Quantum Electronics **45** (10) 927–932 (2015)

PACS numbers: 42.66.Ct; 42.66.Lc; 79.20.Eb
DOI: 10.1070/QE2015v045n10ABEH015732

Effect of laser radiation wavelength and reepithelization process on optical quality of eye cornea after laser correction of vision

M.S. Kitai, A.V. Semchishen, V.A. Semchishen

Abstract. **The optical quality of the eye cornea surface after performing the laser vision correction essentially depends on the characteristic roughness scale (CRS) of the ablated surface, which is mainly determined by the absorption coefficient of the cornea at the laser wavelength. Thus, in the case of using an excimer ArF laser (λ = 193 nm) the absorption coefficient is equal to 39000 cm^{-1}, the darkening by the dissociation products takes place, and the depth of the roughness relief can be as large as 0.23 μm. Under irradiation with the Er:YAG laser (λ = 2940 nm) the clearing is observed due to the rupture of hydrogen bonds in water, and the relief depth exceeds 1 μm. It is shown that the process of reepithelization that occurs after performing the laser vision correction leads to the improvement of the optical quality of the cornea surface.**

Keywords: *eye cornea, medium darkening due to photochemical reaction, laser ablation, characteristic roughness size, effect of UV radiation, effect of IR radiation.*

1. Introduction

The laser refractive surgery is intended to remove the abnormalities of vision refraction of a patient and to achieve the level of improvement, exceeding that provided by glasses or contact lens corrections. The correction of refraction abnormalities consists in the formation of the front surface of the eye cornea by means of laser ablation. The clinical observations after performing the laser ablation of the eye cornea, as well as the data of experimental studies, show that the contrast sensitivity of vision, particularly under the twilight and night conditions, appears to be lower than it was before the operation in a patient using glasses. We suppose that one of the possible reasons of the contrast sensitivity loss can be the forward light scattering at the rough interface between two media with different refractive indices. In Ref. [1] it was shown that in order to save the contrast visual acuity the roughness depth in the ablation zone should be significantly less than 3–4 μm.

According to the theoretical models describing the laser ablation of biomaterials, the biological macromolecules dissociate, producing fragments of micro- and nanometre size. This process can be induced by photochemical reactions, mechanical stresses and the increased temperature that arise in the material under the action of laser pulses. The ablation process may be characterised by the achievable spatial resolution d_p, i.e., the size of the minimal biotissue fragment that can be removed in the process of laser ablation using the extreme delta-shaped profile of the exciting laser radiation. In the present paper it is shown that d_p has the similar order of magnitude in the longitudinal and transverse directions. Therefore, the value of the effective absorption coefficient that determines the size d_p in these directions appears to be important. In fact, d_p is the characteristic scale of the roughness that arises at the surface of the eye cornea after laser ablation. In the present paper we analyse the dependence of d_p on the laser radiation wavelength.

During the cornea healing after the performed laser correction the regeneration of the epithelium removed in the course of operation occurs. In this process the surface of newly formed epithelial layer is partially smoothed (this process is theoretically considered in Ref. [2]) and the optical quality of the interface between the epithelium and the lacrimal film is improved. We use a simplified one-dimensional mathematical model of reepithelization (based on the theory, described in Ref. [2]) of the post-operation rough surface of the eye cornea stroma, assuming that the local epithelium thickness under the equilibrium conditions depends on the slope of the surface, on which the epithelium grows. The one-dimensional approximation allowed us to solve the differential equation describing the reepithelization process by quadratures.

2. Absorption spectrum of eye cornea

The eye cornea consists of collagen (type II) and water in the proportion 22:78 [3].

It is known that a sharp increase in absorption of UV radiation by water begins from the vacuum ultraviolet wavelengths (λ < 170–180 nm) (see, e.g., [4]). This range of wavelengths is not used in biomedical applications. In the region of wavelengths $\lambda \geqslant$ 190–200 nm of the UV range, absorption is mainly caused by the organic component of the cornea. At these wavelengths, absorption by water is inessential and can be neglected. The initial absorption coefficient of the cornea weakly changes within the range of wavelengths 190–210 nm and decreases by nearly an order of magnitude at λ > 240 nm. Therefore, among UV lasers the ones practically used for the vision correction are the excimer ArF laser (λ = 193 nm) and the 5th harmonic of the Nd:YAG laser (λ = 213 nm) [5].

