Spherical and irregular aberrations are important for the optimal performance of the human eye

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The Importance of Spherical and Irregular Aberrations for the Human Eye
Abstract

Contrast sensitivity measured psychophysically at different levels of defocus can be used to evaluate the eye optics. Possible parameters of spherical and irregular aberrations, e.g. relative modulation transfer (RMT), myopic shift, and depth of focus, can be determined from these measurements. The present paper compares measured results of RMT, myopic shift, and depth of focus with the theoretical results found in the two eye models described by Jansonius and Kooijman (1998). The RMT data in the present study agree with those found in other studies, e.g. Cambell and Green (1965) and Jansonius and Kooijman (1997). A new theoretical eye model using a spherical aberration intermediate between those of the eye models described by Jansonius and Kooijman (1998) and an irregular aberration with a typical S.D. of 0.3-0.5 D could adequately explain the measured RMT, myopic shift, and depth of focus data. Both spherical and irregular aberrations increased the depth of focus, but decreased the modulation transfer (MT) at high spatial frequencies at optimum focus. These aberrations, therefore, play an important role in the balance between acuity and depth of focus.
Introduction

Measuring contrast thresholds for sinusoidally modulated gratings of a range of spatial frequencies is an established way of evaluating spatial vision.\textsuperscript{1} Contrast sensitivity is the inverse of contrast threshold and its representation as a function of spatial frequency results in a contrast sensitivity function. This contrast sensitivity function is determined by optical and neural modulation transfer functions, i.e. contrast sensitivity at a certain spatial frequency is the product of optical and neural modulation transfers at the same spatial frequency. Earlier studies have shown that the optical modulation transfer function can be determined objectively using a double-pass laser technique.\textsuperscript{2,3} Optical modulation transfer deteriorates when a certain amount of defocus is applied, while neural modulation transfer remains the same. This occurs because defocus reduces the amplitude of sinusoidally modulated gratings on the retina.\textsuperscript{4} Thus, in order for gratings to be perceived, contrast has to be increased. The amount of the increase in contrast compared with the contrast needed at optimum focus depends on the optical properties of the eye. Therefore, measurements of contrast sensitivity at different levels of defocus can be used to evaluate eye optics.

Contrast threshold measurements under defocusing circumstances have been reported by Campbell and Green,\textsuperscript{1} Green and Campbell,\textsuperscript{5} Charman,\textsuperscript{6} Kay and Morrison,\textsuperscript{7} Jansonius and Kooijman,\textsuperscript{8} and Nio \textit{et al.} and show the attenuating effect of defocus on contrast sensitivity. Within the framework of describing this effect, Charman\textsuperscript{6} defined relative modulation transfer (RMT) at one spatial frequency as the ratio of contrast sensitivity at a certain level of defocus to contrast sensitivity at optimum focus. According to this definition, RMT depends only on optical modulation transfer, because neural modulation transfer is eliminated in the ratio of the two contrast sensitivity values. So, although it is measured psychophysically, RMT is determined solely by eye optics.

Besides experimental RMT, which is derived from the actual measuring of subjects, one can also distinguish theoretical RMT, which is calculated using eye models. Studies have shown that experimental RMT surpasses the theoretical RMT of aberration-free optics at spatial frequencies above 4 cpd.\textsuperscript{5,10} When subjects are measured without cycloplegic drugs, some of this discrepancy at positive values of defocus may be explained by the further relaxation of the crystalline lens.\textsuperscript{7} The remaining discrepancy, which is still noticeable when measuring subjects given cycloplegic drugs, can be ascribed to optical aberrations other than the applied defocus.

Jansonius and Kooijman\textsuperscript{10} studied the effect of spherical and other optical aberrations on RMT in different eye models. They found that chromatic aberrations decreased contrast sensitivity, while they hardly influenced RMT. In contrast, monochromatic aberrations showed a more pronounced effect upon
RMT. Monochromatic aberrations can be roughly subdivided into spherical, irregular, and coma-like aberrations, each having its own effect on RMT. Spherical aberration increases RMT more at negative than positive defocus, particularly at lower spatial frequencies (typically around 4 cpd) and/or with a large pupil diameter. It also decreases modulation transfer at zero defocus and increases the depth of focus. Irregular aberration comprises various other optical aberrations that are present in the eye with large inter-subject variability. Jansonius and Kooijman found that irregular aberration increases RMT at spatial frequencies above 2 cpd. Coma-like aberrations, in contrast, have no significant effect on either modulation transfer at optimum focus or RMT.

