Retina ray-tracing technique for eye-refraction mapping

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ABSTRACT

In photorefractive sight correction, pre-operational computations of to-be-ablated layers are usually based on information about cornea shape that is one of the causes of aberrations. To obtain high-quality results of operation, contributions to aberrations of other origins are to be taken into account. Technique of eye-aberration mapping has been investigated, we called retina ray-tracing. It consists in directing into the eye a narrow beam, scanned (translated) in parallel to itself. Computer controls trajectory of scanning. Beam projection (spot of light) is formed on the retina. Aberrations result in varying position of the spot on retina in the course of scanning. Deviations from initial position are measured and reconstructed into wave aberration function. Mathematical relations, using Zernike polynomial expansions, were found to transform these data into necessary cornea shape correction with ablation technologies. In our experimental setup, we used the technique of acousto-optic scanning with frame time less than 10 ms for 65 sensed points. Eye-aberration mapping is realized with optical power resolution 0.1 diopter.

Key words: photorefractive sight correction, retina ray-tracing, acousto-optic scanning, Zernike polynomials, ablation technology, eye-aberration mapping, wave aberration function.

1. INTRODUCTION

Two major factors are to be taken into account when studying eye aberrations: cornea shape and refraction distribution inside an investigated eye. Cornea shape measurement is one of the well-known problems for medical instrumentation [1, 2, 3]. Studies of eye non-homogeneity attracted attention of ophthalmologists earlier [4]. In the last years, new technology has been proposed, based on wave-front measurements using principles of adaptive optics [5, 6]. In this work, we study another approach called retina ray-tracing technique [7].

It is based on initial information about local aberrations yielding from local non-homogeneity, this information being generalized into wave-front deformation function, or wave aberration function, that is represented in polar pupil coordinates. It contains information on variable part of refraction, size and shape of the blur spot on retina and its intensity distribution, accommodation depth, physical sight acuity. Important from the practical point of view is information necessary to make computations of the required correction of the cornea shape in photorefractive sight operations based on laser ablation, that must take into consideration not only initial cornea shape, but also all other factors, including refraction non-homogeneity inside the eye.

2. CONCEPT OF RETINA RAY-TRACING TECHNIQUE

Fig. 1 illustrates main concepts of the technique [7]. Initial laser beam is shaped (due to square stop *8*) to have rectangular (quadrate) cross-section and is scanned over the studied area (elements *6, 7, 8, 1, 9, 10*). Alignment and control subsystem consists of elements *2, 3, 11 – 20*. It is responsible for alignment of the line of sight relative to the optical axis, and correct setting of the working distance. Signal detection and its preliminary amplification is done by the elements *12, 5, 21, 22*. Computer *23* is processing all the information and controls measurement procedures. Positioning of optical components (*1 – 5*) is clear from the positions of their front and back foci points ($F_1 – F_5$).

Laser beam is positioned consecutively into the specific sites of the lattice shown in fig. 2. Image of a laser spot, projected onto the eye fundus, is formed by objective lens *5* in the plane of coordinate-sensitive photodetector *21*, having four quadrants. It produces four output signals ($U_1 – U_4$), containing information about the position of light spot on eye fundus (fig. 3). If image centroid coincides with the center of photodetector's coordinate system, then equilibrium must take place: $U_1 = U_2 = U_3 = U_4$. Otherwise, transversal shift of the centroid results in shift signals:

$$\delta y' = \frac{(U_1 + U_2) - (U_3 + U_4)}{U_1 + U_2 + U_3 + U_4} \cdot \frac{b}{2} \quad , \tag{1}$$

$$\delta x' = \frac{(U_1 + U_4) - (U_2 + U_3)}{U_1 + U_2 + U_3 + U_4} \cdot \frac{b}{2} \quad . \tag{2}$$

Operations (1), (2) are performed with computer *23*, controlling and synchronizing also analog-to-digital converter *22* and deflector *7*.