In the near IR range, absorption of light by the eye cornea is determined by the absorption spectrum of water with the maximum at 3 μm [6], which is related to the valence vibrations of the O–H groups. Figure 1 presents the absorption spectrum of the eye cornea in the UV and IR regions of the spectrum.

M.S. Kitai, A.V. Semchishen, V.A. Semchishen Institute on Laser and Information Technologies, Russian Academy of Sciences, ul. Pionerskaya 2, 142190 Moscow, Troitsk, Russia; e-mail: mkitai@mail.ru

Received 15 December 2014; revision received 13 May 2015
Kvantovaya Elektronika **45** (10) 927–932 (2015)
Translated by V.L. Derbov

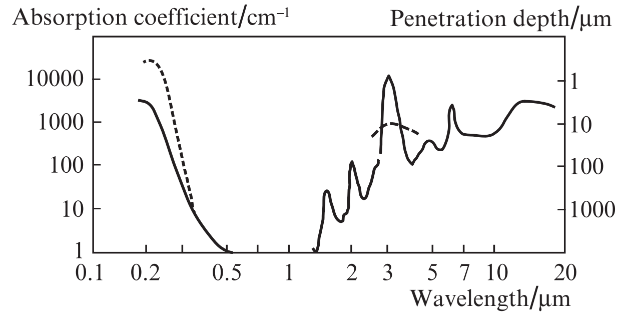

Figure 1. Absorption spectrum and depth of radiation penetration into the eye cornea. The dashed curves show the spectra of cornea effective absorption for ablation with an excimer ArF laser (λ = 0.193 μm, darkening) and an Er:YAG laser (λ = 2.94 μm, clearing).

According to review [7], the dependence of the ablation depth on the energy density of the laser pulse F can be approximated as

$$d(F) \approx \frac{1}{\alpha_{\text{eff}}} \ln\left(\frac{F}{F_{th}}\right), \quad (1)$$

where α_{eff} is the effective absorption coefficient, allowing for the darkening/clearing in the process of material ablation; F_{th} is the threshold energy density of the laser pulse. The ablation is a threshold process and is possible only if $F > F_{\text{th}}$.

3. Change in eye cornea absorption under the effect of the excimer ArF laser

The cornea absorption coefficient α at λ = 193 nm measured in Ref. [4] at pre-ablation energy densities was equal to $\sim 2.7\times10^3$ cm^{-1}.

The fundamentals of the theory, describing the process of the laser ablation of eye cornea under the impact of a single pulse of the ArF laser radiation, are presented in Ref. [8]. The basic stroma component is collagen, the typical molecule of which consists of 1155 repeating groups of three amino acids (glycine, proline, hydroxyproline) (see Fig. 2 in Ref. [9]).

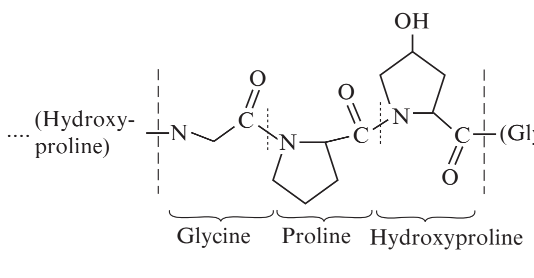

Figure 2. Chemical formula including three amino acids that compose the most of the collagen macromolecule. Peptide groups absorb radiation at a wavelength of 193 nm.

