Comparison of Monochromatic Ocular Aberrations Measured with an Objective Cross-Cylinder Aberroscope and a Shack-Hartmann Aberrometer

XIN HONG, PhD, FAAO, LARRY N. THIBOS, PhD, FAAO, ARTHUR BRADLEY, PhD, RUSSELL L. WOODS, PhD, FAAO, and RAYMOND A. APPLEGATE, PhD, FAAO

School of Optometry, Indiana University, Bloomington, Indiana (LNT, AB), Alcon Laboratories, Ft. Worth, Texas (XH), Schepens Eye Research Institute, Harvard Medical School, Boston, Massachusetts (RLW), College of Optometry, University of Houston, Houston, Texas (RAA)

ABSTRACT: Repeated measures of wavefront aberrations were taken along the line-of-sight of seven eyes using two instruments: an objective, cross-cylinder aberroscope (OA) and a Shack-Hartmann (SH) aberrometer. Both instruments were implemented on the same optical table to facilitate interleaved measurements on the same eyes under similar experimental conditions. Variability of repeated measures of individual coefficients tended to be much greater for OA data than for SH data. Although Zernike coefficients obtained from a single measurement were generally larger when measured with the OA than with the SH, the averages across five trials were often smaller for the OA. The Zernike coefficients obtained from the two instruments were not significantly correlated. Radial modulation-transfer functions and point-spread functions derived from the two sets of measurements were similar for some subjects, but not all. When average Zernike coefficients were used to determine optical quality, the OA indicated superior optics in some eyes, but the reverse trend was true if Zernike coefficients from individual trials were used. Possible reasons for discrepancies between the OA and SH measurements include difference in sampling density, quality of data images, alignment errors, and temporal fluctuations. Multivariate statistical analysis indicated that the SH aberrometer discriminated between subjects much better than did the objective aberroscope. (Optom Vis Sci 2003;80:15–25)

Key Words: aberrometry, aberroscope, visual optics, wavefront aberrations, individual variability

The subjective and objective crossed-cylinder aberroscope has been a mainstay of visual optics research for more than a quarter of a century. The crossed-cylinder aberroscope has been used to describe the aberrations of the normal human eye, changes in aberrations with accommodation, aberrations associated with myopia, aberration changes with aging, and aberration changes with wavelength. These pioneering studies contributed a body of knowledge that is the foundation of our current understanding of the aberration structure of human eyes.

Although the objective aberroscope (OA) technique has been widely used in the past, its accuracy and reliability have been questioned. Calibrations on model eyes with known optical aberrations (i.e., spherical aberration and coma) showed good accuracy, but the accuracy on human eyes appears to be limited. Cox and Walsh evaluated the intrainmage variability and interimage variability and found that standard deviations could be large for individual aberration coefficients. Smith et al. and Cox and Calver assessed theoretically the accuracy of the crossed-cylinder aberroscope technique and suggested that a variety of factors related to the operation of the aberroscope and data analysis could lead to inaccurate measurements. Some of those factors could be eliminated by careful design of the optical system and data analysis. Others, such as difficulty determining grid intersection points and the low sampling density across the pupil, were inherent in the aberroscope design and might be difficult to resolve.

In recent years, the Shack-Hartmann (SH) aberrometer has become a popular method for ocular aberrometry which has been independently validated against a variety of techniques. Salmon and colleagues demonstrated good agreement between aberrations measured by Smirnov's psychophysical method and the SH aberrometer. Moreno-Barriuso and Navarro compared the SH aberrometer with laser
OA methods were long enough to reduce laser speckle and short
(ANSI) standard. Exposure times of 200 ms for SH and 100 ms for
than 2 log units below American National Standards Institute
laser intensity was controlled by neutral density filters to be more
for OA measurements by manipulating the aperture PH2. The
worse than the mean MTF
relative advantages and disadvantages of both technologies.
and differences between these two approaches and illustrate the
experimental conditions. Our results demonstrate the similarities
subjects was responsible for the discrepancies between two
ability, however, is that individual variability between two groups of
methods.

