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Analysis of the Spatial Distribution of Detector Sensitivity in a Multilayer Randomly Inhomogeneous Medium with Strong Light Scattering and Absorption by the Monte Carlo Method

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Abstract—The spatial distribution of sensitivity in the domain of detection of a fiber-optic sensor used for spectrophotometric studies of skin and other biological tissues is studied. The method and results of modeling the propagation of optical radiation in multilayer randomly inhomogeneous media with strong light scattering and absorption are presented. Owing to the small distances between the source and detector (100–800 μm), the propagation of radiation in the medium under study is modeled by the Monte Carlo method combining the calculation of true paths and the use of statistical weights. For the same reason, we represent the surface and interfaces of layers of skin as rough randomly periodic surfaces corresponding to the actual structure of human skin. The method presented can be recommended as a means for the optimum selection of an arrangement for radiation incoupling and outcoupling.

INTRODUCTION

Since the 1950s, spectral methods have been widely used for various diagnostic biomedical studies of skin and blood [1]. At present, the study of diffusely reflected absorption spectra of blood-containing biological tissues and skin is of great interest for physiology and medicine [2].

Because of the inhomogeneous spatial distribution of blood vessels [3, 4], the absolute quantitative determination of the degree of saturation of blood in human skin with oxygen from the experimental absorption spectra is extremely difficult and requires that the contribution of a vascular channel to the detected signal be clearly known. An analogous problem is well known in laser Doppler anemometry [5, 6].

In addition, one should take into account the fact that skin, as an object of study by optical methods, is a rather complicated multicomponent medium containing many chromophores absorbing radiation propagating in the medium in accordance with their characteristic absorption bands [7, 8]. The greatest contribution to absorption in the visible and near infrared spectral regions is made by such chromophores as oxyhemoglobin (HbO_2), reduced hemoglobin (Hb), melanin, and water [1, 7–9]. It is the quantitative relation between these chromophores in the volume measured that determines the shape of the absorption spectrum [1, 8, 9].

In this paper, we present the results of the analysis of the sensitivity distribution of a detected signal in a multilayer randomly inhomogeneous medium with strong scattering and absorption of light within the framework of the study of possibilities for spatial localization of the signal detected in upper skin layers containing capillary loops. The small (less than 1 mm) distance between emitting and detecting optical fibers (the source and detector) in a fiber-optic sensor, predetermined by a shallow subsurface position of capillary loops (at a depth of the order of 100 μm), prohibits the use of the diffusion approximation [10] for this study. In this connection, we used the Monte Carlo method for calculating the spatial distribution of detector sensitivity inside the medium as a function of the distance between the source and the radiation detector.

MODELING OF THE PROPAGATION OF RADIATION IN A MULTILAYER RANDOMLY INHOMOGENEOUS MEDIUM WITH STRONG LIGHT SCATTERING

Following [8, 11–14], we represent skin as a half-space ($z > 0$) containing several layers, each characterized by its own geometrical dimensions (the mean thickness d, the scattering μ_s and the absorption μ_a coefficients, the anisotropy factor g, and the refractive index n). Typical values of these parameters for normal human skin are presented in Table 1.

Taking into account the different cell structures of skin layers [23–25] and inhomogeneous distributions of blood vessels [3, 4] and chromophores [7, 8, 25] with

Table 1. Optical properties of skin layers used in modeling (λ = 633 nm). The data were taken from [8, 15–24]

k	Layer	d, μm	μ_s, mm^{-1}	μ_a, mm^{-1}	g	n
1	Horny-tissue layer	20	100	0.8	0.9	1.5
2	Epidermis	80	60	0.15	0.85	1.34
3	Dermis with capillary loops	150	30	0.07	0.9	1.39
4	Dermis with surface plexus of vessels	80	35	0.1	0.95	1.4
5	Dermis	1500	25	0.05	0.76	1.39
6	Dermis with depth plexus of vessels	80	35	0.15	0.95	1.4
7	Hypodermic fat	6000	15	0.075	0.8	1.44