Under the impact of UV photons the photochemical rupture of macromolecules occurs in the eye cornea collagen, which leads to the efficient increase in the absorption coefficient of the biomaterial at the laser radiation wavelength [8–10]. It is known that in the UV range the absorption cross section for the products of the radiation photochemical action, formed at the place of macromolecule rupture, is noticeably larger than that of the initial macromolecule [11]. The effect of 'darkening' (an increase in absorption in the UV range, induced by the laser radiation) agrees with the experimental data on the dependence of the ablation depth on the pulse energy density. On average, for λ = 193 nm the effective absorption coefficient $\alpha_{\text{eff}} \approx 3.9\times10^4$ cm^{-1} [8, 12]. In Fig. 1 the dashed line shows the spectrum of α_{eff} in the region 0.2 μm. Thus, at this wavelength the ratio of the effective absorption coefficient to the initial cornea collagen absorption equals β = 15, which is confirmed by the data of Ref. [9].

4. Changes in the eye cornea absorption spectrum under the impact of Er:YAG laser radiation (λ = 2.94 μm)

It is known that for water the absolute absorption maximum in the IR range is near 3 μm. With the content of water in the eye cornea taken into account, the initial value of the eye cornea absorption coefficient α (at room temperature and in the absence of laser radiation) in this range lies in the interval $(2.2-4.4)\times10^3$ cm^{-1}. This value is close to the initial absorption coefficient of the cornea in the UV range, or can exceed it. This fact gives rise to the hope for the possibility to use near-IR lasers for the eye cornea ablation. Similar to the case of UV lasers, the absorption coefficient α depends on the laser pulse energy density. However, for the IR range α decreases with increasing energy density F, while for the UV range α increases. This is the principal difference between using the IR and UV radiation for cornea ablation. The cause of this unusual behaviour of the absorption coefficient is that the absorption at 2.94 μm is due not only to the valence vibration of the hydrogen atom, belonging to the intramolecular vibrations of the H_2O molecule. An essential contribution to the absorption at this wavelength comes from the hydrogen bonds. According to Ref. [13], in pure water nearly 88% of O–H bonds belong to the intermolecular hydrogen bonds. One should not mix the 'bound' water in biotissues understood as water ingressed in the solvate shell of the collagen (see, e.g., [13]) with the intermolecular hydrogen bonds in water. The latter are a more general notion than the 'bound' water in biotissues. According to [14], the universal dependence exists between the change of the hydrogen bond enthalpy (ΔH) and the intensity of the spectral line

$$-\Delta H \sim \Delta A^{1/2}, \quad (2)$$

where $\Delta A^{1/2} = A^{1/2} - A_0{}^{1/2}$ is the increment of the spectral line intensity of the hydrogen bond; A is the intensity of the line at the temperature T; and A_0 is the intensity of the initial line.

It is known that

$$\Delta H(T_0, T_1) = \int_{T_0}^{T_1} C_p dT, \quad (3)$$

where C_p is the heat capacity at constant pressure. Under the impact of an IR laser pulse the temperature T_1 increases; correspondingly, the variation of the enthalpy $\Delta H(T_0, T_1)$ grows and $A^{1/2}$ decreases. Note that even a relatively small rise in temperature T causes a significant decrease in hydrogen bond enthalpy due to its small energy. As far back as in 1991 this fact was experimentally demonstrated in pure water [15]. For λ = 2.94 μm and the absorbed energy density 9×10^3 J cm^{-3} the decrease in optical density under the impact of intense laser radiation approaches 88% of the initial value. In fact, under these conditions the intermolecular hydrogen bonds completely disappear. For biotissues the cause of the 'clearing' at λ =2.94 μm is the destruction of hydrogen bonds with a biotissue temperature growth and a sharp decrease in the vibration lifetime [15]. According to the data of Ref. [16], the measured depth of the eye cornea roughness after the ablation with the homogenised radiation of an Er : YAG laser having the energy density $\sim$1 J cm^{-2} could exceed 1 μm, which corresponds to the change in the effective absorption coefficient by nearly an order of magnitude, i.e., at this wavelength $\beta \approx 0.1$ (see also Fig. 1).

