly wearing their correction device or from wearing one with an imprecise correction. There is little hard evidence to support this conjecture.

The effect of chronic blur in myopia actually may improve rather than reduce VA. Blur adaptation studies, in which subjects wear fogging lenses during extended periods of natural viewing, have shown that VA improves in both myopes and emmetropes after a couple of hours of adaptation (George & Rosenfield, 2004; Mon-Williams, Tresilian, Strang, Kochhar, & Wann, 1998; Rosenfield, Hong, & George, 2004). The underlying mechanism appears to be cortical in locus, as studies have shown that this phenomenon exhibits interocular transfer to the untrained eye (Mon-Williams et al., 1998). Webster, Georgeson, and Webster (2002) have shown that a similar effect of neural sharpening can occur in mere seconds after viewing a spatially filtered natural image. It has been proposed that this is related to the spatial frequency specific adaptation shown by Georgeson and Sullivan (1975), whereby the spatial frequency content of the visual world is optimized so as to give the sharpest percept. Adaptation state therefore is something that must be carefully considered in visual performance studies. Some researchers have shown that normal persons exhibit some adaptation to their own particular pattern of aberrations (Artal et al., 2004). Neural adaptation after laser eye surgery has also been suggested as a possible reason that improvements in VA after laser refractive surgery are not seen for weeks (Pesudovs, 2005). It is likely that there are at least two different adaptational mechanisms at work: one that is fast-acting and works to optimize the percept based upon the spatial frequency content of the entire visual scene (i.e., Webster et al., 2002) and another that acts over a longer time course (i.e., Artal et al., 2004) to provide the sharpest image by minimizing the effect of persistent optical aberrations.

To examine the extent to which high-order aberrations limit VA in myopia, we measured the VA of 10 low myopes and compared them to 9 emmetropes. We chose to examine low myopes so as to avoid any major retinal pathology related to high myopia, as well as for technical limitations, which make it difficult to image high myopes in the AOSLO. Furthermore, it may be the case that previous researchers have found reduced VA in myopia due to the fact that the high myopes in their studies may have biased the results. Limiting ourselves to studying low myopes eliminates this possibility.

Methods

Participant screening and clinical testing

Participants were recruited from the student population of the University of California, Berkeley. Informed consent was obtained from the participants after the nature of the study and its possible complications were explained verbally and in writing. This experiment was approved by

the University of California, Berkeley, Committee for the Protection of Human Subjects. Each participant was refracted clinically using monocular subjective refraction (with fogging) and placed into the myopic or emmetropic group. As our intention was to only include persons with low myopia, we excluded any potential subjects with more than 4 D of myopia and/or more than 1.25 D of astigmatism. The emmetropic group was defined as those participants having between −0.25 and +0.75 D of spherical refractive error and not more than 0.25 D of astigmatism.

Mean spherical refractive error was −2.45 D (SEM = 0.34, range = −0.5 to −3.75) for the myopes and 0.06 D (SEM = 0.1, range = −0.25 to +0.75) for the emmetropes. Mean cylindrical refractive error was −0.55 D (SEM = 0.16, range = −1.25 to 0) for the myopes and −0.04 D (SEM = 0.03, range = −0.25 to 0) for the emmetropes. All participants had best corrected VA of 20/20 (MAR of 1′) or better, as measured clinically using the Bailey–Lovie chart (Bailey & Lovie, 1976). Mean MAR of all subjects was 0.83′ (SEM = 0.02′).

The mean age of the myopic group was 25 years (SD = 4.9, range = 20–35), and the mean age of the emmetropic group was 29 years (SD = 7.5, range = 22–41). Participant screening included a self-report questionnaire about ocular health history. Persons who had undergone refractive surgery and those with a history of eye problems were excluded from this study. For the myopic participants, questions related to their history of myopia, correction type, and wearing habits were asked. These data are summarized in Table 1.

Axial length measurements

Axial length was measured using ultrasound biometry (Mentor Scan-A III, Mentor Corporation, California).

Axial lengths were measured five times; all measurements reported and used in the calculations are the mean of the five measures. Axial lengths were not measured on the same day as the psychophysics experiments so as to avoid any possible short-term deformation or desquamation of the corneal epithelium due to measurement. The myopes had axial lengths ranging from 23.57 to 25.48 mm (M = 24.65, SEM = 0.22). The emmetropes had axial lengths ranging from 21.74 to 23.8 mm (M = 22.91, SEM = 0.19). Axial length as a function of spherical equivalent refractive error is shown in Figure 1. All myopes with a spherical equivalent refractive error greater than −2.5 D had an axial length longer than 24.5 mm. There was a large range of axial lengths for the emmetropes that were classified as plano during subjective refraction, spanning just over 2 mm, from 21.74 to 23.8 mm.

