The field of aberrations of the human eye is moving rapidly, being driven by the desire to monitor and optimise vision following refractive surgery. It is important for ophthalmologists and optometrists to have an understanding of the magnitude of various aberrations and how these are likely to be affected by refractive surgery and other corrections. In this paper, I consider methods used to measure aberrations, the magnitude of aberrations in general populations and how these are affected by various factors (for example, age, refractive error, accommodation and refractive surgery) and how aberrations and their correction affect spatial visual performance.

The paper included a short section on asphericity, which is helpful for relating aberrations of corneas to their shapes. Using this as a framework, in this review I consider methods used to measure aberrations, the magnitudes of aberration in the general population and how these are affected by factors such as age, refractive error, accommodation and refractive surgery, how aberrations affect spatial visual performance, how feasible it is to correct these aberrations and how much correcting aberrations could improve vision beyond that currently enjoyed by most people (the so-called ‘super-vision’). A reasonable understanding of the contents of the companion paper, although not necessarily of the equations, is important to make sense of most of this review.

The aberration system used in this paper...
and the companion paper is that of the American National Standard for Ophthalmics—Methods for reporting optical aberrations of eyes ANSI Z80.28-2004. Two recent books by MacRae, Krueger and Applegate and MacRae have extensive coverage of wave aberrations, particularly related to the aim of reducing aberrations following refractive surgery.

**MEASUREMENT OF ABERRATIONS**

Aberrations of the eye have been measured since at least the time of Thomas Young. In this review, I will concentrate on current methods that are being applied commercially. For a brief summary of some older methods that have been used for research applications, see Atchison and Smith. One technique of particular note is the Howland crossed-cylinder aberroscope, the development of which from a subjective to an objective technique can be traced through a series of papers. Two recent books by MacRae, Krueger and Applegate and MacRae have extensive coverage of wave aberrations.

Aberration measuring instruments bear a similar relationship to autorefractors as corneal topographers do to keratometers. As well as providing the same information, refraction in the case of aberration instruments and anterior corneal power in the case of the corneal topographers, they provide the additional information of higher-order aberrations in the case of aberration instruments and departure of the anterior corneal surface from a toric shape in the case of corneal topographers. In my opinion, the success of aberration measuring instruments depends on them being able to supplant autorefractors by providing a reliable, accurate refraction and on reduction in cost.

**Types of measurement techniques**

For some instruments, an image is formed on the retina and this is re-imaged out of the eye for analysis. This can be referred to as ‘into-the-eye’ aberrometry but MacRae, Applegate and Krueger refer to it as ‘retinal imaging’ aberrometry. Other instruments project a narrow beam into the eye and trace the rays from the retina out of the eye. This can be referred to as ‘out-of-the-eye aberrometry’ or ‘outgoing optics aberrometry’. In other instruments, the inclinations of beams into and out of the eye vary and this is referred to as ‘adjustable aberrometry’. All but one of the techniques to be described is objective (subjects do not make any judgments). Transverse aberrations are measured in all but the retinoscopic technique that measures longitudinal aberrations. In both cases, the aberrations are derived for various points in the pupil and typically using a least squares technique are converted into wave aberration functions.

Other terms used to describe aberrometers are ‘simultaneous’ when measurements are taken simultaneously at a number of pupil positions and ‘sequential’ when measurements are taken at one pupil position at a time.

Table 1 shows my version of the advantages and disadvantages of the different instrument types. Some of the disadvantages may be of more or less concern with some instruments than with others of a particular type and improvements can be expected to reduce some of the disadvantages.

**ABERROMETER**

<table>
<thead>
<tr>
<th>Approximate measuring time</th>
<th>Dynamic range</th>
<th>Sampling</th>
</tr>
</thead>
<tbody>
<tr>
<td>Laser ray tracing</td>
<td>seconds</td>
<td>high</td>
</tr>
<tr>
<td>Tscherning aberrometer</td>
<td>milliseconds</td>
<td>moderate</td>
</tr>
<tr>
<td>Hartmann-Shack sensor</td>
<td>milliseconds</td>
<td>moderate</td>
</tr>
<tr>
<td>Retinoscopy</td>
<td>second</td>
<td>high</td>
</tr>
<tr>
<td>Subjective ray tracing</td>
<td>minutes</td>
<td>high</td>
</tr>
</tbody>
</table>

* varies with different instruments
** limited radially but not angularly

**ABERROMETRY—TSCHERNING**

A narrow laser beam is directed into the eye through a range of pupil positions in sequence (Figure 1). A mirror or combination of two mirrors must be repositioned between stimulations to move the laser beam. For each pupil position, the retinal image is re-imaged back through the whole pupil onto a CCD array (although the effective size of the pupil on the outgoing pass can be reduced by an external stop). Because of this second pass, the images will be further degraded but should be similar in appearance except for their locations. The centroids of the images are compared with that of the reference image, corresponding to the pupil centre, to give transverse aberrations. Currently Marcos’s group can measure up to 37 pupil positions for a 6.5 mm pupil in four seconds. Considerable computing is required and eye movements may affect results. The technique can measure a wide range of refractive errors and aberrations (large dynamic range). Unlike the ‘simultaneous’ techniques, it does not suffer from the possibility that centroids corresponding to different pupil locations can be confused.

**ABERROMETRY—LASER RAY TRACING**

A narrow laser beam is directed into the eye through a range of pupil positions in sequence (Figure 1). A mirror or combination of two mirrors must be repositioned between stimulations to move the laser beam. For each pupil position, the retinal image is re-imaged back through the whole pupil onto a CCD array (although the effective size of the pupil on the outgoing pass can be reduced by an external stop). Because of this second pass, the images will be further degraded but should be similar in appearance except for their locations. The centroids of the images are compared with that of the reference image, corresponding to the pupil centre, to give transverse aberrations. Currently Marcos’s group can measure up to 37 pupil positions for a 6.5 mm pupil in four seconds. Considerable computing is required and eye movements may affect results. The technique can measure a wide range of refractive errors and aberrations (large dynamic range). Unlike the ‘simultaneous’ techniques, it does not suffer from the possibility that centroids corresponding to different pupil locations can be confused.
Measurement of ocular aberrations

Atchison

several thin beams at the same time can reduce the time taken for the laser ray tracing technique (Figure 2). However, on re-imaging, it would not be possible to distinguish between the different intersections, let alone to which pupil positions they corresponded. In the Tscherning aberrometer,16,17 a defocusing lens is placed in front of the eye to spread the image across the retina. A grid mask in front of the eye allows radiation to pass through only certain pupil positions. As for the laser ray tracing technique, the retinal image is re-imaged through the whole pupil (although as for the laser ray tracing technique the effective pupil size can be limited by an external stop, Figure 2). Provided aberrations are not very high, it will be clear which image point coincides with which pupil position. To help reduce the problem for high aberrations, the defocusing lens power can be increased. Transverse aberrations are determined by comparing the image positions with those of a reference schematic eye, which will induce some error because real eyes will have different paraxial image sizes from the schematic eye.