The present study was designed to directly compare the OA and
SH techniques on the same group of subjects and under the same
experimental conditions. Our results demonstrate the similarities
and differences between these two approaches and illustrate the
relative advantages and disadvantages of both technologies.

METHODS

Apparatus

We implemented a SH aberrometer and an objective crossed-
cylinder aberroscope within one optical system on the same optical
table (Fig. 1). Light from a He-Ne laser (633 nm) was collimated
by spatial filter (L1, PH1, and L2) to provide incoming parallel
light for the eye. The diameter of the incoming light beam was
adjusted to be <2.0 mm for SH measurements and about 8.0 mm
for OA measurements by manipulating the aperture PH2. The
laser intensity was controlled by neutral density filters to be more
than 2 log units below American National Standards Institute
(ANSI) standard. Exposure times of 200 ms for SH and 100 ms for
OA methods were long enough to reduce laser speckle and short
enough to avoid reflex blinking and minimize eye movement dur-
during exposure.

The lenslet array (400-μm spacing in a square array) of the SH
wavefront sensor was rendered optically conjugate to the eye’s
entrance pupil by means of a unit-magnification telescope system
(L4 and L5). The video sensor CCD1 was located at the back focal
plane of the lenslet array. The aberroscope lens, kindly supplied by
D. Atchison, consisted of a uniform, undistorted, square grid
(spacing 1.2 mm) sandwiched between two cylinder lenses with
+5 and −5 D powers and with perpendicular axes. This grid
spacing was in the range used by previous investigators.2, 7–10, 12, 13
The aberroscope was aligned with the measuring axis of the optical
system and was placed a fixed distance before the entrance pupil of
the eye. The sensor CCD2 was located in the back focal plane of a
camera lens that was focusing at infinity. Both CCD cameras in the
system were conjugate to the retina of the subject’s eye.

The optical axes of the SH and the OA measurement channels
were aligned with the eye’s line-of-sight (in object space this con-
nects the fixation point to the pupil center) by the following two-
step procedure. First, the subject fixated a target located on the
combined optical axis of both measuring channels. This placed the
eye’s fixation point on the instrument axis. Second, while main-
taining fixation, the experimenter translated the subject’s eye
vertically and horizontally until the eye’s pupil was concentric
with the instrument axis. This placed the center of the entrance pupil
on the instrument axis. To perform this second step for OA measure-
ments, we used a television monitor and video camera to view the
eye’s pupil simultaneously with an alignment annulus that was
concentric to the instrument axis (Fig. 2A). Alignment was com-
plete when the eye’s pupil and the alignment ring appeared con-
centric on the monitor. To perform this second step for SH mea-
surements, we used the SH data image to monitor pupil position
(Fig. 2B). The center of the SH image provides a good estimate of
the center of the entrance pupil because the spot array is a geo-
metrical mapping of the entrance pupil onto the lenslet array. Align-
ment was complete when this array of spots was concentric with a
reference pixel on CCD1. This reference pixel, representing the
intersection of the instrument axis and sensor plane, was estab-
lished with the aid of a mirror that reflected a narrow beam of light
along the measurement axis. (The mirror was removed when mea-
suring eyes.)

Subjects

This study followed the tenets of the Declaration of Helsinki.
Subjects were informed of the nature and risks of the study, and
each subject signed an informed consent statement approved by
the institutional review board at Indiana University. The right eyes
of seven observers were measured. All observers had normal vision
with low or mild refractive errors. Spherical refractive errors ranged
from −0.5 to −3.5 D, and cylindrical refractive errors ranged
from 0 to −3 D. To help expose the higher-order aberrations for
study, these manifest refractive errors were corrected with trial
lenses during the experiment. For OA measurements, an addi-
tional spectacle with positive power was used to compensate for
propagation of the grid pattern from aberroscope plane to the
entrance pupil plane.1 The required amount of compensation was
estimated from simple vergence equations, but the final value was

FIGURE 1.
The setup of an optical system with both objective aberroscope and
Shack-Hartmann wavefront sensor. L, lens; M, mirror; CB, cube beam-
splitter; PB, pellicle beamsplitter; PH, pinhole; NDF, neutral density filter;
SH, shutter.
decided by optimizing the rectilinear appearance of the retinal grid. Any residual errors in the determination of this compensation lens would be expected to affect primarily the second-order aberration coefficients, not the higher-order terms.

