Optics of aberroscopy and super vision

Raymond A. Applegate, OD, PhD, Larry N. Thibos, PhD,
Gene Hilmantel, OD, MS

ABSTRACT

This paper (1) reviews the fundamental limits to visual performance imposed by optical imaging and photoreceptor sampling to determine the limits to the potential gains offered by ideal corrections; (2) examines the predicted losses in vision induced by chromatic aberration, phase shifts, typical ocular aberrations, and the gains possible by correcting the monochromatic aberrations of the eye; (3) discusses the principles of aberration measurement in the eye; and (4) presents methods for measuring and classifying monochromatic aberrations of the eye. *J Cataract Refract Surg* 2001; 27:1093–1107 © 2001 ASCRS and ESCRS

For the first time in history, it is possible to clinically measure the optical defects of the eye beyond sphere and cylinder quickly and efficiently in the clinical environment. By all indications, this technology will soon work its way into clinical practice. Several companies currently use or plan to use this technology to quantify

ocular aberrations to design ideal refractive corrections. These optimal corrections are intended to improve the optical quality of the retinal image beyond what is achievable with traditional spherocylindrical corrections. “Ideal” corrections may come in the form of customized corneal ablative corrections, contact lenses, intraocular lenses, or a combination of these.¹ Once the aberrations of the eye are quantified, they can be minimized using adaptive mirrors or other means to improve the ophthalmoscopic view of the fundus. Such techniques are referred to collectively as adaptive optics and are being used to improve the ophthalmoscopic view of the retina down to the level of individual photoreceptors.^{2,3} Similarly, adaptive optics have been used to improve vision and show potential patients the benefit of reducing ocular aberrations.⁴

Currently, it is not clear that ideal corrections can be implemented routinely or that such corrections will routinely result in “super vision”; ie, vision that is significantly improved over that provided by more traditional forms of correction. Surgeries generally increase corneal^{5–8} and total eye⁹ aberrations. Nonetheless, preliminary results with wavefront-guided surgery are encouraging (a decrease in the surgically induced aberrations¹⁰ and in 1 report, a reduction in the normal

Accepted for publication March 28, 2001.

From the Department of Ophthalmology, University of Texas Health Science Center, San Antonio, Texas (Applegate, Hilmantel), and the School of Optometry, Indiana University, Bloomington, Indiana (Thibos), USA.

Supported by National Institutes of Health (National Eye Institute) grants R01-EY08520 (Applegate), R01-EY05109 (Thibos), and an unrestricted grant from Research to Prevent Blindness to the Department of Ophthalmology, University of Texas Health Science Center, San Antonio, Texas.

Dr. Applegate is a consultant to Alcon/Summit/Autonomous and Sarver and Associates. Dr. Thibos is president of Quarrymen Optical and a consultant to Sarver and Associates.

Slack Inc. granted permission to reproduce portions of chapters 6 and 7 from MacRae SM, Krueger R, Applegate RA, eds, Customized Corneal Ablation: The Quest for Super Vision, 2001.

Reprint requests to Raymond A. Applegate, OD, PhD, Department of Ophthalmology, University of Texas Health Science Center, 7703 Floyd Curl Drive, San Antonio, Texas 78284-06230, USA. E-mail: applegate@uthscsa.edu.

ametropic eye's higher-order aberrations after refractive surgery [M. McDonald, MD, "Treatments Based on Wavefront Guided Custom Cornea Corrections," presented at the Refractive Surgery Interest Group Subspecialty Day, American Academy of Ophthalmology, Dallas, Texas, USA, October 2000]]. Even if the optical image is improved, it is not clear that the neural retina will be able to interpret the improved image. The visual system, despite improved retinal images, may have refractive amblyopia.¹

In this article, we (1) review the fundamental limits to visual performance imposed by optical imaging and photoreceptor sampling to determine the limits to the potential gains offered by ideal corrections; (2) examine the predicted losses in vision induced by chromatic aberration, phase shifts, typical ocular aberrations, and the gains possible by correcting the monochromatic aberrations of the eye; (3) discuss the principles of aberration measurement in the eye; and (4) present methods for measuring and classifying monochromatic aberrations of the eye.

Fundamental Limits to Visual Performance

What gains in visual performance are expected with ideal refractive corrections? Improving the optical transfer function (OTF) by removing optical aberrations increases the contrast and spatial detail of the retinal image. In a healthy visual system, the retinal image gains obtained from improving the OTF are reflected in percepts having higher contrast (increased modulation transfer) and sharper edges (decreased phase errors). These 2 components (modulation transfer and phase transfer) of the OTF are pupil-size dependent and key to understanding the optical quality of the retinal image.

Diffraction-Limited Modulation Transfer Function

The importance of pupil size^{1,3,11,12} to modulation transfer can be easily seen by viewing plots of the diffraction-limited modulation transfer function (MTF) of the eye for a variety of pupil sizes (Figure 1). The diffraction-limited MTF displays the ratio of image modulation (contrast) to object contrast for perfect ocular optics as a function of the spatial frequency of a sinusoidal grating. The advantage of using sinusoidal gratings is 2-fold. Sinusoidal objects are transferred to the image plane as

Figure 1. (Applegate) Diffraction-limited MTF for 4 pupil sizes. If the object has 100% contrast, the yellow-shaded area identifies the spatial frequencies and contrast levels over which detection and recognition ("clear vision") are theoretically possible in the normal eye. The gray-shaded area identifies the spatial frequencies and contrast levels over which detection of aliased images is possible. The area above the uppermost curve represents contrast that cannot be achieved in the retinal image. Calculations of the MTFs were made using 555 nm light and methods detailed by Smith equation 11.38.⁴⁶

demodulated (decreased contrast) sinusoidal images of the same frequency. Phase changes are reflected as a shift in the location of the grating image with respect to the object location and do not alter the nature of the sinusoidal luminance pattern or its image modulation. Nonetheless, phase changes are very important and will be discussed further.

