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Changes in through-focus spatial visual performance with adaptive optics correction of monochromatic aberrations

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Abstract

We determined the influence of adaptive optics correction on through-focus illiterate-E visual acuity and through-focus contrast sensitivity under monochromatic conditions. In two subjects, adaptive optics improved high and low (12 %) contrast in-focus visual acuity by 0.1 to 0.15 logMAR, but resulted in more rapid and more symmetrical deterioration in visual acuity away from in-focus. In one subject, adaptive optics improved in-focus contrast sensitivity and resulted in more symmetrical and greater loss of contrast sensitivity about the peak sensitivity because of correction of higher order aberrations. The results show that full correction of higher order aberrations may worsen spatial visual performance in the presence of some defocus.

Keywords: Aberrations; Adaptive optics, Contrast sensitivity; Defocus, Spatial visual performance; Visual Acuity

1. Introduction

Over the last 150 years, visual scientists and clinicians have been interested in the influence of optical defects of the human eye on visual performance and perception of the world. In the past decade, there have been marked improvements in the ability to measure optical imperfections of the eye in terms of objective measurement of higher order aberrations. These improvements have been driven by the potential to correct these aberrations using either contact lenses or through refractive surgery (corneal ablation and intraocular lens implants). Despite these improvements, predictions of visual performance have often not been successful, largely because of limited understanding of the interaction of defocus with other aberrations.

The main optical defect of the eye is defocus as uncorrected refractive errors and presbyopia. Several studies have investigated decreases in spatial visual performance with defocus, with the most common visual functions tested being visual acuity and the contrast sensitivity function. Findings were influenced by several optical related factors including luminance, spectral distribution and contrast of the target, pupil size, and the Stiles-Crawford effect. As an example for visual acuity, this has a maximum for a subject's best correction and decreases with both positive (simulating myopia) and negative (simulating hypermetropia) defocus. This decrease is ameliorated to a minor extent by the Stiles-Crawford effect, but only with large pupils (eg (Atchison, Scott, Strang & Artal, 2002). Usually the decrease is more rapid in the positive than in the negative direction (Atchison et al., 2002)). This is attributed to positive spherical aberration, which occurs in most unaccommodated eyes (Cheng, Barnett, Vilupuru, Marsack, Kasthurirangan, Applegate & Roorda, 2004; Porter, Guirao, Cox & Williams, 2001; Thibos, Bradley & Hong, 2002; Thibos, Hong, Bradley & Cheng, 2002) ameliorating the effect of negative defocus on visual acuity.

Decreases in the contrast sensitivity function (CSF) of the human eye caused by defocus have been measured by Atchison *et al.* (Atchison, Marcos & Scott, 2003; Atchison & Scott, 2002; Woods, Bradley & Atchison, 1996). The CSF is the visual system's ability to detect variation in luminance of sinusoidal grating targets of various spatial frequencies. Contrast sensitivity declined more quickly for positive than for negative defocus, again attributable to the interaction between spherical aberration and defocus. "Notches" (depressions of sensitivity surrounded by more sensitive areas) were demonstrated in the defocused CSF, a result expected according to theoretical modulation transfer functions (MTFs) of defocused optical systems. Defocused CSFs can be compared with predictions based on the in-focus CSF and MTFs derived from measured aberrations according to

CSF prediction (defocus) = *CSF measured* (in-focus) x *MTF* (defocus)/*MTF* (in-focus) (1)

where MTFs are determined from the aberrations. Predictions regarding the shape of the contrast sensitivity function were often good particularly for negative defocus. Again, the Stiles-Crawford effect played a minor role (Atchison et al., 2003; Atchison & Scott, 2002).

Adaptive optics, used in astronomy to compensate for atmospheric turbulence, has been applied recently to the correction of higher-order ocular aberrations such as spherical aberration. Adaptive optics requires sensing of aberrations using a wavefront sensor along with phase modulators to correct aberrations. Liang *et al*. (1997) placed a deformable mirror between the light source and the eye (also between the eye and the sensor), with the mirror being conjugate with both the sensor and the entrance pupil of the eye. The mirror was deformed iteratively until the image closely matched a reference image. Such adaptive optics can produce up to 2 times improvement in contrast sensitivity at high spatial frequencies and 1.2-1.4 times improvement in visual acuity under white light, with even greater relative improvement under monochromatic conditions (Yoon & Williams, 2002). Additional optics can take fundus photographs and measure spatial vision.