To preserve physiological conditions, no drugs were used during the experiments. A dental-impression bite bar stabilized head position and the optically distant fixation target helped relax accommodation. Five SH measurements were interleaved with five aberroscope measurements. After each measurement, the subject moved to the other side of the optical table, which then required small adjustments in alignment before taking the next measurement. This experimental paradigm allowed us to evaluate the between-trial repeatability of each instrument as well as between-instrument correlations.

Data Analysis

Typical raw data images are shown in Fig. 3. Aberration information is contained in the displacement of intersection points (OA) (Fig. 3a) or dots (SH) (Fig. 3b) away from the ideal positions. The difference in clarity evident in these two images provides a visual indication of their relative ability to yield reliable measurements for our implementation of these two methods.

The maximum pupil diameter common to all eyes was 5.2 mm, which was chosen for all the analysis reported below. Raw images were reduced to the required size by multiplication with a numerical mask. For this pupil size, the OA images usually had four sampling points horizontally across the pupil, whereas the SH images had 13 sampling points. Spacing between lines in the retinal grid of OA images was approximately 40 pixels, and spacing between dots in SH images was roughly 60 pixels.

The procedures for reconstructing the wave aberrations from raw data were similar for the two methods. We first located the intersection points of grid lines (OA) or centroids of dots (SH) and then fit their displacements from ideal positions with derivatives of Zernike polynomials. The formulas relating the displacement of intersection points with the wavefront aberrations were described by Howland and Howland and Smith et al., and the formulas for the displacements of centroids and aberrations were detailed by Liang et al. Zernike fitting was performed for each raw data image using a least-squares algorithm. To avoid numerical problems in the fitting procedure, the number of Zernike polynomials should be less than the number of sampling points within the desired pupil. Thus, the sampling density of the instrument effectively limits the number of Zernike modes that can be fitted to the raw data. For OA data, the Zernike coefficients were fitted up to 15 modes (fourth order). For SH data, the Zernike coefficients were fitted up to 66 modes (10th order), but only the first 15 modes were used for comparison with OA results. Nomenclature used in this paper to identify the Zernike coefficients follows the Optical Society of America-recommended standards.

Aberroscope data were fit with Zernike polynomials as reported previously. This Zernike fitting approach is different from the earlier approach of fitting Taylor polynomials and has some advantages over the latter, e.g., the orthogonality of each mode and the simplicity of calculating the wavefront variance. To verify our Zernike fitting software, we used the fitted coefficients to reconstruct the retinal grid for comparison with the original data image. Two examples of reconstructed retinal grids are shown in Fig. 4. The intersection points from the original retinal grid are shown as crosses superimposed on top of the reconstructed retinal grid. Visual assessment confirms the accuracy of our Zernike fitting procedures. The root mean squared (RMS) fitting error of intersection points between measured and reconstructed is within 1 pixel for all eyes participating in our experiments. Considering the width of blurred lines (several pixels) and limited fitting modes (only to
fourth order), we judged the algorithm for Zernike fitting and reconstruction to be adequate for quantifying ocular aberrations. By comparison, the determination of centroids of SH images normally has subpixel resolution.

**FIGURE 3.**

Analysis of the grid pattern in aberroscope images was performed with NIH Image software written for the cross-cylinder aberroscope. All other data analysis was performed in Matlab (Mathwork). The programs were tested in connection with custom software that simulates the retinal grid pattern from known aberration coefficients.