5. Thermomechanical processes in laser ablation of eye cornea

Under the impact of a short laser pulse the volume of the eye cornea material increases as a result of photochemical reactions (for UV radiation) and/or water boiling due to a temperature rise. The resulting mechanical stress has a complex nature. Near the surface irradiated by the laser (at the depth of radiation penetration), the elastic compression stress arises, which gives rise to an acoustic wave propagating into the sample. After the propagation of the acoustic wave, due to the mechanical boundary conditions at the free surface of the irradiated material, in the volume near this surface a quasistatic tensile stress is produced. This mechanical tensile stress (its tensor is denoted by σ_{ik}) is the main cause of the rupture of biological macromolecules, which leads to the laser ablation of the eye cornea. Naturally, the photochemical dissociation of collagen macromolecules under the action of UV radiation also provides a certain contribution to the rupture of macromolecules.

The kinetics of mechanical rupture of macromolecules for the one-dimensional case can be described within the frameworks of the model proposed in Ref. [17]. Then, the increment of the number of mechanically broken bonds N is described by the equation

$$\partial N / \partial t = GN\exp\,[-(U_0 - \gamma\sigma^*)/kT]\,\sinh\,[\gamma(\sigma - \sigma^*)/kT], \quad (4)$$

where G is the pre-exponential factor (the frequency of molecular vibration of the sections of collagen macromolecules), equal to $10^{12} - 10^{13}$ s^{-1}; N is the total concentration of the collagen macromolecule sections in the eye cornea; γ is the structure factor (for the eye cornea ablation γ is equal by the order of magnitude to the volume of nearly three elementary sections of the collagen macromolecule); σ^* is the mechanical tensile stress, at exceeding of which the mechanical destruction occurs; U_0 is the destruction energy of skeletal intermolecular bonds in collagen; and σ is the mechanical tensile stress acting along a definite axis.

Equation (4) describes the rupture of macromolecules because of one-dimensional tension of the sample. For an arbitrary transverse profile of laser radiation propagating along the z axis, the components of the mechanical stress tensor perpendicular to this axis arise. Correspondingly, alongside with σ_{zz}, the nondiagonal components σ_{xz} and σ_{yz} of the mechanical stress tensor appear [18]. Calculations show that the nondiagonal components of the mechanical stress are most essential near the geometric boundaries of the laser beam. According to the Hooke law, the contribution of the diagonal component of the mechanical stress tensor to the free energy of the deformation Φ is proportional to K, the bulk modulus, and the contribution of nondiagonal components is proportional to μ, the shear modulus [18]. By the order of magnitude, the coefficients K and μ are close to each other. The quantity $\gamma\sigma$ in the exponent in Eqn (4) is actually the free energy of the deformation Φ.

For any profile of the laser beam, the nondiagonal elements of the mechanical stress tensor arise near the beam boundaries (where the intensity is maximally changing). Consequently, near the boundary the roughness in the plane xy appears with the characteristic scale $1/\alpha_{\rm eff}$. Therefore, to obtain a high-quality ablated surface one should maximally increase the effective absorption coefficient. The greater the $\alpha_{\rm eff}$, the higher the instantaneous temperature, the greater the mechanical stress, the lower the laser energy density necessary for the ablation, and the smaller the characteristic roughness scale (CRS) of the biotissue inevitably arising in the process of laser ablation. The existing data allow the estimate of CRS in the case of cornea ablation with the radiation of the excimer ArF laser as $d \approx 1/\alpha_{\rm eff} \approx 0.23$ μm. Similarly one can estimate the CRS in the case of laser ablation of the cornea using the Er : YAG laser (2.94 μm); in this case, the CRS is $\sim$22 μm. The ratio of the CRS values for the lasers operating in the UV and IR range is 95. It is clear that because of the laser refractive operation the relief irregularities arise, with the characteristic depth satisfying the condition $h \gg \lambda$ (so-called strong perturbations and deep relief).

With the roughness taken into account, the spatial profile of the eye cornea surface after the laser correction of vision is a random function of two variables x and y (we neglect the eye cornea surface curvature, for which the mean value of radius is $\sim$8 mm). According to the simplest processing of the experimental results [19], when the ArF laser radiation is used for vision correction, the maximal absolute value of the ablation relief height is $h \approx 4.2$ μm (the appropriate root-mean-square deviation being $\pm$1.53 μm), and the root-mean-square roughness size in the xy plane equals $\langle l_0 \rangle \approx 280$ μm. The corresponding maximal slope is $\eta = h/\langle l_0 \rangle = 0.015 \ll 1$.

