Influence of eye micromotions on spatially resolved refractometry

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ABSTRACT

The influence eye micromotions on the accuracy of estimation of Zernike coefficients from eye transverse aberration measurements was investigated. By computer modeling, the following frequently found eye aberrations have been examined: defocusing, primary astigmatism, spherical aberration of the 3rd and the 5th orders, as well as their combinations. It was determined that the standard deviation of estimated Zernike coefficients is proportional to the standard deviation of angular eye movements. Eye micromotions cause the estimation errors of Zernike coefficients of present aberrations and produce the appearance of Zernike coefficients of aberrations, absent in the eye. When solely defocusing is present, the biggest errors, cased be eye micromotions, are obtained for aberrations like coma and astigmatism. In comparison with other aberrations, spherical aberration of the 3rd and the 5th orders evokes the greatest increase of the standard deviation of other Zernike coefficients.

Key words: eye micromotions, wave-front deformation, eye transverse aberrations, Zernike polynomials, astigmatism, spherical aberration, spatially resolved refractometer

1. INTRODUCTION

Laser surgical sight correction techniques as photorefractive keratectomy and laser keratomiles are widely applied in recent years[1-3]. They enable to correct ametropia and/or astigmatism and to improve sight acuity by changing a shape of the anterior cornea surface.

To perform the cornea surgery, the information is required about the eye's sight characteristics. A function of wave aberration is the most important one, because it enables to determine the shape of to-be-deleted cornea layer and to evaluate ametropia and astigmatism parameters, as well as sight acuity.

It is possible to estimate the function of wave aberration with the help of spatially resolved refractometers. These devices are intended to measure eye refraction at small pupil zones (measurement points). Such zones are not overlapped (in other words, they are spatially separated).

Achievements of new laser surgical techniques enabled to perform the more precise surgical correction of the comea shape in a custom, flexible manner. Therefore, the necessity in accurate measurement of a spatial refraction distribution in preoperative and postoperative period has appeared.

Obviously, the measurement of a spatial refraction distribution of an eye, which continuously changes its orientation, is accompanied by errors. Indeed, such random micromotions of the eye as tremor, jumps, and drift[4] are the factors, limiting the measurement accuracy. For this reason, a purpose of the present paper is to determine the influence of eye micromotions on sight characteristics estimation errors and to work out practical recommendations for increasing the accuracy of spatially resolved refractometry.

2. THEORY

To evaluate sight parameters, the wave aberration function with respect to the visual axis is to be estimated. For this purpose, let us consider a thin light beam, exiting from an axial object point and being set sequentially at all measurement points with pupil coordinates ρ_j, φ_j (fig. 1).

In a general case, the light spot intersects the retina plane at a point with coordinates ΔX_j, ΔY_j (point *A* in fig. 1). These coordinates are transverse aberrations of the beam for the *j*-th measurement point. If the eye were an ideal optical system (i. e., without aberrations), the light spot would be placed on the visual axis and ΔX_j, ΔY_j would be equal to zero for all measurement points (for all indexes *j*). The beam trajectory for this case is shown in fig. 1 by a dashed line.

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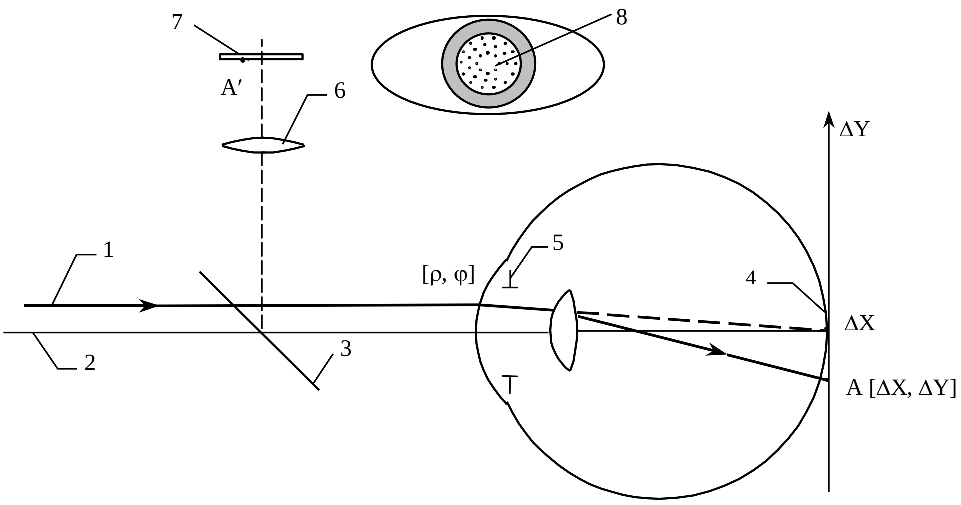