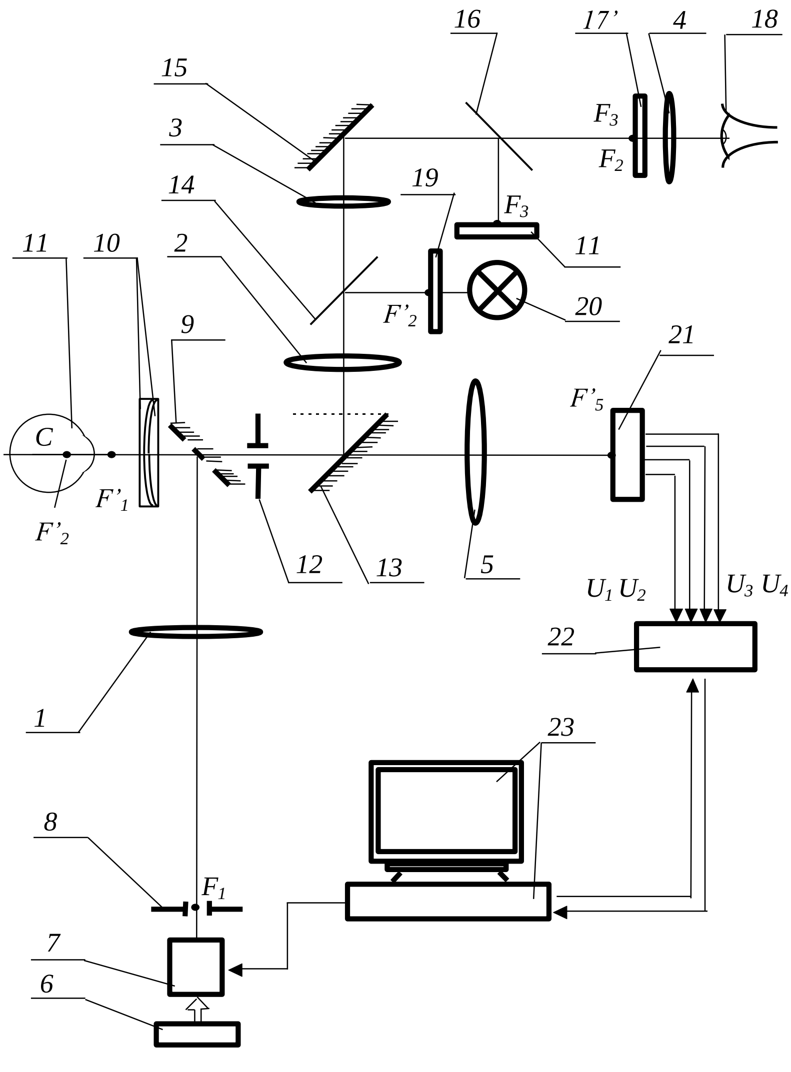

Fig. 1. Functional structure of the instrument

Mark *19* forms an image for fixating patient's sight. Mark *17* is needed to form an autocollimator image of the cornea surface. Patient's eye is considered to be centered and adjusted to the working distance along axis *z*, only if patient has fixated his sight onto the image of the mark *19*.

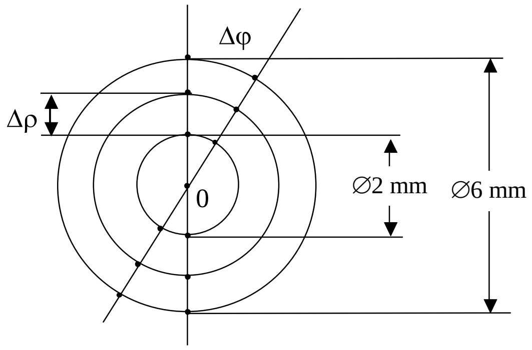

Fig. 2. System of polar coordinates for eye investigations

Once the eye centering procedure has been finished, measurements begin running. Laser beam scans through all the specific sites of the lattice, starting from point 0 and finishing with outer ring. Scanning is performed by deflector *7*. The total time of scanning through the whole lattice is 0.01 sec.

