Variable hyperemia of biological tissue as a noise source in the input optical signal of a medical laser Doppler flowmeter

D. G. Lapitan[1,2] and D. A. Rogatkin[1]

[1]*M. F. Vladimirskiĭ Moscow Regional Scientific Research Clinical Institute, Moscow, Russia*
[2]*e-mail: lapitandenis@mail.ru*

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As applied to the problems of medical laser Doppler flowmetry and based on a modified Kubelka–Munk two-flow model, analytic expressions have been obtained for the radiation power backscattered by biological tissue, taking into account the variable hyperemia of its microvasculature. An estimate is made of the power contribution of the Doppler component of the flow to the overall backscattered-radiation signal recorded by the device, which appears when light is scattered at moving erythrocytes. It is shown that the power contribution of the Doppler component to the overall backscattered radiation flux is no greater than 5% on average. The variable hyperemia that results from various physiological processes causes the radiation flux recorded by the Doppler flowmeter to be amplitude modulated. The power of the amplitude-modulated component can be of the same order of magnitude as, and in certain cases even greater than, the power of the useful Doppler signal, creating noise in the input signal of the device. © *2016 Optical Society of America.*

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INTRODUCTION

The widespread incorporation of modern noninvasive methods of optical (laser) diagnostics in biology and medicine has served as an impetus for a revival of interest in the theoretical description of light propagation in turbid media, which in particular includes human skin. Some of the important tasks in the creation of diagnostic spectrophotometric equipment are to theoretically investigate and model the input radiation, and this provides engineers with a basis for the technical requirements on devices of this class [1]. For example, it is necessary to know the relationship between the signals in each diagnostic channel in order to correctly set the radiation power of the lasers, the amplifications of the photodetectors, etc. in order to develop complex diagnostic devices that combine laser Doppler flowmetry (LDF) and optical tissue oximetry [2]. On the other hand, it is useful in LDF to sum the signal of the backscattered radiation from moving erythrocytes having a Doppler frequency shift and from stationary inhomogeneities of the intracellular biological tissue at the original radiation frequency [3,4]. As shown by Bonner and Nossal [4], these two signals form low-frequency *beats* by summing them at the photodetector, the recording and processing of which provide important medical information concerning blood flow in the microvasculature of the skin and the mucous membranes of the organs. However, the basic model for LDF given in [4] contains a number of essential assumptions and simplifications (low erythrocyte concentration in the diagnostic volume of interest, isotropic illumination of the erythrocytes, etc.). One such crude assumption is that the reference beam scattered at stationary inhomogeneities inside the cellular tissue has a steady-state amplitude.

At the same time, it is well known that the overall amplitude of the backscattered radiation flux strongly and nonlinearly depends on hyperemia in the test volume of biological tissue [5]. Blood is a strong absorber of light in the visible region. This restricts, in particular, the sensitivity of medical optical tissue oximeters [1]. The absorption of light by blood must result in a non-steady-state amplitude of the reference beam in LDF as a consequence of the variable hyperemia of the tissue's microvasculature because of various physiological processes in the organism, and this in turn must be reflected in the input signal of the flowmeter. If the optical field initially imposed on the tissue is written in the form

$$E_0 = A_0 \exp(-i\omega_0 t), \tag{1}$$

where A_0 is the amplitude of the incident field, ω_0 is its circular frequency, t is the time, and the field backscattered from the tissue and experiencing a Doppler frequency shift can be written as

$$E_d = A_d \exp[-i(\omega_0 + \omega_d)t], \tag{2}$$

where A_d is the amplitude of the radiation backscattered from the moving erythrocytes, and ω_d is the Doppler frequency

shift, then, according to the results of [4], the total backscattered field incident on the photodetector surface can be written in the form

$$E_\Sigma = A_S \exp(-i\omega_0 t) + A_d \exp[-i\omega_0 + \omega_d)t], \qquad (3)$$

where A_S is the amplitude of the reference field E_S scattered at stationary inhomogeneities inside the biological tissue. However, if the non-steady-state flux amplitude E_S is taken into account—for example, if one introduces into the treatment the amplitude modulation of the flux in the form

$$E_S = A_S(1 + \kappa \cos \Omega t) \exp(-i\omega_0 t), \qquad (4)$$

where $k \ll 1$ is the modulation depth, and Ω is the modulation frequency, then the total recorded signal is written as

$$E_\Sigma = A_S \exp(-i\omega_0 t) + \frac{\kappa A_S}{2}\{\exp[-i(\omega_0 - \Omega)t] + \exp[-i(\omega_0 + \Omega)t)]\} + A_d \exp[-i(\omega_0 + \omega_d)t]. \qquad (5)$$

Here the factor $kA_S/2$ is the amplitude of the modulated component of the field incident on the photodetector; we denote it as $A_{\text{am}} = kA_S/2$. The exponentials in the second and third terms of Eq. (5) do not differ, especially when Ω and ω_d are of the same order of magnitude. The signals that they express will be called *beats* with the signal described by the first term, and this can result in errors when the blood flow is being computed. The ratio between the amplitudes A_{am} and A_d in Eq. (5) is consequently one of the key factors in interpreting LDF data [2].