As well as attempting to correct some or all of the eye's monochromatic aberrations, adaptive optics allows for manipulation of aberrations to investigate their influence on spatial visual function visual and subjective acceptability of vision. An example of this was the replacement of aberrations by proportional or rotated versions (Artal, Chen, Fernandez, Singer, Manzanera & Williams, 2004; Chen, Artal, Gutierrez & Williams, 2007). This work indicated a degree of neural adaptation to one's own aberrations, at least in the short term, as people preferred their own aberrations to rotated versions and preferred having some proportion of their aberrations remaining rather than having them fully corrected.

The asymmetry of image quality loss about best focus will be reduced with adaptive optics, and might be expected to influence the accommodation feedback loop. However, Chen Kruger, Hofer, Singer & Williams. (2006) found that.most subjects can accommodate equally as well without and with adaptive optics correction.

One concern with applying adaptive optics in the form of contact lenses or refractive surgery is that people might be more susceptible to small amounts of defocus, such as when there is accommodative lag or lead. Higher-order aberrations may have a buffering effect in that, although vision at best focus is reduced, it deteriorates more slowly away from this location. This may mean that an unacceptable level of vision is reached at lower defocus levels when higher-order aberrations are eliminated or that an unacceptable *loss* in vision is reached at lower defocus levels. Some experimental evidence for the former has been found, with Piers, Fernandez, Manzanera, Norrby & Artal (2004) finding greater rates of loss of visual acuity and contrast sensitivity away from best focus when spherical aberration was eliminated compared with spherical aberration at levels expected for spherical intraocular lens patients. Piers, Manzanera, Prieto, Gorceix & Artal (2007) found a greater loss in contrast sensitivity away from best focus when higher order aberrations were corrected as compared with more typical levels of aberration.

As a contribution to understanding the importance of higher-order aberrations to vision, we have measured the influence of defocus on spatial visual performance with and without higher order aberrations. We hypothesise that correcting aberrations increase the symmetry of vision loss about the in-focus position.

2. Methods

This study followed the tenets of the Declaration of Helsinki and received ethical clearance from the Queensland University of Technology's Human Research Ethics Committee.

2.1 Subjects

The subjects were two of the authors. Refractive corrections of their healthy right eyes were -2.25DS/-0.25DC x 70 (DAA) and Plano (BJB). The eyes were cyclopleged with 1 drop of topically applied 1% cyclopentolate, with an additional drop applied at least every 2 hours if sessions lasted longer than 2 hours. Most sessions for the vision acuity experiments lasted about 2 hours, but the two sessions for the contrast sensitivity experiment lasted 6-8 hours each.

2.2 Experimental system

The system consisted of five channels: laser calibration, radiation source, pupil position monitoring, wavefront operations and visual stimulus (Figure 1). The laser calibration channel consisted of a 543 nm He-Ne laser, a 40× microscope objective, a 10 µm pinhole filter and a 120 mm focal length collimating lens. The collimated beam was joined to the radiation source and wavefront operation channels at uncoated pellicle beamsplitter BS_1 (transmission 92%). Aperture A_1 was adjusted to alter the laser beam for different pupil size calibrations.

The radiation source channel consisted of an infrared superluminescent diode (Hamamatsu Photonics, 830 nm, FWHM 25 nm), collimating aspheric lens (*f* 3.1mm, *D* 6.33mm, *NA* 0.68), 1mm aperture A_2, mirror M_4, beamsplitter BS_2, and beamsplitter BS_1 which reflected the radiation into the eye. In order to reduce reflection from the cornea, which may generate noise for the wavefront measurement sensor, we decentred the diode and aperture A_2 by 1.2 to 1.5 mm. The irradiance at the cornea was 14 μW, which is 50 times lower than the Australian/New Zealand laser safety standard for continuous viewing (Australia & Zealand, 2004).

