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ABSORPTION SPECTRA AND LIGHT PENETRATION DEPTH OF NORMAL AND PATHOLOGICALLY ALTERED HUMAN SKIN

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A three-layered skin model (stratum corneum, epidermis, and dermis) and engineering formulas for radiative transfer theory are used to study absorption spectra and light penetration depths of normal and pathologically altered skin. The formulas include small-angle and asymptotic approximations and a layer-addition method. These characteristics are calculated for wavelengths used for low-intensity laser therapy. We examined several pathologies such as vitiligo, edema, erythematosus lupus, and subcutaneous wound, for which the bulk concentrations of melanin and blood vessels or tissue structure (for subcutaneous wound) change compared with normal skin. The penetration depth spectrum is very similar to the inverted blood absorption spectrum. In other words, the depth is minimal at blood absorption maxima. The calculated absorption spectra enable the power and irradiation wavelength providing the required light effect to be selected. Relationships between the penetration depth and the diffuse reflectance coefficient of skin (unambiguously expressed through the absorption coefficient) are analyzed at different wavelengths. This makes it possible to find relationships between the light fields inside and outside the tissue.

Key words: *biological tissue, skin, laser light, absorption spectrum, penetration depth, radiative transfer theory, melanin, blood*

Introduction. Low-intensity laser light (LILL) is used to produce therapeutic effects in an organism. The effect depends on the light wavelength, its penetration depth into the tissue, the power of the incident and absorbed light, the duration of the effect, the structure of the medium, etc. Herein features of light transfer in human skin are studied to evaluate several of these properties. We think that the optical properties and tissue structure do not change during irradiation because of its low intensity. Transfer in biological tissues has been studied in many countries and for a rather long time. Important results have been achieved on the construction of models of structural and optical properties of tissues [1–3]. Computer programs that rely on the broad capabilities of modern computer technology have been developed for calculating light distribution characteristics [4, 5]. However, the employed modeling methods are cumbersome, require large amounts of machine time, and frequently are not efficient and precise enough for practical application because of the huge number of parameters that in the overwhelming majority of instances are unknown and vary considerably. Therefore, they are usually included in calculation procedures by selecting all reasonable values for them. Engineering methods of light transfer theory that were developed in the Institute of Physics of the National Academy of Sciences of Belarus have great potential for research on the optics of tissues [6]. They give the final result in an analytical form that can rather simply and quickly reveal important and unimportant parameters of light distribution, thereby reduce their number, and estimate the a priori uncertainty of one critical parameter or another on the precision of the calculated light fields in the medium. The structural and optical properties of skin must be given in order to apply these analytical methods. Experimental determination of the penetration depth z_0 is fraught with serious complications related to placing a sensor under the skin and, therefore, its effect on it. The goal of the present work

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was to calculate the absorption coefficient A and z_0 for wavelengths used for LILL for normal skin and several of its pathologies and to evaluate the possibility of predicting z_0 from spectral values of the diffuse reflectance coefficient.

Skin Structure and Optical Properties of its Components. These properties have been rather completely studied for "average" or "nominally normal" skin [3, 7–9]. A layer model, the characteristics of which change along a single coordinate, the depth z, is usually used. Three basic layers are identified [8]: the upper, stratum corneum, of thickness d_1 = 0.01–0.02 mm; epidermis (d_2 = 0.04–0.15 mm), and dermis (1–4 mm). The dermis, in turn, consists of several characteristic sublayers [7, 9, 10] with different contents of blood vessels. It is understood that such a structure is to a certain extent arbitrary. There are no clear physical boundaries between the layers and no sharp differences in the optical properties. We propose that the only boundary is the interface of the stratum corneum with the external medium. The stratum corneum has a refractive index relative to air of about 1.5 [3, 7, 9]. Based on this value and considering the skin roughness parameters [9], the Fresnel coefficients for its surface reflectance are calculated below. The refractive index changes smoothly between skin layers so that Fresnel reflectance does not occur. According to the model [2, 11], we think that the epidermis consists of tissue-bases that absorb and scatter light and an absorbing component, melanin (the tissue-basis is frequently called "bloodless tissue"). The dermis consists of the same tissue-basis in which, we think, blood vessels, capillaries, are evenly and randomly distributed. This layer is infinitely thick with respect to the optics, i.e., the structure of biological tissue under the dermis has no effect on the light fields in skin.