In their 1998 paper, Jansonius and Kooijman reported that the focus at which contrast sensitivity is the highest depends on the imaged spatial frequency. This dependency has been previously discussed by Green and Campbell, who noted an approximately −0.9 D shift in optimum focus in an eye with a 6-mm pupil at a spatial frequency of 3 cpd compared with the optimum focus found at 45 cpd. Spherical aberration could be one explanation for this shift. Jansonius and Kooijman further showed that neither the amount of spherical aberration nor the pupil diameter itself influences the optimum focus at spatial frequencies near the resolution of the eye. The optimum focus at decreasing spatial frequencies, however, depends increasingly on the amount of spherical aberration and on pupil diameter. Moreover, as irregular and coma-like aberrations themselves do not cause a myopic shift, the extent of the myopic shift present at low spatial frequencies can be used to estimate the amount of spherical aberration in the human eye. Irregular aberration can, however, influence myopic shift by interfering with spherical aberration, as will be discussed later.

Although spherical and irregular aberrations compromise modulation transfer at optimum focus, they increase RMT and therefore render the eye more tolerant to defocus. This increase in the depth of focus may be a positive contribution of aberrations to eye optics. Because this effect of aberrations on eye optics is directly related to visual function, depth of focus is an interesting visual parameter in the evaluation of present-day ophthalmic techniques (e.g. cataract and refractive surgery) that interfere with eye optics.

Cataract extraction and refractive surgery may influence visual performance by altering the amount of aberrations. Obviously, the evaluation of these techniques on the basis of the Snellen test is not sufficient because this test can explain complaints of glare, halos, diminished depth of focus, and reduced contrast sensitivity only at the highest spatial frequency. Measuring RMT may be important in this respect because of its relation to aberrations and depth of focus. An aberration-free optical system has a high modulation transfer but a low RMT and is, therefore, vulnerable to defocus. Thus, the aim of cataract and refractive surgery should not be confined to the creation
of perfect aberration-free optics for the eye, but to the formation of a perfect balance between modulation transfer (MT) and RMT, i.e. aberrations should be optimized rather than minimized. In order to find this balance, it is important to determine RMT in a large, healthy reference group under various conditions of defocus and pupil diameters.

Earlier studies on the measurement of contrast sensitivity under defocusing conditions were performed with either a small number of subjects, a limited number of conditions, or both. Kay and Morrison,\textsuperscript{7} for example, took through-focus measurements of only 12 subjects. Moreover, they did not apply negative defocusing conditions and studied only small pupils in which hardly any effect of spherical aberrations can be expected. Although Legge \textit{et al.}\textsuperscript{13} performed their measurements on large pupils and negative values of defocus, they only used two subjects. Other through-focus measurement studies included similar small numbers of subjects and only positive defocus values.\textsuperscript{1,8}

Objective measurements of retinal image quality and modulation transfer function of the human eye can be performed with a double-pass laser technique, resulting in a point spread function.\textsuperscript{14,3,15} Several aberrometers use the crossed-cylinder aberroscope technique described by Howland and Howland.\textsuperscript{16} There is as yet no consensus on the optimal use of this technique in measuring high order aberrations accurately.\textsuperscript{17} This may clarify the differences found in the absolute amount of spherical aberration measured with aberrometers and that reported in subjective studies.\textsuperscript{18} The Hartmann-Shack wave-front sensor can be used to measure optical aberrations of the human eye\textsuperscript{19,20,21} and to control adaptive optics and corneal ablation in order to correct the aberrations of the eye optics.\textsuperscript{22,23,24} Both objective and subjective studies of optical aberrations may complement each other in the quest to understand the role of aberrations in the human eye.