OUT-OF-THE-EYE ABERROMETRY—HARTMANN-SHACK SENSOR

Most commercial aberrometers are of this type. A narrow beam (for example, 1 mm wide) from a point radiation source is imaged by the eye and the light passed back from the fundus travels through a Hartmann-Shack wave-front sensor18,19 consisting of an array of micro-lenses (typically of the order of 0.5 mm diameter) and onto the array of a CCD camera (Figure 3). The Hartmann-Shack sensor is conjugate with the pupil and its focal plane is at the camera array. For a perfect eye, the wavefront at the sensor would be a plane wave. Each micro-lens isolates a small bundle of radiation passing through a small region of the pupil. The transverse ray aberration (slope of the wavefront) associated with each micro-lens can be determined from the departure of the centroid of its corresponding image from the ideal image position.

An image can be taken in a matter of milliseconds and fluctuations in aberrations over the order of seconds can be followed. The method can be used as part of an adaptive optics system to correct aberrations of the eye20 and may be adapted to measure scatter within the eye.21 Another advantage of this technique is that it is robust to eye position, as the software algorithm can be used to determine pupil centre accurately, particularly if there is high sampling in the pupil (>> 100 positions). Spot overlapping due to high aberrations or defocus is a major concern, that is, the instrument may have a limited dynamic range. Auxiliary optics is needed on the outgoing path to correct most of

Figure 1. Representation of laser ray tracing aberrometry, showing two ingoing beams and two outgoing beams corresponding to central and peripheral pupil positions. The ingoing beams are narrow but the outgoing beams pass through the whole pupil. Although the outgoing pass degrades the retinal images O’ and A’, the external images O” and A” are of similar appearance and are at similar positions to each other as are the retinal images to each other (allowing for magnification of the external optical system).

Figure 2. Representation of Tscherning aberrometry, showing ingoing and outgoing beams. For the ingoing pass, the ray traces corresponding to the limits and middle of the beam are shown. For the outgoing beam, rays are shown from the centre of the retinal image only. The external stop limits the diameter of this beam. Also shown is the appearance of the retinal image in a commercial instrument for a perfect eye.
the defocus of the eye. Even so, there still may be problems, which vary in magnitude according to the focal lengths and element size of the Hartmann-Shack sensor and the software algorithms.

SEQUENTIAL RETINOSCOPY

A narrow infrared slit beam enters the eye, scanning both radially and angularly across the retina (Figure 4). The former is achieved by using a chopper drum and the latter by rotating the beam and drum.22-24 The retina reflects some of the radiation. An aperture in the optical system is conjugate with the cornea and rotates to be aligned with the ‘sweeping’ direction of the ingoing beam. The time at which peak intensity of radiation strikes each photodetector compared with that for each reference detector is determined and converted into a longitudinal aberration. In the commercial device, measurement time is less than 0.5 second, measurements are taken for up to 1,440 pupil positions (more than for other instrument type) and it has an extremely high dynamic range claimed to be up to ±20 D. Measurements are taken every degree but only four measurements can be taken in the radial direction and interpolation is required between the pupil positions corresponding to the detectors (and in the approximately 2.0 mm diameter centre of the pupil). Measurements are restricted to 6 mm diameter. The instrument has similarities with Placido disc type keratographers in that measurement at any pupil position is in a radial direction only and this must affect the accuracy of the aberrations determined.

SEQUENTIAL SUBJECTIVE RAY TRACING OR ‘SPATIAL RESOLVED REFRACTOMETRY’

Many variants of this technique25-27 have been used for research purposes over the past 40 years.28-31 In its application in a commercial instrument, two narrow beams of visible light enter the eye at one time, a reference beam passing through the centre of the pupil and a test beam (Figure 5). The test beam passes through a variety of pupil positions in sequence. If there is no aberration, a test beam entering parallel to the reference beam will intercept it on the retina and the subject will see them overlapped. If there is aberration, the subject will see them separated and the task for the patient is to alter the angle of the test beam until the retinal images coincide. This angle is a measure of the transverse aberration. Transverse aberrations at many

Figure 3. Representation of Hartman-Shack aberrometry. The left side shows ingoing and outgoing beams, the middle part shows the transverse aberrations and the top left shows an image for a particular eye. The Hartmann-Shack sensor is conjugate with the pupil.

Figure 4. Principle of sequential retinoscopy. An incoming beam passes through a chopper wheel so that it scans across the retina. In this myopic eye, the aperture of the system is conjugate with a point in front of the retina. The outgoing beam passes through to the detectors, which are conjugate with the cornea. This myopic eye, as the retina is scanned from top to bottom, the radiation reaching the detectors passes across the cornea from bottom to top (vice versa in hypermetropia). The time displacement and the order in which the detectors are stimulated determine the aberrations across the meridian. This figure is based on Figure 17.1 of Buscemi.24

The aperture vignettes the light passing through the optical system so that at any moment light from only part of the pupil passes through to a set of photodetectors, four of which are on either side of the axis containing two reference detectors. The array is conjugate with the cornea and rotates to be aligned with the ‘sweeping’ direction of the ingoing beam. The time at which peak intensity of radiation strikes each photodetector compared with that for each reference detector is determined and converted into a longitudinal aberration. In the commercial device, measurement time is less than 0.5 second, measurements are taken for up to 1,440 pupil positions (more than for other instrument type) and it has an extremely high dynamic range claimed to be up to ±20 D. Measurements are taken every degree but only four measurements can be taken in the radial direction and interpolation is required between the pupil positions corresponding to the detectors (and in the approximately 2.0 mm diameter centre of the pupil). Measurements are restricted to 6 mm diameter. The instrument has similarities with Placido disc type keratographers in that measurement at any pupil position is in a radial direction only and this must affect the accuracy of the aberrations determined.
pupil positions can be converted into a wave aberration function.

This method is by far the most time-consuming of those described, taking a few minutes for approximately 36 pupil positions. Because of the lengthy time, pupil position needs to be monitored and corrections made as required. The method is similar to the outgoing Hartmann-Shack method, as the aberrations occur in object rather than image space.

Reference axis for measuring aberrations

Aberrations should be referenced to the pupil centre. At least one major instrument that measures both corneal topography and wave aberrations uses the keratometric axis as the reference axis. This axis passes from the fixation point to the centre of curvature of the anterior corneal surface. On this basis, it can be expected to give results different from those of most other instruments, particularly when the intersections of the line-of-sight and the keratometric axis with the cornea (corneal sighting centre and vertex normal, respectively) are considerably displaced from each other. If the positions of the corneal sighting centre and vertex normal are known, a correction can be made to reference the aberrations to the pupil centre.