The goal of this paper was to estimate the possible relationship between a signal with a Doppler frequency shift and the total backscattered radiation signal in typical diagnostic problems. To achieve the formulated goal, an analytical model of the backscattered optical radiation in the case of two-layer biological tissue with variable hyperemia was constructed, based on a modified two-flow Kubelka–Munk model [6]. It should be stated that simple analytical models are of special interest in tasks of medical diagnosis, because they make it easy to study how the parameters of the medium affect the behavior of the scattered radiation field. In the optics of biological tissues, wide use in this area is made of simple methods for solving transfer equations, such as the two-flow Kubelka–Munk model, and the three-, four-, and seven-flow models [7]. However, they have been considered until recently to have low accuracy. Nevertheless, it was shown comparatively recently that the classical Kubelka–Munk model could be used to obtain accurate values of the radiation fluxes at the boundary of a medium by slightly modifying the coefficients of the original equations [6]. Moreover, this is the only approach of all those currently known that explicitly introduces an additional optophysical parameter of the radiation-propagation medium in the form of the scatterer density (μ_ρ, cm^{-1}), and this is indispensable for computing the intensity of the Doppler signal. This determined the choice of the given model for purposes of solving the problem formulated in this paper.

MODEL OF A TWO-LAYER MEDIUM

We shall represent the main biological tissue to be studied (human skin) as two layers that differ in their optical properties: the epidermis and all the rest of the tissue (the dermis, the subdermal fatty tissue, etc.). Such differentiation is based on the fact that the epidermis contains no blood vessels—i.e., this layer contains no blood, and all the rest of the tissue is filled with it in one degree or other. Let the first layer have finite thickness H_1, while we shall consider the second layer (the rest of the tissue) to be optically semi-infinite ($H_2 \to \infty$), since in most cases light does not pass through it.

The first layer on the left is illuminated by the initial probing radiation flux F_0 (Fig. 1). We shall designate the backscattered flux recorded by a device as F_{BS}.

In Fig. 1, the light fluxes transmitted through the first layer are designated by solid arrows, while the backscattered radiation fluxes are designated by dashed arrows. The optical properties of the layers in this model are determined by generalized attenuation and backscattering transport coefficients [6]. Let β_{11} and β_{12} be the corresponding transport coefficients that determine the optical properties of the first layer, and let β_{21} and β_{22} be those that determine the optical properties of the second layer, which will be different when the two layers possess different levels of hyperemia. We shall designate the capillary hyperemia level of the second layer as V_b. In general, V_b is a distinctive integral parameter that characterizes not only the volume of mixed blood circulating in the observation zone but also the degree to which the surface capillaries have opened [8]. However, for the purposes of solving the formulated problem, it can be regarded as the relative fraction of blood ($V_b = 0$–1 rel. units) or, more precisely, of hemoglobin in the overall volume of the tissues of the second layer of the model medium under consideration. Then $V_b = 1$ will in fact denote only blood, while $V_b = 0$ denotes skin with no blood.

The Gurevich relationships given in [6] make it possible to immediately write the transmitted and backscattered light fluxes in explicit form for one layer of turbid medium. Using these relationships, an expression was obtained that describes the backscattered optical radiation power in the case of a two-layer medium with multiple light scattering:

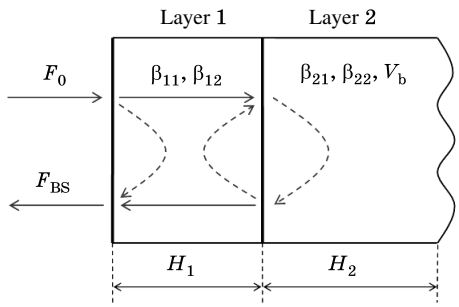

FIG. 1. Light fluxes transmitted forward and back inside a two-layer biological tissue.

