Pseudophakic Implants, Aspherical Optics, Quality of Vision for Cataract Patients

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Introduction

Vision is a complex phenomenon involving a sequence of events that starts with the eye capturing photons emitted by an object and culminates in a visual sensation generated by the activation of specialised neuronal structures in the occipital cortex.

The first stage in this process may be described as optical since it involves the refraction of light waves through the various optical structures of the eye (lachrymal film, cornea, lens, vitreous body) until they reach the photoreceptors on the retina. This stage is important since it conditions the quality of vision for subjects with no visual pathway pathology.

Quality of vision is dependent on a number of aspects of the visual function, such as light sensitivity, colour vision, differential sensitivity (perception of contrast), etc.; the combined performance of these various aspects determines the overall quality of the retinal image. Both low-order and high-order optical aberrations impair quality of vision, as can the loss of transparency of the ocular media. The clinical definition of a cataract is the development of opacity (or cloudiness) in the cortex and/or nucleus of the lens. This opacity results from the breakdown of the cortical or nuclear fibres of the lens, which increases the absorption and diffusion of light. Changes to the indexes and geometry of the crystalline lens also lead to the development of optical aberrations, the most typical example of which is index myopia. It has also been demonstrated that these changes to the lens can induce an increase in positive ocular spherical aberration.

Artificial intraocular lens replacements for the natural lens can restore the transparency of the media and correct the hypermetropisation induced by the aphakia. However, all the studies conducted on pseudophakic subjects have shown that the spherical geometry of classical implants does not offer optical quality equivalent to that measured for young, phakic subjects with no lens opacity.

This is because the young lens partially compensates for the positive spherical aberrations caused by the cornea by inducing a negative spherical aberration. The insertion of a spherical optic after ablation of a cataracted lens eliminates this compensation mechanism, and may even accentuate the total positive spherical aberration (through the addition of the spherical aberration of the implant to that of the cornea). Studies conducted on older cataract patients fitted with a spherical posterior chamber implant found that their spherical aberration was greater than that of patients of a similar age who had not been operated on for cataract (comparison with the degree of pre-operative spherical aberration is generally compromised by the very lens opacity that prompted the surgery since it impairs the reliability of the pre-operative measurements). In addition to the increase in spherical aberration, we have recently identified an increase in coma-type asymmetric optical aberrations following the insertion of spherical pseudophakic implants into the posterior chamber which may result from lens decentration or tilt. In summary, a biconvex spherical implant of appropriate power can restore the transparency of ocular media and compensate for post-operative refraction, but is not designed to correct (and may even accentuate) certain high-order optical aberrations such as positive spherical aberrations. This aberration is particularly detrimental to the sensitivity to contrast when the pupil diameter is greater than 4.5 mm.

New manufacturing processes and the use of eye models that can incorporate specific physiological conditions (mesopic vision inducing a large pupil diameter) have recently been applied to the development and marketing of pseudophakic implants with aspherical surfaces. The principal anticipated benefit is improved retinal image quality for pseudophakes, which must result in better visual acuity and better perception of contrasts in mesopic and scotopic environments than they would experience with a spherical lens.

Good sensibility to contrasts guarantees better adaptation to scotopic conditions (beneficial for activities such as driving at night, or moving around in dimly-lit areas).

The sensitivity of the optical quality of a thin implant to decentration tends to increase (if the optic is aspherical) and to induce negative spherical aberrations. Consequently, above a certain degree of decentration or tilt, the asphérisation of an implant may prove to be detrimental, resulting in poorer optical quality than a spherical implant of the same central power positioned identically. It is therefore important to exclude some patients who present with capsular pseudo exfoliation or in the event of zonular deinsertion or preoperative capsular rupture. Benefits may also be drawn from refining the theoretical model used to design lens implants to take advantage of the physiological decentration of the pupil and foveal eccentricity.

Recent progress in the fields of instrumental optics and computing has opened up new opportunities for the detailed study of the optical properties of the human eye. The best illustration of this is the use of aberrometers in routine clinical ophthalmology. Aberrometry is a modern medical science concerned with studying the optical properties of the human eye. It is now possible to study in detail the optical properties of isolated ocular structures such as the cornea and some internal optical structures (crystalline lens).

Aberrometry lets the ophthalmologist qualify and quantify, in a context of enhanced treatment delivery, acquired or congenital optical imperfections that cannot be corrected by spectacles (since these only correct so-called "low-order" aberrations: myopia, hypermetropia, and astigmatism). Once the aberrometer has recorded accurately and quantitatively these various "high-order" aberrations, a comprehensive analysis of the optical properties of the eye may be performed, such as a calculation of the theoretical best corrected and uncorrected visual acuity, a prediction of the optical component of the sensitivity to contrasts, etc.

Ophthalmologists often encounter an understandable difficulty getting to grips with the theoretical principles, whose origins are in physical optics, and the terminology that may be unfamiliar. The technical terms used in aberrometry are in fact the same as those used by astronomers and engineers to characterise the optical quality of telescopes and other optical instruments. Becoming familiar with these concepts nonetheless remains essential for a full understanding of the principles and results of the aberrometric examination. The main purpose of this article is to describe these concepts.

I. Basic principles of optics relevant to the study of quality of vision

1. Stigmatism and optical path

Ophthalmologists conventionally show the propagation of light through a lens as "rays" whose path obeys the laws of geometrical optics. With this representation, the lens is able to focus light by transforming an incident beam of parallel rays (from a distant luminous point source) into a refracted beam that converges at the focal point of the lens.

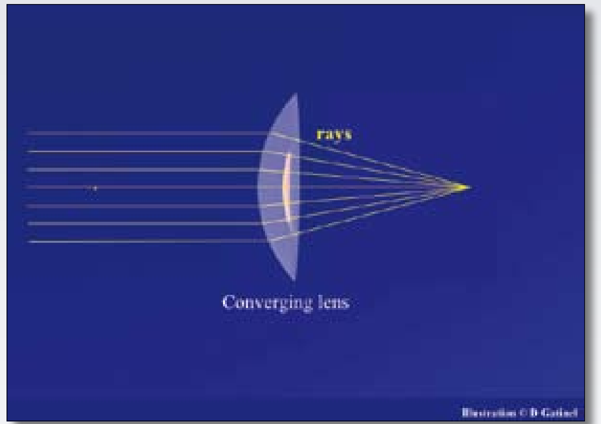


Figure.1

Figure.1: *Convex plano lens to focus light rays.*

The power of the lens and its aberrations can be predicted by calculating the path of the rays through the lens (ray tracing). A wave model for the propagation of light can also be considered.

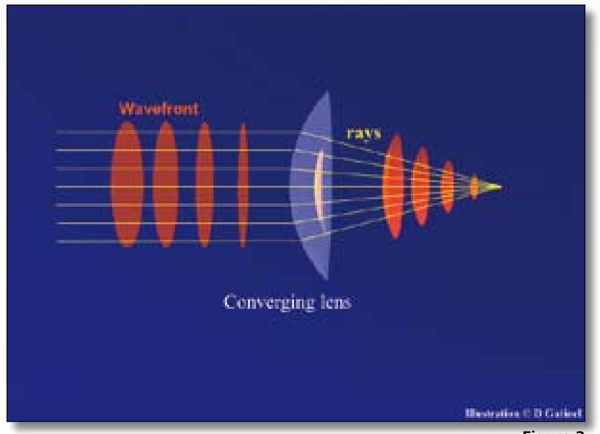


Figure.2

Figure.2: *Wavefront light propagation: the wavefront envelope and ray path are mutually perpendicular.*

This approach is useful in that it reveals aspects that are valuable when evaluating the optical quality of an eye, such as the optical aberrations of the wavefront and diffraction. Knowledge of a few principles of physical optics is a prerequisite to the understanding of this wave model of light propagation.

1.1 Propagation of light waves

A few basic principles govern the path of light waves through transparent media such as the ocular tunics. These principles result from the interaction between the electromagnetic field corresponding to the propagated light radiation and the molecular structures of the media through which the light passes.

- A light source emits an electromagnetic signal that propagates by oscillating between two extreme values of electromagnetic field strength. Its direction of propagation can be exemplary constituted by rays of light. The surface of the wavefront is perpendicular to these rays. The distance between any two successive peaks defines the wavelength of the electromagnetic radiation.

- The set of points in space that have the same phase for the electromagnetic field at a given time defines the surface of the wavefront. A spherical wavefront is emitted by a point source that is embedded in an isotropic media.

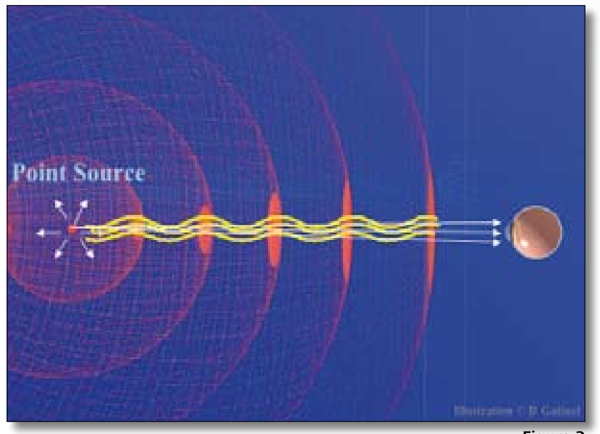


Figure.3

Figure.3: *The wavefront (WF) joins the emerging light waves at a single point in time. The light rays travel perpendicular to the WF at all points.*

- A wavefront entering a transparent material whose refractive index is greater than 1 slows down, with the velocity of propagation decreasing as the index increases. The wavelength of the light radiation under consideration also has an impact on the velocity of propagation through a given medium. This phenomenon is called material dispersion and is the cause of chromatic aberrations in image formation of polychromatic sources.

- When, at a given point in space, several electromagnetic wave surfaces "superimpose" on each other with no significant phase difference, the amplitude of oscillation of the field increases at this point by summation, resulting in an increase in light intensity. For this phenomenon to occur, the "optical path length" from the point source to this image point must be the same (accurate to an integer multiple of wavelengths) for all the ray paths.

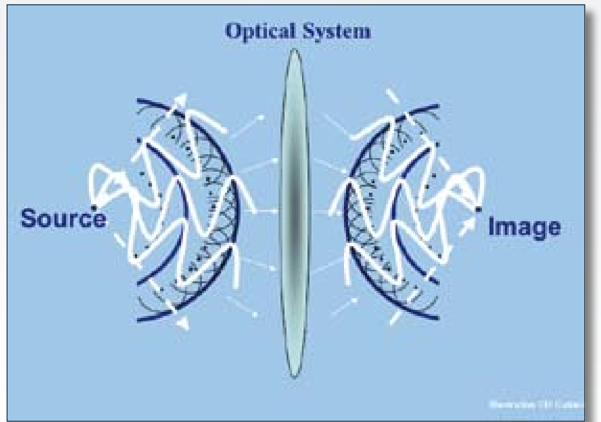


Figure.4

Figure.4: *A propagating wavefront of light can be defined by the focus of the points lying at the same optical path from the source. When the optical path length is the same for all the rays emitted by a source, they interfere constructively to produce a sharp image of the source. This is the consequence of the Huygens-Fresnel's principle of elementary wavelets (shown in dark blue).*

- The formation of the new wavefront propagation can be explained by the Huygen-Fresnel principle of elementary wavelets. This principle states that any point on a wavefront of light may be regarded as the source of secondary waves and that the surface that is tangential to the secondary waves can be used to determine the future position of the wavefront. When a wavefront encounters an obstacle or opening then diffraction occurs; the obstacle or opening tends to form a secondary source for the wavefront, which affects the direction of propagation of the emitted or diffracted wavefront.