AOSLO imaging and psychophysics

The AOSLO was used to project a high-contrast stimulus onto the retina of each observer. The stimulus (a Snellen E) was scanned onto the retina in a raster fashion with a 658-nm (red) diode laser. Scanning was carried out with a resonant scanner–galvanometric scanner combination (Electro-Optics Products Corp., Flushing Meadows, NY). The beam is scanned at 16 kHz in a sinusoidal pattern by the resonant horizontal scanner, which is coupled to the vertical galvanometric scanner that operates in a sawtooth pattern at 30 Hz. Each frame consists of 525 horizontal lines. Scan amplitude sets the imaging field size and can be adjusted manually to be between 0.5° and 3°. A calibration grid placed at the retinal plane during system setup and calibration allows for the precise setting of field size (Grieve, Tiruveedhula,

	Age of myopia onset (years)	Is myopia progressing?	Correction method	No. of years wearing correction device	Frequency of wearing correction device (hr/day)
M1	8	No	S	17	8
M2	16	No	S	15	16
M3	15	No	S	20	8
M4	16	No	S	5	1
M5	12	Yes	CL(T)	8	10
M6	17	Yes	S	11	15
M7	13	No	CL	11	12
M8	12	Yes	S	6	16
M9	18	Yes	S	1	16
M10	6	No	CL	5	16

Table 1. History of myopia and correction-wearing habits for myopic group. S = spectacles, CL = soft contact lenses, CL(T) = toric soft contact lenses.
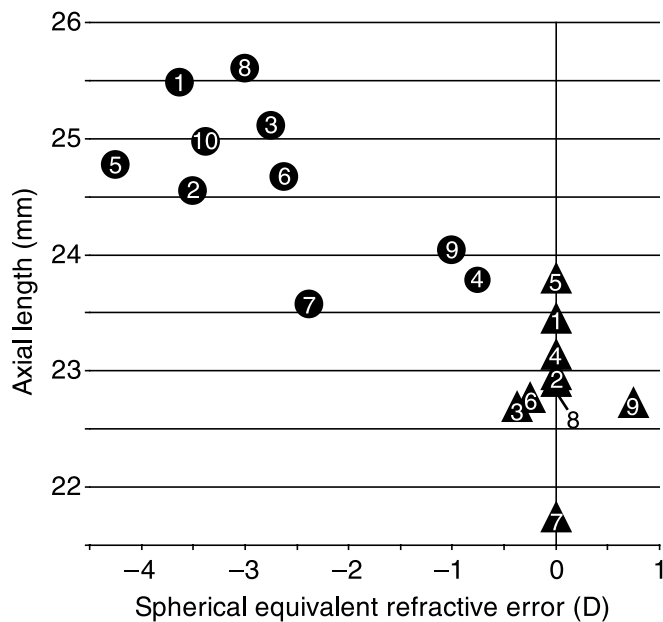
Figure 1. Spherical equivalent refractive error as a function of axial length for the myopes (circles) and emmetropes (triangles). The observer's number is given inside the symbol.

Zhang, & Roorda, 2006; Poonja at al., 2005; Roorda et al., 2002).

The field size used in this study was approximately $30' \times 30'$, with the central $10' \times 10'$ portion optimized to occur over the most linear section of the sinusoidal horizontal scan. Linearization of the central portion of the scan was accomplished by projecting a checkerboard target of known pixel dimensions onto the calibration grid and setting them to be in register. This resulted in the central $10' \times 10'$ section being approximately linear, with ~16 pixel lines corresponding to 1 arcmin. Optimization of the central portion of the raster scan was essential for ensuring that the Snellen E stimulus used for acuity testing was not distorted horizontally, which would have been a cue to stimulus orientation and would have invalidated any of our measures of acuity.

To produce the Snellen E in the raster scan, the beam was modulated using an acousto-optic modulator (AOM; Brimrose Corp., Baltimore, MD) that is placed in the path of the beam prior to the entrance pupil of the system. The AOM, under computer control, can be set to deflect the beam into or out of the system, acting essentially as a switch that can turn on or turn off the beam. The beam is switched on during the forward section of the horizontal scan (because of the sinusoidal nature of the horizontal scanner, it scans in both a forward and return path, turning it on only during the forward path limits light exposure; Poonja at al., 2005). On those lines where portions of the Snellen E were present, the AOM switched the beam off to create the desired stimulus features. This resulted in a

Snellen E that appeared to the observer as black on a bright red background; the resulting contrast was nearly 100%.

Light that is reflected back out of the eye is descanned by the scanning mirrors as it is back reflected along the path of the beam. A small portion of the light is directed via a beamsplitter to a Shack–Hartmann wavefront sensor (SHWS), while the rest passes through and is focused onto a confocal pinhole. The light passing through the pinhole is then detected by a photomultiplier tube (PMT; H7422-20, Hamamatsu, Japan). The PMT signal is sent to a frame grabber board that builds a video image of the retina pixel by pixel. Wavefront compensation is provided by a 37-actuator DM (Xinetics, Andover, MA) placed in the path conjugate to the entrance pupil of the eye prior to scanning. The AOSLO system and stimulus delivery technique is described in detail elsewhere (Grieve et al., 2006; Poonja at al., 2005; Roorda et al., 2002).

VA was measured in the following two conditions: (a) spectacle corrected without AO and (b) spectacle corrected with AO.

