1

Optics of the Human Eye

W. N. Charman

Introduction

Optically, the human eye is fairly typical of vertebrate eyes in general and is not distinguished by any remarkable characteristics. It lacks, for example, the exquisite resolution of the eyes of many raptors, the enormous range of focussing power of some diving ducks and cormorants or the subtle duplicated optical system of the four–eyed fish, anableps. Nevertheless its optical performance is well matched to the capabilities of the neural network which it serves. As Helmholtz remarked with characteristic acuteness over a century ago, '*The eye has every possible defect that can be found in an optical instrument and even some which are peculiar to itself; but they are so counteracted, that the inexactness of the image which results from their presence very little exceeds, under ordinary conditions of illumination, the limits which are set to the delicacy of sensation by the dimensions of the retinal cones*' (Helmholtz, 1962).

Knowledge of the way in which light propagates through the eye is crucial to the understanding of the abilities of the whole visual system, since formation of an optical image on the retina is the first stage in the complex processes which lead to perception. In general the retinal image is not a simple, reduced-scale reproduction of the external world but differs from it in such factors as spatial form, spectral composition, polarization and light flux, due to the transmission and imaging characteristics of the ocular media.

In this review we shall consider first the optical characteristics of the individual components of the eye and then the way in which these combine to determine the overall optical performance. Related reviews which set these human characteristics in the wider context of the evolution of vertebrate and invertebrate eyes will be found in Volume 2 of this series (Gregory, 1991).

Optical Components of the Eye

The main optical features of the eye are illustrated in Fig. 1.1. Full details of these structures are given in numerous texts (e.g. Duke-Elder and Wybar, 1961; Records, 1979; Davson, 1980; Moses and Hart, 1987). The techniques used to determine the dimensions and optical characteristics of each component are discussed by Charman, Chapter 16 and Henson, Chapter 17.
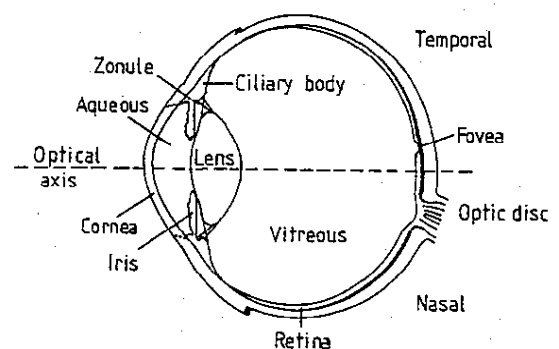

Fig. 1.1 *Schematic horizontal section of the human eye.*

General Shape of the Eye

The globe of the adult eye can be approximated as a sphere with an average radius of ~ 12 mm. It is completed anteriorly by the transparent cornea, which forms a roughly spherical cap with a radius of curvature ~ 8 mm, the distance between the centres of the spheres being about 5 mm (e.g. Le Grand and El Hage, 1980). Most eyes are, in fact, somewhat flattened posteriorly and the larger sphere is slightly conical anteriorly. There is also often some asymmetry about the anterior-posterior axis (Deller *et al.*, 1947;

Sorsby, 1948). The eye at birth is only some 18 mm long, as compared to ~ 24 mm in adulthood, so that the relative growth rates of different parts of the eye need somehow to be synchronized if the potential for a sharp retinal image is to be maintained throughout early life: the implications of this for the development of refractive error are discussed by Young (Chapter 2).

Cornea

The vertical and horizontal diameters of the adult cornea are about 12 and 11 mm respectively. It is covered by the tear film, some 5 μm thick (Ehlers, 1965; Maurice, 1967). Among other functions, this serves to maintain the smoothness of the optical surface. Undoubtedly the irregularities in the thickness of the components of this film, of the order of the wavelength of light, which progressively arise during the inter-blink period as a result of evaporation (Guillon, 1982; Josephson, 1983), must have some effects on the quality of the retinal image but little attention has so far been devoted to these.

The cornea is a relatively complex structure made up of several distinct layers – the epithelium, Bowman's membrane, the stroma, Descemet's membrane and the endothelium, the stroma making by far the greatest contribution (~ 90%) to the overall thickness of about 0.5 mm at the centre of the cornea and 0.7 mm at the periphery (Steindorff, 1947; Martola and Baum, 1968; Hirji and Larke, 1978; Azen et al., 1979). The optical inhomogeneity contributes to the scattering of light from the cornea that allows this element to be seen in optical section with the slit-lamp microscope. Of particular importance is the way in which the constituent collagen fibres, each 25–33 nm in diameter, are arranged in the stroma. Not only do their structure and arrangement lead

to birefringence (see Bour, Chapter 3) but also their quasi-uniform size and regularity are of vital importance to corneal transparency. The fibres form approximately regular lattices. Thus, although scattering occurs from each fibre due to the differences in refractive index between the fibres and the interstitial material, the resultant scattered light from the overall lattice interferes destructively in all directions except that of the incident radiation, leading to good corneal transparency unless the stromal lattice is disturbed by oedema or trauma (Maurice, 1957, 1969, 1970; Hart and Farrell, 1969; Feuk, 1970, 1971; Benedek, 1971).

Although the anterior surface of the cornea approximates to a sphere, the details of its contour have received intensive study as a result of its influence on the fit of contact lenses (e.g. Prechtel and Wesley, 1970; Mandell and St Helen, 1971; Clark, 1972, 1974a, 1974b; Fujii et al., 1972; Mandell, 1974; El Hage, 1976; Kiely et al., 1982; Edmund and Sjontoft, 1985; Guillon et al., 1986; Dingeldein and Klyce, 1989). Since, too, the anterior cornea makes the greatest individual contribution to the overall power of the eye of all its optical surfaces, due to the large change in refractive index from air to cornea, its form is of great interest with regard to its role in determining the aberration of the retinal image.

The general picture of the corneal contour that has emerged is that, typically, the radius of curvature near the anterior pole takes its smallest value of about 7.8 mm and that the surface progressively flattens towards the periphery. There are, however, wide variations in the topography of individual corneas. In some instances the corneal curvature may steepen towards the periphery and complete rotational symmetry is unusual (e.g. Bonnet, 1959; Mandell and St Helen, 1971; Clark, 1974a; Kiely et al., 1982; Guillon et al., 1986; Dingeldein and Klyce, 1989). Examples are shown in Fig. 1.2 (a) and (b).
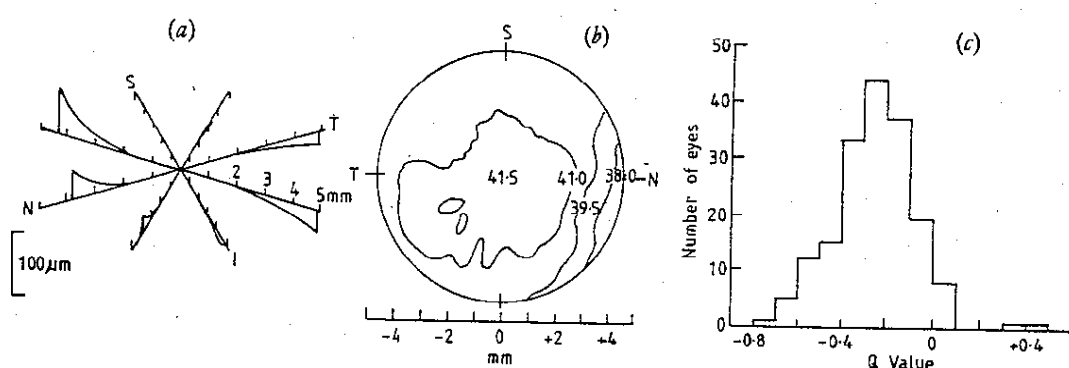

Fig. 1.2 *The corneal contour: (a) An example showing the asymmetries present in an individual cornea. The data are expressed in terms of the departures from a spherical surface which best fits the central cornea, measured along the horizontal, vertical and 45° oblique meridians (after Clark, 1974a). (b) An example showing the variation in the local power in dioptres of the corneal surface of the individual eye (after Dingeldein and Klyce, 1989). (c) Histogram showing the distribution of the corneal asphericity parameter, Q, in a sample of 176 eyes (after Kiely* et al., *1982). $Q = 0$ corresponds to a spherical surface.*

surface as a conicoid with rotational symmetry about the z-axis. The equation of the conicoid can be written in a variety of forms. Kiely et al. (1982) use the equation:

$$x^2 + y^2 + (1 + Q)z^2 = 2zr_0 \tag{1.1}$$

where x and y are Cartesian coordinates perpendicular to the z-axis of rotational symmetry, r_0 is the radius of curvature at the corneal apex and Q is an asphericity parameter which specifies the form of the conicoid: $Q > 0$ is an ellipsoid with the major axis in the x, y plane; $Q = 0$ is a sphere; $-1 < Q < 0$ is an ellipsoid with its major axis in the z direction; $Q = -1$ is a paraboloid with its axis along the z-axis; $Q < -1$ is a hyperboloid.

It should be noted that the asphericity parameter Q is related to the parameters p (shape factor) and e (eccentricity) sometimes used by other authors, by the equations:

$$p = 1 + Q = 1 - e^2 \tag{1.2}$$

It is also obvious that, for a system which is rotationally symmetrical about the z-axis, the terms $(x^2 + y^2)$ that appear in Kiely et al.'s equation can be replaced by a single term y^2, where y is now the distance perpendicular to the axis of symmetry. Kiely et al. (1982) find the means and standard deviations of r_0 and Q to be 7.72 ± 0.27 mm and -0.26 ± 0.18 respectively, implying that the typical surface approximates to an ellipsoid with its major axis along the z-direction, i.e. along the optical axis: they note that this is compatible with the findings of several earlier authors. The histogram in Fig. 1.2(c) shows that, as is indicated by the relatively large standard deviation of the parameter Q, asphericity values for individual eyes are widely distributed and can include some examples in which the cornea steepens rather than flattens in the periphery. The results of Guillon et al. (1986) lead to broadly similar conclusions.

The typical corneal flattening does not appear to be sufficient to fully correct the spherical aberration of the cornea, although it substantially reduces it in comparison with that for a spherical surface (Kiely et al., 1982). This flattening is also of significance to imagery at large field angles, since the ray pencils then pass through peripheral regions of the cornea.

The radius of curvature of the posterior surface of the cornea has received much less attention than that of the anterior surface, but it is approximately 6.5 mm centrally, so that the cornea forms a meniscus which is thinner at the centre than at the edge. Although there are microscopic irregularities the typical bulk refractive index of the cornea is about 1.377, depending upon the wavelength and temperature at which it is measured. This gives to the anterior and posterior surfaces of the cornea paraxial powers of about + 49 and – 6 D respectively.

transparent liquid. It is being formed and circulated continuously and normally constitutes an optical medium, scattering very little light unless cellular debris accumulates as the result of pathology or the after effects of surgery. The refractive index is about 1.337. The depth of the anterior chamber reduces throughout adult life, from about 3.8 to 3.0 mm, due to the corresponding increase in thickness of the lens (e.g. Weale, 1963).

Pupil

This circular opening in the iris, lying approximately tangential to the anterior surface of the lens, plays the important role of aperture stop to the optical system of the eye. It thus not only controls the amount of light flux contributing to the retinal image, but also affects the quality of that image through its influence on diffraction, aberration and ocular depth-of-focus. The finite thickness of the iris slightly reduces the cross-section of the ray pencils contributing to off-axis images.

Although ambient lighting has, perhaps, the most important influence on pupil diameter, the latter is also affected by such factors as age, accommodation, emotion and drugs (e.g. Hess, 1965; Lowenstein and Loewenfeld, 1969; Zinn, 1972; Janisse, 1973; Thompson, 1987). Woodhouse (1975) and Woodhouse and Campbell (1975) conclude that the major role of pupillary construction to light is to reduce retinal illumination and to prepare the eye for a return to darkness. A mobile pupil allows substantially better stimulus detection during the first few minutes of dark adaptation as compared to a fixed pupil.

Lens

The lens (Fig. 1.3) is built up from fine fibres which are hexagonal in cross-section and stretch in arcuate fashion from the anterior to the posterior pole of the lens. Neighbouring fibres are rigidly linked through ball-and-socket joints along their edges so that relative movement cannot occur. It is, perhaps, the most remarkable of the optical components of the eye in that it continues to grow throughout life, through the addition of new lens fibres originating at the lens equator (see, e.g. Weale, 1963, 1982). The dimensions increase by about 50% between birth and the age of 70, so that the usually-quoted thickness of 4 mm and diameter of 9 mm, together with anterior and posterior radii of curvature of about 10 and 6 mm respectively, can only be considered as representative. A further consequence of this continued addition of new lens fibres is that the central region or nucleus of the lens is older than the outer layers, or cortex, the lens having an

since the vitreous tends to liquify. The consistency normally increases from the lens towards the retina and from the centre towards the periphery (Le Grand and El Hage, 1980). Optically, the vitreous body in the young is almost clear and free of refractive irregularities, although these become more prevalent during old age (Weale, 1963). The refractive index of the vitreous is about 1.336; Millodot (1976) has suggested that this may increase slightly with age.

The Retinal Surface

Although an eye may produce a sharp image, it is necesary that the retinal receptors be in an appropriate spatial position to accept this image, i.e. the receptor outer segments should lie close to the surface of optimal image quality. On the optical axis this implies that close constraints exist on the length of the eye: if it is too long myopia will occur and if too short hypermetropia (see below). These effects have their counterparts at larger field angles. While the retinal surface approximates to a sphere, this need not necessarily match the curvature of the image surface. Moreover, the retina does not always have rotational symmetry about the axis of the eye, so that even if the optical components had perfect rotational symmetry, a mismatch would occur between the optical image and the retinal surface at some parts of the field. Fortunately, as the surfaces of the eye are approximately concentric about the pupil, these mismatches are, in general, surprisingly small. At the retina the transmission and absorption of light in the receptor outer segments is strongly affected by optical waveguide effects: these are fully discussed by Enoch and Lakshminarayanan (Chapter 12) and will not be considered further here.

Ocular Asymmetries and the Optical Axis of the Eye

As has already been noted in the cases of the anterior cornea and retina, the optical surfaces of the eye may lack rotational symmetry and their nominal centres of curvature may not lie on a common axis: meridional changes in radius of curvature lead to ocular astigmatism (see Howland, Chapter 18). The pupil centre may also be displaced. These effects can be demonstrated by the observation of the Purkinje images (see Henson, Chapter 17). Nevertheless, these various decentrations, asymmetries and tilts are small in most eyes and a nominal optical axis can be identified. It is interesting to note that this optical axis does not in general intersect the retina at the fovea but rather some 1.5 mm (5°) nasal and 0.5 mm (1.5°) superior to the fovea (Emsley, 1952). If we define the visual axis as the two parallel lines passing from the fixation point to the first nodal point of the eye and from the second nodal point to

the fovea, the angle between the optical and visual axes is called angle α and typically takes a value of about 5°. A variety of other, closely-related angles whose use is claimed to be advantageous under some circumstances, are defined in the literature (e.g. Duke-Elder and Abrams, 1970; Le Grand and El Hage, 1980).

Eye Models and Refractive Error

Given a knowledge of the parameters of a normal eye it is obviously possible to construct a model with similar imaging characteristics. Such models have many uses, e.g. in predicting the dimensions of retinal images and their changes with accommodation and correcting lenses, in relating the image dimensions to those of retinal receptors, in predicting the effects of changes in ocular parameters on overall refractive state, and in relation to surgery, particularly in determining the power of intraocular lens that is required (see Ridgway, Chapter 7).