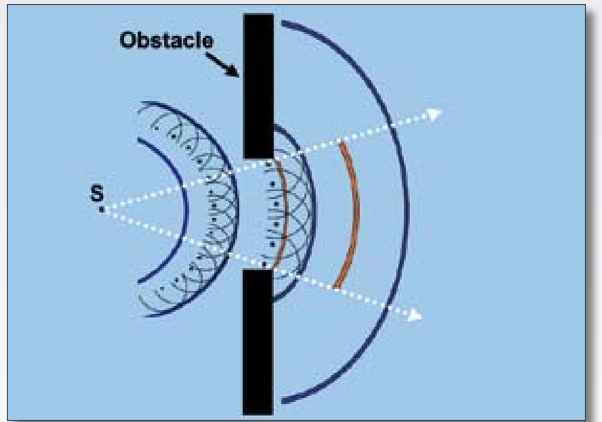


Figure.5

Figure.5: *Light is diffracted in directions not predicted by geometrical optics. When a propagating wavefront of light encounters an obstacle with a small hole, the Huygen-Fresnel principle states that its propagation will not be confined to the sectorial space bounded by the open angle from the source and the edges of the aperture source (orange wavefront portion). Rather, it will propagate through nearly all the space behind the aperture (blue propagating wavefront). This is due to the fact that points located near the edge of the aperture act like single source points. Hence, when the aperture becomes very small, it acts as a secondary point source.*

These principles apply also to non-uniform media; e.g., the propagation of a wavefront through a partially opaque lens will be altered by a combination of refractive and diffractive phenomena, whose significance will vary depending on the wavelength of the light.

1.2 Stigmatism

The eye provides living organisms with a visual representation of their environment. The fidelity of this picture depends on a number of variables that may be separated into two main categories: the first comprises those variables that characterise the quality of the optical elements (cornea, lens), and the second consists of the system for receiving and transmitting visual information (photoreceptors, visual pathways). Stigmatism is a fundamental component of the performance of an optical system and it is worth spending some time on this important property. In simple terms, stigmatism is the ability of the optical system to create a representation of a point source that is as much like a point as possible. For an eye, the collection plane for this image must correspond to that of the photoreceptors of the fovea. An observed image is rarely a point source. It may, however, be considered to be a set of elemental point sources, and the stigmatism must comprise a sufficient number of these points. The eye must therefore have a certain axial and lateral depth of field. For the human eye, the density of photoreceptors (which determines the maximum resolution) decreases rapidly away from the fovea. As a good approximation it may be assumed that if the image of the points located on the optical axis is a point in the foveolar plane, then all the neighbouring points will also be created as points. For this simplified approach we have deliberately ignored diffraction, which limits the stigmatism of any optical system, and we have considered only a monochromatic system. Note, however, that under optimal conditions for stigmatism (absence of optical aberration, large pupil diameter to limit diffraction) the diameter of a cone in the fovea (about two microns) captures the diameter of the light spot formed from an elemental point source. The maximum sensitivity of the human eye for polychromatic radiation is found in the green spectrum, at about 555nm for photopic vision. For scotopic vision the maximum shifts to 507nm towards blue.

1.3 Optical path

Consider a lens that offers excellent stigmatism and let us ignore once again any diffraction phenomena. Such a lens allows parallel incident rays emitted by a distant point source to converge at a single point at the focal point of the lens. Since the rays emitted by such a source arrive in a parallel beam at the lens, the corresponding incident wavefront (perpendicular to these rays) is planar before its meets the surface of the lens. Since the lens is perfectly stigmatic, it must transform a planar wavefront that crosses through it into a perfectly spherical wavefront that is centred at its focal point (Figure.2). In this case, the spherical wave surfaces converge towards this point where they arrive in phase. The refractive index of a transparent material is proportional to the velocity of propagation at which light waves pass through the material. As light passes through a lens, its wavelength is shortened (although the frequency of oscillation of the electromagnetic field is unchanged). The optical path corresponds to the total number of oscillations performed by the light during its journey from the point source to the image point. Stigmatism and preservation of the optical path are closely related. In order to obtain stigmatism, the optical path length between the source and the image must be identical for all ray paths.

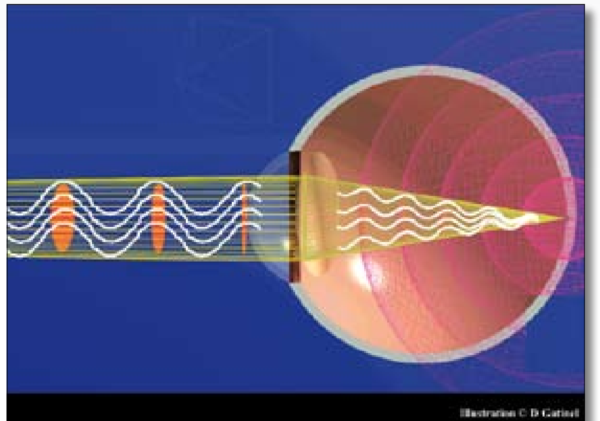


Figure.6

Figure.6: *Stigmatism (yellow rays), wavefront (orange and red surfaces) and optical path length (white oscillating waves). The optical path is identical for all the rays: all the light waves will arrive, in phase, at the image plane to form a perfect ("diffraction limited") image.*

In other words, the geometrical design of the lens must have been calculated such that, after passing through the lens, the points in the electromagnetic wave that are in phase are distributed over a spherical wave surface whose centre coincides with the focal point of the lens. Or, alternatively, the optical path length travelled by the light waves from the point source must be identical for all the waves refracted by the lens. For a given ray, the optical path for a stigmatic lens may be divided into three sections: from the source to the anterior face of the lens, inside the lens, and from the posterior face of the lens to the image point.

1.4 Spherical aberration

At a given instant in time, all points on a wavefront that encounter the anterior face of a curved lens are not in phase since, for a convex lens, the path travelled from the light source lengthens slightly with increasing distance from the optical axis.

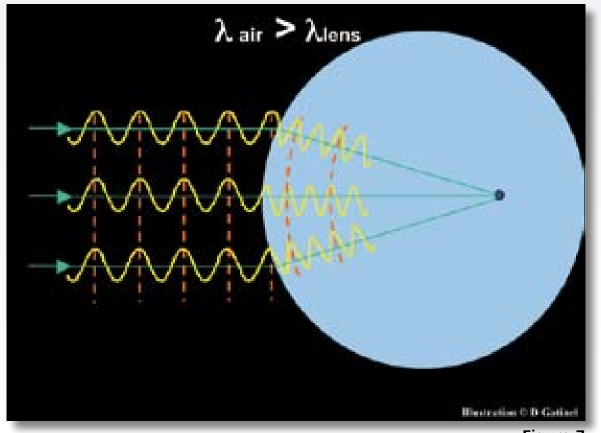


Figure.7

Figure.7: *If the optical distance taken by every ray travelling through the lens is the same, then all the rays will arrive at the image plane in phase to form a perfect image. All the points on a wavefront that encounter the anterior face of a curved lens are not in phase since, for a convex lens, the path travelled from the light source lengthens slightly with increasing distance from the optical axis. Due to this optical path difference, the peripheral rays may not interfere in phase with the central ones at the lens focal plane.*

This phase difference becomes greater as the width of the incident beam increases (i.e. as it includes points located further away from the optical axis of the lens). If the geometry of the lens is not adjusted to compensate for this phase difference then image collected at the focus will not be a point but will be dispersed in a concentric pattern. The spherical aberration at the focal point of the lens is thus a result of the phase difference between the lights waves that pass through the central portion compared with those from the edge of the wavefront (due to the difference in optical path length). Due to the geometrical optics involved, the light rays refracted at the periphery do not converge at the same point on the optical axis as those refracted at the centre of the lens.

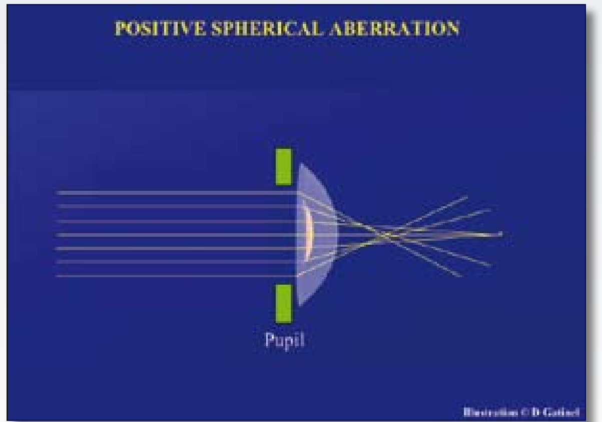


Figure.8

Figure.8: *Positive spherical aberration of a lens. The peripheral rays are refracted in front of the central rays.*

For a broad incident beam that encounters an optical system consisting of a single lens, the spherical aberration may be eliminated by designing lenses with aspherical surfaces.

2. Principles of optical aberration compensation

2.1 Capture and reconstruction of the wavefront
The map of high-order aberrations provided by the Shack-Hartmann aberrometer is effectively a record of differences in optical path length. The device emits a wavefront (through a small aperture which ensures that any aberrations associated with the path into the eye are negligible) that reflects off the fovea and is captured outside the eye after passing through all the optical elements of the eye and having been "diaphragmed" by the pupil. An array of small lenses, called lenslets, in the aberrometer all with identical focal lengths refract the captured wavefront and divide it into elemental portions.

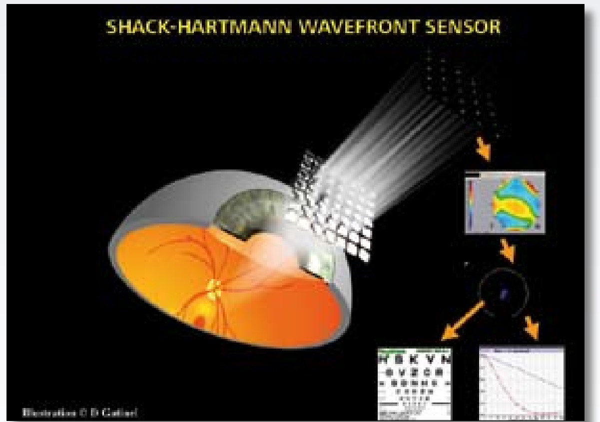


Figure.9

Figure.9: *Monochromatic light reflects off the fovea and is captured outside the eye after passing through all the optical elements of the eye. An array of lenslets refract the captured wavefront and divide it into elemental portions. These portions are focused onto a CCD sensor. From the deviation of the spots, the higher order wavefront is reconstructed. From the wavefront shape, different metrics can be calculated.*

Each of these portions is focused onto a CCD sensor, which records the offset of the "focal spot" from the optical axis of the lenslet.

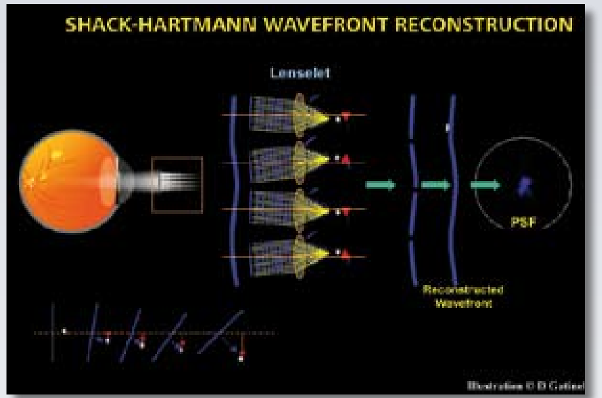


Figure.10

Figure.10: *The deviation of the spots of the focal lenslets (red arrows) is proportional to the slope of the wavefront. Measuring the distance of the focal spot to the optical axis on the lenselet allows to measure the local slope of the wavefront.*

If the eye does not induce any aberrations, the optical path length is identical at all points across the plane of the pupil. Indeed, in this case, the fovea is at the focal point of the ocular optical system, and the emerging wavefront is a flat disk parallel to the plane of the pupil (the emerging rays, perpendicular to the plane of the pupil, are "parallel" to each other). Each portion of this planar wavefront will itself be a plane and will be refracted identically and along the optical axis of each lenslet. If the eye does produce aberrations, the wavefront will be distorted (the points in phase will not be distributed over a plane, and the emerging rays, locally perpendicular to the wavefront, will no longer be absolutely parallel to each other). Consequently, the lenslets will focus each portion of the wavefront differently (off axis) since they form an angle with the axis of the lenslet. The distance separating the centre of the focal spot produced and the optical axis of the lenslet is proportional to the inclination of the incident portion of the wavefront (Figure.10).

It is then possible to reconstruct the surface of the wavefront by combining the various individual "facets" that form the wavefront. Remember that these facets correspond to phase differences generated by differences in optical path length for various points across the plane of the pupil. In order to produce an intelligible representation, the surface of the wavefront is broken down into a sum of elemental terms (Zernike's polynomials), each of which is weighted by a coefficient.