The second advantage is that through Fourier analysis, more complex patterns can be constructed as a summation of sinusoidal gratings of various frequencies, orientations, modulations, and phases. Conversely, objects can be decomposed into a series of sinusoidal gratings of various frequencies, orientations, modulations, and phases. Such decomposition allows the modulation of each frequency component of an object to be reduced to reflect the impact of the visual system on object contrast. In turn, these demodulated spatial frequencies permit the reconstruction of an image as the eye would see it. For these reasons, the MTF and the phase transfer function (PTF) are standard measures of optical quality in the optical industry.

Figure 2. (Applegate) Luminance as a function of location for a sinusoidal object and image showing 25% demodulation from object to image with a 100 degree phase shift causing a change in image location.

In Figure 1, the eye's diffraction-limited MTF reveals several interesting facts: Only for a uniform field (0 spatial frequency) is the object modulation (contrast) transferred to the image plane with 100% efficiency (a modulation factor of 1); this efficiency is independent of pupil size. At all other frequencies, the modulation (contrast) transfer ratio is less than 1 and is pupil-size dependent. At higher spatial frequencies, the modulation (contrast) transfer ratio increases as pupil size increases. The cutoff frequency (the spatial frequency at which the modulation goes to 0) increases as pupil size increases.

Diffraction-Limited Phase Transfer Function

The OTF is made up of 2 principle components: the MTF and the PTF. The PTF describes the phase

shift of the image with respect to the object (Figure 2) as a function of spatial frequency. In a diffraction-limited optical system, object location and image location are identical and there are no phase shifts. In an aberrated optical system, phase shifts can become very important and can lead to what is commonly called "spurious resolution" and decreased visual performance. These effects will be discussed.

Retinal Limitations

In addition to optical considerations, there is a fundamental retinal limitation to visual performance: the ability of the photoreceptors to sample the retinal image. Photoreceptors in the foveola are approximately $2\ \mu\text{m}$ in diameter. The response of the photoreceptor to an input signal is graded based on photon capture within the photoreceptor; ie, within a single photoreceptor spatial (shape) information is lost. Consequently, to differentiate the letter *E* from a period, the components of the letter *E* must be distributed over an adequate number of receptors to allow the components of the letter to be detected by separate receptors (Figure 3). The coarseness of the foveolar photoreceptor mosaic limits letter acuity, independent of the quality of the optics, to somewhere between 20/8 and 20/10 (depending on the biological variation in foveolar receptor diameters in a particular eye). At spatial frequencies beyond 75 cycles/degree (cpd) (20/8), the retinal image will be undersampled, causing the visual percept to be distorted (aliased), thereby limiting the ability of the nervous system to interpret a high-quality retinal image (Figure 1). In an optically aberrated eye capable of neural-limited acuity (20/8 to 20/10), improving the optics cannot improve acuity but will improve contrast for larger pupils.

Receptor-sampling limits does not mean that the visual system is incapable of seeing targets having finer

Figure 3. (Applegate) A: If a letter *E* is imaged so it falls within the borders of a single photoreceptor, the letter cannot be differentiated from a period. B: To be seen as a letter, *E* must be sampled by enough photoreceptors to differentiate the letter's component parts. Reprinted with permission from Applegate.¹

detail. The visual system can see targets having finer detail; however, they will not appear in their true form. Because of receptor undersampling, the appearance of the image is distorted, forming what is commonly called an alias percept of the actual object (Figure 4); ie, the percept of the object takes on an appearance that can be quite different from the object. Consequently, photoreceptor sampling fundamentally limits acuity for larger pupils in the diffraction-limited eye.

Modulation Transfer Function and Aberrations

Counter to the diffraction-limited case, in an optically aberrated system, large pupils can be a disadvantage. First, the modulation transfer ratio can go negative, indicating a phase shift (light bars are now dark and dark bars are light). Second, in the diffraction-limited case (Figure 1), the modulation transfer ratio first goes to zero at the cutoff frequency and remains at zero. In the aberrated case, the function can cross zero several times before reaching the cutoff frequency. The extent of these effects is dependent on the magnitude and type of residual optical aberration. To illustrate this point, Figure 5 shows MTFs for 4 pupil diameters in the presence (thinner lines) and absence (thicker lines) of a 0.50 diopter (D) defocus.

These effects are better illustrated by looking at the change in the modulation transfer ratio and the first zero crossover of the MTF caused by moderate aberrations (Figure 6). Both parts of this figure show that with moderate defocus, increasing pupil diameter decreases optical performance. To maximize the benefit of increasing the pupil diameter (decreased f number), the system has to be essentially aberration free.

In Figures 5 and 6, counter to the diffraction-limited case, the larger the pupil, the larger the loss in image contrast and first zero crossover; the defocus induced losses in image contrast and first zero crossover for the 2 mm pupil are less than those for larger pupils, showing why pinhole testing is effective clinically.