Wavelength correction

All the techniques described here, apart from the subjective ray tracing, use infrared radiation in the approximate range 780 to 900 nm. There are several advantages in using near infrared instead of visible radiation: it is comfortable because the eye is not sensitive to infrared radiation, the source cannot influence accommodation, pupil dilation is usually not required because pupillary responses are not sensitive to infrared and fundus reflectance is higher than in visible radiation. The aberrations and in particular the second-order defocus must relate to visible wavelengths to be relevant to vision. Infrared radiations penetrate deeper into the fundus than do visible wavelengths, which would give a small myopic shift in refraction but there is a much stronger hypermetropic shift because of the longitudinal chromatic aberration of the eye.

Using Hartmann-Shack and laser ray tracing methods, Llorente and colleagues found that aberrations changed little from the infrared (787 nm) to the visible (543 nm) radiation, except for the defocus term. For most of their 36 eyes, the defocus difference was predicted well by the Indiana chromatic eye model. The mean difference was 0.78 ± 0.29 D. Other studies found small changes only in aberrations other than defocus.

Comparison of techniques/instruments

Salmon, Thibos and Bradley compared the Hartmann-Shack technique with a psychophysical ray tracing technique and obtained similar results on two subjects. Moreno-Barriuso and Navarro compared Hartmann-Shack and laser ray tracing techniques with a model eye and real eyes. The results were similar for the two techniques, although the aberrations for one eye were too large to be measured by their Hartmann-Shack technique. Moreno-Barriuso and colleagues extended this work to compare Hartmann-Shack, laser ray tracing and psychophysical ray tracing techniques in two subjects. Again despite the different principles, the techniques gave very similar results.

Except for the psychophysical ray tracing of Moreno-Barriuso and colleagues, these studies were done with laboratory instruments. There have been few studies of the aberration results with commercial aberration measuring instruments but several can be expected in the next few years. The accuracy (validity) of instruments is difficult to compare because a model eye, even if its aberrations have been measured by alternative means, cannot always be aligned accurately in front of an instrument. It is much easier to compare the repeatability of different instruments and ophthalmologic literature abounds with examples of this. Aberration instruments can be compared with each other or with automated refractors with respect to refraction. Durrie and Stahl compared six instruments, four using the Hartmann-Shack principle, one using the laser ray tracing principle and one using the retinoscopic principle. The instruments with the higher sampling densities performed slightly better on repeatability (three measurements) with the more highly aberrated eyes than did those with lower sampling densities.

MAGNITUDES OF ABERRATIONS

It is often of interest to determine the corneal and internal/lenticular contributions to the total ocular aberrations. These will be discussed in the following sub-sections.

Population studies

There have been recent large-scale studies of aberrations in normal populations. When comparing results from these studies, there are some confounding factors, as there were different ranges of refractive error and age. Porter and colleagues, Castejón-Mochón and colleagues and Wang and Koch used natural pupils whereas Thibos, Bradley and Hong, Brunette and colleagues and Wang and colleagues used cycloplegia, different numbers of orders were used and different pupil sizes were analysed. Despite
Measurement of ocular aberrations

2. The magnitude of aberrations increases can be drawn: these inconsistencies, some conclusions can be drawn:

1. Monochromatic aberrations show considerable variability within populations (Figure 6). Mean higher-order RMS for 6 mm pupils is approximately 0.3 μm.45-47

2. The magnitude of aberrations increases with increase in pupil size.45,46 As this happens, the relative contribution of the orders increases as the order number increases, for example, Castejón-Mochón and colleagues43 found that the third and fourth orders contributed 6.4 per cent and 2.6 per cent, respectively, of the total RMS at 5 mm pupil diameter in their subjects.41,52 or age was a confounding parameter.44 Llorente and colleagues53 found that spherical aberration changed with spherical aberration and vertical coma with vertical tilt (Table 1 of Atchison).1

3. The vast majority of aberrations are in the second orders, even for well-corrected eyes40,43 and the magnitudes of aberrations decrease as the aberration order becomes higher (Figure 6).

4. There is considerable between-eye symmetry, with many higher-order terms being highly correlated between the two eyes.40,43,45

5. Several aberration terms within eyes are significantly correlated,41,42 which is to be expected as many Zernike aberration polynomials have common monomial terms, for example, defocus with spherical aberration and vertical coma (Figure 6). Thibos and colleagues42 found that when expected left-right eye mirror symmetries were taken into account,1 7/38 other co-efficients in the fourth to seventh orders were also significantly different from zero, although these were very small compared with \( c_4 \). The mean magnitude of \( c_4 \) is approximately 0.10 ± 0.10 μm for 6 mm pupils.41,45,46

Refractive error

A number of studies has found that myopia is accompanied by moderate or no increase in higher-order aberrations.47,49,51 Spherical aberration should increase with increase in myopia if the only variation between eyes with different refractive corrections is axial length.47 Although Carkeet and colleagues43 found children with myopia less than 3 D had significantly less spherical aberration than other refractive error groups, Cheng and colleagues47 did not find that spherical aberration changed significantly with the degree of myopia in a young adult group.

Accommodation

Accommodation is the ability of the eye to change its power to bring objects of interest at different distances into focus.6 Accommodation is not always accurate and deteriorates at both low stimulus levels, where the response can be in lead of the stimulus by 1 D or more, and at high stimulus levels, where the response tends to lag the stimulus. Accuracy also decreases when the accommodative stimulus becomes degraded, such as occurs at low luminances and with low contrast detail or when the retinal image is little affected by defocus as occurs with small pupils. In such circumstances, the accommodative...
response reverts to an intermediate tonic accommodation. Ciuffreda\textsuperscript{46} provided a good review of these aspects of accommodation.

With accommodation, ray paths through the eye will change, from the object to the retina or in the reverse direction and it is not unexpected that the aberrations should change. There is considerable older literature on the possibility of accommodative astigmatism,\textsuperscript{60,73} but the instrumentation was too crude to adequately measure any claimed changes. More recent studies using autorefractors have found results ranging from negligible\textsuperscript{74} to small changes of the order of 0.1 D\textsuperscript{75} or 0.5 D\textsuperscript{76} and changes well beyond 1 D.\textsuperscript{77,79} The studies finding the greater changes may be viewed with some suspicion because of the possibility of alignment away from the line-of-sight. In the recent large-scale study by Cheng and colleagues\textsuperscript{48} using a Hartmann-Shack wavefront sensor, all astigmatic changes were less than 0.2 D.

