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Influence of skin surface roughness degree on energy characteristics of light scattered by a biological tissue

V.V. Barun, A.P. Ivanov

Abstract. **We present the results of modelling of photometric characteristics of light in soft tissues illuminated by a parallel beam along the normal to the surface, obtained with allowance for the skin roughness parameters and the angular structure of radiation approaching the surface from within the tissue. The depth structure of the fluence rate and the spectra of the diffuse reflection of light by the tissue in the interval of wavelengths 300–1000 nm are considered. We discuss the influence of the tilt angle variance of rough surface microelements and light refraction on the studied characteristics. It is shown that these factors lead to the reduction of the radiation flux only in the near-surface tissue layer and practically do not affect the depth of light penetration into the tissue. On the other hand, the degree of the surface roughness and the conditions of its illumination from within the tissue essentially affect the coefficient of diffuse reflection of light and lead to its considerable growth compared to the cases of a smooth interface and completely diffuse illumination, often considered to simplify the theoretical problem solution. The role of the roughness of skin surface is assessed in application to the solution of different direct and inverse problems of biomedical optics.**

Keywords: *rough surface, skin, radiation transport theory, light reflection, refraction.*

1. Introduction

Photometric (quadratic in electromagnetic field strength) characteristics of radiation inside and outside a biotissue are the basis for solving a wide scope of biomedical optics problems, e.g., the optimisation of light therapy techniques, laser hyperthermia, and optical diagnostics. In the process of tissue illumination, the light passes through the interface between the skin and the environment, and to a certain degree changes its energy and angular parameters. Therefore, the skin surface affects the spatial distribution of the radiation power, absorbed and scattered in the tissue, the depth of light penetration, and the spectra of the effect on the tissue chromophores and other characteristics of the radiation–biotissue interaction.

In the majority of papers, the light transport in biotissues is theoretically considered neglecting the roughness of the surface. This is due to a few reasons. First, the solution of the radiation transport problem in the medium is essentially simplified, since the transmission of light through the interface is described by the well-known Fresnel formulae. Second, the shortage of experimental data on the parameters of skin surface roughness that could provide a base for the statistical description of its structure exists. The main difficulty in the study of light propagation in a scattering medium with a rough boundary is to account for the interaction of multiply scattered radiation with the surface and the mutual influence of these two processes. Thus, in the terahertz frequency range (the wavelengths $\lambda = 100-1000$ μm) they usually consider transparent (dielectric) or strongly absorbing (conducting) objects [1–3], so that the contributions from their volume and rough surface can be separated and considered independently. In other words, in the above interval of wavelengths the multiple scattering in a bulk medium is neglected. This assumption is quite correct for biotissues, since the characteristic size of their inhomogeneities is usually much smaller than λ. This fact manifests itself in the opacity of biotissues, clothes, different packing materials for goods, and similar objects for the radiation of terahertz frequencies.

In the optical range of wavelengths the situation is different in principle. Here the multiple light scattering by biotissues cannot be eliminated, and it is necessary to allow for the interaction of light with the medium surface. A few publications are known, devoted to the numerical solution of the radiation transport equation in a biotissue using the Monte Carlo method with the skin roughness taken into account [4–9]. The authors of Ref. [4] considered the surface with the Gaussian probability density of the surface eminences and the Gaussian correlation function, and the authors of Refs [5–9] considered the randomly inhomogeneous quasi-periodic (sinusoidal) surface. Note that the Monte Carlo simulation, alongside with the known advantages (the possibility to consider practically any configurations of the medium and any spatial distributions of optical properties), possesses definite drawbacks. One of the major drawbacks is the laboriousness of the computation process that requires large memory and computer time. Thus, in Ref. [2] the computation of only one variant took almost two hours.

The aim of the present paper is to develop a semianalytic method that allows 'pure' selection of the effect of skin roughness parameters on the light characteristics in a biotissue and, thus, an answer to the question how essentially the approximation of the real surface by a smooth interface affects these characteristics. In other words, is it generally required to complicate the problem, and in which situations the interface roughness is important to be taken into account?

Note that polarisation and coherence effects also accompany the transmission of light through a rough surface.

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However, their investigation is a separate problem and they are not discussed in the present paper.