In the present study, RMT was determined in a large population at both positive and negative levels of defocus and with a wide range of pupil diameters. In order to characterize average eye optics, we compared the myopic shift and the depth of focus determined from these data with values obtained from a theoretical eye model that used various degrees of spherical and irregular aberrations. The possibility of age-related changes was also studied. This way of characterizing eye optics adds extra value to the objective measurements of optical aberrations by providing independent confirmation or incongruity of results.

**Methods**

**Experimental set-up**

Contrast threshold measurements were performed in order to determine the average RMT of a large population. The population, setup, and psychophysical method used in this study were the same as in the Nio \textit{et al.} study.\textsuperscript{9} In brief, the population consisted of 100 subjects, aged 20-69 years. Each subject
underwent routine ophthalmic screening, including measurements of visual acuity, optical correction, corneal curvature, intraocular pressure, stray light, and biometry. To prevent accommodation and fluctuation of the pupil diameter during contrast threshold measurements, each subject was administered two drops of 1% cyclopentolate hydrochloride prior to the measurements, with a 30-minute interval between drops.

Contrast sensitivity was determined per subject at spatial frequencies of 1, 2, 4, 8, 16, and 32 cpd. This was done for three different pupil diameters (2, 4, and 6 mm) and six different levels of defocus (-1, -0.5, 0, +0.5, +1, and +2 D), resulting in 18 contrast sensitivity functions. Each contrast threshold, 108 in total, was measured in a random sequence. Before the experiment started, each subject was optimally corrected in mydriasis for the viewing distance of two meters using an ETDRS letter chart. This correction defined Defocus Level Zero. Optical correction as well as defocusing lenses and artificial pupils were then put in a trial frame.

We measured contrast thresholds according to the von Békésy tracking method, using a retinal illumination of approximately 600 td. The monitor (Joyce DM4, P31 phosphor, peak wavelength 520 nm), displaying vertical sinusoidal gratings, was surrounded by a square equiluminant screen. At the viewing distance used, the square monitor covered 5.7 degrees of visual angle and the surrounding screen 41.4 degrees. By controlling the subject's head movements, we were able to minimize decentration of the artificial pupil from the viewing axis to less than 1 mm.

Data processing

The focus at which contrast sensitivity is highest varies with spatial frequency. In our study, Defocus Level Zero was defined as the optimum focus for high spatial frequencies: i.e. the smallest letters read on the ETDRS letter chart (approximately 30 cpd). RMT at a given spatial frequency was defined as the ratio of contrast sensitivity at a certain level of defocus to contrast sensitivity at Defocus Level Zero. Contrast sensitivity is the inverse of the Michelson contrast at threshold, with contrast being defined as:

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

where $L_{\text{max}}$ is the maximum luminance and $L_{\text{min}}$ the minimal luminance of the sine wave pattern. The RMT of our population was the Chauvenet-corrected average of all individual RMT values. For 100 subjects, the Chauvenet correction excludes any value outside the range of ±2.81 S.D. from the average of the population. This correction was performed to reduce the noise that originated from the limited number of reversals used in the von Békésy tracking method. This limited number of reversals was chosen so that we could
measure the 108 contrast thresholds within a reasonable amount of time, in order to avoid fatigue of the subjects.

Some subjects could not detect the gratings at a spatial frequency of 32 cpd at any level of defocus. Therefore, none of the data concerning the spatial frequency of 32 cpd were regarded in further analyses. The myopic shift was estimated from the difference between the optimum focus at 16 cpd and that at 4 cpd. The optimum focus at both spatial frequencies was determined by fitting a parabola to the averaged contrast sensitivity of the population measured as a function of defocus. The parabola was based on three data points: the highest point and its two neighboring points. To assess the average spherical aberration of our subjects, we compared the myopic shift we found to that of the two eye models described by Jansonius and Kooijman: one model estimated the upper limit of spherical aberration [eye (1)], while the other estimated the average spherical aberration [eye (2)]. Table 1 compares the spherical aberrations found in various studies, on the basis of which they calculated eye (1) and eye (2).

Because irregular aberration influences RMT, we evaluated its effect upon myopic shift and depth of focus by applying various amounts of irregular aberration to eye (1) and eye (2). In his study, Van den Brink showed an irregular distribution of dioptric power in eye optics; i.e. each location on the optical surface varied randomly in dioptric power from its neighboring location. The distribution with which this dioptric power varied could be described by a Gaussian curve with a typical S.D. of approximately 0.5 D.

