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Increased quality of vision by innovative intraocular lens and human eye modeling

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ABSTRACT

The accommodative ability of the natural eye lens is lost when it is replaced by an artificial intraocular lens in cataract surgery. While monofocal intraocular lenses are usually intended for distance vision, multifocal lenses allow for good vision at various distances, thus enabling the patient to live without glasses. In this work, several innovative multifocal intraocular lens design concepts are analyzed in terms of vision quality. Results from simulations of these lenses in different human eye models are compared to results from in-vitro measurements. We further demonstrate how the choice of the eye model impacts intraocular lens design and we show how biometric parameters that are usually not considered in intraocular lens power calculation may influence vision quality.

1. Introduction

When the crystalline lens of the human eye becomes opaque, currently the only treatment is cataract surgery involving extraction of the natural lens fibers and implantation of an intraocular lens (IOL) usually made of acrylic, silicone or poly(methyl methacrylate) (PMMA). Cataract surgery is well-established and is generally considered effective and safe in restoring good vision. However, the majority of implanted IOLs are monofocal allowing only far vision. Thus, most cataract patients require glasses for vision in near and intermediate distances. Nowadays many patients undergo surgery long before serious vision impairment occurs, leading to an increasing demand for independence from glasses. For this reason, multifocal IOLs have been developed which provide good vision at various distances, comparable to the accommodative ability of the natural lens. Multifocal IOLs simultaneously form two or more images on the patient's retina. One of the images has, for example, objects in far distance in focus while intermediate and near objects are out of focus. Another image has near objects in focus while objects in other distances are out of focus. Although the patient's brain adapts to picking the in-focus image, the intensity of this image is reduced compared to a monofocal IOL and, further, out-of-focus portions of other images are superimposed on the focused image. Thus, usually losses in contrast sensitivity for far vision occur in exchange for improved near vision. Most multifocal IOL designs put more light into the far focus than in the near and intermediate foci [1]. This accounts for the fact that far vision light intensity cannot be easily controlled, while it is often possible to increase near vision light intensity by using additional light sources. Several multifocal IOL design concepts have been proposed, which are mainly distinguished into refractive and diffractive concepts [2]. Refractive multifocal IOLs consist of at least two surface regions with different optical powers, which can be arranged for example concentrically, in radial sections (sectors), or even in hexagonally shaped regions. Pupil dependency is one of the major disadvantages of concentrically designed refractive multifocal IOLs. Refractive concepts are characterized by the fact that diffractive effects are negligible because of the small wavelength of light compared to the surface structures of the IOL. In diffractive multifocal IOLs the diffractive effects are produced by small and closely spaced grooves cut into the IOL surface. These grooves induce phase shifts and create thereby an infinite number of foci, although the main intensity lies within the first two or three orders of diffraction. By adjusting the spacing and the shape of the grooves, the optical properties of the IOL, e.g. the intensity distribution between far, intermediate and near focus, can be altered. In

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diffractive IOLs the drawbacks of refractive IOL designs, i.e. pupil dependency and uncontrolled phase shifts that appear due to the necessary transition regions between zones of different refraction, are avoided. This also leads to better separation of the light intensity into distinct peaks with increased optical image quality. One disadvantage of diffractive multifocal IOLs is the intensity loss due to light deflected into higher orders of diffraction. First-generation multifocal IOLs showed in many cases glare and night-vision problems [3]. These phenomena are alleviated in new multifocal IOLs, e.g. in diffractive concepts by reducing the groove height toward the periphery of the lens (apodization). Further, this new generation of multifocal IOLs decreases the number of patients who require spectacles after IOL surgery. New multifocal IOLs often make use of hybrid diffractive/refractive designs [1]. For the optical design of new IOLs it is essential to have accurate eye models representing the anatomic and optical behavior of the human eye. These models are used to predict and/or optimize various parameters that are linked to quality of vision, e.g. the modulation transfer function (MTF) and the residual spherical aberration (SA) [4,5]. Furthermore, eye models are used for prediction of the required dioptric power of an IOL in cataract surgery, which is of high importance, as a wrongly chosen dioptric power outpaces every other known aberration in loss of visual acuity.

2. Methods

A brief introduction into eye modeling encompassing biometric techniques and different modeling paradigms is given here. Further, the goals of IOL optimization are explained and methods for simulating the optical behavior of IOLs are introduced.