2. Model of biotissue and skin surface

We consider a three-layer tissue, consisting of a stratum corneum (sc), epidermis (e) and homogeneous dermis (d). The optical and structural characteristics of each layer were obtained based on the critical analysis and generalisation of the appropriate literature data [10–17] and the results of our own studies [18, 19] in the spectral range 300–100 nm. The tissue model includes the geometrical thicknesses d_{sc} and d_e, the refractive indices n_{sc}, n_e, and n_d, as well as the optically significant biophysical parameters, namely, the volume concentrations of melanin in the epidermis f_m and blood capillaries in dermis C_v, the degree of blood oxygenation S, and the mean diameter of capillaries D. The model allows direct calculation of the characteristics of absorption and scattering by the tissues layers based on the known specific absorption indices of melanin and haemoglobin derivates [12, 13, 16]. The influence of the 'sieve' effect (the localised absorption of light by blood vessels and erythrocytes) on the optical properties of the medium that occurs in the blue-violet spectral region was assessed by means of the technique of Refs [17, 18], and for the effect of blood haematocrit, or erythrocytes packing density, the data of Refs [19, 20] were used. The calculations using the analytical formulae from Ref. [18] have shown that the mean value of D allows adequate estimation of the contribution from the network of microvessels of different size to the absorption index of the tissue. In this case, the desired energy characteristics of multiply scattered light differ only by a few percent from those calculated using the capillary diameter distribution function.

The effective (reduced) scattering indices of the layers were calculated using the approximate formulae [12, 16] comprising two terms. The first of them describes the scattering by large-scale fibrils, and the second one by small-scale particles of the tissue. The spectral dependence of the mean cosine of indicatrix (asymmetry coefficient) was adopted from Ref. [11]. Below for definiteness we assume the following parameters to be constant: $d_{sc} = 20$ μm, $d_e = 100$ μm, $C_v = 0.02$, $S = 0.75$. The dermis is considered as a semi-infinite layer. The refractive indices of epidermis and dermis are assumed to be close to each other, $n_e = n_d = 1.33$ [14], and determined by water, the dominant component. The discontinuity of the refractive index inside the tissue exists at the interface between the stratum corneum ($n_{sc} = 1.55$) and the epidermis.

Below, like in Refs [4–9], we will consider two types of roughness. Their characteristics are commonly specified with respect to the macroscopically planar medium surface ($z = 0$). From this surface, we will also measure the depth of the observation point inside the tissue. Let the probability densities $P(\gamma)$ of random tilts of surface elements relative to the plane $z = 0$ for the Gaussian and sinusoidal rough surfaces have the form [21, 22]

$$P(\gamma) = \frac{1}{2\pi D_\gamma}\exp\left(-\frac{\gamma^2}{2D_\gamma}\right), \quad (1)$$

$$P_s(\gamma) = \begin{cases} [\pi\sqrt{\gamma_{max}^2 - \gamma^2}]^{-1} \text{ for } |\gamma| \leqslant \gamma_{max}, \\ 0 \text{ for } |\gamma| > \gamma_{max}, \end{cases} \quad (2)$$

respectively, where D_γ is the variance of tilt angle distribution; $\gamma_{max} = \omega\zeta_{max}$ is the maximal slope of the sinusoid relative to the plane $z = 0$; ω and ζ_{max} are the angular frequency and the amplitude of the sinusoid. The values of parameters (1) and (2) typical for human skin are presented in review [22]. Using distributions (1) and (2) and applying the technique [21, 23] to take into account the mutual shadowing of the interface relief elements, one can calculate the angular structure of radiation [24], reflected and (or) transmitted by the rough surface, as well as the integral parameters, namely, the coefficients of reflection (R) and transmission ($1 - R$) [25] for any angular pattern of illumination. We emphasise that the values of R are calculated [21, 23] taking into account the constructive interference of light, propagating in the far-field zone in the direction of the local Fresnel reflection angle and the destructive interference for other angles. In a similar way, the transmission coefficient is found. The technique of Refs [21, 23] allows also one to determine polarisation characteristics of radiation, reflected and refracted by the rough surface. However, as pointed out above, they are not considered at the present stage. The results of Refs [21, 23–25] are applicable to large-scale (compared to the wavelength of radiation incident on the tissue) surface irregularities, which is usually typical for human skin in the visible and near-infrared spectral regions. Note that in works [1, 2] devoted to the terahertz frequency region, a similar approach was used, but without the mutual shadowing of the interface relief elements taken into account.

3. Method for calculating the energy characteristics of multiply scattered light

The authors of papers [26–28] proposed an analytical scheme for the calculation of integral radiation characteristics in a multilayer biotissue. By 'integral' we mean here the fluence rate, or spatial luminance [29]

$$W(\lambda, z) = \int_{4\pi} I(\lambda, z, \theta, \varphi)\,\mathrm{d}\Omega$$

(I is the luminance of light at the depth z; θ and φ are the polar and azimuthal angles; $\mathrm{d}\Omega = \sin\theta\mathrm{d}\theta\mathrm{d}\varphi$ is the solid angle element) and the tissue diffuse reflection coefficient R_{sk}, equal by definition to the ratio of the reflected flux to the incident one. The quantities W and I will be normalised below to the illumination of the medium surface by the incident light.

