Simulation of Speckle Structure of Scattered Light inside Multi-Layered Skin Tissue

V. V. BARUN[1,2], S. K. DICK[1], A. P. IVANOV[2], N. D. ABRAMOVICH[1]

[1]*Belarus State University of Informatics and Radioelectronics, Brovka Str. 6, Minsk 220013, Belarus*
[2]*B.I. Stepanov Institute of Physics, Belarus National Academy of Sciences, Nezavisimosti Pr. 68, Minsk 220072, Belarus*
barun@dragon.bas-net.by

***Abstract* — Calculation results on speckle patterns formed by multiply scattered light in multi-layered biological tissue are given. They are obtained by using the earlier-designed analytical procedure that is based on the engineering methods for solving to the radiative transfer equation and the relations between the coherence theory and the theory of light propagation through a scattering medium. The given examples illustrate the opportunities of the procedure. A special part of the paper is devoted to the inverse problems of biomedical optics that can be posed and solved by the designed procedure. In particular, there is shown the sensitivity of the contrast of the interference patterns formed by multiply scattered light to blood volume fraction and blood oxygen saturation.**

***Index Terms* — multi-layered biological tissues, skin, multiple light scattering, speckle pattern, radiative transfer equation.**

I. INTRODUCTION

The speckle structure of scattered light is currently used in both scientific and practical problems for determining various characteristics of biological tissues, e.g. tissue particle sizes and blood flow rate, for diagnosing different kinds of pathologies and monitoring therapy efficiencies. An analytical approach based on engineering approximations of the solutions to the radiative transfer equation under multiple scattering conditions was firstly proposed in [1] to describe some parameters of such a structure. The features in the formation of the speckles in a semi-infinite biotissue slab are studied and their characteristic dimensions are evaluated as applied to a macroscopically uniform medium. However, real tissues are practically always multi-layered. This paper extends the general approach [1] to the estimation of interference pattern parameters to the case of multi-layered human skin tissue. A procedure for calculating the characteristic dimensions of the speckles and their contrast inside the medium is proposed at wide variations of structural and biophysical tissue parameters. While constructing the procedure, the scattering centers are assumed to be fixed, as in [1]. Under their movement, the speckle contrast would be only lower. In other words, the upper estimate of the contrast will be given here. The below results are valid for mobile scattering particles too under their pulse illumination, when the pulse duration is essentially smaller than the characteristic time of the particle movement. The scatterers are obviously can be considered as "frozen" or immobile in this case.

II. CALCULATION PROCEDURE

The calculation algorithm [1] is based on the known relation [2] between the coherence theory for light field in a scattering medium and the radiative transfer theory (RTE) [3]. This approach enables the speckle structure of the laser light field to be analyzed by the widely used engineering methods of RTE with specific applications to a biological object.

Following to [1], for calculating the coherent component of the light field in multi-layered biotissue, the multi-component method [4] of the RTE will be used below, which is applicable for media with highly forward-extended phase function $p(\beta)$, where β is the scattering angle. Just the same situation would usually occurs for biotissues [5], because large (as compared with light wavelength λ in the visible) collagen fibers of bloodless tissue and blood erythrocytes contribute mainly to light scattered in the forward direction. Due to this reason, one can clearly isolate two components of $p(\beta)$ [4] substantially differing each another by their angular scales. The first one is a sharp peak (diffraction peak) near the direction of the incident beam with a halfwidth on the order of an arc degree, which depends on the sizes of scattering particles. The second component is an essentially more diffuse light with a halfwidth of the order of dozens of arc degrees, which depends mainly on the relative refractive index of scattering particles. The said two components of $p(\beta)$ are usually computed by the Fraunhofer diffraction formulas and the anomalous diffraction or geometrical optics approximation [6]. Using the said separation of phase function into the above angular functions, one can show that there will be three components of multiply forward-scattered light. According to the terminology [1], they are a non-scattered component with angular divergence of the incident light beam (it will be denoted below by subscript $i = 0$), a diffraction one with the divergence on the order of an arc degree ($i = 1$), and a diffuse one with the divergence on the order of dozens of arc degrees ($i = 2$).