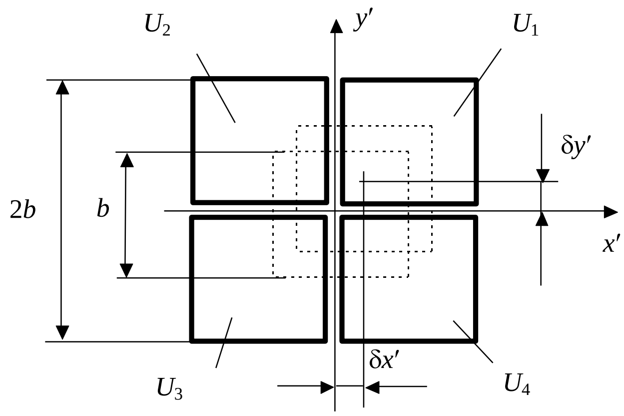

Fig. 3. Light spot displacement in the plane of photodetector

Quantity of specific sites (knots) in the lattice is defined by necessary spatial resolution (Δρ and Δφ in polar coordinates). Transition time of the beam from one position to another is less than $1.5 \cdot 10^{-5}$ sec. Spot of light on eye bottom will have transversal displacements δ_{xi} and δ_{yi} along corresponding axes, if optical system of an eye has aberrations. These displacements related to initial point are as follows:

$$\delta y_i = \frac{\beta}{2}\left[\frac{(U_{1i}+U_{2i})-(U_{3i}+U_{4i})}{U_{1i}+U_{2i}+U_{3i}+U_{4i}} - \frac{(U_{10}+U_{20})-(U_{30}+U_{40})}{U_{10}+U_{20}+U_{30}+U_{40}}\right]\cdot b \quad , \qquad (3)$$

$$\delta x_i = \frac{\beta}{2}\left[\frac{(U_{1i}+U_{2i})-(U_{3i}+U_{4i})}{U_{1i}+U_{2i}+U_{3i}+U_{4i}} - \frac{(U_{10}+U_{20})-(U_{30}+U_{40})}{U_{10}+U_{20}+U_{30}+U_{40}}\right]\cdot b \quad , \qquad (4)$$

where $\beta = -\, a_0' \,/\, f_5'$, a_0' is axial distance from back principal plane to the retina, f_5' is back focal distance of objective lens *5*. A set of measured values δy_i , δx_i is used to find wave aberration function of eye's optical system.

3. WAVE-FRONT DISTORTIONS

Let *A* be an object point, located on the line of sight of an ideal eye. Its image (point *A*'), formed by optical system of an ideal eye, will be located at the intersection point of an homocentric beam (fig. 4*a*). This point is also a center of spherical wave front W_E, shown in fig. 4*a* as a sphere tangential to the eye's cornea.