Let the parallel beam of light be incident along the normal (along the positive direction of the z axis) to the skin surface. The multiple scattering of light in each of the tissue layers causes the formation of diffuse reflected (backscattered) fluxes. Within the biotissue, the radiation flux is formed by multiple scattering of directed and diffuse fluxes and multiple repeated reflection of radiation between the layers of the medium and the skin surface. Detailed analytical expressions for the calculations are presented in Refs [26–28]. The coefficients of reflection (R) and transmission ($1 - R$) of the incident directed beam by the surface that enter these expressions together with the reflection (R^*) and transmission ($1 - R^*$) coefficients of the surface, illuminated from the inside by the diffuse radiation, determine the specific features of the rough tissue-environment interface.

Naturally, the values of $R^*(\lambda)$ depend on the directional pattern of the radiation that illuminates the skin surface. The

polar (η) and azimuthal (φ) angular intensity distribution of light approaching the surface from within the tissue can be presented as

$$I(\eta,\varphi)=\begin{cases}T_{sc}\,T_e I_e(\eta)I_d(\eta)n^2+T_{sc}I_{1e}n^2\\ +\dfrac{T_{sc}\delta(1-\cos\eta)\,\delta(\varphi)(n-1)^2}{(n+1)^2}\ \text{for}\ \ \eta\leqslant\arcsin\dfrac{1}{n},\\ 0\ \ \text{for}\ \ \arcsin\dfrac{1}{n}<\eta\leqslant\dfrac{\pi}{2},\end{cases}\tag{3}$$

where T_{sc} and T_e are the coefficients of transmission of the directional radiation by the stratum corneum and the epidermis, respectively; $n=n_{sc}/n_e$ is the relative refractive index; δ is the delta function; I_{1e} is the intensity of backscattered light; and the angle η is measured from the negative direction of the z axis and specifies the direction of light incidence on the skin surface from within the tissue. This angle is related to the polar angle θ of the radiation propagation inside the medium as $\theta=\arcsin(n\sin\eta)$ and allows for the refraction in the course of light propagation from the epidermis having the refractive index n_e to the skin surface having the refractive index n_{sc}. The first term is proportional to the product of the luminance coefficient $I_d(\eta)$ of the light backscattered by the dermis, and the coefficient $I_e(\eta)$ of the transmission of this light by the optically thin epidermis. The second term is proportional to the intensity and corresponds to the reflection by the epidermis of the radiation that did not reach the dermis. The third term corresponds to the Fresnel reflection of light from the interface between the stratum corneum and the epidermis.

In Eqn (3), the reflection from the stratum corneum is not taken into account because of its small optical thickness. We also neglect in Eqn (3) the angular divergence of the beam, normally incident on the rough skin surface, since it weakly affects the reflection coefficient R^* [25]. The spectral dependences of $I_e(\eta)$ and I_{1e} were calculated in the small-angle approximation [30, 31], and $I_d(\eta)$ – in the asymptotic approximation [30, 32] of the transport theory. The applicability of these approximations to the media having the optical parameters similar to those of biotissues has been repeatedly confirmed by comparison with the experimental data [30], the results of Mote Carlo simulations [33] and other numerical methods [34]. Let us write the functions that enter Eqn (3) in the explicit form

$$I_e(\theta)=\exp\left[-\frac{d_e(\mu_{ee}-\mu_{sc}F_e)}{\cos\theta}\right],\tag{4}$$

$$I_{1e}=\frac{\Lambda_e}{2\pi}\frac{1-F_e}{1-F_e\Lambda_e}\int_0^1\left\{1-\exp\left[-\mu_{ee}(1-\Lambda_eF_e)d_e\frac{1+\cos\theta}{\cos\theta}\right]\right\}\times\frac{\cos\theta}{1+\cos\theta}\mathrm{d}(\cos\theta),\tag{5}$$

$$I_d(\theta)=\frac{\rho_0(\theta)-\{1-\exp[-4G(\theta)\,G(0)\sqrt{\mu_{ad}/(3\mu'_{ed})}]\}}{\pi}.\tag{6}$$

Here $F_e=1-(1-g_e)/3$ [33]; μ_{ee}, μ_{se} and g_e are the attenuation and scattering indices and the coefficient of the indicatrix asymmetry for the radiation scattered by the epidermis, respectively; $\Lambda_e=\mu_{se}/\mu_{ee}$ is the probability of photon survival; $\rho_0(\theta)=0.5(1+4\cos\theta)/(1+\cos\theta)$ is the luminance factor of the semi-infinite non-absorbing medium [30]; $G(\theta)=3(1+2\cos\theta)/7$; $\mu'_{ed}=\mu_{ad}+\mu_{sd}(1-g_d)$ is the effective attenuation index of dermis; and μ_{ad}, μ_{sd} and g_d are the attenuation and scattering indices and the coefficient of the indicatrix asymmetry for the radiation, scattered by the dermis, respectively. Due to the weakly expressed angular structure of the indicatrix of the single scattering by the epidermis at $\gamma>\pi/2$ [11, 12] it was assumed that the second term in Eqn (3) does not depend on the polar angle θ. In other words, the light formed by backscattering of the incident radiation in the epidermis is practically diffuse. Note that all above parameters of the elementary volume depend on the incident light wavelength λ, in particular, $g_d=g_e=0.62+0.00029\lambda$ [11], where λ is expressed in nanometres.

