ПРИЛАДИ І СИСТЕМИ БІОМЕДИЧНИХ ТЕХНОЛОГІЙ

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ANALYSIS OF WAVE ABERRATIONS OF INTRAOCULAR LENSES (IOL), IMPLANTED INTO A PHYSICAL MODEL OF THE HUMAN EYE

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Wave aberrations of seven models of various intraocular lenses (IOL) which are located inside a physical model of the human eye are studied and compared. Statistically significant amplitudes of aberration modes of various IOLs are determined with the help of a ray tracing aberrometer on the basis of the difference in amplitudes of aberration modes of the human eye model with IOL and the human eye model without IOL. RMS values of IOL wave aberration are found on the basis of amplitudes of these modes. Increment in RMS values, caused by IOL decentring or IOL inclination within the eye, is revealed. Acceptable values of IOL decentring and inclination are found.
Keywords: *physical model eye, IOL aberrations, IOL aberrometry, IOL RMS, IOL decentring and inclination.*

Introduction

Implantation of intraocular lenses (IOL) into the human eye is an effective method for treatment of cataract. At the present time, there are a lot of uniform IOL models that are produced by various manufacturing companies and may be chosen by ophthalmologists. Therefore, an ophthalmologist is urged to choose the best IOL model. However, it is not always possible to make the right selection of an IOL model on the basis of those descriptions that are enclosed and that contain the data on the specific optical properties. These descriptions do not contain sufficient information on aberrational properties of specific IOL models, on the existing manufacturing errors, as well as on potential adverse effects of their incorrect implantation into the human eye. Many manufacturing companies produce the IOLs that were processed according to the procedure of aspherisation of their optical surfaces in order to decrease spherical aberration of the human eye's optical system, but this procedure raises the question concerning possible induction of the higher-order aberration modes by such IOLs. There remain questions about influence of IOL decentring relative to the cornea on the quality of retinal image, as well as about acceptable levels of such decentring.

The studies of aberration properties of various IOL models are described in works [1-6]. Authors of these studies have examined the influence of IOL aberrations on the quality of a retinal image with the help of various methods. In works [1-3] the method of calculation of path parameters of rays through a mathematical model of the eye was used with the help of such software applications as ZEMAX, OSLO, etc. The studies described in works [4-6] were conducted using physical models of the optical system of the eye, in which the studied IOL model was incorporated. The quality of retinal image in an optical system of the eye with an IOL was mainly evaluated with the help of the modulation transfer function (MTF) or the limit of spatial resolution. In addition, a method has been used for analysis of a retinal image of test objects in a form of radial test patterns/mires, Snellen chart signs [4], luminous dots [5], besides, the method is known of optical Fourier analysis of the light field distribution near the pupil of the eye model [6]. Special physical models of the eye [7, 8], as well as special equipment [8, 9] for IOL testing were developed in order to investigate IOL aberrations in accordance with the ISO 11979 standard.

Numerical methods of studying IOL aberrations with the help of computer simulation of the optical system of the eye can provide accurate results. Still, they idealize the IOL itself, therefore not allowing to establish the effect of actual IOL manufacturing errors (namely) optical heterogeneities of used materials, possible deviations from nominal geometrical, and optical parameters, shape distortions of optical surfaces) on quality of the retinal image.

Ophthalmologists have been often exposed to difficulties, when they tried to describe quality of the retinal image of the data concerning *MTF*, since sometimes they cannot adequately evaluate the extent, to which certain changes can affect visual acuity, when they are described in the form of graphs of these functions, caused by IOL aberrations. The same is valid to other characteristics, such as the point spread function (*PSF*).

In addition, the spatial resolution limit (as the specific parameter) does not always reveal aberration changes in the optical system of the eye caused by the IOL. In this way, two lenses, which have different aberration properties and the same values of aperture and lens power, can have the same resolution limit.

In view of the aforesaid, we suggest to use the root-mean-square (*RMS*) value of wave aberration in the area of the pupil as an integral evaluation of this function in order to ensure the comparative analysis of IOL aberration properties. Using *RMS* values is convenient for comparative evaluation of wave aberrations of different IOL models. It considerably simplifies establishing the criterion of acceptable values of errors in the course of IOL implantation into the human eye.