Fig. 1 Functional diagram of a single-beam spatially resolved refractometry: *1* – light beam; *2* – visual axis; *3* – beamsplitter; *4* – retina; *5* – pupil; *6* – lens; *7* – photodetector; *8* – measurement point at the pupil plane

Since the eye possesses aberrations, the point *A* shifts at the retina plane. This displacement depends on coordinates ρ_j, φ_j. Values $\Delta X_j = \Delta X(\rho_j, \varphi_j)$ and $\Delta Y_j = \Delta Y(\rho_j, \varphi_j)$ can be determined by photo-electric two-coordinate meter *3, 6, 7* (fig. 1), measuring a position of the light spot image (point *A*) on the photodetector's light-sensitive surface. Thus, a set of ΔX_j and ΔY_j is obtained, enabling to estimate the eye's wave aberration function and other parameters of sight imperfections. For this purpose, the following technique may be used:

1. The wave aberration function $W(\rho_j, \varphi_j)$ is assumed to be expanded in terms of Zernike polynomials[5]:

$$W(\rho_j, \varphi_j) = \sum_n \sum_m R_n^m(\rho_j)\left[C_{nm} \cdot \cos m\varphi_j + S_{nm} \cdot \sin m\varphi_j\right], \tag{1}$$

where *R* is a radius of a reference sphere, having a center at the origin of retinal coordinate system (the axes ΔX_j and ΔY_j) (fig. 2); *j* is an index of a pupil measurement point; $n' = 1.337$ is a refraction index of the vitreous humor; R_n^m are radial Zernike polynomials[5]; C_{nm}, S_{nm} are coefficients of Zernike polynomials.

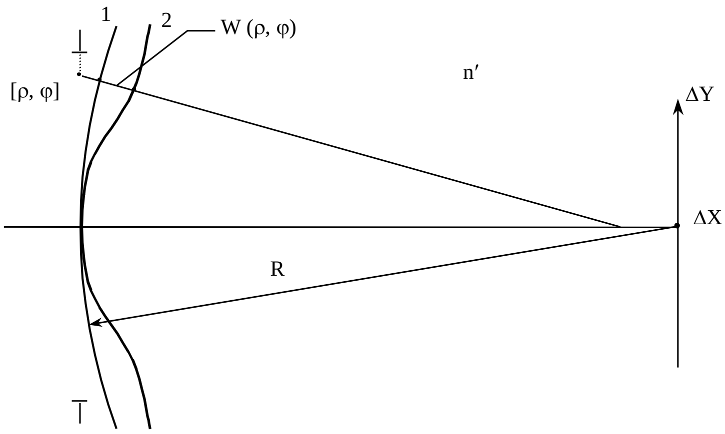

Fig. 2 Wave-front deformation of the eye's optical system:
1 – reference sphere with a center at the retina plane, *2* – wave front, deformed by aberrations

2. By using relations[5] $\Delta X = \frac{R}{n'}\frac{\partial W}{\partial X}$ and $\Delta Y = \frac{R}{n'}\frac{\partial W}{\partial Y}$, one can get a set of equations (2):

$$\begin{aligned} &\frac{R}{n'}\left(\frac{\partial W(\rho,\varphi)}{\partial\rho}\cdot cos\,\varphi - \frac{\partial W(\rho,\varphi)}{\partial\varphi}\cdot\frac{sin\varphi}{\rho}\right)\Bigg|_{\rho=\rho_j,\varphi=\varphi_j} = \Delta X_j; \\ &\frac{R}{n'}\left(\frac{\partial W(\rho,\varphi)}{\partial\rho}\cdot sin\varphi + \frac{\partial W(\rho,\varphi)}{\partial\varphi}\cdot\frac{cos\,\varphi}{\rho}\right)\Bigg|_{\rho=\rho_j,\varphi=\varphi_j} = \Delta Y_j, \end{aligned} \tag{2}$$

In this set, the C_{nm} and S_{nm} coefficients are unknown. The ΔX_j and ΔY_j aberrations are measured by a spatially resolved refractometer.