Thus, the technique for modelling the integral characteristics inside and outside a multilayer medium includes two stages. At the first stage the spectral dependences of R and R^* are calculated [25] under the variation of the structural and biophysical parameters of the tissue using Eqns (1)–(6), and then, using the technique of Refs [27, 28], the desired characteristics at the chosen values of the above medium parameters are calculated. The calculations that take fractions of a minute do not require particularly large computer memory. Note that when D_γ changes within the range 0–0.44 [22], the reflection coefficient of the medium illuminated from outside along the normal to the macroscopic surface ($z=0$) changes weakly, $R=0.047$–0.027 [25] and, therefore, practically does not affect the photometric characteristics inside and outside the biotissue.

4. Results and discussion

4.1. Reflection coefficient of skin surface illuminated from within the medium

As mentioned above, probability densities (1) and (2) were used to describe the roughness properties of the surface. The earlier calculations have shown [25] that if the variance values for the tilt angles of the relief elements in Eqns (1) and (2) are set to be similar, i.e., $D_\gamma=\gamma^2_{max}/2$, then the relative difference in the reflection coefficient R^* does not exceed 5%–7% within a wide interval of D_γ, typical for human skin [22]. Therefore, below we consider only the Gaussian rough surface with the probability density (1).

By definition, the quantity R^* is equal to the ratio of the total reflected radiation flux to the incident one:

$$R^*=\frac{\int_{2\pi}r^*(\eta^*)I[\arcsin(n\sin\eta)]\sin\eta\cos\eta\,\mathrm{d}\omega}{\int_{2\pi}I[\arcsin(n\sin\eta)]\sin\eta\cos\eta\,\mathrm{d}\omega},\tag{7}$$

where $r^*(\eta^*)$ is the reflection coefficient of the skin surface illuminated from within the tissue at the local angle η^* of light incidence on surface elements; and $\mathrm{d}\omega=\sin\eta\mathrm{d}\eta\mathrm{d}\varphi$. The relation between the angles η^* and η is given in Refs [24, 25]. As seen from Eqns (3)–(7), the coefficient R^* is formed by three fluxes with different angular patterns, the partially and completely diffuse light [the first and the second terms in Eqn (3), respectively], and the normally directed radiation (the third term). Obviously, for the above three components of intensity (3) the reflection coefficient is maximal for the second flux and minimal for the third one. This is due to the appropriate dependences of the Fresnel reflection coefficient on the angle of incidence η^*. To clarify the physical mechanisms of interaction between the multiply scattered radiation and the rough surface, and their influence on the formation of the fluence rate in the medium, let us analyse the contributions to R^*

from the three terms in Eqn (3) at different wavelengths, i.e., from the radiation fluxes, backscattered in the dermis (Φ_1), in the epidermis (Φ_2), and at the interface between the epidermis and the stratum corneum (Φ_3). By definition, the fluxes Φ_{1-3} are such:

$$\Phi_1 = \int_{2\pi} r^*(\eta^*)\, T_{sc}\, T_e I_e[\arcsin(n\sin\eta)] I_d[\arcsin(n\sin\eta)] n^2 \times \sin\eta\cos\eta\, d\omega, \quad (8)$$

$$\Phi_2 = \int_{2\pi} r^*(\eta^*)\, T_{sc} I_{1e} n^2 \sin\eta\cos\eta\, d\omega, \quad (9)$$

$$\Phi_3 = \left(\frac{n-1}{n+1}\right)^2 r^*[\eta^*(\eta = 0)]\, T_{sc}, \quad (10)$$

where in Eqn (10) the reflection coefficient r^* is calculated for the incidence angle η^* corresponding to $\eta = 0$, and the interface between the epidermis and the stratum corneum is assumed to be plane. In particular, for the smooth skin surface $r^* = [(n_{sc} - 1)/(n_{sc} + 1)]^2$. Note that the third term in Eqn (3) weakly affects the reflection coefficient R^*, and its contribution Φ_3 (0.0002–0.0005 depending on the surface roughness) is small compared to the contributions Φ_1 and Φ_3.