Phase and Aberrations

As important as the effects of optical aberrations on retinal image contrast and the resulting visual perception are shifts in spatial phase induced by optical aberrations. It is well known, for example, that defocus can introduce phase reversals into images so that dark bars become light and light bars become dark, an effect some-

Figure 4. (Applegate) A simulated retinal view of an 80 cpd grating (notice the edges of the pattern shows the grating) being sampled by a primate foveolar retinal receptor mosaic. Undersampling creates an alias percept of the grating that appears as zebra stripes. Photograph courtesy of David Williams. Reprinted with permission from Applegate.¹

Figure 5. (Applegate) Diffraction-limited (no blur) MTFs (thick color-coded lines) and 0.50 D of defocus MTFs (color-coded thin lines) for 4 different pupil sizes. Calculations of the diffraction-limited MTFs were made using 555 nm light and methods detailed by Smith equation 11.36, and calculations of the defocus MTFs were made using a geometric approximation detailed by Smith equation 11.40.⁴⁶ Negative values reflect a phase shift and are discussed in the next section. (Note: A small error in the geometric approximation can be appreciated most clearly by the crossover of the 2 mm defocus MTF and the diffraction 2 mm MTF at low spatial frequencies. In larger pupils in which the wavefront error is greater than 1 wavelength, the error in the geometric approximation of the defocus case is, for all practical purposes, inconsequential.)

times called spurious resolution (Figure 7). Similar effects occur for aberrations besides defocus, which can lead to spatial-frequency-dependent losses in con-

Figure 6. (Applegate) A: The change in the MTF (0.50 D blur–no blur) as a function of spatial frequency up to the first zero crossover. Notice that the loss in MTF increases as pupil size increases. (The small gain in the MTF for the 2 mm pupil between 0 and approximately 7 cpd is an error in the calculation due to ignoring the effects of diffraction in the calculation of the blur MTF.) In larger pupils, this error is essentially zero. B: The MTF first zero crossover frequency as a function of pupil size for 3 levels of defocus.

Figure 7. (Applegate) Schematic of a defocused MTF showing phase reversal. On the right are focused and blurred images of the same pattern showing phase reversal for higher frequencies (ie, the higher-frequency white part of the pattern just off center aligns with the dark part of the pattern in the periphery). Figure courtesy of Bradley and coauthors.⁴⁷

trast^{13,14} and distorted letter appearance. A potential benefit of correcting ocular aberrations is that errors in spatial phase are corrected. This benefit is likely to be substantial for many aspects of visual function.

For example, the visual task of letter discrimination is strongly affected when optical defocus introduces phase reversals into some spatial frequency components of the target but not others. The upper row of images in Figure 8 demonstrates the devastating effect of phase reversals on letter legibility caused by optical blur. However, when these phase reversals are corrected, as shown in the bottom row of images, the decrease in contrast induced by blur has relatively little impact. This simple

demonstration illustrates that the phase shifts and phase reversals caused by optical aberrations can be more important than the loss of contrast.

Chromatic Aberration, Spherical Aberration, and the Neural Transfer Function

Thus far, the discussion has been limited to monochromatic light, simple defocus, and phase and has ignored the effects of chromatic aberration and the relationship between the neural threshold function and the optical MTF. In Figure 9, these interactions are demonstrated for a 6 mm pupil using the Indiana model eye.¹⁵ Introducing chromatic defocus (green line) into the model reduces the MTF significantly from the diffraction-limited case (black line). The MTF is affected more when a spherical aberration is introduced, and performance is even worse when chromatic and spherical aberrations are combined in the model.

The vertical separation between these curves shows that correcting the eye's chromatic aberration should result in about a 5-fold improvement in image contrast at 30 cpd. The improvement in image contrast would be approximately 12-fold if the eye's spherical aberration were corrected and approximately 25-fold if both chromatic and spherical aberrations were corrected.

Anticipated improvements in spatial resolution are not as dramatic as the improvements in contrast. The spatial frequency limit of detection without aliasing for the visual system is indicated graphically by the intersection of the optical MTF with the neural threshold function. This intersection in the Indiana eye occurs at

Figure 8. (Applegate) A blurred letter *E* with phase reversals at higher frequencies (top) and without phase reversal (bottom). Notice how phase reversal degrades the percept of the letter to the point that for the smaller letters (higher frequencies), the *E* is no longer legible. Figure courtesy of Bradley and coauthors.⁴⁸

Figure 9. (Applegate) Modulation transfer functions for a diffraction-limited case (black line), the chromatic aberration (green line), typical spherical aberration (blue line), both spherical and chromatic aberration (red line), and the foveal neural threshold function (purple line). The foveal neural thresholds are measured by using gratings formed on the retina through interference, bypassing the optics of the eye. The points at which each MTF crosses the foveal neural threshold function define the limiting spatial frequency the model can “see” (dashed arrows).

approximately 30 cpd (20/20) when both aberrations are included in the model. The intersection increases to 60 cpd (20/10), a 2-fold increase, when both aberrations are removed. This analysis indicates that the benefits of aberration correction are proportionally greater when measured in the contrast domain than when measured in the spatial domain.

Measures of Optical Quality

The quality of an optical system can be specified in 3 different but related ways. The first is a description of the loss of contrast experienced when an image of a sinusoidal grating object is cast. As discussed above, the MTF and PTF together define the OTF of an imaging system.

The second way is to describe the detailed shape of the image for a simple geometrical object such as a point of light or a line. The distribution of light in the image plane is called a point-spread function (PSF) for a point object or a line-spread function (LSF) for a line object. Simple measurements derived from these functions, such as the width (blur circle diameter) or height (Strehl ratio) of the intensity distribution, are taken as figures of merit that capture the blurring effects of optical imperfections.

The third way of specifying optical quality is by the underlying optical aberrations rather than the secondary effect of these aberrations on image quality. This description can be couched in terms of the deviation of light rays from perfect reference rays (ray aberrations) or the deviation of optical wavefronts from the ideal reference wavefront (wavefront aberrations). This aberration method is a more fundamental approach to the description of optical imperfections in eyes from which all the secondary measures of optical quality (PSF, LSF, MTF, PTF, and OTF) can be derived. It is also the approach that is most useful for customized corneal ablation and

contact lenses since the aberration function of an eye is a prescription for optical perfection.