Optical properties that determine light transfer effects in the medium include $\beta = \sqrt{3k\varepsilon}$, the depth attenuation index; k and $\varepsilon = \varepsilon'(1 - g)$, the absorption index and effective attenuation index, respectively; ε', the attenuation index; and g, the average cosine of the scattering indicatrix. For the stratum corneum, epidermis, and dermis, they have the indices 1, 2, and 3, respectively:

$$k_1 = k_t, \quad k_2 = f_m k_m + (1 - f_m)\, k_t, \quad k_3 = C_V \alpha k_b + (1 - C_V)\, k_t, \tag{1}$$

$$\varepsilon_1 = \varepsilon_t, \quad \varepsilon_2 = (1 - f_m)\, \varepsilon_t, \quad \varepsilon_3 = C_V \varepsilon_b + (1 - C_V)\, \varepsilon_t. \tag{2}$$

Here ε_t and ε_b are the effective attenuation indices of tissue-basis and blood; k_t, k_m, and k_b, absorption indices of tissue-basis, melanin, and blood; f_m and C_V, bulk concentration of melanin in epidermis and capillaries in dermis. The blood absorption index is

$$k_b = Hf\,[Sk_{HbO_2} + (1 - S)\, k_{Hb}], \tag{3}$$

where H is hematocrit (volume fraction of erythrocytes in blood); f, volume fraction of hemoglobin in erythrocytes; S, degree of blood oxygenation; and k_{HbO_2} and k_{Hb}, absorption indices of oxyhemoglobin (HbO_2) and deoxyhemoglobin (Hb). We set H = 0.4 and f = 0.25 [9]. Coefficient α in Eq. (1) takes into account localized absorption of light by vessels. Let us find its meaning. The concentration of capillaries that absorb light in biological tissue is usually small. This suggests that a "sieve" (or "hole") effect occurs in the medium for such nonuniformity [12], where much light passes through the less absorbing portions and the total penetration is greater than expected considering the even absorptive capability of a unit volume. This effect has been studied in detail [13]. It was shown that

$$\alpha = 2\sqrt{3}\,\frac{1 - \exp\left[-\pi k_b D\,(1 - 0.043 k_b D)/(2\sqrt{3})\right]}{\pi k_b D}, \tag{4}$$

where D is the average diameter of capillaries. Equation (4) shows that for $k_b D < 1$, $\alpha \to 1$.

Figure 1 shows the skin optical characteristics k_{HbO_2}, k_{Hb}, k_m, k_t, ε_b, and ε_t as functions of wavelength. They were obtained by critical analysis, selection, and averaging of previous results [2, 3, 8, 9, 11, 14–16]. Table 1 lists typical values of C_V and f_m for normal and pathologically altered skin. Vitiligo, erythematosus lupus, edema, and subcutaneous wound were chosen as several types of pathologies. These data are used below in the calculations.

Method for Calculating Skin Diffuse Reflectance Coefficient. As noted, the light absorption coefficient of skin (A) is of practical importance for phototherapy. By definition A is the fraction of absorbed energy relative to the

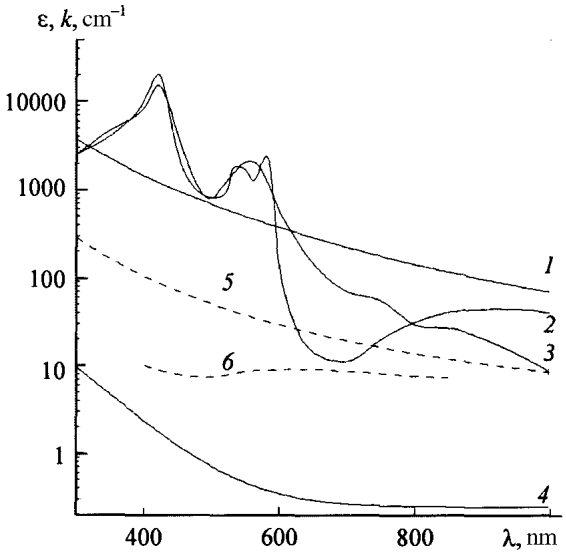

Fig. 1 Spectral dependences of absorption index (solid lines) and effective attenuation index (dashed) for skin components: melanin (1), oxyhemoglobin (2), deoxyhemoglobin (3), skin bases (4, 5), and blood (6).