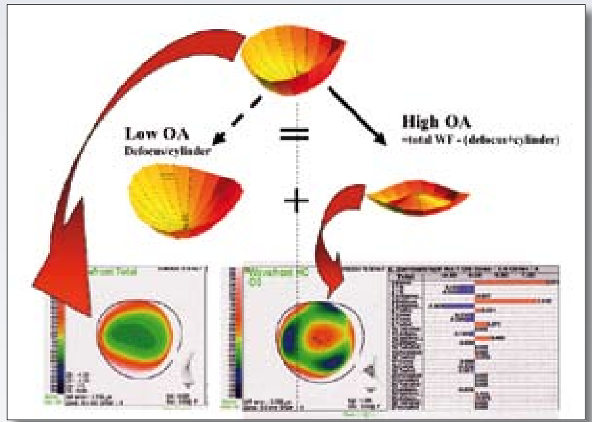


Figure.11

Figure.11: *The presence of a spherocylindrical ametropia dictates the overall shape of the wavefront, because the rate of high order aberration is usually much lower. In emmetropic patients, the central portion of the wavefront is usually flat and distorsions prevail at the edges of the pupil. The mathematical extraction of higher order aberrations allows to visualize the isolate effects of these aberrations. The contribution of higher order aberrations to the wavefront distorsion is better visualized on the higher order wavefront map, where first and second order aberrations are removed. Coma and trefoil induce an asymmetric distorsion of the wavefront envelope. Spherical aberration induces a distorsion of the central area of the wavefront relative to its edges. A purely spherically altered wavefront would have a "sombrero" shape. In some particular conditions, asymmetry can be visible on the total wavefront map as an effect of the presence of a large amount of higher order aberrations, as in the case shown.*

These terms are "orthogonal", and offer a means of describing a given wavefront in a unique manner. They correspond to the "classical" optical aberrations: tilt, spherical and cylindrical defocus (astigmatism), coma, trefoil, spherical aberrations, etc.

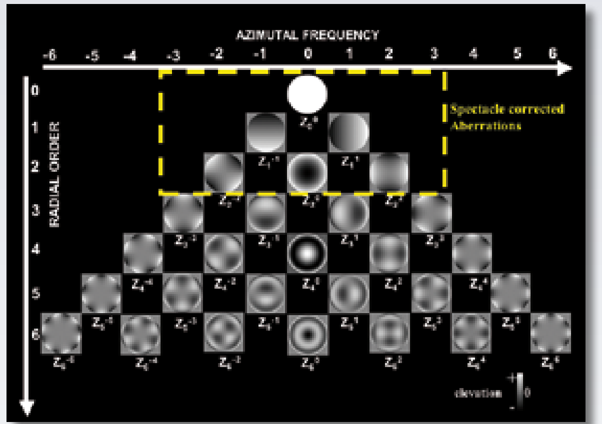


Figure.12

Figure.12: *The first Zernike polynomials correspond to classical optical aberrations (prism, cylindrical and spherical defocus).*

The value of the coefficient that weights a given term (e.g.: 0.1 micron) is proportional to the significance of the aberration in the wavefront under consideration. Since they represent a phase difference, i.e. a difference between the "peaks" and "troughs" of the electromagnetic radiation at a given time, these aberrations may be quantified by a number corresponding to a distance. In view of the magnitude of the wavelengths of visible light, this unit is the micron (a λ number for phase difference wavelength is also used in instrumental optics). For a given aberration term, the RMS (root mean square) value expressed in microns corresponds to the mean value of the squares of the differences measured between the aberration and the ideal wavefront.

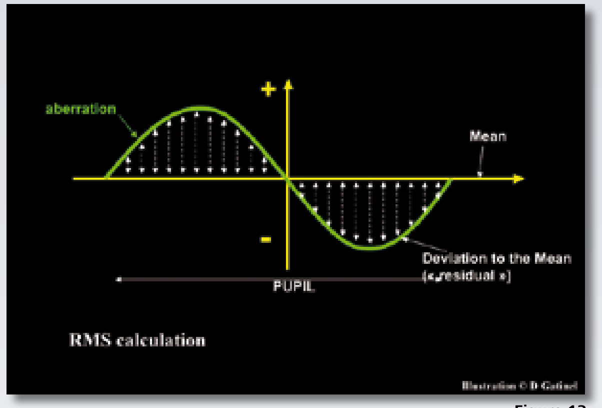


Figure.13

Figure.13: *An aberration corresponds to a wavefront distorsion. The RMS calculation is expressed in microns and corresponds to the square root of the mean value of the squared residuals.*

Considering the squares of the deviations and not their absolute value avoids having to compensate for positive (phase advance) and negative (phase lag) phase differences.

2.2 Effect of optical aberrations on the wavefront

An eye that does not produce any optical aberrations "transforms" a planar wavefront into a spherical wavefront, centred on the fovea. A spherical wavefront transmitted from the fovea through an eye that does not generate any optical aberrations emerges as a flat disk whose periphery corresponds to the periphery of the eye's optical exit pupil. The resolving power of such a system for monochromatic light only appears to be limited by diffraction. An eye that is only affected by a pure spherical ametropia (defocus) would "transform" a planar incident wavefront into a spherical wavefront; which would not be centred on the fovea.

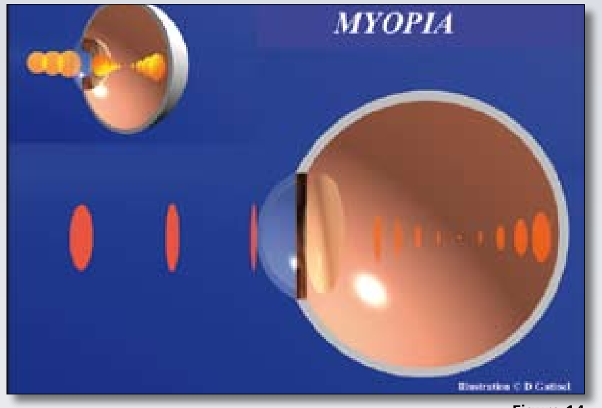


Figure.14

Figure.14: *A myopic eye transforms a planar wavefront in a spherical wavefront centred on a point located in front of the fovea.*

A spherical wavefront transmitted from the fovea through an eye that is affected only by myopic defocus emerges as a perfectly spherical dome, whose periphery corresponds to the periphery of the eye's pupil. The centre of curvature of the dome coincides with the punctum remotum of the myopic eye.

In reality, every eye, even an "emmetropic" eye, has some high-order optical aberrations that primarily take the form of coma, trefoil and spherical aberrations. Remember that the human eye is made up of aspherical optical surfaces that do not have a common optical axis, which is the reason why there are always some aberrations. These aberrations may therefore be characterised by analysis of the deformation of the wavefront, and are expressed in microns or units of phase difference wavelength compared with a "normal" theoretical planar wavefront.

Spherical aberration produces a characteristic distribution of focal spots refracted by the lenslets of an aberrometer. In fact, since the waves that are far from the centre of the wavefront have experienced a certain phase difference, the wavefront there is no longer planar but curved: for this reason, the focal spots produced by the lenslets around the peripheries tend to be similar to those produced by the lenslets at the center.

Coma-type aberration induces an asymmetrical distortion of the peripheral focal spots produced by the lenslets.

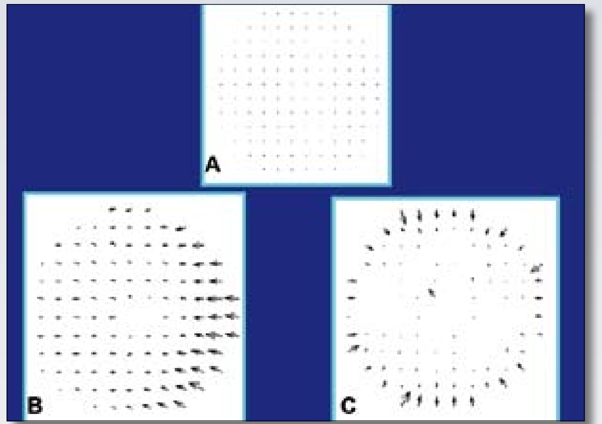


Figure.15

Figure.15: *Examples of CCD video of a Shack-Hartmann wavefront sensor after focusing of the array of lenslets of different altered wavefronts:*
A: diffraction limited eye
B: coma
C: spherical aberration

For an eye that is healthy and emmetropic, these aberrations primarily deform the "periphery" of the wavefront, whose central portion remains relatively flat; this explains why, when the pupil diameter is between 2 and 3 mm, the eye can appear to be a perfect optical system. If the eye is not emmetropic, the overall curvature of the wavefront is spherical (the role of defocus), however, the aberrations cause a peripheral distortion. After conventional refractive surgery, or the implantation of a spherical pseudophakic implant, the degree of spherical-type aberrations and coma increases significantly compared with the physiological level measured for a young, phakic subject, and in proportion to the pupil diameter.

II. Asphericαl surfaces and spherical aberration

1. Asphericity

1.1 Definition

Curvature is the amount by which a geometric object deviates from being flat. The primordial example is that of a circle which has curvature equal to the inverse of its radius everywhere. Smaller circles bend more sharply, and hence have higher curvature. The curvature of a smooth curve is defined as the curvature of the circle that “kisses” or just touches the curve at the given point of interest (osculating circle). The radius of curvature of a curve at a point is the radius of the osculating circle at that point.

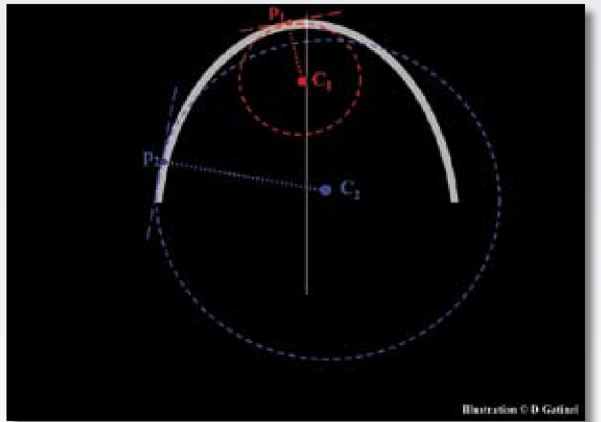


Figure.16

Figure.16: *The curvature of a smooth curve (thick grey line) is defined at any point p by the inverse of the value of the local osculating circle of center C. In this example, the curvature is greater at p1 than at p2.*

Curvature of two-dimensional surfaces curved in a three-dimensional space (such as the IOL surface) can be measured along a plane perpendicular to the surface at the point of interest. Only with a sphere is the radius of curvature equal to the radius of a sphere at every point of the spherical surface.

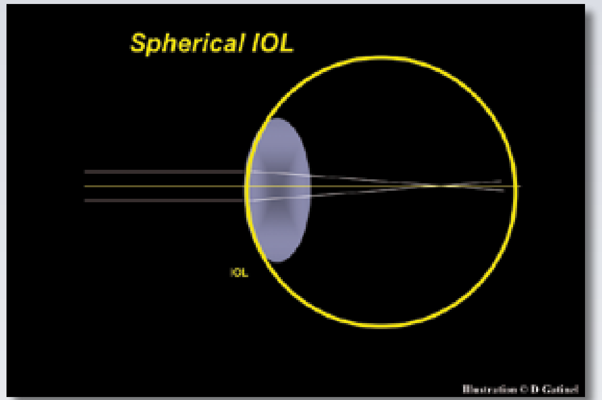


Figure.17

Figure.17: *The surface of the anterior (and posterior) profiles are spherical: in cross-section, the entire profile of the IOL surface is perfectly circular. The paraxial power of the lens for an object located at infinity is determined from the values of the anterior and posterior apical radii of curvature, and from its refractive index.*

An optical surface is said to be aspherical when its contour cannot be overlaid on a sphere (i.e. where the profile of the meridians of the surface are not circles). Unlike a sphere, the curvature of an aspherical surface is not identical at all points on the surface.