Paraxial Models

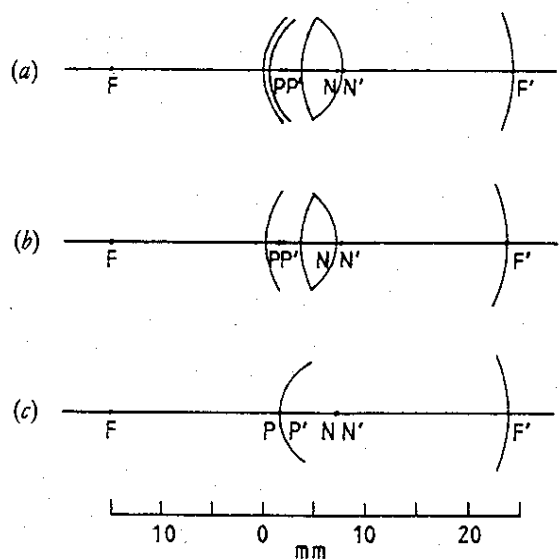Any model is likely to involve some simplification in comparison with real eyes and most earlier models assumed

Fig. 1.5 *Examples of paraxial models of the human eye. In each case F, F'; P, P'; N, N' represent the first and second focal, principal and nodal points respectively. (a) Unaccommodated schematic eye with four refracting surfaces (Le Grand and El Hage, 1980). (b) Simplified, unaccommodated schematic eye with three refracting surfaces (Emsley, 1952). (c) Reduced eye with a single refracting surface (Emsley, 1952). Note that progressive reduction in the number of optical surfaces produces only minor changes in the positions of the cardinal points.*

that the lens was homogeneous and that imagery was paraxial. In reality, as Le Grand and El Hage (1980) point out in an excellent discussion of the limitations of such models, the pupil diameters and other dimensions of the eye are such that, even with the restricted pupil appropriate to photopic conditions, the ray angles exceed those for which the paraxial approximation $\sin\omega = \tan \omega = \omega$ holds. Watkins (1972) notes that for calculations accurate to five significant figures the paraxial approximation is only adequate for object field angles up to ± 2° and entrance pupil diameters of < 0.5 mm. These models must therefore be used with care.

Paraxial models have been proposed by a variety of authors (e.g. Gullstrand, 1924; Emsley, 1952; Ogle, 1968; Duke-Elder and Abrams, 1970; Le Grand and El Hage, 1980; Fincham and Freeman, 1980; Bennett and Rabbetts, 1988), the models differing chiefly in their exact choice of parameters and the degree of simplification involved. Examples are shown in Fig. 1.5 and the associated values of the parameters are listed in Table 1.1. In schematic eyes the cornea and lens are each usually represented by a pair of surfaces: a single surface is used for the cornea in the simplified schematic eye. Since the two nodal and two principal points lie very close together in all the models, considerable simplification can be achieved by using a reduced eye model consisting of a single refracting surface. A reduced eye model of this type is perfectly adequate for many calculations in visual optics (e.g. Emsley, 1952; Obstfeld, 1978).

Wide-Angle Models

With advances in understanding of the form of the surfaces of the eye and the refractive index distribution in the lens, and in the application of computers to ray tracing,

there have been an increasing number of attempts to develop non-paraxial models which give useful predictions of performance away from the axis. The simplest approach is to introduce aspheric surfaces of revolution while retaining homogeneous media (Lotmar, 1971; Drasdo and Fowler, 1974; Drasdo and Peaston, 1980; Kooijman, 1983; Navarro et al., 1985). Such models can be validated by comparing their predictions of off-axis aberration with that observed in real eyes. More elaborate models may involve the use of inhomogeneous lenses. The earliest of these (although strictly a paraxial rather than a wide-angle model) was Gullstrand's 'No 1' schematic eye, involving a high-index core and a lower index cortex, and the idea has been progressively extended through a variety of models where the lens is approximated by a finite number of 'shells' of differing index to include smooth gradients of refractive index (e.g. Gullstrand, 1924; Lotmar, 1971; Pomarantzeff et al., 1972, 1983; Watkins, 1972; Blaker, 1980). There is little doubt that, with progressive refinement in the light of better experimental data, these models will make valuable contributions to our understanding of the off-axis optical performance of the eye.

Ocular Ametropia

Real eyes are not standardized in their dimensions but, like the rest of the body, can be expected to show variation in at least some of their individual parameters. The spread of the values of the radius of curvature of the cornea has already been remarked upon and a similar scatter is found in other ocular dimensions (see below). It is reasonable to ask, then, whether our eye models can indicate the magnitude of change in any parameter which is likely to lead to a clinically-significant refractive error.

Table 1.1 *Parameters of some paraxial models of the human eye.*

		Schematic eye (Le Grand and El Hage, 1980)	Simplified schematic eye (Emsley, 1952)	Reduced schematic eye (Emsley, 1952)
Radii of surfaces (mm)	Anterior cornea	7.80	7.8	5.55
	Posterior cornea	6.50	—	—
	Anterior lens	10.20	10.0	—
	Posterior lens	−6.00	−6.00	—
Distances from anterior cornea (mm)	Posterior cornea	0.55	—	—
	Anterior lens	3.60	3.60	—
	Posterior lens	7.60	7.20	—
	Retina	24.20	23.90	22.22
Refractive indices	Cornea	1.3771	—	
	Aqueous humour	1.3374	1.333	4/3
	Lens	1.4200	1.416	
	Vitreous humour	1.3360	1.333	

Consider, for simplicity, Emsley's reduced eye model (Fig. 1.5c). It is evident that a refractive error KD can arise if the power of the dioptric elements $F_E = (n'-1)/r$ (where n' is the refractive index of the eye media and r is radius of curvature of the refracting surface) does not match the dioptric length of the eye $K' = n'/k'$ (where k' is the axial length of the eye), i.e.

$$K = K' - F_E = (n'/k') - [(n'-1)/r] \qquad (1.3)$$

To determine the change in K which arises as a result of change in any one of the parameters we may write:

$$\partial K = \frac{\partial K}{\partial k'}.\delta k' + \frac{\partial K}{\partial n'}.\delta n' + \frac{\partial k}{\partial r}.\delta r$$

or

$$\delta K = -(n'/k'^2)\delta k' + ([1/k'] - [1/r])\delta n' + ([n'-1]/r^2)\delta r . \qquad (1.4)$$

Substituting the standard values for k', n' and r from Table 1.1 we find that the changes required to produce 1 D of myopia are approximately $\delta k' = +0.37$ mm, $\delta n' = +0.0074$, $\delta r = -0.093$ mm: the signs of these changes are reversed for 1 D of hypermetropia.

Obviously considerable approximation is involved in determining these values. A similar analysis could, of course, be extended to more sophisticated eye models but nevertheless the finding that changes of about 1/3 mm in axial length and 1/10 mm in corneal radius of curvature give refractive changes of about 1 D constitutes a useful approximate rule-of-thumb. Refractive index values vary very little from one eye to another (Le Grand and El Hage, 1980) except for the lens in some pathological conditions (see Howe, Chapter 3).

Distribution of Parameters in Real Eyes

Surface curvatures, component separations and axial lengths show considerable variation. Various studies have shown that each follows an approximately normal distribution.(e.g. Stenstrom, 1946; Sorsby *et al.*, 1957, 1981) (Fig. 1.6(a) to (d)). The spread of values is such that, in the light of the previous section, a high prevalance of large refractive errors might naively be expected to occur in the general population if the overall refraction of the eye arose from random combinations of parameters having the distributions shown. In practice this is not the case
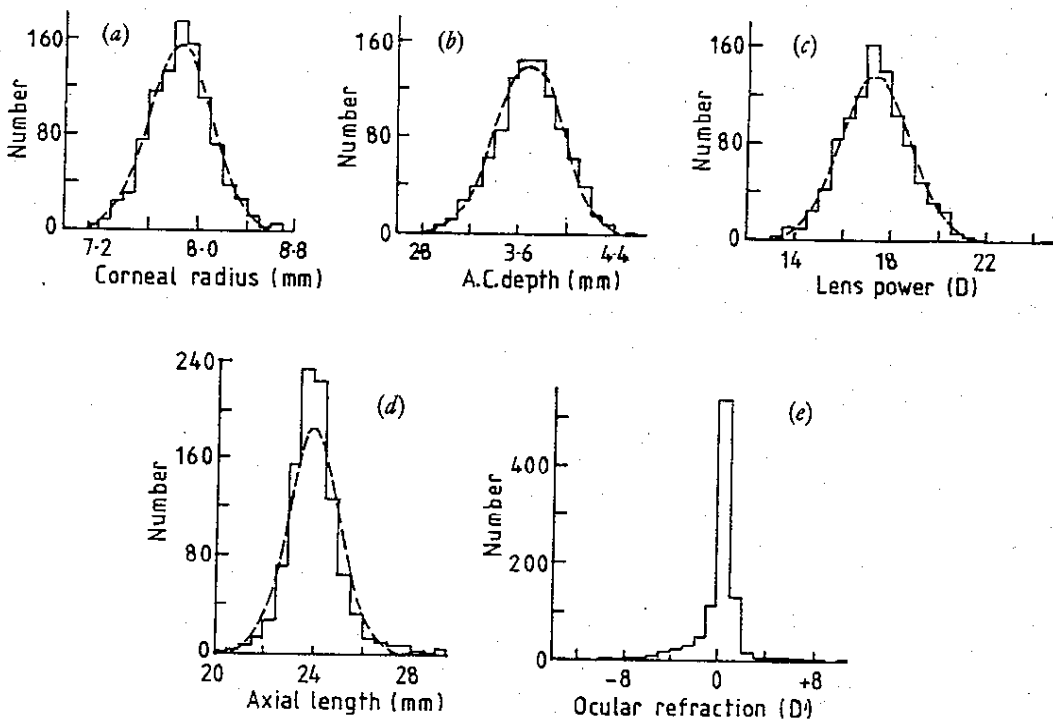

Fig. 1.6 *Distribution of some ocular parameters and of refractive error (after Stenstrom, 1946). In each case, the dashed curve represents the corresponding normal distribution. Note that while the individual parameters are distributed approximately normally, the refractive errors are strongly peaked to near-emmetropia. (a) Corneal radius of curvature. (b) Anterior chamber depth. (c) Lens power. (d) Axial length. (e) Refractive error.*

aets and Berry (1968) have made further measurements on whole eye and other preparations. Such measurements are difficult to make and the results are influenced by the proportion of scattered light included (see also Henson, Chapter 17). Nevertheless, the general form of the results is reasonably consistent, showing a rapid rise in transmittance at around 400 nm followed by high values through the visible and near infra-red: the transmittance
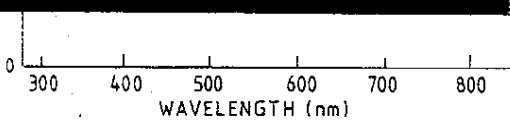

Fig. 1.8 *Transmittance of normal, human, crystalline lenses of different ages. Although the measurements may not be entirely representative of in vivo performance, the relative transmittanc are likely to be realistic (after Lerman, 1980).*

d absorption
Average data
1.7, together
or the average
ety of earlier
s to be largely
lividuals who
vity at short
ldock, 1972).
vith age (e.g.
also Fig. 1.8)

)0   1400

human eye.
npiled by
authors.
23–78 years,
bsorption
· of macular

cular pigment
for light

—— 0·5

—— 8

—— 25
——54
—— 47

——82

800

ne lenses of
he entirely
·ansmittances

due partly to the increasing pathlength through the progressively thicker cortex of the lens and partly to increased pigment deposition and, perhaps, scattering in the nucleus (Said and Weale, 1959; Mellerio, 1971, 1987). The energy of some of the short-wave radiation that is absorbed may appear as fluorescence at longer wavelengths (e.g. Lerman, 1972; Satoh *et al.*, 1973; Spector *et al.*, 1975; Bando *et al.*, 1976; Lerman, 1980), although the effects of this light in reducing retinal image contrast are believed to be small (Boynton and Clarke, 1964). Some investigators have suggested that the strong and increasing short-wave absorption of the lens is helpful in protecting the ageing retina from the 'blue light hazard' (e.g. Sliney and Wolborsht, 1980; Marshall, 1985) and that, for this reason, intraocular implant lenses should have similar absorption characteristics (see Ridgway, Chapter 7).

It is worth noting that, because of the bi-convex form of the crystalline lens, lenticular absorption may have some influence on measurements of the Stiles-Crawford effect (Stiles and Crawford, 1933, see Enoch and Lakshminrayanan, Chapter 12) at shorter wavelengths: the pathlength through the lens is substantially shorter for rays near the edge of the pupil, leading to a smaller absorption loss and hence a reduction in the measured Stiles–Crawford effect over that occurring at the receptor level (Weale, 1961; Vos and Van Os, 1975). As already noted, it is not strictly true to assume that the attenuation per unit path in the lens is position-independent, since regions of higher scattering and absorption obviously exist (e.g. Brown, 1973; Sasaki *et al.*, 1980; Mellerio, 1987).

The light is further attenuated by the retina itself before it reaches the receptor outer segments. Not only is there some loss due to scattering and reflection, contributing some 30% of the total light scatter within the eye (Vos and Bouman, 1964) but there are further screening effects from the retinal blood vessels (Kishto, 1970). In the foveal

region, the macular pigment, extending over the central few degrees of the retina (Stanworth and Naylor, 1955; Kilbride *et al.*, 1989) and lying anterior to the receptor outer segments (Snodderly *et al.*, 1984) absorbs heavily at short wavelengths (Fig. 1.7). It has been argued that this pigment plays a useful role in reducing the blurring effects of the longitudinal chromatic aberration of the eye, and hence improving acuity (e.g. Reading and Weale, 1974). However the amount of pigment varies widely between individuals and may also be age-dependent (Bornstein, 1977): as yet there have been no investigations to explore the possibility of any correlation between the amount of pigment and the acuity achieved.

Since the cornea, lens and the oriented molecules of macular pigment all show birefringence, the polarization characteristics of light entering the eye are modified before the receptor outer segments are reached: likewise, the polarization of light reflected back out of the eye undergoes further modification. Bour discusses these effects in detail in Chapter 13.

Ocular Radiometry and Retinal Illumination

If we confine ourselves to uniform object fields subtending at least 1° at the eye so that blurring due to aberration or defocus has negligible effect, the retinal image would also be expected to be uniform across its area, except at the edges, provided that the field angles were moderate. Wyszecki and Stiles (1967) show that the retinal irradiance, or internal stimulus, in a wavelength interval $\delta\lambda$, corresponding to an external stimulus of spectral radiance $L_{e\lambda}(\theta,\phi)$ W per unit wavelength interval per cm² per unit
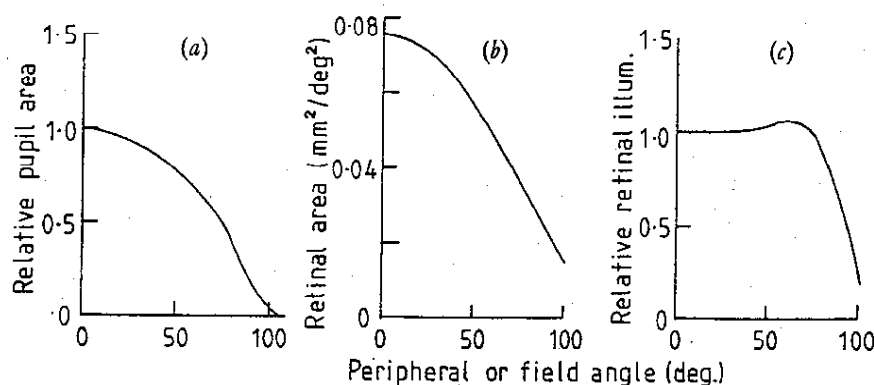
Fig. 1.9 *Influence of peripheral (field) angle on area of entrance pupil, retinal image area and retinal illuminance for an object field of uniform luminance (Ganzfield). (a) Relative area of entrance pupil (after Jay, 1962). (b) Area of retinal image corresponding to one square degree of field (after Dresdo and Fowler, 1974). (c) Curve (a) divided by curve (b) and normalized to unity on the visual axis i.e. relative retinal illuminance for a uniform object field (after Charman, 1983, 1989).*

solid angle of emission, in a direction with respect to the eye given by the angular coordinates (θ, ϕ) is then given by:

$$\frac{L_{e\lambda}(\theta,\phi).\delta\lambda.p(\theta,\phi).t(\theta,\phi,\lambda)}{m(\theta,\phi\lambda)} \qquad (1.5)$$

where $p(\theta, \phi)$ cm² is the apparent area of the pupil as seen from the direction (θ, ϕ); $t(\theta, \phi, \lambda)$ is the fraction of the incident radiant flux which is transmitted through the eye; and $m(\theta, \phi, \lambda)$ is an area magnification factor (cm²) relating the area of the retinal image to the angular subtense of the stimulus at the eye, which will vary somewhat with the parameters of the individual eye.