Mydriasis and cycloplegia were induced with one drop of 2.5% phenylephrine and one drop of 1% tropicamide ~20 min prior to the start of the experimental session and were maintained throughout with an additional drop, if necessary. If observers were unable to keep their eye open throughout the duration of the experiment, they were excluded from the analysis; one emmetrope was excluded for this reason. Head stabilization was maintained with a dental impression bite bar mounted on an X-Y-Z stage.

Aberrations were measured using the AOSLO's SHWS, which has 241 lenslets over a 5.81-mm pupil. A digital CCD camera detects the focused spots, and aberrations are fit to a 10th-order Zernike polynomial (Grieve et al., 2006; Roorda et al., 2002). Low-order aberrations (sphere and astigmatism) were corrected using standard trial lenses placed into the AOSLO system near the spectacle plane (~14 mm from the entrance pupil). The RMS error from the SHWS was used to optimize this correction objectively to within 0.1 D.

Participants were then subjectively refracted while viewing a static 20/20 Snellen E stimulus through their spectacle correction in the AOSLO system. This was done because (a) RMS wavefront error has been shown to be a poor predictor of subjective image quality (Applegate, Ballentine, Gross, Sarver, & Sarver, 2003; Applegate, Marsack, Ramos, & Sarver, 2003; Chen, Singer, Guirao, Porter, & Williams, 2005) and (b) although the AO system corrects aberrations, it does not necessarily focus the corrected image onto the photoreceptor plane. A fixed defocus level was determined by placing small amounts of defocus onto the DM and asking the participant to report which looked clearer. If a fixed defocus level was needed, it was placed onto the DM for the experimental trials. This was done separately for each condition. For the AO condition, aberrations were corrected through the best spectacle lens correction and a second subjective refraction

was performed. Most subjects did not require any additional defocus.

High-contrast photopic letter acuity was assessed using a four-alternative forced-choice (4AFC) tumbling E test. The threshold, as determined by QUEST (Watson & Pelli, 1983), was the MAR that the observer could correctly identify 82.5% of the time. One eye was imaged for each subject (typically the right eye); the fellow eye was occluded. The stimulus is presented in Maxwellian view (Maxwell, 1860, cited in Westheimer, 1966), at a retinal illuminance of 6.79 log Trolands. Retinal illuminance in trolands was calculated based upon a laser power of 10 μW over an area of 0.25 deg^2 (Wyszecki & Stiles, 1982). This power level is less than 1% of the American National Standards Institute (ANSI) maximum permissible exposure for continuous viewing of a 658-nm source of this size (ANSI, 2000). This high retinal illuminance was used so that high-contrast retinal images could be obtained during the psychophysical task. Although, at this light level, the photopigment is nearly 100% bleached, there is no indication that this would hinder the subjects' performance (Wyszecki & Stiles, 1982). Subjects adapted quickly to the bright field and had no problem performing the task comfortably.

Each trial was initiated with a keyboard press by the participant. After a brief delay ($\sim$200 ms), the stimulus was presented in one of four randomly chosen orientations and a video of the retina was acquired. Stimulus duration was 500 ms. Video of the retina was acquired for 2 s. Videos were acquired to determine the preferred fixation locus and cone spacing of the observer. At the end of each presentation, the participant indicated the orientation by pressing an arrow key on the keyboard. Experiment control and data acquisition were accomplished through a custom MATLAB (The MathWorks, Natick, MA) graphical user interface, which controlled custom C++ software developed in this laboratory. QUEST was implemented in MATLAB using the Psychophysics Toolbox extensions (Brainard, 1997; Pelli, 1997).

After one initial practice run, five threshold measurements were made for each condition, with AO and no-AO runs interleaved. Runs consisted of 60 trials each. Thresholds presented are the average of five runs. Participants were instructed to rest during the breaks between runs and at their discretion throughout the experiment to minimize fatigue (this was easily accomplished as the observer self-initiated each trial). During the AO condition, an AO correction was done prior to the beginning of a run and again halfway through (after trial 30). AO correction was optimized by the experimenter to give the lowest RMS wavefront error. If a defocus level was preferred by the observer, it was placed onto the DM. The experimenter monitored AO compensation by viewing the live AO-corrected image of the retina on a CRT. If the image quality deteriorated, the subject was instructed to pause and another AO correction was performed. Typically, only two AO corrections were made during each AO condition run.

Calculation of magnification factor

Due to the fact that low-order aberrations (sphere and astigmatism) were corrected using trial lenses in the AOSLO system, slight spectacle magnification effects might tend to exaggerate or eliminate any small differences that may exist between groups. Using both the spectacle magnification factor, which we can calculate from the trial lens power in the system, and the axial length, which we have measured, we can calculate the size of features in the retinal image. This is important for relating the size of a detectable stimulus feature to the size of individual photoreceptors in the retina.