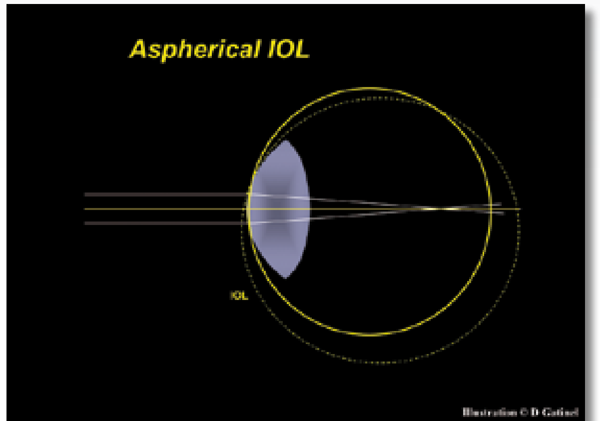


Figure.18

Figure.18: *Aspherical profile geometry : the entire profile of the IOL is no longer spherical. The IOL curvature (defined at every point as the reciprocal of the radius of the local “osculating circle”) varies from the center to the edges. In this diagram, the anterior surface of the IOL is aspherical; the local radius of curvature increases (the curvature decreases) toward the periphery.*

Although this definition suggests a certain simplicity, more complex concepts are required to describe how the aspherical surface deviates from a spherical surface.

The asphericity of a surface may be defined by a variable, usually called Q. This variable describes how the curvature of a curve or surface varies from its central portion out to its periphery. This variation confers special optical properties to aspherical surfaces. Q can be defined from the geometry of a simple aspheric curve such as an ellipse.

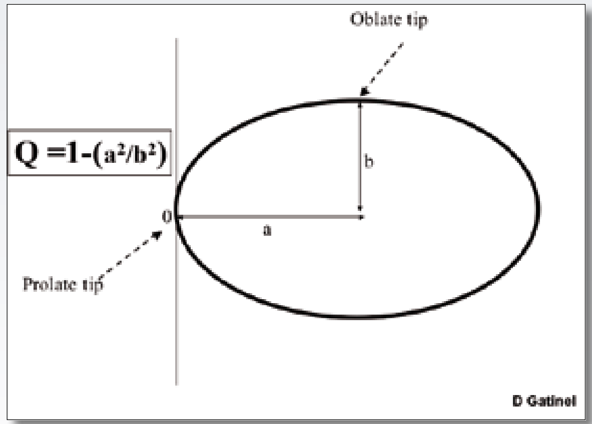


Figure.19

Figure.19: *An ellipse is an aspherical curve; the value of the asphericity (Q) of the prolate tip can be computed from the value of the ellipse’s semi major axis.*

Asphericity is a geometrical concept and is not specifically related to optics. The asphericity of an optical structure is not sufficient by itself to predict all aspects of its optical behaviour, which depends also on its apical curvature, its refractive index and the distance to the viewed object. The optical behaviour of an aspherical lens varies depending on whether the rays in the incident beam that cross the lens are parallel or not. For an intraocular implant this beam is convergent as a result of its refraction by the cornea.

1.2 Characterisation of an aspherical surface

To start with, let us consider the profile of an aspherical surface. The profile of an aspherical surface may be described with reference to one of several families of curves. The family of simple or figurate conic sections are the curves most commonly used.

1.2.1 Conic sections

The family of conic sections (ellipse, circle, parabola, and hyperbola) are the model most commonly used to describe the profile of the anterior face of the cornea, since they provide a more accurate representation of reality than the spherical model, and particularly for the central 8 mm. These sections have also been used to describe the asphericity of the anterior and posterior faces of the crystalline lens, and are used more generally to establish the profiles of the surfaces used for models of theoretical eyes or aspherical implants.

As their name implies, conic sections are formed by the intersection of the surface of a cone and a plane.

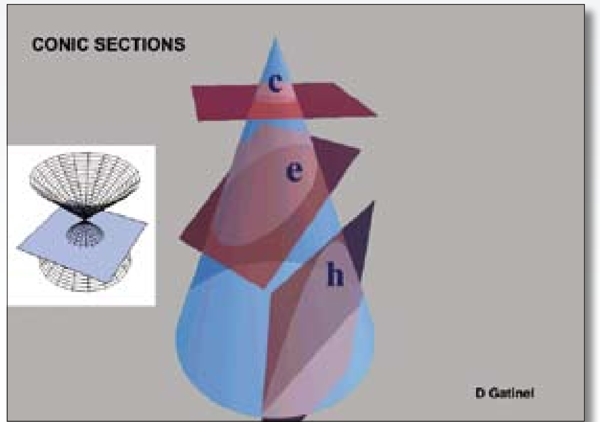


Figure.20

Figure.20: *The family of conic sections. The nature of the curve (circle, ellipse, parabola, hyperbola) depends on the angle of the plane relative to the cone.*

Other than for a circle, each member of this family of curves has a curvature that varies continually. Two parameters are sufficient to describe these curves fully: the apical radius of curvature and the asphericity factor.

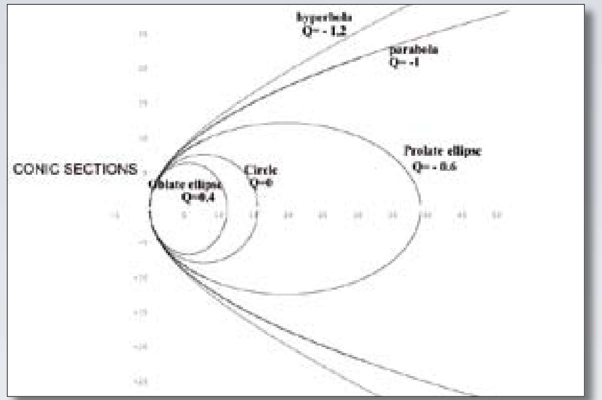


Figure.21

Figure.21: *Conic sections that have the same apical radius, but different asphericities. Curves with negative Q values (prolate ellipse, parabola, hyperbola) are said to be prolate, whereas curves with positive Q value (oblate ellipse) are said to be oblate. Here, the circle (Q=0) is oscillating with all the other represented curves. Therefore, its radius is the apical radius.*

All conic sections thus have a common mathematical equation in which these two parameters feature (Baker's equation):

$$y^2 = 2R_0x - (1-Q)x^2$$

Where Ro is the radius of curvature measured at the apex of the cone: this is the radius of curvature of the circle tangential to the apex of the conic section, also known as the osculating circle.

Q is the factor of asphericity: it characterises the variation in the radius of curvature with increasing distance from the apex.
Its value determines the type of conic section:
Q<-1: the curve is a hyperbola
Q= -1: the curve is a parabola
-1<Q<0: the curve is a prolate ellipse
Q=0: the curve is a circle
Q>0: the curve is an oblate ellipse

Other descriptors of the asphericity, called p and e, are also present in the scientific literature. They can all be calculated from each other since Q=p-1 and p=1-e^2 (**2**).

Ro determines the curvature of the surface in the immediate vicinity of the optical axis. For the cornea, it is used to determine the central keratometric power, and for an implant the nominal focal power.

Q governs the optical properties of the surface at a distance from the optical axis (non-paraxial conditions). Indeed, the sign of Q determines how the radius of curvature varies from the optical axis to the periphery (positive: the radius of curvature reduces compared with Ro, negative: the radius of curvature increases compared with Ro). When Q is zero, the profile of the surface is spherical; its radius of curvature is constant at all points (but not its optical power, this distinction is important). Conic sections thus offer a model that is both simple and "intelligible" in clinical terms.

If the changes to the profile of an optic are represented by different values of Q for the same Ro (for a constant apical radius), the modifications induced by the variation in asphericity are found to be of the order of a micron, but are more apparent with increasing distance from the apex of the conic section.

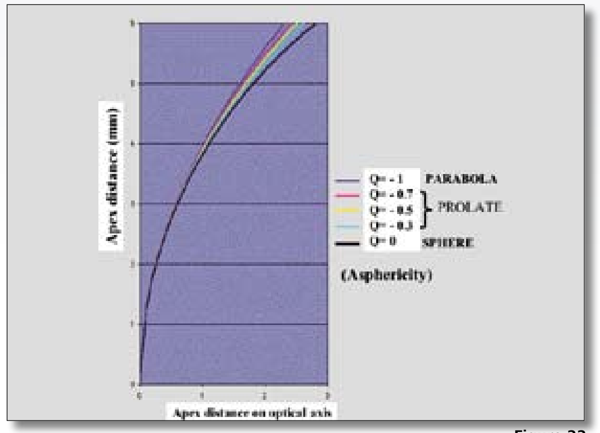


Figure.22

Figure.22: *Conic sections with the same apical radius of curvature but different Q values are drawn to scale.*

In optical terms, the role of asphericity increases if the system's aperture is wide. In other words, since the asphericity of the various ocular elements (cornea, lens) determines to a large extent the level of sphericity aberrations, then the repercussion of these aberrations on the quality of vision will be greater with increasing pupil diameter (since it lets through more peripheral rays). It is known that the difference in profile induced by the asphericity of a lens has an impact on the optical path of the waves that pass through it. In other words, the asphericity of a lens may be selected so as to equalise all the optical paths between the object point and the image point, irrespective of whether they pass through the centre or around the periphery of the system's aperture.

1.2.2 Figurate conic sections, conoid surfaces

Figurate conic sections are conic sections whose profile is modified by adding coefficients:

$$y^2 = 2R_0x - (1-Q)x^2 + kx^4 + lx^6 + \dots$$

This adjustment to the profile of the conic section around the periphery may be useful in increasing the accuracy of the modelling of the anatomical variations of the surfaces of the cornea and crystalline lens, or alternatively to provide optimal adjustment of the optical path within a lens.

Purely spherical or aspherical (non toroidal) surfaces are generated by rotating a conic section around its central axis. By definition, they have rotational symmetry (they can be pivoted around their central axis without changing their geometrical and optical properties). This is the case for the faces of non-toroidal phakic or pseudophakic implants. Toroidal surfaces are characterised by the difference in curvature between the meridians, and clearly do not have rotational symmetry.

The human cornea has two convex surfaces (anterior and posterior) that are naturally aspherical, insofar as away from the region located immediately next to its apex (within the central 3 mm) they are not simply a spherical surface. If they were, the representation of the topography of corneal curvature for our patients would show a uniformly colored elevation map. The corneal profile studied beyond the central 6 mm may be modelled using a figurate conic section.

The cornea is often also slightly toroidal. In this case, the region located near to its apex is no longer a sphere but rather a toroid. With this configuration, each meridian of the cornea considered in isolation remains aspherical, since its curvature varies from the centre to the periphery. It can only be likened to a toroid near to its apex.

These principles do not relate only to the cornea; they may be applied to the description of the surface of a spectacle lens or contact lens.

1.3 Physiological variations

1.3.1 Asphericity of the cornea

In mathematical terms, the profile of an aspherical surface such as that of an anterior corneal section is similar to that of the pointed (prolate) apex of an ellipse. The average value of the asphericity factor Q for the anterior face of the cornea is approximately -0.2.

The curvature of the cornea reduces slightly from the apex to its periphery. At the apex, which is generally the most curved part of the corneal surface, the average radius of curvature is 7.8 mm (approximately 43 D). The curve then eases (the radius of curvature increases) towards the periphery for a prolate cornea.

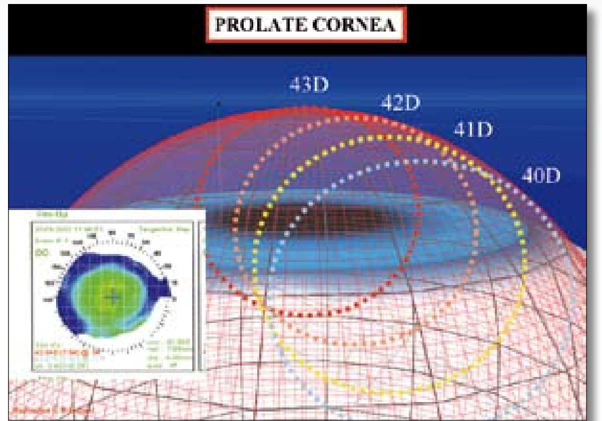


Figure.23

Figure.23: *The cornea is slightly prolate. Its curvature decreases from the center to the periphery.*

This variation, which has important optical consequences, has little significance in macroscopic terms and cannot be detected by biomicroscopic examination. The posterior face of the cornea is also aspherical and toroidal. Its radius of curvature is less than that of the anterior face, although its asphericity is more negative. It separates a more light-refracting medium (the stroma) from a less light-refracting medium (the aqueous humor) and for this reason has a negative optical power.