Use of this equation close to the axis of the eye is straightforward. In the peripheral field the situation becomes more complicated. The pupil then appears as an ellipse of increasing eccentricity and reduced area and, due to the retina lying on a spherical surface, the distance of the retina from the exit pupil of the eye and the retinal area corresponding to an object of constant angular subtense change markedly. Remarkably, theoretical calculations (Fitzke,1981; Bedell and Katz, 1982; Charman, 1983, 1989; Kooijman, 1983; Pflibsen *et al.*, 1988) show that these various factors combine so that an object field of constant luminance (i.e. a Ganzfield) results in a retinal illuminance which is almost constant with peripheral angle (Fig. 1.9). This theoretical result has been broadly confirmed by practical measurements on both rabbit and human eyes (Kooijman and Witmer, 1986) showing that, from a photometric point of view, the design of the human eye as a wide-angle optical system is remarkably effective.

A full discussion of the photometric aspects of point and extended sources is given by Wright (1949).

Retinal Image Quality

The aberrations of the eye and their effects on retinal image quality have long been of interest and most of the major figures of optics and vision have contributed to the advancement of our knowledge of this topic, from Newton onwards. The retinal image is inevitably degraded by pupil–diameter and wavelength–dependent diffraction and the regular, monochromatic, Seidel aberrations that any system of centred, spherical surfaces is prey to, i.e. spherical aberration, coma, oblique astigmatism, field curvature and distortion. It also has the defects to be expected from a biological system with various tilts, decentrations and other asymmetries.

Specification of Image Quality

A variety of methods exist for specifying the quality of the image formed by any optical system. Perhaps the most obvious is in terms of the image that it forms of a point object, i.e. the point-spread function or PSF, $P(x,y)$, where x,y are appropriate coordinates in the image surface. In general $P(x,y)$, which varies with field position, wavelength and focus, lacks radial symmetry, making it difficult to describe simply. Moreover, the illuminance in the outer parts of the image will be very low, making it difficult to measure experimentally, even though an important fraction of the light flux is contained in this part of the image. From the measurement point of view, there are advantages in determining the orientation-dependent line-spread function or LSF, $L(x)$, which, since a line is a series of points, is simply the line integral of the point-spread function, i.e.

$$L(x) = \int\limits_{-\infty}^{+\infty} P(x,y)\,dy \qquad (1.6)$$

It is assumed here that, as is normally the case with the eye, different points in the object plane are incoherently illuminated and that the aberrations can be considered as constant over the relevant areas of field. Although, like the PSF, the LSF gives a good intuitive indication of the likely acuity for objects of similar form, because much of the image flux is contained in the outer parts of the LSF it is often difficult to recognize the effects that this spread of light will have when extended objects are imaged. Fortunately, it is possible to move directly from the LSF to a description of imaging performance for extended objects in the form of cosinusoidal gratings of varying spatial frequency, R. Any optical system images such an object as a similar grating but, usually, of reduced contrast or modulation (Selwyn, 1948). The images may also be shifted slightly perpendicular to the length of the grating bars, i.e there may be a spatial phase shift. To move from the LSF to the optical transfer function (OTF), $O(R)$, which describes these modulation and phase changes as a function of R, we use the Fourier transform relationship:

$$O(R) = \int\limits_{-\infty}^{+\infty} L(x)e^{-2\pi iRx}dx = T(R)e^{i\phi(R)} \qquad (1.7)$$

where $T(R)$ and $\phi(R)$ are the modulation and phase transfer functions respectively (MTF and PTF). The PTF becomes zero if the corresponding LSF is symmetrical.

A further alternative description of the optical performance of the eye is in terms of the wavefront aberration. As is well known, the wavefronts in any system are surfaces which are orthogonal to the corresponding ray pencils. Hence, if in terms of geometrical optics an optical system forms a perfect point image, all the imaging rays intersect at that point or, alternatively, all the imaging wavefronts are spherical. If aberration is present there is no longer a point focus and hence the wavefronts are no longer spheri-

s of a point
SF, $P(x,y)$,
image sur-
ld position,
, making it
tminance in
, making it
though an
in this part
view, there
-dependent
:e a line is a
the point-

(1.6)

ith the eye,
rently illu-
sidered as
zh, like the
if the likely
ich of the
: LSF it is
spread of
ed. Fortu-
LSF to a
ed objects
:patial fre-
object as a
or modu-
:e shifted
g bars, i.e
i the LSF
?), which
as a func-
ship:

(1.7)

ase trans-
he PTF
tetrical.
perform-
ation. As
: surfaces
pencils.
al system
intersect
avefronts
longer a
r spheri-

cal. Thus we can usefully describe the aberration in terms of the distances by which different portions of the wavefront in the exit pupil depart from the ideal sphere. This is called the wavefront aberration: each of the Seidel aberrations corresponds to a characteristic form of wavefront aberration (Hopkins, 1950; Born and Wolf, 1975; Smith, 1978). Rayleigh made use of this description when he introduced his quarter-wavelength criterion. This effectively states that the aberration in any optical system has negligible effect on the image if the wavefront aberration is everywhere less than a quarter of a wavelength (path differences of less than this amount mean that the wave disturbances add up almost exactly in phase at the image point). The OTF can be deduced mathematically from the wave aberration (e.g. Born and Wolf, 1975; Hopkins, 1962; Smith, 1978), so that it is possible to compute each of the descriptors of imaging performance from any other (although the LSF and OTF must be known for all orientations).

The Aberration-Free Eye

If an eye is free of aberration, the wavefront aberration is always zero. The only factor contributing the blur of the in-focus retinal image is the diffraction arising from the finite wavelength of light, λ, and the finite diameter, p, of the eye pupil: for this reason an aberration-free eye is often called a diffraction-limited eye. The point-spread function is then an Airy diffraction pattern and the LSF and MTF take similarly standard forms (the PTF will be zero since the PSF and LSF are symmetrical). Blur increases on either side of focus (Lommel, 1884; Linfoot and Wolf, 1956; Born and Wolf, 1975). For the in-focus case, angular resolution for two object points by the Rayleigh criterion is given by:

$$\theta_{min} = (1.22\lambda)/p \text{ rad} \qquad (1.8)$$

Fry (1955) has given a clear description of the way in which the PSF of an aberration-free eye changes as a function of focus: these results have important implications in indicating the precision required for the accommodation mechanism and during refractive procedures. The corresponding through-focus MTFs have been illustrated for various wavelengths and pupil diameters by several authors (e.g. Westheimer, 1964, 1966; Campbell and Green, 1965; Campbell and Gubisch, 1967; Charman and Tucker, 1977; Charman and Jennings, 1976a; Charman and Heron, 1979; Charman, 1983). Much of this data is based on tables given by Levi (1974). Smith (1982) has given detailed consideration to the question as to the general applicability of physical and geometrical optical descriptions to imagery of the in and out-of-focus, diffraction-limited eye.

The Regular Monochromatic Aberrations of the Eye

As Guidarelli (1972) has noted, the eye crudely approximates to a homocentric system, since all the surfaces, including the retina, have centres of curvature lying near the centre of the aperture stop. In a truly homocentric system, the chief ray, passing through the centre of the aperture stop, is always an optical axis, so that no off-axis aberrations occur. This is by no means exactly true for the eye but nevertheless, as Young (1801) noted many years ago, the retinal curvature does very closely match the curvatures of the eye's image surfaces, so that off-axis aberrations are quite well controlled.

In practical terms, we would expect that, if the eye were indeed a regular optical system, the only significant ocular aberration on the optical axis would be spherical aberration. The other regular aberrations would become increasingly important away from the optical axis, although it might be anticipated that distortion, a field-angle dependent magnification change, could potentially be corrected for by the higher visual centres. The latter is, indeed, almost certainly the case and it is of interest that similar compensation or adaptation can be observed to occur for the distortions associated with correcting spectacle lenses (Ogle, 1950).

Most investigators attempting to determine the nature of individual aberrations have, in fact, concentrated on spherical aberration and oblique astigmatism and have often assumed that it is adequate to consider that the optical and visual axes of the eye coincide (i.e. angle α is zero), so that any aberration measured for foveal vision should be pure spherical aberration. There is, of course, particular interest in determining the aberration associated with the foveal image in that it is in the fovea that the neural network makes the greatest demands on optical image quality.

Spherical Aberration

Earlier work on spherical aberration has been reviewed by several authors (Koomen *et al.*, 1949; Rosenblum and Christensen, 1976; Charman, 1983). As spherical aberration involves a regular, radially symmetric, change in power, most measurements have been made using apertures, often annular, to isolate different small regions of the entrance pupil and hence measure the associated power by appropriate subjective or objective techniques.

Results, as found by a variety of different investigators, are shown in Fig. 1.10; although the individual data vary, they suggest that spherical aberration is generally of the order of 1 D at the edge of a 4 mm diameter pupil. Also included in Fig. 1.10 is the spherical aberration for a schematic eye with spherical surfaces, together with the curve showing the amount of primary, positive, under-corrected
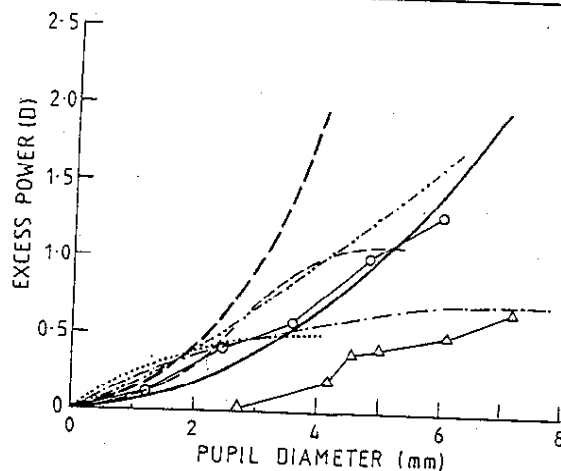
Fig. 1.10 *Spherical aberration of the human eye as found by various investigators: ——·—— Koomen et al. (1949); ———— Françon (1951); ...... Ivanoff (1953); -O-O-O Schober et al. (1958); —·— Jenkins (1963a); -△-△-△ Millodot and Sivak, 1979). The heavy continuous curve represents the amount of under-corrected, primary, spherical aberration which van Meeteren (1974) assumed was typical and the heavy dashed curve is for a schematic eye with spherical surfaces (Jenkins, 1963a).*

spherical aberration which Van Meeteren (1974) suggested was representative for real eyes. For the latter, the excess power of a zone of radius r is given by $\Delta P = 4Ar^2$ and the wavefront aberration by $W(r) = Ar^4$: van Meeteren's assumed value of A is $4 \times 10^4 \, \mathrm{m}^{-3}$. Not surprisingly, the amount of aberration shows considerable variation between individual eyes and is accommodation-dependent, there apparently being minimal aberration when the eye is accommodating for an object at about 0.7 m from the eye (Koomen et al., 1949; Ivanoff, 1947, 1953, 1956; Jenkins, 1963a; Berny, 1969), a distance which, intriguingly, corresponds to the equilibrium or 'resting' state of the accommodation system (see Ciuffreda, Chapter 11).

It is clear from Fig. 1.10 that the eye has less spherical aberration than would be expected from an eye with spherical surfaces and that the aberration is almost negligible for pupil diameters less than 2–3 mm. This is largely attributable to the asphericity of the eye's surfaces, particularly the cornea, and to the index gradients in the lens. There is some disagreement as to whether the cornea and lens act in combination to minimize spherical aberration (El Hage and Berny, 1973) or whether they are each separately minimized (Millodot and Sivak, 1979).

Fry (1955) has computed some through-focus, retinal, point-spread functions, using the spherical aberration data of Ivanoff (1947). Berny (1969) has used wavefront aberration data to calculate the effect of spherical aberration on

the ocular MTF and has demonstrated the loss of modulation transfer at intermediate spatial frequencies that can occur in comparison with a diffraction-limited eye. Similar MTF measurements and calculations have been presented by Charman and Jennings (1976a) and Charman et al. (1978).

As will be discussed further below in the context of the overall wavefront aberration, many of the above studies emphasize that, when an appropriate measurement technique is used, it is found that the power variation in the pupil of any individual eye rarely has true radial symmetry. Thus the concept of the eye displaying only pure spherical aberration near its axis has only limited validity. Even when eyes suffer from supposedly typical amounts of spherical aberration, the subjective refraction varies very little with pupil size (Koomen et al., 1949, 1951; Charman et al., 1978). This is presumably due to irregular aberration, the Stiles-Crawford effect (which reduces the contribution of the outer zones of a dilated pupil under photopic conditions) and to the fact the spherical aberration has only a small effect on optimal focus for periodic detail near the cut-off frequency for the eye ($\sim 30 \, \mathrm{c \, deg}^{-1}$).

Oblique Astigmatism

In practice, measurements of ocular oblique astigmatism inevitably involve the determination of the positions of the astigmatic focal lines with respect to the retina (which may itself lack rotational symmetry about the axis), rather than the true oblique astigmatism of the dioptrics. Most attention has been concentrated on the horizontal (nasal-temporal) meridian of the eye, using retinoscopy (skiascopy) or objective optometers (Ferree et al., 1931, 1932; Rempt et al., 1971; Hoogerheide et al., 1971; Millodot and Lamont, 1974; Millodot, 1981).

Representative data are shown in Fig. 1.11. The observed amount of oblique astigmatism is lower than that calculated for eye models with spherical surfaces: Dunne et al. (1987) have been able to achieve good agreement between the measured values and those for a model eye with aspheric surfaces, including that of the retina. It is of interest that the experimental data, plotted with respect to the visual axis, appear to be symmetrical about a nasally displaced point, presumably due to angle α. Variations in the relationship between the radial and tangential refractions and the central refractions for emmetropic, myopic and hypermetropic groups (Millodot, 1981) seem to be explicable in terms of variations in the axial lengths of the eyes concerned (Charman and Jennings, 1982). The amount of aberration measured increases slightly with the level of accommodation being exercised, presumably due to the shape changes in the crystalline lens (Smith et al., 1988).
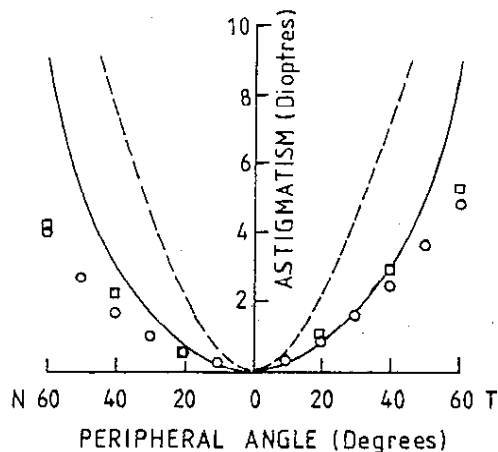
of modula-
ies that can
l eye. Simi-
e been pre-
Charman *et*

ntext of the
ove studies
:ment tech-
ation in the
·adial sym-
g only pure
ed validity.
al amounts
:tion varies
949, 1951;
to irregular
·educes the
·upil under
: spherical
l focus for
or the eye

stigmatism
tions of the
which may
rather than
/ost atten-
tal (nasal-
:opy (skia-
931, 1932;
illodot and

1.11. The
·r than that
es: Dunne
agreement
model eye
na. It is of
· respect to
t a nasally
riations in
tangential
:metropic,
981) seem
lengths of
982). The
y with the
nably due
iith *et al.*,

Fig. 1.11 *The oblique astigmatism of the human eye. N and T represent cases where the image falls on the nasal and temporal retina respectively. Squares are mean experimental data from Lotmar and Lotmar (1974) based on the original data of Rempt et al. (1971) for 726 eyes, and circles are data from Millodot (1981) for 62 eyes. The broken and continuous curves are theoretical predictions from the schematic eye models of Le Grand (1967) and Lotmar (1971), using spherical and aspheric surfaces respectively.*

The average values shown in Fig. 1.11 conceal a range of variations in the amount, sign and symmetry of the oblique astigmatism that is found in individual eyes (e.g. Ferree *et al.*, 1931, 1932; Ferree and Rand, 1932, 1933; Rempt *et al.*, 1971; Hoogerheide *et al.*, 1971). Using calculations with a schematic eye, Barnes *et al.* (1987) have plausibly argued that these individual differences are probably associated with small translations and rotations of the surfaces of the eye with respect to the axis of symmetry.

In general, correction of oblique astigmatism brings about, at best, only very modest improvements in visual performance for peripheral tasks (see, e.g. the review by Charman, 1983) due presumably to the limits to performance being set primarily by neural rather than optical factors. Wang *et al.* (1983) have, however, demonstrated that the sharpness of fundus camera photographs of the peripheral retina can be usefully improved by the use of cylindrical correcting lenses.