TABLE 1 Structural and Biophysical Parameters of Normal and Pathologically Altered Skin

Pathology	Epidermis	Melanin concentration, f_m, %	Capillary concentration, C_V, %
Normal skin	Present	5–15	1–3
Vitiligo	Present	0–2	1–3
Erythematosus lupus	Present	5–15	5–15
Edema	Present	5–15	0.2–0.6
Subcutaneous wound	Absent, instead of it, blood layer 100 μm thick	–	1–3

incident energy, which is equal to $1 - R$, where R is the diffuse reflectance coefficient of the light. Therefore, we will examine the method for calculating R.

Let the skin be illuminated with a directed light beam. The skin surface reflects a certain fraction of the incident flux with a reflectance coefficient r. Light passing through the skin surface falls on its three successive layers as a directed beam as a result of the small optical thicknesses of the stratum corneum and epidermis and their scattering indicatrix, which is highly elongated "forward" [8]. The fluxes are multiply reflected between layers and practically diffuse. The ratio of the total reflected flux to the incident one is the diffuse reflectance coefficient (R) of skin. This scheme assumes that the R values depend on the transmission coefficients and reflection of the directed and diffuse light by stratum corneum and epidermis and on the reflectance coefficient of directed and diffuse light by the dermis. Therefore, the formulas for the transmission and reflectance coefficients of directed and diffuse light by the corresponding layers are given below. Values that refer to diffuse light are marked with an asterisk (*). As mentioned, stratum corneum and epidermis have rather thin optical thicknesses. Therefore, a small-angle approximation [6] of transfer theory is used to describe the distribution in them of the directed light; an asymptotic approximation, the diffuse light [6, 17]. The latter is precise if the angular structure of the scattered light brightness coincides with its angular distribution beneath the surface [6] regardless of the optical thickness of the medium. We think that such coincidence is approximated in biological tissue. On the other hand, dermis is an optically thick layer for which the asymptotic approximation works well [6, 17].

The coefficients of light transmission and reflectance (T and R with the appropriate indices) by stratum corneum and epidermis illuminated along the normal have the form

$$T_{1,2}(d_{1,2}) = \exp[-\tau_{1,2}(1-\Lambda_{1,2}F_{1,2})], \quad R_{1,2}(d_{1,2}) = \Lambda_{1,2}\frac{1-F_{1,2}}{1-F_{1,2}\Lambda_{1,2}}\int_0^1\left\{1-\exp\left[-\tau_{1,2}(1-\Lambda_{1,2}F_{1,2})\frac{1+\mu}{\mu}\right]\right\}d\mu, \quad (5)$$

where τ is the optical thickness ($\tau_{1,2} = \varepsilon'_{1,2}d_{1,2}$, $d_{1,2}$ is the geometric thickness of stratum corneum or epidermis), Λ is the probability of photon survival, $\Lambda = (\varepsilon' - k)/\varepsilon'$; F is the fraction of light scattered "forward" (at scattering angles 0–90°) relative to the totally scattered flux, $F = 1 - (1 - g)/3$ [6]; and μ is the cosine of the scattering angle. For diffuse illumination of stratum corneum and epidermis

$$T^*_{1,2}(d_{1,2}) = \frac{\sinh[4\sqrt{k_{1,2}/(3\varepsilon_{1,2})}]}{\sinh[d_{1,2}\beta_{1,2} + 4\sqrt{k_{1,2}/(3\varepsilon_{1,2})}]}, \quad R^*_{1,2}(d_{1,2}) = \frac{\sinh(d_{1,2}\beta_{1,2})}{\sinh[d_{1,2}\beta_{1,2} + 4\sqrt{k_{1,2}/(3\varepsilon_{1,2})}]}. \quad (6)$$

For directed illumination of dermis along the normal

$$R(\infty) = \exp[-(36/7)\sqrt{k/(3\varepsilon)}]. \quad (7)$$

For diffuse illumination of dermis

$$R^*_3(\infty) = \exp[-4\sqrt{k_3/(3\varepsilon_3)}], \quad (8)$$

which is analogous to Eq. (6) as $d_{1,2} \to \infty$.