Research Technique

RMS values of wave aberration of to be studied IOL models can be determined from the results of their aberrometry. Wave aberrations of modern high-tech IOLs are very small linear values, commensurable with fractions of wavelengths. It is possible to identify and compare aberrations of such IOLs with the help of high-precision aberrometers, when the following conditions are met: 1) absolute stillness of the eye with an implanted IOL relative to the aberrometer during aberrometry; 2) there are no changes in the position of the eye with an implanted IOL relative to the aberrometer during repeated aberrometry sessions, which are necessary for statistical estimation of accuracy of the measurement results; 3) accurately controlled relocations of the IOL within the eye (IOL decentring and inclination); 4) stable state and operating conditions of the IOL, identical to real intraocular conditions, and 5) providing absolutely identical IOL position within the eye when changing IOL models.

There is no need to prove that it is impossible to implement all the above mentioned conditions in the eye of the same patient in vivo. These conditions can be performed when using a physical model of the eye adequately simulating its optical system.

IOL aberration properties depend on the crosswise sizes of the cone of rays on IOL surfaces, the convergence angle of the cone of rays falling on the IOL, the refractive index before and after the IOL. We have manufactured a device (Figure 1a), capable to reproduce such conditions. The model simulates the cornea, intraocular environment, and retina. The cornea is represented by a BK7 glass meniscus lens having spherical surfaces. The radii of its spherical surfaces and axis thickness correspond to average shape of the cornea. The device has special mechanisms and scales for accurate installation of IOL relative to the cornea, accurate IOL relocation along and perpendicularly to the optical axis, as well as for accurate IOL turns around one of the axes that is perpendicular to the optical axis.

The intraocular media in the model are represented by an aqueous solution of NaCl at the concentration providing coefficients of anterior and posterior chambers refractive index identical to those of the human eye. A diffusing polyethylene film that can depolarize the light of the aberrometer laser emitter reflected from it serves as retina simulator.

The aberrometry of the eye model was carried out using the TRACEY-VFA aberrometer (Figure 1b), which restores the wave aberration with the help of the ray tracing method [10].

Figure 1a. The eye model

Figure 1b. The TRACEY-VFA aberrometer

We received the data on IOL aberrations based on the results of aberrometry of models of the human eye with and without IOL. Aberrometry of the model without IOL provides the data on the amplitudes of aberration modes of the cornea lens. The data received from aberrometry of the model with IOL give an insight into the levels of amplitudes of the same modes modified by IOL.

It is known that the modes representing certain types of classical aberrations are independent from each other, and the aberration of the model of the human eye with IOL is the sum of the aberrations of the cornea lens and the IOL aberrations. This makes it possible to find the values of amplitudes of modes of the IOL itself based on the data on the values of mode amplitudes of the model without IOL and the data on the values of the same modes of the model with implanted IOL. With this aim in view, it is necessary to subtract the average (based on multiple aberrometry sessions) values of modes amplitudes of the model without IOL from the corresponding average (also based on multiple aberrometry sessions) values of modes amplitudes of the model with a centered IOL.

Centering the IOL in the eye model was achieved by superpositioning the centers of symmetry of images of the system of centering the LEDs of the aberrometer in the light reflected from cornea lens surfaces and IOL surfaces.

The resulting differences of the amplitudes of each mode of IOL models were tested for statistical significance. Student's t-test was used for this purpose. The null hypothesis was that the IOL, implanted into the eye model, does not change the amplitude of a particular aberration mode of a model of the eye without IOL. The null hypothesis is discarded at the significance level of $\alpha \leq 0.05$. Statistically significant differences were seen as the amplitudes of aberration modes of the IOL that are to be studied. The following table 1 provides the *RMS* calculation results, which were obtained by this method for the specified IOL models.