Other Individual Monochromatic Aberrations

No attempt appears to have been made to study possible ocular coma in isolation. A variety of small-scale, local refractive irregularities may modestly degrade retinal image quality: these irregularities can best be detected by the entoptic phenomena that they produce (see Palmer, Chapter 15).

Overall Monochromatic Wavefront Aberration

Determination of the overall wavefront aberration has the advantage that, since each individual Seidel aberration produces a characteristic form of wavefront distortion (e.g. Hopkins, 1950; Born and Wolf, 1975; Smith, 1978), in principle it is possible to deduce from the measured wavefront both the contributions of the individual aberrations and their combined effects.

Up to the present time, wavefront aberration has only been measured on the visual axis. The earliest study was that of Smirnov (1962), who used a very laborious coincidence technique in conjunction with a moveable 0.4 mm pupil to map the wavefront aberration in a square mesh across the full pupils of 10 eyes: a substantial period of data analysis was also required. Somewhat similar measurements of the local dioptric power of small regions of the pupil were made by Van den Brink (1962) and, very recently, Howland and Buettner (1989) have shown that these may also be used to deduce the wavefront aberration.

Objective methods were pioneered by Berny and Slansky (1969), who adapted the powerful Foucault knife edge method to secure the necessary information in two flash photographs, taken with two mutually perpendicular orientations for the knife edge. At that time, however, analysis of the photographs took several months, so that only a single eye was studied. Walsh *et al.* (1984) and Walsh and Charman (1985) modified the ingenious subjective aberroscope technique of Howland and Howland (1976, 1977) to allow objective aberroscope recording. The method involves viewing a point source through a grid which is sandwiched between two crossed cylinder lenses: each aberration produces a characteristic distortion in the retinal shadow image of the grid. Lastly, the wavefront aberration can be inferred from measurements of the retinal point-spread function (Artal *et al.*, 1988a, 1988b).

Although there are differences in the form of aberration between individuals, which are not surprising in view of the known variation in such factors as corneal contour, and in the resolution in the detail of the wavefront which can be achieved by each method, all these studies agree in demonstrating that the wavefront aberration rarely displays the radial symmetry that would be expected if only spherical aberration was present. Some examples are shown in Fig. 1.12. In general, coma-like aberration often seems to be more important in the individual eye than spherical aberration. Nevertheless, if the wavefront errors of many eyes are averaged then the aberration shows much greater radial symmetry (Charman and Walsh, 1985) and, indeed, the radial change in power that can be deduced from this average wavefront aberration corresponds closely to the classically-measured amounts of under-
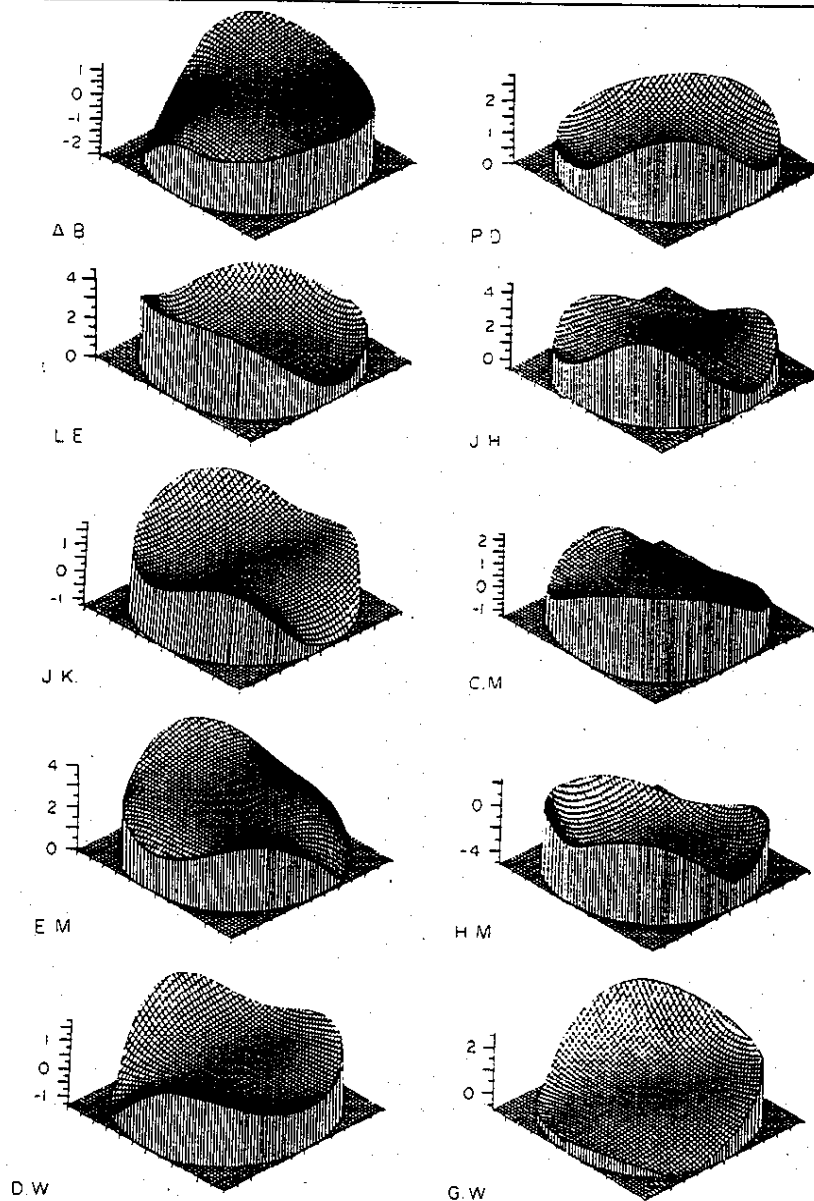
Fig. 1.12 *Pseudo-isometric representation of the wavefront aberration in the right eyes of 10 subjects. The vertical scale represents the wavefront aberration in µm, with the positive direction being towards the subject. Zero wavefront aberration would correspond to a flat surface in this form of plot. The near left and near right sides of each diagram represent the left and bottom of the subject's pupil as seen by an observer facing the subject (after Walsh and Charman, 1985).*

corrected spherical aberration that are shown in Fig. 1.10. The wavefront aberration of the individual eye is generally less than a quarter wavelength within the central 2–3 mm diameter of the pupil.

Chromatic Aberration

The longitudinal chromatic aberration of the eye has been a subject of study since the time of Newton (Charman,

1985). Typical results of modern subjective measurements are shown in Fig. 1.13. Objective studies produce broadly compatible data (Lau *et al.*, 1955; Ronchi and Millodot, 1974; Charman and Jennings, 1976b). As can be seen, the aberration typically amounts to about 2 D across the full visible spectrum. It is at first sight surprising that correction of this aberration with a suitable achromatizing lens produces little effect upon acuity for white light (Hartridge, 1947) but Campbell and Gubisch (1967) point
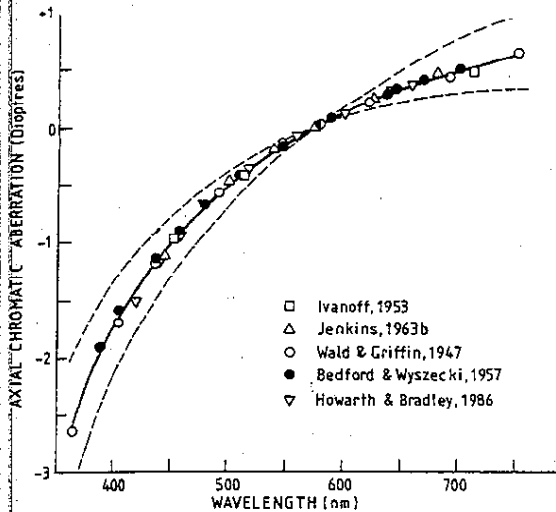
Fig. 1.13 *Average axial, or longitudinal chromatic aberration of the eye as measured by several authors. In each case, the mean data have been displaced so that the chromatic aberration is zero at 578 nm. ○ Wald and Griffin, 1947 (14 eyes); □ Ivanoff, 1953 (11 eyes); ● Bedford and Wyszecki, 1957 (12 eyes); △ Jenkins, 1963b, (32 eyes); ▽ Howarth and Bradley, 1986 (20 eyes). The dashed curve represents the overall range of individual observers, as found by Bedford and Wyszecki.*

out that this is because the major effects of the aberration are to degrade modulation transfer at intermediate, rather than high, spatial frequencies. Some authors (Polack, 1923; Ivanoff, 1953; Le Grand, 1967; Millodot and Sivak, 1973; Bobier and Sivak, 1978; Sivak, 1982) have suggested that the accommodation system may make use of the range of focus conferred by longitudinal chromatic aberration to 'spare the accommodation', i.e. to minimize accommodative effort: on the other hand, there is little evidence for markedly different steady-state accommodation characteristics in white and monochromatic light (Charman and Tucker, 1978), nor is depth-of-focus as conventionally measured markedly greater in white light (von Bahr, 1952; Oshima, 1958; Ogle, 1960; Legge et al., 1987).

There have been suggestions in recent years that the longitudinal chromatic aberration may be age-dependent, possibly due to index changes in the various ocular media (Millodot, 1976; Millodot and Newton, 1976; Mordi and Adrian, 1985) but other recent studies (Ware, 1982; Howarth et al., 1988) deny this and as yet an age dependence cannot be regarded as being firmly established.

Transverse, or lateral chromatic aberration must also occur, i.e. the retinal images of a given object should be slightly different in size at differing wavelengths. Calculations (Van Meeteren, 1974; Howarth, 1984; Thibos, 1987) suggest that the aberration plays a significant role in reducing retinal image quality in white light. While the

effects are of increasing significance in the periphery, they may also occur at the fovea, due to the misalignment between the visual and optical axes of the eye (angle α). The effects are orientation dependent. The subjective measurements of Ogboso and Bedell (1987) appear to give values which are somewhat lower than the theoretical predictions, the transverse chromatic aberration between wavelengths of about 435 and 572 nm remaining less than 10′ for field angles up to 40°. At the fovea, the measured transverse chromatic aberration is also less than would be expected for an angle α of 5° (Simonet and Campbell, 1990; Thibos et al., 1990).

Intraocular Scatter and Related Effects

In all the foregoing discussion of aberrations, it has been assumed that the refractive indices of the media are either constant or smoothly varying and that the optical surfaces themselves are also smooth on the scale of the wavelength of light. In practice, a variety of regular and irregular small-scale inhomogeneities exist and these contribute to further light loss and a spread of light in the retinal image. The qualitative aspects of these effects exercised considerable fascination for earlier authors: the associated entoptic phenomena are described in Palmer (Chapter 15).

Quantitative measurements of the forward-scattered light were pioneered by Holladay (1926) and Stiles (1929), who used a technique based on the masking effect of a glare source. They expressed their results in terms of an equivalent veiling luminance, L_{eq} cd m^{-2}, that would produce the same masking effect as a glare source giving an illuminance E at the eye, as a function of the angular distance, α degrees, between the glare source and fixation. They found the approximate relationship:

$$L_{eq} = \frac{10E}{\alpha^2} \cdot (4° < \alpha < 100°) \qquad (1.9)$$

Later work, well summarized by Vos et al. (1976) led to the modification of this formula to take account of effects closer to fixation:

$$L_{eq} = \frac{29E}{(\alpha + 0.13)^{2.8}} \quad (0.15° < \alpha < 8°) \qquad (1.10)$$

These equations apply to young adult eyes. The magnitude of the scattering undoubtedly increases throughout life, by a factor of at least 2–3 times, although the same angular dependence is retained (Allen and Vos, 1967; Hemenger, 1984). Roughly a quarter of the effective stray light comes from the cornea (Vos and Boogaard, 1963; Boynton and Clarke, 1964) and a further quarter from reflections off the retina (Vos, 1963; Vos and Bouman, 1964): the rest comes almost entirely from the lens, there being little contribution from the aqueous or vitreous. A theoretical analysis of the contribution of lens fibres to

intraocular scatter has been presented by Hemenger (1988). Attempts have recently been made to develop a clinical method for assessment of the glare effect caused by scattering, by measuring the impairment in the contrast sensitivity function in the presence of a glare source (Paulsson and Sjostrand, 1980; Abrahamsson and Sjostrand, 1986): the method shows particular promise as an aid to following changes in cataractous eyes.

Final Retinal Image Quality

Estimates of the quality of the final retinal image can be made in three main ways: by calculation from the wave aberration; by a psychophysical method; and by direct measurement of the light distribution on the retina using a double-pass ophthalmoscopic technique.

Calculation from the Wavefront Aberration

Calculation of the optical transfer function (OTF) can be carried out by autocorrelation of the complex pupil function with its complex conjugate, using numerical methods originally devised by Hopkins (1962). The pupil function gives the variation in amplitude and phase across the exit pupil of the system. The local phase can be deduced directly from the wavefront aberration (each wavelength of aberration corresponds to 2π of phase) and it may either be assumed that the amplitude across the pupil is uniform or, if desired, pupil-weighting functions can be introduced in an attempt to take account of the effects of pupil-dependent lenticular absorption or the Stiles-Crawford effect (Stiles-Crawford apodization, see, e.g. Metcalf, 1965). The point-spread (PSF) and line-spread (LSF) functions may similarly be calculated from the wavefront aberration (e.g. Hopkins, 1962; Born and Wolf, 1975; Smith, 1978).

This approach was used by Berny (1969) to estimate ocular MTFs from the wavefront aberration as deduced from Foucault's knife-edge measurements, and by Berny and Slansky (1969) to compare the retinal PSF with the ideal Airy disc: these latter authors concluded that the eye was essentially diffraction-limited for pupil diameters up to 2 mm. Van Meeteren (1974) took a more indirect approach, in that he estimated typical values for wavefront aberration from published measurements of the individual Seidel aberrations and other sources of image degradation. Interestingly, he found that making allowance for the Stiles-Crawford effect had little influence on the overall MTF for the natural pupil diameters (<5 mm) which occur under photopic conditions: in white light, chromatic aberration appeared to be the most important single aberration for typical photopic pupils.

The subjective aberroscope results of Howland and Howland (1976, 1977) are of particular interest in that a large sample (55) of eyes was studied. Hence it was pos-

sible to obtain some idea of the spread in performance between individual eyes. Fig. 1.14(a) shows the rank ordered, calculated MTFs for monochromatic light of wavelength 555 nm and a 5 mm diameter pupil. Although Howland and Howland's technique demanded that subjects sketch the image of the aberroscope grid that the saw, very similar results were obtained when using the objective variant of the method (Walsh et al., 1984, Walsh and Charman, 1985), as shown in Fig. 1.14(b). It will be noted that, for these 5 mm pupils, the performance of all eyes falls well below that set by the diffraction limit. The calculated phase transfer functions (PTFs) suggest that pupil dependent optical phase shifts may have some role in limiting the visual system's ability to discriminate spatial phase (Charman and Walsh, 1985).
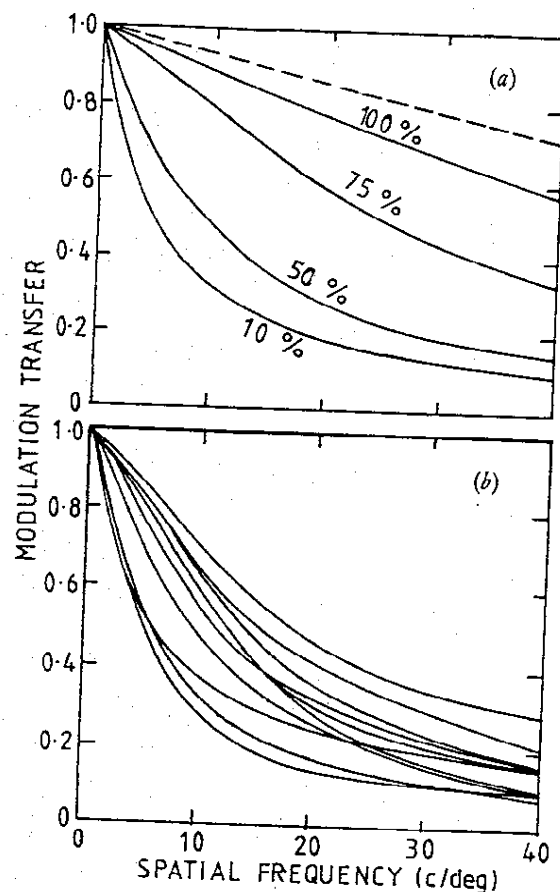

Fig. 1.14 *Ocular MTFs for an eye with a 5 mm pupil as calculated from wavefront data. (a) Rank-ordered MTFs for a group of 55 eyes and light of wavelength 555 nm. The 100% curve represents the best eye of the set (after Howland and Howland, 1977), the dashed curve being the MTF for a diffraction-limited eye. (b) MTFs for a sample of 10 eyes at a wavelength of 590 nm and a pupil diameter of 5 mm (after Walsh and Charman, 1985).*

Calculations of retinal image quality from wavefront data suffer from the obvious disadvantage that no allowance can be made for the deleterious effects of light scatter in the media. Hence they may lead to too optimistic an estimate of retinal image quality. As already noted, it is possible to weight the pupil function for lenticular absorption or the Stiles-Crawford effect, but such weighting necessarily rests on assumptions rather than direct measurement.