**FIGURE 4.**
Examples of reconstructed retinal grids and intersection points from original retinal grids. “×” represents the intersection points from original retinal grids, “+” stands for the centers of retinal grids. The grids are reconstructed from Zernike coefficients.
Calibration

Instrument calibration was validated with a model eye constructed by using a commercially available doublet with known aberrations (E32309, Edmund Scientific) as the optical component and a rotating diffuser as the retina. Aberrations of the model eye were calculated from the manufacturer’s specifications with the help of the commercial lens design program OSLO (Sinclair Optics). The RMS spherical aberration of this model eye was predicted to be 0.0868 μm for a 6.4-mm pupil. Both instruments gave results that were extremely close to the theoretical prediction: 0.0899 ± 0.0055 μm for OA data and 0.0858 ± 0.0005 μm for SH data (mean ± SD) for a 6.4-mm pupil. As predicted, the other high-order aberrations had nearly zero values. We conclude that both instruments accurately measure the aberrations of a commercial glass lens.

RESULTS

Examples of reconstructed wavefronts from three eyes (subjects RW, SD, and XQ) are shown in Fig. 5. The left panel shows wavefronts derived from individual OA measurements, and the middle panel is from individual SH measurements. Visual inspection indicates that the wavefronts are qualitatively different. These differences are quantified in the magnitudes of the corresponding Zernike coefficients (histograms on right side of Fig. 5). As these examples show, individual coefficients derived from individual OA and SH measurements can be very different.

To explore the correlation of results obtained by the two instruments, Zernike coefficients obtained for five pairs of repeated measurements on the same eye are displayed in scattergrams in Fig. 6. If the two instruments produced identical results, every pair of measurements would fall on the y = x diagonal line. It is clear from a visual inspection of this figure that paired measurements from the two instruments rarely agreed. From 98 comparisons (14 modes, seven subjects), only three instances of significant correlation between the OA and SH aberrations were observed at the p ≤ 0.05 level, and none at the p ≤ 0.02 level (t-test). To reveal the differences in measurement variability of the two instruments, we also display in Fig. 6 the 95% probability ellipses for each Zernike mode. Assuming each pair of measurements is drawn from a bivariate Gaussian distribution, 95% of repeated measurements should fall inside the ellipse. Notice that most ellipses are elongated vertically, which implies that OA measurements were more variable.

[FIGURE 5.](Comparison of Ocular Aberrations Methods—Hong et al. 19)

Pairs of wave aberration contour maps derived from successive objective aberroscope (OA) and Shack-Hartmann (SH) measurements on three eyes (subjects RW, SD, and XQ are shown in panels A, B, and C, respectively). The interval between contour lines is 0.05 μm. Right-hand panels show histograms of the values of each Zernike coefficient (micrometers). Filled bars are objective aberroscope and open bars are Shack-Hartmann.

[FIGURE 6.](Comparison of Ocular Aberrations Methods—Hong et al. 19)

Scattergrams of Zernike coefficients (in micrometers) derived from the objective aberroscope (OA) measurements (ordinate) are plotted against the values from immediately consecutive measurements taken with the Shack-Hartmann (SH) aberrometer (abscissa). Each graph represents five repetitions on both instruments for one eye. Panels A and B show data from RLW and AB, respectively. Ellipses indicate the region where 95% of repeated measurements are expected to fall. The major axis of the ellipse represents the first principal component, which is also the least-squares regression line.
than SH measurements. For example, the average standard deviations of repeated measurements of fourth-order Zernike coefficients were five times greater for OA than for SH. The major axis of each probability ellipse represents the least-squares regression line for the data. It was rare for these regression lines to have a slope near 1.0, which is further evidence that paired measurements on the two instruments rarely agreed.

Because each eye was measured five times on each instrument, we averaged the Zernike coefficients across trials to obtain a mean vector of Zernike coefficients for each instrument on each subject. Correlations between instruments for these mean values are shown in Fig. 7 for all seven subjects. Of these 14 comparisons, only one (vertical prism: \( N = 1, f = -1 \)) had a statistically significant correlation coefficient at the \( p = 0.05 \) level. Repeated measures, two-way analysis of variance (subjects and instruments as factors) conducted on individual modes confirmed that differences between instruments are statistically significant (\( p < 0.05 \)) for all but two of the third- and fourth-order modes. Zernike coefficients of the same order were then combined to yield a measure of RMS error for each trial. Results from two-way analysis of variance on these data indicated that RMS error measurements from the two instruments were significantly different for second and third orders, but not for the fourth order.

