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Aberration Correction with Aspheric Intraocular Lenses

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Abstract

The shape of the normal human cornea induces positive spherical aberration (SA) which causes image blur. In the young phakic eye, the crystalline lens compensates for a certain amount of this corneal aberration. However, the compensation slowly decreases with the aging lens and is fully lost after cataract extraction and implantation of a standard intraocular lens (IOL). Conventional spherical IOLs add their intrinsic positive SA to the positive SA of the cornea increasing the image blur. As a useful side effect, this also increases the depth of focus—often referred to as pseudo-accommodation. Aspheric intraocular lenses have been introduced to be either neutral to SA or to compensate for a certain amount of corneal SA. A customized correction for the individual eye seems to be the most promising solution for tailored correction of SA. In this chapter we will provide detailed information on the various concepts of aspheric intraocular lenses to elucidate that the term “aspheric intraocular lens” is being used for a large amount of different lens designs.

Keywords: spherical aberration, aspheric surface, customized intraocular lens, decentration, tilt

1. Introduction

The disease pattern of cataract comprises pathologic conditions of the human eye resulting from an opacification of the crystalline lens. The most frequent causes for the development of cataract are age-related transformation processes. Although research on pharmacologic treatment of cataract has been in focus for many years, the surgical extraction of the cloudy crystalline lens and implantation of an artificial intraocular lens (IOL)—referred to as cataract surgery—represent the only available treatment. Cataract surgery is one of the most frequently performed surgical procedures with several million surgeries being performed worldwide each year.

First IOL developments were primarily targeted on biocompatible materials and new fixation techniques rather than on correction of ocular aberrations other than defocus and astigmatism. First lens implants were made from polymethyl methacrylate, therefore being rigid and requiring large incisions for implantation. Furthermore, the optimum site of implantation (anterior chamber, iris, ciliary sulcus, or capsular bag) still had to be found, and adequate haptics for proper fixation had to be developed. Surgical results were therefore less predictable [1, 2].

In the early 1980s, foldable silicone materials and later acrylic materials allowed implantation through smaller ports and therefore caused less damage to the corneal structure allowing a faster rehabilitation. This finally facilitated ambulant cataract

surgery. In the following years, the capsular bag was identified as the optimum position for an IOLs, and the development of new lens power calculation formulas dramatically increased the predictability of the refractive outcome [1].

2. Aspheric lenses

With the improvement in IOL power calculation, the goal of cataract surgery became predictable; the focus of cataract surgery shifted from “restoration of vision” to “refractive surgery.” Manufacturers started optimizing the IOL optic from an equiconvex spherical lens design to different aspheric surface profiles and finally multifocal and free-form surface designs. The buzzword of those days was “spherical aberration” (SA) which should be eliminated to improve contrast sensitivity and visual acuity. Spherical aberration is one of the monochromatic aberrations that is caused by the difference in focal length (or optical power) for varying aperture diameter of a lens. For positive spherical aberration, the optical power increases from the lens center to the periphery, and rays far from the optical axis will intersect the optical axis in front of the paraxial focus (**Figure 1**).

Any spherical optical surface causes SA. To achieve an equal distribution of optical power across the lens diameter, the optical surfaces have to be tailored accordingly. SA can only be reduced by varying the spherical radii of curvature of anterior and posterior surface yielding a so called best-form lens, but it cannot be eliminated. This can be achieved by implementing aspheric surfaces. There are basically two types of aspheric surfaces that have been described in the literature, the first one is referred to as “continuous asphere” and can be described by the formula

$$z = \frac{\frac{1}{r} \cdot \rho^2}{1 + \sqrt{1 - (1 - Q) \cdot \frac{1}{r^2} \rho^2}} + \sum_{n=1}^{n=8} a_{2n} \cdot \rho^{2n} + X \quad (1)$$