The spectral dependences of Φ_1 and Φ_2 are presented in Fig. 1a. It is seen that in the short-wavelength region the main contribution to the reflection coefficient R^* is made by the radiation backscattered from the epidermis (Φ_2). This is due to the strong absorption of light by melanin in the near-UV and blue regions of the spectrum, so that a considerably attenuated flux reaches the dermis, which practically does not affect the values of R^*. The greater is the concentration f_m and the thickness d_e of the epidermis, the more the dermis contribution Φ_1 to R^* exceeds that of Φ_2, starting from some greater wavelengths λ. Thus, for $f_m = 0.04$ we have $\Phi_1 > \Phi_2$ for $\lambda >$ 500 nm, and for $f_m = 0.16$ the above inequality is valid for $\lambda >$ 550–600 nm. In the red spectral region at $\lambda > 650$ nm the reflection coefficient R^* is mainly affected by the light (Φ_1), multiply scattered by the dermis. In the spectra of the flux Φ_1 the absorption bands of blood haemoglobin derivates manifest themselves at $\lambda \approx 418$ and 540–580 nm. As follows from Eqn (6), in the maxima of these bands the light backscattered by the dermis is concentrated near $\theta = 0$, which leads to a slower growth of the flux Φ_1 with an increase in λ as compared to the wavelengths beyond the above absorption peaks, particularly at low concentrations f_m. The reason is a decrease in the reflection coefficient r^* at the light incidence near the normal to the macroscopic surface ($z = 0$).

The above features of formation of the fluxes Φ_{1-3}, naturally, manifest themselves in the spectrum of R^*. We consider them below by analysing the dependences of R^* on the wavelength (Fig. 1b), calculated using Eqns (3)–(7) with the mutual shadowing of the relief elements taken into account [25]. Attention should be payed to both the spectral behaviour of the reflection coefficient R^* and its magnitude. Above we mentioned two simplifying assumptions, frequently used to solve the problems of biomedical optics. First, they neglect the surface roughness. Second, they assume $I(\eta)$ = const (absolutely diffuse illumination). Then $R^* = 0.625$ and does not depend on λ. As seen from Fig. 1b, with the angular structure of radiation $I(\eta)$ and roughness taken into account, the values of R^* are essentially smaller. In the near UV and blue

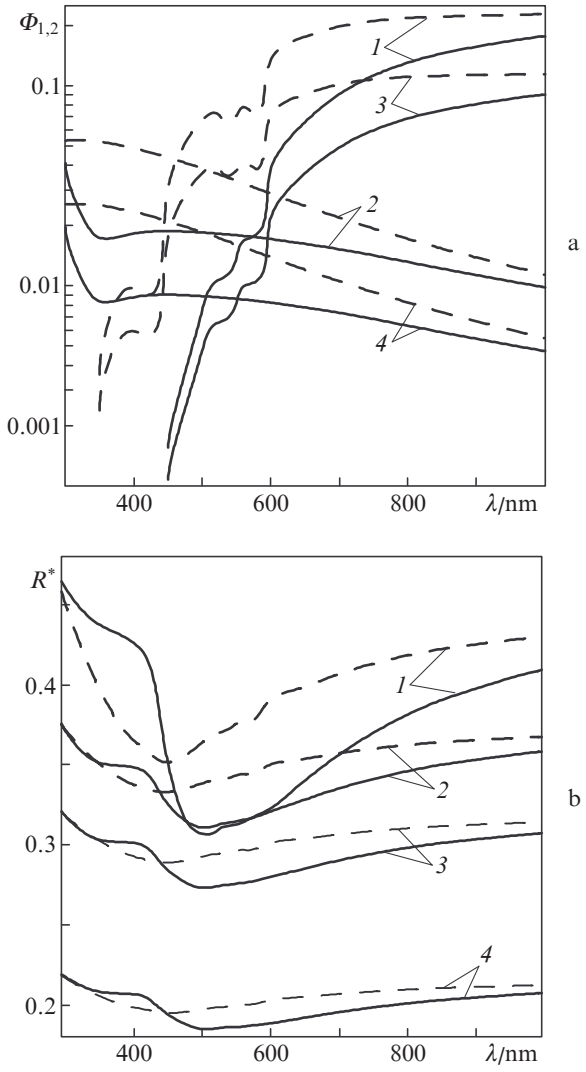

Figure 1. Dependences of (a) (*1*, *3*) Φ_1, Φ_2 (*2*, *4*) and (b) R^* on λ for $f_m = 0.16$ (solid curves) and 0.04 (dashed curves); (a) $D_\gamma = (1, 2)$ 0.0256 and (*3*, *4*) 0.444; (b) $D_\gamma = (1)$ 0, (*3*) 0.2 and (*4*) 0.444.

spectral regions ($\lambda < 450$ nm) the reflection coefficient R^* is formed by entirely diffuse fluxes backscattered in the epidermis. However, due to the refraction near the skin surface, these fluxes are concentrated in the cone with the half apex angle $\eta_0 = \arcsin(1/n) \approx 60°$. As a result, the absolute values of R^* are essentially smaller than in the case $I(\eta)$ = const even for a smooth surface [curves (*1*)]. As the variance D_γ grows, such a decrease in R^* becomes even more significant.