We used the standard thin-lens equation to calculate the magnification factor due to the spectacle lenses:

$$M_{\text{spec}} = 1/(1 - dP), \tag{1}$$

where M_{spec} is the spectacle magnification factor (a unitless value), d is the distance from the entrance pupil to the spectacle lens (in mm), and P is the lens power (in D). Using a constant value of 14 mm for d might seem simplistic, as there is indeed some variation in the distance from the lens well to the entrance pupil of the eye. These are due to individual differences in head and face shape, bite bar, and constraints imposed by the optical bench. However, this equation is very insensitive to small variations; for a -4 D myope (the largest in this study), an error in distance of $\pm$2 mm would only change the magnification factor by $\sim\pm$0.007, or less than 1% in either direction.

Calculating the size of retinal features

We used Bennett's adjusted axial length method to calculate the size of features in the retinal image (Bennett, Rudnicka, & Edgar, 1994). This method is advantageous in that one needs only the axial length to convert degrees of visual angle to millimeters on the retina. This requires the calculation of q, which is a scaling factor relating the two units. Multiplying a known visual angle by q gives the size of the corresponding retinal features in millimeters. The following equation is used to obtain q:

$$q = 0.01306(x - 1.82), \tag{2}$$

where x is the axial length in millimeters, 1.82 is the distance from the corneal vertex to the eye's second principal point in millimeters (taken from the Bennett–Rabbett's model eye), and 0.01306 is a constant that converts

radians to degrees and takes into account n, the refractive index, which is taken to be 1.336 (Bennett & Rabbetts, 1989; Bennett et al., 1994). For a 24-mm emmetropic eye, $q = 0.298$, and 1° of visual angle equals 0.298 mm or 298 μm on the retina. Here, again, we are approximating by using 1.82 mm as the distance from the corneal vertex to the eye's second principal point, but we are confident that any individual variations would have only a small effect on q. Bennett states that individual variations in the distance between the corneal vertex and the eye's second principal point are unlikely to exceed ±0.55 mm. The resulting maximum error in q is only the ratio of 0.01306 and 0.55, or 0.007 (Bennett et al., 1994).

Results

Mean raw (unadjusted for spectacle magnification) AO-corrected MAR was 0.61′ ($SEM = 0.02′$) for the myopes and 0.49′ ($SEM = 0.03′$) for the emmetropes. These values correspond to Snellen acuities of 20/12.1 and 20/9.9, respectively. The difference between groups is significant ($p = .0017$, t test, two tailed). Results for the myopic group are shown in Figure 2, and those for the emmetropic group are shown in Figure 3. Mean MAR for all observers before AO correction was 0.81′, which is very similar to the 0.83′ MAR measured clinically with the Bailey–Lovie chart (Bailey & Lovie, 1976).

Within each group, there was a significant improvement in VA after AO correction of high-order aberrations. The myopic group improved from a mean of 0.83′ in the no-AO condition to a mean of 0.61′ with AO—a 27% reduction in MAR. The emmetropes improved to an even greater extent, from a group mean of 0.8′ without AO to a mean of 0.49′ with AO—a 39% reduction in MAR. These reductions are consistent with the 33% reduction in MAR

Figure 3. Raw MAR for the emmetropic group with (filled triangles) and without (open triangles) AO correction. Error bars represent the standard error of the mean.

found by Poonja et al. (2005) and the 37.5% reduction found by Yoon and Williams (2002).

The difference between groups is less significant when taking into account magnification effects from the spectacle lenses used to correct low-order aberrations. The mean spectacle-magnification-adjusted AO-corrected MAR is reduced to 0.58′ for the myopes; it remains essentially unchanged, at 0.49′, for the emmetropes. Spectacle magnification reduced the angular retinal image size by ~5.2% for the myopes. Spectacle-magnification-adjusted AO-corrected MAR for both groups is shown in Figure 4. Although the groups become more similar after adjusting for spectacle magnification effects, there is still a statistically significant difference between them ($p = .0082$, t test, two tailed).

The difference between the myopic and emmetropic groups is enhanced even over the raw MAR values when accounting for both spectacle magnification and axial length ($p = .00013$, t test, two tailed). VA is typically defined as the finest spatial detail that the visual system
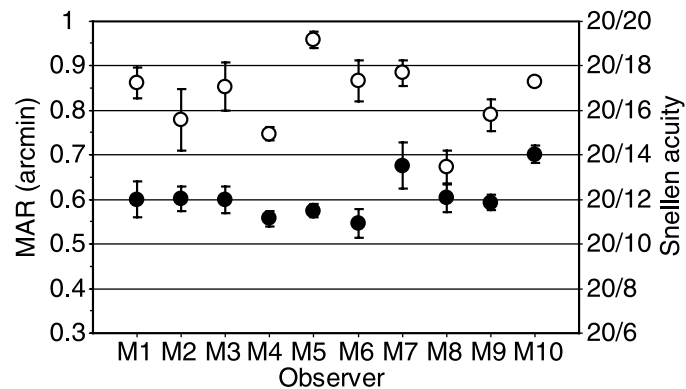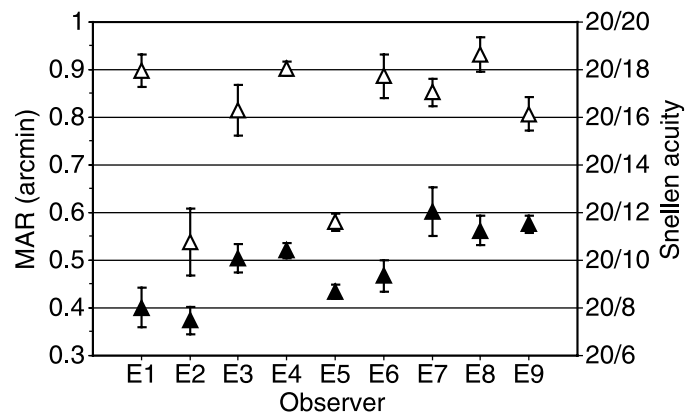