Some studies have, however, highlighted a large inter-individual variability. Moreover, an oblate profile is found in almost 100% of corneas after radial keratotomy (RK), refractive photokeratectomy (RPK) and LASIK for myopia. There is no statistical link between the value of the corneal asphericity and that of the central keratometry in the general population.

Neither does there appear to be any relation between asphericity and ametropia, other than for severe myopia for which a study found a higher percentage of oblate corneas than the average. We have not found any difference in corneal asphericity between myopes and emmetropes.

1.3.2 Asphericity of the crystalline lens

The central curvature and asphericity of the refractive surfaces of crystalline lenses is more difficult to evaluate due to the geometrical variations associated with accommodation and age, and the relative inaccuracy of the instruments that measure the anterior segment (evaluation of the asphericity requires accuracy of the order of a micron). Different values of Q between -6 and +1 have been proposed for the anterior and posterior faces of the crystalline lens. The estimations determined with a view to mimicking the optical behaviour of the human eye attribute a value of approximately -1 for the anterior face of the lens. The Liou and Brennan theoretical eye model attributes a value of -0.94 for the anterior face and +0.96 for the posterior face of the lens in addition to a longitudinal and transversal aspherical refractive index gradient.

The insertion of an implant with spherical surfaces cannot therefore recreate the optical quality provided by a young, non-accommodating natural lens. The increase in volume of the natural lens with age and certain structural modifications (variation in the refractive index gradient and spherisation of the surfaces) explains the increase in spherical aberration over time.

1.4 Optical role of asphericity

It is important to make a clear distinction between curvature and optical power. Optical power depends not only on the curvature, but also on the variation in refractive index and on the angle of incidence of the rays refracted by the surface under consideration.

A spherical lens with a given index (e.g.: n=1.49) and a constant curvature will have greater optical power at its periphery compared to at its centre.

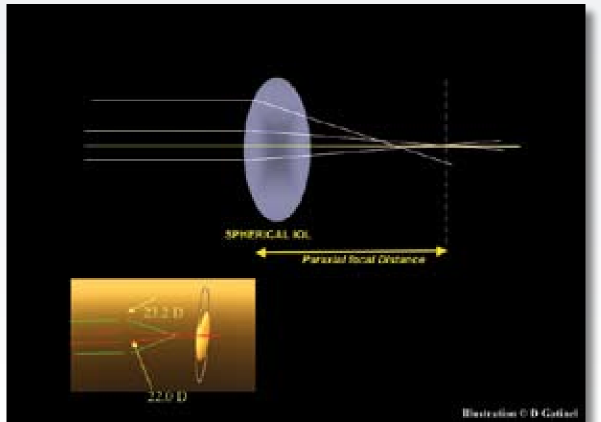


Figure.24

Figure.24: *IOL with positive spherical aberration. Despite constant curvature, the optical power of the IOL increases from the center to the periphery.*

This is because of the phase difference induced between the waves refracted by the centre and periphery of the lens. For a planar incident wavefront (emitted by a distant point source), the points that reach the surface of the lens are not in phase, and this phase difference is maximal between the points located far from the axis of the lens.

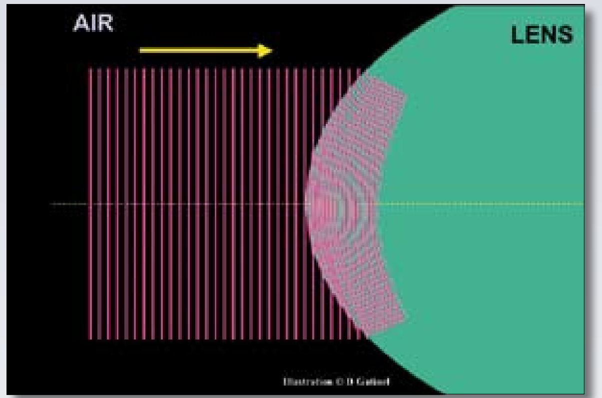


Figure.25

Figure.25: *When the incident planar wavefront meets the surface with the lens its propagation speed decreases (the wavelength shortens). Since the central portion of the incident wavefront reaches the lens medium before its periphery. The latter, which still travels in the air, gain some phase advance. This causes the wavefront to take on some curvature and converge. Depending on the asphericity and refractive index values of the lens, the wavefront may take on a spherical curvature and converge to a sharp focal point (no spherical aberration). Positive spherical aberration would be caused by the induction of excessive phase advance for the wavefront periphery; due to this optical path difference, the peripheral rays may not interfere in phase with the central rays at the lens focal plane.*

This variation in optical power between the centre and the periphery is expressed, for a refracted wavefront, by a phase difference between the points that leave its centre and its periphery. At the focal point of the lens, this phase difference causes a dispersion of the zone of superimposition of the in-phase waves, and equally by an enlargement of the focal spot.

In geometrical optics, this phenomenon may be interpreted as an increase in the angle of incidence of the rays that are not close to the optical axis, and that are responsible for their excessive refraction. The peripheral rays are then focused ahead of those refracted at the centre (positive spherical aberration).

An aspherical lens has a non-constant curvature, but may have a constant optical power if the reduction in curvature (peripheral flattening) exactly compensates for the effect of an increase in the angle of incidence so as to produce a constant lens power. In this case, the optical path length is identical for all the paths from the point source to the image point through the optical system. Peripheral corneal flattening (a prolate profile) thus reduces the sphericity aberrations or spherical aberrations that would be observed if the cornea was truly spherical, but is not extreme enough to totally cancel out the ocular sphericity aberrations.

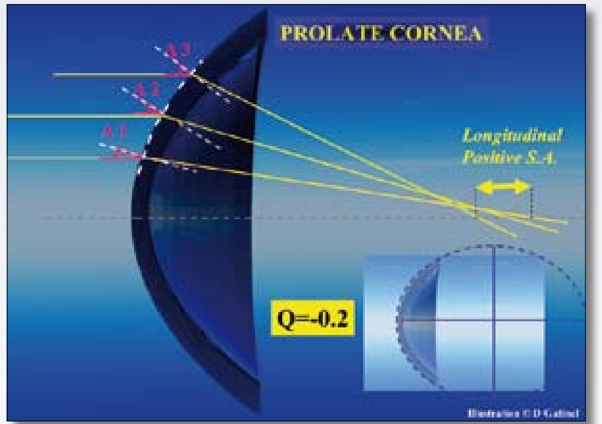


Figure.26

Figure.26: *The corneal profile is usually slightly prolate (mean Q value close to -0.2). The peripheral corneal flattening (a prolate profile) reduces the amount of positive spherical aberrations that would be observed if the cornea was truly spherical, but is not extreme enough to totally cancel out the ocular sphericity aberrations. The crystalline lens (not shown here) also contributes to the reduction of the ocular positive spherical aberration.*

Traditionally, the natural lens is thought to attenuate some of the remaining positive spherical aberrations, thanks to its aspherical geometry and it refractive index gradient (the index is higher at the centre of the lens than at its periphery).

The precise study of the optical properties of the natural lens runs into practical difficulties. Although it is possible to measure (using a Shack Hartmann type aberrometer) the optical aberrations of the eye as a whole, it is not possible to measure the aberrations induced by the lens alone in vivo. In vitro, it is possible to study the optical properties of the lens; however, structural modifications result from the extraction process and the analysis conditions differ compared to the physiological conditions. Particularly, the optical behaviour of the lens is related to that of the cornea that converges the light rays before these encounter the front face of the lens. The natural lens does not have a unique refractive index since it reduces from the centre to the periphery. This particularity would appear to make the lens generate a negative spherical aberration.

2. Spherical aberration

Spherical aberration is defined in geometrical optics as being the difference in refraction between the rays refracted at the centre and the rays refracted at the periphery of an optical system (Figure.8 and 27).

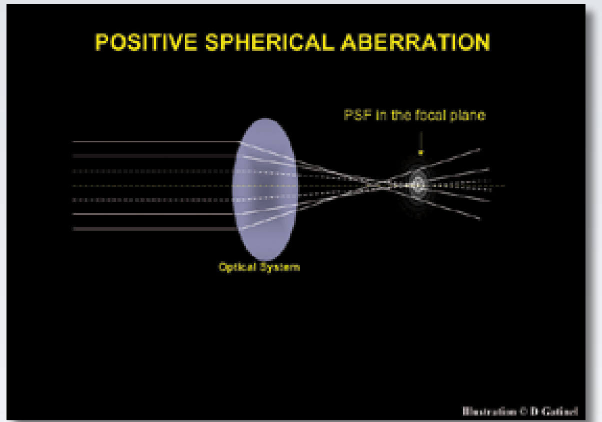


Figure.27

Figure.27: *Positive spherical aberration is due to an excessive bending of the refracted peripheral rays, which comes to focus ahead of the refracted paraxial rays. This causes the image of a point at the focal plane (point spread function: PSF) to become enlarged.*

In physical optics, the definition is an optical phase difference with rotational symmetry between the periphery and the centre of the wavefront. For this reason, pupil diameter is highly influential on the degree of spherical aberration. Spherical aberration is conventionally positive when the peripheral rays are refracted more than the central rays, and is negative when the opposite is true. Physiologically, there is a positive difference in total ocular optical power of approximately 0.5D between the periphery and centre of the pupil (for a pupil diameter of 6 mm). When calculated from the reconstruction of the wavefront followed by its breakdown into Zernike's polynomials, the average degree of spherical aberration for the entire eye is between 0.10 and 0.15 microns for a pupil diameter of 6 mm.

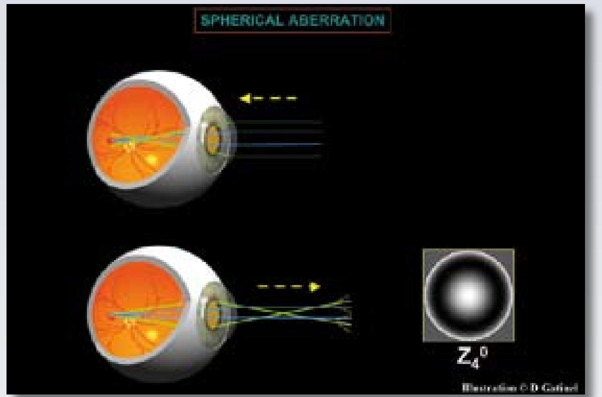


Figure.28

Figure.28: *The cornea and crystalline lens both reduce the positive spherical aberration (Z_4^0) of the eye to a residual value which is generally between 0.10 and 0.15 microns for a 6-mm pupil.*

In clinical practice, spherical aberration only starts to significantly impair the quality of vision when the pupil diameter is greater than about 4 mm.

Figure.29

Figure.29: *Comparison of the intensity Point Spread Function obtained in an eye purely aberrated with positive spherical aberration, and in an eye with no aberration (diffraction limited). The enlargement of the PSF for large pupil diameters may be responsible for visual symptoms such as halo and glare in mesopic or scotopic conditions.*

This impairment is not necessarily the reason for a loss of lines of visual acuity, as determined using high-contrast optotypes, but does reduce sensitivity to contrasts, notably at low and medium spatial frequencies.

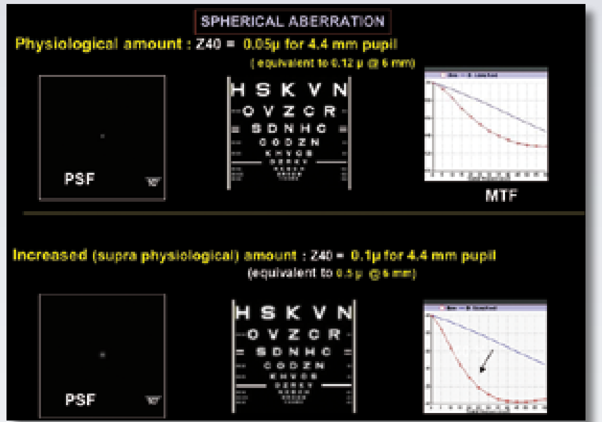


Figure.30

Figure.30: *Effect on the Optotype chart image and Modulation Transfer Function curve of the increase in positive spherical aberration (Z_4^0 term of the Zernike polynomial classification). Note the reduction in the area under the MTF curve, indicative of lower contrast sensitivity.*

It is in this frequency range (e.g.: 3 to 9 cycles per degree, which corresponds to an acuity of between 1/10 and 3/10) that the human eye's sensitivity to contrasts reaches its physiological maximum. The increase in spherical aberration may also cause visual signs, such as the impression of hazy mesopic vision, with impaired contrast sensitivity, or the perception of halos around bright light sources (headlights, streetlamps, etc.).