It has long been known that most eyes suffer from positive spherical aberration when unaccommodated, with a trend to negative spherical aberration on accommodation.\textsuperscript{30,29,46,60,89} Cheng and colleagues\textsuperscript{48} have recently conducted the most comprehensive study in this area. They found that although many Zernike aberration coefficients changed with accommodation, the spherical aberration coefficient $\zeta_0$ showed the greatest changes. It was the only one that changed systematically with accommodation; the change being nearly always negative and proportional to the change in the accommodative response (mean -0.043 µm per dioptre of accommodation for 5 mm pupils) (Figure 7). It became zero at about 1.7 D accommodation, which was slightly less than earlier estimates.\textsuperscript{10,85} The population mean total RMS (excluding defocus) was relatively constant for accommodation up to 3.0 D, beyond which it increased.

As well as its lack of accuracy, in ‘steady-state’ vision accommodation exhibits short-term fluctuations (or short-term temporal instabilities) that are related at least partly to heart and respiration effects.\textsuperscript{90,91} The magnitude of these fluctuations is typically less than 0.25 D and is restricted mainly to frequencies less than 2 Hz. As well as fluctuation errors in accommodation, that is, the defocus, there are much smaller fluctuations in the higher-order aberrations of the eye.\textsuperscript{92,93} The shapes of the power spectra for many of the higher-order aberrations are considerably different from those of the defocus. Dynamic adaptive optical systems can correct the fluctuations in the eye’s aberrations in real-time and can accentuate the improvement in retinal image quality provided by a ‘static’ adaptive optical system.\textsuperscript{96,98}

### Ageing

Spatial visual performance decreases with increase in age.\textsuperscript{99,103} This has both optical and retinal-neural causes. As an example of vision loss, Elliot\textsuperscript{100} found a reduction in contrast sensitivity of 0.4 log units at 16 cycles/degree between corrected young (mean 22 years) and corrected older (mean 72 years) age groups, of which half could be attributed to each of retinal-neural and optical factors. At lower frequencies, the differences were smaller and retinal-neural factors were the more important.

Age-related changes take place in all ocular tissues.\textsuperscript{6} Because of its accessibility, the anterior shape of the cornea has been extensively studied. The anterior surface of the cornea becomes more curved with age but more so in the horizontal meridian.\textsuperscript{104} The mean anterior cornea is slightly prolate at about $Q = -0.20 \pm 0.20$.\textsuperscript{105,107} The anterior cornea becomes less prolate (Q less negative) or changes little with increase in age.\textsuperscript{107,109} Less is known of the shape of the posterior cornea surface. Mean Q estimates are -0.42,\textsuperscript{110} -0.66,\textsuperscript{111} and -0.38 ± 0.27,\textsuperscript{107} with Dubbelman and colleagues\textsuperscript{107} finding that the surface becomes more prolate (Q more negative) with increase in age. This trend (-0.007/year) will have negligible effect on aberrations because of the small difference between corneal and aqueous refractive indices.\textsuperscript{112}

The most dramatic age-related changes in the eye take place within the lens, which continues to grow throughout life because new cells are formed without older cells being discarded. Its shape, size and mass change markedly. The central thickness increases throughout life because of increases in cortical thickness, with corresponding decreases in anterior chamber thickness\textsuperscript{36,115,116} but there is little or no change in its equatorial diameter.\textsuperscript{59} The surfaces of the unaccommodated lens become more curved with increase in age, particularly the anterior surface.\textsuperscript{54,117}

Some estimates of lenticular surface asphericity have been made using \emph{in vivo} Scheimpflug photography.\textsuperscript{54,56,117} Dubbelman and van der Heijde\textsuperscript{107} found mean anterior and posterior surface Q values of -5 and -4, claiming the back surface becomes less prolate (but non-significantly so) with increase in age.

The refractive index of the lens decreases away from its centre. Most recent studies have used ray tracing through the equator of the \emph{in vitro} lens\textsuperscript{118} or destructive Fresnel reflectance measurements of the \emph{in vitro} lens.\textsuperscript{119} To model the gradient index as an equivalent refractive index, the latter must be higher than the refractive index in the lens centre. Despite the surfaces of the lens in the unaccommodated eye becoming more curved with age, the eye does not become more myopic (the
lens paradox) and various models have been proposed for age-related changes in the gradient index distribution to account for this. An in vitro study using magnetic resonance imaging (MRI) indicated that the refractive index in the nucleus reduces at a rate of 0.00034/year, which is similar to model calculations of changes in equivalent refractive index. However, a more recent MRI study indicates that there is little if any change in nuclear refractive index with age, and the refractive index distribution changes less quickly near the centre of the lens with increase in age. This can also be modelled as a decrease in equivalent refractive index.

The eye loses its ability to accommodate with increase in age until this capability is lost in the mid-50s. Our current understanding is that this is due to changes in the lens and its capsule. Nothing is known about how accommodation affects the age-related refractive index distribution — this may be of considerable importance in determining the rate at which accommodative amplitude is lost.

The transmission of the eye decreases with increase in age, mainly attributable to changes in the lens and senile miosis. The transmission of light to the retina decreases by up to 60 per cent between the ages of 20 and 60 years. Losses in transmission can be compensated in many situations by an increase in luminance. Light loss is more marked for shorter wavelengths. There is increase in both forward and backward scatter, particularly after the age of 40 years.

Refractive errors are relatively stable between the ages of 20 and 40 years, after which there is a shift in the hypermetropic direction by an average of about 1.5 D over the next 20 years. After the age of 70 years, there is a shift of the mean refractive error in the myopic direction associated with the development of nuclear cataract.

With all these changes to the optics of the eye, it is not unexpected that the aberrations will also change with age. A number of studies has found that total higher-order aberrations increase with age throughout adulthood, although Brunette and colleagues, who investigated a wide range of ages of eight to 80 years, found that the total and component higher-order aberrations including spherical aberration actually declined until about the fourth decade, after which they increased. Artal and colleagues found a three-times increase in RMS aberrations over the age range 20 to 70 years (5.9 mm pupils) (Figure 8a). Amano and colleagues found the RMS of third-order coma (combined $c_3$ and $c_5$ co-efficients) increased as age increased. Several studies have produced a range of results. Oshika and colleagues found significant increases in third-order aberrations with increase in age, while Guirao, Redondo and Artal reported significant increases in higher-order RMS, third-order coma and fourth-order spherical aberration, with spherical aberration becoming more positive with increase in age (note that due to the sign convention adopted by these authors at that time, spherical aberration is negative in the paper). Wang and colleagues claimed that higher order RMS and coma RMS (combined third- and fifth-order co-efficients) but not spherical aberration RMS (combined fourth- and sixth-order co-efficients) increased significantly with age. Amano and colleagues found increased third-order coma with increase in age but like Wang and colleagues found no change in spherical aberration with increase in age. Fujikado and colleagues reported no changes in third-order RMS aberrations nor in fourth-order RMS aberrations with age (4 mm pupils).