Contrary to the trends in Fig. 6, the probability ellipses in Fig. 7 tend to be elongated horizontally. This implies that mean aberration coefficients were different for different subjects when measured with the SH, but not when measured with OA. Horizontal ellipses also imply that the slopes of the regression lines are different from the expected value of 1.0. Taken together, the results of Figs. 6 and 7 indicate that the OA does not discriminate between subjects as well as the SH for two reasons. First, repeated measures on the same subject are highly variable, and, second, average measurements on different subjects are much the same. These results suggest that single OA measurements are unreliable and may overestimate the actual level of aberrations present. To the contrary, the average of repeated measures with the OA are often close to zero, which suggests that the OA has a tendency, on average, to underestimate some aberrations compared with the SH measurements.

If the two instruments gave the same readings, then the difference between Zernike coefficients for any given mode should be zero. We tested this hypothesis by examining the statistical distribution of the difference between paired measurements (SH - OA) obtained for individual coefficients. The means, 95% confidence intervals for the means, and the standard deviations of paired differences for all trials on all subjects are presented in Fig. 8. There is no evidence from this figure that one instrument consistently measures more or less aberration than the other. The SH instrument consistently measured larger amounts of some aberrations, but smaller amounts of other aberrations, and in some cases, the OA and SH-measured coefficients had opposite sign. Differences were statistically significant for several modes, most notably oblique astigmatism (\( a_3^2, #3 \)), spherical defocus (\( a_0^2, #4 \)), quadrafoil (\( a_{4}^4, #10 \)), and spherical aberration (\( a_4^4, #12 \)). Hotelling’s \( T^2 \) test rejected the null hypothesis that both instruments gave the same vector of second- through fourth-order coefficients with very high confidence (\( p < 0.0001 \)). The same result was obtained for the vector of third- through fourth-order coefficients.

Radial-Averaged Modulation Transfer Function (rMTF)

The MTF, which evaluates the overall effect of wave aberrations on the eye’s optical quality, was calculated from the third- and

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**FIGURE 7.**
Zernike pyramid of scatter-grams plotting the mean (\( N = 5 \) trials) coefficient values determined from the objective aberroscope (OA) and Shack-Hartmann (SH) methods. Each symbol represents a different eye. Vertical and horizontal lines (often too short to be visible) show \( \pm 1 \) SEM for each subject.

**FIGURE 8.**
Paired differences between Shack-Hartmann (SH) and objective aberroscope (OA) coefficient values. Vertical extent of individual rectangles indicate 95% confidence intervals (c.i.) for the mean (horizontal line through center of rectangle). Error bars indicate \( \pm 1 \) SD of 35 paired differences (five pairs on seven subjects).
fourth-order Zernike coefficients. Two averaging methods were used to identify the MTF for each eye. First, MTF’s were calculated from Zernike coefficients from each individual trial, and these MTF’s were averaged (Fig. 9). These MTF’s reflect the level of aberration reported on any given trial (Fig. 6), and because the OA aberrations were often quite large on a single trial, the MTF’s calculated from the OA measurements are significantly lower for some eyes (e.g., see the radially averaged MTF’s for subjects DM, RW, and XQ in Fig. 9). The error bars, which indicate ±1 SEM across trials, clearly indicate more between-trial variability for OA results than for SH results for all subjects.

The second averaging method was to average coefficients across trials, and the MTF’s were then calculated from these mean values. By this method the radial MTF’s derived from the OA were frequently much better than those from the SH measurements (Fig. 10). This reflects our observation that the mean coefficient values were often at or around zero for the OA measurements (Fig. 7). Interestingly, although we found little evidence of any correlation between individual modes, the resulting MTF’s derived from the OA and SH are very similar in most eyes (Figs. 9 and 10).