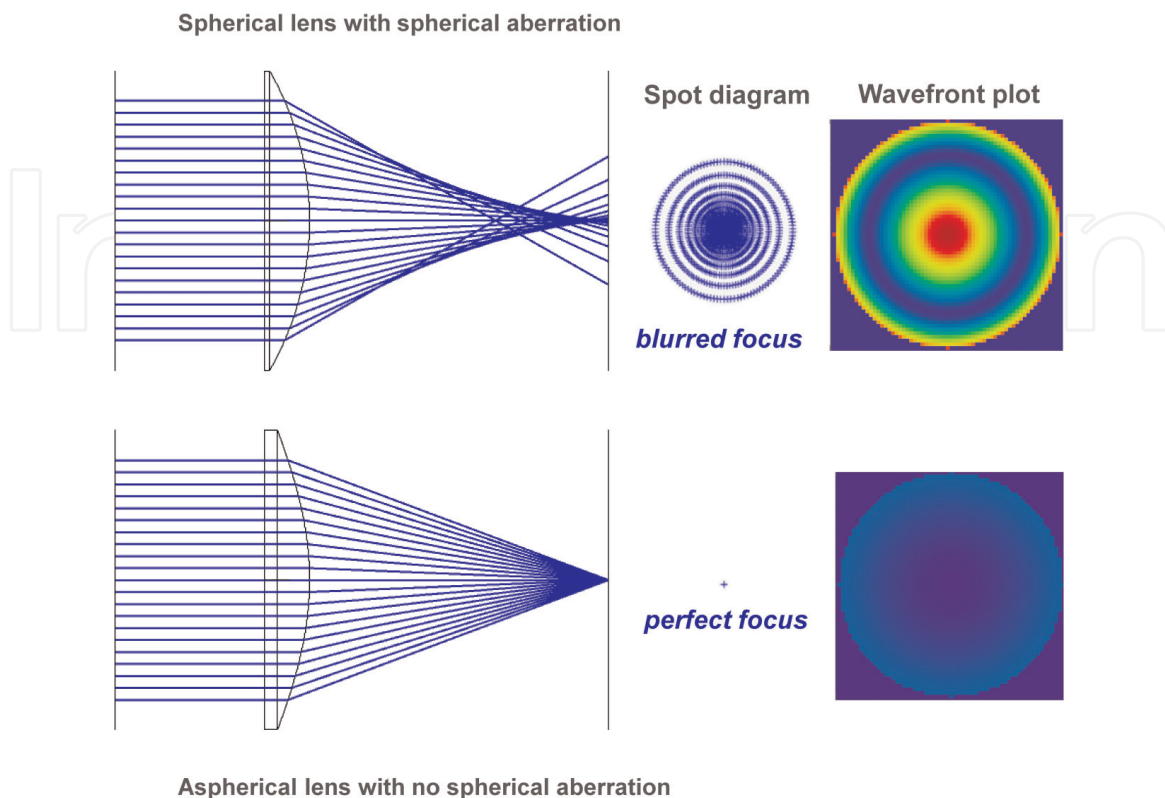


Figure 1. Rays focused by a lens with positive spherical aberration (top) compared to a lens without spherical aberration (bottom) [4].

where z is the height of the surface from the apex ($= 0$ mm), r is the radius of curvature, and ρ is the radial coordinate from the center to the periphery. Q is called “asphericity” [3], and a_{2n} are higher-order aspheric coefficients. X is a placeholder for additional polynomials, such as Zernike polynomials, which can be used to define additional surface shapes. From this equation numerous aspheric surface profiles can be generated (**Figure 2**), and most of current aspheric intraocular lens designs are based on the formula above. Equation (1) can be expanded to represent toric and biconic surfaces as well. The asphericity Q is identical to the conic constant κ (often used in optical design software) and can be transformed from other shape definitions for the aspheric surface such as the eccentricity e or the index of eccentricity e^2 [3]:

$$Q = -e^2 \quad (2)$$

The second type of aspherical surfaces, called “zonal asphere,” is constructed from a set of annular rings with varying radius of curvature and asphericity. For a detailed description of these surfaces, please refer to the literature [5, 6].