Knowing $T_{1,2}$ and $R_{1\text{-}3}$ of each skin layer, its diffuse reflectance coefficient R can be found. Light fluxes that are multiply reflected by the layers form an infinitely decreasing geometric series. It is easy to find its sum and obtain the desired reflectance coefficient:

$$R = r + \frac{(1-r)(1-r^*)R_{123}(d_1, d_2, \infty)}{1 - r^* R^*_{123}(d_1, d_2, \infty)}, \quad (9)$$

where $r = 0.04$; $r^* = 0.2$ is the reflectance coefficient of the skin surface illuminated by a diffuse flux from within; R_{123} and R^*_{123} are reflectance coefficients of the layered structure from the three layers [stratum corneum (1), epidermis (2), and dermis (3)] without an external boundary for directed and diffuse illumination, respectively. The arguments of the reflectance coefficients in Eq. (9) (and transmission coefficients below) indicate the layer thicknesses. The coefficients R_{123} and R^*_{123} can be found analogously to Eq. (9) by summing the corresponding geometric series:

$$R_{123}(d_1, d_2, \infty) = R_1(d_1) + \frac{T_1(d_1)\,T^*_1(d_1)\,R_{23}(d_2, \infty)}{1 - R^*_1(d_1)\,R^*_{23}(d_2, \infty)}, \quad (10)$$

$$R^*_{123}(d_1, d_2, \infty) = R^*_1(d_1) + \frac{[T^*_1(d_1)]^2\,R^*_{23}(d_2, \infty)}{1 - R^*_1(d_1)\,R^*_{23}(d_2, \infty)},$$

where R_{23} and R^*_{23} are determined, like in Eqs. (9) and (10), by sums of geometric series:

$$R_{23}(d_2, \infty) = R_2(d_2) + \frac{T_2(d_2)\,T^*_2(d_2)\,R_3(\infty)}{1 - R^*_2(d_2)\,R^*_3(\infty)}, \quad (11)$$

$$R_{23}^{*}(d_2, \infty) = R_2^{*}(d_2) + \frac{[T_2^{*}(d_2)]^2 R_3^{*}(\infty)}{1 - R_2^{*}(d_2) R_3^{*}(\infty)},$$

The transmission and reflectance coefficients in Eqs. (9)–(11) are given by Eqs. (5)–(8). It should be noted that Eqs. (9)–(11) are similar to each other taking into account that by definition $(1 - r)$ and $(1 - r^{*})$ are transmission coefficients of the skin surface for illumination directed from outside and diffused within, respectively.

Method for Calculating Illumination within Skin. Let us write the expression for the distribution of the light field within tissue. Let the incident light beam create illumination E_0 on the skin surface. The total (from two sides of the area) illumination as a function of depth $E(z)$ has the form:

$$E(z) = \begin{cases} E_{1\downarrow} + E_{1\uparrow} & \text{at} \quad 0 \le z \le d_1, \\ E_{2\downarrow} + E_{2\uparrow} & \text{at} \quad d_1 \le z \le d_1 + d_2, \\ E_{3\downarrow} + E_{3\uparrow} & \text{at} \quad z \ge d_1 + d_2, \end{cases} \tag{12}$$

where arrows next to indices indicate illumination from the side of the laser beam and from the opposite side. In stratum corneum

$$\frac{E_{1\downarrow}}{E_0 (1 - r)} = T_1(z) \left[1 + \frac{R_{123}(d_1 - z, d_2, \infty) R_1'^{*}(z)}{1 - R_{123}^{*}(d_1 - z, d_{2,} \infty) R_1'^{*}(z)} \right], \tag{13}$$

$$\frac{E_{1\uparrow}}{E_0 (1 - r)} = \frac{T_1(z) R_{123}(d_1 - z, d_2, \infty)}{1 - R_{123}^{*}(d_1 - z, d_2, \infty) R_1'^{*}(z)}, \tag{14}$$

where $R_1'^{*}(z)$ is the reflectance coefficient of part of the stratum corneum of thickness z from the substrate (skin surface) for diffuse illumination from within the medium:

$$R_1'^{*}(z) = R_1^{*}(z) + \frac{r^{*} [T_1^{*}(z)]^2}{1 - r^{*} R_1^{*}(z)}. \tag{15}$$