3. The C_{nm} and S_{nm} coefficients may be found by the least squares method, according to known formula:

$$\mathbf{C} = (\mathbf{A}^{\mathbf{T}}\ \mathbf{A})^{-1}\mathbf{A}^{\mathbf{T}}\ \mathbf{X} = \mathbf{B}\,\mathbf{X}, \tag{3}$$

where **C** is a vector of unknown coefficients C_{nm} and S_{nm}; **A** is a matrix with dimensions $2q \times t$, whose elements are numerical factors at $\boldsymbol{C_{nm}}$ and $\boldsymbol{S_{nm}}$ in the set (2), t is a total number of the vector **C** elements; **X** is a vector, comprising right parts of the equation set (2).

4. According to expression (1), the $W(\rho_j, \varphi_j)$ function is obtained. Then parameters of ametropia (mean power) and astigmatism may be calculated as follows [6]:

$$A_D = \frac{8 \cdot C_{20}}{D}, \tag{4}$$

$$|A_s' - A_m'| = \frac{8}{D}\sqrt{C_{22}^2 + S_{22}^2}\ , \tag{5}$$

where A_D and $|A_s' - A_m'|$ are the ametropia and primary astigmatism values in diopters, respectively; D is a diameter of the entrance pupil in mm, C_{20}, C_{22}, and S_{22} are Zernike coefficients in µm.

From the equations (2) and (3), one can find that the measurement errors (i. e., errors of the vector **X** elements) lead to errors of C_{nm}, S_{nm} estimation. These errors define the accuracy of the $W(\rho_j, \varphi_j)$ function estimation as well as the A_D and $|A_s' - A_m'|$ parameters evaluation.

To investigate the influence of eye micromotions on the accuracy of transversal aberration measurement, let us consider fig. 3, in which an ametropic eye is shown. The thin light beam (the beam *1*) enters the eye in parallel to the visual axis. If the visual axis of the eye and the optical axis of the refractometer coincide, then due to ametropia $(\Delta \neq 0)$ the beam will hit the retina at point A (fig. 3). Its image A' on the photodetector's light-sensitive surface will be located at distance $F_l' A' = f_l' \cdot \varepsilon_0$ from the optical axis. Here f_l' is the focal length of lens *3*. If we divide the indicated distance into linear magnification $\beta = f_l' / f$, we shall get a value of segment $AF_0' = \Delta Y_j$, where f is the front focal length of the eye. Similarly, one can determine the ΔX_j value. Exit angle ε_0 of the light beam can be found from equation $\varepsilon_0 = \frac{\Delta Y_j}{S+R}$, because the reflected ray passes through the eye's nodal points N, N'. Taking in account that $-\frac{\Delta Y_j}{h} = \frac{\Delta}{f_a'}$, one can get the expression for ε_0:

$$\varepsilon_0 = -\frac{h \cdot \Delta}{(S+R) \cdot f_a{}'}. \tag{6}$$

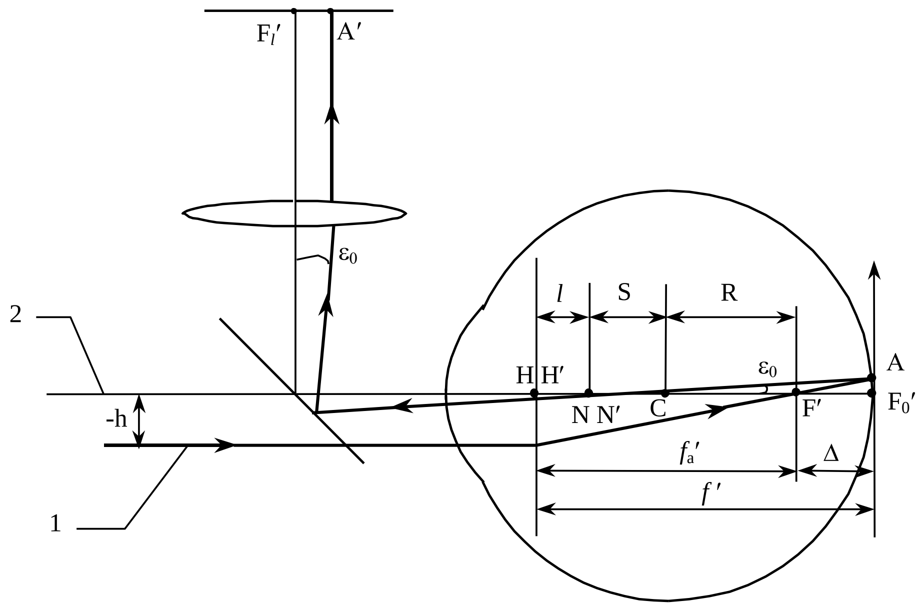

Fig. 3. Position of a light spot image at planes of the retina and the photodetector's light-sensitive surface in the ametropic eye: *1* – light beam; *2* – visual axis; *H, H′* – principal points; *N, N′* – nodal points; f_a', f' – back focal length of the ametropic and emmetropic eye, respectively; *C* – eye rotation center; *F′*, F_0' – focal points of the ametropic and emmetropic eyes, respectively

Let the eye turned (due to micromotions) around its geometric center *C* by some small angle ψ and the center of turn remained on the optical axis of a refractometer (fig. 4).