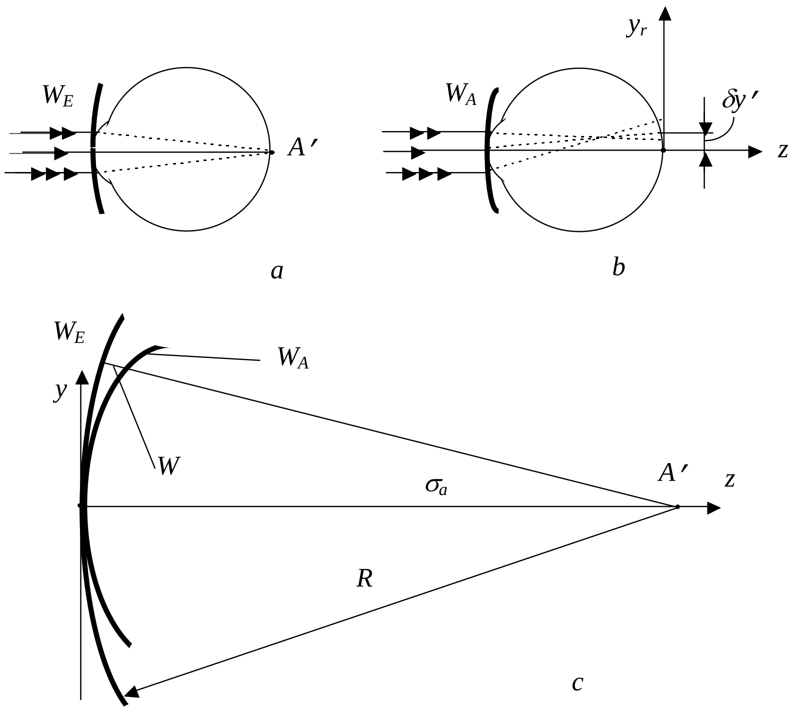

Fig. 4. Wave-front distortions

A real eye, having defects in its optical system, distorts homocentricity (fig. 4.1*b*). As a result, wave front W_A will deviate from sphericity. Fig. 4.1*c* represents wave front deviation *W*, being a distance between surfaces W_E and W_A along radius of the sphere W_E. If intersection point of the surface W_E with line of sight *z* at a plane perpendicular to the axis *z* is an origin of orthogonal coordinate system, then *W* for small aperture angles σ_a' may be described as a function $W=W(y, x)$. In polar system (ρ,φ), it can be represented as $W = W(\rho,\varphi)$, where $y = \rho\cdot\cos\varphi$, $x = \rho\cdot\sin\varphi$. Function *W* is wave aberration function of an optical system and is related to geometrical transversal ray aberrations $\delta y'$, $\delta x'$ in a simple way[8]:

$$\delta y' = \frac{R}{n'}\cdot\frac{\partial W(y,x)}{\partial y} \quad \text{and} \quad \delta x' = \frac{R}{n'}\cdot\frac{\partial W(y,x)}{\partial x} \quad , \qquad (5)$$

where *n*' is refraction index of vitreous (space of image). On the other hand, it is common to express functions $W(x, y)$ and $W(\rho, \varphi)$ in terms of power series expansion:

$$W(\rho,\varphi) = \sum_i \sum_j W_{ij}\cdot\rho^i \cos^j\varphi \quad ,$$

or in the form of Zernike polynomials [8, 10]:

$$W(\rho,\varphi) = \sum_n \sum_m C_{nm} R_n^m(\rho)\cdot\cos m\varphi \quad , \qquad (6)$$

where $n \geq m$, $n + m$ is an even number, C_{nm} are coefficients at polynomials $R_n^m(\rho)$, and

$$R_n^m(\rho) = \sum_{k=0}^{\frac{1}{2}(n-m)} (-1)^k \frac{(n-k)! \cdot \rho^{n-2k}}{k! \left[\frac{1}{2}(n-m)-k\right]! \cdot \left[\frac{1}{2}(n-m)-k\right]!} \quad . \tag{7}$$

Zernike orthogonal polynomial expansion is preferable in our case for two reasons. First, according to the Nijboer's classification, coefficients C_{nm} and polynomials $R_n^m(\rho)$ characterize both an order and type of aberration. For example, when $m = 0$, wave function describes spherical aberration caused by defocusing ($n = 2$), or spherical aberration of the third order ($n = 4$), and so on. When $m = 1$, we obtain distortion ($n = 1$), primary coma (third order coma, $n = 3$), and so on. Index $m = 2$ is present at astigmatism from lower to higher orders. Thus, non-zero index at C_{nm} certifies the fact of corresponding aberration, while the value C_{nm} itself permits computation of this aberration. Second, approximation of the function $W(\rho,\varphi)$ in terms of Zernike polynomials, based on the results of measurement of transversal beam translation on retina, is more accurate.

It should be noted, that expression (6) is used for description of wave aberrations of centered optical systems with axial symmetry. As far as transversal aberration function and wave function are even relatively the coordinate φ, (sin $m\varphi$) expansion terms are absent in expression (6). Optical system of a real eye has not an axial symmetry. That is why, functions of transversal aberrations and wave aberrations are not even relatively the coordinate φ, that is proved by numerous experimental investigations. Thus, $W(\rho,\varphi)$ expansion in terms of Zernike orthogonal polynomials may be written as follows [9]:

$$W(\rho,\varphi) = \sum_n \sum_m R_n^m(\rho) \left[C_{ynm} \cdot \cos m\varphi + C_{xnm} \cdot \sin m\varphi \right], \tag{8}$$

where coefficients C_{ynm} and C_{xnm} have the same physical meaning, but represent non-symmetry of aberration function in orthogonal directions.

4. ZERNIKE POLYNOMIALS AND TRANSVERSAL ABERRATIONS

Since Zernike polynomials are given in polar system of coordinates, corresponding transformation must be made with function $W(y, x)$ and partial derivatives. Let us consider this procedure in more details.