Several clinical studies have demonstrated the benefits offered by aspherical implants in terms of sensitivity to contrasts in the low and medium frequency range. An improvement in maximum visual acuity in mesopic environments has also been reported. However, the routine restoration of a level of optical quality comparable with that of young, phakic subjects has not yet been reported in pseudophakes.

3. Perspectives

Detailed analysis of the path of light waves through ocular optical structures should culminate in the development of optimised optical surfaces that can offer a degree of optical quality that is equivalent to that of a phakic eye.

The attainment of this objective is facilitated by the use of a realistic theoretical model of an eye, i.e. one that features aspherical refractive surfaces and relatively decentred optical surfaces. The model proposed by Liou and Brennan has four aspherical refractive surfaces, and a visual axis whose inclination is similar to that of the physiological mean (5°).

Optical calculation software tools (ray tracing) make it possible to test the behaviour of different implant geometries when certain ocular parameters are varied (mean central keratometry, asphericity of the anterior ocular surface, pupil diameter, etc.).

There is a choice of comparison criteria available: the degree of high-order optical aberration, the point-spread function (Strehl ratio) and the Modulation Transfer Function (MTF). The latter is particularly relevant since it determines the degree of attenuation of contrast at different degrees of resolution.

The refractive surfaces of the implant may be modelled using conic sections, or even by figurate conic sections since these offer the advantage of increasing the number of degrees of freedom and thus of refining the geometry of the tested lens.

The choice of optimal geometry is determined based on the limitations that must include realistic phenomena such as decentration and the slight tilt of the implant.

The choice of optimal geometry is determined based on the limitations that must include realistic phenomena such as decentration and the slight tilt of the implant

III. Impact of Modulation Transfer Function (MTF) on quality of vision, cataract surgery and aspherical IOLs

The restoration of quality of vision is a legitimate goal after cataract surgery. Ablation of the opaque crystalline lens and insertion of an implant into the capsular bag restores transparency to the ocular areas, while suppressing the attenuation and the light dispersion caused by the opaque crystalline lens. The reduced size of the corneal tunnels and the sophistication of the modern extraction techniques of the crystalline lens are quasi-neutral procedures for the non-crystalline ocular structures such as the cornea.

Despite perfectly performed surgery, certain subjects, often younger and more active than the average cataract patient, present with some bothersome visual phenomena, such as the sensation of a loss of sharpness or slight doubling of images. These symptoms occur more in a mesopic environment and are not reduced or even slightly reduced by spherical-cylindrical optical correction. They are caused by a rise in the level of certain high-order aberrations, themselves secondary to the spherical geometry of the implant. Indeed, it has been shown that spherical implants do not restore a quality of vision equivalent to that of a young pseudophakic eye after cataract surgery, due to the induction of extensive optical aberrations. The measure of visual acuity at strong contrast (done for example by means of black optotypes test chart on a white background) is not an exhaustive indicator of the quality of vision. It is however quite sensitive to the presence of spherical (myopia, hypermetropia) or cylindrical (astigmatic) defocalization; a demi-diopter of myopic defocalization is enough, for example, to reduce the visual acuity to a maximum contrast of two decimal lines. On the other hand, certain higher-order aberrations (spherical aberration, coma) are compatible with the maintenance of visual acuity tested in the classical fashion, but they induce a marked reduction of contrast sensitivity.

The optical performance of the human eye in terms of resolution and of contrast sensitivity may be evaluated using the “Modulation Transfer Function (MTF).“ Besides the purely basic approach that involves the understanding of optical concepts involved in its elaboration, this parameter opens interesting clinical and therapeutic perspectives to which this article is principally dedicated.

1. Images and spatial frequencies

The MTF represents the manner in which the optical system under consideration attenuates the contrast of the image that it forms with regard to the observed object.

This indicator may be calculated on the basis of the study of the aberrations deforming the wavefront. It is applied to any isolated optical system (cornea, implant) or to a system composed of different optics (entire eye, camera lens, telescope, astronomical telescope, etc.).

To understand the relevance of the MTF, it is necessary to be familiar with the decomposition properties of an image into its elementary constituent parts.

The decomposition of an image into a set of elementary points is quite intuitive (e.g. pixels). Each point of the image observed is defined as a luminous point source of given intensity. If one recognizes the way in which the optical system treats the image of an ”elementary” point (point spread function or PSF), it is then possible to apply this transformation to the set of points composing the initial image to obtain a simulation of the image rendered.

This operation is called convolution of the image by the point spread function (Figure.31).

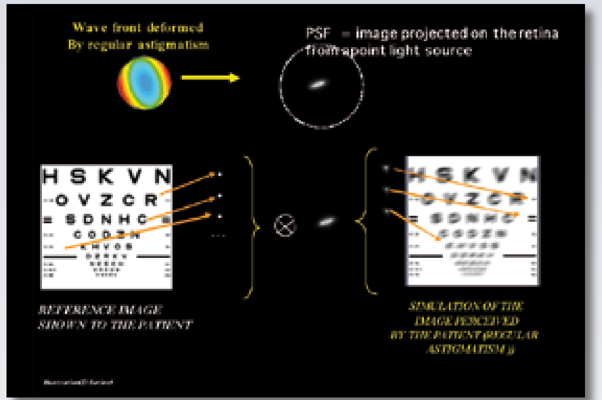


Figure.31

Figure.31: *The convolution, denoted by the symbol composed of a circle with a cross, consists in predicting the effect of an optical aberration on the image perceived by the patient. Besides the optical aberration responsible (here the regular astigmatism), this operation requires knowing the retinal point spread function (PSF). Its mathematical formulation is relatively complex and will not be illustrated here.*

Schematically, the convolution operation may be understood by considering the following three stages, which occur successively:
1) The image shown is broken down into a sum of elementary light points.
2) Each elementary point undergoes the PSF deformation.
3) The representation of the perceived image is done by recomposition of the elementary "deformed" points.
In the absence of optical aberration, the PSF is quasi-point (for the purpose of the near diffraction). Consequently, stage 2 does not change the morphology of the elementary points, and the image perceived (after recomposition) is unaltered (this is true only if the diffraction is moderate, that is, if the pupil diameter is more than about 2.5 mm).

The image convolution technique allows objectifying certain visual complaints of optical origin, but does not allow the direct assessment of the effect of the optical aberrations implicated in the contrast sensitivity.

A monochrome image may also be broken down into a combination of spatial frequencies. Each of these frequencies corresponds to an array composed of alternate dark and light bands (monochromatic light), allowing them to be oriented in a variable fashion within the image.

The number of pairs of dark and light bands per unit of distance defines the value of the spatial frequency. Each of the frequencies present in the decomposition of the image is balanced by a value that reflects its "amplitude," that is, the brightness interval between the darkest and brightest part of the array.

The superposition of these different ranges of spatial frequencies allows the recomposition of the fixed image (Figure.32).

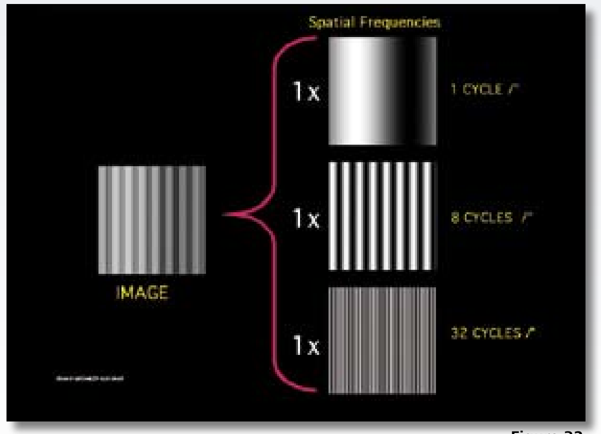


Figure.32

Figure.32: *Schematic example of decomposition of an image into spatial frequencies. The extent of the image presented corresponds arbitrarily to one degree of visual angle. The reference image chosen is deliberately "simple" since it may be fully recomposed with three spatial frequencies of equal contrast. The patterns of each of these spatial frequencies are moreover easily discernable in the image. This decomposition principle (Fourier transform) is applied however to more complex images, where the decomposition calls on a wide range of spatial frequencies of variable contrast and variable orientation. This operation involves assessing the relative importance of different spatial frequencies within the image analyzed (a coefficient proportional to the contrast intensity of each spatial frequency is calculated).*

A visual acuity of 10/10 corresponds approximately to the power to distinguish that allows the discerning of a spatial frequency of 30 cycles per degree (Figure.33).

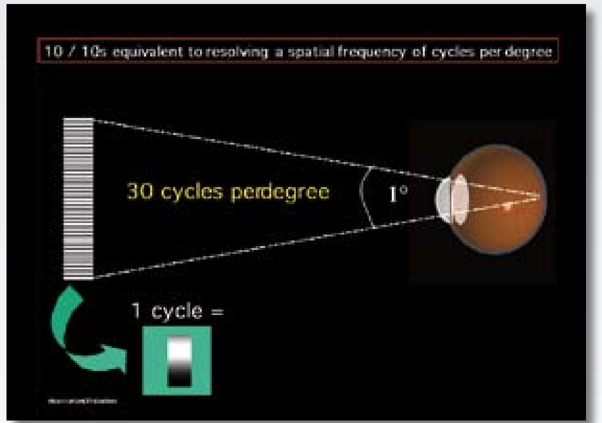


Figure.33

Figure.33: *The ocular separator power is directly linked to the "spatial frequencies." In this example, a visual acuity equal to 10/10 for a maximum frequency contrast is equivalent to the ability to "separate" 30 cycles each composed of a maximum (white) and a minimum (black) of brightness. One degree of angle corresponds to 60 minutes of arc; the total of black and white bars equals 60. If the maximum acuity of a patient is 10/10, this implies that the minimum angle of resolution is 1 minute of arc. When the spatial frequencies are presented with a maximum contrast, most healthy and properly corrected eyes in fact have acuity close to 15/10, even 20/10. Indeed, it can be demonstrated that the size of the foveal photoreceptors and the diameter of the PSF under optimum conditions (low level of optical aberration, weak diffraction) permit a range of samples of spatial frequencies close to 60 cycles per degree.*

The fact that an image may be broken down into a set of elementary periodic signals called spatial frequencies follows from Fourier analysis, which is applied to practically any complex signal. This approach, familiar to the physician, the electronics expert, or the specialist in signal processing, is however less familiar to the clinician who is more accustomed to having a visual pattern represented as the juxtaposition of luminous points.

In order to make it more familiar, this method may be compared to the spectral decomposition of a sound into various sound frequencies. The spectrum of spatial frequencies extends from the very lowest (the "lows," which correspond to the coarsest elements of the image), to the highest (the "highs," which permit the representation of fine details). The MTF calculated for the set of ocular diopters may be approximated for the visual function by the audiometric curve (audiogram), where it is applied to the study of the auditory sensitivity in order to determine the perception threshold of intensity for each sound frequency.

Thus the current examination of visual acuity does not explore the capacity of the horizontal ocular resolution for a target of maximum contrast, and is limited to the information contained on the horizontal axis of the MTF. It is reminiscent of an audiometric exam that would test the auditory perception of various sound frequencies only at maximum volume. For a healthy eye corrected for optical aberrations, the maximum discriminatory power is generally close to 20/10 when the target presents maximum contrast. This resolution corresponds to the maximum sampling capacity of the mosaic of fovea cones.

2. Spectrum of spatial frequencies of an image and the Fourier transform

The Fourier transform of an image involves performing a spectral decomposition. Unlike the human ear, which can break down a chord into its elementary different notes (sound frequencies) and which is thus capable of carrying out an operation similar to a Fourier transform (Figure.34), the eye cannot process the visual information in terms of spatial frequencies. On the other hand, digital imaging techniques exist that can perform this transformation almost instantaneously.