Figure 2. Raw MAR for the myopic group with (filled circles) and without (open circles) AO correction. Error bars represent the standard error of the mean.
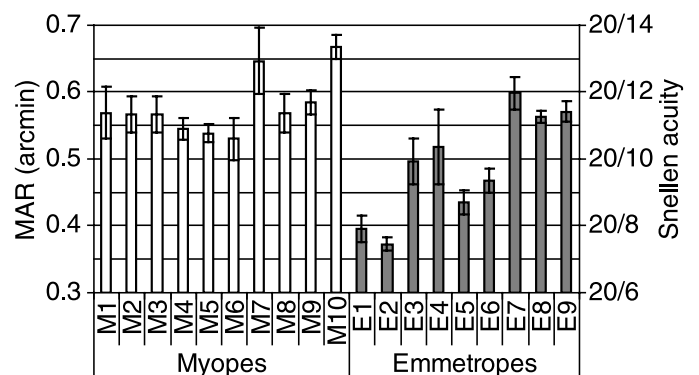

Figure 4. Spectacle-magnification-adjusted AO-corrected MAR. Error bars represent the standard error of the mean. Spectacle magnification slightly reduces the difference between groups.
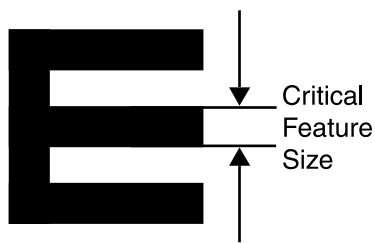
Figure 5. The critical feature of a Snellen E is equal to the distance on the retina (in μm) of one half cycle of the high spatial frequency square wave component.

can resolve. When relating the performance in terms of actual spatial units on the retina, as opposed to MAR, we must define the critical feature size (CFS) of the stimulus. The CFS is simply the size (in μm) of the smallest detectable feature on the retina of the Snellen E that is required to correctly judge its orientation. We define the CFS for a Snellen E as the distance on the retina (in μm) between two successive bright bars in the strokes of the Snellen E, or one half cycle of the high spatial frequency square wave component embedded in the Snellen E (Figure 5). Myopes are able to resolve a critical feature on the retina with a mean size of 2.87 μm (*SEM* = 0.07), whereas emmetropes are able to do so with a mean size of 2.25 μm (*SEM* = 0.1). Acuities in terms of CFS are shown in Figure 6.

The radial average modulation transfer function (MTF) was computed for each observer from the residual wavefront aberration after AO correction and is shown in Figure 7. MTFs after AO correction were quite similar for both groups, revealing that each group achieved similar levels of AO compensation for their optical aberrations. There was no significant difference in the total RMS wavefront error between the myopes and emmetropes before AO correction (myopes: $M = 0.4$ μm,
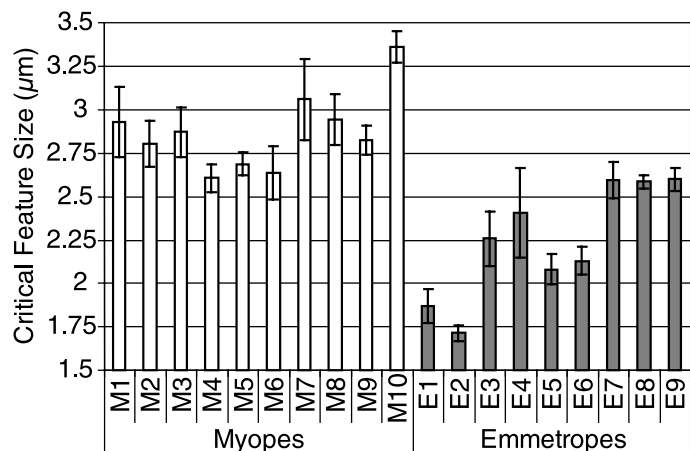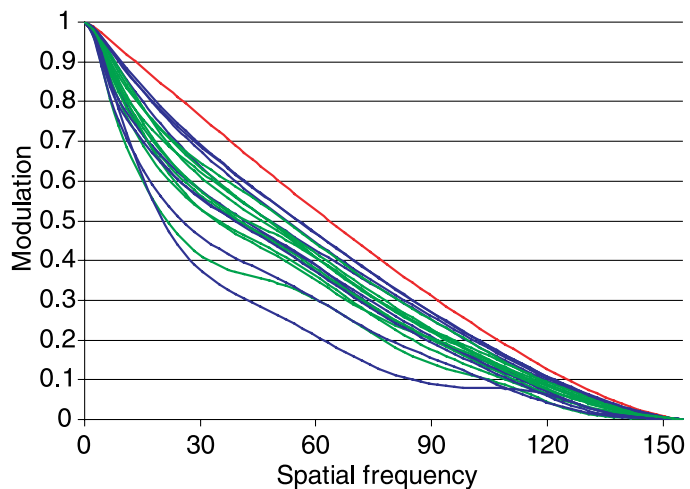