It is well known, that replacement of variables $x = x(\rho, \varphi)$, $y = y(\rho, \varphi)$ in the function $W = W(x, y)$ means, that following transforms with partial derivatives $\frac{\partial W}{\partial x}$ and $\frac{\partial W}{\partial y}$ are to be made:

$$\frac{\partial W}{\partial \rho} = \frac{\partial W}{\partial x} \cdot \frac{\partial x(\rho,\varphi)}{\partial \rho} + \frac{\partial W}{\partial y} \cdot \frac{\partial y(\rho,\varphi)}{\partial \rho} \quad \text{and} \quad \frac{\partial W}{\partial \varphi} = \frac{\partial W}{\partial x} \cdot \frac{\partial x(\rho,\varphi)}{\partial \varphi} + \frac{\partial W}{\partial y} \cdot \frac{\partial y(\rho,\varphi)}{\partial \varphi}.$$

Considering $\frac{\delta W}{\partial y}$ and $\frac{\partial W}{\partial x}$ as unknowns in the given system of equations and taking into consideration, that

$$\frac{\partial x(\rho,\varphi)}{\partial \rho} = \frac{\partial(\rho \cdot \sin\varphi)}{\partial \rho} = \sin\varphi,$$
$$\frac{\partial y(\rho,\varphi)}{\partial \rho} = \frac{\partial(\rho \cdot \cos\varphi)}{\partial \rho} = \cos\varphi,$$

$$\frac{\partial x(\rho,\varphi)}{\partial \varphi} = \frac{\partial(\rho \cdot \sin\varphi)}{\partial \varphi} = \rho \cdot \cos\varphi,$$
$$\frac{\partial y(\rho,\varphi)}{\partial \varphi} = \frac{\partial(\rho \cdot \cos\varphi)}{\partial \varphi} = -\rho \cdot \sin\varphi,$$

we obtain the following system of equations

$$\frac{\partial W}{\partial x} \cdot \sin\varphi + \frac{\partial W}{\partial y} \cdot \cos\varphi = \frac{\partial W}{\partial \rho},$$
$$\frac{\partial W}{\partial x} \cdot \rho\cos\varphi - \frac{\partial W}{\partial y} \cdot \rho\sin\varphi = \frac{\partial W}{\partial \varphi},$$

from which one can get, according to Kramer's rule,

$$\frac{\partial W}{\partial x} = \frac{\begin{vmatrix} \frac{\partial W}{\partial \rho} & \cos\varphi \\ \frac{\partial W}{\partial \varphi} & -\rho\sin\varphi \end{vmatrix}}{\begin{vmatrix} \sin\varphi & \cos\varphi \\ \rho\cos\varphi & -\rho\sin\varphi \end{vmatrix}} = \frac{-\frac{\partial W}{\partial \rho} \cdot \rho\sin\varphi - \frac{\partial W}{\partial \varphi} \cdot \cos\varphi}{-\rho(\sin^2\varphi + \cos^2\varphi)},$$

$$\frac{\partial W}{\partial y} = \frac{\begin{vmatrix} \sin\varphi & \frac{\partial W}{\partial \rho} \\ \rho\cos\varphi & \frac{\partial W}{\partial \varphi} \end{vmatrix}}{\begin{vmatrix} \sin\varphi & \cos\varphi \\ \rho\cos\varphi & -\rho\sin\varphi \end{vmatrix}} = \frac{\frac{\partial W}{\partial \varphi} \cdot \sin\varphi - \frac{\partial W}{\partial \rho} \cdot \rho \cdot \cos\varphi}{-\rho(\sin^2\varphi + \cos^2\varphi)}.$$

Finally,

$$\frac{\partial W(y,x)}{\partial y} = \cos\varphi \cdot \frac{\partial W(\rho,\varphi)}{\partial \rho} - \frac{\sin\varphi}{\rho} \cdot \frac{\partial W(\rho,\varphi)}{\partial \varphi}, \tag{9}$$

$$\frac{\partial W(y,x)}{\partial x} = \sin\varphi \cdot \frac{\partial W(\rho,\varphi)}{\partial \rho} + \frac{\cos\varphi}{\rho} \cdot \frac{\partial W(\rho,\varphi)}{\partial \varphi}. \tag{10}$$

Substituting (9) and (10) into (5), we obtain

$$\partial y' = \frac{R}{n'}\left[\cos\varphi \cdot \frac{\partial W(\rho,\varphi)}{\partial \rho} - \frac{\sin\varphi}{\rho} \cdot \frac{\partial W(\rho,\varphi)}{\partial \varphi}\right], \tag{11}$$

$$\partial x' = \frac{R}{n'}\left[\sin\varphi \cdot \frac{\partial W(\rho,\varphi)}{\partial \rho} + \frac{\cos\varphi}{\rho} \cdot \frac{\partial W(\rho,\varphi)}{\partial \rho}\right]. \tag{12}$$