$$F_{\mathrm{BS}} = F_0\left\{\frac{P_1(1-\exp(-2L_1H_1))}{1-P_1^2\,\exp(-2L_1H_1)} + \frac{P_2(1-P_1^2)\exp(-2L_1H_1)}{[1-P_1^2\,\exp(-2L_1H_1)][1-P_1P_2-P_1(P_1-P_2)\exp(-2L_1H_1)]}\right\}, \tag{6}$$

where

$$L_i = \sqrt{\beta_{i1}^2-\beta_{i2}^2}, \quad P_i = (\beta_{i1}-L_i)/\beta_{i2}, \quad \beta_{i1} = \omega_i\frac{\mu_{\mathrm{ai}}-\mu_{\rho i}\,\ln(1-R_i)+\mu_{\rho i}\,\ln\left[1-\omega_i+\sqrt{\omega_i^2-R_i^2\,\exp(-2\mu_{\mathrm{ai}}/\mu_{\rho i})}\right]}{\sqrt{\omega_i^2-R_i^2\,\exp(-2\mu_{\mathrm{ai}}/\mu_{\rho i})}},$$

$$\beta_{i2} = R_i\,\exp(-\mu_{\mathrm{ai}}/\mu_{\rho i})\frac{\mu_{\mathrm{ai}}-\mu_{\rho i}\,\ln(1-R_i)+\mu_{\rho i}\,\ln[1-\omega_i+\sqrt{\omega_i^2-R_i^2\,\exp(-2\mu_{\mathrm{ai}}/\mu_{\rho i})}]}{\sqrt{\omega_i^2-R_i^2\,\exp(-2\mu_{\mathrm{ai}}/\mu_{\rho i})}},$$

$$\omega_i = \frac{1-(1\text{–}2R_i)\exp(-2\mu_{\mathrm{ai}}/\mu_{\rho i})}{2}, \tag{7}$$

and i is the layer number ($i = 1, 2$). As can be seen, coefficients β_{i1} and β_{i2} are determined explicitly via the actual optophysical properties of the medium: the absorption coefficient μ_{ai} corresponding to the absorption coefficient in the general transfer equation [7], the mean density $\mu_{\rho i}$ of inhomogeneities inside the medium, and the Fresnel reflectance R_i at the boundary of the inhomogeneities inside the medium. In general, the absorption coefficient and the Fresnel reflectance are functions of radiation wavelength λ. In combination, $\mu_{\rho i}$ and R_i determine the scattering coefficient μ_{si} of each layer. In the case of single scattering, for example [9],

$$\mu_{\mathrm{si}} = -\mu_{\rho i}\,\ln(1-R_i). \tag{8}$$

When numerical modeling was used, the absorption coefficient of the first layer was taken from [10] as a function of wavelength in the form

$$\mu_{a1} = 27\,\exp(-0.006\lambda).$$

analogously to the absorption coefficient, $\mu_{\rho 2}$ in the case of the second layer is defined as the sum

$$\mu_{\rho 2} = \mu_{\rho t}(1-V_b) + \mu_{\rho b}V_b, \tag{10}$$

where $\mu_{\rho t}$ and $\mu_{\rho b}$ are the mean densities of scattering inhomogeneities in tissue with no blood and with blood, respectively.

The single-scattering approximation was used to compute the intensity of the Doppler component of the radiation. The same approximation was used in Bonner and Nossal's classical paper [4]. The secondary scattering of the reverse flux in the medium after any first reflection of the forward flux from any inhomogeneities was neglected in this approximation. Using the expression for the power of the transmitted and backscattered fluxes as applied to one layer of turbid medium in the single-scattering approximation [9], the authors of this paper obtained an exact analytical solution to express the power of backscattered radiation in the case of a two-layer medium with single scattering,

$$F_{\mathrm{BS,single}} = F_0R\left[\frac{\exp(-\mu_{a1}/\mu_{\rho 1})(1-Y_1^{\mu_{\rho 1}H_1})}{1-Y_1} + \frac{\exp(-\mu_{a2}/\mu_{\rho 2})(1-Y_1)Y_1^{\mu_{\rho 1}H_1}(1-R)^{\mu_{\rho 1H_1}}}{(1-Y_1)(1-Y_2)-R^2\,\exp(-\mu_{a1}/\mu_{\rho 1})\exp(-\mu_{a2}/\mu_{\rho 2})(1-Y_1^{\mu_{\rho 1H_1}})}\right], \tag{11}$$