Definition and Interpretation of Aberration Maps

From a clinical perspective, perhaps the most useful interpretation of optical aberration maps is by errors of the optical path length (OPL). The OPL concept specifies the number of times a light wave must oscillate in traveling from one point to another. Since the propagation velocity of light is slower in the watery refractive media of the eye than in air, more oscillations will occur in the eye than in the same physical distance in air. By defining OPL as the product of the physical path length with the refractive index, OPL becomes a measure of the number of oscillations executed by a propagating ray of light. This is an important concept because light rays emitted by a point source will propagate in many directions, but if all the rays have the same OPL, every ray represents the same number of oscillations. Consequently, light at the end of each ray will have the same temporal phase and this locus of points with a common phase represents a wavefront of light. A propagating wavefront of light is therefore defined by the locus of points in space lying at the same OPL from a common point source of light. To define the aberrations of an optical system, the OPL for a ray passing through any point (x,y) in the plane of the exit pupil is compared with the chief ray passing through the pupil center $(0,0)$. The result is called the optical path difference (OPD). Thus, the aberration structure of the eye's optical system is summarized by a 2-dimensional map showing how OPD varies across the eye's pupil.

In a perfect imaging system, the OPL is the same for all light rays traveling from the object point to the image point and therefore $OPD = 0$ for all (x,y) locations in the pupil. In the case of an eye, this means that rays of light from a single point object that pass through different points in the pupil will arrive at the retinal image point having oscillated the same number of times. Such rays will have the same temporal phase and therefore will add constructively to produce a perfect image. If, on the other hand, light passing through different points in the pupil arrive with different phases because they traveled along paths of different OPLs, the system is aberrated and the quality of the image will suffer. By conceiving of optical aberrations as differences in optical path length, it is easy to see how aberrations might arise (1) due to

thickness anomalies of the tear film, cornea, lens, anterior chamber, or posterior chamber; (2) because of refractive index anomalies of the ocular media that might accompany inflammation, disease, and aging; or (3) because of decentering or tilting of the various optical components of the eye with respect to each other.

A concrete example of the OPL concept is illustrated in Figure 10. In a myopic eye with no other aberrations, the optical path is shorter for rays passing near the pupil margin than for those passing through the pupil center. Consequently, the best retinal image of a point object will be formed if this variation in optical path length is compensated by placing the point source at the eye's far point. Now the wavefront of light enters the eye as a concave wavefront such that the central rays arrive at the eye before the marginal rays, giving them a head start so when they follow a longer optical path through the eye they arrive in phase with the marginal rays. In short, to obtain the optimum retinal image, the optical distance from each object point to its image must be the same for every path through the pupil. The wavefront aberration map indicates the extent to which this ideal condition is violated.

By reversing the direction of light propagation in Figure 10, a more practical definition of an aberration map for the eye is obtained. It follows from the preceding discussion that if the retinal point P' is a source of light reflected out of the eye, the shape of the emerging wavefront is determined by the variation of OPL across the eye's pupil. If the eye is optically perfect and em-

Figure 10. (Applegate) Example of a diverging wavefront from source P being focused to retinal point P' by a myopic eye. Reversing the direction of light propagation, light reflected from retinal point P' emerges from the eye as a wavefront converging on point P . When referenced to the x - y plane of the pupil, the wavefront shape $W(x,y)$ is also an aberration map of the eye.

metropic, this reflected wavefront would be a plane wave propagating in the positive z -direction. Thus, for distance vision, any departure of the emerging wavefront from the x - y plane is an optical aberration. For near vision, the reflected wavefront emerging from the eye must be compared with a spherical wavefront centered on the fixation point. In practice, the distance $W(x,y)$ between the reflected wavefront emerging from the eye and the corresponding reference sphere is taken as a measure of the wavefront aberration function of the eye for the given viewing distance. By convention, positive aberrations occur when the marginal ray travels a shorter OPL than does the central (chief) ray, as in the case of the myopic eye shown in Figure 10. Therefore, by this sign convention, $W(x,y) = -\text{OPD}(x,y)$.

In summary, the shape of a wavefront of light reflected out of the eye from a point source on the retina is determined by the OPD for rays passing through each point in the eye's pupil. Therefore, a map of OPD across the pupil plane is equivalent to a mathematical description, $W(x,y)$, of the shape of the aberrated wavefront that emerges from the eye. Either may be used as an aberration map of the eye. Such maps are fundamental characterizations of the optical quality of the eye that can be used to compute other common metrics of image quality (eg, the PSF or OTF) from the expected retinal image of any visual target can be computed.

Methods of Measuring Aberration Maps of Eyes

The current interest in optical aberrations of eyes has been spawned by new technology resting on an ancient principle and the promise of super vision. Nearly 400 years ago, the celebrated Jesuit philosopher and astronomer, Christopher Scheiner, professor at the University of Ingolstadt and a contemporary of Kepler and Galileo, published his treatise *Optical Foundations of the Eye*,¹⁶ 75 years before Huygens' wave theory of light. This pioneering book described a simple device that is widely known in ophthalmology as Scheiner's disk (Figure 11). Scheiner reasoned that if an optically imperfect eye views through an opaque disk containing 2 pinholes, a single distant point of light such as a star will form 2 retinal images. If the eye's imperfection is a simple case of defocus, the double retinal images can be brought into register by viewing through a spectacle lens of the appro-

Figure 11. (Applegate) Scheiner's disk isolates rays, allowing their aberrated direction of propagation to be traced. An ametropic eye will form 2 retinal images for each object point when viewing through a Scheiner disk with 2 apertures.

prate power. This design idea for an optometer for measuring refractive errors of eyes was first proposed by Porterfield in 1747 and improved by Young in 1845.