Smith and colleagues found a decreasing negative spherical aberration of the lens with increase in age and Artal and colleagues found increases in lens aberrations, including a positive shift in spherical aberration with age.

For young subjects, the anterior corneal aberrations are usually higher than those of the whole eye, which means that the internal aberrations compensate to some extent for those of the cornea in young eyes. This balance is lost with increase in age, so that the aberrations of the total eye are greater than those of the anterior cornea (Figure 8a). He and colleagues argued that the compensation in young eyes may hold only for those eyes with low aberrations, as they found that the anterior corneal and internal aberrations were often additive for many of their young subjects with higher levels of aberrations.
Despite findings in several studies that aberrations increase with age, it should be remembered that some compensation is provided by pupil sizes decreasing with age. As well as reducing aberrations, smaller pupils increase depth-of-focus, thus compensating to some extent for small refractive errors and loss of accommodation.

Marcos \(^{155}\) considered the consequence on the changes in ocular aberrations with age for refractive surgery. Conventional LASIK increases aberrations and in particular spherical aberration in the positive direction for myopes. Age-related changes in the lens will increase these further. If an ideal corneal ablation is applied (one minimising aberrations), this correction will not last because of the lens changes. Furthermore, the subsequent implantation of conventional IOLs following the development of cataract will undo any aberration minimisation achieved because its aberrations will be different from those of the lens it replaced. This problem may be alleviated in the future by custom IOLs.

**Disease**

A few studies have looked at the magnitudes of higher-order aberration in corneal disease. The non-inflammatory corneal disease keratoconus causes increases in all orders, particularly the third order. \(^{156-158}\) for example, Maeda and colleagues \(^{158}\) found a 7.3 times increase in anterior corneas become more prolate and even oblate (steepening away from centre) to much less prolate and becomes negative after surgery (Figure 9a). Anterior corneal spherical aberration, which moves in the positive direction, \(^{163-166}\) Most of the changes can be accounted for by changes in anterior corneal aberrations. The anterior cornea changes from prolate (flattening away from centre) to much less prolate and even oblate (steepening away from centre) to much less prolate and becomes negative after surgery (Figure 9a), with the change increasing with the magnitude of correction. This is because anterior corneas become more prolate following treatment. \(^{173,175-177}\) As an example of the magnitude of changes, Marcos and colleagues \(^{165}\) found that mean anterior corneal and total RMS higher-order aberrations increased following LASIK by 3.7 and 1.9 times, respectively, and the mean spherical aberration co-efficient \(c_4\) increased by 0.63 µm (pre-surgical mean spherical equivalent -6.8 ± 2.9 D, 6.5 mm pupil).

Aberrations increase following hypermetropic corneal refractive surgery \(^{175,174}\) (Figure 9a). Anterior corneal spherical aberration is positive before surgery and becomes negative after surgery (Figure 9b), with the change increasing with the magnitude of correction. This is because anterior corneas become more prolate following treatment. \(^{173,175-177}\) As an example of the magnitude of the changes, Llorente and colleagues \(^{175}\) found that mean anterior corneal and total RMS higher-order aberrations increased following LASIK by 1.8 and 2.2 times, respectively and the mean spherical aberration co-efficient \(c_4\) decreased by 0.70 ± 0.30 µm (pre-surgical mean spherical equivalent +3.2 ± 1.1 D, 6.5 mm pupil).

Two reasons for the large increases in aberrations following refractive surgery are decentration in the ablation pattern, which particularly affects third-order aberrations, \(^{178,179}\) and aberrations being measured at pupil sizes exceeding the ablation zone. \(^{180}\) The internal aberration contribution changes following myopic and hyperopic LASIK, \(^{160,175}\) indicating some changes.

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\(^{a}\) Kuroda and colleagues did not use normalisation terms so their \(c_4\) co-efficients have been reduced by \(\sqrt{5}\) to be consistent with ANSI Z80.28-2004.
Measurement of ocular aberrations

As discussed above, corneal refractive surgery usually increases aberrations of the eye considerably. A few years ago, there was considerable optimism about the possibility of customising surgery to eliminate aberrations, thus resulting in ‘super vision’. This has been tempered more recently by a greater awareness of the problems in achieving this. The companies manufacturing the instrumentation are developing new algorithms, based on wave-aberration measurements, to improve surgical outcomes. A number of reports27,178,188-204 is appearing, which describe the success of wavefront ‘guided’ surgery. The results of these studies are generally encouraging. Other worthwhile aims are that aberrations following surgery will not be greater than those before surgery and that excessively high aberrations can be reduced to near normal levels. It must be kept in mind that it remains more important to correct the second-order aberrations (the defocus and astigmatism) than the higher-order aberrations.

Intraocular lenses
Following IOL implantation, most eyes have considerable aberration because both the cornea and the spherical IOL have positive spherical aberration. Although the shape (bending) of spherical IOLs can be optimised,205,206 they will always have positive spherical aberration. Thus, aberrations for eyes with intraocular lenses can be expected to be greater than those for young eyes, for which the usually negative internal spherical aberration provides some balance to the positive spherical aberration of the cornea. Aberrations of eyes with intraocular lenses were similar to the aberrations before surgery and also similar to those reported for eyes of similar age in other studies.207 This can be attributed partly to the sign of internal spherical aberration changing from negative to positive with increase in age (Figure 8b).

Clear lensectomy (intraocular lenses placed in eyes that had not developed cataract) produced smaller aberrations in hypermetropic patients than did LASIK in another group of hypermetropic patients of similar age.274 Other studies208,209 have compared the higher-order ocular aberrations with different intraocular lenses.

An aspheric IOL with negative spherical aberration is needed to compensate for the positive spherical aberration of the cornea.210,211 Recently an aspheric front surface IOL was introduced to provide reduced spherical aberration to the majority of patients212 and seems to provide improved visual performance relative to spherical IOLs.213,214 Customised IOLs may become possible by using a photopolymerisable material that enables the shape of lenses to be adjusted after fitting in the eye to correct higher-order as well as the second-order aberrations.215

**Contact lenses**

Despite their low power compared with that of the cornea, contact lenses can dramatically affect ocular aberrations, because the back and front surfaces have very high absolute powers (often over 60 D each). Surface asphericities, particularly if on the front, have considerable importance. Additionally with rigid contact lenses, the back surface of the tear film neutralises approximately 90 per cent of the departure of the anterior corneal surface from a spherical shape. Some earlier studies216,217 have considered the influence of lens surface asphericity on visual performance. In the past two decades, there has been a number of contact lenses with rotationally symmetrical aspheric surfaces with claims that they eliminate most of the ocular aberrations.