The below results are based on the optical skin model [7, 8]. A three-layered tissue is considered consisting of stratum corneum, epidermis, and dermis. For each of

these layers, there are set their geometrical thicknesses (d_1 = 20 μm for stratum corneum and varying d_2 for epidermis), absorption and extinction coefficients μ_a and μ_e, as well as phase functions or a number of their integral parameters [5] required for the computations according to the method of [1, 4]. Coefficients μ_a of epidermis and dermis can vary by the wavelength of an incident beam and volume fractions of their absorbers, namely, of melanin and hemoglobin derivates, respectively. Note that blood is considered as a two-component medium containing the two main hemoglobin derivates, oxy- and deoxyhemoglobin with S being the blood oxygenation degree or blood oxygen saturation.

The calculation procedure is described in detail in [9]. The irradiance at any depth z is assumed to be a sum of the coherent and incoherent components. The first one provides the interference pattern or the speckle structure, but the second one forms the incoherent background. The coherent light, in its own turn, in each skin layer is represented as a sum of non-scattered, diffraction and diffuse component according to [4]. The incoherent background is computed according to the engineering method of [10, 11]. It comprises the known layer-adding method with accounting for multiple re-reflections of light between the skin layers and its surface and a number of engineering approximations of the RTE [3], namely, the small-angle approach to compute light fields in optically this stratum corneum and epidermis and the asymptotic theory to do the same for optically thick dermis. So the fully analytical procedure is designed for the computations of the speckle structure at any tissue depth. It takes only a few seconds for the calculations with using a moderate PC.

III. EXAMPLES OF SIMULATED SPECKLE STRUCTURE INSIDE BIOTISSUE

The speckle structure in dermis will be considered here, because it is of main interest for diagnostic purposes of erythrocytes and whole blood parameters. Fig. 1 illustrates the depth dependence of the irradiance produced by coherent light at various wavelengths of the visible to near IR spectral range. Fig. 1*a* corresponds to varying volume fraction f_m of melanin in epidermis. It plays a role of a spectral filter that does not essentially affect the parameters of the speckle structure in dermis. Epidermis is simply attenuates light penetrating in the depth of the medium, the attenuation being the more noticeable, the more concentration f_m is. Fig. 1*a* corresponds to varying volume fraction C_V of capillaries in dermis. It is rather surprisingly, at the first glance, that the irradiance is practically independent of C_V although the dermis absorption increases substantially in this case. However, there is a physically transparent explanation of this fact. Really, with C_V increasing, the contribution of light scattering by blood erythrocytes to the total dermis extinction grows. But the phase function of erythrocytes is much more forward extended as compared with that of bloodless tissue. Therefore, the coherent component attenuates weaker at large volume fractions of blood. So the competition of two oppositely influencing factors, namely, of the increase of the dermis absorption coefficient with C_V and the more forward extension of the phase function of the medium, gives rise to their approximate balancing, so that irradiance E_2 depends weakly on the capillary volume factor.

Figure 2 shows the spatial distribution of the incoherent background at various wavelengths of the visible to near IR spectral range. This background affects the speckle contrast in the dermis depth. One can see from Fig. 3 that background has a maximum at some depth that depends, mainly, on the wavelength. The maximum moves to the deeper dermis layers with increasing λ. The background irradiance can be two- to threefold larger than the corresponding values at the interface of epidermis and dermis. This is due to multiple re-reflections between deep and surface layers of the medium. The increase in blood volume fraction leads to the reduced background due to the growing tissue absorption coefficient. The increase in the melanin concentration and epidermis thickness playing a role of a spectral filter attenuates the background approximately proportional at all the depths, i.e. the situation similar to that of Fig. 1*a* takes place.