Psychophysical Method

The principle of this method is straightforward. If the subject views an external grating of spatial frequency R and modulation $M_o(R)$, the modulation in the retinal image must be $M_o.T(R)$, where $T(R)$ is the modulation transfer of the ocular dioptrics at this spatial frequency. Suppose now we steadily reduce the modulation of the grating until it appears to the observer to be just at threshold. Then the threshold modulation on the retina, $M_{it}(R)$, will be given by:

$$M_{it}(R) = M_{ot}(R). T(R) \qquad (1.11)$$

where $M_{ot}(R)$ is the measured modulation of the external grating at threshold. The reciprocal of $M_{ot}(R)$ is obviously the contrast sensitivity as conventionally defined, and measurement of $M_{ot}(R)$ as a function of R simply corresponds to the procedure used to establish the contrast sensitivity function.

Clearly, $M_{it}(R)$ corresponds to the threshold for the retina/brain portion of the visual system and, if it can be established independently, then $T(R)$ can be deduced from $M_{it}(R)$ and $M_{ot}(R)$. The internal threshold can, in fact, be determined by bypassing the dioptrics of the eye and forming a system of interference fringes directly on the retina (Le Grand, 1935, 1936; Byram, 1944; Westheimer, 1960; Arnulf and Dupuy, 1960; Campbell and Green, 1965; Berger-Lheureux, 1965; Campbell, 1968; Dupuy, 1968; Burton, 1973; Bour, 1980). Two mutually-coherent point sources are produced close to the nodal points of the eye and the two resultant divergent beams overlap to generate a system of Young's fringes on the retina. The fringe spacing, β rad, is given by $\beta = \lambda/a$, where λ is the wavelength and a is the separation of the two coherent point sources, both measured in air. If the point sources have similar intensity, the fringes will be of unit contrast and it is necessary to have some means of varying the fringe contrast, either by varying the relative intensity of the sources or by adding a suitable background illumination to the retina. The threshold contrast for the retina/brain, $M_{it}(R)$, can then be measured as a function of R, so that we can deduce the MTF of the ocular dioptrics as:

$$T(R) = M_{it}(R)/M_{ot}(R) \qquad (1.12)$$

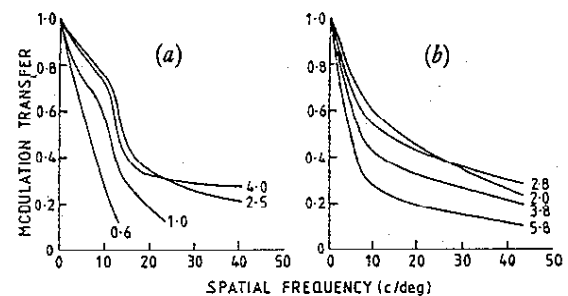Fig. 1.15 gives some typical results for in-focus, foveal

Fig. 1.15 *Ocular MTFs derived from contrast thresholds. The figures on the curves indicate the corresponding pupil diameters in mm (a) After Arnulf and Dupuy (1960), mean of three subjects (b) After Campbell and Green (1965), one subject.*

MTFs as determined by this method. The data show that optimal MTF is achieved with a pupil diameter of around 2–3 mm. Bour (1980) has extended the method to explore through-focus changes in axial MTF and finds that irregular aberration may lead to more than one position of optimal focus. The method has also been applied to off-axis imagery: although Green (1970) and Enoch and Hope (1973) deduced that it was unlikely that optical factors contributed significantly to the well-known loss in acuity with increasing field angle, Frisén and Glasholm (1975) felt that their data suggested that optical degradation was of some significance. This discrepancy has yet to be fully resolved.

The major criticism of the psychophysical technique is that the assumption that the interference fringes are totally unaffected by the ocular media may not be justified. Ocular scattered light may well degrade the fringe contrast, particularly in older eyes. This would tend to lead to an over-estimate of the modulation transfer. It is also difficult to be sure that the subjects are consistent in their judgements of the internal and external thresholds, particularly in those studies where there are differences in the colour and field size of the two grating systems. Indeed, some studies using the technique have produced much less plausible results (Fraser and Morrison, 1987).

Ophthalmoscopic Methods

If the image of a suitable object is thrown onto the retina, a proportion of the flux in the retinal image will be reflected back out of the eye and, given an appropriate optical system, can allow an observer to view the relayed retinal image. This is, of course, the principle of the ophthalmoscope (see Henson, Chapter 17). If, for example, the original object is a line, the relayed image will be a line-spread function. It will be noted that two stages of image degradation are involved in this process and it is necessary to assume that no coherence is preserved in the light reflected from the retina. Under these circumstances the

overall OTF is simply the product of the inward and outward OTFs. Although the scattering geometries involved might conceivably make these OTFs differ, it is usual to assume that they are the same, so that the double-pass OTF is simply the square of the desired single-pass OTF.

This type of measurement was pioneered by Flamant (1955) who used a photographic technique to record the double-pass line-spread function and went on to calculate its single-pass counterpart. Photographic techniques pose considerable problems of non-linearity and lack of sensitivity and all subsequent experimenters have used photo-electric techniques (e.g. Krauskopf, 1962, 1964; Röhler, 1962; Westheimer and Campbell, 1962; Campbell and Gubisch, 1966; Röhler *et al.*, 1969; Jennings and Charman, 1978, 1981; Gorrand *et al.*, 1978; Gorrand, 1979; Santamaria *et al.*, 1987).

Most of the earlier authors measured the external line-spread function, took its Fourier transform to yield the double-pass OTF, square rooted this to give the single-pass OTF and, finally, took the inverse Fourier transform to give the single-pass line-spread function. In fact, in

most instances the images were sufficiently symmet: that phase terms could be ignored, so that only the M was considered. More recently, Santamaria *et al.* (19 using a television camera and image analysis sys (Fig. 1.16), have made direct measurements of the po spread function: this has the advantage that it allows ca. lation of the LSF and OTF for any orientation.

In general, the results of this type of study agree with the findings of other methods. Fig. 1.17 shows so representative axial data. Ocular aberration is found become increasingly significant as the pupil diame exceeds 2–3 mm. Off axis, image quality shows only a si decline for field angles up to about 10°; thereafter image progressively blurs and oblique astigmati becomes more important. The oblique astigmatism det mined by this type of measurement is similar in magnit: and form to that found by retinoscopic or optometer tec niques (Jennings and Charman, 1978, 1981).

One major uncertainty in this double-pass ophth: moscopic technique concerns the nature of the retir reflection. It is assumed that all coherence is lost at t
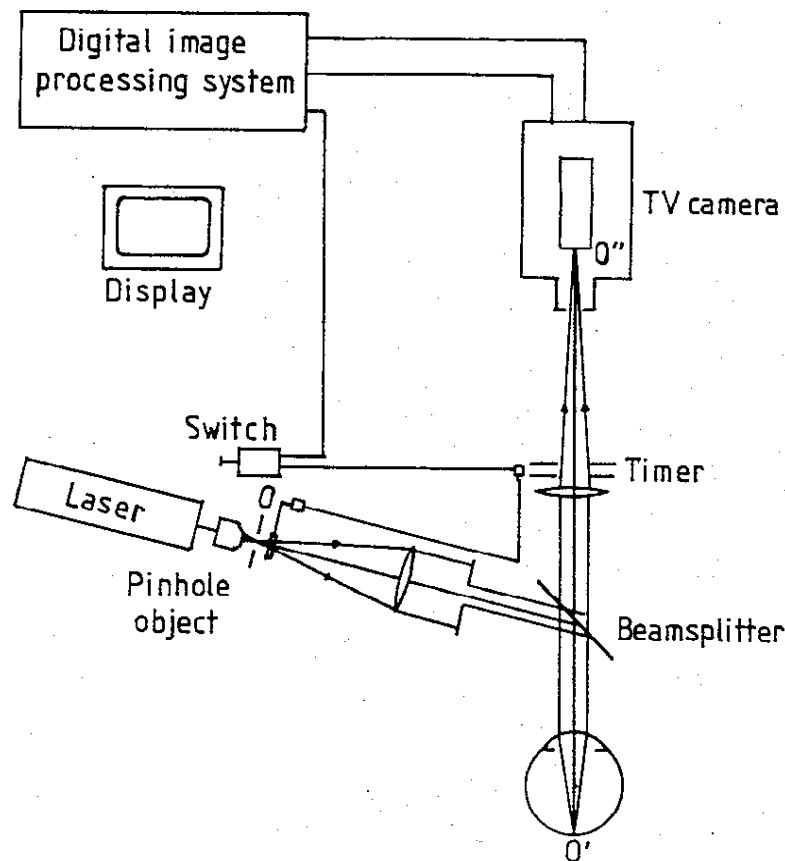

Fig. 1.16 *Arrangement used by Santamaria* et al. *(1987) for recording the double-pass retinal point-spread function.*

ntly symmetrical
at only the MTF
aria *et al.* (1987),
analysis system
nts of the point-
at it allows calcu-
itation.

study agree well
..17 shows some
ion is found to
pupil diameter
1ows only a slow
; thereafter the
1e astigmatism
igmatism deter-
ar in magnitude
optometer tech-
!1).

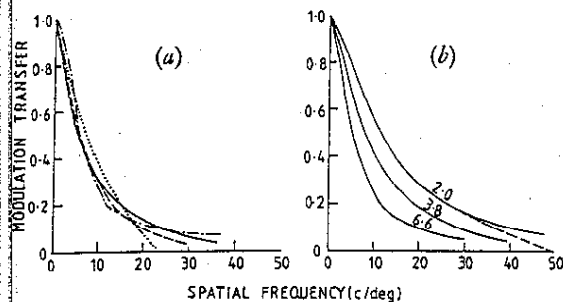:-pass ophthal-
: of the retinal
:e is lost at the

Fig. 1.17 *Ocular MTFs derived by the double-pass ophthalmoscopic technique. (a) Examples of MTFs for 5 mm pupils as found by different investigators. ......: Krauskopf (1962), white light; ————— Campbell and Gubisch (1966), white light; ————— Charman and Jennings (1976), 555 nm; —+—. Santamaria et al. (1987) 633 nm. (b) Mean data of Campbell and Gubisch (1966) for three subjects and the pupil diameters (mm) indicated; the dashed curve is for a 1.5 mm pupil. White light.*

retina and that the reflection itself does not introduce any degradation, i.e. it occurs at a single surface (see, e.g. Charman, 1983, for review). In practice some degradation must occur and the exact extent to which this affects the measured OTFs remains somewhat uncertain. A further problem is that, when recording very faint reflected images, there may be a tendency to truncate the outer parts of the image where the illuminance is very low but which may still contain substantial amounts of light flux due to their considerable area: this can lead to a significant overestimate of modulation transfer (Simon and Denieul, 1973). Vos *et al.* (1976) have attempted to overcome this latter problem by combining ophthalmoscopic measurements of the retinal point-spread function (Campbell and Gubisch, 1966) with the estimates of other authors for the amount of wider-angle, entoptic stray light, to produce a realistic estimate of the light profiles in the foveal image of a white-light point source.

Future Developments

It seems likely that, with the further development and availability of improved low light-level cameras, frame-grabbing, and image analysis techniques, rapid acquisition of data on retinal image quality will become much more routine. This will allow extended study of not only the naked eye but also the contact lens wearing eye, in which aberration as well as overall refraction is modified by lens wear. Clearly, too, there would be considerable interest in investigating image quality in eyes with intraocular implants or after refractive surgery: a full understanding of the image quality, both axial and peripheral, that is achieved should lead to improvements in lens design and surgical techniques.

Analytical approximations for the ocular MTF

Several authors have attempted to approximate the experimentally-determined ocular MTFs by analytic functions (see Charman, 1983, for review). Such functions have obvious merits in simplifying calculation and in allowing the comparison of different individual MTFs. Functions suggested include:

$$T(R) = \exp[-R/R_c]$$

$$T(R) = \exp[-(R/R_c)^2]$$

$$T(R) = \exp[-(R/R_c)^n] \tag{1.13}$$

where R_c is some constant spatial frequency and n is a further constant. The last expression has been found to provide a reasonable fit to ocular MTFs at low and medium spatial frequencies and might, perhaps, be useful under some circumstances (Jennings and Charman, 1974; Drasdo and Cox, 1987; Thompson and Drasdo, 1989).

Ocular Depth-of-Focus

In considering the MTFs in the preceding section, it was tacitly assumed that the eye was optimally focussed. There is, in fact, some ambiguity in defining optimal focus, since the focus at which modulation transfer reaches its highest value may be spatial-frequency dependent in the presence of aberration (e.g. Koomen *et al.*, 1951; Green and Campbell, 1965; Charman and Jennings, 1976a; Charman *et al.*, 1978). Thus the optimal focus may vary somewhat with the spatial frequency content of the object (i.e. its spatial form). In general, however, a shift away from the optimal focus produces an overall loss in modulation transfer, corresponding to an increase in image blur. In a geometrical optical approximation, the blur increases linearly with the pupil diameter and the dioptric defocus, although this approximation is poor near the region of focus. Fig. 1.18 plots the through-focus modulation transfer for a diffraction-limited eye, at a number of spatial frequencies.

We might expect, then, that the ability of the observer to detect defocus would depend upon the characteristics of the object under observation (form, spectral content, contrast, luminance), the pupil diameter and the visual characteristics at the retina/brain level. In particular, since only low spatial frequency information can be perceived at low luminances (Van Nes and Bouman, 1967), one might anticipate a corresponding increase in blur thresholds. For the same basic reason, it would be expected that low vision patients would show greater tolerance to blur, and hence greater depth-of-focus (Legge *et al.*, 1987).

As the range, or depth, of focus over which blur has negligible effect is of considerable practical significance with respect to the need for refractive correction, the pre-
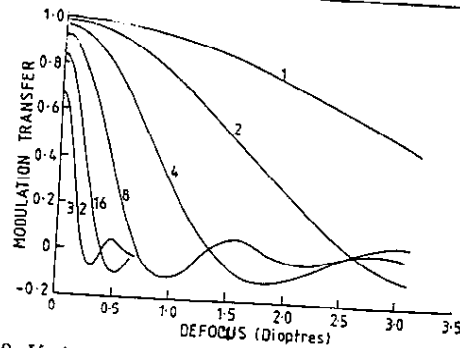
Fig. 1.18 *Variation in theoretical modulation transfer with defocus for a diffraction-limited eye with a 4 mm diameter entrance pupil and monochromatic light of wavelength 555 nm. Each curve refers to the single spatial frequency (c deg^{-1}) indicated. Note that high spatial frequencies are much more rapidly degraded by defocus than are lower spatial frequencies.*
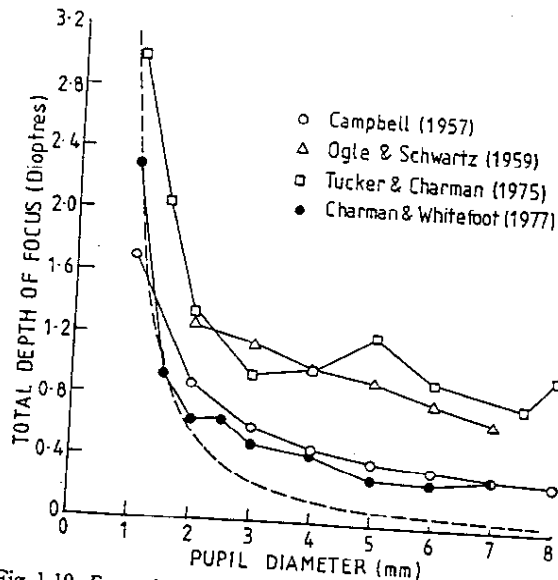

Fig. 1.19 *Examples of experimental measurements of photopic, total, monocular depth-of-focus as a function of pupil diameter (optimal focus lies midway through the total depth-of-focus). ○ Campbell (1957) Based on the just perceptible blur for a small disc, one subject, white light. △ Ogle and Schwartz (1959) based on 50% probability of resolving a 20/25 chequerboard; mean of three subjects, white light. □ Tucker and Charman (1975) based on 80% probability of achieving 90% of the optimal Snellen acuity; mean of two subjects, white light. ● Charman and Whitefoot (1977) based on the detectable movement of laser speckles; mean of six subjects, 633 nm. The dashed line gives the depth-of-focus based on Rayleigh's quarter-wavelength criterion for an aberration-free eye in monochromatic light of wavelength 555 nm.*

...sion required in that correction, and for the design of instrumentation, it has been measured by many investigators using a variety of techniques. Fig. 1.19 shows some

typical data as obtained under photopic conditions. . that depth-of-focus decreases less rapidly with $_\Gamma$ diameter than would be predicted from either geomet or physical optical considerations for an aberration-eye. This may be due partly to the apodizing role o: Stiles-Crawford effect and partly to the increasing abc tion at the edge of dilated pupils.