To determine an intersection point of the ray *1* with the retina plane, an auxiliary ray is to be used. Such a ray (the ray *3*) enters the eye in parallel to the ray *1*, passes through the nodal points N, N', and intersects the back focal plane of the ametropic eye at the point *O*. The ray *1* passes through this point too, but it intersects the retina at the point *A* (fig. 4). The retinal surface is shown in fig. 4 as a plane, perpendicular to the visual axis, because the actual angle ψ is very small. The ray, reflected from the retina, passes through the nodal points N, N' and defines a position of the point A'_ψ (fig. 4).

It is obvious from fig. 4 that for small values of angle ψ one can determine the ε_ψ value from equation $\varepsilon_\psi \cong \frac{AD}{DN'} = \frac{AD}{S+R}$. Since the *ADO* and *OLM* triangles are geometrically similar, the expression $\frac{AD}{LM} = \frac{\Delta}{f_a{}'}$ is valid. On the other hand, *LM* = *MK* + *KL*. Besides, *MN* ≅ -*h* and $KL = KH - LH = (l+S) \cdot \psi - l \cdot \psi = S \cdot \psi$ for small values ψ. Then $LM = -h + S \cdot \psi$, $AD = \frac{\Delta}{f_a{}'}(S \cdot \psi - h)$, and

$$\varepsilon_\psi = \frac{(S \cdot \psi - h) \cdot \Delta}{(S+R) \cdot f_a{}'}. \tag{7}$$

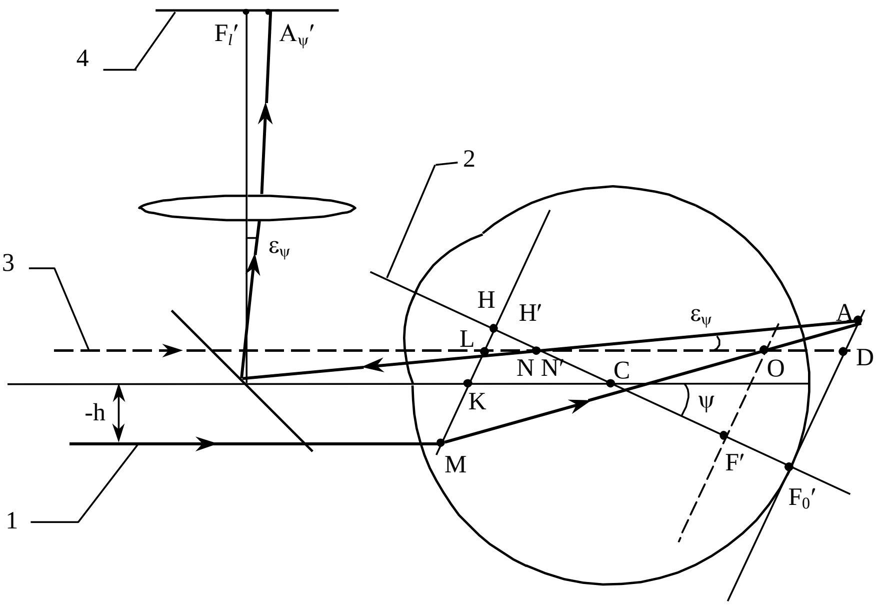

Fig. 4 Position of light spot image at the planes of the retina and the photodetector’s light-sensitive surface in the rotated ametropic eye: *1* – light beam; *2* – visual axes; *3* – auxiliary ray; *4* – photodetector’s light-sensitive surface

The $(\varepsilon_\psi - \varepsilon_0)$ value, multiplied by segment $(S + R)$, gives the aberration measurement error $\partial\Delta\mathrm{Y}$ of the ray *1* at the retina in the meridional plane. The $\partial\Delta\mathrm{X}$ error is determined by a similar way.