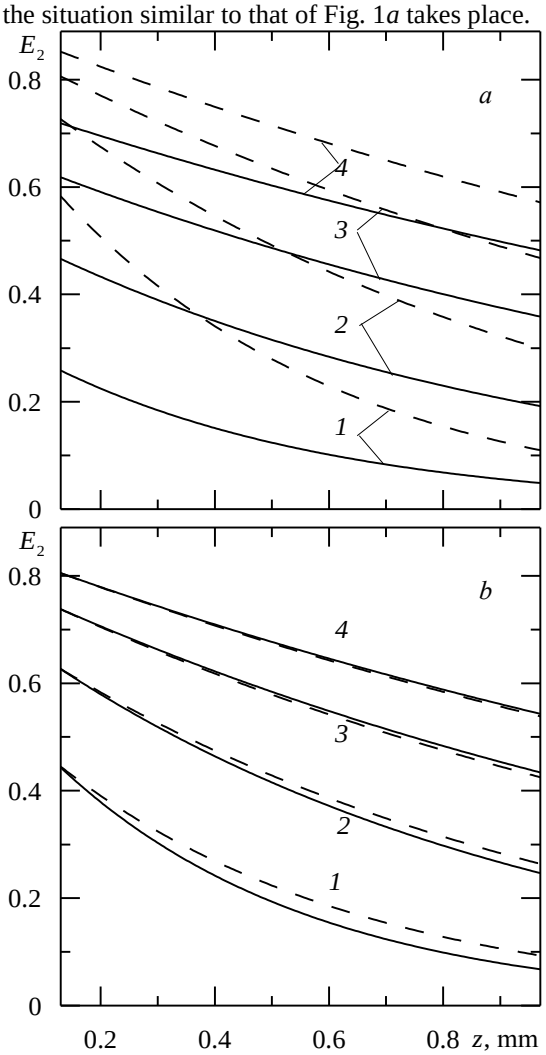

Fig. 1. Depth structure of coherent component in dermis at λ = 500 (*1*), 600 (*2*), 700 (*3*), and 800 nm (*4*), S = 0.75, d_2 = 100 μm, *a* – f_m = 0.04 (dashed curves) and 0.16 (solid), C_V = 0.04; *b* – C_V = 0.02 (dashed) and 0.08 (solid), f_m=0.04

Consider now the interference patterns or speckle structures in the dermis depth. The radial dependence of the scattered light field is the sum [9] of periodic

functions with various spatial frequencies corresponding to the correlation radii (dimensions) for the direct light, the diffraction and diffusion components. Calculations show that the characteristic speckle dimensions are mainly determined by the diffusion component. They depend on the optical tissue properties and, in particular, on the dermis phase function. Fig. 4 illustrates the speckle structure at dermis depth 1 mm with accounting for the incoherent background. The speckle radii can vary from fractions to several μm.

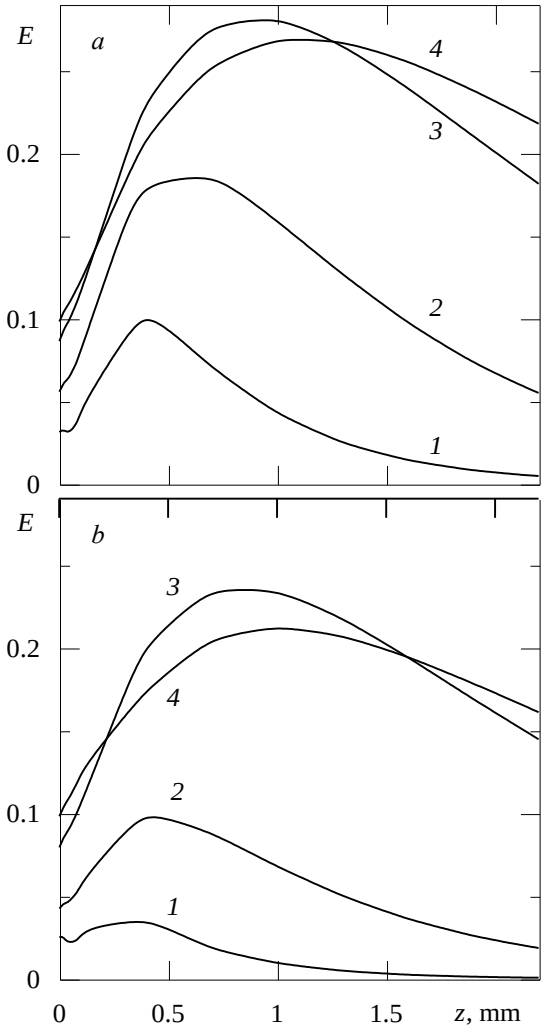

Fig. 2. Depth structure of incoherent background at λ = 500 (curves *1*), 600 (*2*), 700 (*3*), and 800 нм (*4*) for C_V = 0.02 (*a*) and 0.08 (*b*), d_2 = 100 μm, f_m = 0.08, and S = 0.75

Figures 4 give *z*-dependences of characteristic scales $L_{1,2}$ (produced by the diffraction component) and $L_{2,2}$ (diffusion one) of the speckle structure in dermis. The spectral range of the presented results is λ = 500 to 800 nm. Speckle dimensions grow with wavelength increasing. This is directly followed from the analytic formulas of [9]. For phase functions with relatively small forward extension, radius $L_{1,2}$ depends weakly on *z*, whereas $L_{2,2}$ changes several-fold, while going to the deep dermis layers up to depth 2 mm. For phase functions with relatively large forward extension, both $L_{1,2}$, and $L_{2,2}$ decrease with *z* increasing. The speckles are larger for skin irradiation in the near-IR as compared with their dimensions under visible irradiation. The difference in $L_{1,2}$ and $L_{2,2}$ can achieve 2 to 4 times at λ = 800 and500 nm as a function of the depth.