<table>
<thead>
<tr>
<th></th>
<th>h = 1 mm</th>
<th>h = 2 mm</th>
<th>h = 3 mm</th>
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<tbody>
<tr>
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<td>0.60</td>
</tr>
<tr>
<td>Koomen et al. (1949)</td>
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<td>Von Bahr (1945)</td>
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<td>Eye (1)*</td>
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<td>1.71</td>
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<td>Eye (2)*</td>
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<td>0.93</td>
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*Taken from Jansonius and Kooijman

By varying the S.D., we were able to realize different amounts of irregular aberration in the eye models. The resulting values for myopic shift and depth of focus were then compared with the values measured in our population.

The spherical aberration of the entire eye originates in the cornea and the crystalline lens. The average spherical aberration of the cornea of our population was determined by using topographic pictures (TMS-1 version 1.61 cornea topographer, USA). Unfortunately, this version of the cornea topographer was not equipped with software to compute corneal aberrations. Therefore, we determined this parameter on the basis of the topographic
pictures. First, we calculated the spherical aberration of the cornea of each subject, assuming a spherically shaped cornea and using Equation (2): \[ P_{sa} = \frac{n^2 h^2 P}{2n'^2 R^2} \] where \( P_{sa} \) is the spherical aberration in diopters, \( n \) the refractive index in object space (1.00), \( n' \) the refractive index in image space (1.33), \( h \) the ray height (distance from the center of the cornea) in meters, \( R \) the radius of the surface in meters, and \( P \) the power of the corneal surface in diopters. \( P \) was estimated using the picture derived from the cornea topographer. The dioptric power was read at four points 1 mm from the optical axis. These points were spaced 90 degrees from each other and placed on the cylindrical axes determined by the cornea topographer. \( P \) was the mean value of the four points and \( R \) was calculated from \( P \) using Equation (3): \[ R = \frac{n' - n}{P} \]

In reality, the cornea flattens towards its periphery. Therefore, the actual spherical aberration of the cornea is less than that expected from Equation (2). We corrected the spherical aberration of the total cornea for a 6-mm pupil by subtracting the difference in average corneal power measured at 3- and at 1-mm distances from the optical axis from the spherical aberration calculated with Equation (2). Then, we averaged the results to obtain the spherical aberration of the population. Regression analysis was performed to study any age-related changes in spherical aberration of the cornea.

The depth of focus for a specific spatial frequency can be defined as the dioptric range within which contrast sensitivity exceeds half its maximum value.\(^{13}\) We used a standard spline routine (EasyPlot V4; Spiral Software, Bethesda, MD, USA) to fit a curve through our averaged contrast sensitivity data points as a function of defocus at 8 cpd (Figure 1). This spatial frequency was chosen because it represents an intermediate between frequencies important for reading newspaper letters, i.e. 12 cpd, and for detecting edges, i.e. 3 cpd.\(^{29,8}\) Because the measured dioptric range was limited on the negative side, the depth of focus was defined as twice the positive half of the dioptric range in which the contrast sensitivity exceeded half its maximum value. The depths of focus we calculated for eye (1) and eye (2) were compared with both the experimental results and the results obtained using the theoretical model described by Nio et al.\(^9\)
Figure 1. 
Average contrast sensitivity data points from Nio et al. (2000) at 8 cpd as function of defocus. A curve was fitted using a standard spline routine (EasyPlot V4; Spiral Software, Bethesda, MD, USA). Depth of focus (DOF) was defined as twice the positive half of the dioptric range in which the contrast sensitivity exceeded half its maximum value.

Results

The average contrast sensitivity functions of the population at different levels of defocus determined for a 4-mm pupil are shown in Figure 2A. As can be seen, contrast sensitivity is hardly affected by defocus at 1 cpd. Defocus does, however, decrease contrast sensitivity progressively with increasing spatial frequencies until 4 cpd is reached. The contrast sensitivity functions run approximately parallel at higher spatial frequencies, as described in earlier papers.¹⁷ The effect of defocus is more pronounced in larger pupils than in smaller ones, as can be seen in the RMT graphs: Figure 2 B-D for 2-, 4-, and 6-mm pupils, respectively. The difference between 4- and 6-mm pupils is relatively small compared with that between 2- and 4-mm pupils.