A simple lens will not always bring the 2 retinal images into coincidence, however, so a more general method of quantifying the refractive imperfection of the eye at each pupil location is needed. Smirnov¹⁷ was the first to extend Scheiner's method by using a fixed light source for the central reference pinhole and a moveable light source for the outer pinhole as illustrated in Figure 12. By adjusting the moveable source horizontally and vertically, the isolated ray of light is redirected until it intersects the fixed ray at the retina and the patient now reports seeing a single point of light. Having made this adjustment, the displacement distances Δx and Δy are measures of the ray aberration of the eye at the given

Figure 12. (Applegate) Smirnov's aberrometer used the principle of Scheiner's disk to measure the eye's optical imperfections separately at every location in the eye's entrance pupil.

pupil point. By independently determining Δx and Δy for many different rays entering the eye's pupil with respect to a foveal fixated reference ray entering the center of the pupil, one can deduce the local slope of the aberrated wavefront as a function of pupil entry for a foveal image. Mathematical integration of the slope data yields the shape of the aberrated wavefront, which can be displayed as an aberration map. Such an approach has recently been implemented in the subjective spatially resolved refractometer^{18–21} and the objective ray-tracing refractometer.^{22–24}

Tscherning demonstrated that deviations of rays from their ideal position can also be subjectively quantified using a grid superimposed on a blurring lens and viewing a point source (Figure 13).²⁵ Sixty years later, Howland invented the crossed-cylinder aberroscope,²⁶ (Figure 14) which was first used subjectively²⁷ and later objectively²⁸ to quantify the aberrations of the eye. Recently, the principles of the Tscherning aberrometer have been incorporated in an objective aberrometer^{29,30} and used to design ablative corrections³¹ to reduce the higher-order aberrations of the normal and highly aberrated eye.¹⁰

Figure 13. (Applegate) The Tscherning aberroscope is shown at the left. The aberroscope consists of a +5 D lens with a 1 mm square superimposed grid. To use it, the subject views a distant point source of light that is focused in front of his or her retina by the aberroscope lens. The grid is then shadowed on the subject's retina, and the distortions of the grid are drawn by the subject. A grid corresponding to a positive spherical aberration is depicted. Figure courtesy of Howland.⁴⁸

Figure 14. (Applegate) The optics of the crossed-cylinder aberroscope. The construction and use are similar to those of the Tscherning aberroscope. However, instead of a +5 D spherical lens, a ± 5 D crossed-cylinder lens is used. This causes the grid to be shadowed in the interval of Sturm. In this arrangement, diffraction smears the images of the lines along their lengths, making the image sharper than that of the Tscherning aberroscope. Figure courtesy of Howland.^{27,48}

Independently of these developments in ophthalmology, Scheiner's simple idea was reinvented by Hartmann for measuring the ray aberrations of mirrors and lenses.³² Hartmann's method was to perforate an opaque screen with numerous holes, as shown in Figure 15. Each hole acts as an aperture to isolate a narrow bundle of light rays so they can be traced to determine any errors in their direction of propagation. Since rays are perpendicular to the propagating wavefront, any error in ray direction is also an error in wavefront slope. Thus, Hartmann's method is commonly referred to as a wavefront sensor.

Seventy years later, Shack and Platt³³ invented a better Hartmann screen using an array of tiny lenses that focuses the light into an array of small spots, 1 spot for each lenslet. Their technique came to be known as Shack's modified Hartmann screen or Shack-Hartmann for short. To see how the array of spot images can be used to determine the shape of the wavefront, look at the wavefront in cross-section as shown in Figure 16.

In a perfect eye, the reflected plane wave will be focused into a perfect lattice of point images, each image falling on the optical axis of the corresponding lenslet. In contrast, the aberrated eye reflects a distorted wavefront as illustrated in Figure 17. The local slope of the wavefront is now different for each lenslet, and therefore the wavefront will be focused into a disordered collection of spot images. By measuring the displacement of each spot from its corresponding lenslet axis, we can deduce the

Figure 15. (Applegate) The Hartmann screen used to measure aberrations objectively is a Scheiner disk with numerous apertures.

slope of the aberrated wavefront as it entered the corresponding lenslet. Mathematical integration of slope yields the shape of the aberrated wavefront, which can then be displayed as an aberration map.

The Shack-Hartmann wavefront sensor was first used in 1994 by Liang and coauthors³⁴ to measure aberrations in human eyes, completing this historically meandering path to the development of a fast, objective, reliable method of assessing the aberration structure of human eyes. Liang's concept of a Scheiner-Hartmann-Shack aberrometer is shown schematically in Figure 18. Because the shape of an aberrated wavefront changes as the light propagates, it is important to analyze the reflected wavefront as soon as it passes through the eye's pupil. To do this, a pair of relay lenses focuses the lenslet array onto the entrance pupil of the eye. Optically, then, the lenslet array appears to reside in the plane of the eye's

Figure 16. (Applegate) The Shack-Hartmann wavefront sensor forms a regular lattice of image points for a perfect plane wave of light.

entrance pupil where it can subdivide the reflected wavefront immediately as it emerges from the eye. The array of spot images formed by the lenslet array is captured by a video sensor and then analyzed by computer to estimate the eye's aberration map.