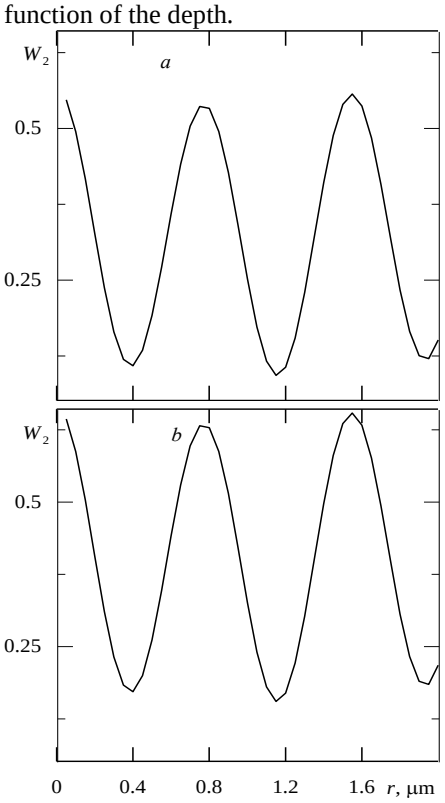

Fig. 3. Radial structure of light field in dermis at depth 1 mm for S = 0.5 (*a*) and 0.97 (*b*), λ = 600 nm, f_m = 0.08, C_V = 0.04, and d_2 = 100 μm

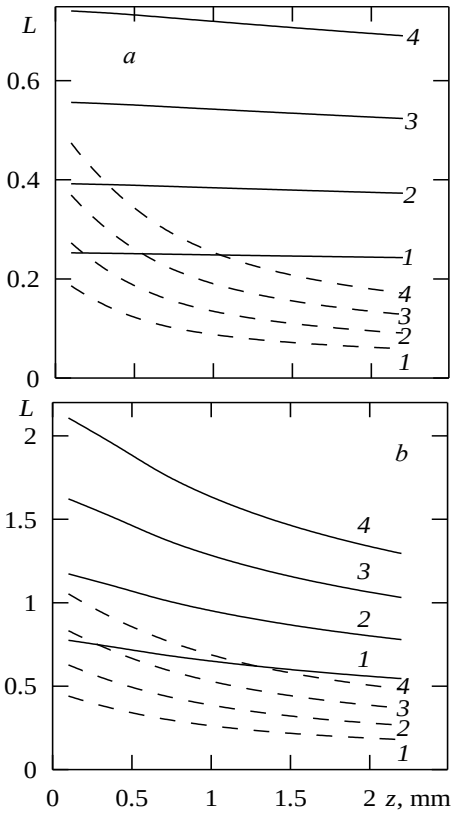

Fig. 4. Depth dependences of characteristic speckle scales $L_{1,2}$ (solid curves) and $L_{2,2}$ (dashed) at relatively small (a) and large (b) phase function extension, λ = 500 (curves *1*), 600 (*2*), 700 (*3*), and 800 nm (*4*), f_m = 0.08, C_V = 0.04, and d_2 = 100 μm

IV. ON THE SOLUTION TO INVERSE PROBLEMS WITH USING SPECKLE MEASUREMENTS

Consider contrast K of the speckle pattern of scattered light field in dermis. This quantity can be really observed in experiments, and its dependence on structural and biophysical tissue parameters can be used for setting and solving the inverse problems of biomedical optics, for example, for retrieving blood volume fraction and blood oxygenation degree from measurement results. Let the contrast be defined similarly to its definition for an interference pattern:

$$K = (W_{max} - W_{min})/ (W_{max} + W_{min}),$$

where W_{max} and W_{min} are, respectively, the maximal and minimal values of the total irradiance produced by both the coherent and incoherent light. These quantities can be directly calculated by the relations of [9].