The roughness of the eye cornea surface affects the contrast sensitivity of vision, i.e., the quality of the image formed at the eye retina. The light passes the eye cornea, which is the major lens that forms the image at the retina. The contrast acuity of vision may be characterised by the ratio of intensities for the diffuse ($I_{\rm dif}$) and directional ($I_{\rm dir}$) components of the scattered light at the so-called visual axis (see [1] for details):

$$I_{\rm dif}/I_{\rm dir} \sim (h/\lambda)^2, \quad (5)$$

i.e., the smaller the h, the weaker the diffuse component as compared to the directional one, and the sharper the image. In Ref. [1] it is shown that the intensity $I_{\rm dif}$ comes up with the intensity $I_{\rm dir}$ that forms the image at the eye retina if the depth h exceeds 3–4 μm. Therefore, to save the contrast acuity of vision after the laser correction, the height of the roughness relief should be much less than 3–4 μm. From the physics of the cornea ablation process it follows that such smoothness of the ablated surface can be provided only by the radiation of the excimer ArF laser (λ = 193 nm). In Ref. [19] the algorithm of laser beam scanning determined the roughness scale.

6. Influence of the reepithelization process on optical quality of the cornea surface after laser vision correction

In the course of cornea healing after the photorefractive keratectomy (PRK), the removed epithelium undergoes regeneration. We assume that during the regeneration process the newly formed epithelium partially smoothes the ablated surface. As a result, the optical quality of the operated surface of the eye cornea is improved.

The outer layers of the cornea epithelium consist of 6–8 rows of regularly arranged cells, linked by intercellular bridges. Their thickness amounts to 10%–20% of the cornea thickness. The cells of the deepest row (the basal layer) have a clavate shape and contain large nuclei. These cells multiply and fill the epithelium defects. Above them a few rows of polygonal cells are located that become more and more flat towards the outer surface of the cornea. The cells of the outer layer are quite flat, and the epithelium surface is smooth. The epithelium has a high regeneration capability. It plays the protective role and regulates the water content in the cornea.

We suppose that the cause of smoothing is the dependence of the epithelium cells mobility on the slope of the surface, at which they are located. It is known that the stronger the interaction with the adjacent cells, the more suppressed the cells' capability of migration [20, 21], affecting the biochemistry of surface cells [22]. Therefore, the epithelium cells that find themselves in the relief wells interact stronger with their neighbours, restricting their migration ability. The cause is the stress in the epithelium cells, which arises due to the interaction of the given cell with the adjacent ones. As a result, as will be shown by the model calculation, the epithelium thickness in the wells in the equilibrium state appears to be greater than at the tops. In Ref. [2] it is shown that the layer of epithelial cells 'feels' the well (or any other 'perturbation' of the eye cornea) at the migration length, determined by the balance between the migration capability and the rate of epithelial desquamation (cell loss), $L \approx 500$ μm.

Thus, there are two types of epithelial cells: the low-mobility ones located at the stroma surface, their fraction being $g = m/n$ (where n is the total number of cells), and the rest ones, not contacting with stroma and having normal mobility, their fraction being $1 - m/n$. Due to the presence of first-type epithelial cells, the mobility of all cells, μ, appears to be lower than the nominal one by nearly $1 - g$ times.

By the order of magnitude the mobility of epithelial cells at the surface of the eye cornea is equal to the spatial derivative of their migration velocity $\langle \varphi_x \rangle \approx 2\times10^{-6}\,\text{cm s}^{-1}$. In a well of the initial relief, the mobility is lower because the epithelial cells adjoin the stroma. Assuming the mobility of the cells directly linked to the stroma to be zero, we can consider the mobility of cells in a well to be proportional to $D = 1 - m/n$. In this case, the mobility of epithelial cells is naturally considered as a function of the coordinates x, y.