Figure 7. MTFs of myopes (green) and emmetropes (blue) were quite similar after AO correction. Diffraction-limited MTF is shown in red for comparison.

SEM = 0.12 μm; emmetropes: $M = 0.32$ μm, *SEM* = 0.08 μm; $p > .5$, *t* test, two tailed). Nor was there a significant difference between groups in total RMS wavefront error after AO correction (myopes: $M = 0.079$ μm, *SEM* = 0.017 μm; emmetropes: $M = 0.081$ μm, *SEM* = 0.022 μm; $p > .9$, *t* test, two tailed).

The magnitudes of Zernike Modes 3–21, before and after AO correction, are shown in Figures 8 and 9, respectively. These were computed by first taking the absolute value of the residual RMS error for each Zernike mode for each observer and then averaging within groups. It can be seen from Figures 8 and 9 that the magnitudes of Zernike Modes 3–21 are very similar between groups prior to and after AO correction. Zernike modes above Mode 21 are excluded because their magnitudes are so

Figure 6. AO-corrected CFS. Error bars represent the standard error of the mean.
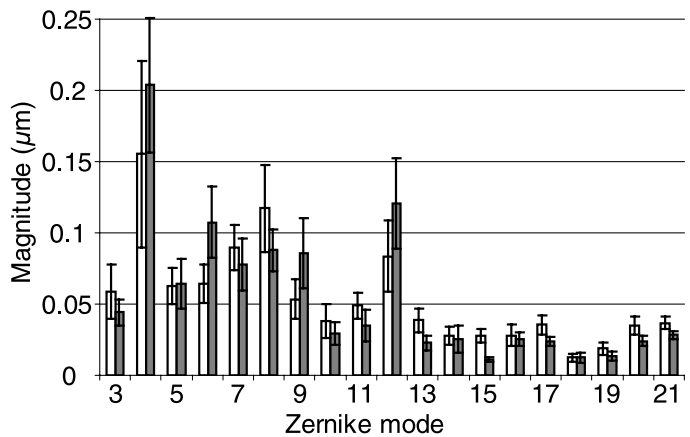

Figure 8. Magnitude of Zernike Modes 3–21 for myopes (open bars) and emmetropes (shaded bars) after spectacle correction but without AO correction. Error bars represent the standard error of the mean.
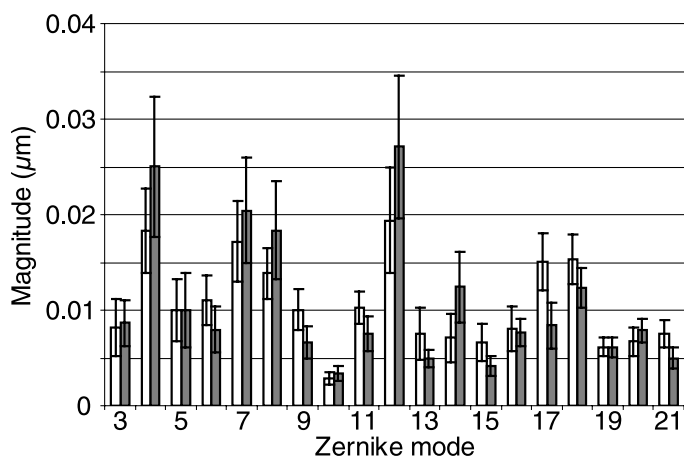
Figure 9. Magnitude of Zernike Modes 3–21 for myopes (open bars) and emmetropes (shaded bars) after AO correction. Error bars represent the standard error of the mean. Note the difference in scale from Figure 8.

small that they are of little consequence. It is difficult to interpret the impact of individual modes simply by their magnitude, as they tell us little about their effect on vision or about the subjective image quality of the observer. This is because not all Zernike modes have the same effect on vision and can have complex interactions with one another that can either enhance or degrade subjective image quality and VA (Applegate, Ballentine, et al., 2003; Applegate, Marsack, et al., 2003; L. Chen et al., 2005).