Taking into account the equations (7) and (6), one can get that

$$\partial\Delta\mathrm{Y} = (\varepsilon_\psi - \varepsilon_0)\cdot(S + R) = \frac{S\cdot\psi}{f_a'}\cdot\Delta \qquad (8)$$

The analysis of the formula (8) enabled to draw the following conclusions:

1. The $\partial\Delta\mathrm{Y}$ error is a consequence of non-coincidence of the eye nodal points N, N' with the eye rotation center C $(S \neq 0)$ and presence of ray aberration $(\Delta \neq 0)$.
2. The $\partial\Delta\mathrm{Y}$ error is absent if the eye has no aberrations for the ray *1* $(\Delta = 0)$.
3. If $\Delta = const$ for all measurement points, then the $\partial\Delta\mathrm{Y}$ error does not depend on height *h* of the ray,.
4. If the Δ magnitude depends on coordinates ρ_j and φ_j, i.e., $\Delta = \Delta(\rho_j, \varphi_j)$, then the $\partial\Delta\mathrm{Y}$ error is proportional to the Δ distance. In this case, $\partial\Delta\mathrm{Y}$ is not a constant and depends on *j* even if $\varphi = const$.
5. According to fig. 3, $\Delta Y = AF_0' = -\frac{h\cdot\Delta}{f_a'}$. Therefore, by using the equation (8), one can estimate the relative error $\partial\Delta\mathrm{Y}/\Delta\mathrm{Y}\cdot 100\%$:

$$\partial\Delta\mathrm{Y}/\Delta\mathrm{Y}\cdot 100\% = -\frac{S\cdot\psi}{h}\cdot 100\%. \qquad (9)$$

It is evident from (9) that this error does not depend of ray aberration value.

The formulas (8) and (9) enable to evaluate quantitatively the errors in a general case. Particularly, for the standard eye (with $f' =$ 22.89 mm, $S =$ 5.12 mm) at h = -1 mm, $\psi = 20' = 5.8\times10^{-3}$ radians, $\Delta_1 =$ 1.12 mm (equivalent to ametropia of 3 D), and Δ_2 = 2.13 mm (equivalent to ametropia of 6D), the $\partial\Delta Y$ error, caused by angular eye turn by angle ψ, has values: $\partial\Delta Y_1 = \pm$ 1.5 μm, $\partial\Delta Y_2 = \pm$ 3.05 μm, $\partial\Delta Y/\Delta Y\cdot 100\%$ =3%. Then absolute errors of ametropia evaluation equal to 0.09 D and 0.18 D, respectively. Obviously, such errors are essentially non-zero.

The carried out calculations enabled to draw one more important conclusion. The jumps of the eye's axis and drift, which take units up to tens of angular minutes, may influence essentially on the accuracy of ΔX_j, ΔY_j evaluation. Tremor, having an angular range of about units up to tens of angular seconds, should not be considered as a factor affecting substantially on the indicated accuracy.

On the other hand, the effect of eye micromotions can be considered as indeterminacy of coordinates of the ray *1* over the pupil. When $\Delta = \Delta(\rho_j, \varphi_j)$, the aberrations of the ray *1* at a measurement point with actual coordinates $(\rho_j + \partial\rho_j, \varphi_j + \partial\varphi_j)$, $(\rho_j + \partial\rho_j, \varphi_j + \partial\varphi_j)$ are referred to a measurement point with coordinates ρ_j, φ_j. This causes the following errors:

$$\begin{aligned} \partial\Delta X_j &= \Delta X(\rho_j + \partial\rho_j, \varphi_j + \partial\varphi_j) - \Delta X(\rho_j, \varphi_j), \\ \partial\Delta Y_j &= \Delta Y(\rho_j + \partial\rho_j, \varphi_j + \partial\varphi_j) - \Delta Y(\rho_j, \varphi_j), \end{aligned} \tag{10}$$

which induce the errors of C_{nm}, S_{nm} estimation because of distortion of the **X** elements.

To provide investigation of the influence of eye micromotions on the accuracy of estimation of indicated parameters, we have chosen a method of computer modeling random angular position of the eye's visual axis relative to the measurement beam (fig. 4). The method is based on using artificial distortion of the ρ_j, φ_j coordinates of measurement points by adding random numbers $\Delta\rho_j$,$\Delta\varphi_j$. These numbers are obtained by using either a random number generator for a specified errors distribution law or an array of numbers, obtained from actual eye micromotion measurements and stored in computer memory. An important advantage of such a numerical method is the possibility of getting the results for different micromotions including natural micromotion realizations.

The method include the following sequence of steps:

1. The $W(\rho_j, \varphi_j)$ function is specified a priori. The left part of equation (2) and the **A** matrix are formed.
2. The ΔX_j, ΔY_j values are computed at distorted pupil coordinates $\rho_j + \Delta\rho_j, \varphi_j + \Delta\varphi_j$.
3. The $\partial\Delta X$ and $\partial\Delta Y$ components, caused by eye rotation, are added to aberration values, indicated in step #2. The vector **X** is formed.
4. The **C** vector is calculated by formula (3).
5. Steps #2...#5 are repeated many times.