Consider the stroma wells characterised by two basic parameters, the depth $2B$ and the width at half-depth level $2c$. Assume also that the epithelial cells, located in different wells, do not interact with each other. This is true if the distance separating wells is much greater than $2c$. The separations between the wells is also much greater than the migration length L.

Let us formalise the information about the squamous surface epithelial cells, coating the surface of the eye cornea (their thickness is of the order of 2–2.5 μm, and their diameter is of the order of 10 μm). Using the known formula for the volume of an ellipsoid having the semiaxes a, b, c: $V = (4\pi/3)abc$, we find that the volume of an epithelial cell is $V_C \approx 523\ \mu\text{m}^3$. Then the characteristic mean linear size of the cell is $l_C \approx (V_C)^{1/3} \approx 8.1$ μm, and the total area of the cell surface is $S_C \approx 660\ \mu\text{m}^2$. Under the normal growth conditions, the epithelial cells lie layer by layer. If the depth of the irregularities is of the order of 4 μm and their diameter is greater than 500 μm, then the epithelium cells completely fill the relief wells, since the thickness of the squamous epithelial cells is nearly by two times smaller than the characteristic well depth (4 μm), and the well width (~500 μm) is much greater than the diameter of the epithelial cells.

Without the loss of generality, we can consider a single well. To simplify the calculations let us consider a one-dimensional problem, when the surface shape depends on a single coordinate x. Let us describe the shape of the ablated surface by the function

$$Z(x) = -B - B(\pi x/c), \tag{6}$$

where x varies from $-c$ to c. Figure 3 presents the model profiles of the irregularities (wells), used in the calculations. Since the corneal curvature radius (typically 8 mm) is essentially greater than B and c, one can neglect the cornea curvature. The cornea surface is considered plane and the level $z = 0$ corresponds to the unperturbed surface, the well width being greater than the linear size of a cell. We assume that the epithelial cells migrate in the plane of the eye cornea.

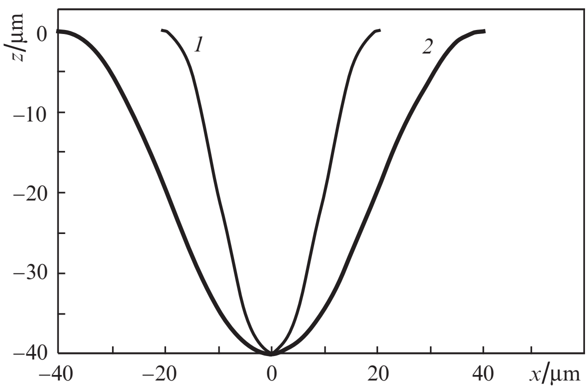

Figure 3. Model profiles of the wells of the cornea surface, used in the calculations at different values of the half-depth width c = (*1*) 10 and (*2*) 20 μm; B = 20 μm.

The equation describing the dynamics of the epithelial cells migration over the planar eye cornea has the form

$$\frac{\partial r}{\partial t} = a - br(x,t) + \operatorname{div}\{\varphi(x)\operatorname{grad}[r(x,t)]\}, \tag{7}$$

where $r(x, t)$ is the height (thickness) of the layer of epithelial cells (μm); a is the rate constant of the epithelial cell generation (μm s^{-1}); b is the time constant of the cell loss (desquamation) (s^{-1}); and the function $\varphi(x)$ (μm^2 s^{-1}) describes the dependence of the migration (diffusion) rate of the epithelial cells on the coordinate in the well. Generally, Eqn (7) corresponds to the model described in Ref. [2], but the simplifications made by us make the solution possible. The solution of the equation, analogous to Eqn (7), is described in monograph [23]. Formally, Eqn (7) is a Helmholtz equation with respect to the function $r(x, t)$.