Discussion

AO correction of the eye's high-order aberrations improves VA in both myopes and emmetropes but not to the same extent. Because both groups received similar levels of optical quality through AO compensation for their aberrations, it is unlikely that this difference is due only to residual optical factors. The MTFs (Figure 7) and residual wavefront RMS for Zernike Modes 3–21 (Figures 8 and 9) are very similar for both groups after AO correction. Although individual Zernike modes tell us little about subjective image quality, it has been shown that very low levels of RMS error (<0.05 μm) have no significant effect on VA (Applegate, Ballentine, et al., 2003). All aberrations after AO correction fell below this level (Figure 9). We can get some indication of the image quality that the observers in this study experienced by convolving their individual point spread function (PSF) with a properly scaled Snellen E.

We have done this for all of the participants in this study and show the results for M10, the worst performing myope in the AO condition, in Figure 10. The corresponding wavefront aberration maps, MTFs, and phase transfer functions (PTFs) before and after AO correction
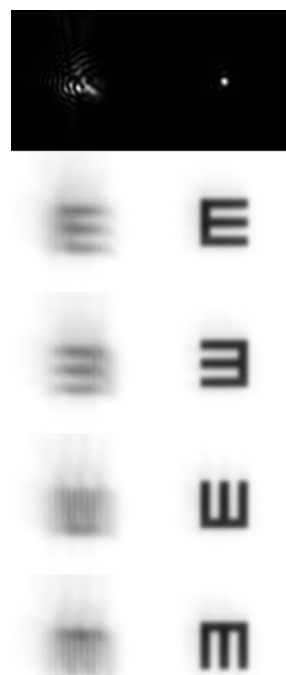

Figure 10. PSFs (top panel) and convolved Snellen Es for observer M10; scaled to threshold without (left column) and with AO (right column).

are shown in Figures 11, 12, and 13, respectively. The top panel of the right column of Figure 10 shows the AO-corrected PSF, and the corresponding convolved Snellen Es at the four orientations tested, scaled to be at threshold (0.7′ MAR), are shown below it. We compare them here to the PSF and convolved Es without AO, again scaled at this observer's threshold for that condition (0.87′ MAR). We scale them to be at threshold so that we may examine the effects of the PSF on the image quality independent of letter size. It is obvious that the optical degradations caused by the eye's normal optics are minimized in the AO condition so that they no longer are a limiting factor for determining stimulus orientation.
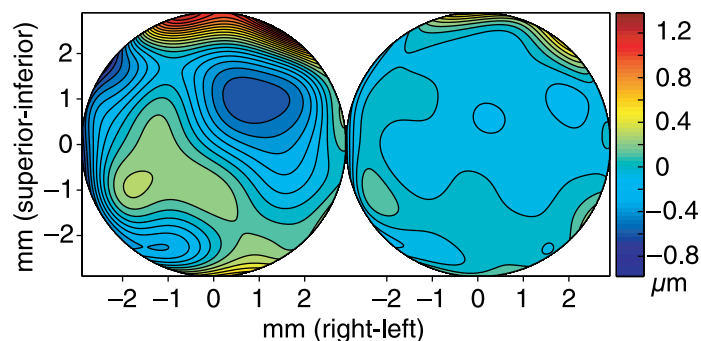

Figure 11. Wavefront aberration for subject M10 before (left) and after AO (right). Contour lines are separated by 0.1 μm; color bar shows color relations in microns.
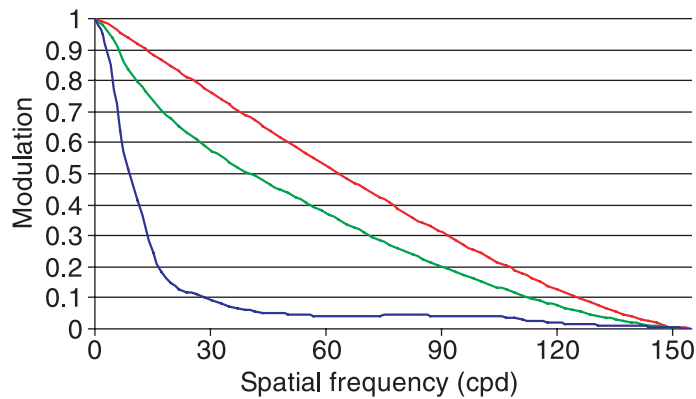
Figure 12. MTFs before (blue) and after AO (green). Diffraction-limited MTF (red) is shown for comparison.

It can be seen from Figure 10 that without AO correction, the PSF of a normal eye has two main effects on a Snellen E. First, it reduces the overall contrast, which can easily be inferred by looking at the corresponding MTF (Figure 12, blue curve). Secondly, and likely more important, it disrupts the relationship between the different spatial frequency components of the Snellen E. Depending upon the orientation of the letter, this effect can be moderate, as in the normally oriented E, or it could be more severe, as it is for the vertically oriented Es.

A striking feature of Figure 10 is how much information about the orientation still exists in the stimulus even when a great deal of the high spatial frequency information has been blurred out by the optical aberrations of the eye. This illustrates how oversimplistic the notion of VA based upon Snellen letters (or any optotype) is. It is likely that a trained observer might easily be able to pick up on the low spatial frequency cues to orientation, such as the gradient of contrast present in the blurred letters of Figure 10. This could explain why some persons are able to improve in VA tasks with training (Bondarko & Danilova, 1997).