2.1. Biometric data acquisition

High quality experimental input data are required for modeling the optical properties of the human eye. These data are usually obtained by in-vivo biometric measurements using ultrasound waves or light waves that are reflected at the different layers (e.g. cornea, lens, retina) of the eye, giving information about the eye's structure. Optical biometry is typically noninvasive, while for the application of ultrasound mechanical contact for coupling the ultrasound source and receiver to the eye is needed. The axial eye length and locations of different structures can be determined using ophthalmic ultrasonography. In this technique, reflections of sound waves at ocular structures of different densities are detected. The resolution of this method is typically limited to 0.10 mm [6]. The accuracy of the technique is even less when the applanation technique is used, which often yields a falsely short axial eye length and widely scattered results due to varying degrees of corneal compression.

More recently, optical coherence biometry has been developed, which employs partially coherent light for measurement of the positions or distances between ocular structures [7,8]. Based on this technique the IOLMaster has been developed (Zeiss, Jena, Germany), a device that allows for noninvasive measurement of the axial eye length with an accuracy of 10–20 μm [6]. Optical coherence tomography uses partially coherent light for acquiring 2-dimensional tomograms or 3-dimensional volumes of semi-transparent ocular layers with high resolution, e.g. for the study and diagnosis of corneal or retinal diseases or for the measurement of retinal blood flow [9,10].

The radius of curvature of the anterior corneal surface is measured optically in Keratometry. A ring of known size is placed in front of the eye. The reflection of this ring forms a virtual image (the first Purkinje image) behind the cornea. The size of this image is related to the radius of curvature. If the cornea shows astigmatism,

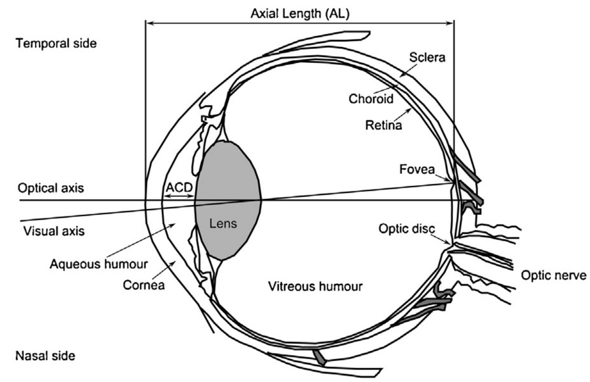

Fig. 1. *Schematic diagram of the human eye*. Horizontal section of the right eye as seen from above.

the ring image becomes elliptical, and the degree of astigmatism can be obtained by comparing the major and minor axes of the ellipse. Placido disk imaging works similar to Keratometry, but instead of one ring, several concentric rings are placed in front of the cornea to obtain the radius of curvature and astigmatism at several radial positions of the cornea. With this method, the 3-dimensional shape of the anterior corneal surface can be obtained. A profile through the anterior segment is given by Scheimpflug imaging, where a narrow slit is projected into the eye and imaged via a tilted lens onto a camera. The Scheimpflug technique allows for imaging the anterior and posterior surfaces of cornea and lens and for measurement of the anterior chamber depth (ACD).

2.2. Human eye modeling

A starting point to introduce eye modeling is a schematic human eye as shown in Fig. 1. It can be seen that an optical model of the eye should comprise the cornea, the crystalline lens, the retina, and the media in between these parts, i.e. the aqueous humour in the anterior chamber and the vitreous humour between lens and retina. As refraction always occurs on refractive index changes, the optical properties of the eye are determined by the surfaces and the distances between these surfaces. The majority of eye models take the following parameters into account: (1) the shape of the anterior and (2) posterior corneal surface, (3) the thickness of the cornea, (4) the distance to the crystalline lens or IOL, i.e. the ACD, (5) parameters of the natural lens or IOL (shape and refractive index gradient), (6) the distance to the retina, (7) the curvature of the retina, and (8) the refractive indices of all involved media. Moreover the different light detection properties of the fovea and off-axis regions should be included in the model. The fovea is the spot of highest photoreceptor density corresponding to sharpest vision and lies 4 to 8 degrees temporal to the optical axis. For more details on the optics of the human eye we refer to the recent literature [11,12].

The first step to actually finding a quantitative eye model is to fill the optical and anatomic parameters with as detailed biometric data as possible. This can be done within (1) a population based approach or (2) a personalized approach. The origin and complexity of the surface data and the targeted application are the key parameters defining the eye model. Most models decrease complexity by assuming rotationally symmetric surfaces. The classic eye model from Gullstrand has all surfaces spherical [13]. However, this model does not sufficiently describe the effect of spherical aberration of the cornea [4].

As a second step an eye model has to be validated and analyzed with regard to effects that can be approximated sufficiently. For this purpose, predictions from the model need to be compared to results from clinical studies. The principal problems in modeling