In the steady state case considered below, the time derivative is zero. Correspondingly, Eqn (7) with respect to the unknown function $r(x) \equiv R$, independent of time, is transformed into the ordinary differential equation

$$a - bR + \frac{d}{dx}\left[\varphi(x)\frac{dR}{dx}\right] = 0. \qquad (8)$$

The constants a and b are assumed to be independent of x. At smooth regions of the x axis, having no roughness, the dimensionless spatial derivative $Y \equiv dR/dx \approx 0$ so that $R = a/b$. If the relief is described by Eqn (6), this boundary condition holds for $|x| \geqslant c$. Differentiating Eqn (8), we obtain

$$-bY + Y\frac{d\varphi}{dx} + \varphi\frac{dY}{dx} = 0. \qquad (9)$$

We can write the solution of Eqn (9) in the form

$$\frac{dY}{Y} = \frac{dx}{\varphi(x)}\left(b - \frac{d\varphi}{dx}\right). \qquad (10)$$

According to the available literature data, the thickness of the epithelial layer of smooth eye cornea amounts to 30–50 μm; the mobility of epithelial cells, moving over the stroma surface, can be evaluated by the characteristic mean value of the spatial derivative $<\dot{\varphi}_x> \approx 2\times10^{-6}$ cm s^{-1}. The irregularities (wells) at the eye cornea typically 'occlude' during the typical healing (reepithelization) time of $\tau \approx 3-5$ days [or $(2.5-4.3)\times10^5$ s]. The characteristic distance, at which the wells manifest themselves, is $\tau\,\langle\dot{\varphi}_x\rangle \approx 500$ μm, which agrees with the migration length L from [2]. We neglect the spatial features of the surface at the distances, exceeding 500 μm. These data allow the estimation of the constants $b \approx 3\times10^{-6}$ s^{-1} and $a \approx 1.5\times10^{-8}$ cm s^{-1}. Therefore, the mean epithelium thickness amounts to $a/b = 50$ μm.

The solution of Eqn (10) is a functional of $\varphi(x)$. The precise calculation of this function is a subject of a separate study. Our aim here is to show that the reduction of $\varphi(x)$ in the well leads to the smoothing of the surface due to the growth and migration of epithelial cells. Assume that the dependence $\varphi(x)$ is expressed as

$$\varphi(x) = \varphi_0\left[1 - \delta\cos\left(\frac{\pi x}{c}\right)\right]. \qquad (11)$$

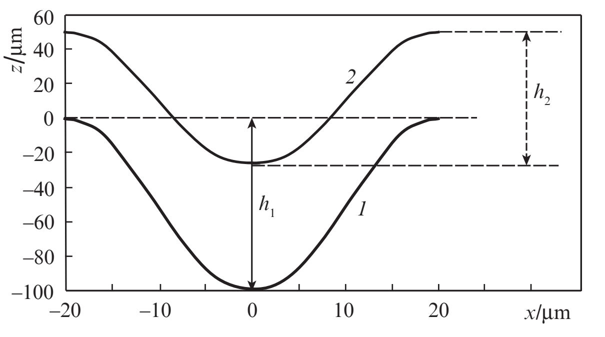

Figure 4. Shape of the cornea surface after (*1*) laser ablation at $B = 20$ μm, $c = 40$ μm and (*2*) the end of the reepithelization process.

Here, $\delta < 1$ is a dimensionless positive quantity, characterising the change in the mobility of epithelial cells. Figure 4 illustrates the cornea surface shape after laser ablation at $B = 20$ μm and $c = 40$ μm (*1*) and after the end of reepithelization process (*2*). After ablation, the depth of an individual well is equal to h_1, and after the occlusion of the well by the epithelial cells, its depth becomes equal to h_2. It is seen that $h_2 < h_1$, and for the given B and c the difference $h_2 - h_1$ amounts to 30 μm. The simple model consideration has shown that the process of post-operation healing (reepithelization) facilitates the improvement of the cornea surface optical quality after laser vision correction using the PRK method [24].