Figure 3 shows the optimum foci on the basis of the averaged contrast sensitivity measured at spatial frequencies of 4 and 16 cpd for 2-, 4-, and 6-mm pupils. The difference between the optimum foci at 4 and 16 cpd, i.e. the myopic shift, increases with pupil diameter (p<0.05). The largest increase occurs between the 4- and 6-mm pupils, from -0.27 to -0.40 D.
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Figure 2A.
Average contrast sensitivity functions of the population at different levels of defocus determined for a 4-mm pupil.

Figure 2B.
Figure 2B-D.
The relative modulation transfer function (RMTF) for 2-, 4-, and 6-mm pupils, respectively. The typical SE shown in the graphs applies to all points. Data from Nio et al. (2000).
The average individual myopic shift with a 6-mm pupil ±1 S.E. was -0.33 ± 0.05 D, which is in reasonable agreement with the myopic shift found using the averaged contrast sensitivity values: -0.40 D. No significant correlation was found between the individually determined myopic shift measured with a 6-mm pupil and age ($p=0.69$).

The average myopic shift of our population (Figure 4) of -0.40 D, measured with a 6-mm pupil, was somewhat less than the corresponding theoretical value of eye (1) and more than that of eye (2). When eye (1) is compared with eye (2) at different pupil diameters and different amounts of irregular aberration, it can be seen that the shift in eye (1) is larger than in eye (2). Furthermore, the shift increases in both eyes with increasing pupil diameter.

Figure 5A shows the experimentally acquired values of the depth of focus. One set of data was calculated by doubling the positive half of the dioptric range in which the contrast sensitivity at 8 cpd exceeds half its maximum value (Figure 1). The depths of focus based on the theoretical model of Nio et al.\textsuperscript{9} concerning the same population and the Charman\textsuperscript{6} data obtained from a single subject are shown for comparison. The theoretical values of the depth of focus in eye (1) and eye (2) at different values of irregular aberration are plotted as a function of pupil size in Figure 5B. These two data sets differ only slightly: the experimental and theoretical data show the same pattern and are in the same range. The depth of focus that was obtained by doubling the positive range coincides with the theoretical values obtained with 0.3-0.5 D irregular aberration in eye (2).

**Figure 3.**
Optimum focus based on the average contrast sensitivity of all subjects measured at spatial frequencies of 4 and 16 cpd for different pupil diameters. The myopic shift, defined as the difference between the optimum foci at 4 and 16 cpd, increases with increasing pupil diameter ($p<0.05$).
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Figure 4.
Experimental and theoretical myopic shift with various amounts of irregular aberration (IA). At large pupil diameters, the experimental results show less myopic shift than the eye (1) model and somewhat more than the eye (2) model.

Figure 5A.
Comparison of different depths of focus of a spatial stimulus of 8 cpd. Model Nio (2000) is calculated using the mixed effect model, based on measurements described by Nio et al. (2000). Double positive range is the depth of focus calculated by doubling the positive half of the dioptric range at which contrast sensitivity exceeds half its maximum value (see Figure 1). Charman (1979) comprises of the data measured in one subject.
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Figure 5B.
Comparison of different depths of focus of a spatial stimulus of 8 cpd.
Depth of focus as calculated using the eye (1) and eye (2) models at different amounts of irregular aberration (IA): 0.3, 0.5, and 0.7 D S.D. The experimental results of the double positive range (Figure 5A) are also shown.

Table 2. Spherical aberration of the cornea: $C_1$ and $C_3$ are local corneal powers (D) measured 1 and 3 mm from the optical axis, respectively; $R$ (mm) is the central corneal radius; $P_{sa,s}$ (D) is the spherical aberration for a ray height ($h$) of 3 mm, i.e. a 6-mm pupil, based on the central radius; $P_{sa,c}$ (D) is the spherical aberration, corrected for the attenuating effect of the peripheral cornea on the spherical aberration.