The Shack-Hartmann approach has been reduced to an objective clinical aberrometer.³⁵ The wave aberration measured by the aberrometer is in turn used to design a customized corneal ablative pattern designed to reduce not only the spherical and cylindrical (astigmatic) aberrations (lower-order aberrations) but also the higher-order aberrations of the eye.³⁵ Using the LADARVision® system, McDonald recently reported the first eye in which the higher-order aberrations of a normal ametropic eyes were reduced by wavefront-guided customized ablation (M. McDonald, MD, "Treatments Based on Wavefront Guided Custom

Figure 17. (Applegate) The Shack-Hartmann wavefront sensor forms an irregular lattice of image points for an aberrated wavefront of light.

Figure 18. (Applegate) The modern aberrometer built on the Scheiner-Hartmann-Shack principle uses relay lenses to image the lenslet array into the eye's pupil plane. A video sensor (CCD) captures an image of the array of spots for computer analysis.

Figure 19. (Applegate) Examples of higher-order aberration maps for 4 normal, healthy individuals reconstructed from measurements taken with an Scheiner-Hartmann-Shack aberrometer similar to that shown in Figure 18. Light areas in the map indicate the reflected wavefront is phase-advanced; dark areas indicate phase-retardance. The maximum difference between high and low points on each map is about $1\ \mu\text{m}$. (Zernike orders 0–2 omitted for clarity.)

Cornea Corrections,” presented at the Refractive Surgery Interest Group Subspecialty Day, American Academy of Ophthalmology, Dallas, Texas, USA, October 2000).

Normal and Clinical Examples

Four examples of aberration maps for normal healthy eyes are shown in Figure 19. By using a gray scale to encode wavefront height, the human visual system’s natural ability to infer depth and structure from shading is used. The maximum difference between the highest and lowest points on each of these maps is about $1\ \mu\text{m}$, which is a little more than 1 wavelength of the light used to measure the eye’s aberrations ($0.633\ \mu\text{m}$). Perhaps the most distinctive feature of these maps is the irregular shape of their smoothly varying shapes. Another important feature of aberration maps of normal eyes is the tendency to be relatively flat in the center of the pupil, with aberrations growing stronger near the pupil margin. This is consistent with the literature showing that image quality is relatively good for medium-sized pupils but deteriorates as pupil diameter increases.^{36,37}

Figure 20. (Applegate) Examples of higher-order aberration maps from eyes with 4 different clinical conditions. A: Dry eye. B: Keratocornus. C: Laser in situ keratomileusis surgery. D: Cataract. The maximum difference between high and low points on each map is about $10\ \mu\text{m}$, except D which is closer to $1\ \mu\text{m}$. (Zernike orders 0–2 omitted for clarity.)

For comparison, Figure 20 shows 4 examples of clinically abnormal eyes. Qualitatively, these maps have the same irregular, smoothly varying shapes as in normal eyes. The main difference is that the magnitude of the aberrations is about 10-fold larger, and therefore image quality is about 10-fold worse than normal. Another important abnormality of the keratoconic patient¹⁹ and to a lesser extent the dry-eye patient³⁸ is the tendency to have large aberrations in the middle of the pupil. The implication of this result is that image quality will be subnormal for small pupils as well as for large pupils.

Limitations

Classical analysis of data from a Shack-Hartmann wavefront sensor does not consider the quality of individual spots formed by the lenslet array. Only the displacement of spots is needed for computing local slope of the wavefront over each lenslet aperture. However, experience has shown that the quality of dot images can vary dramatically over the pupil of a human eye as illustrated in Figure 21. The presence of blurred spots indicates a violation of the underlying assumption that the

Figure 21. (Applegate) Examples of selective loss of image quality in individual spots for eyes with 2 different clinical conditions.

wavefront is locally flat over the face of the lenslet. Two possible reasons are illustrated in Figure 22.

The first possibility is that the gross aberrations of the eye are so large that the wavefront is significantly curved over the area of the lenslet. The result is a blurry spot that is difficult to localize. If the aberrations are large enough, neighboring spots can overlap, which complicates the analysis. The second possibility involves irregular aberrations on a very fine spatial scale. Perturbations of the wavefront within the lenslet aperture are too fine to be resolved by the wavefront sensor using classical methods. Rather than displacing the spots laterally, these “microaberrations” scatter light and blur the spots formed by the aberrometer. Although these blurry spots are problematic, they contain useful information about the degree and location of scattering sources inside the eye that may prove useful in clinical applications.^{38,39}

Taxonomy of Optical Aberrations

One systematic method for classifying the shapes of aberrations maps is to conceive of each map as the weighted sum of fundamental shapes or basis functions. One popular set of basis functions are the Zernike polynomials. This set of mathematical functions are formed as the product of 2 other functions, one of which depends only on the radius r of a point in the pupil plane and the other depends only on the meridian θ of a point in the pupil plane. The former is a simple polynomial of

Figure 22. (Applegate) Blurring of individual spots detected by a Shack-Hartmann wavefront sensor may indicate the presence of gross aberrations of large magnitude or microaberrations. In either case, the blur is due to violation of the underlying assumption that the wavefront is locally flat over the lenslet aperture.

the n th degree, and the latter is a harmonic of a sinusoid or cosinusoid. A pictorial dictionary of the first 28 Zernike polynomials is arranged in the form of a pyramid of basis functions in Figure 23, which displays the relationship between the Zernike terms and the classical concepts of aberration theory such as spherical aberration and coma. Every aberration map can be represented uniquely by a weighted sum of these functions. The process of determining the weighting coefficient required to describe a given aberration map is a least-squares curve-fitting process called Zernike decomposition, which results in a vector of Zernike coefficients. Mathematical details may be found in several standard reference works.^{40,41}

Figure 23. (Applegate) Pictorial dictionary of Zernike modes used to systematically represent the aberration structure of the eye.