Given the wide range in ocular aberrations and the fact that the majority of aberrations are in the third order, these claims are exaggerated.218

Recent studies219-223 using wavefront aberration measuring instruments confirm the important influence of contact lenses on ocular aberrations.

**Orthokeratology**

This technique reshapes the cornea using contact lenses with flat-fitting back surfaces to reduce myopia. It has been practised for 40 years but has received recent impetus with corneal topography to monitor progress, new oxygen-permeable materials and improved technology to give precise lathing of surfaces. Together these make treatment possible through overnight wearing of contact lenses.226 The procedure reduces the peripheral flattening of cornea and may even cause it to become oblate. Joslin and colleagues227 reported increases in higher-order aberrations in one orthokeratology trial, similar to those found in refractive surgery. Mean total RMS increased from 0.31 µm to 0.73 µm, with mean spherical aberration RMS (combined $c_4$ and $z_4$ coefficients) increasing from 0.08 ± 0.16 µm to 0.39 ± 0.15 µm and mean horizontal coma co-efficient $c_4$ increasing from +0.05 ± 0.08 µm to +0.35 ± 0.14 µm (9 subjects, 6 mm pupils, mean myopic reduction 3.1 ± 0.9 D).

**Spectacle lenses**

Spectacles are distinct from contact lenses and intraocular lenses because they are not attached to the eye and do not rotate with it. The higher-order aberrations of
lenses are not important by themselves as the low overall power and surface powers of typical meniscus-shaped lenses means that they have little effect on the ocular aberrations. The main aberrations of concern are those associated with the eye rotating to look at objects that are not aligned along the optical axis of the lenses—off-axis defocus and astigmatism and transverse chromatic aberration. Spectacle lens bending and asphericity are designed to minimise the first two of these aberrations. It is likely that progressive addition lenses might introduce higher-order aberrations because of rapid changes in surface curvature, particularly along the progressive corridor. Villegas and Artal measured small amounts of coma, trefoil and astigmatism along the intermediate corridor of a commercial progressive addition lens with a +2 D addition. Astigmatism was high outside the corridor, as is well known, but the higher-order aberrations were similar to those at the lens centre (RMS 0.03 to 0.06 µm for order aberrations were similar to those at one year). They considered a variety of options, such as correcting those aberrations for which longer-term variability was not significantly greater than short-term variability and correcting only those aberrations, the co-efficients of which would be of a particular sign for 90 per cent or more of the time. Interestingly, they found that the aberration correction based on the average aberrations over a day, regardless of the above considerations, provided the best optical quality across a period of a few months.

**Off-axis aberrations**

While the aberrations that we have examined are those associated with foveal (on-axis) vision, we should be interested also in the aberrations associated with off-axis (peripheral) vision, because the quality of the off-axis optics is important for detection thresholds, detection of movement and limiting the ability to observe the peripheral fundus.

**OFF-AXIS REFRACTION**

Over the past 70 years, several techniques, including retinoscopy, manual objective refractometers, subjective refraction, line spread and point spread functions, photorefraction and the Hartmann-Shack sensor, have been used to measure off-axis refractive errors. In most of these studies, measurements were in the horizontal visual field. The components of refraction in the horizontal and vertical meridians typically shift in the negative or positive directions, respectively, with increase in horizontal visual field angle.

Figure 10 shows the refractive components, $M$, $J_{45}$, across the horizontal visual field for groups of emmetropic and myopic subjects with small amounts...
of on-axis astigmatism. Measurements go to only ±35° because of field restrictions with the instrument, but other studies have measured out to 60°.247,254,256,258 Off-axis refractions are often high in the periphery. They vary considerably between subjects, are asymmetric with the nasal visual field usually having higher levels of $J_{180}$. 90-180° astigmatism than the temporal visual field and mean values of $J_{45}$ 45-135° astigmatism are small.

For the emmetropic group in Figure 10, there is a myopic (negative) shift of mean $M$ into the periphery. $J_{180}$ has a mean of -1.5 D at 35° in the nasal visual field (approximately half the value of the conventional cylinder). Seidemann and colleagues269 found that as well as a nasal-temporal asymmetry, the inferior visual field is usually considerably more myopic than the superior visual field.

Shapes of refractive profiles change as a function of on-axis refractive error. On increase in visual field angle, there are shifts of mean sphere $M$ in the hypermetropic (positive) and myopic (negative) directions with increases in central myopia and central hypermetropia, respectively.254,259,270,271,272 Shapes of refractive profiles change as a function of on-axis refractive error. On increase in visual field angle, there are shifts of mean sphere $M$ in the hypermetropic (positive) and myopic (negative) directions with increases in central myopia and central hypermetropia, respectively.254,259,270,271,272 Pre-surgical refraction was -5.25 D. I thank the Journal of Cataract and Refractive Surgery for permission to show these data.

As already mentioned, conventional LASIK changes the asphericity as well as the curvature of the anterior cornea, such that hypermetropic LASIK causes the cornea to become more prolate (flattening away from the centre), while myopic LASIK causes the cornea to become less prolate or even oblate (steepening away from the centre). It can be predicted, and Ma and colleagues271 have shown, that in corrected hypermetropia and corrected myopia this results in trends towards increased hypermetropia and increased myopia as off-axis angle increases (Figure 11). This is in the opposite direction to the changes in refraction with increase in angle for the uncorrected situations.

**HIGHER-ORDER ABERRATIONS**

Only three studies have measured the higher-order off-axis aberrations. Navarro, Moreno and Dorrorsoro266 adapted their laser ray-tracing technique to investigate the nasal visual field in four subjects, Guirao and Artal267 used the point spread function in the temporal visual field of four subjects, and Atchison and Scott270 modified the Hartman-Shack sensor technique to measure in both directions along the horizontal visual field for five subjects. In the last study, allowance was made for unequal sampling in the horizontal and vertical directions. Where comparable, the results in the three studies were similar. In the Atchison and Scott270 study, results were standardised for 6 mm pupils. There were considerable variations between subjects in the patterns of Zernike aberrations but as for $J_{180}$ astigmatism, these were generally greater in the nasal than in the temporal visual field. Figure 12 shows the contributions of the second to fourth orders to the root-mean squared aberrations for one subject and Figure 13 shows the group higher-order RMS aberrations. These figures show that the second-order aberrations dominate in the peripheral visual field, as they usually do in the...
The RMS third-order aberrations increased up to four times from fixation to 40° in the visual field but the fourth- to sixth-order contributions changed little across the visual field (Figure 13).

The picture that emerges from the three studies is that, while the higher-order aberrations increase into the periphery, the rate of increase is not dramatic and the second-order aberrations remain the most important. This supports earlier studies\(^262,263,265,266\) that found that off-axis image quality is not much poorer than on-axis quality once off-axis refractive errors are corrected.