In epidermis

$$\frac{E_{2\downarrow}}{E_0 (1 - r)} = T_1(d_1) T_2(z - d_1) \left[1 + \frac{R_{23}(d_1 + d_2 - z, \infty) R_{21}'^{*}(z - d_1)}{1 - R_{23}^{*}(d_1 + d_2 - z, \infty) R_{21}'^{*}(z - d_1)} \right], \tag{16}$$

$$\frac{E_{1\uparrow}}{E_0 (1 - r)} = \frac{T_1(d_1) T_2(z - d_1) R_{23}(d_1 + d_2 - z, \infty)}{1 - R_{23}^{*}(d_1 + d_2 - z, \infty) R_{21}'^{*}(z - d_1)}, \tag{17}$$

where $R_{12}'^{*}(z - d_1)$ is the reflectance coefficient of the layered structure consisting of part of the epidermis of thickness $(z - d_1)$, the whole stratum corneum, and the skin surface for diffuse illumination from within the medium:

$$R_{21}'^{*}(z) = R_2^{*}(z) + \frac{R_1'^{*}(d_1) [T_2^{*}(z)]^2}{1 - R_1'^{*}(d_1) R_2^{*}(z)}. \tag{18}$$

In dermis

$$\frac{E_{3\downarrow}}{E_0\,(1-r)} = T_1\,(d_1)\,T_2\,(d_2)\,T_3\,(z-d_1-d_2)\left[1+\frac{R_3 R'^{*}_{321}\,(z-d_1-d_2)}{1-R^{*}_3 R'^{*}_{321}\,(z-d_1)}\right], \tag{19}$$

$$\frac{E_{3\uparrow}}{E_0\,(1-r)} = \frac{T_1\,(d_1)\,T_2\,(d_2)\,T_3\,(z-d_1-d_2)\,R_3}{1-R^{*}_3 R'^{*}_{321}\,(z-d_1-d_2)}, \tag{20}$$

where $R'^{*}_{321}(z - d_1 - d_2)$ is the reflectance coefficient of the layered structure consisting of part of the dermis of thickness $(z - d_1 - d_2)$, the complete stratum corneum and epidermis and also the skin surface:

$$R'^{*}_{321}\,(z) = R^{*}_3\,(z) + \frac{R'^{*}_{21}\,(d_2)\,[T^{*}_3\,(z)]^2}{1-R'^{*}_{21}\,(d_2)\,R^{*}_3\,(z)}. \tag{21}$$

Transmission coefficients $T^{*}_3(z - d_1 - d_2)$ in Eqs. (19) and (20) and reflectance $R^{*}_3(z)$ in Eq. (21) are calculated using Eq. (6) by substituting for d_1 or d_2 of this argument and the optical properties of dermis:

$$T_3\,(z) = \frac{\sinh\left[(36/7)\sqrt{k_{3,2}/(3\varepsilon_{3,2})}\right]}{\sinh\left[z\beta_{3,2} + 4\sqrt{k_{3,2}/(3\varepsilon_{3,2})}\right]}. \tag{22}$$

Absorbed Light Dose by Normal and Pathologically Altered Skin. Laser therapy should take into account that not all light impinging on biological tissue is absorbed by it. Part of it is reflected. Only the absorbed light has a therapeutic effect. It is important in principle to maintain an accurate dosing. The energy absorbed by biological tissues for laser therapy is

$$F_{\mathrm{a}} = N_{\mathrm{a}}\,st = N_{\mathrm{i}}\,st\,(1-R), \tag{23}$$

where N_{a} and N_{i} are the power density of the incident and absorbed light, s is the area of the laser beam, and t is the illumination time. The products $N_{\mathrm{a}}st$ and $N_{\mathrm{i}}st$ are the absorbed and incident doses, respectively.

Table 2 gives data for the absorbed dose as a function of wavelength and degree of oxygenation for incident energy 1 J. Average values of f_{m} and C_V from Table 1 for normal and pathologically altered skin are used for calculations in Eq. (23). The epidermis thickness d_2 is set at 60 μm. Estimating F_{a} at various N_{i} is trivial because the absorbed energy is proportional to the incident energy.