The population-average MTF’s computed across seven eyes using both instruments are shown in Fig. 11. As expected from the results shown in Figs. 9 and 10, when MTF’s are calculated from

**FIGURE 9.**
Radial-averaged modulation transfer functions and point-spread functions calculated from objective aberroscope (OA) and Shack-Hartmann (SH) measurements for seven subjects. Left: radial-averaged modulation transfer functions were calculated from each OA (solid line) and SH (dashed line) trial and averaged. The error bars represent the SEM of radial modulation transfer functions. Right: The size of point-spread functions for each image is 6.7 arc min × 6.7 arc min. The intensity of the image was normalized to the maximum intensity of each image to illustrate the details of each point-spread function. The calculations of radial-averaged modulation transfer functions and point-spread functions only include the third- and fourth-order Zernike modes.

**FIGURE 10.**
Radial-averaged modulation transfer functions and point-spread functions derived from averaged (across five repeat trials) Zernike series. All graphical details are the same as Fig. 9.
aberration measurements on individual trials, the average OA rMTF’s are slightly lower than those calculated from the SH measurements (Fig. 11B). However, the reverse is true when individual Zernike coefficients are averaged across trials and the MTF’s are calculated from these average values. In this case, the OA produces slightly higher rMTF’s (Fig. 11A). Although on average the SH MTF’s are slightly higher, the two functions are not statistically different from each other for a wide range of spatial frequencies, which suggests that individual variability is greater than the variability between the OA and SH techniques.

Point-Spread Functions

Another way to compare the optical performance inferred from the two different aberrometry methods is to examine the point-spread function (PSF) calculated from their respective Zernike coefficients of third and fourth orders. The results of this exercise, shown in Figs. 9 and 10, indicate that PSF’s computed by the two methods are generally similar in overall size, but often quite different in appearance. Extra lobes in the PSF’s are typically associated with phase reversals in the eye’s optical transfer function. Phase reversals are not represented in the radial MTF’s, which might explain why the MTF’s for the two aberrometry methods agree even though their PSF’s do not appear the same.

DISCUSSION

Comparison of Results with the Literature

Our OA results are consistent with previously published aberration measurements with the objective aberroscope. In our study, for a 5.2-mm pupil, the RMS wave variance averaged across all trials was $0.104 \pm 0.030 \mu m$ (mean ± SD) for third-order aberrations and $0.128 \pm 0.10 \mu m$ for fourth-order aberrations. Walsh et al. reported that for a 5.0-mm pupil, RMS wave variance was $0.140 \pm 0.40 \mu m$ for third-order aberrations and $0.066 \pm 0.031 \mu m$ for fourth-order aberrations based on their 11 subjects. The average RMS wave variance was also found to be $0.120 \pm 0.62 \mu m$ for third order and $0.056 \pm 0.025 \mu m$ for fourth order in 10 subjects in the study conducted by Cox and Walsh. Thus, although the third-order RMS wave variance from our study is in good agreement with the literature, our fourth-order RMS measurements with the OA were approximately twice that reported previously. Between-trial variability (Fig. 6) and the slightly larger pupil size in our study could have contributed to this discrepancy.

The between-trial variability of our OA data are similar to those reported by Cox and Walsh (see their Table 4). In their study, the between-trial SD’s were 38% and 37% of the mean for third- and fourth-order RMS, respectively. Our between-trial SD’s for OA measurements were on average 33% and 43% of the third- and fourth-order mean RMS. Our SH data were, however, more consistent, with average between-trial SD’s of 24% and 15% of the mean RMS for the third- and fourth-order aberrations, respectively. The better test-retest reliability of the SH can also been in Fig. 6.

In Fig. 12, the 50th and 75th percentile MTF’s are adapted from published results on 55 normal eyes. The pupil size used in that study was 5.0 mm, only slightly smaller than the 5.2-mm pupil we used in our study. The 50% curve indicates the radial modulation transfer function of the eye with optical quality that is better than 50% of eyes in their study group, and the 75% curve is better than 75% of eyes. The averaged radial MTF’s of our eyes fall between these two population curves, which is additional evidence that our OA results are typical of those reported in the literature.