3. RESULTS AND DISCUSSION

Numerical investigation of the influence of eye micromotions on the accuracy of C_{nm}, S_{nm} estimation was carried out under such conditions:

1. The investigated eye is assumed to be standard with the back focal length 22.89 mm, pupil diameter 6 mm, having the following values of longitudinal aberration (in diopters):
    - Solely ametropia as aberration of defocusing: +3 and +6;
    - Solely primary astigmatism: +3 and +6;
    - Combination of ametropia and astigmatism: (+3, +3), (+3, +6), (+6, +3), and (+6, +6);

- Solely spherical aberration of the 3rd order at the pupil edge: +3 and +6;
- Solely spherical aberration of the 5th order at the pupil edge: +3 and +6.

2. The standard deviation of angular tilt of the visual axis (under normal error distribution) was equal to 2, 6, 10, 14, and 18 angular minutes.
3. The total number q of measurement points at the pupil plane was set to 33, 56, 66, and 110.
4. The number q of measurement points must be sufficient to exclude the influence of the $^{2q}\!/_{t}$ factor on the accuracy of Zernike coefficients C_{nm}, S_{nm} estimation.
5. The $\Delta\rho_j$, $\Delta\varphi_j$ values are assumed to have no mutual correlation for different measurement points.

The results are illustrated in fig. 5, where the standard deviations of Zernike coefficients C_{nm}, S_{nm} are indicated. These standard deviations are computed over 100 realizations of the **C** vector.