Even if a constant technique is used, the measu depth-of-focus depends upon the criterion used to asse and it is striking that, in accommodation experime accommodation may be stimulated by very small char ($\leqslant 0.1$ D) in the vergence (distance) of the target (Ludlam *et al.*, 1968). The limited dioptric range o which a target can be moved without the detection of $\mathfrak{r}$ implies of course, the need for a focussing or accommo tion system, so that the retinal image quality is usu: slightly worse that would be obtained at optimal focus

Matching Between Optical and Neural Quality

Helmholtz long ago suggested that, for resolution of grating, an unstimulated retinal element must be prese between those elements which are stimulated by neig bouring bright bars of the grating (Helmholtz, 1924). Th basic concept is now formalized within the framework sampling theory. It is easy to show that if it is assumed th the retinal receptors are packed in a hexagonal array, wit neighbouring receptors having centre-to-centre spacin D, the highest sinusoidal spatial frequency that can unam biguously be resolved by the array is given by:

$$v = d\pi/180\,D(3)^{1/2} \ \text{c deg}^{-1}$$

d is the posterior nodal distance (approximately 16.7 mm (Snyder and Miller, 1977; Miller and Bernard, 1983. Gratings of higher frequency than this Nyquist limit, : are undersampled and may appear as spurious gratings o lower frequency – the phenomenon known as 'aliasing' Such aliasing can be observed under the very abnorma viewing conditions where interference fringes of high spa-tial frequency are generated on the retina, as in the psychophysical method for determining the ocular OTF. Several authors have used these aliasing effects to make deductions about the spatial arrangement of the cone mosaic in the retina (e.g. Williams, 1985; Coletta and Wil-liams, 1987; Smith and Cass, 1987; Thibos *et al.*, 1987; Williams, 1988).

In general, aliasing is likely to be unfavourable and ideally the cut-off frequency of the optics should be com-parable to that of the receptor array. For incoherently-illuminated objects viewed in the normal way it would appear that this matching is at least approximately achieved in the foveal region. The attenuation provided by

the modulation transfer of the eye's dioptrics substantially reduces the chances of aliasing being observed. It is possible that there would, in fact, be no advantage in maintaining the diffraction-limited performance of the eye for pupil sizes greater than 2–3 mm. The superior image quality available from a diffraction-limited eye with, say, a 6 mm pupil would mean that the effective optical cut-off frequency would substantially exceed that of the retina. In this sense, then, the axial aberration of the eye may not be disadvantageous. It would indeed appear reasonable that the optical quality should decline for large pupils, since the natural pupil only dilates when the light level is low and the retinal cut-off frequency is lowered. It is interesting to note that there is evidence that the degradation in image quality that occurs off-axis may serve to maintain the optical and neural matching into the peripheral regions, where the retinal ganglion cell intervals and receptive field sizes are larger (Jennings and Charman, 1978, 1981).

Factors other than the MTF of the eye's optics may, of course, also play a role in reducing the possibility of aliasing. Both disorder in the receptor array (Yellot, 1982; Hirsch and Miller, 1987) and averaging over the receptor aperture (Miller and Bernard, 1983) may be of importance. Recent measurements of human foveal cone spacing by Hirsch and Curcio (1989) lead them to suggest that within about 2° from the centre of the fovea conventional measurements of photopic visual acuity correlate quite well with those predicted from the cone spacing, implying that the optical performance of the eye can support its neural capabilities: only at the foveal centre is there a suggestion that optical performance may fall short of the ideal.

Much remains to be done to clarify certain aspects of the optics of the eye, particularly those relating to the refractive index distribution and shape of the lens as a function of age and accommodation, and to optical performance in the peripheral field. Nevertheless, it is unlikely that improved understanding of the eye will in any way diminish our respect for the way in which its optical performance closely matches the needs of the neural parts of the visual system.

References

Abrahamsson, M. and Sjostrand, J. (1986). Impairment of contrast sensitivity function (CSF) as a measure of disability glare. *Invest. Ophthalmol. Vis. Sci.*, 27, 1131–1136.

Allen, M. J. and Vos, J. J. (1967). Ocular scattered light and visual performance as a function of age. *Am. J. Optom. Arch. Am. Acad. Optom.*, 44, 717–727.

Arnulf, A. and Dupuy, O. (1960). La transmission des contrastes par le système optique de l'oeil et les seuils de contrastes rétiniens. *C. R. hebd. Séanc. Acad. Sci. Paris*, 250, 2757–2759.

Artal, P., Santamariá, J. and Bescós, J. (1988a). Retrieval of wave aberration of human eyes from actual point-spread-function data. *J. Opt. Soc. Am. A*, 5, 1201–1206.

Artal, P., Santamariá, J. and Bescós, J. (1988b). Phase-transfer function of the human eye and its influence on point-spread function and wave aberration. *J. Opt. Soc. Am.*, A5, 1791–1795.

Axelrod, D., Lerner, D. and Sands, P. J. (1988). Refractive index within the lens of a goldfish eye determined from the paths of thin laser beams. *Vision Res.*, 28, 57–65.

Azen, S. P., Burg, K. A., Smith, R. E. and Maguen, E. (1979). A comparison of three methods for the measurement of corneal thickness. *Invest. Ophthalmol. Vis. Sci.*, 18, 535–538,

Bando, J., Ishii, Y. and Nakajima, A. (1976). Changes in blue fluorescence intensity and coloration of human lens protein with normal lens aging and nuclear cataract. *Ophthalmic Res.*, 8, 456–463.

Barnes, D. A., Dunne, M. C. M. and Clement, R. A. (1987). A schematic eye model for the effects of translation and rotation of ocular components on peripheral astigmatism. *Ophthalmic Physiol. Opt.*, 7, 153–158.

Bedell, H. E. and Katz, L. M. (1982). On the necessity of correcting peripheral target luminance for pupillary area. *Am. J. Optom. Physiol. Opt.*, 59, 767–769.

Bedford, R. E. and Wyszecki, G. (1957). Axial chromatic aberration of the human eye. *J. Opt. Soc. Am.*, 47, 564–565.

Benedek, G. B. (1971). Theory of transparency of the eye. *Appl. Opt.*, 10, 459–473.

Bennett, A. G. and Rabbetts, R. B. (1988). Schematic eyes – time for a change? *Optician*, 196 (5169), 14–15.

Berger-Lheureux, S. (1965). Mesure de la fonction de transfert de modulation du système optique de l'oeil et des seuils de modulation rétiniens. *Rev. Opt. Théor. Instrum.*, 44, 294–333.

Berny, F. (1969). Etude de la formation des images rétiniennes et détermination de l'aberration de sphéricité de l'oeil humaine. *Vision Res.*, 9, 977–990.

Berny, F. and Slansky, S. (1969). Wavefront determination resulting from Foucault test as applied to the human eye and visual instruments. In *Optical Instruments and Techniques*. ed. Home Dickson, J. pp. 375–385. London: Oriel Press.

Blaker, J. W. (1980). Towards an adaptive model of the human eye. *J. Opt. Soc. Am.*, 70, 220–223.

Bobier, C. W. and Sivak, J.G. (1978). Chromoretinoscopy. *Vision Res.*, 18, 247–250.

Boerttner, E. A. and Wolter, J. R. (1962). Transmission of the ocular media. *Invest. Ophthalmol.*, 1, 776–783.

Bonnet, R. (1959). Stéréophotogrammetrie de la cornée humaine. *Rev. d'Optique*, 38, 447–462.

Born, M. and Wolf, E. (1975). *Principles of Optics*. 5th edn. Oxford: Pergamon.

Bornstein, M. H. (1977). Developmental pseudocyanopsia: ontogentic change in human colour vision. *Am. J. Optom. Physiol. Opt.*, 54, 464–469.

Bour, L. J. (1980). MTF of the defocused optical system of the human eye for incoherent monochromatic light. *J. Opt. Soc. Am.*, 70, 321–328.

Boynton, R. M. and Clarke, F. J. J. (1964). Sources of entoptic scatter in the human eye. *J. Opt. Soc. Am.*, 54, 110–119, 717–719.

Brown, N. (1973). The change in shape and internal form of the lens of the eye on accommodation. *Exp. Eye Res.*, 15, 441–459.

Brown, N. (1974). The shape of the lens equator. *Exp. Eye Res.*, 19, 571–576.

Burton, G. J. (1973). Evidence for non-linear response processes in the human visual system from measurements on the thresholds of spatial beat frequencies. *Vision Res.*, 13, 1211–1255.

Byram, G. M. (1944). The physical and photochemical basis of visual resolving power. I. The distribution of illumination in retinal images. *J. Opt. Soc. Am.*, 34, 571–591.

Campbell, F. W. (1957). The depth of field of the human eye. *Optica Acta*, 4, 157–164.

Campbell, F. W. (1968). The eye as an optical filter. *Proc. IEEE*, 56, 1009–1014.

Campbell, F. W. and Green, D. G. (1965). Optical and retinal factors affecting visual resolution. *J. Physiol. (Lond.)*, 181, 576–593.

Campbell, F. W. and Gubisch, R. W. (1966). Optical quality of the human eye. *J. Physiol. (Lond.)*, 186, 558–578.

Campbell, F. W. and Gubisch, R. W. (1967). The effect of chromatic aberration on visual acuity. *J. Physiol. (Lond.)*, 192, 345–358.

Campbell, M. C. W. (1984). Measurement of refractive index in an intact crystalline lens. *Vision Res.*, 24, 409–415.

Campbell, M. C. W. (1987). The influence of the crystalline lens on the image quality of the eye. Proc. 14th Congress ICO, Quebec, 24–28th August 1987, *Proc. SPIE*, 813, 47–48.

Campbell, M. C. W. and Hughes, A. (1981). An analytic gradient index schematic lens and eye for the rat which predicts aberrations for finite pupils. *Vision Res.*, 21, 1129–1148.

Campbell, M. C. W. and Hughes, A. (1987). The refractive index distribution and image quality of a human crystalline lens. *Advances in Diagnostic Visual Optics.* eds. Fiorentini, A., Guyton, D. and Siegel, I. M. pp. 42–46. Heidelberg: Springer.

Chan, D. Y. C., Ennis, J. P., Pierscionek, B. K. and Smith, G. (1988). Determination and modelling of the 3-D gradient refractive indices in crystalline lenses. *Appl. Optics*, 27, 926–931.

Charman, W. N. (1983). The retinal image in the human eye. In *Progress in Retinal Research.* eds Osborne, N. and Chader, G. Vol. 2. pp. 1–50. Oxford: Pergamon.

Charman, W. N. (1985). The flawed retinal image. *Optician*, 190 (5020), 31–34 and 190 (5022), 16–20.

Charman, W. N. (1989). Light on the peripheral retina. *Ophthalmic Physiol. Opt.*, 9, 91–92.

Charman, W. N. and Heron, G. (1979). Spatial frequency and the dynamics of the accommodation response. *Optica Acta*, 26, 217–228.

Charman, W. N. and Jennings, J. A. M. (1976a). The optical quality of the retinal image as a function of focus. *Br. J. Physiol. Opt.*, 31, 119–134.

Charman, W. N. and Jennings, J. A. M. (1976b). Objective measurements of the longitudinal chromatic aberration of the human eye. *Vision Res.*, 16, 999–1006.

Charman, W. N. and Jennings, J. A. M. (1982). Ametropia and peripheral refraction. *Am. J. Optom. Physiol. Opt.*, 59, 922–923.

Charman, W. N. and Tucker, J. (1977). Dependence of accommodation response on the spatial frequency spectrum of the observed object. *Vision Res.*, 17, 129–139.

Charman, W. N. and Tucker, J. (1978). Accommodation and color. *J. Opt. Soc. Am.*, 68, 459–471.

Charman, W. N. and Walsh, G. (1985). The optical phase transfer function of the eye and the perception of spatial phase. *Vision Res.*, 25, 619–623.

Charman, W. N. and Whitefoot, H. (1977). Pupil diameter and the depth-of-field of the human eye as measured by laser speckle. *Optica Acta*, 24, 1211–1216.

Charman, W. N., Jennings, J. A. M. and Whitefoot, H. (1978). The refraction of the eye in relation to spherical aberration and pupil size. *Br. J. Physiol. Opt.*, 32, 78–93.

Clark, B. A. J. (1972). Autocollimating photokeratoscope. *J. Opt. Soc. Am.*, 62, 169–176.

Clark, B. A. J. (1974a). Topography of some individual corneas. *Aust. J. Optom.*, 57, 65–69.

Clark, B. A. J. (1974b). Mean topography of normal corneas. *Aust. J. Optom.*, 57, 107–114.

Clark, B. A. J. and Carney, L. G. (1971). Refractive index and reflectance of the anterior surface of the cornea. *Am. J. Optom. Arch. Am. Acad. Optom.*, 48, 333–343.

Coletta, N. J. and Williams, D. R. (1987). Psychophysical estimate of extrafoveal cone spacing. *J. Opt. Soc. Am. A*, 4, 1503–1513.

Davson, H. (1980). *Physiology of the Eye.*, 4th edn. Edinburgh: Churchill Livingstone.

Deller, J. F. P., O'Connor, A. D. and Sorsby, A. (1947). X-ray measurements of the diameters of the living eye. *Proc. R. Soc. B*, 134, 456–467.

Dingeldein, S. A. and Klyce, S. D. (1989). The topography of normal corneas. *Arch. Ophthalmol.*, 107, 512–518.

Drasdo, N. and Cox, W. (1987). Local luminance effects of degraded pattern stimulation. *Clin. Vis. Sci.*, 1, 317–325.

Drasdo, N. and Fowler, C. W. (1974). Non-linear projection of the retinal image in a wide-angle schematic eye. *Br. J. Ophthalmol.*, 58, 709–714.

Drasdo, N. and Peaston, W. C. (1980). Sampling systems for visual field assessment and computerised perimetry. *Br. J. Ophthalmol.*, 64, 705–712.

Duke-Elder, S. and Abrams, D. (1970). *Ophthalmic Optics and Refraction. System of Ophthalmology. Vol. V.* London: Kimpton.

Duke-Elder, S. and Wybar, K. C. (1961). *The Anatomy of the Visual System. System of Ophthalmology. Vol. II.* London: Kimpton.

Dunne, M. C. M., Barnes, D. A. and Clement, R. A. (1987). A model for retinal shape changes in ametropia. *Ophthalmic Physiol. Opt.*, 7, 159–160.

Dupuy, O. (1968). La perception visuelle. *Vision Res.*, 8, 1507–1520.

Edmund, C. and Sjontoft, E. (1985). The central-peripheral radius of the normal corneal curvature. *Acta Ophthalmologica* 63, 670–677.

Ehlers, N. (1965). The precorneal film. *Acta Ophthalmol. (Suppl.)*, 81, 21–34.

El Hage, S. G. (1976). The three-dimensional configuration of the cornea. *Nouv. Rev. Optique*, 7, 205–209.