Possible Reasons for Low Correlation of Zernike Coefficients Obtained with the OA and SH Instruments

Our results provided little evidence of positive correlation between individual Zernike aberration coefficients measured by two different methods on the same eye (Figs. 5 and 6). Furthermore, differences between eyes were not equally reported by the two methods (Fig. 7). In short, we found little evidence that the two...
instruments were measuring the same thing. Below we discuss four possible reasons for this result: (1) low sampling density, (2) poor image quality of retinal grid, (3) alignment errors, and (4) temporal fluctuations.

Both aberrometry methods sampled the optical wavefront in the pupil plane. However, the sampling density for the OA method was considerably lower than for the SH method. Low sampling density can introduce inaccuracies for theoretical as well as numerical reasons. Theoretically, the sampling density must be sufficient to capture any high-frequency, high-order aberrations present in the eye. Otherwise, aliasing of the optical wavefront will occur because undersampled components are mistaken for lower-frequency, lower-order modes. Numerically, the fitting of experimental measurements with spatial derivatives of Zernike polynomials yields coefficients with a degree of uncertainty. Such errors can be reduced by oversampling the wavefront. Unfortunately, the optical quality of the human eye limits our ability to increase the sampling density of the objective aberroscope. We tried to increase the sampling density of our system to a 6 × 6 retinal grid for a 5.2-mm pupil, but found that useable data images could be obtained only for the eye with the best optics in our study group. Thus the problem of low sampling density appears to be an inherent limitation to obtaining reliable aberration coefficients in the OA method.

A fundamental limitation of the OA method is that the aerial image of the distorted retinal grid is imaged through the full pupil. Therefore the quality of the data image is degraded by any uncorrected aberrations of the subject’s eye. This loss of image quality hampers the accurate determination of grid intersection points from which the wave aberrations are computed, which leads to uncertainty in derived aberration coefficients. To estimate the magnitude of this effect, Smith et al. introduced an uncertainty of 1/10 grid element width to intersection points of the retinal grid. Their simulated result showed that the standard deviations of measured coefficients varied from 17% to 315% of the mean values. A blurred retinal image could also result in errors in determining the grid center, which would introduce coma, and grid magnification, which would affect the magnitude of the aberration estimates.

Alignment and centration errors caused by involuntary eye movement might have contributed to the discrepancy between aberration structures measured by the two instruments. If the eye moves transversely in the SH aberrometer, a different set of lenslets will be illuminated, but the pupil coordinates of those lenslets is not in doubt. However, if the eye moves transversely in the OA aberroscope, the sample points in the pupil move to a new, unknown position. All we record is the double-pass image of the retinal grid; the projection of the grid onto the pupil is not monitored. Smith and colleagues pointed out that Howland and Howland were aware of this problem and incorporated in their early programming the selection of the pupil center by the observer.

Because measurements for the OA and SH methods were interleaved over a period of several minutes, physiological changes in the eye’s optical system may have contributed to the low correlation between Zernike coefficients obtained by the two methods. One test of this hypothesis for the future would be to perform the measurements simultaneously so that temporal fluctuations in the eye’s aberration structure would be inconsequential.

**Comparison of Ocular Aberrations Methods—Hong et al.**

This study compared the results obtained from two instruments that we believe to be typical in performance with those reported previously in the literature. However, both instruments could be improved. For example, reducing the grid spacing in the OA method would have three distinct advantages. First, it would reduce the probability of undersampling the wavefront slope, which would otherwise lead to aliasing errors in the reconstructed wavefront. Second, it would permit the calculation of aberration coefficients for terms beyond the fourth order. Third, it would increase the reliability of derived aberration coefficients by making the least-squares fitting of Zernike polynomials an overdetermined problem.