By modulating radius of curvature, asphericity, and aspheric coefficients, the SA induced by the surface can be customized. Additional polynomials can be added on top to create non-rotationally symmetric aspheric surfaces to compensate for higher-order errors such as coma or trefoil. For example, one alternate way to compensate for spherical aberration would be to modulate the aspherical surface with a linear combination of Zernike polynomials representing various orders of spherical aberration:

$$X = C_{11} \cdot Z_4^0 + C_{22} \cdot Z_6^0 + C_{37} \cdot Z_8^0 + \dots \quad (3)$$

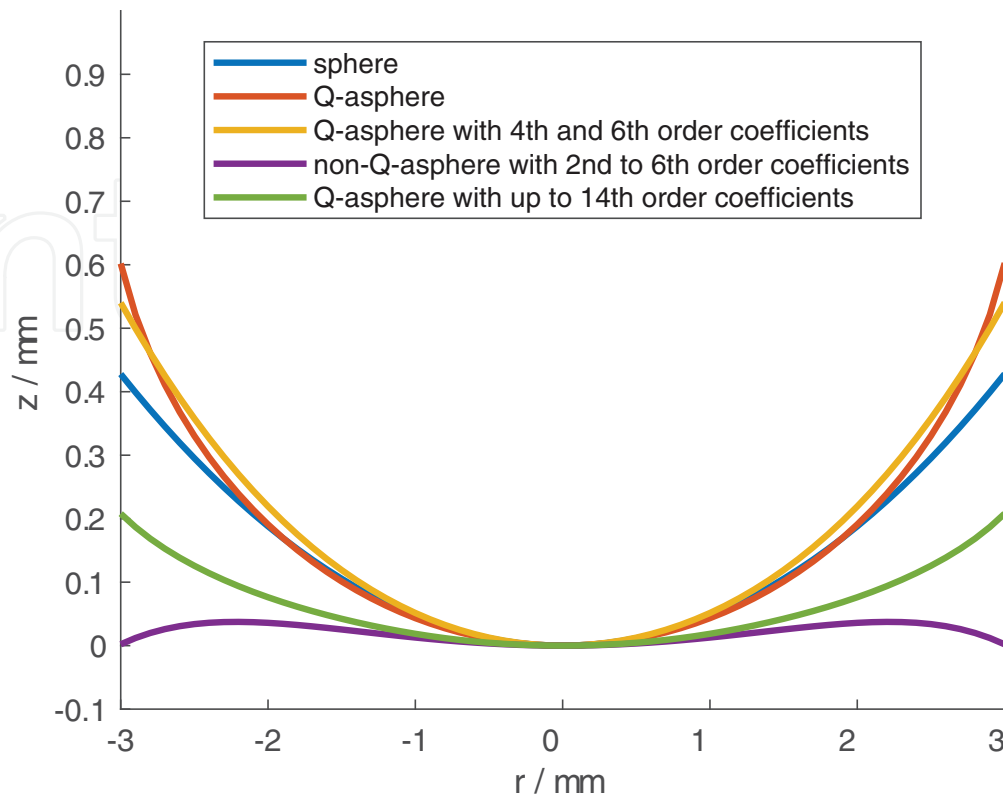


Figure 2. Variation of optical surface section with increasing number of coefficients. z is the elevation relative to the surface apex. All curves are derived from intraocular lens designs for an average power (20 to 22 D) lens.

3. Corneal spherical aberration

As mentioned above, the average human cornea induces a significant portion of positive SA, which is typically being described by the Zernike coefficient Z_4^0 (spherical aberration) on a diameter of 6.0 mm at corneal plane. The amount of SA can be calculated from the corneal surface shape by optical ray tracing. A method to do so was described by Norrby et al. providing a reference value for the Liou-Brennan model eye [7, 8]. Calossi provided an overview of SA values for a limited set of variables [3].

Depending on the underlying database, various authors reported different values for the average corneal SA. Holladay et al. reported that the average SA of the human cornea is about $+0.27 \pm 0.20 \mu\text{m}$ (value misprinted in the original publication [9] and corrected by Norrby et al. [7]). Similar values were found by Beiko and Haigis ($+0.274 \pm 0.089 \mu\text{m}$) [10]. The widely spread Liou-Brennan model eye provides about $+0.258 \mu\text{m}$ of spherical aberration being close to the reported average clinical values [7, 8]. De Sanctis et al. found higher values in their patients ($+0.328 \pm 0.132 \mu\text{m}$) [11], while Shimozone et al. found lower values ($0.203 \pm 0.100 \mu\text{m}$) [12].