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<td>Average</td>
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<tr>
<td>$C_1$ (D)</td>
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</tr>
<tr>
<td>$C_3$ (D)</td>
<td>42.7</td>
</tr>
<tr>
<td>$R$ (mm)</td>
<td>7.66</td>
</tr>
<tr>
<td>$P_{sa,s}$ (D) at $h$ = 3 mm</td>
<td>1.88</td>
</tr>
<tr>
<td>$P_{sa,c}$ (D) at $h$ = 3 mm</td>
<td>1.47</td>
</tr>
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</table>

The average spherical aberration of the cornea ±1 S.E. was calculated to be 1.47 D ± 0.04 (Table 2) and did not show any significant effect caused by age, as determined by regression analysis ($p=0.97$). Note that the average local corneal power is smaller at 3 mm from the optical axis than at 1 mm. This implies an attenuating effect of the peripheral cornea upon spherical aberration, calculated on the basis of the curvature 1 mm from the optical axis.
It is possible to determine RMT by measuring contrast sensitivity psychophysically at different levels of defocus. Relative modulation transfer is an optically determined parameter, because any neural influence is eliminated in the ratio of contrast sensitivity under defocus to contrast sensitivity at Defocus Level Zero. Relative modulation transfer as a function of spatial frequency results in the relative modulation transfer function (RMTF). This function has been investigated in earlier studies using limited number of subjects. Campbell and Green,\textsuperscript{1} for example, performed their classic study on just one subject. Charman\textsuperscript{6} also studied only one subject. Both studies measured contrast sensitivity at comparable levels of defocus and with similar sizes of artificial pupils. Their data fall within the range of ±2 S.D., which is the prediction interval of our mean RMTF. Kay and Morrison\textsuperscript{7} studied 12 subjects using similar levels of defocus and a 3-mm pupil. Although their RMTF was below 2 S.E. of our mean RMTF measured with a 2-mm pupil, most of their data points were within 2 S.E. of our mean RMTF measured with a 4-mm pupil. Most of the points that were out of the range were above our RMTF of the 4-mm pupil. Therefore, their data with a 3-mm pupil are in agreement with ours. Jansonius and Kooijman\textsuperscript{8} took measurements in six subjects not administered cycloplegic drugs. The natural pupil size of their population varied between 4.5 and 7 mm. Student’s t-test showed no significant difference ($p > 0.05$) in three-quarters of their data points when compared with our RMTF of the 4-mm pupil. The remaining quarter (i.e. the data points with a significant difference) was located above our mean RMTF, but still within two S.D. A possible explanation for the discrepancy between the two studies is the uncertain pupil diameter during the measurements and the absence of cycloplegic drugs. Kay and Morrison\textsuperscript{7} found that the effect of positive defocus without cycloplegia is less pronounced than with cycloplegic drugs. They ascribed this effect to lens relaxation in response to positive defocus. In summary, the RMT data presented in this study agree well with the results found in other studies.

Myopic shift is associated with an asymmetry of contrast sensitivity around the Defocus Level Zero at lower spatial frequencies. This is illustrated by the RMTF graphs in Figure 2 B-D: at a spatial frequency of 4 cpd, the graphs show that -1 D defocus has a smaller attenuating effect on contrast sensitivity than +1 D defocus. Therefore, the ratio of contrast sensitivity at -1 D defocus to that at +1 D defocus for a spatial frequency of 4 cpd might also function as a measure of spherical aberration. This ratio can be readily assessed in comparison with myopic shift, because it takes fewer contrast thresholds to measure. This ratio was determined for our population and for the theoretical eye models of Jansonius and Kooijmans\textsuperscript{10} [eye (1) and eye (2)]. The measured ratio of contrast sensitivity at -1 D defocus to that at +1 D defocus, shown in Figure 6, increased linearly with increasing pupil diameter from 1.7 to 6.0 for
the 2- and 6-mm pupils, respectively. The effect of spherical aberration on this ratio of contrast sensitivity was then analysed using model eyes (1) and (2). Figure 6 shows that the calculated ratio in the theoretical eye models (1) and (2) was approximately 1 for a 2-mm pupil, where no clear effect of aberrations was expected. For larger pupil diameters, where the effect of spherical aberration is larger, eye (1) had a larger ratio than eye (2). This ratio also showed an increase with increasing pupil diameter in both eye (1) and eye (2). This indicates that spherical aberration is depicted in the ratio of contrast sensitivity at -1 D defocus to that at +1 D defocus.