Table 2. Parameters used for modeling interfaces of skin layers. The data were taken from [4, 23–25]

k	Upper boundary of the layer	A_{kx}, A_{ky}, μm	$(\pi/\omega)_{kx}$	$(\pi/\omega)_{ky}$	Z_k, μm
1	Horny-tissue layer	2	100	150	0
2	Epidermis	2.5	80	80	20
3	Dermis with capillary loops	20	50	45	100
4	Dermis with surface plexus of vessels	2	20	40	250
5	Dermis	2	20	50	330
6	Dermis with depth plexus of vessels	2	20	50	1830
7	Hypodermic fat	5	20	50	1910
8	Muscle tissue	5	25	30	8000

depth, we represent the interfaces between skin layers as wavy randomly periodic surfaces (Fig. 1)

$$B_k(\mathbf{r}) = Z_k + (A_k \sin(\omega_{kx} x + \phi_{kx}) + a_k \sin(\omega'_{ky} y + \phi'_{kx})) \times (A_k \sin(\omega_{ky} y + \phi_{ky}) + a_k \sin(\omega'_{ky} y + \phi'_{ky})). \quad (1)$$

Here, $Z_k = K_{k-1} + d$ is the mean depth of the upper boundary of the kth layer; A_k and a_k are the maximum deviations from the mean value Z_k determining the height of roughnesses at the boundary of the kth layer; ω_{kx}, ω_{ky}, ω'_{kx}, and ω'_{ky} are the quantities determining the longitudinal and transverse dimensions of irregularities; and ϕ_{kx}, ϕ_{ky}, ϕ'_{kx}, and ϕ'_{ky} are the arbitrarily chosen parameters determining the location of the upper boundary of the kth layer at the OZ axis.

The values of the parameters in (1) are chosen for the boundary of each layer so that the surface shape (Fig. 1) most closely corresponds to the shape of the interface of the corresponding layer in the structure of normal human skin (Fig. 2). The only exception is found in derma layers, where separation into layers is conditioned by the nonuniform distribution of blood vessels over depth and parameters of layer boundaries are chosen in correspondence with typical dimensions of collagen fibers of derma in this connection [23–25] (Fig. 1c). Parameters determining the shape of interfaces for each layer are given in Table 2.

The propagation of radiation in this multilayer medium with strong scattering and absorption of light is modeled by the Monte Carlo method combining the calculation of true paths and the use of statistical weights [26].

The conditions of the introduction of photon packets into the medium and the conditions of their detection are determined by the position of the source and detector relative to the surface of the medium, their diameters, and numerical apertures. The propagation of photon packets in the medium (variation and selection of the direction of their motion owing to scattering, refraction, or reflection at layer boundaries) proceeds in much the same way as in [11–14], with the sole difference being that a decrease in the statistical weight of a photon packet within the framework of our modeling occurs only in its Fresnel reflection or refraction at layer interfaces; i.e., the weight of a photon packet W reaching the region of detection is determined as

$$W = W_0 \prod_{j=1}^{M} \mathbf{R}_j(\alpha_i), \quad (2)$$

where W_0 is the initial weight of a photon and M is the number of acts of interaction of the photon packet with layer boundaries. The function $\mathbf{R}_j(\alpha_i)$ depends on the angle of incidence of a photon packet at an interface