El Hage, S. G. and Berny, F. (1973). Contribution of the crystalline lens to the spherical aberration of the eye. *J. Opt. Soc. Am.*, 63, 205–211.

Emsley, H. H. (1952). *Visual Optics.* Vol. 1. London: Hatton.

Enoch, J. M. and Hope, G. M. (1973). Interferometric resolution determinations in the fovea and parafovea. *Doc. Ophthalmol.*, 34, 143–156.

Ferree, C. E. and Rand, G. (1932). The refractive conditions for the peripheral field of vision. In *Report of Joint Discussion on Vision.* pp. 244–262. London: Physical Society.

Ferree, C. E. and Rand, G. (1933). Interpretation of refractive conditions in the peripheral field of vision. *Arch. Ophthalmol. (Chicago)*, 9, 925–938.

Ferree, C. E., Rand, G. and Hardy, C. (1931). Refraction for the peripheral field of vision. *Arch. Ophthalmol. (Chicago)*, 5, 717–731.

Ferree, C. E., Rand, G. and Hardy, C. (1932). Refractive asymmetry in the temporal and nasal halves of the visual field. *Am. J. Ophthalmol.*, 15, 513–522.

Feuk, T. (1970). On the transparency of the stroma in the mammalian cornea. *IEE Trans. Biomed. Eng. (BME)*, 17, 186–190.

Feuk, T. (1971). The wavelength dependence of scattered light intensity in rabbit corneas. *IEEE Trans. Biomed. Eng. (BME)*, 18, 92–96.

Fincham, W. H. A. and Freeman, M. H. (1980). *Optics.* 9th edn. Chapter 20. London: Butterworths.

Fitzke, F. W. (1981). Optical properties of the eye (Abstract). *Invest. Ophthalmol. Vis. Sci.*, 20 (Suppl.), 144.

Flamant, F. (1955). Etude de la répartition de la lumière dans l'image rétinienne d'une fente. *Rev. Opt. Théor. Instrum.*, 34, 433–459.

Françon, M. (1951). Aberration sphérique, chromatisme et pouvoir séparateur de l'oeil. *Rev. Opt. Théor. Instrum.*, 30, 71–80.

Fraser, I. M. and Morrison, J. D. (1987). Optical contrast ratio at low spatial frequencies in the human eye. *J. Physiol. (Lond.)*, 392, 109.

Frisén, L. and Glansholm, A. (1975). Optical and neural resolution in peripheral vision. *Invest. Ophthalmol.*, 14, 528–536.

Fry, G. A. (1955). Blur of the retinal image. *Br. J. Physiol. Opt.*, 12, 130–152.

Fujii, T., Maruyama, S. and Ikeda, M. (1972). Determination of corneal configuration by the measurement of its derivatives. *Optica*

Acta, 19, 425–430.

Gaydon, A. G. (1938). Colour sensations produced by ultra-violet light. *Proc. Phys. Soc. Lond.*, 50, 714–720.

Geeraets, W. J. and Berry, E. R. (1968). Ocular spectral characterstics as related to hazards from lasers and other sources. *Am. J. Ophthalmol.*, 66, 15–20.

Goldmann, H. (1937). Studien über die Alerskernstreifen der Linse. *Arch. Augenheilk.*, 110, 405–414.

Gorrand, J. M. (1979). Diffusion of the human retina and quality of the optics of the eye on the fovea and the peripheral retina. *Vision Res.*, 19, 907–912.

Gorrand, J. M., Dupuy, O., Farfal, F., Plantegenest, M. and Slansky, S. (1978). Contributions de la rétine et du système optique de l'oeil a l'abaissement de la modulation de l'image par double traversée de l'oeil. *J. Optics (Paris)*, 9, 359–364.

Green, D. G. (1970). Regional variation in the visual acuity for interference fringes on the retina. *J. Physiol. (Lond.)*, 207, 351–356.

Green, D. G. and Campbell, F. W. (1965). Effect of focus on the visual response to a sinusoidally modulated spatial stimulus. *J. Opt. Soc. Am.*, 55, 1154–1157.

Gregory, R. (1991). *Evolution of the eye and visual system.*, Vol. 2. *Vision and Visual Dysfunction*. London: Macmillan.

Guidarelli, S. (1972). Off-axis imaging in the human eye. *Atti. Fond. G. Ronchi*, 27, 407–415.

Guillon, J-P. (1982). Tear film photography and contact lens wear. *J. Br. Contact Lens Ass.*, 5, 84–86.

Guillon, M., Lydon, D. P. M. and Wilson, C. (1986). Corneal topography: a clinical model. *Ophthalmol. Physiol. Opt.*, 6, 47–56.

Gullstrand, A. (1924). Appendices to *Physiological Optics*. Vol. 1, by Helmholtz, H. von, transl. Southall, J. P. C. Rochester: Optical Society of America.

Hart, R. W. and Farrell, R. A. (1969). Light scattering in the cornea. *J. Opt. Soc. Am.*, 59, 766–774.

Hartridge, H. (1947). The visual perception of fine detail. *Phil. Trans. R. Soc. B*, 232, 519–671.

Helmholtz, H. von (1924). *Treatise on Physiological Optics.* Translated from the 3rd German edn. by Southall, J. P. C. Rochester: Optical Society of America.

Helmholtz, H. von (1962). *Popular Scientific Lectures.* pp. 116–117. New York: Dover. (Originally published 1868).

Hemenger, R. P. (1984). Intraocular light scatter in normal vision loss with age. *Appl. Opt.*, 23, 1972–1974.

Hemenger, R. P. (1988). Small-angle intraocular scattered light: a hypothesis concerning its source. *J. Opt. Soc. Am. A*, 5, 577–582.

Hess, E. H. (1965). Attitude and pupil size. *Sci. Am.*, 212(4), 46–54.

Hirji, N. K. and Larke, J. R. (1978). Thickness of human cornea measured by topographic pachometry. *Am. J. Optom. Physiol. Opt.*, 55, 97–100.

Hirsch, J. and Curcio, C. A. (1989). The spatial resolution capacity of the human foveal retina. *Vision Res.*, 29, 1095–1011.

Hirsch, J. and Miller, W. H. (1987). Does cone positional disorder limit resolution? *J. Opt. Soc. Am. A*, 4, 1481–1492.

Holladay, L. L. (1926). The fundamentals of glare and disability. *J. Opt. Soc. Am.*, 12, 271–319.

Hoogerheide, J. K., Rempt, F. and Hoogenboom, W. P. H. (1971). Acquired myopia in young pilots. *Ophthalmologica*, 163, 209–215.

Hopkins, H. H. (1950). *Wave Theory of Aberrations.* London: Oxford.

Hopkins, H. H. (1962). The application of frequency response techniques in optics. *Proc. Phys. Soc.*, 79, 889–919.

Howarth, P. A. (1984). The lateral chromatic aberration of the eye. *Ophthalmic Physiol. Opt.*, 4, 223–236.

Howarth, P. A. and Bradley, A. (1986). The longitudinal chromatic aberration of the human eye, and its correction. *Vision Res.*, 26, 361–366.

Howarth, P. A., Zhang, X. X., Bradley, A., Still, D. L. and Thibos, L. N. (1988). Does the chromatic aberration of the eye vary with age?

J. Opt. Soc. Am. A, 5, 2087–2092.

Howcroft, M. J. and Parker, J. A. (1977). Aspheric curvatures for the human lens. *Vision Res.*, 17, 1217–1223.

Howland, B. and Howland, H. C. (1976). Subjective measurement of high-order aberrations of the eye. *Science*, 193, 580–582.

Howland, H. C. and Buettner, J. (1989). Computing high order wave aberration coefficients from variations of best focus for small artificial pupils. *Vision Res.*, 29, 979–983.

Howland, H. C. and Howland, B. (1977). A subjective method for the measurement of the monochromatic abberations of the eye. *J. Opt. Soc. Am.*, 67, 1508–1518.

Ivanoff, A. (1947). On the influence of accommodation on spherical aberration of the human eye. *J. Opt. Soc. Am.*, 37, 730–731.

Ivanoff, A. (1953). *Les Aberrations de l'oeil.* Paris: Revue d'Optique.

Ivanoff, A. (1956). About the spherical aberration of the eye. *J. Opt. Soc. Am.*, 46, 901–903.

Janisse, M. P. (1973). *Pupillary Dynamics and Behaviour*. New York: Plenum Press.

Jay, B. S. (1962). The effective pupillary area at varying perimetric angles. *Vision Res.*, 1, 418–424.

Jenkins, T. C. A. (1963a). Aberrations of the eye and their effects on vision. Part I. *Br. J. Physiol. Opt.*, 20, 50–91.

Jenkins, T. C. A. (1963b). Aberrations of the eye and their effects on vision. Part II. *Br. J. Physiol. Opt.*, 20, 161–201.

Jennings, J. A. M. and Charman, W. N. (1974). An analytical approximation for the modulation transfer of the eye. *Br. J. Physiol. Opt.*, 29, 64–72.

Jennings, J. A. M. and Charman, W. N. (1978). Optical image quality in the peripheral retina. *Am. J. Optom. Physiol. Opt.*, 55, 582–590.

Jennings, J. A. M. and Charman, W. N. (1981). Off-axis image quality in the human eye. *Vision Res.*, 21, 445–455.

Josephson, J. E. (1983). Appearance of the pre-ocular tear film lipid layer. *Am. J. Optom. Physiol. Opt.*, 60, 883–887.

Kiely, P. M., Smith, G. and Carney, L. G. (1982). The mean shape of the human cornea. *Optica Acta*, 29, 1027–1040.

Kilbride, P. E., Alexander, K. R., Fishman, M. and Fishman, G. A. (1989). Human macular pigment assessed by imaging fundus reflectometry. *Vision Res.*, 26, 663–674.

Kishto, B. N. (1970). Variation of the visual threshold with retinal location. Part 1. The central 20° of visual field. *Vision Res.*, 10, 745–767.

Kooijman, A. C. (1983). Light distribution on the retina of a wide angle theoretical eye. *J. Opt. Soc. Am.*, 73, 1544–1550.

Kooijman, A. C. and Witmer, F. K. (1986). Ganzfield light distribution on the retina of human and rabbit eyes: calculations and *in vitro* measurements. *J. Opt. Soc. Am. A*, 3, 2116–2120.

Koomen, M., Skolnik, R. and Tousey, R. (1951). A study of night myopia. *J. Opt. Soc. Am.*, 41, 80–90.

Koomen, M., Tousey, R. and Skolnik, R. (1949). The spherical aberration of the eye. *J. Opt. Soc. Am.*, 39, 370–376.

Koretz, J. F., Handelman, G. H. and Brown, N. P. (1984). Analysis of human crystalline lens curvature as a function of accommodation and age. *Vision Res.*, 24, 1141–1151.

Krauskopf, J. (1962). Light distribution in human retinal images. *J. Opt. Soc. Am.*, 52, 1046–1050.

Krauskopf, J. (1964). Further measurements of human retinal images. *J. Opt. Soc. Am.*, 54, 715–720.

Lau, E., Mutze, K. and Weber, F. G. (1955). Die chromatische Aberration des menschlichen Auges. *Albrecht v. Graefes Arch. Ophthalmol.*, 57, 92–100.

Legge, G. E., Muller, K. T., Woo, G. C. and Campbell, F. W. (1987). Tolerance to visual defocus. *J. Opt. Soc. Am. A*, 4, 851–863.

Le Grand, Y. (1935). Sur la measure de l'acuité visuelles au moyen de franges d'interference. *C. R. Hebd. Séanc. Acad. Sci. Paris*, 200, 490–491.

Le Grand. Y. (1936). Sur une mode de vision éliminant les défauts

(947). X-ray
. *R. Soc. B*,

hy of normal

of degraded

ction of the
thalmol., 58,

1s for visual
Ophthalmol.,

Optics and
impton.
f the Visual
ipton.
7). A model
iol. Opt., 7,

1507–1520.
al radius of
670–677.
. (Suppl.),

tion of the

crystalline
. Am., 63,

on.
resolution
almol., 34,

ns for the
on Vision.

ive condi-
Chicago),

n for the
717–731.
symmetry
Ophthal-

ammalian

ght inten-
ME), 18,

9th edn.

). *Invest.*

s l'image
-459.
: pouvoir

io at low
392, 109.
lution in

Opt., 12,

ation of
. *Optica*

optiques de l'oeil. *Rev. Opt. Théor. Instrum.*, 15, 6–11.

Le Grand, Y. (1967). *Form and Space Vision*. Transl. Millodot, M. and Heath, G. C. Bloomington: Indiana U.P.

Le Grand, Y. and El Hage, S. G. (1980). *Physiological Optics*. Berlin: Springer-Verlag.

Lerman, S. (1972). Lens proteins and fluorescence. *Isr. J. Med. Sci.*, 8, 1583–1589.

Lerman, S. (1980). *Radiant Energy and the Eye*. London; Balliere Tindall.

Levi, L. (1974). *Handbook of Tables for Applied Optics*. Cleveland: CRC Press.

Linfoot, E. H. and Wolf, E. (1956). Phase distribution near focus in an aberration-free diffraction image. *Proc. Phys. Soc. B*, 69, 823–832.

Lommel, E. (1884). Die Beugungerscheinungen einer Kreisrunden Oeffnung und eines Kreis runden schirchens. *K. Bayer Akadd. Wissenschaft*, 15, 229–328.

Lotmar, W. (1971). Theoretical eye model with aspherics. *J. Opt. Soc. Am.*, 61, 1522–1529.

Lotmar, W. and Lotmar, T. (1974). Peripheral astigmatism in the human eye: experimental data and theoretical model predictions. *J. Opt. Soc. Am.*, 64, 510–513.

Lowenstein, O. and Loewenfeld, I. E. (1969). The pupil. In *The Eye*. Vol. 3. ed. Davson, H. New York: Academic Press.

Ludlam, W. M., Wittenberg, S., Giglio, E. J. and Rosenberg, R. (1968). Accommodative responses to small changes in dioptric stimulus. *Am. J. Optom. Physiol. Opt.*, 45, 483–506.

Mandell, R. B. (1974). *Contact Lens Practice: Hand and Flexible Lenses*. 2nd edn. Springfield, IL: Thomas.

Mandell, R. B. and St Helen, R. (1971). Mathematical model of the corneal contour. *Br. J. Physiol. Opt.*, 26 183–197.

Mann, I. C. (1928). *The Development of the Human Eye*. p. 59. Cambridge: University Press.

Marshall, J. (1985). Radiation and the ageing eye. *Ophthalmol. Physiol. Opt.*, 5, 241–263.

Martola, E. L. and Baum, J. L. (1968). Central and peripheral corneal thickness: a clinical study. *Arch. Ophthalmol.*, 79, 28–30.

Maurice, D. M. (1957). The structure and transparency of the cornea. *J. Physiol. (Lond.)*, 136, 263–286.

Maurice, D. M. (1969). The cornea and sclera. In *The Eye*. Vol. 1. Davson, H. 2nd edn. New York: Academic Press.

Maurice, D. M. (1970). The transparency of the corneal stroma. *Vision Res.*, 10, 107–108.

Mellerio, J. (1971). Light absorption and scatter in the human lens. *Vision Res.*, 11, 129–141.

Mellerio, J. (1987). Yellowing of the human lens: nuclear and cortical contributions. *Vision Res.*, 27, 1581–1587.

Metcalf, H. (1965). Stiles-Crawford apodisation. *J. Opt. Soc. Am.*, 55, 72–74.

Miller, W. H. and Bernard, G. D. (1983). Averaging over the foveal receptor aperture curtails aliasing. *Vision Res.*, 23, 1365–1369.

Millodot, M. (1976). The influence of age on the chromatic aberration of the eye. *Albrecht v. Graefes Arch. Klin. Exp. Ophthal.*, 198, 235–243.

Millodot, M. (1981). Effect of ametropia on peripheral refraction. *Am. J. Optom. Physiol. Opt.*, 58, 691–695.

Millodot, M. and Lamont, A. (1974). Refraction of the periphery of the eye. *J. Opt. Soc. Am.*, 64, 110–111.

Millodot, M. and Newton, I. A. (1976). A possible change of refractive index with age and its relevance to chromatic aberration. *Albrecht v. Graefes Arch. OPhthalmol.*, 201, 159–167.

Millodot, M. and Sivak, J. (1973). Influence of accommodation on the chromatic aberration of the eye. *Br. J. Physiol. Opt.*, 28, 169–174.