![Figure 6](image.png)

**Figure 6.**
Ratio of contrast sensitivity at -1 D defocus to that at +1 D defocus with respect to the Defocus Level Zero as a function of pupil diameter. Experimental results ± 1 SE and theoretical values for eye (1) and for eye (2) are shown. Irregular aberration (IA): 0.3, 0.5, and 0.7 D S.D. The ratio appears to be linearly related to the pupil diameter in the range studied. The experimental results resemble those of eye (1) with an irregular aberration less than 0.3 D.

Although slightly higher, the measured ratio of contrast sensitivity at -1 D defocus to that at +1 D defocus was comparable with that of the theoretical eyes (Figure 6). One cause of this higher ratio could be inherent to the way in which Defocus Level Zero was determined. Our subjects were given positive lenses until they could no longer maintain their best visual acuity; i.e. at 30 cpd, our subjects had their individual depth of focus at their disposal to put up with negative defocus. This yielded an asymmetry that was not caused by spherical aberration. Indeed, with a 2-mm pupil, where the effect of spherical aberration is considered minimal, the ratio of our population was higher than that of both eye (1) and eye (2). Another possible cause of the higher measured ratio values is the low amount of irregular aberration seen in our
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subjects. A third cause could be a higher amount of spherical aberration in our population than that calculated for eye (1). The latter explanation is not very likely, however, because the measured myopic shift of our population was still lower than that of eye (1) (Figure 4).

As mentioned before, irregular aberration influences modulation transfer (MT) and thus also the depth of focus, myopic shift, and the ratio of contrast sensitivity at -1 D defocus to that at +1 D defocus (see Figures 4-6). Although irregular aberration itself does not cause an asymmetry around Defocus Level Zero, it does attenuate the effect of spherical aberration on both myopic shift and the ratio of contrast sensitivity at -1 D defocus to that at +1 D defocus. Furthermore, it increases the depth of focus. Compared with a diffraction-limited system, a larger depth of focus is the major advantage of an optical system with aberrations. The experimental depth of focus found by doubling the positive half of the dioptric range in which the contrast sensitivity exceeds half its maximum value appears to be lower than the theoretical depth of focus found with the mixed effect model described by Nio et al. (Figure 5A). Charman found values of 2.4, 1.4, and 1.3 D for 2-, 4-, and 6-mm pupils, respectively, in his single subject. The values of the depth of focus that we calculated for eye (1) and eye (2) showed an increase with spherical aberration: for comparable amounts of irregular aberration, the values were higher in eye (1) at pupil diameters larger than 2 mm. The effect of spherical aberration on the depth of focus seems to decrease with increasing irregular aberration: using a 6-mm pupil, the difference between eye (1) and eye (2) decreased with increasing irregular aberration. Apparently, there is a maximum to the increase in the depth of focus as a result of the two kinds of aberrations studied. Aberrations in general dim the effect of defocus on MT. This is illustrated in Figure 2 B-D: an increase in pupil diameter from 4 to 6 mm had less effect than an increase from 2 to 4 mm. This implies that, at large amounts of defocus, pupil diameter is of little importance for MT.

Aberrations decrease MT at high spatial frequencies at optimum focus. To assess the effect of spherical and irregular aberrations on this issue, we calculated the MT at 30 cpd for model eyes (1) and (2). Figure 7 shows that the MT of eye (1) is slightly but consistently lower than that of eye (2). Also, the MT decreased in both model eyes with increasing irregular aberration and pupil diameter. This confirms the idea that, at optimum focus, both spherical and irregular aberrations attenuate the MT for high spatial frequencies.
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Figure 7.
Theoretical modulation transfer at optimum focus for 30 cpd in eye (1) and eye (2) as function of pupil diameter. Irregular aberration (IA): 0.3, 0.5, and 0.7 D S.D.