4. Correction of spherical aberration with IOLs

Aberration correction could be best described as a superposition of wave fronts as outlined in **Figure 3**.

During cataract surgery the corneal SA is typically increased by the likewise positive spherical aberration of a spherical IOL. Therefore, lens designers at first created the “aberration-free” or “aberration-neutral” lens concept, a lens design that was meant to eliminate its intrinsic spherical aberration and thus being neutral to the eye’s overall SA [13]. However, the amount of SA is highly depending on the vergence of the incident rays. Therefore, there are differences in the design of “aberration-neutral” lenses: some of them are designed to be neutral to SA in a collimated beam, e.g., a beam as such could be used in measurement instrumentation. Others are designed to be neutral to SA behind some generic model cornea (in a converging beam). Both of them will exhibit a considerable amount of SA when implanted in a real eye; the first will provide a small correction for SA, while the latter may provide

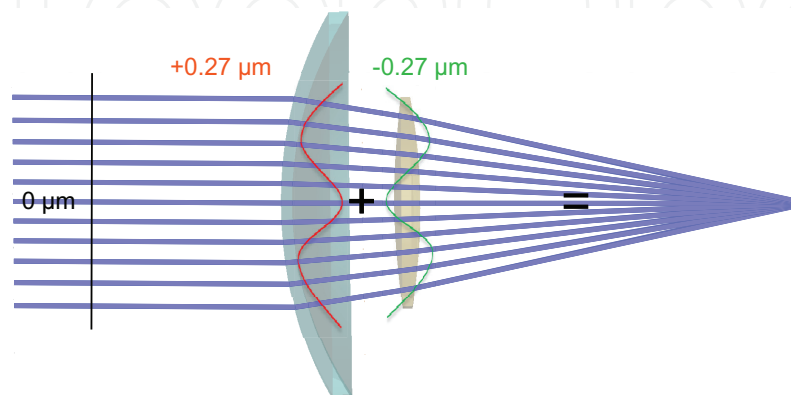


Figure 3.

Simplified sketch of the principle of aberration correction: An impinging plane wave front (collimated beam) is refracted by the cornea and affected by spherical aberration (red); the intraocular lens (yellow) compensates for the same amount of spherical aberration (green) resulting in a perfect wave front at the focus plane. Note: The plotted wave fronts do not account for the defocus.

Manufacturer	Product	SA correction
Johnson & Johnson Vision, Groningen, The Netherlands	TECNIS	-0.27 μm [9]
HOYA, Nagoya, Japan	Vivinex XC1	-0.18 μm [17, 18]
Carl Zeiss Meditec, Berlin, Germany	CT ASPHINA 509MP	-0.18 μm [18]
Alcon Laboratories, Forth Worth, TX, USA	AcrySof IQ SN60WF	-0.17 μm [17]
Bausch + Lomb, Rochester, NY, USA	EyeCee One	-0.14 μm [18]
	Quatrix Evolutive	-0.1 μm [18]
PhysIOL, Liege, Belgium	PODeye	-0.11 μm [18]
Kowa Pharmaceuticals, Düsseldorf, Germany	AvanSee	-0.04 [18]

Table 1.
List of selected intraocular lens models providing correction of spherical aberration.

negligible change to the corneal SA. When being analyzed on an optical bench (in a collimated beam), both lenses will show the opposite characteristics [4].

Aberration-correcting designs evolved subsequently, providing compensation to a fixed amount of corneal SA. One of the first aberration-correcting lenses was presented by Holladay et al. providing a correction of $-0.27 \mu\text{m}$ and thus targeting on the average SA found in human eyes [9].