The Optical Society of America recently sponsored a task force of visual optics researchers to develop standards for reporting optical aberrations of eyes. Recommendations of this task force have been published.⁴²

Relationships Between Measures of Optical Quality

One of the most important results of optical theory in the 20th century was the linking of the PSF, LSF, MTF, PTF, and OTF by means of a mathematical operation known as the Fourier transform.^{43,44} The point-spread function is computed as the squared magnitude of the Fourier transform of a complex-valued pupil function built from the aberration map. Since the PSF represents the intensity distribution of light in the image of a point source, it should be a highly localized, bright spot. Diffraction sets a lower limit to the diameter of the spot and an upper limit to the intensity in the center of the spot. A common metric of image quality called Strehl ratio is computed as the ratio of the actual intensity in the center of the spot to the maximum intensity of a diffraction-limited spot. As pupil diameter increases, the intensity of a diffraction-limited spot increases faster than the intensity of an aberrated spot, which tends to reduce Strehl ratio. It is not uncommon for the PSF of human eyes to have multiple peaks, which complicates the simple notion of Strehl ratio. More important, it signals the formation of 2 or more point images for a single point object. This condition of diplopia or polyopia has great clinical significance because of its implications for visual performance and the quality of visual experience.^{13,14}

Conclusion

Modern aberroscopy allows detailed characterization of the monochromatic aberrations of the eye and, in turn, the design of the ideal optical correction. Furthermore, and perhaps equally important, accurate characterization of the monochromatic aberrations of the eye allows computation of the expected retinal image for any visual object (Figure 24). Such an image computation overcomes the longstanding handicap imposed on clinicians and visual scientists by the natural inaccessibility of the retinal image.⁴⁵ Thus, optical theory, when combined with knowledge of the anatomy

Figure 24. (Applegate) Point-spread functions (left panels) and simulated retinal images of an eye chart (right panels) for the eye in Figure 20,C, analyzed for 2 pupil diameters (2 mm and 6 mm).

and physiology of the eye's neural system, enables us to define the limits to improvement of vision and to predict the visual consequences of optical interventions and/or nonintervention.⁴⁹

References

1. Applegate RA. Limits to vision: can we do better than nature? *J Refract Surg* 2000; 16:S547–S551
2. Williams DR, Liang J, Miller DT. Adaptive optics for the human eye. *Vision Science and Its Applications* 1996. OSA Technical Digest Series. Washington, DC, Optical Society of America, 1996; 13:145–147
3. Liang J, Williams DR, Miller DT. Supernormal vision and high-resolution retinal imaging through adaptive optics. *J Opt Soc Am A* 1997; 14:2884–2892
4. Bille JF. Preoperative simulation of outcomes using adaptive optics. *J Refract Surg* 2000; 16:S608–S610
5. Applegate RA, Howland HC, Sharp RP, et al. Corneal aberrations and visual performance after radial keratotomy. *J Refract Surg* 1998; 14:397–407
6. Martinez CE, Applegate RA, Klyce SD, et al. Effect of pupillary dilation on corneal optical aberrations after photorefractive keratectomy. *Arch Ophthalmol* 1998; 116:1053–1062