The pupil position monitoring channel consisted of beamsplitters BS_1 and BS_2, mirror M_3 and a Pixelink Pl-A741 firewire camera with 35mm focal length lens, together with infrared LED illumination ring IR. The subject's pupil image was displayed on a computer monitor and used to keep the eye aligned by adjusting the position of the bitebar upon which the subjects' head was mounted.

Light reflected from the retina passed along the wavefront operations channel. This channel included wavefront measurement and wavefront correction. A relay lens pair (L_1 and L_2) imaged the eye pupil onto the surface of the deformable micro-electromechanical system mirror. A second relay lens pair (L_3 and L_4) imaged the eye pupil onto the microlens array of a Hartmann-Shack sensor. The deformable mirror was a Boston Micromachines Corporation μDMS-Multi (gold coated reflection membrane, 12 x 12 array of actuators 4.4 mm diameter on side). The sensor consisted of a rectangular array of 0.4 mm diameter, 24 mm focal length lenslets (Adaptive Optics Associates) and a progressive scan 1008 ×1018 pixels CCD Camera (TM-1020-15, JAI Pulnix, Inc.). Magnifications were 0.667 between the pupil and the deformable mirror and 1 between the pupil and the HS sensor. The channel included an optical trombone (precision 0.1 mm or 0.088 D) between lenses L_1 and L_2 to vary defocus independent of the mirror.

The visual stimulus channel was split from the wavefront operations channel at cold mirror BS_3. The mirror reflected the light from the visual stimulus (letter Es or sinusoidal gratings) towards the eye. The stimuli were provided by a liquid crystal display based high resolution, high brightness projector (Hitachi Ltd, 1280 × 960 pixels, 86 Hz), under control under control by a visual stimulus generator (VSG 2/5 video-card, Cambridge Research System), projecting targets onto a high resolution rear projection screen (Praxino Ltd). The pixel size on the screen was 0.268 mm (0.28 min arc) and the display area was 343 mm width by 254 mm height. The display was rendered monochromatic with a green interference filter (550 nm, FWHM 10 nm). A 5.5 mm diameter stop A_3 conjugated with the entrance pupil was the limiting aperture for the eye. Distance between the screen and the stop was 3.33 m.

2.3. Aberration measurement and correction

When the system was set up, a good quality plane mirror was used in place of the deformable mirror. The collimation of the green laser beam was checked with a shear interferometry at a number of locations in the system (between L_2 and the mirror and between L_4 and the sensor). The reference Hartmann-Shack image was taken for this situation. The plane mirror was then replaced by the deformable mirror. The visual performance tests were conducted without and with the mirror correcting the eye's aberrations. For the null condition, in which the power supply to the mirror was turned off, the root mean square aberration (RMS) of the whole optical system, except for defocus that can be altered by moving the trombone, was less than 0.05 μm as measured by the calibration laser. We checked the aberrations of some real eyes and model eyes with the system and found similar values of spherical aberration coefficient C_4^0 as for a COAS-HD aberrometer (Wavefront Sciences). Incorporating defocus by moving the optical trombone or incorporating astigmatism by introducing trial lenses gave expected values. During trials of the application of adaptive optics we also incorporated a camera to qualitatively examine point spread functions to ensure that aberration correction was working well.

Aberrations were reconstructed from the sensor's slope measurement signals and decomposed as Zernike polynomial expansions up to 12th order with up to the 10th order terms used to drive the deformable mirror and do theoretical calculations. Customized software written with computer language Visual C++ (Microsoft Visual Studio 6.0, Microsoft Pty. Limited) measured and controlled aberrations in real time. The frequency of aberrations measurement and correction was limited mainly by the sensor's grabbing rate (about 15Hz), but software limitations such as determining centroids and/or displaying the results on the screen limited this to about 12Hz. As our experiments required actuators to hold their deforming positions for a few hours, the maximum permissible voltage was reduced from 275 V to 260 V to protect the mirror. We determined aberrations and corrections based on a 5.6 mm pupil, for which there were 68 active actuators.

For measurements without adaptive optics, before an experiment aberrations were measured at 830 nm and the optical trombone moved to a reference position at which the defocus co-efficient C_2^0 was within ±0.05 μm (±0.044 D with 5.6mm pupil size). At this reference position, aberrations were measured continually at 12 Hz for 5-6s and the residual aberration co-efficients and the residual RMS were taken as averages across this time.