With increasing λ, the scattering index of epidermis considerably decreases [12, 15, 31]. Hence, the flux Φ_2 related to the reflection of light from the epidermis also decreases. Simultaneously, the flux from the dermis approaching the internal surface of skin rapidly increases. As seen from Eqn (6), the luminance of this light has a maximum at $\theta = 0$. Besides that, in the short-wavelength spectral region, where the epidermis melanin strongly absorbs light, the oblique beams (large angles θ) are visibly weakened [see Eqn (4)], so that the radiation, incident on the interface from within the medium, is concentrated near $\eta = 0$. The Fresnel reflection coefficient of light incident on the skin surface near the normal is not large. For these reasons, mathematically expressed by Eqns (4) and (6), as well as due to the reduction of the scat-

tering index of epidermis, the values of $R^*(\lambda)$ in the short-wavelenght spectral region become smaller. In the case of small variance D_γ [curves (*1*) and (*2*)] the spectral behaviour of the albedo R^* is evident. The position of the R^* minimum corresponds to $\lambda = 500-600$ nm depending on the melanin concentration. With an increase in D_γ the concentration of the incident light near the normal to the surface becomes not so essential, since the fraction of oblique surface elements increases, and the spectral behaviour of R^* is smoothed. With an increase in λ and a decrease in f_m, the epidermis absorbs weaker, and the angular radiation pattern near the surface is wider. The Fresnel reflection coefficient of the interface increases and, therefore, R^* increases as well.

Let us estimate, which of the three factors, the angular structure of the dermis luminance coefficient [the first term in Eqn (3)], the attenuation of oblique beams in the epidermis (the second term), or the refraction in the stratum corneum, stronger affects the reflection coefficient R^*. The results of R^* calculations (not presented in Fig. 1) for the red and near-IR spectral regions at $I_d(\theta) \equiv 1$ have shown that the angular dependence $I_d(\theta)$ affects the albedo R^* weaker than the other factors mentioned above, especially at large D_γ. Therefore, the variations of the volume concentration C_v of the dermis capillaries within 0.02–0.08 and the values of S within 0.5–1 cause almost no change in the reflection coefficient R^*. The calculations with $I_e(\theta) \equiv 1$ and the dependence $I_d(\theta)$, calculated according to Eqn (6), show that R^* is virtually independent of λ: $R^* \approx 0.44$, 0.37, and 0.22 for $D_\gamma = 0$, 0.2, and 0.44, respectively. It follows that refraction is the main factor affecting the spectral behaviour and magnitude of R^*.

Figure 2 presents the dependences of the reflection coefficient R^* on the variance of the tilt angle of the rough skin surface at different wavelengths. It is seen that depending on λ [curves (*2*) and (*3*)] for small D_γ (practically smooth surface) a weakly expressed maximum of R^* can be observed. With the increase in the variance, the effect of mutual shadowing of microscopic relief elements becomes more evident, and the reflection coefficient R^* monotonically decreases. For a rough surface with $D_\gamma > 0.15$, the relative variations of R^* within the wavelength interval 300–1000 nm amount to nearly 5%–10% under a wide-range variation of the structural and biophysical parameters of the tissue, so that for many problems the reflection coefficient can be considered nearly constant in this spectral region and dependent only on the variance D_γ.

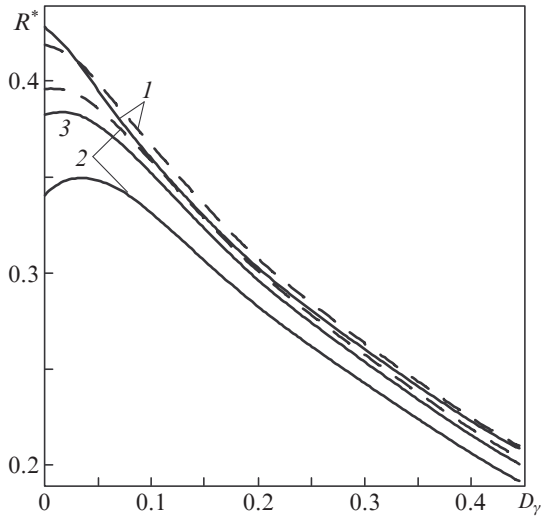

Figure 2. Dependences of R^* on D_γ for $f_m = 0.16$ (solid curves) and 0.04 (dashed curves); λ = (*1*) 400, (*2*) 600 and (*3*) 800 nm.