7. Oliver KM, Hemenger RP, Corbett MC, et al. Corneal optical aberrations induced by photorefractive keratectomy. *J Refract Surg* 1997; 13:246–254
8. Oshika T, Klyce SD, Applegate RA, et al. Comparison of corneal wavefront aberrations after photorefractive keratectomy and laser in situ keratomileusis. *Am J Ophthalmol* 1999; 127:1–7
9. Seiler T, Kaemmerer M, Mierdel P, Krinke H-E. Ocular optical aberrations after photorefractive keratectomy for myopia and myopic astigmatism. *Arch Ophthalmol* 2000; 118:17–21
10. Mrochen MC, Bueeler M, Seiler T. Wavefront-guided treatments for the management of decentered ablations. ARVO abstract 3246. *Invest Ophthalmol Vis Sci* 2001; 42:S604
11. Liang J, Williams DR. Effect of higher order aberrations on image quality in the human eye. *Vision Science and Its Applications*. OSA Technical Digest Series. Washington DC, Optical Society of America, 1995; 1:70–73
12. Yoon GY, Williams DR. Visual benefits of correcting the higher order monochromatic aberrations and the longitudinal chromatic aberration in the eye. *Trends Optics Photonics* 2000; 35:205–211
13. Woods RL, Bradley A, Atchison DA. Monocular diplopia caused by ocular aberrations and hyperopic defocus. *Vision Res* 1996; 36:3597–3606
14. Woods RL, Bradley A, Atchison DA. Consequences of monocular diplopia for the contrast sensitivity function. *Vision Res* 1996; 36:3587–3596
15. Thibos LN, Ye M, Zhang X, Bradley A. Spherical aberration of the reduced schematic eye with elliptical refracting surface. *Optom Vis Sci* 1997; 74:548–556
16. Scheiner C. *Oculus hoc est fundamentum opticum*. . . Innsbruck, Agricola, 1619
17. Smirnov MS. [Measurement of the wave aberration of the human eye]. [Russian] *Biofizika* 1961; 6:687–703
18. Webb RH, Penney CM, Thompson KP. Measurement of ocular wavefront distortion with a spatially resolved refractometer. *Appl Optics* 1992; 31:3678–3686
19. Penney CM, Webb RH, Tiemann JT, Thompson KP. Spatially resolved objective autorefractometer. US Patent No 5.258,791, 1993
20. He JC, Marcos S, Webb RH, Burns SA. Measurement of the wave-front aberration of the eye by a fast psychophysical procedure. *J Opt Soc Am A* 1998; 15:2449–2456
21. Burns SA, Marcos S. Measurement of image quality with the spatially resolved refractometer. In: MacRae SM, Krueger RR, Applegate RA, eds, *Customized Corneal Ablation; the Quest for Super Vision*. Thorofare, NJ, Slack Inc, 2001; 201–209
22. Molebny VV, Pallikaris IG, Naoumidis LP, et al. Retinal ray-tracing technique for eye-refraction mapping. *Proc SPIE* 1997; 2971:175–183
23. Pallikaris IG, Panagopoulou SI, Molebny VV. Clinical experience with the Tracey technology wavefront device. *J Refract Surg* 2000; 16:S588–S591
24. Molebny VV, Pallikaris IG, Panagopoulou SI, et al. Tracey retinal ray tracing technology. In: MacRae SM, Krueger RR, Applegate RA, eds, *Customized Corneal Ablation; the Quest for Super Vision*. Thorofare, NJ, Slack Inc, 2001; 193–196
25. Tscherning M. Die monochromatischen aberrationen des menschlichen auges. *Z Psychol Physiol Sinnesorgan* 1893; 6:456–471
26. Howland B. Use of crossed cylinder lens in photographic lens evaluation. *Appl Optics* 1960; 7:1587–1599
27. Howland HC, Howland B. A subjective method for the measurement of monochromatic aberrations of the eye. *J Opt Soc Am* 1977; 67:1508–1518
28. Walsh G, Charman WN, Howland HC. Objective technique for the determination of monochromatic aberrations of the human eye. *J Opt Soc Am A* 1984; 1:987–992
29. Mrochen M, Kaemmerer M, Mierdel P, et al. Principles of Tscherning aberrometry. *J Refract Surg* 2000; 16: S570–S571
30. Mrochen M, Kaemmerer M, Mierdel P, Seiler T. Wavefront-guided LASIK using a Tscherning wavefront analyzer. In: MacRae SM, Krueger RR, Applegate RA, eds, *Customized Corneal Ablation; the Quest for Super Vision*. Thorofare, NJ, Slack Inc, 2001; 185–191
31. Mrochen M, Kaemmerer M, Seiler T. Wavefront-guided laser in situ keratomileusis: early results in three eyes. *J Refract Surg* 2000; 16:116–121
32. Hartmann J. Bemerkungen uber den Bau und die Justirung von Spektrographen. *Z Instrumentenk* 1900; 20:47
33. Shack RV, Platt BC. Production and use of a lenticular Hartmann screen (abstract). *J Opt Soc Am* 1971; 61:656
34. Liang J, Grimm B, Goetz S, Bille JF. Objective measurement of wave aberrations of the human eye with the use of a Hartmann-Shack wave-front sensor. *J Opt Soc Am A* 1994; 11:1949–1957
35. Pettit GH, Campin JA. CustomCornea using the LADARVision system. In: MacRae SM, Krueger RR, Applegate RA, eds, *Customized Corneal Ablation; the Quest for Super Vision*. Thorofare, NJ, Slack Inc, 2001; 165–172
36. Campbell FW, Gubisch RW. Optical quality of the human eye. *J Physiol* 1966; 186:558–578
37. Liang J, Williams DR. Aberrations and retinal image quality of the normal human eye. *J Opt Soc Am A* 1997; 14:2873–2883
38. Thibos LN, Hong X. Clinical applications of the Shack-Hartmann aberrometer. *Optom Vis Sci* 1999; 76:817–825
39. Applegate RA, Thibos LN. Localized measurement of scatter due to cataract. Abstract 13. *Invest Ophthalmol Vis Sci* 2000; 41(4):S3
40. Born M, Wolf E. *Principles of Optics; an Electromag-*

- netic Theory of Propagation, Interference and Diffraction of Light, 7th ed. Cambridge, Cambridge University Press, 1999
41. Malacara D, ed. Optical Shop Testing, 2nd ed. New York, NY, John Wiley & Sons, Inc, 1992
 42. Thibos LN, Applegate RA, Schwiegerling JT, Webb R. Standards for reporting the optical aberrations of eyes. *Trends Optics Photonics* 2000; 35:232–244
 43. Gaskill JD. Linear Systems, Fourier Transforms, and Optics. New York, NY, John Wiley & Sons, Inc, 1978
 44. Goodman JW. Introduction to Fourier Optics. San Francisco, CA, McGraw-Hill, 1968
 45. Sarver EJ, Applegate RA. Modeling and predicting visual outcomes with VOL-3D. *J Refract Surg* 2000; 16:S11–S616
 46. Smith WJ. Modern Optical Engineering: The Design of Optical Systems, 2nd ed. New York, NY, McGraw-Hill, Inc, 1990; 327
 47. Bradley A, Hong X, Chung STL, Thibos LN. The impact of defocus-induced phase reversals on letter recognition is different for hyperopes and myopes. Abstract 187. *Invest Ophthalmol Vis Sci* 1999; 40(4):S35
 48. Howland HC. Ophthalmic wavefront sensing; history and methods. In: MacRae SM, Krueger RR, Applegate RA, eds, Customized Corneal Ablation; the Quest for Super Vision. Thorofare, NJ, Slack Inc, 2001; 35–38
 49. Doshi JB, Sarver EJ, Applegate RA. Schematic eye models for simulation of patient visual performance. In press, *J Refract Surg*