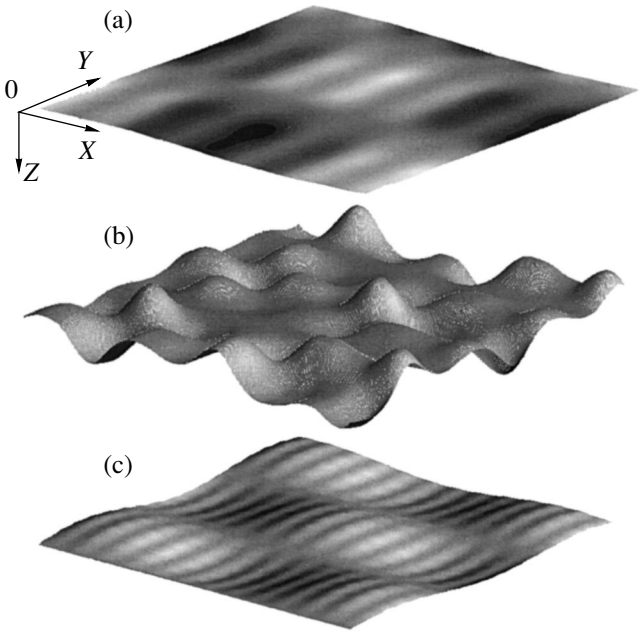

Fig. 1. Boundaries of some layers of the model medium determined by Eq. (1). (a) Surface of the medium, (b) interface between epidermis and derma, and (c) surface determining interfaces between derma layers. The parameters used for modeling layer boundaries are presented in Table 2.

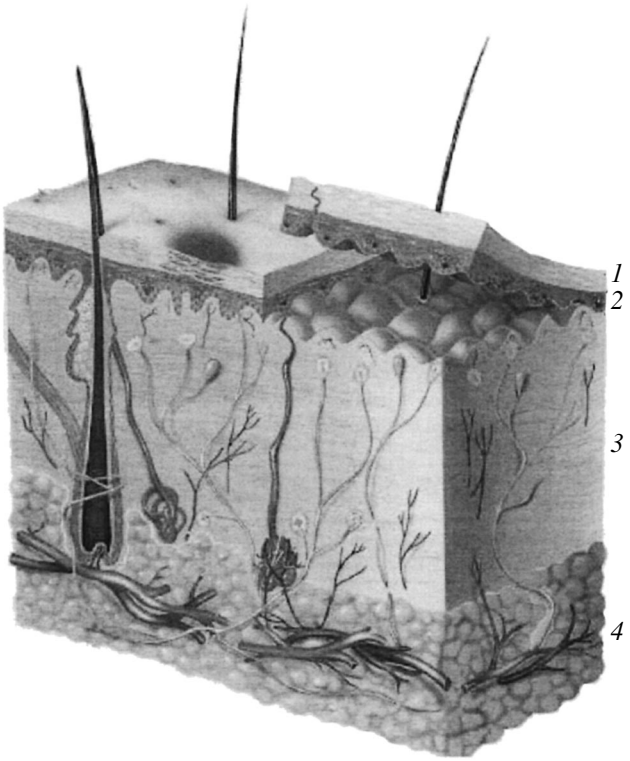

Fig. 2. Structure of the skin. The following layers are usually distinguished on the basis of different cell structure: (*1*) horny layer, (*2*) epidermis, (*3*) derma, and (*4*) hypodermic fat.

between layers α_i and is determined by the Fresnel reflection coefficient $R_{k,m}$. If a photon packet is reflected from a layer boundary, then $\mathbf{R}_j(\alpha_i) = R_{k,m}$, and if it passes through an interface, then $\mathbf{R}_j(\alpha_i) = 1 - R_{k,m}$. According to [27], we have

$$R_{k,m} = \begin{cases} \left(\dfrac{n_m - n_k}{n_m + n_k}\right)^2, & \alpha_i = 0^\circ \\ \dfrac{1}{2}\left[\dfrac{\sin^2(\alpha_i - \alpha_t)}{\sin^2(\alpha_i + \alpha_t)} + \dfrac{\tan^2(\alpha_i - \alpha_t)}{\tan^2(\alpha_i + \alpha_t)}\right] & \\ \qquad 0^\circ < \alpha_i \le \sin^{-1}\left(\dfrac{n_k}{n_m}\right) & \\ 1, \quad \sin^{-1}\left(\dfrac{n_k}{n_m}\right) < \alpha_i \le 90^\circ, & \end{cases} \tag{3}$$

where α_t is the angle of refraction of a photon packet at the interface between the layers *m* and *k*.

The path of a photon packet can be traced until the moment when the statistical weight of the packet becomes smaller than the preset value of 10^{-4}. The history of random walks of photon packets (i.e., paths and variations of weights of photon packets reaching the region of detection) is recorded in the data file. The number of paths modeled depends on the particular conditions of the problem. In this study, we restrict ourselves to the detection of 50000 photon packets.