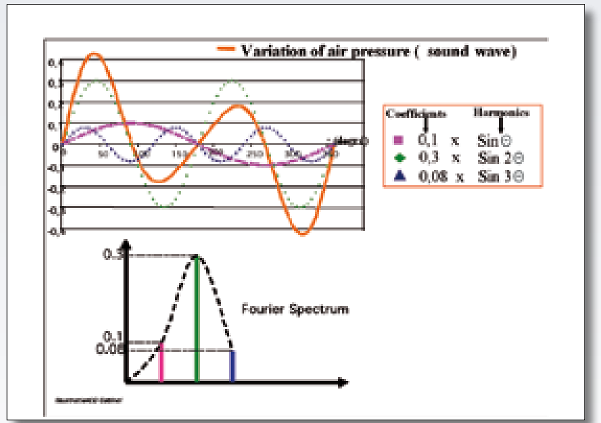


Figure.34

Figure.34: *Schematic representation of a cycle of air pressure variations causing a sound (full curve in orange). It can be broken down into three "pure" sounds (dashed curves) and conversely, the sum of these three sound vibrations reproduces the overall perception of "sound" (the variations in pressure are added or subtracted according to their sign). A musical, trained ear can decompose this sound into its three constituent harmonics. The Fourier spectrum represents the importance of the respective contribution of each of the harmonics present within the sound perceived. The human ear is capable of making Fourier type transforms. The decomposition of an image into spatial frequencies is an analogous optical procedure, where the spatial frequencies correspond to "pure harmonic sounds" from the acoustic domain. And given the two-dimensional nature of an image, the representation of its Fourier spectrum is itself two-dimensional.*

The frequency spectrum is represented by an image in grey levels. The intensity of each pixel of this spectral image is proportional to the amplitude of the coded spatial frequency. The frequencies are distributed from the center toward the edges of the image based on their level. The richer the image in terms of fine detail, the more marked will be the presence of higher spatial frequencies within the decomposition (Figure.35). Conversely, an image poor in details has a spectral content dominated by low and medium spatial frequencies (Figure.36).

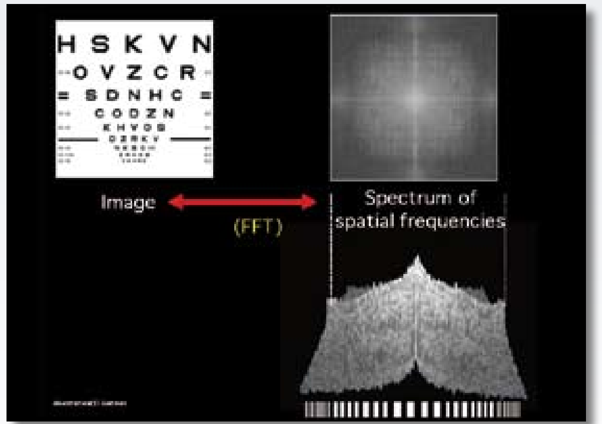


Figure.35

Figure.35: *Example of decomposition by Fast Fourier Transform (FFT) of the image of an optotype test chart of decreasing size (Snellen acuity chart). The frequency spectrum is represented in grey levels on the right. The luminosity of each pixel is proportional to the importance of the spatial frequency present in the image, considered for that of this pixel code. Below, the 3D representation of the spectrum allows the visualization of "peaks" of these frequencies - the higher the peak, the greater the brightness amplitude of the corresponding frequency is raised within the analyzed image. The peaks for the low frequencies are clustered at the center of the spectral image; the peaks for the high frequencies are close to the edges. The cross appearance with "bulges" visible at the edges of the spectrum presented in this example corresponds to the fact that the optotypes are presented in a fully aligned manner both horizontally and vertically, and are themselves coded by the spatial frequencies oriented primarily in these same directions. This spectral representation of an image should not be confused with the two-dimensional representation of the MTF in grey levels.*

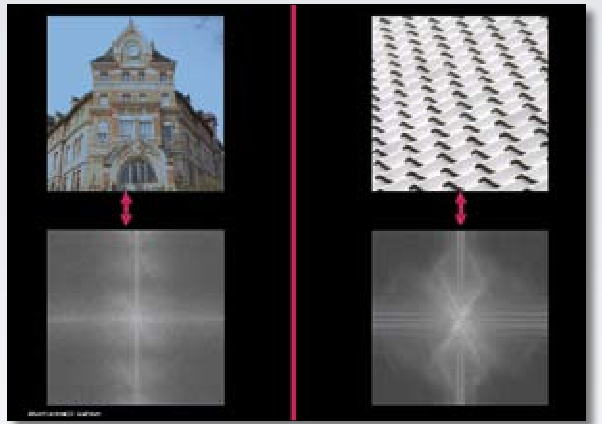


Figure.36

Figure.36: *Examples of images (above) accompanied by their respective Fourier transforms (below). On the left, the image represents the front wall of the Rothschild Foundation and has many details (various architectural patterns decorating the facade). The spectral content extends homogeneously from the low frequencies (center of the spectral image) to the high frequencies (edge of the spectral image). On the right, the image corresponds to an arrangement of tiles from a Zen temple pavilion (Kyoto, Japan). It presents a regular, repeated pattern that is made up of primarily low and medium frequencies. It is interesting to note that the spectral content in this example is clustered toward the center of the diagram (weak expression of high spatial frequencies in this image). The oblique orientation of the most important peaks in the frequency space indirectly reflects the oblique arrangement of the tiles in the actual image.*

These frequencies have a variable effect on visual perception. For example, it has been shown that the recognition of patterns like familiar faces and landscapes is done essentially on the basis of correct perception of medium spatial frequencies (Figure.37). This notion is reflected in the relative tolerance by certain elderly persons with regard to a moderate myopic ametropia that does not hamper them or hampers them only slightly in their daily life in identifying things close to them or to navigate within a familiar environment. Moreover, neuro-ophthalmologic processing of the visual information acts as a filter in attenuating the perception of the low spatial frequencies in favour of the medium frequencies.

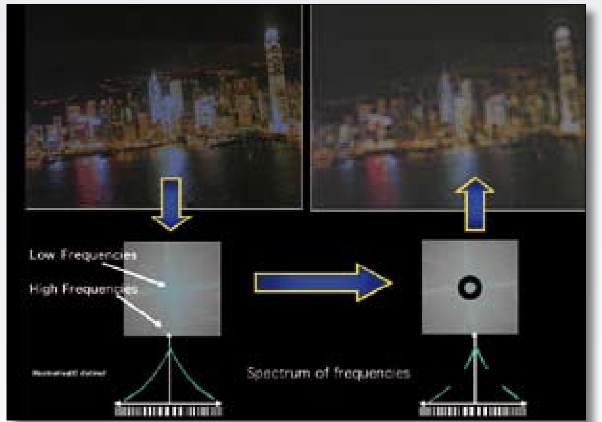


Figure.37

Figure.37: *Effect of removal of some medium and high frequencies by a frequency filter in the decomposition into spatial frequencies of an image representing Hong Kong Bay. After reconstitution of the image (without the suppressed spatial frequencies), the landscape is identifiable, but the contrast is reduced, and certain details are absent. The absence of certain frequencies in the final recomposition also explains the presence of "double contours" in the image. This type of filtering may be used to rid an image of "noise", such as rasterization (for that it is sufficient to remove one or more spatial frequencies that code the undesirable raster).*

3. Interest in the MTF

The MTF corresponds to a ratio between the respective contrasts of the projected image and the image formed for each spatial frequency. As a function of the diffraction and the importance of optical aberrations of the system studied, it is possible to determine how the fashion in which the optical system attenuates the contrast between specific spatial frequencies, and to deduce from this the optical quality of the image that is rendered (Figure.38). The closer this relationship is to 100%, the better is the optical quality of the system tested for this spatial frequency. For a "perfect" optical system, devoid of optical aberrations, the MTF curve initially decreases linearly with a negative slope, slightly flattening towards higher frequencies, due to the effects of the diffraction, which reduces the contrast of the high spatial frequencies. The presence of high-order optical aberrations reduces "the height" of the curve, since the aberrations reduce the contrast that is transmitted.

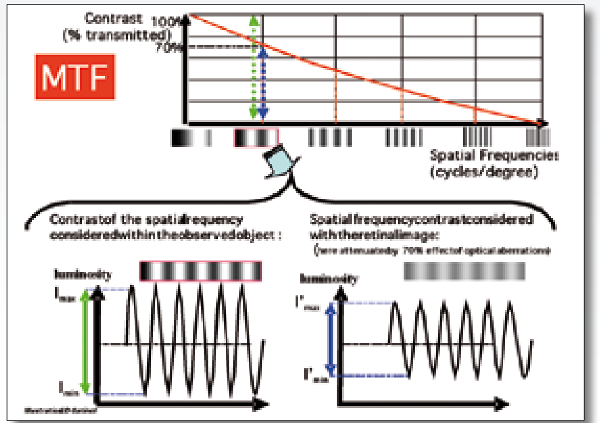


Figure.38

Figure.38: *Principle of contrast transfer modulation. The MTF curve represents the loss of optical contrast for each spatial frequency, brought about by the combined effects of refraction errors and diffraction.*

Each of the spatial frequencies constituting the fixed object corresponds to a sinusoidal vertical array (brightness variation). After refraction by the ocular optical surfaces, this array has a reduced contrast within the image because of diffraction and because of optical aberrations. The modulation for this frequency is given by: Modulation (M) = (Imax –Imin) /(Imax + Imin).

In this example, 70% of the contrast is transmitted for the spatial frequency corresponding to 2 cycles per degree. The spatial frequencies have a variable orientation, and the MTF is at first calculated for each of these frequencies (two-dimensional MTF generally represented in grey levels). The curve presented as a graph is calculated as the mean of the respective MTF curves of each of the hemimeridians of the pupil of the optical system under consideration. When the optical system is free of optical aberrations, only the effect of the pupil diffraction reduces the optical quality. As a result, the theoretical MTF curve of a system limited by diffraction is not a horizontal straight line, but one slightly inclined to the right. The effect of the dispersion is not taken into account when the MTF is calculated after receiving the wavefront, but may be so if the MTF is calculated on the basis of the direct reception of the PSF. The phase shift effect caused by certain aberrations, in particular asymmetrical or unpaired aberrations (coma), is not represented in this curve.

Calculation of the MTF thus allows predicting the effect on the visual envelope of a disturbance of the wavefront (optical aberrations of low and high order). An equal visual acuity tested for maximum optotype contrast depends on the emmetropization of the rays crossing the center of the pupil. The visual acuity tested for reduced optotype contrast is correlated to the emmetropization of the periphery of the pupil (Figure.39). This is particularly interesting for aberrations in which the type and the rate are insufficient to induce a decrease of the best corrected visual acuity, but which provoke a reduction of the sensitivity to optical contrasts for certain spatial frequencies, such as the coma or spherical aberration. For example, these last items induce a reduction of the predominant MTF for the medium spatial frequencies. The value of the RMS rate (Root Mean Square – mean square or effective value) of the high-order aberrations of the eye is not always proportional to the deterioration of the optical quality that they induce, since certain high-order optical aberrations compensate for them. The MTF curve is a more pertinent piece of information than the RMS level of a given aberration or a group of aberrations because it illustrates the expected effect of these aberrations on the various spatial frequencies and for different pupil diameters (Figure.40). The MTF curve can be calculated for the entire set of optical errors (poor focus, astigmatism, high-order aberrations). Nevertheless, it is generally calculated for the best spherical-cylindrical correction in order to reveal the effect of the high-order aberrations alone.