Absorption coefficient A for all biological tissue layers was discussed above. Let us note that values of A for local sections of the medium are interesting for several practically important problems in biomedical optics, for example, light therapy using a sensitizer. Equations (12)–(22) can be used to calculate these values taking into account actual structural and optical properties of the analyzed section. However, this issue is beyond the scope of the present work.

Penetration Depth of Light into Tissue. The penetration depth of light into tissue z_0 is an important practical property for phototherapy and tomography. It determines the light dose distribution within the medium, the temperature regime, and many other properties that are significant in clinical practice. The quantity z_0 represents the depth to which the light density decreases by 10 times relative to the incident density, i.e., satisfies the equation

$$E\,(z_0) = 0.1E_0. \tag{24}$$

Figure 2 shows ranges over which the light penetration depth varies in normal and pathologically altered skin at wavelengths used in laser therapy. These z_0 values correspond to the lower and upper limits for the change of f_{m} and C_V from Table 1. The following wavelengths were used: 441, 470, 630, 670, 780, 812, 815, 830, 852, 855, 890,

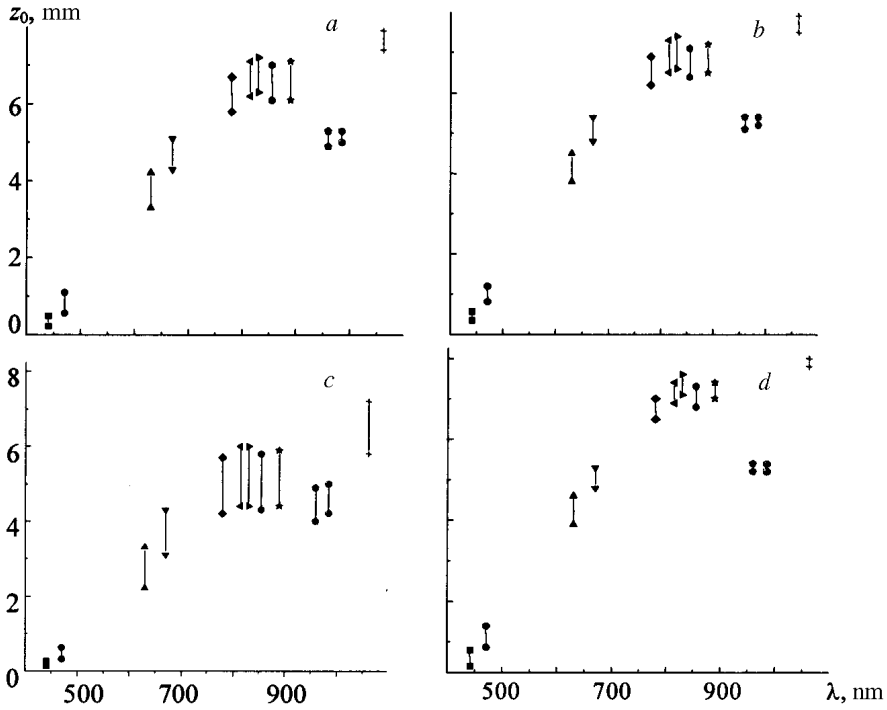

Fig. 2 Range of light penetration depth change for normal skin (a) and vitiligo (b), erythematosus lupus (c), and edema (d) as functions of wavelength.