According to equations (7) and (8), partial derivatives of the wave front function $W(\rho,\varphi)$, expressed in terms of Zernike polynomials, will have the following form:

$$\frac{\partial W(\rho,\varphi)}{\partial \rho} = \sum_n \sum_m \left[C_{ynm} \cdot \cos m\varphi \; + \; C_{xnm} \cdot \sin m\varphi \right] \cdot \frac{\partial R_n^m(\rho)}{\partial \rho} \quad , \tag{13}$$

$$\text{where } \frac{\partial R_n^m(\rho)}{\partial \rho} = \sum_{k=0}^{\frac{1}{2}(n-m)} (-1)^k \frac{(n-k)! \cdot (n-2k) \cdot \rho^{n-2k-1}}{k! \cdot \left[\frac{1}{2}(n+m)-k\right]! \cdot \left[\frac{1}{2}(n-m)-k\right]!} , \tag{14}$$

$$\frac{\partial W(\rho,\varphi)}{\partial \varphi} = \sum_n \sum_m m R_n^m(\rho) \left[C_{xnm} \cdot \cos m\varphi - C_{ynm} \cdot \sin m\varphi \right] \; . \tag{15}$$

Making substitutions in expressions (11) and (12) from (13), (14), and (15), we obtain equations, where transversal aberration values are on the left side, and polynomial expansions in terms of coordinates ρ, φ - on the right side. Expressions (11) and (12) give dependencies of the polynomial coefficients C_{ynm} and C_{xnm} on the values of $\delta y'$, $\delta x'$, determining wave aberration function $W(\rho, \varphi)$ of the optical system of an eye, and enabling evaluation of the contribution of any aberration, computation of the cornea profile to be ablated, etc.

5. COMPUTATION OF POLYNOMIAL COEFFICIENTS

Values of $\delta y'$ and $\delta x'$ are functions of retina ray-tracing coordinates (ρ, φ), or coordinates of crossing the retina by a ray. Having measured $\delta y'$ and $\delta x'$ at each specific site of a lattice, described by its coordinates ρ_k, φ_k, where $k = 1....q$ (in our case $q = 65$), we get the data for left part of equations (11), (12). Thus, we have a system of $2q$ linear equations relatively the coefficients C_{ynm} and C_{xnm}. The number of polynomial terms is defined by accuracy necessary for description of the function $W(\rho, \varphi)$. In other words, the number of coefficients C_{ynm} and C_{xnm} is determined by values of n and m.

The number of unknowns C_{ynm} and C_{xnm} in above mentioned system of $2q$ equations may be not equal to $2q$. As a rule, it is smaller. Therefore, the task of wave-front interpolation is transformed into the task of approximation, that may be solved using least-square method. These coefficients can be found by multiplying corresponding matrices [12]:

$$\mathbf{C} = (\mathbf{A}^T \mathbf{E}\, \mathbf{A})^{-1} \mathbf{A}^T \mathbf{E}\, \mathbf{F}, \tag{16}$$

where **C** is a column matrix with unknown coefficients C_{ymn}, C_{xmn} and $\begin{bmatrix} a_{11} & a_{12} & \cdots & a_{1t} \\ a_{21} & a_{22} & \cdots & a_{2t} \\ . & . & \cdots & . \\ a_{s1} & a_{s2} & \cdots & a_{st} \end{bmatrix} = \mathbf{A}$ is a matrix of numerical coefficients for unknowns C_{ymn}, C_{xmn} in the system of $s = 2q$ linear equations, based on expressions (11) and (12); $\mathbf{A}^T$ is a transposed matrix **A**; **E** is a unit matrix, **F** is a column matrix of measured transversal aberrations $\delta y'$ and $\delta x'$, containing s elements. Matrix **A** is to be calculated only once for all given combination of numbers m and n. Diagonal matrix with weight coefficients may be used instead of matrix **E**, reflecting degree of confidence to the accuracy of aberration measurements. All matrix operations according to the expression (16) are computerized.