Today, surgeons may choose from a variety of aspheric IOLs with different amount of compensation for SA (**Table 1**). Theoretically, one could choose the IOL providing the optimum correction for an eye. This would require preoperative examination of corneal topography and analysis of corneal aberrations. Diagnostic instrumentation for the anterior segment such as the Pentacam (Oculus Optikgeräte GmbH, Wetzlar, Germany) or the CASIA2 (Tomey Corp., Nagoya, Japan) allow direct readout of the SA amount over 6 mm diameter. The SA value calculated from corneal tomographic data could then be used to select an IOL model that provides best correction. Still, since the range of IOLs with different SA corrections is limited, not every eye could be supplied with optimum correction. Clinical results with this “selection method” are controversial but indicate the potential for improvement [14–16]. Piers et al. found that contrast sensitivity peaks with $0 \mu\text{m}$ of SA [17]. On the contrary, other investigators found that a residual SA of about $+0.1 \mu\text{m}$ may be beneficial for visual performance [18–20]. Manzanera and Artal argued that changes in SA between -0.17 and $+0.2 \mu\text{m}$ are merely noticeable by patients [21]. This may be an explanation why the differences in visual performance between aberration-free and aberration-correcting lenses are usually small.

The next logical step is a compensation procedure based on the true individual SA, rather than on average values. Wang et al. found that not only SA should be considered but the full spectrum of corneal aberrations [22–24]. Especially eyes with high amounts of spherical aberration such as eyes after laser refractive surgery or eyes with forme fruste keratoconus could benefit more from customized correction of SA [22, 23, 25] than normal eyes, if centration of the implant can be kept within strict limits. Therefore, an optimum solution could be the customization of intraocular lenses [26, 27]. Several researchers provided theoretical basics and theoretical results showing the potential of customized intraocular lenses [28–31]. The design process of such IOLs requires the implementation of customized model eyes based on biometric data and the use of ray tracing technology [28, 32–36]. The first clinical results with this method have recently been published showing promising results [37].

5. Limitations of aberration-correcting lenses

A major limitation for the selection of the appropriate IOL is the accuracy and repeatability of the preoperative corneal topography. The calculation of corneal SA requires highest precision of corneal topography in the periphery, since the difference in elevation between an aspheric corneal surface and a spherical surface is only some microns (**Figure 4**). Schröder et al. investigated the measurement repeatability and precision of several corneal topographers and tomographers and found that the repeatability of these devices is decreasing from the center to the periphery and may not be sufficient to detect small changes in corneal asphericity [38].

Another limitation arises from the concept of aberration correction itself. As outlined in **Figure 3**, the method requires the alignment of the IOL in relation to the cornea to be as perfect as possible. But even if an ideal positioning of the IOL is achieved intraoperatively, the risk of decentration or tilt remains in the postoperative course.

Altmann et al., Eppig et al., and others analyzed the effects of decentration and tilt of spherical and aspheric IOLs on the image quality and found that it is more affected by decentration than by tilt and that the susceptibility of lens misalignment increases with the amount of SA to be corrected [9, 39–46]. Some authors defined that a range of decentration within a SA-correcting IOL would perform better or equal than a standard spherical IOL. This range was reported to be between 0.0 and 0.3–0.8 mm, depending on the design of the lens and simulation conditions [9, 40–42]. In a previous publication, we summarized the data on the IOL decentration from various sources and found that the clinically observed decentration is between 0 and 1 mm but most frequently about 0.3 mm [33, 40, 47–56]. Others showed that there is a tendency for IOLs decentrating and tilting into nasal direction with mirror symmetry between both eyes [51].

Gillner et al. showed in a previous publication that IOL designs with a more conservative correction of SA may provide a larger range of tolerance to decentration [41]. Examples thereof are the ZO/ASPHINA design (Carl Zeiss Meditec AG, Berlin, Germany) and the Aspheric Balanced Curve Design (ABCD) (Hoya Corporation, Tokyo, Japan). Both designs are based on higher-order