In addition, the absorption of layers of the medium is taken into account according to the modified Beer–Lambert law [28]. To do this, we recalculate the weight of each individual photon packet in the medium following the history of its motion in accordance with absorption of layers

$$W = \left[W_0 \prod_{j=1}^{M} \mathbf{R}_j(\alpha_i)\right] \exp\left(-\sum_{q=1}^{N} \mu_a(\mathbf{r}_q) l_q\right), \tag{4}$$

where *N* is the number of scattering events experienced by the photon packet in the process of propagation, l_q is the path length between the $(q-1)$th and the qth acts of scattering, and $\mu_a(\mathbf{r})$ is the absorption coefficient of the medium at the point **r**.

Knowing the probability *W* of diffuse reflection within the limits of the detector area for each *f*th photon packet, we define the spatial distribution of detector sensitivity $Q(\mathbf{r})$ inside the medium under study as the probability of diffuse reflection of radiation by the

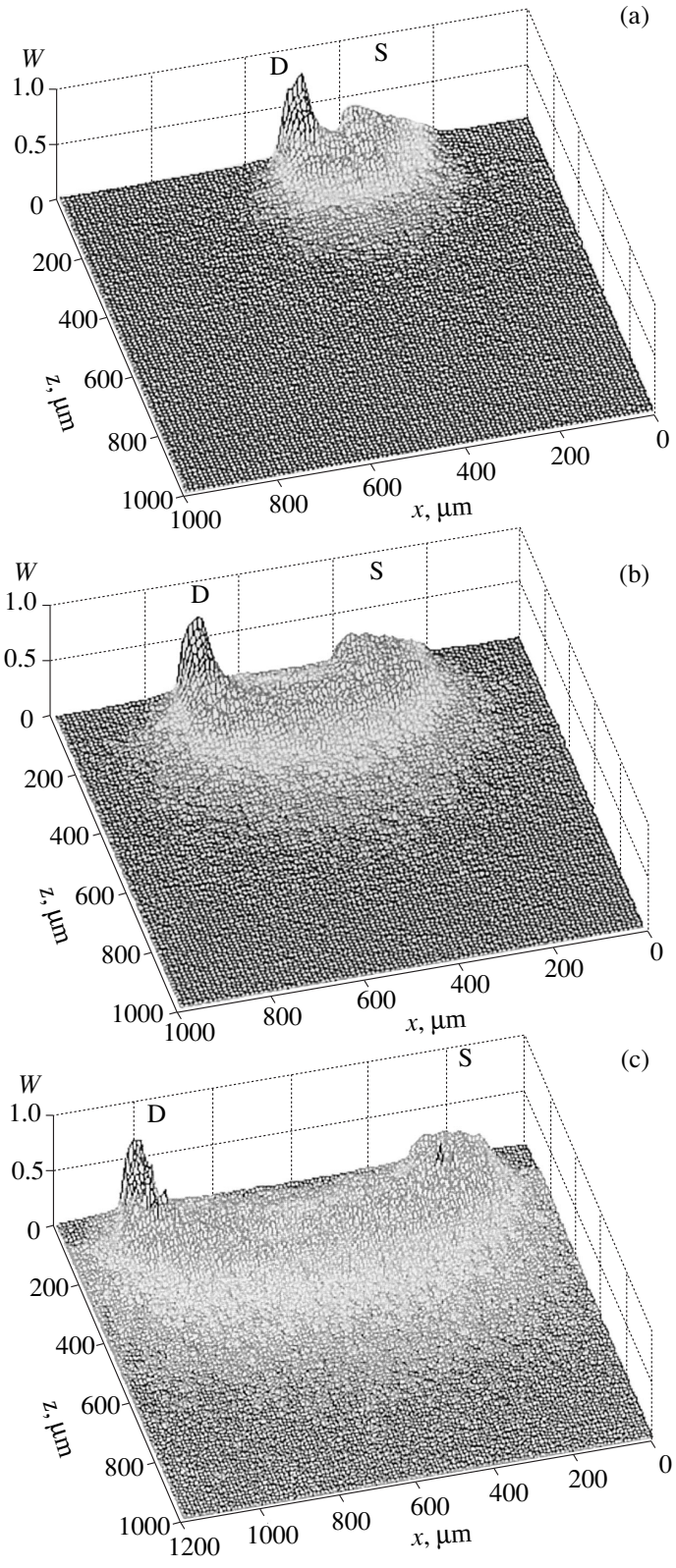

Fig. 3. Spatial distribution of detector sensitivity inside of a multilayer medium with strong scattering and absorption of light. The distance between the source S and the detector D is (a) 250, (b) 400, and (c) 800 μm. Diameters of the source and the detector are 200 and 50 μm, respectively. The numerical apertures of the source and the detector are 0.2. The distance between the source and the detector is counted from center to center. Optical and structural parameters of the medium are presented in Table 1.

point **r** [29, 30]. Then,

$$Q(\mathbf{r}) = -\frac{1}{I}\frac{\partial I}{\partial \mu_a} = \frac{\sum_{f=1}^{N_{ph}} \sum_{j=1}^{N_s} U_j W_f(\mathbf{r})}{\mu_a(\mathbf{r}) \sum_{f=1}^{N_{ph}} W_f}, \quad (5)$$

where I is the radiation intensity detected, N_{ph} is the total number of photon packets detected, N_s is the number of scattering events experienced by the fth photon on its way from the source to the detector, and U is the function determining the residence of the fth photon packet at the point **r**. $U = 1$ each time a photon packet is present at the point **r** and $U = 0$ if a photon bypasses this point.

RESULTS

Results of calculations of the spatial distribution of detector sensitivity inside a multilayer medium with strong scattering and absorption of light for a fiber-optic sensor are presented in Fig. 3. Distances between the source and detector are 250, 400, and 800 μm.

Note that for the smallest (250 μm) distance between the source and detector, the detected signal is localized in the medium so that the major contribution to the signal is produced only by the upper layers of the medium, which do not contain blood (Fig. 3a).