Millodot, M. and Sivak, J. (1979). Contribution of the cornea and lens to the spherical aberration of the eye. *Vision Res.*, 19, 685–687.

Mordi, J. A. and Adrian, W. K. (1985). Influence of age on chromatic aberration of the human eye. *Am. J. Optom. Physiol. Opt.*, 62, 864–869.

Moses, R. A. and Hart, W. M. (1987). Eds. *Adler's Physiology of the Eye: Clinical Application*. 8th edn. St Louis: Mosby.

Nakao, S., Mine, K., Nishioka, K. and Kamiya, S. (1969). The distribution of refractive indices in the human crystalline lens. *Jap. J. Clin. Ophthalmol.*, 23, 41–44.

Navarro, R., Santamaria, J. and Bescos, J. (1985). Accommodation-dependent model of the human eye with aspherics. *J. Opt. Soc. Am. A*, 8, 1273–1281.

Norren, D. V. and Vos, J. J. (1974). Spectral transmission of the human ocular media. *Vision Res.*, 14, 1237–1244.

Obstfeld, H. (1978). *Optics in Vision*. London: Butterworths.

Ogboso, Y. U. and Bedell, H. E. (1987). Magnitude of lateral chromatic aberration across the retina of the human eye. *J. Opt. Soc. Am. A*, 4, 1666–1672.

Ogle, K. N. (1950). *Binocular Vision*. Philadelphia, PA: Saunders.

Ogle, K. N. (1960). Blurring of the retinal image and contrast thresholds in the fovea. *J. Opt. Soc. Am.*, 50, 307–315.

Ogle, K. N. (1968). *Optics*, 2nd edn. Springfield, IL: Thomas.

Ogle, K. N. and Schwartz, J. T. (1959). Depth-of-focus of the human eye. *J. Opt. Soc. Am.*, 49, 273–280.

Oshima, S. (1958). Studies on the depth of focus of the eye. *Jap. J. Ophthalmol.*, 2, 63–72.

Parker, J. A. (1972). Aspheric optics of the human lens. *Can. J. Ophthalmol.*, 7, 168–175.

Patnaik, B. (1967). A photographic study of accommodative mechanisms: changes in the lens nucleus during accommodation. *Invest. Ophthalmol.*, 6, 601–611.

Paulsson, L-E. and Sjostrand, J. (1980). Contrast sensitivity in the presence of a glare light. *Invest. Ophthalmol. Vis. Sci.*, 19, 401–406.

Pflibsen, K. P., Pomarentzeff, O. and Ross, R. N. (1988). Retinal illuminance using a wide-angle model of the eye. *J. Opt. Soc. Am. A*, 5, 146–150.

Pierscionek, B. K., Chan, D. Y. C., Ennis, J. P., Smith, G. and Augusteyn, R. C. (1988). Non-destructive method of constructing three-dimensional gradient index models for crystalline lenses: I Theory and experiment. *Am. J. Optom. Physiol. Opt.*, 65, 481–491.

Pokorny, J., Smith, V. C. and Lutze, M. (1987). Aging of the human lens. *Appl. Opt.*, 26, 1437–1440.

Polack, A. (1923). Le chromatisme de l'oeil. *Bull. Soc. Ophthalmol. (Paris)*, 36, 401–563.

Pomarantzeff, O., Dufault, P., and Goldstein, R. (1983). Wide-angle optical model of the eye. In *Advances in Diagnostic Visual Optics*. eds. Breinin, G. M. and Siegel, I. M. pp. 12–21. Berlin: Springer.

Pomarantzeff, O., Govignon, J. and Schepens, C. I. (1972). Wide-angle optical model of the eye. *Optica Acta*, 19, 387–388.

Prechtel, L. A. and Wesley, N. K. (1970). Corneal topography and its application to contact lenses. *Br. J. Physiol. Opt.*, 25, 117–126.

Reading, V. M. and Weale, R. A. (1974). Macular pigment and chromatic aberration. *J. Opt. Soc. Am.*, 64, 231–234.

Records, R. E. (1979). *Physiology of the Human Eye and Visual System*. Hagerstown: Harper and Row.

Rempt, F., Hoogerheide, J. and Hoogenboom, W. P. H. (1971). Peripheral retinoscopy and the skiagram. *Ophthalmologica*, 162, 1–10.

Röhler, R. (1962). Die Abbildungseigenschaften der Augenmedien. *Vision Res.*, 2, 391–429.

Röhler, R., Miller, U. and Aberl, M. (1969). Zur Messung der Modulationsubertragungs-Funktion des lebenden menschichen Auges in reflektierten Licht. *Vision Res.*, 9, 407–428.

Ronchi, L. and Millodot, M. (1974). The cortical counterpart of the chromatic aberration of the eye. *Am. J. Optom. Physiol. Opt.*, 51, 635–641.

Rosenblum, W. M. and Christensen, J. L. (1976). Objective and subjective aberration measurement of the human eye. In *Progress in Optics*. Vol. 13. ed. Wolf, E. pp. 69–91. Amsterdam: North Holland.

Ruddock, K. H. (1972). Light transmission through the ocular media and macular pigment and its significance for psychophysical investigation. *Handbook of Sensory Physiology. Vol. VIII/4*. eds.

Jameson, D. and Hurvich, L. Chapter 17. Berlin: Springer.

Said, F. S. and Weale, R. A. (1959). The variation with age in the spectral transmissivity of the living human crystalline lens. *Gerontologica*, 3, 213–231.

Sample, P. A., Esterson, F. D., Weinreb, R. N. and Boynton, R. M. (1988). The aging lens: *in vivo* assessment of light absorption in 84 human eyes. *Invest. Ophthalmol. Vis. Sci.*, 29, 1306–1311.

Santamaria, J., Artal, P. and Bescós, J. (1987). Determination of the point-spread function of human eyes using a hybrid optical-digital method. *J. Opt. Soc. Am. A*, 4, 1109–1114.

Sasaki, K., Shibata, T., Fukada, M. and Hockwin, O. (1980). Changes in lens transparency with aging. A clinical study with human volunteers using a Scheimpflug camera. In *Ageing of the Human Lens*. eds. Regnault, F., Hockwin, O. and Courtois, Y. Amsterdam: Elsevier.

Satoh, K., Bando, M. and Nakajima, A. (1973). Fluorescence in human lens. *Exp. Eye Res.*, 16, 167–172.

Schober, H., Munker, H. and Zolleis, F. (1958). Die aberration des menschilichen Auges und ihre Messung. *Optica Acta*, 15, 47–57.

Selwyn, E. W. H. (1948). The photographic and visual resolving power of lenses. Part 1. Visual resolving power. *Photogr. J. B*, 88, 6–12.

Simon, J. F. and Denieul, P. M. (1973). Influence of the size of test field employed in measurements of modulation transfer function of the eye. *J. Opt. Soc. Am.*, 63, 894–896.

Simonet, P. and Campbell, M. C. W. (1990). The optical transverse chromatic aberration on the fovea of the human eye. *Vision Res.*, 30, 187–206.

Sivak, J. G. (1982). The contribution of the crystalline lens to chromatic and spherical aberrations of the eye. *Can. J. Optom.*, 44, 89–91.

Sivak, J. G. and Mandelman, T. (1982). Chromatic dispersion of the ocular media. *Vision Res.*, 22, 997–1003.

Sliney, D. H. and Wolborsht, M. (1980). Safety standards and measurement techniques for high intensity light sources. *Vision Res.*, 20, 1133–1141.

Smirnov, M. S. (1962). Measurement of the wave aberration of the human eye. *Biofizikia*, 6, 687–703 and *Biophysics*, 6, 766–795.

Smith, G. (1982). Ocular defocus, spurious resolution and contrast reversal. *Ophthalmic Physiol. Opt.*, 2, 5–23.

Smith, G., Millodot, M. and McBrien, N. (1988). The effect of accommodation on oblique astigmatism and field curvature of the human eye. *Clin. Exp. Optom.*, 71, 119–125.

Smith, R. A. and Cass, P. F. (1987). Aliasing in the parafovea with incoherent light. *J. Opt. Soc. Am. A*, 4, 1530–1534.

Smith, W. J. (1978). Image formation: geometrical and physical optics. In *Handbook of Optics*. ed. Driscoll, W. G. and Vaughan, W. Section 2. New York: McGraw-Hill.

Snodderly, D. M., Auran, J. D. and Delori, F. C. (1984). The macular pigment. II Spatial distribution in primate retinas. *Invest. Ophthalmol. Vis. Sci.*, 25, 674–685.

Snyder, A. W. and Miller, W. H. (1977). Photoreceptor diameter and spacing for highest resolving power. *J. Opt. Soc. Am.*, 67, 696–698.

Sorsby, A. (1948). X-ray measurement of the diameters of the living eye. In *Modern Trends in Ophthalmology*. Vol. II. ed Sorsby, A. pp. 103–110. London: Butterworth.

Sorsby, A., Benjamin, B. and Bennett, A. G. (1981). Steiger on refraction: a reappraisal. *Br. J. Ophthalmol.*, 65, 805–811.

Sorsby, A., Benjamin, B., Davey, J. N., Sheridan, M. and Tanner, J. M. (1957). Emmetropia and its aberrations. *Spec. Rep. Ser. Med. Res. Counc.*, 293.

Spector, A., Roy, D., and Stauffer, J. (1975). Isolation and characterisation of an age-dependent polypeptide from human lens with non-tryptophan fluorescence. *Exp. Eye Res.*, 21, 9–24.

Stanworth, A. and Naylor, E. J. (1955). The measurement and clinical significance of the Haidinger effect. *Trans. Ophthalmol. Soc. UK*, 75, 67–79.

Steindorff, K. (1947). Deskriptive Anatomie des Auges der Wirbeltiere und des Menschen. *Tabul. Biol. Berl.*, 22(1), 166–297.

Stenstrom, S. (1946). Untersuchungen über die Variation und Kovariation der optischen Elements des menschlichen Auges. *Acta ophthal. Kbh. (Suppl.)*, 26.

Stiles, W. S. (1929). The effect of glare on the brightness threshold. *Proc. R. Soc. Lond. B*, 104, 322–351.

Stiles, W. S. and Crawford, B. H. (1933). The luminous efficiency of rays entering the eye pupil at different points. *Proc. R. Soc. Lond. B*, 112, 428–450.

Thibos, L. N. (1987). Calculation of the influence of lateral chromatic aberration on image quality across the visual field. *J. Opt. Soc. Am. A*, 4, 1673–1680.

Thibos, L. N., Bradley, A., Still, D. L., Zhang, X. and Howarth, P. A. (1990). Theory and measurement of ocular chromatic aberration. *Vision Res.*, 30, 33–49.

Thibos, L. N., Cheney, F. E. and Walsh, D. J. (1987). Retinal limits to the detection and resolution of gratings. *J. Opt. Soc. Am. A*, 4, 1524–1529.

Thompson, D. A. and Drasdo, N. (1989). The effect of 0.5% thymoxamine on the pattern-onset electroretinogram. *Doc. Ophthalmol.*, 72, 47–54.

Thompson, H. S. (1987). The pupil. In *Adler's Physiology of the Eye: Clinical Application*. eds. Moses, R. A. and Hart, W. M. Chapter 12. St Louis: Mosby.

Tucker, J. and Charman, W. N. (1975). The depth-of-focus of the human eye for Snellen letters. *Am. J. Optom. Physiol. Opt.*, 52, 3–21.

Van den Brink, G. (1962). Measurements of the geometrical aberrations of the eye. *Vision Ress.*, 2, 233–244.

Van Meeteren, A. (1974). Calculations on the optical modulation transfer function of the human eye for white light. *Optica Acta*, 21, 395–412.

Van Nes, F. L. and Bouman, M. A. (1967). Spatial modulation transfer in the human eye. *J. Opt. Soc. Am.*, 57, 401–406.

Von Bahr, G. (1952). Studies on the depth of focus of the eye. *Acta Ophthalmol.*, 30, 39–44.

Vos, J. J. (1963). Contribution of the fundus oculi to entoptic scatter. *J. Opt. Soc. Am.*, 53, 1449–1451.

Vos, J. J. and Boogaard, J. (1963). Contribution of the cornea to entoptic scatter. *J. Opt. Soc. Am.*, 53, 869–873.

Vos, J. J. and Bouman, M. A. (1964). Contribution of the retina to entoptic scatter. *J. Opt. Soc. Am.*, 54, 95–100.

Vos, J. J. and Van Os, F. L. (1975). The effect of lens density on the Stiles-Crawford effect. *Vision Ress.*, 15, 749–751.

Vos, J. J., Walraven, J. and Van Meeteren, A. (1976). Light profiles of the foveal image of a point source. *Vision Res.*, 16, 215–219.

Wald, G. (1949). The photochemistry of vision. *Doc. Ophthalmol.*, 3, 94–137.

Wald, G. and Griffin, D. R. (1947). The change in refractive power of the human eye in dim and bright light. *J. Opt. Soc. Am.*, 37, 321–336.

Walsh, G. and Charman, W. N. (1985). Measurement of the axial wavefront aberration of the human eye. *Ophthalmic Physiol. Opt.*, 4, 23–31.

Walsh, G., Charman, W. N. and Howland, H.C. (1984). Objective technique for the determination of monochromatic aberrations of the human eye. *J. Opt. Soc. Am. A*, 1, 987–992.

Wang, G-J., Pomerantzeff, O. and Pankratov, M. M. (1983). Astigmatism of oblique incidence in the human model eye. *Vision Res.*, 23, 1079–1085.

Ware, C. (1982). Human axial chromatic aberration found not to decline with age. *Albrecht v. Graefes Arch. Ophthalmol.*, 218, 39–41.

Watkins, R. D. (1972). *A Finite Aperture Model of the Optical System of the Human Eye*. Flinders University, Australia. PhD thesis.

Weale, R. A. (1956). Problems of peripheral vision. *Br. J. Ophthalmol.*, 40, 392–415.

Weale, R. A. (1961). Notes on the photometric significance of the

human crystalline lens. *Vision Res.*, 1, 183–191.

Weale, R. A. (1963). *The Aging Eye*. London: H. K. Lewis.

Weale, R. A. (1974). Natural history of optics. In *The Eye*. Vol. 6. eds. Davson, H. and Graham, L. T. pp. 1–110. London: Academic Press.

Weale, R. A. (1979). Sex, age and the birefringence of the human crystalline lens. *Exp. Eye Res.*, 29, 449–461.

Weale, R. A. (1982). *A Biography of the Eye*. London: Lewis.

Werner, J. S. and Wooten, B. R. (1980). Age changes in ocular media density and consequences for colour vision. In *Colour Vision Deficiencies, Vol. V.* pp. 355–359. Bristol: Adam Hilger.

Westheimer, G. (1960). Modulation thresholds for sinusoidal light distributions on the retina. *J. Physiol. (Lond.)*, 152, 67–74.

Westheimer, G. (1964). Pupil size and visual resolution. *Vision Res.*, 4, 39–45.

Westheimer, G. (1966). Focusing responses of the human eye. *Am. J. Optom. Arch. Am. Acad. Optom.*, 43, 221–232.

Westheimer, G. and Campbell, F. W. (1962). Light distributions in the image formed by the living human eye. *J. Opt. Soc. Am.*, 52, 1040–1045.

Williams, D. R. (1985). Aliasing in human foveal vision. *Vision Res.*, 25, 195–205.

Williams D. R. (1988). Topography of the foveal cone mosaic in the living human eye. *Vision Res.*, 28, 433–454.

Woodhouse, J. M. (1975). The effect of pupil size on grating detection at various contrast levels. *Vision Res.*, 15, 645–648.

Woodhouse, J. M. and Campbell, F. W. (1975). The role of the pupil light reflex in aiding adaptation to the dark. *Vision Res.*, 15, 649–653.

Wright, W. D. (1949). *Photometry and the Eye*. London: Hatton Press.

Wyszecki, G. and Stiles, W. S. (1967). *Colour Science*. p. 214–219. New York: Wiley.

Yellot, J. I. (1982). Spectral analysis of sampling by photoreceptors, topographical disorder prevents aliasing. *Vision Res.*, 22, 1205–1210.

Young, T. (1801). On the mechanism of the eye. *Phil. Trans. R. Soc. B.*, 92, 23–88.

Zinn, K. M. (1972). *The Pupil*. Springfield, Ill: Thomas.