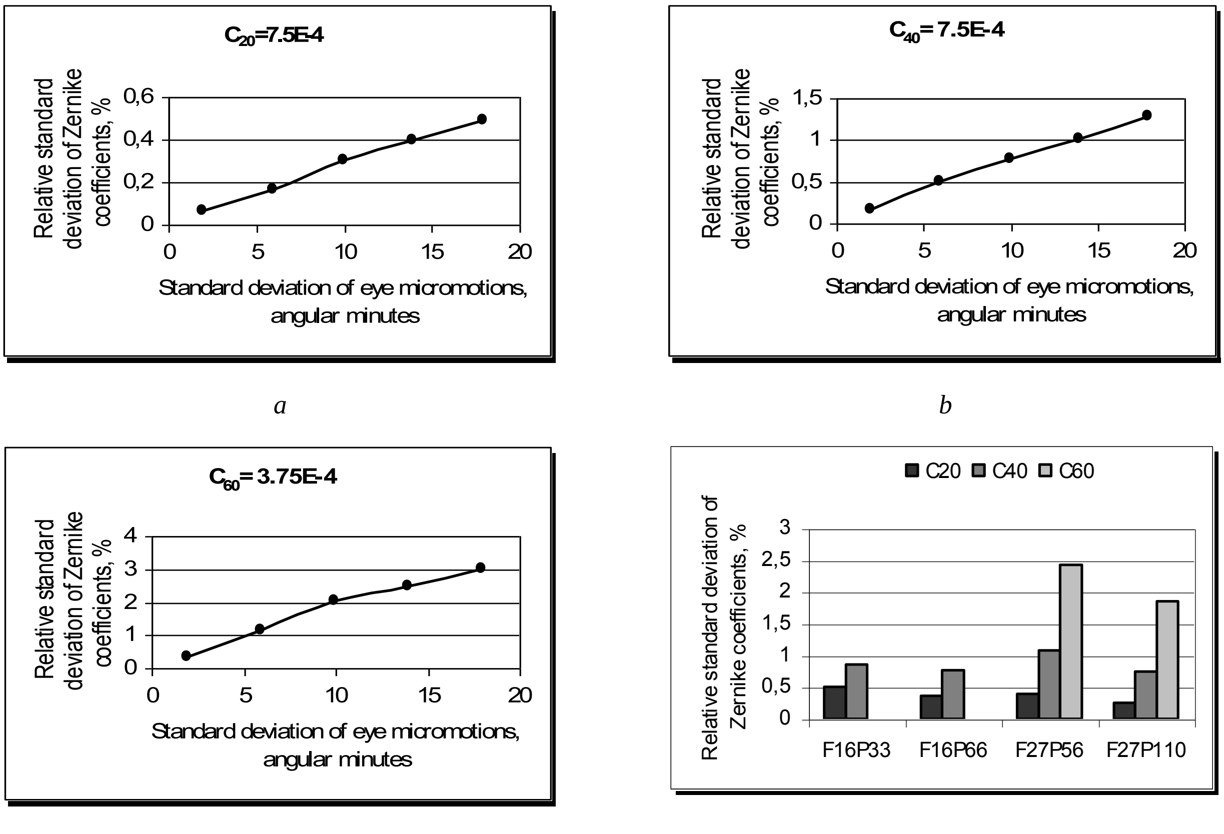

Fig. 5 *a*, *b*, *c* – plots of relative standard deviation of Zernike coefficients C_{nm} versus standard deviation of angular eye micromotions (obtained under 27 Zernike polynomials and 110 measurement points); *d* – diagram of relative standard deviation of the same coefficients versus the number of Zernike polynomials and the number of measurement points (F16P33 means 16 reconstruction polynomials and 33 measurement points)

As it is may be seen from fig. 5, the standard deviation $\sigma_{C,S}$ of Zernike coefficients estimation is directly proportional to σ_ψ in range $\sigma_\psi = 0'…20'$. For this reason, numerical results of $\sigma_{C,S}$ are indicated in table 1 only for $\sigma_\psi = 10'$.

The diagram *d* in fig. 5 shows that $\sigma_{C,S}$ depends on the number *q* of pupil measurement points and it decreases when *q* increases.

Table1 Results of computer modeling, for $\sigma_\psi = 10'$

Coef.	Given Zernike coefficients, normalized to 3-mm pupil radius, $\times 10^{-3}$											
C20	0.75	1.5	0	0	0.75	0.75	1.5	1.5	0	0	0	0
C22	0	0	0.75	1.5	0.75	1.5	0.75	1.5	0	0	0	0
C40	0	0	0	0	0	0	0	0	0.75	1.5	0	0
C60	0	0	0	0	0	0	0	0	0	0	0.375	0.75
Coef.	Standard deviation of estimated Zernike coefficients, normalized to 3-mm pupil radius, $\times 10^{-6}$											
C20	1.88	3.51	1.10	2.19	2.58	3.26	3.72	5.16	9.73	19.5	6.97	13.9
C22	3.78	7.05	1.98	3.95	3.96	5.17	7.23	7.91	8.82	17.6	8.65	17.3
C31	3.13	5.18	1.57	3.13	3.39	3.60	5.83	6.78	12.4	24.7	9.93	19.9
C33	3.12	7.17	1.59	3.18	3.76	3.52	7.17	7.52	8.23	16.5	7.61	15.2
C40	1.50	3.39	0.83	1.66	1.82	2.27	3.42	3.65	5.61	11.2	5.32	10.6
C42	2.24	3.22	1.18	2.35	2.58	3.41	3.57	5.16	10.3	20.6	8.58	17.2
C44	2.56	5.19	1.28	2.57	3.02	3.85	5.48	6.04	8.52	17.0	8.59	17.2
C51	1.30	2.22	0.69	1.37	1.55	2.12	2.61	3.10	7.73	15.5	14.3	28.6
C53	1.05	2.51	0.51	1.02	1.40	1.68	2.61	2.79	9.89	19.8	21.0	42.0
C55	1.74	3.05	0.78	1.57	1.73	2.21	3.17	3.46	13.1	26.2	26.5	53.1
C60	0.72	1.73	0.37	0.75	0.90	1.07	1.73	1.80	4.98	9.97	6.93	13.9
C62	1.02	1.89	0.56	1.12	1.19	1.60	2.07	2.37	6.73	13.5	10.7	21.4
C64	1.11	2.19	0.48	0.96	1.28	1.59	2.33	2.55	9.39	18.8	19.5	38.9
C66	1.38	2.43	0.73	1.47	1.53	2.06	2.66	3.05	9.84	19.7	20.4	40.9
S11	3.92	10.9	2.70	5.40	3.28	2.57	8.57	6.57	10.8	21.5	8.38	16.8
S22	3.52	7.35	1.67	3.34	3.00	3.91	7.50	8.00	8.68	17.4	8.06	16.1
S31	2.53	5.91	1.40	2.80	2.55	2.87	5.34	5.10	12.7	25.5	9.54	19.1
S33	3.09	5.64	1.30	2.59	3.06	3.75	5.74	6.12	8.60	17.2	8.08	16.2
S42	2.26	3.41	1.08	2.15	2.46	3.09	3.55	3.92	10.4	20.9	8.83	17.7
S44	3.03	5.50	1.24	2.47	3.06	3.76	5.68	6.11	8.20	16.4	7.46	14.9
S51	1.06	2.52	0.66	1.33	1.13	1.36	2.30	2.25	8.17	16.3	13.3	26.5
S53	1.01	2.60	0.52	1.03	1.43	1.71	2.68	2.86	11.7	23.4	24.7	49.3
S55	1.40	3.21	0.87	1.75	1.81	2.34	3.30	3.61	13.2	26.4	26.9	53.8
S62	1.03	2.18	0.49	0.99	1.24	1.55	2.29	2.49	8.28	16.6	12.7	25.4
S64	1.12	1.96	0.46	0.92	1.10	1.37	2.03	2.20	7.76	15.5	16.0	32.0
S66	1.25	2.39	0.63	1.27	1.39	1.80	2.52	2.79	10.1	20.2	21.1	42.2