6. APPLICATION AND RESULTS

In our practical application, measurements with the instrument of fig. 1 are oriented on cornea profile computations necessary for highly accurate photorefractive operation of keratectomy, including Lasik operation [13]. Having measured values δy_i, δx_i, we use them to find wave aberration function $W(\rho, \varphi)$ of eye's optical system and polynomial coefficients

C_{ynm} and C_{xnm}. This function describes wave front increment, caused by aberrations, and is responsible for point image formation (fig. 5*a*).

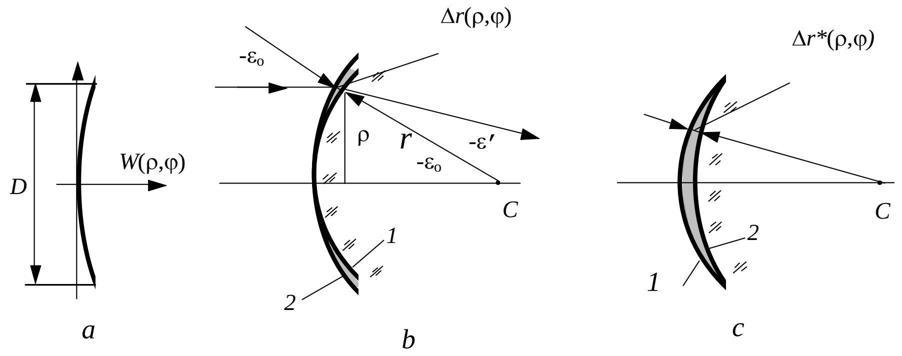

Fig. 5. Computation of ablation profile: *1* - cornea shape before correction, *2* - cornea profile after correction

To compensate for distortions occurring in the eye due to aberrations, theoretically two ways are possible. First one could consist in building up the cornea in its thickness Δr (fig 5*b*) dependent on coordinates (ρ, φ) [14]:

$$\Delta r(\rho,\varphi) = \frac{W(\rho,\varphi)}{k}, \tag{17}$$

where $k = n_c \cdot cos\,\varepsilon_0' - cos\,\varepsilon_0 = \sqrt{n_c^2 - (\rho/r)^2} - \sqrt{1-(\rho/r)^2}$, n_c - cornea refraction, r - radius of the outer corneal surface.

The second solution, that we use in our practice, consists in cornea upper layer ablation with excimer (or other type) laser in the circular zone with its diameter *D*, at the periphery of which $\Delta r^*(\rho,\varphi) = 0$ (fig. 5c). We get it as

$$\Delta r^*(\rho,\varphi) = \Delta r(\rho,\varphi) - \frac{W(\rho = 0.5\cdot D,\varphi)}{k_0}, \tag{18}$$

where $k_0 = \sqrt{n_c^2 - D^2/4r^2} - \sqrt{1 - D^2/4r^2}$.

For n_c = 1.376, r = 7.98 mm and variations of ρ from 0 to 3 mm, denominator in (17) varies insignificantly within the limits from 0.376 to 0.397 correspondingly, that can be regarded as constant. At D = 6 mm, k_0 = 0.397.

7. DISCUSSION AND CONCLUSIONS

Integral action of cornea shape and eye refraction non-homogeneity, leading to image distortions, has been studied using the technique called retina ray-tracing. Initial information on light spot displacements on retina is reconstructed into wave aberration function. This function is used for computations of necessary cornea ablation to correct the sight.

We have got accuracy of aberration measurement equivalent to optical power deviation as high as 0.1 diopter. Limiting factors are photodetector noise, and corresponding signal-to-noise ratio that is restricted by maximal allowable light intensity on retina. Polar system of coordinates is optimal, as far as circular shapes are studied. Our investigations showed, that wave aberration function of an eye is rather plain. It means that limited number of measurements is needed to reconstruct the function. We use 65 points in 4 circles (16 in each) plus central point. Interpolation algorithms have been developed for its reconstruction.

We plan to continue our theoretical and experimental studies to optimize measurement errors and computation procedures from the point of view of polynomial length. We plan also to reconstruct in optimal way other ophthalmologic parameters of an eye, necessary to estimate its functions, and to choose right strategy in eye treatment and, particularly, in sight correction. Influence of eye aberrations in peripheral zone is also of interest, its study is in the scope of our future work.

8. ACKNOWLEDGMENTS

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