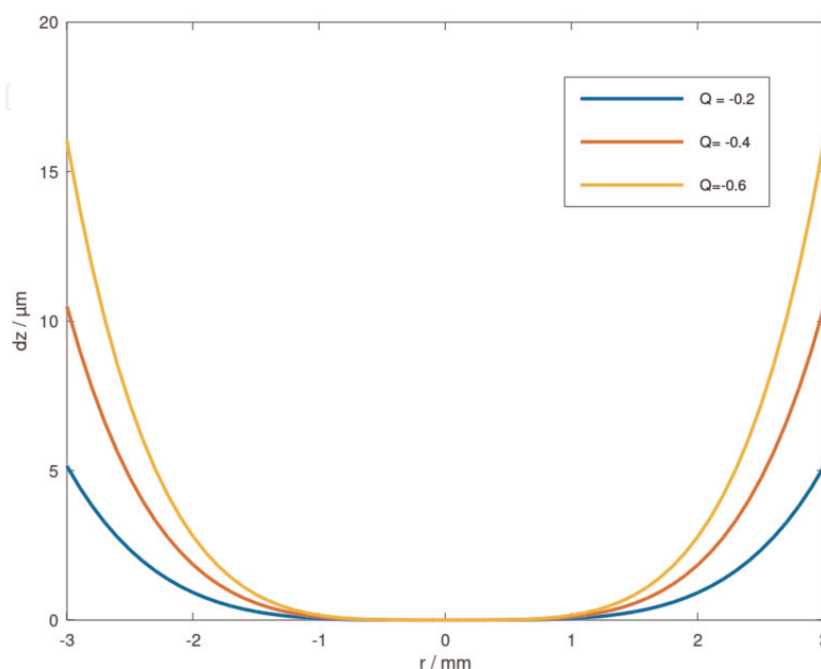


Figure 4. Difference in corneal elevation for three surfaces with $R = 7.77$ mm radius of curvature and several values of Q compared to a sphere.

aspherics including coefficients a_4 and higher (see Eq. 1) and were specifically designed considering some reasonable amount of IOL decentration. The effect of decentration on image performance of some selected IOL designs is shown in **Figures 5** and **6**. The graphs exhibit a drop of the image quality with aspheric lenses below the image quality of a spherical IOL when decentration exceeds 0.4 and 0.3 mm, respectively.

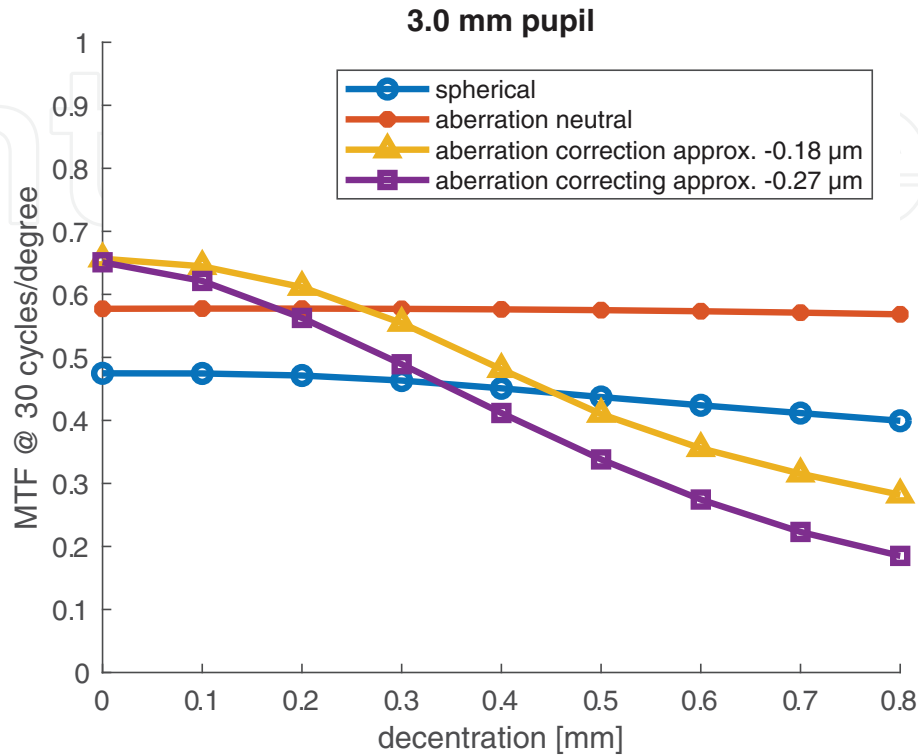


Figure 5. Simulation of modulation transfer function at 30 cycles per degree for four different intraocular lenses and a pupil diameter of 3.0 mm in the Liou-Brennan model eye as a function of decentration [40, 41, 57].