Bondarko and Danilova (1997) did a Fourier analysis of two of the most widely used optotypes: the Landolt C and the Snellen E. They showed that the difference in amplitude spectra of orthogonal orientations yielded a peak in information at spatial frequencies lower than those corresponding to the gap size. Of the two stimuli, the Snellen E was determined to be a better predictor of actual VA, as it contained less power in the difference spectra at low spatial frequencies (Bondarko & Danilova, 1997). Nevertheless, low spatial frequencies in optotype stimuli may drive acuities for the most experienced observers. Only one observer in this study, E2 (one of the authors), had any prior experience in the task; thus, it is unlikely that any experience-dependent effects might have biased the results we obtained. Preliminary results from this laboratory suggest that any improvements in this acuity task due to learning are minimal (Rossi & Roorda, 2006).

Taking into account all of the available information we have about the optical and retinal image quality after AO correction, we are confident that optics are no longer a limiting factor for this acuity task. It then remains to be determined which of the other possible explanations for the difference that we have measured seems the most plausible. The remaining possible factors limiting visual performance in low myopia are retinal and/or cortical in nature. The main retinal limitation would be the sampling grain of the photoreceptor mosaic. Changes in the myopic retina may occur at either the photoreceptor level or a postreceptoral level within the retina. Cortical limitations are more difficult to isolate, but they could be related to adaptational or developmental effects.

The question of adaptation state is an interesting one that needs further investigation. If, as Artal et al. (2004) have suggested, there is, in fact, some adaptation to one's own aberrations, then there remains the question of the time course and mechanism of this adaptation. It is possible that in removing the aberrations of the eye, we may not have gained the full potential increase in visual performance because the previous adaptation state was still in effect. Furthermore, in natural viewing conditions, the eye's aberrations change with accommodation (e.g., spherical aberration) and pupil size (Cheng et al., 2004); hence, the notion of adaptation to one's own aberrations becomes even more complex. Presumably, the brain would have to have an infinite number of adaptation states to cover all of these conditions or the adaptation mechanism would only apply to those aberrations that are static. Even the latter situation would seem to require multiple adaptation states because the complex interaction of aberrations with one another in different viewing conditions would have different effects on retinal image quality. Artal et al. suggest that there may be a rough adaptation to the general shape of the PSF, which does not change too drastically in different viewing conditions. Whether or not a difference in the adaptation state of myopes versus emmetropes would result in the differences that we have measured requires further investigation.
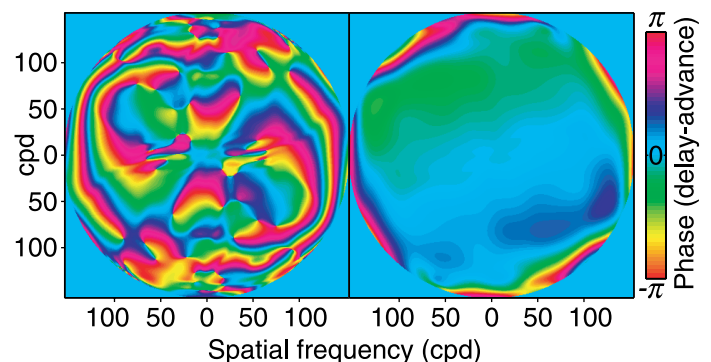
Figure 13. PTFs before (left) and after AO (right). Color indicates phase delay/advance.

The value of 0.49′ MAR obtained with AO correction for the emmetropic group is precisely what would be predicted from the photoreceptor spacing at the fovea, which ranges from 0.42′ to 0.54′ (*M* = 0.51′) (Yoon & Williams, 2002). This value is equivalent to those obtained using interferometric methods, which are considered to be an accurate approximation of the cone spacing in the fovea (Williams, 1985). It is likely that we have achieved a rough estimate of the sampling grain of the photoreceptor mosaic with an AO-corrected VA task. If so, the results from the myopic group, MAR 0.58′ (spectacle magnification adjusted), would predict an increase in photoreceptor spacing.

In retinal units, these results suggest a difference in cone spacing of 0.62 μm (a 28% increase). It is not clear whether this magnitude of stretching is plausible for the low myopes in this study. Further data on the relationship between photoreceptor spacing and visual performance in myopia are required. We attempted to measure foveal cone spacing in these observers with low coherence infrared light, but the cones were not resolved at the center of the fovea. Analysis of the relationship between cone spacing and visual performance near the fovea is a topic of current research. If photoreceptor spacing is similar in emmetropia and myopia, then there must be some postreceptoral differences in sensitivity between myopes and emmetropes, either within the retina or at a later stage in visual processing. This neural insensitivity might be considered a subclinical form of amblyopia.

Conclusions

1. Myopes perform worse than emmetropes in a VA task after AO correction in both angular (MAR) and retinal (CFS) units.
2. Residual optical aberrations after AO correction do not limit VA in emmetropia or low myopia.
3. There is no difference in the high-order aberrations of low myopes and emmetropes.
4. Retinal and/or cortical factors limit VA in low myopes after AO correction.

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