For measurements with adaptive optics, all mirror actuators were initially given a uniform voltage value to move the mirror surface to half its maximum displacement so that the mirror could operate in both directions about the shape that this gave to the mirror. After this was done, the optical trombone was moved to a reference position at which the defocus co-efficient C_2^0 was within ±0.05 μm. Just before starting the closed loop correction, subjects blinked deeply and then keep their eyes open widely for 5-6 s. During the first 1.7 s of this period, the wavefront aberrations were measured and in the following 1-1.4 s, depending on the convergence rate of the correction, the deformable mirror changed shape to minimize the wavefront RMS value. Closed loop adaptive optics correction, but ignoring defocus, was accomplished by driving each relevant actuator with a voltage which was obtained from the reconstructed aberration (Zhang & Roorda, 2006) and the "deflection calibration parabolic curve" for single actuators provided by the manufacturer. A loop gain of 0.1 to 0.15 was used because each actuator's response is influenced by the positions of its adjacent actuators. Those actuators out of the active area remained at their initial voltages, except for the actuators just adjacent to the active area which were given voltages related to the voltage(s) of their neighbour actuator or actuators in the active area. Usually 12-16 loops were needed to minimize the RMS wavefront aberrations of the eye and the optical system. The mirror's shape was now maintained for the experimental session. In the remaining approximately 3 s of the 5-6 s period, the wavefront aberrations continued to be measured and the residual aberration co-efficients and the residual RMS were taken as averages across this time. The residual RMS (excluding defocus) was lower than 0.10 μm in nearly all cases.

Measurements were repeated at the end of sessions with the trombone at its reference position. In all cases the residual RMS was no more that 20 % greater than the pre-experiment value.

2.4. Visual acuity and contrast sensitivity measurements

Each experiment was conducted first without and then with adaptive optics. After measuring the eyes' aberration, the subjects adjusted the optical trombone to best focus for a 0.25 logMAR E letter. The average of 6 determinations was used as the in-focus reference position relative to which defocus was altered. For adaptive optics correction, the radiation source was blocked after the mirror reached its desired shape. For subject DAA, a correcting cylindrical lens -0.25 DC x 70 was placed next to the stop of the visual stimulus channel to correct his small astigmatism during the visual acuity experiment. This trial lens was removed when adaptive optics correction was applied.

For visual acuity, each subject performed a four-alternative forced choice illiterate E experiment for high (95 %) and low (12 %) Michelson contrast at 8 cd/m^2 background luminance. The subjects pressed one of four buttons on a small signal box to indicate letter orientation following a 1s presentation. 160 presentations were given in a run across a 0.4 log unit range (5 presentations × 4 orientations × 8 sizes). The data were fitted with a maximum likelihood estimate methods and the 62.5% probability level was taken as the visual acuity (50% with correction for guessing). Following threshold determination at the in-focus position, the optical trombone was moved in the negative defocus direction (optical path length reduces, simulating hypermetropia) followed by the positive direction. Approximately 3 sets of measurements were averaged for each defocus out to ±1.33 D.

Contrast sensitivity for subject DAA and horizontal gratings was performed at 35 cd/m^2 mean luminance at spatial frequencies of 2.5, 5, 10, 20 and 30 cycles/degree using 0.5° Gabor patches (0.5° was the angle from the centre of the pattern by which contrast reduced to 60.6% of the central value). Each stimulus was presented for 1 second in the form of a temporal square wave function. We used a visible/no-visible choice staircase algorithm to determine the threshold. The subject’s task was to press one of two buttons depending upon whether or not the grating was visible. The button press triggered the next presentation. The initial contrast for all spatial frequencies in a run was -0.4 log unit. If initially visible, the contrast decreased in 0.4 log steps until the grating was not visible, whereupon the contrast changed in 0.2 log units until it was again visible. From the next reversal, step size was 0.1 log unit. The first two reversals for a spatial frequency were ignored and the mean was taken as the average of 6 subsequent reversals.

A set of measurements was taken for each spatial frequency across the focus range, in order of lowest to highest spatial frequency. For each defocus and spatial frequency, results of two runs were averaged. Where the variability between the runs was greater than 0.2 log unit, an additional run was made. For the majority of defocus/spatial frequency combinations, standard deviations were < 0.1 log unit.