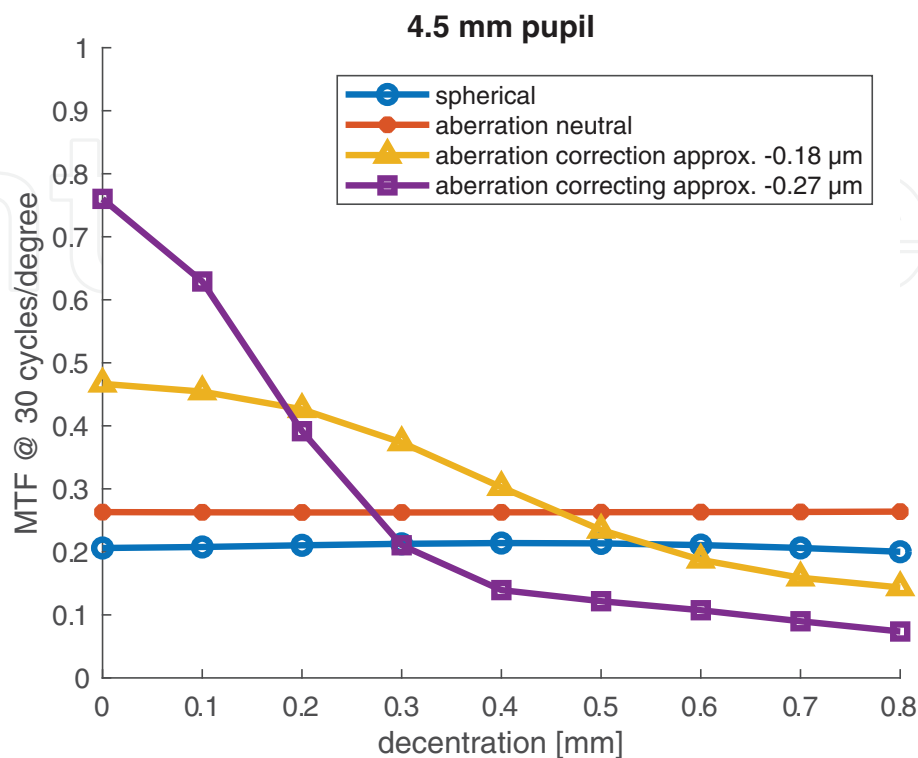


Figure 6. Simulation of modulation transfer function at 30 cycles per degree for four different intraocular lenses and a pupil diameter of 4.5 mm in the Liou-Brennan model eye as a function of decentration [40, 41].

6. Conclusions

Correcting the spherical aberration of the cornea by intraocular lenses may improve the visual outcome compared to standard spherical lenses. Especially patients with high aberrations after corneal refractive surgery may benefit from a reduction of the overall aberrations. However, the prospects for a 100% correction of SA or aiming to a residual SA of $+0.1 \mu\text{m}$ are limited with respect to an ideal and stable IOL. Therefore, any generic or customized IOL concept pursuing an aberration correction of aberrations, such as astigmatism, spherical aberration, coma, etc. must be designed with a tolerance according to the average expected misalignment in normal eyes (approximately 0–0.3 mm decentration and 0–3 degrees of tilt) [40, 57]. Consequently, this likewise limits the correctability of some higher-order aberrations. Eyes after corneal refractive surgery usually show very high values of SA and require special attention in the planning of cataract surgery. While eyes after myopic refractive procedures might benefit from a negative SA IOL [22], eyes after hyperopic refractive procedures often show high-negative SA and would require an IOL with positive SA for compensation [23]. Due to the high variability of SA in cataract patients, the “one-size-fits-all” approach may only provide optimum correction for a small amount of patients. Therefore, customized intraocular lenses tailored to correct for the individual spherical aberration may provide a better solution for a wide range of patients.

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