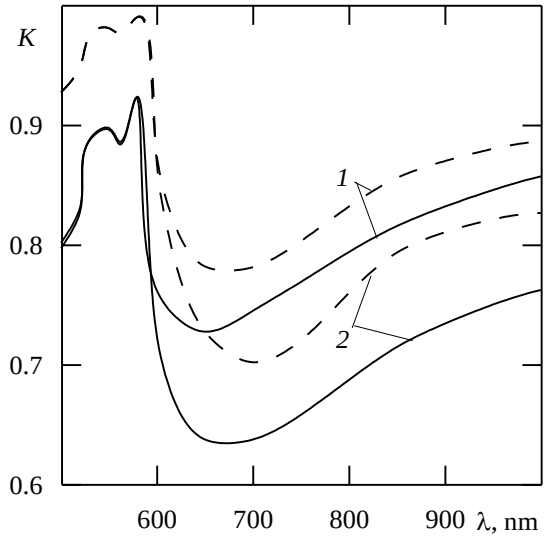

Fig. 5. Spectral dependence of the contrast of the speckle patters at depth 1 mm (curves *1*) and 1 cm (*2*) in dermis for C_V = 0.02 (solid lines) и 0.08 (dashed), f_m = 0.08, S = 0.75, d_2 = 100 μm

Figure 5 illustrates the spectral dependence of the contrast in the visible and near-IR ranges. Note, first, that the K spectrum is similar to that of the blood absorption coefficient $k(\lambda)$. The more the $k(\lambda)$ values are, the larger the contrast. This is due to the fact that the background of the interference pattern is formed by the multiply scattered light. When the absorption increases, the background decreases. One can see from Fig. 5 that there are rather large contrast values exceeding 0.5 up to relatively big deprhs (up to 1 cm). The contrast is minimal in the range of 600 to 700 nm, where the tissue absorbs light relatively weakly. As depth z increases, the minimal K values move to the long-wavelength spectral region. The similar movement is observed for increasing blood volume fraction. All these features can be explained by the comparison of the irradiances created by coherent and incoherent light.

When Fig. 1*b* has been considered, it is noted that mean irradiance E_2 of the coherent light field depends weakly on C_V, and the corresponding explanation has been made. Nevertheless, the contrast of the interference pattern depends essentially on the capillary volume fraction. This can be explained by rather a strong dependence of the incoherent background on C_V. When C_V increases, the contrast increases too. This can be seen by the comparison of the solid and dashed curves in Fig. 5. Changes in K over the red – near-IR spectral range are approximately 0.1. This quantity can be rather simply recorded in experiments. The above said enables the approaches and the methods to be developed in the new biomedical optics scientific direction, namely, for determining the capillary volume fraction by using the characteristics of the spectral pattern in biotissues.

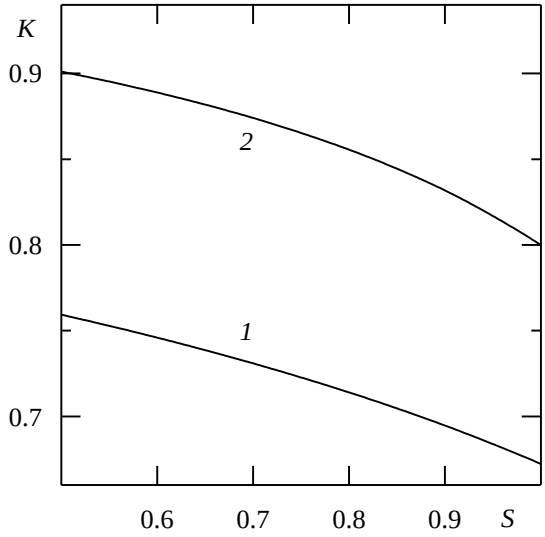

Fig. 6. Dependence of K at z = 5 mm on S for C_V = 0.02 (curve *1*) and 0.08 (*2*), f_m = 0.08, S = 0.75, d_2 = 100 μm, and λ = 600 nm

Figure 6 illustrates once more example of possible posing and solving the inverse problem on retrieving blood oxygen saturation S by using the contrast of the interference pattern in dermis. If one returns to above-shown Fig. 3, where the calculations are given for two values S = 0.5 и 0.97, one can turn attention that the radial structure of the total irradiance produced by coherent and incoherent light in dermis depends on S values at wavelength 600 nm. At other λ of the visible to near IR spectral range, there are weak corresponding dependences. This is due to the sharp changes of the oxyhemoglobin absorption at the said wavelength. One can see from Fig. 6 that there is rather a strong dependence of the contrast on S. Although the data of Fig. 6 are given for z = 0.5 cm only, the calculations have showed that the similar behavior of K can be observed at other depths. Note that the same wavelength of 600 nm was also recommended in [12] for retrieving the blood oxygenation degree by the completely another tool, namely by the spectrophotometry of skin diffuse reflection.