As the distance between the source and detector increases to 400 μm, the domain of detection becomes wider (Fig. 3b). Moreover, the signal detected, in this case, becomes localized in the medium so that the detector sensitivity, with respect to layers located at depths of 100–200 μm, increases along with decreasing detector sensitivity in the upper layers of the medium.

As the distance between the source and detector further increases to 800 μm, the spatial distribution of detector sensitivity varies so that the sensitivity maximum is located at depths of 200–250 μm. However, the contribution of other layers of the medium, including deeper ones, turns out to be significant as well (Fig. 3c).

It is interesting to note that the distribution of the spatial sensitivity immediately in the region of the source and detector is asymmetric, which is explained by the fact that the diameters of the source and detector are different: 200 and 50 μm.

CONCLUSION

The model calculations were carried out with the aim of studying the possibility of spatial localization of the signal detected in the upper (containing capillary loops) skin layers. Modeling of the propagation of optical radiation in a multilayer medium with strong scattering and absorption of light with optical and structural parameters typical of human skin was carried out by the Monte Carlo method combining the calculation of true paths and utilization of statistical weights.

The modeling results give a graphic idea of how the domain of detector sensitivity is localized in a multilayer medium with strong scattering and absorption of light between the regions of radiation incoupling and outcoupling. Analysis of the spatial distribution of detector sensitivity under various conditions of illumination of the medium allows one to optimize the system of incoupling and outcoupling of probing radiation.

It should be noted that the algorithm used in our calculations for step-by-step modeling of scattering and absorption appears to be the most justified and promising because it allows one to save considerable time in the calculation of the distribution of radiation intensity and its various derivatives for new absorption coefficients for unchanged configurations of the source and detector. With allowance made for the fact that physiological processes in biological tissues are characterized, for the most part, only by variations in the absorption coefficient, this approach to the modeling of the propagation of optical radiation appears particularly interesting for physiological and diagnostic biomedical studies of skin and other biological tissues.

The approach presented in the research can be successfully applied to problems of colloid chemistry, materials technology, and so on.

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