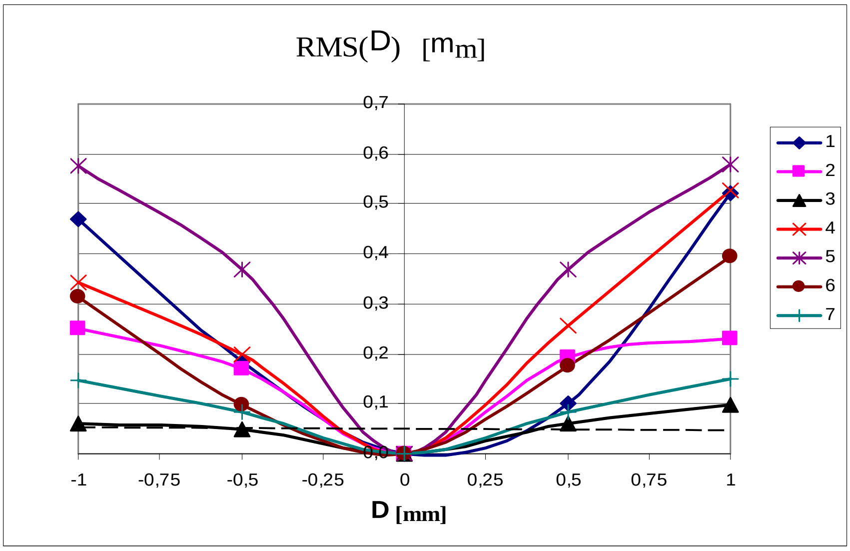
Table 1.

The company	**№**	**The IOL model**	**Lower *RMS* [μm]**	**Higher *RMS* [μm]**	**Total *RMS* [μm]**
Abbott Medical Optics, USA	1	ZCB00	0,116	0,139	0,181
Bausch & Lomb, USA	2	ADAPT-AO	0,167	0,048	0,173
Alcon Laboratories, USA	3	SN60AT	0,137	0,053	0,147
Alcon Laboratories, USA	4	SN60WF	0,240	0,077	0,252
1stQ GmbH, Germany	5	B1ADY0	0,389	0,104	0,403
НПП Репер-НН, Russia	6	МИОЛ-2	0,133	0,159	0,208
«US Optics», Ukraine	7	SL-907	0,044	0,044	0,062

The optical power of the IOL models listed in the table is 20 diopters, while aberrometry zone diameter in the cornea lens plane is equal to 4 mm.

Another factor that was investigated in the similar manner was the changes in the values of IOL aberration modes, which were determined by their decentring factor in respect to the optical axis of the cornea lens or IOL inclination to this optical axis. The changes in the amplitude values were determined by the difference in average values of the corresponding amplitudes of aberration modes of the model of the eye with IOL decentring or inclination and eye model with the same IOL, which was already centered. The *RMS* value calculated by increments of aberration modes amplitudes of wave aberration of the eye model served as the integrated estimate of eye model aberrations increase due to IOL decentring or inclination. It was conditionally considered that there is no wave aberration in the optical system of the model of the eye with a centered IOL.

The dependency of *RMS* value increments obtained in this way from the decentring value (Δ) or the rotation angle (φ) is presented in a form of the diagrams shown in Figure 2.



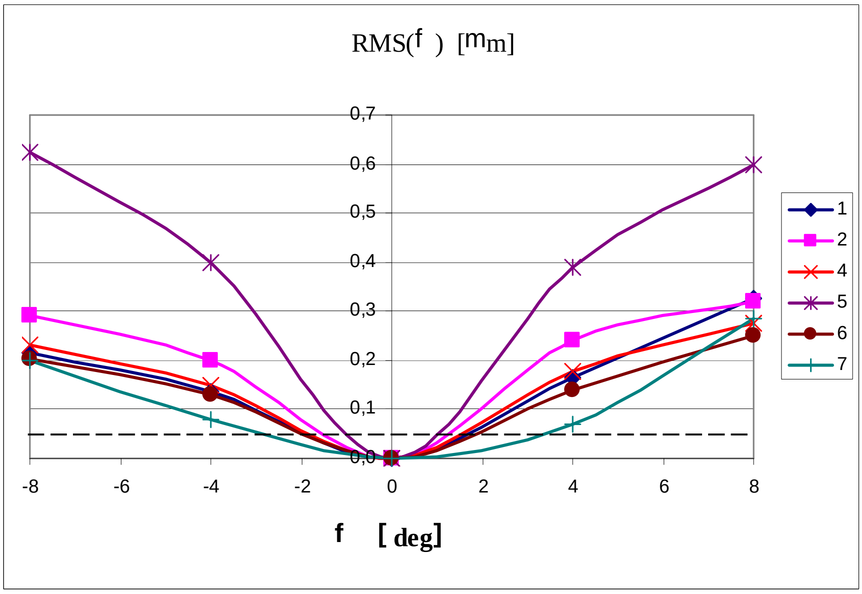


Figure 2. *RMS* increment graphs.