7. Conclusions

We have established that the quality of laser-ablated eye cornea surface depends on the effective (with the darkening/clearing taken into account) absorption coefficient at the radiation wavelength. The advantages of using an excimer ArF laser (λ = 193 nm) as compared to the radiation of an Er:YAG laser (λ = 2.94 μm) for the laser correction of vision are demonstrated. The mathematical model that allows the description of the post-operation cornea reepithelization process and its smoothing is developed. It is shown that the process of the cornea surface reepithelization improves the optical quality of the surface.

Acknowledgements. In conclusion, the authors thank A.P. Sviridov for stimulating discussions. The work was supported by the Russian Science Foundation (Grant No. 14-25-00055 in part of an ablation model) and the Russian Foundation for Basic Research (Grant No. 14.29.10211 in part of reepitalization model description).

References

1. Semchishen A.V., Seminogov V.N., Semchishen V.A. *Kvantovaya Elektron.*, **42** (4), 345 (2012) [*Quantum Electron.*, **42** (4), 345 (2012)].
2. Huang D., Tang M., Shekhar R. *Amer. J. Ophtalmolog.*, **135** (3), 267 (2003).
3. Trokel S.L., Srinivasan R., Brayen B. *Amer. J. Ophtalmolog.*, **96**, 710 (1983).
4. http://www.genon.ru/GetAnswer.aspx?qid=dc60f6a8-a53d-4091-9c52-8c18c959927b.
5. Pettit G.Y., Ediger M.N. *Appl. Opt.*, **35** (19), 3386 (1996).
6. Wieliczka D., Weng S., Querry M. *Appl. Opt.*, **28**, 1714 (1989).
7. Anisimov S.I., Luk'yanchuk B.S. *Usp. Fiz. Nauk*, **172** (3), 301 (2002) [*Phys. Usp.*, **45** (3), 293 (2002)].
8. Kitai M.S., Popkov V.L., Semchishen V.A., Kharizov A.A. *IEEE J. Quantum Electron.*, **27** (2), 302 (1991).
9. Fisher B.T. *Dissertation* (University of Florida, 2004).
10. Puliafito C.A., Steinert R.F., Deitch T.F., et al. *Ophtalmology*, **92**, 741 (1985).
11. Nikitin V.N., Perskii E.E., Utevskaya L.A. *Vozrast i evolutsionnaya biokhimiya kollagenovykh struktur* (Age and Evolutionary Biochemistry of Collagen Structures) (Kiev: Naukova Dumka, 1977).
12. http://medbiophysics.professorjournal.ru/c/document_library/get_file?p_l_id=471225&folderId=465647&name=DLFE-8930.pdf.
13. http://window.edu.ru/resource/960/29960/files/sgu014.pdf.
14. Iogansen A.V. *Spectrochimica Acta*, **55**, 1585 (1999).
15. Vodopyanov K.L. *J. Chem. Phys.*, **94** (8), 5389 (1991).
16. Mrochen M., Funk W., Semchishen V., Seiler T. *J. Refractive Surgery*, **16**, 51 (2000).
17. Regel' V.R., Slutsker A.I., Tomashevskii E.E. *Kineticheskaya priroda prochnosti tverdykh tel* (Kinetic Nature of Solid Strength) (Moscow: Nauka, 1984).

18. Landau L.D., Lifshits E.M. *Theory of Elasticity* (Oxford, New York: Pergamon Press, 1970; Moscow: Nauka, 1987).
19. Mrochen M., Schelling U., Wuelner C., Donitzky C. *J. Cataract Refaract Surg.*, **35**, 363 (2009).
20. Garrod D.R., Steinberg M.S. *Nature*, **244**, 568 (1973).
21. Vesely P., Weiss R.A. *Intern. J. Cancer*, **11**, 64 (1973).
22. Huttenlocker A., Lakonishok M., Kinder M., et al. *J. Cell Biol.*, **141**, 515 (1998).
23. Tikhonov A.N., Samarskii A.A. *Uravneniya matematicheskoi fiziki* (Equations of Mathematical Physics) (Moscow: Nauka, 1977).
24. Semchishen A.V., Semchishen V.A. *Ross. Oftalmol. Zh.*, **6** (1), 43 (2012) [*Russ. Ophthalmol. J.*, **6** (1), 43 (2012)].