TABLE 2 Average Absorbed Doses (J) of Laser Light

S	Pathology	Wavelength, nm														
		441	470	630	670	780	812	815	830	852	855	890	960	982	985	1062
0.97	Normal skin	0.85	0.82	0.60	0.56	0.52	0.52	0.52	0.52	0.53	0.54	0.54	0.64	0.65	0.65	0.58
	Vitiligo	0.69	0.60	0.36	0.36	0.39	0.41	0.41	0.42	0.45	0.45	0.47	0.60	0.62	0.62	0.54
	Erythematosus lupus	0.88	0.86	0.63	0.58	0.58	0.59	0.59	0.60	0.61	0.61	0.63	0.69	0.70	0.69	0.61
	Edema	0.83	0.79	0.59	0.55	0.50	0.50	0.50	0.50	0.51	0.51	0.52	0.63	0.64	0.64	0.57
	Subcutaneous wound	0.98	0.96	0.39	0.36	0.43	0.46	0.46	0.47	0.51	0.51	0.53	0.66	0.67	0.66	0.56
0.75	Normal skin	0.86	0.81	0.61	0.57	0.52	0.52	0.52	0.52	0.53	0.53	0.54	0.64	0.65	0.65	0.57
	Vitiligo	0.73	0.58	0.39	0.37	0.40	0.41	0.41	0.42	0.44	0.45	0.47	0.60	0.61	0.61	0.53
	Erythematosus lupus	0.88	0.86	0.67	0.62	0.59	0.59	0.59	0.59	0.61	0.61	0.62	0.68	0.69	0.69	0.60
	Edema	0.83	0.79	0.59	0.56	0.51	0.50	0.50	0.50	0.51	0.51	0.52	0.63	0.64	0.64	0.57
	Subcutaneous wound	0.98	0.95	0.48	0.42	0.44	0.46	0.46	0.47	0.50	0.50	0.53	0.65	0.65	0.66	0.55
0.50	Normal skin	0.87	0.81	0.62	0.58	0.52	0.52	0.52	0.52	0.53	0.53	0.54	0.64	0.65	0.65	0.57
	Vitiligo	0.75	0.57	0.41	0.39	0.40	0.41	0.41	0.41	0.44	0.44	0.46	0.60	0.61	0.61	0.53
	Erythematosus lupus	0.89	0.76	0.71	0.65	0.59	0.58	0.58	0.59	0.60	0.60	0.61	0.68	0.68	0.67	0.58
	Edema	0.84	0.79	0.59	0.56	0.51	0.50	0.50	0.50	0.51	0.52	0.52	0.63	0.64	0.64	0.56
	Subcutaneous wound	0.98	0.93	0.55	0.47	0.45	0.45	0.45	0.46	0.49	0.49	0.52	0.64	0.65	0.64	0.55

960, 982, 985, and 1062 nm. It can be seen that z_0 varies from fractions of a millimeter in the blue region of the spectrum to ~8 mm in the near-IR region. The z_0 values increase only slightly for vitiligo relative to normal skin although their range of change slightly narrows. This is due to the thinness of the epidermis (60 μm) so that the penetration depth is determined mainly by the light distribution in the dermis. Thus, the z_0 values decrease relative to normal skin for erythematosus lupus, where the average diameter of the capillaries and, therefore, the volume concentration of vessels and the absorption index of the dermis increase. The depth z_0 increases for edema, which is manifested by an increased concentration of intertissue fluid. The range of change of z_0 for edema expands into the blue

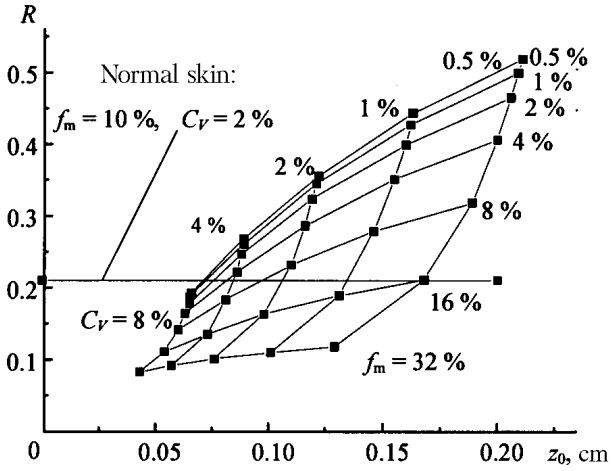

Fig. 3 Nomogram of reflectance coefficient vs. light penetration depth for λ = 570 nm.