4.2. Fluence rate distribution over the biotissue depth

It was shown above that the coefficient R^* of the diffuse reflection of light by the skin surface essentially depends on the degree of its roughness and the angular structure $I(\eta)$ of the radiation, approaching the surface from within the tissue. The dependence $I(\eta)$ is determined by three factors, the major of which is the refraction in the stratum corneum. Commonly, this refraction as well as the growth of the variance D_γ lead to the reduction of R^*. Therefore, the refraction and the interface roughness taken into account must lead to smaller values of W near the skin surface. Such a behaviour of W is clearly seen in Fig. 3, where the depth profiles of the fluence rate at the wavelength 800 nm are presented. In the upper layers of the dermis, the values of W are really decreasing under the effect of the roughness degree and the refraction. This is particularly clearly seen in the maximum of the $W(z)$ profiles at low melanin concentrations f_m. However, inside the tissue at relatively large depths z, neither the surface roughness degree nor the refraction exert practical influence on the value of W. The physical reason is clear. The light, scattered by the surface, is sufficiently diffuse, and its intensity decreases with the growth of z faster than the intensity of the directed incident radiation and the radiation, scattered mainly in the forward direction. In particular, it follows that with the interface roughness and the refraction in the stratum corneum taken into account the light penetration depth remains practically unchanged.

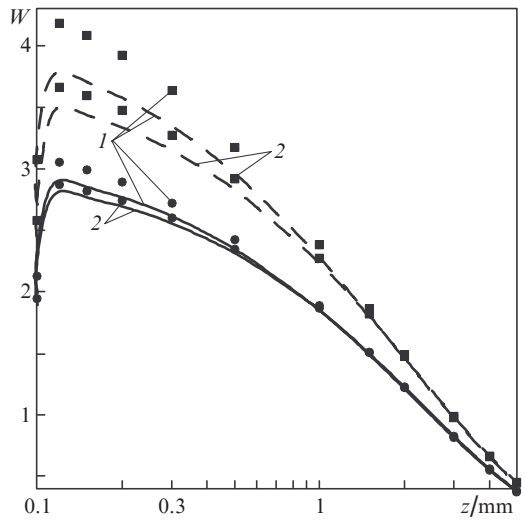

Figure 3. Dependences of W on z for $f_m = 0.16$ (solid curves, ●) and 0.04 (dashed curves, ■) with (lines) and without (●, ■) refraction taken into account; D_γ = (*1*) 0 and (*2*) 0.44, λ = 800 nm.

The data of Fig. 3 are indirectly confirmed by the results of Ref. [6] (for a rough surface) and Ref. [35] (for a smooth surface), where it is shown that the detected unpolarised signal is localised mainly in the upper layer of the tissue both for the change of the relief parameters [6] and for the variation of R^* [35].

4.3. Coefficient of diffuse reflection of light by biotissue with the skin surface roughness taken into account

Obviously, the change in the light diffuse reflection coefficient R^* is accompanied with the variation of the transmission coefficient $T^* \equiv 1 - R^*$. The interface roughness and refraction lead to the reduction of R^* and, therefore, the increase in T^* or the light flux that leaves the biotissue volume. The spectra of the skin reflection coefficient R_{sk}, obtained with the influence of the surface roughness, the angular pattern of the radiation intensity $I(\eta)$, Eqn (3), and the multiply repeated reflection between the tissue layers taken into account, are presented in Fig. 4. Here the points show the results of the calculation with $I(\eta) \equiv 1$ and the refraction neglected. It is seen that the skin surface roughness and the refraction lead to considerable transformation of both the shape of the spectra and the absolute values of R_{sk}. The albedo R_{sk} generally increases with increasing D_γ due to a smaller part of light reflected by the surface into the tissue. Particularly clear manifestation of this effect is observed in the red and near-IR wavelength ranges. Note also that at a large melanin concentration (solid curves) the specific features related to the absorption by blood at 400–600 nm vanish in the spectrum $R_{sk}(\lambda)$. The reason is the strong shielding of dermis by epidermis in this case.

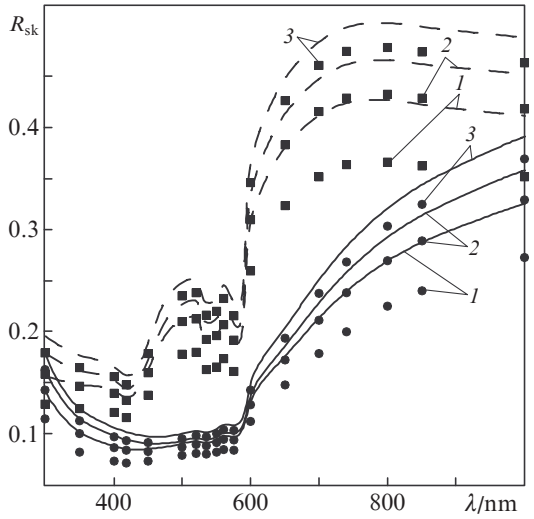

Figure 4. Spectrum of the coefficient of diffuse light reflection by skin for $f_m = 0.16$ (solid curves, ●) and 0.04 (dashed curves, ■); points – without refraction, D_γ = (*1*) 0, (*2*) 0.2 and (*3*) 0.44.