2.5 Point spread function simulations and contrast sensitivity predictions

For simulations and predictions, corrections had to be made to the measured aberrations. Defocus was corrected according to (Atchison, 2004)

$$\Delta C_2^0 = (2\Delta l/0.15^2 + 0.79 - 0.3)2.8^2/(4\sqrt{3}) \qquad (2)$$

where Δl is the movement of the trombone (m) from its position at which aberrations were measured to the in-focus position chosen by a subject, 0.15 m is the focal length of lens L_1, 0.79 D is a combination of the -0.07 D chromatic aberration of the system and of the 0.86 D chromatic focus difference of the eye according to Thibos et al. (Thibos, Ye, Zhang & Bradley, 1992) between 830 nm and 550 nm, 0.3 D is the inverse of the stimulus distance, 2.8 mm is the pupil semi-diameter used for measurement, and $2.8^2/(4\sqrt{3})$ converts from a longitudinal defocus to a defocus coefficient.

Astigmatic co-efficient corrections were applied for the -0.25 x 70 lens subject DAA used in the visual acuity experiment as (Atchison, 2004)

$$\Delta C_2^{-2} = (0.25/2)\sin(140)2.8^2/(2\sqrt{6}) \quad \Delta C_2^{2} = (0.25/2)\cos(140)2.8^2/(2\sqrt{6}) \qquad (3)$$

All aberration coefficients C_z, except for defocus, were then corrected from 830 nm to 550 nm using

$$C_{z(550)} = C_{z(830)}[(n_{550} - 1)/(n_{830} - 1)] \qquad (4)$$

where n_{550} and n_{830} are estimated equivalent refractive indexes of the eye for 550nm and 830nm (Thibos et al., 1992). This correcting equation is used by Wavefront Sciences with their aberrometers, and makes change to the aberrations between 830 nm and 550 nm of only 2%, which is consistent with experimental studies finding little change in aberrations between the infrared and visible wavelengths (Llorente, Diaz-Santana, Lara-Saucedo & Marcos, 2003; Marcos, Burns, Moreno-Barriuso & Navarro, 1999). Finally, as aberrations were measured for 5.6 mm stop, compared with the 5.5 mm stop for the visual stimulus channel, all aberration coefficients were interpolated to 5.5mm (Campbell, 2003).

Aberrations were used for point spread function simulations and modulation transfer function predictions with the aid of the optical design program Zemax-EE (Zemax Development Corporation). DAA's Stiles-Crawford function had recently been determined with a two-channel Maxwellian viewing system (Atchison & Scott, 2002) as

$$\exp[-0.14(x - 0.3)^2 - 0.10(y + 0.6)^2]$$

where (x, y) are the pupil co-ordinates relative to pupil centre, and this was included for this subject as a pupil apodisation for determining the point spread function and modulation transfer function. Contrast sensitivity was then predicted for DAA according to equation (1).

3. Results

3.1. Aberration measurements and corrections

Figure 2 shows the two subjects' in-focus wave aberrations maps and point spread functions without and with adaptive optics. The root mean squared (RMS) of wavefront aberrations reduces from 0.45 μm to 0.09 μm for DAA and from 0.36 μm to 0.08 μm for BJB following AO correction.

Figure 3 shows through-focus point spread functions. Without adaptive optics correction, the point spread functions show considerable asymmetry about the in-focus position focus, as has been previously shown by Wilson, Decker & Roorda (2002). The spread is much greater with positive defocus that with negative defocus. The symmetry of the point spread functions is much improved with adaptive optics correction.

3.2. Visual acuity

Figure 4 shows visual acuity as a function of defocus for subjects DAA (top) and BJB (bottom), for high and low contrast and without and with adaptive optics. Positive defocus means that defocus is produced as if a positive lens were placed in front of a emmetropic eye. The error bars represents standard deviation for 2-3 runs. Without adaptive optics, in-focus high contrast visual acuity was better than in-focus low contrast visual acuity by approximately 0.2 logMAR, and in general visual acuity for both contrasts deteriorated at similar rates away from in-focus. Visual acuity deteriorated more quickly for positive focus than for negative focus.