V. CONCLUSION

The designed analytic procedure for evaluating the parameters of speckle patterns produced by multiply scattered light from multi-layered biological tissues has important scientific and practical applications. First, it enables one to calculate the characteristics of the interference pattern at any biotissue depth without using complicated and cumbersome algorithms and computer

codes. This creates the prerequisites for producing new methods and upgrading known ones for investigating light interaction with biological tissues. Second, the above procedure in combination with the optical model of the medium enables the parameters of the speckle structure to be directly related with optically meaningful biotissue characteristics. The vivid illustrations of the above said are Figs. 5 and 6. It follows from these Figures that the real physical quantity, which can be experimentally determined, namely, the contrast of the interference pattern, is sensitive to the blood vessel volume fraction and to the blood oxygen saturation. This direct relation opens wide opportunities for developing new methods of the non-invasive diagnostics of pathologically altered tissues and for optimizing these methods by irradiation wavelengths. Third, the analytical character of the calculation procedure provides the simplicity of its usage by various researchers, who are not specialized in computer engineering and programming. This allows the number of its users to be increased.

It is planned in the future to develop the obtained results, to apply them for mobile scattering particles, and to construct analytical relations for parameters of the speckle patterns and characteristics of moving particles, for example, of erythrocytes, and of a medium, in which the movement occurs. The functional opportunities of the speckle-optical methods are obviously to be essentially expanded in this case.

REFERENCES

[1] A.P. Ivanov and I.L. Katsev, "On the Speckle Structure of Light Field in Disperse Medium Illuminated by a Laser Beam," *Quant. Electron*., vol. 35, pp. 670– 674, 2005.

[2] L.S. Dolin, "Equation for Energy Density of Wave Beam Field," *Izv. Vuzov. Radiophys. Series*, vol. 23, pp. 332 – 338, 1980.

[3] E.P. Zege, A.P. Ivanov, and I.L. Katsev, Зеге, *Image Transfer through a Light Scattering Medium*, Springer – Verlag, 1988.

[4] E.P. Zege, I.L. Katsev, and I.N. Polonsky, "Multicomponent approach to light propagation in clouds and mists," *Appl. Opt.*, vol. 32, pp. 2803 – 2812, 1993.

[5] M.J.C. Van Gemert et al., "Skin optics," *IEEE Trans. Biomed. Eng.*, vol. 46, pp. 1146 – 1154, 1989.

[6] C.F. Boren and D. Huffman, *Absorption and Scattering of Light by Small Particles*, J. Wiley & Sons, 1983.

[7] V.V. Barun and A.P. Ivanov, "Thermal Effects of a Short Light Pulse on Biological Tissues. I. An Optical and Thermal Model," *Biophysics*, vol. 49, pp. 1125 – 1133, 2004.

[8] V.V. Barun, A.P. Ivanov et al., "Absorption Spectra and Light Penetration Depth of Normal and Pathologically Altered Human Skin," *J. Appl. Spectrosc*., vol. 74, pp. 430 – 439, 2007.

[9] N.D. Abramovich, V.V. Barun, S.K. Dick, er al., "Analytical Procedure for Evaluating Contrast of Light Field Scattered by Soft Biological Tissues," in *Proc. 5th Troitsk Conf. on Medical Phys. and Innovations in Medicine*, vol. 1, pp. 212 – 214, 2012.

[10] V.V. Barun and A.P. Ivanov, "Light and Thermal Fields in Multi-Layered Skin Tissue Exposed to Laser Irradiation," *Optics Spectrosc.*, vol. 100, pp. 139 – 147, 2006.

[11] V.V. Barun and A.P. Ivanov, "Light Absorption in Blood during Low-Intensity Laser Irradiation of Skin," *Quantum Electron*., vol. 40, pp. 371 – 376, 2010.

[12] A.P. Ivanov and V.V. Barun, "Light Reflection Spectra as a Diagnostic Tool for the Structural and Biophysical Parameters of Skin," *Optics Spectrosc.*, vol. 104, pp. 300 – 307, 2008.