Results, presented in table 1, indicate that:

1. If the eye has just one type of aberration, for example, solely ametropia (defocusing) or primary astigmatism, then eye micromotions results in appearing non-zero values C_{nm}, S_{nm} of all other types and orders of aberrations, absent in the eye. The contribution of each present aberration to $\sigma_{C,S}$ of absent aberration is independent and directly proportional to C_{nm}, S_{nm} of present aberrations.

2. The ratio between $\sigma_{C,S}$ of absent aberrations and C_{nm}, S_{nm} of present aberrations is unequal. For example, this ratio for C_{20} is two times greater than for C_{22}. This confirms the bigger influence of present ametropia on values $\sigma_{C,S}$ of absent aberrations in comparison with present astigmatism of the same value.

3. When $\sigma_\psi = 10'$, the $\sigma_{C,S}$ value of all aberrations, indicated in table 1, does not exceed 7.2% of the present aberration value (this cases are marked by a gray background).
4. When solely defocusing is present, the biggest increase of $\sigma_{C,S}$ of absent aberrations is observed for aberrations like primary astigmatism (the coefficients C_{22}, S_{22}) and primary coma (the coefficients C_{31}, S_{31}).
5. The biggest increase of $\sigma_{C,S}$ of absent aberrations is observed when spherical aberration of the 3rd and 5th orders is present (see four last columns in table 1). Obviously, it may be explained by the greater values of derivative of this aberration with respect to spatial coordinate ρ.

5. CONCLUSIONS

1. When ametropia is not compensated, then due to eye micromotions the retina image at the photodetector's light-sensitive surface moves over this surface. The range of such movements is proportional to the longitudinal aberration of the beam as well as to the eye rotation angle. The eye rotation leads to essential measurement errors of eye transversal aberrations. Elimination of these errors may be achieved by hardware compensation of ametropia.
2. Eye micromotions like jumps, which are not correlated for different measurement points and are of units or tens of angular minutes, may have an essential influence on Zernike coefficients estimation errors. Therefore, the ametropia and astigmatism parameters and the wave aberration function are affected too.
3. The greater are values of present aberrations, the greater is the effect of the factor, indicated in item #2. This concerns especially to those aberrations, which vary rapidly when changing pupil coordinates (field aberrations and high-order aberrations).
4. To diminish the influence of the factors, indicated in items #2 and #3, it is necessary to reduce the σ_ψ values. It can be achieved by shortening the time interval of transverse aberration measurement session.
5. The presented method has a universal character. It can be used when investigating the estimation errors of sight imperfection parameters at design of not only single-beam refractometers, but also other spatially resolved refractometers.

REFERENCES

1. J. Marsall, S. Trokel, S. Rothery, and R. R. Krueger, “Photoablative reprofiling of the cornea using an excimer laser: photorefractive keratectomy,” *Lasers Ophthalmol.*, **1**, pp. 21–48, 1986.
2. I. Pallikaris, M. Papatzanaki, E. Stathi, E. Frenschock, and A. Georgiadis “A corneal flap technique for laser in situ keratomileusus: human studies,” *Laser Surg. Med.*, **10**, pp. 463–468, 1990.
3. T. R. Littlefield, R. G. Koepnick, P. S. Binder, and H. S. Geggel, “New method for reshaping the cornea,” *J. Biomedical Optics*, **2**, pp. 106–114, 1997.
4. A. L. Yarbus, *The role of eye micromotions in a sight process*, Science, Moscow, 1965.
5. M. Born, E. Wolf, *Principles of optics*, Pergamon, New York, 1983.
6. V. V. Molebny, I. H. Chyzh, and V. M. Sokurenko, “Calculation of primary eye aberrations by Zernike polynomials,” *Scientific news of NTUU “KPI”*, No. 1. pp. 85–88, 2000.