Unfortunately, ocular aberrations limit the utility of finer aberroscope grids. As a first approximation, the angular subtense of the retinal pattern cast by the OA grid is equal to the product of cross-cylinder lens power and pupil diameter. For a ± 5 D aberroscope lens and a 5.2-mm pupil, the retinal image will be 1.5° in

**Discrimination Performance**

One way to judge the relative merit of the OA and SH methods is to assess their ability to discriminate between different eyes. In the statistical analysis of variance, measurement variability is partitioned into between-subject variability and between-trial variability. In our present context, the former is of interest and the latter is measurement noise. Canonical variate analysis is a multivariate statistical method that looks for a new variable space of just a few dimensions that maximizes the difference between subjects. The new dimensions, called canonical variates, are linear combinations of the original variates (Zernike coefficients). If the measured differences between eyes are large compared with the measurement errors, then data points from the same eye will form an isolated cluster in the new space, well separated from clusters associated with other eyes. Conversely, if measurement errors dominate, then canonical variate analysis will fail to separate the data from different subjects into separate clusters. This is a means of assessing the ability of a given method to discriminate between eyes.

Fig. 12 displays the OA and SH measurements for our seven subjects in the two-dimensional space spanned by the two most significant canonical variates. These two variates account for 80% of between-subject variance in the SH measurements and 82% of between-subject variance in OA measurements. Notice that the SH data clearly separate into six nonoverlapping clusters when plotted in this canonical variate space. Each of these clusters corresponds to a different subject, with the exception of one cluster that contains the data from two subjects. By comparison, the OA data appear randomly distributed with no clear separation of measurements from different subjects into different clusters. Thus, we conclude that our SH aberrometer was better at discriminating between subjects than our OA aberroscope and thus would be a preferred instrument to examine differences between subjects or to monitor changes in aberrations within a given eye.

**Technical Limitations**

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Unfortunately, ocular aberrations limit the utility of finer aberroscope grids. As a first approximation, the angular subtense of the retinal pattern cast by the OA grid is equal to the product of cross-cylinder lens power and pupil diameter. For a ± 5 D aberroscope lens and a 5.2-mm pupil, the retinal image will be 1.5° in
diameter. To achieve the same pupil sampling density provided by a SH wavefront sensor with 0.4-mm lenslet spacing would require 13 grid lines. In this case, the retinal object would be a pattern with fundamental frequency equal to 13 lines/1.5°/H11005 8.7 cpd. The aberroscope captures an image of this object formed by the eye using the full pupil. According to Fig. 11, the modulation transfer of a normal eye is high at this fundamental frequency, but would be expected to be much lower for a clinically abnormal eye. Further loss of image contrast results from using a grid made of thin lines that place much of the stimulus energy in the higher harmonic frequencies where contrast transfer of the eye is lower. For this reason, an improved design would shift energy into the lower, fundamental frequency by using a grid with 50% duty cycle (i.e., line width equal to the spacing between lines). A square-wave grid pattern would maximize the object contrast at the fundamental frequency, but phase shifts in the aberroscope image caused by ocular aberrations could corrupt the image and hamper the detection of grid intersection points. Thus a sinusoidal grid pattern might be preferred. Another option for finer sampling of the pupil is to use a stronger cross-cylinder lens to increase the size of the pupil projection on the retina and thus decrease the spatial frequencies in the retinal pattern. However, this approach would risk introducing significant aberrations into the aberroscope itself and it would be less effective at representing optical quality for any single retinal location (such as the foveola).

Other improvements in design of the OA and SH instruments are also possible. For example, the use of an incoherent light source could reduce laser speckle noise, use of confocal optics may reduce the effect of unwanted scatter, and the development of improved algorithms for localizing grid intersection points might significantly improve the performance of both instruments (H. Howland, personal communication). Image processing algorithms can also be used to advantage by improving the quality of the data image before it is analyzed. Improved methods for utilizing all of the information in the data images, not just the intersection points (OA) or spot location (SH), might also yield practical benefits. Ultimately, a consideration of cost, system complexity, and performance will determine which instrument is preferred for any given application.

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Larry N. Thibos
School of Optometry
Indiana University
Bloomington, IA 47405-3680
e-mail: thibos@indiana.edu