Figure 8.
Through-focus MT curves of a diffraction-limited eye model and a realistic physiological eye model based on eye (2) with an irregular aberration (IA) of 0.5 D S.D. The latter curve has been shifted +0.55 D in order to study the differences with the diffraction-limited eye.
In the case of aberration-free or diffraction-limited optics in the human eye, the MT is not attenuated by aberrations, possibly allowing for higher values of contrast sensitivity and visual acuity. In order to estimate the profitability of such optics, through-focus MT curves at 8 cpd of a diffraction-limited eye were compared with an approximation of a realistic physiological eye: model eye (2) with an irregular aberration of 0.5 D (Figure 8). At this spatial frequency, the diffraction-limited optics showed an MT of nearly 1.0 at Defocus Level Zero, whereas the model eye with aberrations had an optimum MT of 0.37 at a focus of –0.55 D. There are, however, some critical notes to this seemingly large advantage of a diffraction-limited eye. First of all, its depth of focus is approximately half that of an eye with aberrations. Any object outside the focal plane will appear blurred and with decreased contrast. Even more important, the absolute MT of the physiological eye model is higher for a defocus value outside the +0.38 to -0.38 D range. This can be observed when the curve in Figure 8 for the physiological eye is shifted +0.55 D. The relative blur of objects outside the focal plane may be a trigger of accommodation, which would compensate for the limited depth of focus. However, continuous accommodation could cause complaints of asthenopia. Furthermore, it is known that accommodation and aging alter the spherical aberration of the eye, thereby reducing the effect of minimizing aberrations in eye optics.

Another consideration is that the retinal cone mosaic can maximally reconstruct a spatial frequency of approximately 70 cpd. The cutoff frequency of the MTF in a diffraction-limited eye is approximately 180 cpd, which should allow for a visual acuity of 6.0 or 120/20. However, if frequencies higher than 70 cpd are offered to the retina, aliasing may occur in which spurious gratings are perceived that distort the image.

Nevertheless, the advantage of diffraction-limited eye optics is a higher MT level in the focal plane at all of the spatial frequencies the retina can resolve. This does not, however, always lead to a better modulation of the increased contrast of the image offered to the retina. Although P retinal-ganglion cells may profit, M retinal-ganglion cells that are responsible for perception of low contrast saturate at approximately 15% contrast. This seems to be in accordance with most of the contrast of objects viewed during activities of daily life. So, if the eye optics are no longer the limiting factors of vision, the retinal characteristics become the decisive element in vision. Snyder et al. consider retinal characteristics to be a product of biological needs and evolution. Eye optics, therefore, may have developed in such a way to best serve the retinal cone mosaic. If these optics are freed from aberrations, Schwiegerling predicts a theoretical limit of foveal vision between 20/12 and 20/5 for common pupil diameters. The true advantage of an aberration-free system in daily life remains to be seen.

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Chapter 4
Conclusion

Relative modulation transfer described as a combination of myopic shift, a ratio of contrast sensitivity at -1 D defocus to that at +1 D defocus, and depth of focus, gives an indication of the amount of spherical and irregular aberrations present in a healthy population. Our experimental myopic shift data show a best fit with a theoretical eye that has a spherical aberration between those of the eye (1) and eye (2) models of Jansonius and Kooijman\(^1\) and an irregular aberration of 0.3-0.5 D. Recently, Artal \textit{et al.}\(^{37}\) showed that internal optics compensate for the spherical aberration of the cornea. Our results, showing a spherical aberration between 0.93 and 1.71 D for the entire eye and 1.47 D for the cornea, agree with their conclusion.

Our results of the ratio of contrast sensitivity at -1 D defocus to that at +1 D defocus agree most with the eye (1) model with an irregular aberration below 0.3 D S.D.

The experimental results of the depths of focus approximate the values calculated for eye (1) with an irregular aberration of 0.3 D and for eye (2) with an irregular aberration between 0.3 and 0.5 D. On the basis of these numbers, the experimental estimation of the depth of focus seems to be better than the results of the mixed effect model of Nio \textit{et al.}\(^9\)

Sufficient depth of focus, contrast sensitivity, and visual acuity, among others, are essential for good visual performance in daily life. In order to achieve this goal, aberrations should be optimized rather than minimized whenever eye optic corrections are carried out, e.g. in cataract and refractive surgery.
Chapter 4

References