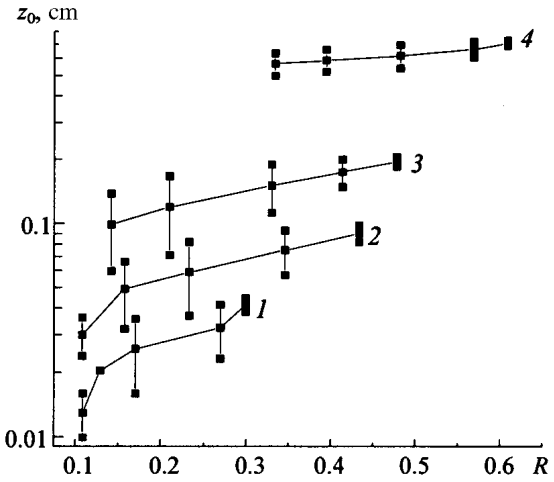

Fig. 4 Range of z0 change as a function of reflectance coefficient for λ = 420 (1), 450 (2), 570 (3), and 800 nm (4).

region of the spectrum and narrows in the green to near-IR regions compared with erythematosus lupus. This is due to the behavior of the absorption spectra of oxy- and deoxyhemoglobin and the concentration of vessels. For a subcutaneous wound, z_0 = 36 μm at λ = 441 nm and full illumination falls by 10 times even in the near-surface region. The penetration depth is similar to the corresponding z_0 values for normal skin at other examined wavelengths. Such behavior is due to the absence of epidermis and absorption by blood rather than melanin. In general, the light penetration depth into normal and pathologically altered skin does not vary widely.

Light Penetration Depth as a Function of Reflectance Coefficient. How are the spectral reflectance coefficient $R(\lambda)$ and the light penetration depth into tissue related to each other? Or in other words, can $z_0(\lambda)$ be predicted by measuring $R(\lambda)$? It is understandable that the structural and optical properties of skin are related in this way. It is also clear that the relationship is ill defined because of the unknown concentrations of melanin and blood. Figure 3 shows a nomogram in coordinates of reflectance coefficients vs. penetration depth for λ = 570 nm that is constructed for one of the isosbestic points of the spectra of oxy- and deoxyhemoglobin in order to avoid an effect from the de-

gree of oxygenation of blood. (Remember that an isosbestic point corresponds to equal absorption indices for oxy- and deoxyhemoglobin.) The volume concentration of blood vessels varies along the curves with smaller slopes relative to the abscissa; the melanin concentration, along the "vertical" curves intersecting them. The corresponding values are given in the figure. A horizontal line is also drawn in the nomogram for "nominally normal" skin. The range of change of C_V and f_m practically encompasses the limits of change for these concentrations for normal and pathologically altered skin. In principle, the nomogram enables the light penetration depth to be determined from the diffuse reflectance coefficient at this wavelength if one of the concentrations is known and, therefore, it enables the light field inside the medium to be estimated from the measured reflectance coefficient. Unfortunately, neither the melanin concentration nor the vessel concentration is usually known in practice (if additional measurements of C_V or f_m by biochemical or other methods are not made). The nomogram shows that a key interval of z_0 values can be set at constant R. It varies for different R (see the corresponding curve in Fig. 4). Analogous nomograms can be constructed for other isosbestic points and the possible interval of z_0 values can be determined (Fig. 4). The maximum variations of z_0 occur in the visible spectrum. Here the penetration depth varies by ~1.5–2 times. Such uncertainty in the determination of z_0 by measuring the reflectance coefficient is often acceptable for estimates. The limits of change of z_0 are lower in the near-IR region. Naturally the range of change of z_0 narrows at the maximum and minimum (not shown in Fig. 4) R values corresponding to the lower and upper boundaries of C_V and fm.

Conclusion. Absorption coefficients of human skin and the light penetration depth into skin at wavelengths used for low-intensity laser therapy were calculated using a multilayer model of skin and several of its pathologies (vitiligo, erythematosus lupus, edema, subcutaneous wound) that is based on engineering methods of light transfer theory. The potential for predicting the penetration depth by measuring the spectral diffuse reflectance coefficient of skin was evaluated. It was shown that the error of such an estimation in the visible region of the spectrum is less than two times and decreases in the near-IR region. This is often acceptable in practice. Relationships of the diffuse reflectance coefficient and the penetration depth were found and are apparently the only noninvasive method for determining the light properties within biological tissues from light fields outside the medium. The results can be used in the clinic and research to develop procedures for optimizing noninvasive methods for light treatment of inner parts of the medium at the radiation dose and light wavelength.

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