Application of adaptive optics improved in-focus visual acuity by approximately 0.1 to 0.15 logMAR (1.2 to 1.4x), which is similar to the 1.2x and 1.4x average improvement in 7 subjects found by Yoon and Williams (Yoon & Williams, 2002) for 6 mm pupils at 20 cd/m^2 and 2 cd/m^2 under white light conditions. The deterioration of visual acuity was more symmetric about in-focus, and generally at a greater rate, than without adaptive optics. Except for the combination of positive defocus and low contrast, at sufficient levels of defocus visual acuity became worse with adaptive optics than without adaptive optics.

The more rapid deterioration of visual acuity away from in-focus with adaptive optics than without it is most apparent in the region around in-focus. For example, at high contrast without adaptive optics, subject DAA has a 0.6 D defocus range over which visual acuity varies by less than 0.04 logMAR. Over the same range, visual acuity with adaptive optics varies by about 0.10 logMAR.

3.3. Contrast sensitivity

Figure 5 shows contrast sensitivity at different spatial frequencies as a function of defocus for subject DAA. The left and right columns show results with and without adaptive optics, respectively. Measurements are given by the solid plots with symbols and predictions are given by the dotted plots.

Without adaptive optics, DAA show a shift in the contrast sensitivity peak in the negative (hyperopic) direction as spatial frequency decreases; this can be explained by a large positive spherical aberration (Zernike C_4^0 co-efficient = +0.17 µm at 5.5 mm pupil) and is consistent with previous through-focus measurements (Atchison & Scott, 2002; Atchison, Woods & Bradley, 1998). In general, the predictions are similar to the measurements, although the predicted notches are sometimes deeper and there are some notches that appear in the predictions but not the measurements (eg 20 c/d in the top left figure). In addition, the predictions without adaptive optics are displaced to the positive defocus side by up to 0.4 D. The agreement between measurements and predictions is better for negative than for positive defocus, consistent with our previous studies

With adaptive optics correction, the in-focus sensitivities are higher than without adaptive optics by 0.06 log unit (2.5 c/d), 0.11 log unit (5 c/d), 0.11 log unit (10 c/d), 0.25 log unit (20 c/d) and 0.39 log unit (30 c/d). These improvements are similar to those found by Yoon and Williams

(2002) for 6 mm pupils under white light conditions of 0.4 and 0.3 log units (at 24 c/d for two subjects) and 0.2 log unit (at 32 c/d for one subject) All peaks occurred within 0.1 D of the in-focus position, and measurements were much more symmetrical about the in-focus position than occurred without adaptive optics. Some notches with adaptive optics were very deep eg the 1.3 log unit notch for 20 c/d (top right). Predicted peaks coincide closely with the measured peaks. The agreement between prediction and measurement was better for positive defocus than was the case without adaptive optics, but the agreement was poorer for negative defocus. In particular, the loss in contrast sensitivity with defocus is not as nearly marked as predicted.

The results in Fig. 5 are for vertically gratings. Because of asymmetries in aberrations, it is expected that the effects of adaptive optics correction on contrast sensitivity would be different for other grating orientations.

4. Discussion

In this study we determined the influence of adaptive optics correction on through-focus visual acuity and contrast sensitivity under monochromatic conditions for two subjects. Defocus steps were as small as 0.09 D so that we could accurately track nuances in visual performance, particular for contrast sensitivity, but overall trends were similar to previous studies using bigger steps (Piers et al., 2004; Piers et al., 2007). Adaptive optics correction improved in-focus high (95 %) and low (12 %) contrast visual acuity by 0.1 to 0.15 logMAR, but it also caused more rapid and more symmetrical deterioration in visual acuity away from in-focus. Adaptive optics correction improved in-focus contrast sensitivities and caused a more symmetrical and greater loss of contrast sensitivity about the peak sensitivities. The results demonstrate that correction of astigmatism and higher order aberrations may worsen spatial visual performance in the presence of some defocus.