For the horizontal meridian and five subjects, Atchison\(^276\) found that both component and overall Zernike aberrations are greater for the nasal than for the temporal visual field. Usually anterior corneal aberrational components are much higher than the overall aberrations across the visual field and are balanced to a considerable degree by the internal aberrational component.

**Distortion**

Distortion in peripheral vision refers to how far the retinal image is displaced from its (expected) paraxial position. Ames and Proctor\(^277\) referred to work of Donders\(^278\) with a subject with exophthalmia (protruding eyeballs) and of Druault,\(^279\) who examined extracted eyes and two eyes of people with exophthalmia. Both studies gave the peripheral angle external to the eye and the position of the retinal image position relative to the corneal margin. Ames and Proctor converted these retinal positions to internal angles without giving any details. Using the Gullstrand number 1 schematic eye, Atchison and Smith\(^6\) used the results to estimate peripheral distortion within the eye, finding the retinal image is closer to the optical axis than that for a distortion-free eye.

**INFLUENCE OF HIGHER-ORDER ABERRATIONS ON VISION**

Two important issues regarding higher-order aberrations are the effect that they have on vision and whether these effects are the same or different for aberrations with the same co-efficients. One way of investigating these issues is to present displays on a monitor as if they have been viewed through aberrated optics and then have people with well-corrected optics view the ‘aberrated’ displays and perform visual performance tasks such as visual acuity determination.

Applegate and colleagues\(^280,281\) used this approach. Letter charts were ‘aberrated’ by convolving the unaberrated chart details with the point spread function that would result from particular aberrations at a 6 mm pupil size. Well-corrected subjects viewed the charts through small (3 mm) pupils. This approach presumes that the aberrations of the subjects at this pupil size will have negligible additional effects on the appearance of the displays. Different Zernike polynomials with the same co-efficient, and hence contribution to the RMS, affected visual acuity differently. Aberrations near the vertical centre line of the Zernike pyramid such as \(Z_4^2\) affected visual acuity more than aberrations well away from the centre line\(^291,292\) (Figure 14).

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**Figure 13.** Group RMS aberrations in third to sixth aberration orders for five subjects. Pupil size 6 mm. Error bars indicate standard deviations. Data of Atchison and Scott.\(^270\) I thank the Optical Society of America for permission to show these data.

**Figure 14.** Simulated retinal image of a 6/6 letter E aberrated with 0.25 µm RMS wave aberration for different Zernike aberration polynomials, a 6 mm pupil diameter and wavelength 500 nm. The polynomials are arranged as a pyramid for second to fourth orders. The polynomials near the centre line of the pyramid affect image quality more than polynomials away from the midline. Based on Figure 20 of Applegate.\(^282\)
work to combining Zernike polynomial functions while maintaining a fixed RMS (0.25 μm for 6 mm pupils). Polynomials that are two radial orders apart and the same sign and angular frequency tend to combine to decrease visual acuity.

These studies do not fully quantify the effects that aberrations have on vision because of the ‘uncoupled nature’ of the optics from display to the subjects’ retinas. They indicate that different aberrations have different effects on vision and furthermore that RMS wavefront error is not a good predictor of visual acuity. A second way to investigate the influence of aberrations on vision is to perform visual tasks by viewing through an optical system containing adaptive optics that can effectively correct or manipulate the system containing adaptive optics from display to the subjects’ retinas.

THE POSSIBILITY OF CORRECTING OCULAR ABERRATIONS AND CONSEQUENCES FOR VISION

Methods for reshaping the cornea by excimer laser ablation, using computer-controlled, flying-spot scanning, and for making contact lenses or intra-ocular lenses of any desired surface form have made rapid progress. Correction of both the second-order and higher-order monochromatic aberrations of the eye is becoming possible. The aberrations of the eye have to be known and ideally the contribution of the element that is to change should be known (for example pre-surgical cornea versus post-surgical cornea or ocular lens versus intraocular lens).

There are many problems associated with correcting ocular aberrations. In the case of corneal refractive surgery, ‘the cornea is not a piece of plastic’. There are variable healing rates and incomplete understanding about the biomechanical forces that determine the final corneal shape. Lipshitz presented an extensive list of mainly surgical and post-surgical problems with correcting the optics in corneal refractive surgery. Problems with contact lenses and intra-ocular lenses include decentrations, tilts and rotations (dynamically in the case of contact lenses). Guirao, Williams and Cox have made estimations of the misalignment that can occur in custom contact lenses before they cease to provide any benefit.

A consideration of the likely improvements in vision by correcting the aberrations of the eye requires an understanding of the optical and the neural (or retinal-brain) limits to vision. Here I will use one of the useful measures of optical quality of the eye, the optical transfer function (OTF). This has two components. One is the modulation transfer function (MTF) that gives the ratio of the modulation of the image of a sinusoidal grating to that of the object, as a function of the spatial frequency and orientation of the grating (Figure 15). The modulation transfer function has values between 1 (at zero spatial frequency) and zero. The second component is the phase transfer function (PTF) that gives the shift in phase (in periods) as a function of spatial frequency of the grating. Both the MTF and PTF depend on the orientation of the grating.

Diffraction provides an upper limit to the highest spatial frequency that is passed by an optical system (that is, has some modulation of the image of a sinusoidal grating). The shape of the diffraction-limited MTF is approximately linear with spatial frequency, and its high frequency cut-off is \( \pi d/(180 \lambda) \) cycles/degree where \( d \) is the diameter of the pupil and \( \lambda \) is the wavelength of light. For a wavelength of 550 nm, the cut-off is approximately 32 \( d \) cycles/degree, where \( d \) is in mm. For well-corrected eyes in monochromatic light, the optical quality is close to diffraction limited for small (2 mm) pupils but deteriorates with increase in pupil size because of residual second-order and the higher-order aberrations. Optimum pupil size with the natural optics is about 3 mm in monochromatic light. As the diffraction limited cut-offs for 2 mm and 6 mm pupils are about 60 and 180 cycles/degree, respectively, the capacity for improvement in retinal image quality by correcting the aberrations is greater with large pupils than with small pupils. This capacity is about 10 times for a 6 mm pupil at a spatial frequency of 40 cycles/degree, which is a high spatial frequency in terms of the resolving ability of the retina (Figure 15).

There is difficulty with providing diffraction-limited vision because monochromatic aberrations are not stable. As discussed above, they fluctuate over different time scales involving microfluctuations of accommodation, varying thickness of the tear film during the blink cycle, diurnal changes in the cornea, possible changes in the overall length of the eye and ageing changes.