The dependences $R_{sk}(\lambda)$ are often used to solve the inverse problem aimed at the reconstruction of different parameters of soft tissues [36]. In this case, both the absolute values of the reflection coefficient and its spectral behaviour are important. The theoretical base of the solution is formed by the model calculations. Various analytical and numerical methods for calculating $R_{sk}(\lambda)$ are known. First, this is the Kubelka–Munk approximation (see, e.g., [37, 38]) and the approach [27, 28, 39] where R^* is a fitting parameter. We are not aware of any publications, where the analytical solution scheme would account for the refraction of light in the biotissue and the roughness of the skin surface. In the numerical solution of the transport equation using the Monte Carlo method with the different refractive indices specified and the layer boundaries [5–9], the refraction is automatically taken into account in the course of drawing the photon trajectories. However, in the majority of published papers (see, e.g., the references in [36]) the surface roughness is not considered. What are the consequences? To answer this question let us consider, e.g., curves (*1*) and (*3*) in Fig. 4, corresponding to $D_\gamma = 0$ and 0.44. In the visible spectral region at $\lambda < 600$ nm the absolute value of the difference of ordinates of these curves amounts to nearly 0.01–0.03, i.e., is comparable with the measurement error of the reflection coefficient. However, in the red and near infrared ranges the difference can approach 0.15 (particularly at small melanin concentrations). This is obviously unacceptable for solving the inverse problems.

5. Conclusions

The presented results allow the answer to the question, when in the solution of direct and inverse problems of biomedical optics one should account for the skin surface roughness. In the assessment of the radiation effect on the tissue chromophores (direct problem), the interest is commonly focused at the integral characteristics of light inside the medium. The data of Fig. 3 unambiguously confirm that in the study of light power absorbed by blood erythrocytes and the accompanying effects the surface roughness is practically not essential. The similar conclusion follows for the depth of light penetration into the medium. A different situation takes place in the study of tissue heating using the radiation in the red or near-IR spectral region. Thus, as shown in Ref. [31], in this case the major heating mechanism is the heat transfer from epidermis to dermis. In the epidermis, the thin near-surface layer of skin, the radiation density W decreases with increasing surface roughness degree due to the reduction of the coefficient R^* of light reflection into the medium. This is particularly evident at small concentrations of melanin. Such a reduction of W leads to smaller heating of the epidermis and, therefore, less intense heat transfer from this layer. As a result, the growth of the dermis temperature will be smaller than in the case of a smooth interface.

In the solution of inverse problems, special attention is paid to noninvasive methods, i.e., those, producing no destruction. Various methods of reconstructing the structural and biophysical parameters of soft tissues from the parameters of backscattered light are known. For example, in Ref. [39] the method of noninvasive determination of f_m, d_e, C_v, and S from the measurements of albedo R_{sk} is proposed. From the above consideration it unambiguously follows that for the solution of the inverse problem it is necessary to take the skin surface roughness and the light refraction in the stratum corneum, or the angular pattern $I(\eta)$, into account. In this case, the solution is complicated by two factors: the appearance of an additional unknown D_γ, and the dependence of R^* and, therefore, the measured quantity R_{sk}, on the sought structural and biophysical parameters of epidermis and dermis. In this context, the slight effect of C_v and S on R^* is favourable, so that one can put $I_d(\theta) \equiv 1$. Then the function $I(\eta)$ will depend only on the product $f_m d_e$, and one can develop a technique similar to that of Ref. [39] to determine C_v together with S from the measurements of R_{sk} at few wavelengths. Note that the reconstruction of the parameter D_γ, characterising the degree of the skin surface roughness, can be of interest in dermatology and cosmetology. However, the detailed consideration of this issue is beyond the scope of the present paper.

To summarise, we emphasise that the skin surface roughness practically does not affect the energy characteristics of the scattered light in the deep (more than 1 mm) layers of dermis. At $z < 1$ mm the spatial illuminance decreases and its reduction is particularly affected by the model parameters D_γ, f_m, d_e, as well as the wavelength λ. On the contrary, outside the medium the refraction of radiation and the surface roughness lead to the growth of the skin reflection coefficient, which should be taken into account in the solution of a number of problems of biomedical optics.

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