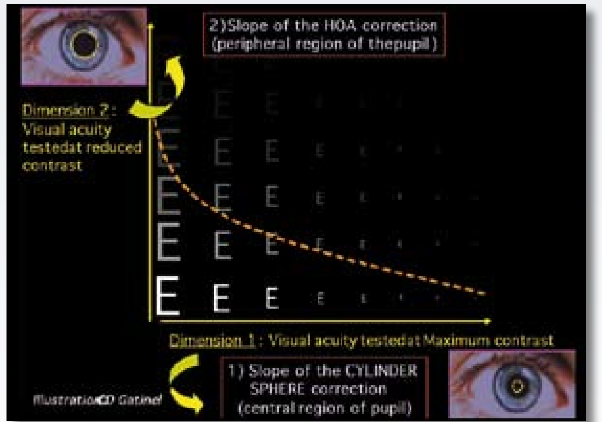


Figure.39

Figure.39: *Correlation among MTF, clinical factors, and the visual envelope. The emmotropization of the light rays passing through the centre of the pupil is correlated to good visual acuity at maximum contrast, but does not guarantee the maintenance of this acuity at reduced contrast. The MTF curve, which may be matched to sensitivity to predicted contrasts by considering optical defects, is as "high" as the "emmotropized" pupil surface is large.*

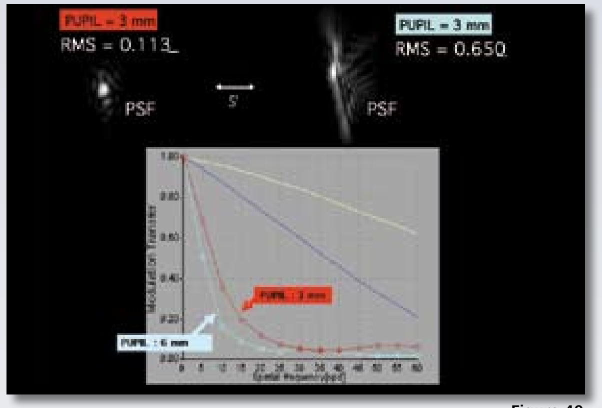


Figure.40

Figure.40: *Representation of the PSF and the average MTF curve for the same eye analyzed at two different pupil diameters (3 mm and 6 mm) and best corrected in eyeglasses for sphere and cylinder. The eye analyzed is pseudophakic (hydrophobic acrylic spherical biconvex implant), and the patient complains of a reduction in the quality of night vision on this side. The corrected visual acuity tested with high contrast optotypes is about 10/10. The MTF is calculated after reception of the wavefront for aberrations of high order only (i.e. for the best spherical-cylindrical correction). Note the broadening and the vertical distortion of the PSF with pupil dilatation (deleterious effects of coma type optical aberrations present in the wavefront). The contrast transfer is more reduced for the 6 mm pupil diameter than for the 3 mm pupil diameter for the set of spatial frequencies. The yellow and dark blue straight-line curves correspond respectively to the average theoretical MTF curve in the absence of optical aberration (diffraction only) for the 6 mm and 3 mm diameters.*

The presence of a high level of optical aberrations has deleterious consequences for the MTF and thus for the visual acuity at maximum contrast and/or the sensitivity to contrast at certain spatial frequencies. Although the MTF is loosely linked to the function of contrast sensitivity, the measurement of the latter in clinical practice depends also well on neurological factors and is limited essentially to lower and medium spatial frequencies. In general, the visual cortex performs a special "processing" of the information transmitted by the receiver (eye) and the visual pathways, and can "compensate for" or conversely "accentuate" the effect of certain optical aberrations on visual perception. The natural crystalline lens is aspherical and compensates for some of the aberrations induced by the corneal diopter; those that do not allow the spherical implants. Reduced quality of vision after cataract surgery and insertion of a pseudophakic spherical implant is connected principally to the inducing of a higher level of high-order spherical aberrations, such as spherical aberration and coma type aberration. Certain aspherical implants improve contrast sensitivity by reducing the rate of total post-operative spherical aberration.This gain may be reduced or voided in case of imperfect centration due to the induction of a coma-type aberration. The threshold between the gain made by aspherization and the loss linked to the inducing of coma is difficult to evaluate based on the study of the respective RMS rates of these aberrations. Moreover, the existence of physiologic decentration among the various ocular diopters is neglected in the design of aspherical implant suppliers of first-generation negative spherical aberration.

4. MTF and the design of a pseudophakic implant

Calculation of the MTF may be used to refine the design of an aspherical pseudophakic implant. The CT ASPHINA™ 604P and the CT ASPHINA™ 603P (Carl Zeiss) are the first intraocular lenses to have been specifically designed based on the predicted effect on the quality of vision by the calculation of the ocular MTF (Figure.41).

A theoretical model of the eye incorporating the aspherical corneal surfaces, the physiologic pupil decentration, and the eccentricity of the fovea has been chosen ("Liou and Brennan" model). The MTF calculated for the whole eye and implant allows the incorporation of the combined effects of spherical aberration and coma. For each design of an implant tested in this theoretical eye, the calculation of the total eye MTF is done. The geometry used for the design of the aspherical implant is the one that provides the best MTF curve on the set of the tested spatial frequencies. Calculation of the MTF on the basis of the reception of the wavefront by the patient who has received this type of implant must logically allow the verification of the theoretical relevance of this model and has a significant clinical application.

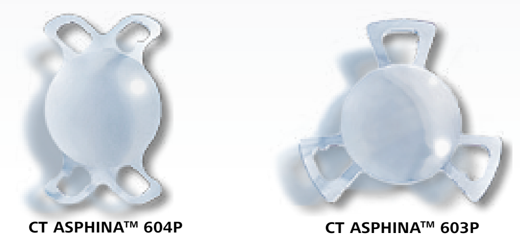


Figure.41: *ZEISS Optic aspherical lenses.*

Conclusion

Modern optical correction devices, are today elaborated not only to restore good visual acuity, but als to better preserve or to improve contrast sensitivity. This opportunity appears to be particularly interesting for patients who are active and who perform certain activities in mesopic conditions and/or in a context of severe visual stresses such as night driving. The aspherization of the optics of posterior chamber implants is an important first step in this direction. In view of the potentially deleterious role of other optical aberrations, such as coma or the existence of physiological pupil decentration, represents an additional theoretical advance for second-generation aspherical implants. The MTF allows the integration of the combined effect of all high-degree aberrations. This parameter should thus occupy a major place in the evaluation of the optical quality of the eye and in the development of innovative optical correction devices.

References

Gatinel D. Wavefront Analysis, pp117-146. Refractive Surgery, 2nd Edition, Azar DT, Gatinel D, Hoang-Xuan T. Elsevier Mosby, 2006.

Liang J, Grimm B, Goelz S, Brille JF. Objective measurement of wave aberration of the human eye with use of a Hartmann-Shack wavefront sensor. J Opt Soc Am A,1994;11:1949-1957.

Cheng X, Bradley A, Thibos LN. Predicting subjective judgment of best focus with objective image quality metrics. Journal of Vision, 2004;4(4):310-321.

Marsack JD, Thibos LN, Applegate RA. Metrics of optical quality derived from wave aberrations predict visual performance. Journal of Vision, 2004; 4:322-328.

Gatinel D. Aberrations monochromatiques de haut degré: définitions et conséquences sur la fonction visuelle. High-Degree Monochromatic aberrations: definitions and results for the visual function. Gatinel D, Hoang-Xuan T: Le LASIK: de la théorie à la pratique, (LASIK - from theory to practice), Elsevier, 2003 ;pp151-159.

Thibos LN, Applegate RA. Assessment of optical quality. In:"Customized corneal ablation: the quest for supervision". MacRae S, Krueger R, Applegate R, pp67-78. Slack inc, 2001, Thorofare, NJ, USA.

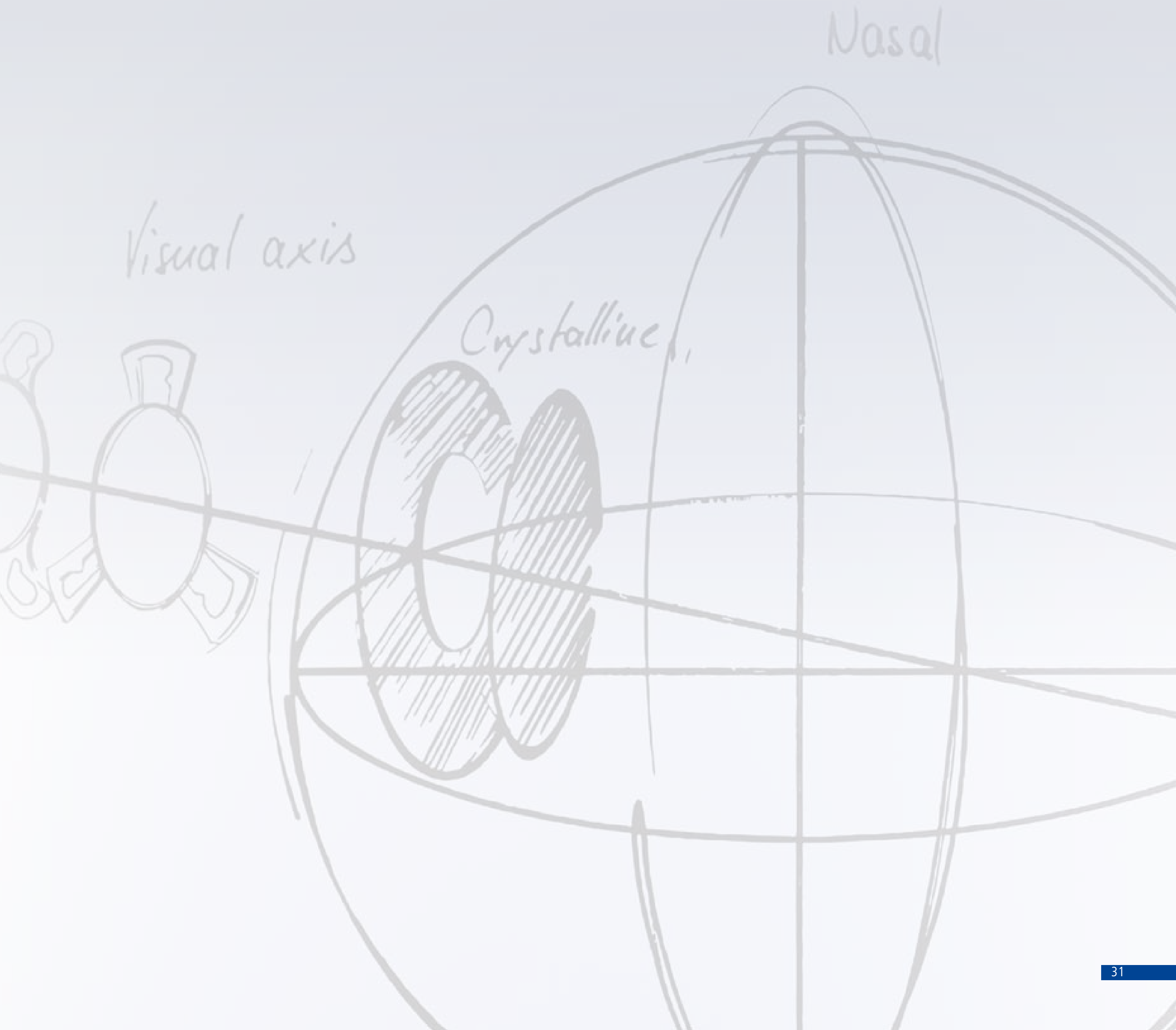
Gatinel D, Malet J, Azar DT, Hoang-Xuan T. Effects of the Pupil Constriction on the Wavefront Zernike Terms. Invest. Ophthalmol. Vis. Sci. 2003 44: E-Abstract 966.

Thibos LN, Applegate RA, Schwiegerling JT, Weeb R, VSIA standards taskforce members. Standards for reporting the optical aberrations of eyes. J Refract Surg,2002;18: S652-S660.

Artal P, Guirao A, Berrio E, Williams DR, Compensation of corneal aberrations by the internal optics in the human eye. J Vis. 2001;1(1):1-8.

Barbero S, Marcos S, Jimenez-Alfaro I. Optical aberrations of intraocular lenses measured in vivo and in vitro. J Opt Soc Am A Opt Image Sci Vis. 2003;20(10):1841-51.

Rohart, B Le Marinel, T Hoang-Xuan, D Gatinel. Comparison of the ocular aberrations after cataract surgery with hydrophobic (MA60AC, ALCON) and hydrophilic (XLSTABI, Ioltech) acrylic intraocular lens: a comparative study. J Cataract Refract Surg, 2006;32(7):1201-5.



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