Results Discussion and Conclusions

The results of studying *RMS* of wave aberrations of implanted IOL models indicate the significant difference in their values. Model #7 has the smallest *RMS* value, while model #5 has the largest. It is significant that some IOL models with aspheric surfaces show smaller values of spherical aberration, this fact confirming their purpose. Still, in comparison with other models there may be observed some increased values of higher-order aberrations. Second-order aberration modes are predominant in virtually all IOL models.

RMS increment graphs, when such increment is caused by IOL decentring or inclination within the eye model, demonstrate significant differences in various IOL models as well.

The graphs shown in Figure 2 allow finding acceptable values of IOL decentring and inclination. In work [10], it is shown that the spatial resolution of two point light sources does not actually become worse in the optical system of the eye unless the increase of wave aberration is accompanied by *RMS* value increase by more than 0.1 λ, where λ is the wavelength. When $\lambda = 0.55$ microns, the condition $RMS \leq 0.055$ microns ensures the absence of changes in visual acuity evident to the eye. In the given graphs, the dotted horizontal line drawn at the height of 0.055 microns indicates acceptable limits of decentring and inclination of specific models of implanted IOL in the points, where the dotted line intersects with the charts. As it may be seen, the acceptable decentring of some IOL models ranges ±(0.1–0.3) mm and the acceptable inclination ranges $\pm(1.5^{o} - 3.5^{o})$.

The main conclusion should be as follows: the implanted IOL models should pass an additional third party inspection test. Such test will allow to determine actual aberration characteristics of the lenses. Taking into account the aberration characteristics of the IOL will allow more reasonable application of those in clinical practice, which can improve cataract treatment effectiveness and even restore a patient's normal or maximum possible visual acuity.

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РЕАЛІЗАЦІЯ МОДИФІКОВАНОЇ ФОТОДИНАМІЧНОЇ ТЕРАПІЇ З ВИБІРКОВИМ ЛАЗЕРНИМ СКАНУВАННЯМ ПУХЛИНИ ЗАЛЕЖНО ВІД НАЯВНОСТІ ФЛЮОРЕСЦЕНЦІЇ

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Розглянуто метод фотодинамічної терапії із вибірковим лазерним опроміненням в області флуоресценції пухлини. Запропонований метод полягає у визначенні границь флуоресценції із подальшим «поточковим» опроміненням сфокусованим лазерним пучком із фіксованою часовою затримкою в кожній локальній ділянці тканини, терапія пухлини відбувається лише у тій зоні, в якій наявна флуоресценція. Для реалізації даного методу використано нову лікувально-діагностичну установку на основі скануючого лазерного пристрою. Для реєстрації флюоресценції ФС пухлини використано кольорову відеокамеру SD на базі ПЗЗ-матриці, підключену до ПК. Програмне забезпечення лікувально-діагностичного комплексу реалізовано у вигляді авторського програмного пакету «ControLS». Дана діагностична система може дозволити ефективніше використовувати лазерне випромінювання та враховувати коливання концентрації ФС в пухлині, залежно від наявності флюоресценції в конкретний момент часу.

Ключові слова: *фотодинамічна терапія, лазерне сканування, флуоресцентна діагностика.*

Вступ

Зростання ракових захворювань в сучасному світі стимулює пошук нових прогресивних підходів до вирішення онкологічних проблем. Одним із відомих методів є фотодинамічна терапія (ФДТ), яка дозволяє поєднати діагностику та лікування світлом характерної довжини хвилі в одній процедурі при введенні у пухлину фотосенсибілізатора (ФС) [1-3]. Поглинання молекулами ФС квантів