Also as mentioned previously, accommodation is not always accurate, with characteristic ‘leads’ and ‘lags’ of accommodation at low and high accommodative stimulus levels, respectively. A higher-order aberration-free eye is more susceptible to errors in focus caused by accommodation (or by errors in the

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**Figure 15.** Monochromatic modulation transfer functions for 2, 3 and 6 mm pupils for a diffraction limited eye (no aberrations) and an eye with near average levels of aberration. Wavelength 550 nm. For the aberrated eye, the image quality is similar for 2 mm and 3 mm pupils.
Cheng and colleagues concluded that if one state of accommodation. Interestingly, aberrations can be corrected at only second-order correction) than are normal eyes, that is, the depth-of-focus is reduced. This is shown in Figure 16. Accommodative accuracy tends to worsen at the low light levels for which natural pupils are largest and hence would otherwise give the maximum benefit of aberration-free vision.

Even if accommodation were accurate, accompanying the changing shape of the lens during accommodation in young subjects are changes in monochromatic aberrations (see Accommodation). In other words, aberrations can be corrected at only one state of accommodation. Interestingly, Cheng and colleagues concluded that if aberrations were corrected for the no-accommodation condition, most eyes would still have reduced aberrations over a zero to 6 D accommodative range.

As well as changes in pupil size with changes in illumination and accommodation, the pupil centre moves by up to 0.4 mm. This will not prevent the correction of monochromatic aberrations provided that during surgery the pupil remains within the region over which aberrations were determined and some fixed eye structure such as the limbus is used as a reference for ablation. Aberrations will reappear for foveal vision if the pupil size under low luminance exceeds that size for which an aberration correction applies.

Even if monochromatic aberrations are corrected, image quality is still affected by chromatic dispersion by the ocular media in the form of the longitudinal and transverse chromatic aberrations. Longitudinal chromatic aberration is a wavelength-dependent defocus and in comparison with the monochromatic aberrations, is very similar between people. It amounts to approximately 2 D across the visible spectrum for most people, which in wave aberration terms is equivalent to \( \Delta c_1 \) (or \( \Delta c_1 \)) = 0.36 \( \mu \)m for a 5 mm pupil. Despite its small magnitude relative to longitudinal chromatic aberration, this can cause considerable degradation of image quality because of the wavelength dependent spatial phase shifts. In the future, it may be possible to correct the chromatic aberrations, for example with a diffractive intraocular lens in which the diffraction acts in the opposite direction to the chromatic aberration of the ocular media, but alignment of such devices will be critical in order to not worsen transverse chromatic aberration and thus vision.

Another optical limitation that will not be improved by correcting monochromatic aberrations is scatter within the ocular media but in young subjects at least this is not likely to be important.

Once we get to the retina we must consider the neural limitations to vision, namely, the spacing of cones on the retina and the pathways from them to the visual cortex. The spacing of ganglion cells probably sets resolution but for foveal vision there is one-to-one correspondence between cones and ganglion cells and thus the cone spacing can be used to determine resolution limits. The cones are packed in a hexagonal array, which produces a resolution limit of \( \frac{1}{\sqrt{3}} \) where \( s \) is the centre-to-centre spacing of receptor units. This is the so-called Nyquist limit, which is half the sampling frequency. By assuming that the closest spacing of human foveal cones is 3 \( \mu \)m and that 0.29 mm on the retina corresponds to 1 degree, Williams and colleagues calculated the Nyquist limit to be 59 cycles/degree (Figure 17).

Thus, although diffraction-limited optics provide resolution of 180 cycles/degree or better with large pupils, this cannot be appreciated by the visual system.
At frequencies above the Nyquist limit, a sinusoidal grating may be seen by the visual system but the pattern will not look like the object, a phenomenon known as aliasing (Figure 18). Improvements in the resolving capacity of the visual system following correction of aberrations can be determined by plotting appropriate MTFs along with the neural contrast sensitivity function (neural-CTF) by bypassing the optics of the eye. The CTF is the contrast needed for sinusoidal gratings to be resolved by the visual system as a function of spatial frequency. Contrast is given by:

$$\frac{L_{\text{max}} - L_{\text{min}}}{L_{\text{max}} + L_{\text{min}}}$$

where $L_{\text{max}}$ and $L_{\text{min}}$ are the maximum and minimum luminances of a sinusoidal grating. The resolution of the visual system is given by where its MTFs and neural-CTF intersect, which is 43 cycles/degree in Figure 17. Within this limit, the extent to which the MTF exceeds the neural-CTF indicates how much the contrast of the sinusoidal grating may be reduced from a value of 1 for it still to be resolved. In Figure 17, at 40 cycles/degree the MTF for the aberrated eye is 40 per cent greater than the neural-CTF, which means that a sinusoidal grating of 0.7 contrast could just about be resolved. However, the MTF of the diffraction-limited eye is 10 times greater than the neural-CTF and so a sinusoidal grating of 0.1 contrast could be resolved. The potential improvements in vision in the figure represent upper limits for large pupils that are naturally present only for mesopic and scotopic luminances. At these luminances, because neural integration occurs at the retina (there is no longer a one-to-one correspondence between receptors and ganglion cells), the neural-CTFs are higher and rise more quickly than at photopic luminances. Although 6/6 is often considered to be a normal acuity, most healthy eyes up to the eighth decade exceed this in photopic conditions and a realistic value for young healthy eyes is slightly better than -0.1 logMAR (that is, better than 6/4.8). The bars of a 6/6 letter subtend one minute of arc, corresponding approximately to a spatial frequency of 30 cycles/degree, although there are other frequency components within the letter. An aberration-free resolution limit of 59 cycles/degree corresponds to approximately 6/3, so the likely improvement in visual acuity by correcting monochromatic aberrations across a range of pupil sizes is 0.14 logMAR (1.4 times) for the example in Figure 17. Taking into account the chromatic aberrations, this factor should be reduced to some extent.

Based on a consideration of the factors above, Charman and Chateau concluded that perfect correction of aberrations is likely to achieve only minor improvements in visual acuity or resolution in normal eyes but considerable improvements are possible in contrast sensitivity under mesopic and scotopic conditions. Reduction of high levels of aberration, such as still occurs in refractive surgery or other cases of distorted optics, should be most useful.

Experiments using adaptive optics to correct monochromatic aberrations yield improvements in contrast sensitivity and visual acuity. In white light, Yoon and Williams obtained 0.3 log (two times) improvement in contrast sensitivity for two subjects at 16 to 24 cycles/degree and about 0.08 ± 0.03 mean logMAR improvement in visual acuity for seven subjects. When chromatic aberration was also avoided by using monochromatic light, the improvement became 0.6 log (four times) improvement in contrast sensitivity and 0.19 ± 0.04 mean logMAR improvement in visual acuity. These results match predictions reasonably well, although the improvement in contrast sensitivity is more modest than expected, probably partly because of the difficulties in correcting aberrations exactly.

Charman and Chateau and Williams and colleagues provide further reading regarding the extent to which vision can be improved by correcting higher-order aberrations.

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