The results of the study have implications for the corrections of ocular aberrations, such as might become possible with ophthalmic instruments and which is a longed-for goal of static corneal refractive surgery and custom contact lens correction. If it is possible to compensate for all aberrations except defocus, people may become less tolerant of defocus such as might occur with a lag or lead of accommodation or with a residual refractive error. In other words, the depth-of-focus, which may be defined as the vergence range of focusing error which does not result in objectionable deterioration in image quality, may be reduced. The reduction in depth-of-focus is for at least two reasons. Firstly, the vision may be poorer than when aberrations are available to "dampen" the effects of defocus. Secondly, if a person has become used to better in-focus spatial vision that they previously experienced, they may be aware of the more rapid loss of vision away from in-focus. We are currently investigating subjective depth-of-focus under different aberration correction conditions (chromatic and monochromatic).

Reduction in depth-of-focus will depend on the nature of the stimulus. For example, for simple grating targets we found that the adaptive optics induced changes in the rate of contrast sensitivity loss in opposite directions for positive and negative defocus, with loss increased and decreased for negative defocus and positive defocus, respectively (Figure 5). However for the more complex illiterate-E targets (and possibly by extension to Snellen letters) the loss of visual acuity was increased with adaptive optics in both defocus directions (Figure 4). It may be that changes in the phase transfer function rather than only losses of contrast sensitivity are partly responsible for this.

Changes in spatial vision with adaptive optics correction in the study were marked, and as noted in the Results section, in broad agreement with the in-focus results of Yoon & Williams (2002) for visual acuity and contrast sensitivity. However as noted by those authors the changes were less than those predicted for complete correction of aberrations. We were able to correct the aberrations (other than defocus) to within 0.07 μm to 0.10 μm RMS for 5.6 mm pupils. It is possible during some measurements that these increased but it was not feasible to check aberrations during the lengthy measurements. Both subjects were experienced psychophysical observers with little movement during measurements, and measurement checks at the end of sessions showed little change in residual aberrations. Like other researchers in the field, we will be endeavouring to improve the quality of our adaptive optics system. In addition to the limitations in the AO system, the predictions in the CSF did not always match the experimental results closely, and this is likely to be due at least partly to limitations in the quality of our projector-screen stimulus system and the effect of small unintended eye movements over one second presentation times.

Our experiments were conducted in monochromatic light rather than broad spectrum radiation, thus negating the effects of longitudinal and transverse aberration. A more thorough analysis could have included white light conditions. Introducing chromatic aberrations reduces in-focus acuity and contrast sensitivity (Yoon & Williams, 2002) and increases subjective depth of focus (Campbell, 1957). If white light targets had been used, we would have expected less distinct differences without and without adaptive optics, particularly regarding the flucuations in contrast sensitivity..

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Figure Captions

Figure 1. Instrumentation for experiment. See text for further details.

Figure 2. Wave aberration maps and simulated point spread functions both without and with adaptive optics correction for the two subjects at in-focus. RMS aberrations are indicated. The aberration maps include second and higher-order aberrations including defocus and take into account trial lens correction of subject DAA during visual acuity measurements without adaptive optics. The point spread function image sizes are 15.6 min. arc x 15.6 min.arc. Pupil size 5.5 mm. For DAA, the most important aberrations without adaptive optics correction were defocus (C_2^0= 0.32 μm), vertical coma C_3^{-3} = +0.16μm, and spherical aberration C_4^0 = +0.17μm. For BJB, the most important aberrations without adaptive optics correction were defocus (C_2^0= 0.27 μm), astigmatism C_2^2 = -0.17μm, and spherical aberration C_4^0 = +0.14μm.

Figure 3. Through-focus simulated point spread functions for subject DAA without and with adaptive optics correction. The point spread function image sizes are 20.6 min. arc x 20.6 min.arc Other in-focus details are as for Figure 2. Pupil size 5.5 mm.

Figure 4. Through-focus visual acuity (logMAR) at high and low contrasts both without and with adaptive optics correction for the two subjects. Positive defocus means that defocus is produced as if a positive lens were placed in front of an emmetropic eye. Error bars show standard errors across runs. Pupil size 5.5 mm.

Figure 5. Through-focus contrast sensitivities for subject DAA both without adaptive optics (left) and with adaptive optics correction (right). Positive defocus means that defocus is produced as if

a positive lens were placed in front of an emmetropic eye. Spatial frequencies are indicated. Pupil size 5.5 mm.

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