The absorption coefficient of the second layer was defined as the sum [8]

$$\mu_{a2} = \mu_{a1}(1-V_b) + [\mu_{a\mathrm{HBO2}}\mathrm{S}_t\mathrm{O}_2 + \mu_{a\mathrm{HB}}(1-\mathrm{S}_t\mathrm{O}_2)]V_b, \tag{9}$$

where $\mu_{a\mathrm{HBO2}}$ is the absorption coefficient of the blood's oxyhemoglobin, $\mu_{a\mathrm{HB}}$ is the absorption coefficient of the blood's deoxyhemoglobin, and $\mathrm{S}_t\mathrm{O}_2 = 0$–1 rel. units is the mean tissue saturation of oxyhemoglobin [8] (a typical value of $\mathrm{S}_t\mathrm{O}_2 = 0.7$ was used below in the calculations for mixed arterial–venous blood).

Since the first layer contains no blood, the mean inhomogeneity density in it corresponds to some inhomogeneity density $\mu_{\rho t}$ of biological tissue with no blood, whereas, where $Y_i = (1-R_i)\exp(-2\mu_{\mathrm{ai}}/\mu_{\rho i})$.

To compute the power of the Doppler component F_d from the total backscattered flux $F_{\mathrm{BS,single}}$, the difference in the $F_{\mathrm{BS,single}}$ values when $V_b \neq 0$ and $V_b = 0$ was used; i.e.,

$$F_d = F_{\mathrm{BS,single}}(\mu_{\rho b} \neq 0) - F_{\mathrm{BS,single}}(\mu_{\rho b} = 0). \tag{12}$$

RESULTS OF THEORETICAL MODELING

To check the correctness of the model at the first stage, the spectra of the power of the backscattered flux F_{BS} were computed at different wavelengths in the range $\lambda = 440$–950 nm (Fig. 2), using Eqs. (6)–(10) and with different theoretical hyperemia levels of the second layer of biological tissue

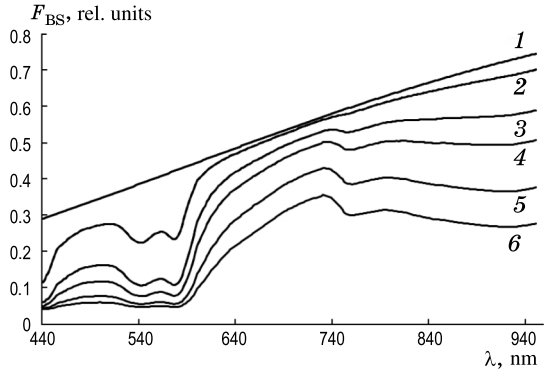

FIG. 2. Spectra of back-scattered radiation F_{BS} at different hyperemia levels of the medium $V_b = 0$ (*1*), 0.02 (*2*), 0.1 (*3*), 0.2 (*4*), 0.5 (*5*), and 1 (*6*).

($V_b = 0$, 0.02, 0.1, 0.2, 0.5, 1). It was assumed in the computations that $F_0 = 1$ and $H_1 = 200$ μm, while inhomogeneity densities of the tissue and the blood close to those typical of human skin were taken from [6]: $\mu_{\rho t} = \mu_{\rho b} = 100$ cm^{-1}.

It can be seen from Fig. 2 that the radiation flux F_{BS} weakens as V_b increases, especially in the range $\lambda = 440$–600 nm. This dependence entirely agrees with the actual situation, since blood is a good absorber of optical radiation. The minima at wavelengths in the 540- and 576-nm regions correspond to the absorption peaks of oxyhemoglobin, while those at a wavelength in the 757-nm region correspond to deoxyhemoglobin in the absorption spectrum of hemolyzed blood [11]. To check the validity of the spectra constructed by means of the spectrometer of the LAKK-M diagnostic system (NPP Lazma, RF) [12], an experiment was run in the course of which similar experimental reflection spectra of white light from the skin and mucous membrane of the oral cavity of a healthy person were obtained (Fig. 3). As can be seen, these curves are very similar in shape and amplitude to the theoretically computed spectra (Fig. 2), and this supports the objectivity and reliability of the constructed model.

The normal hyperemia values of the surface layers of human tissue lie within the limits 0.05–0.2 rel. units [13,14]. As can be seen from Fig. 2, the recorded flux varies appreciably within these limits—by almost a factor of 2 on certain sections of the spectrum. For instance, when an occlusion test is carried out, V_b sharply increases by a sizable factor at the instant of postocclusion reactive hyperemia [15]; this results in a sharp falloff of F_{BS}, and hence failures may occur in the operation of the device. The hyperemia level of the second layer is thus a key parameter among all the other properties of two-layer biological tissue.

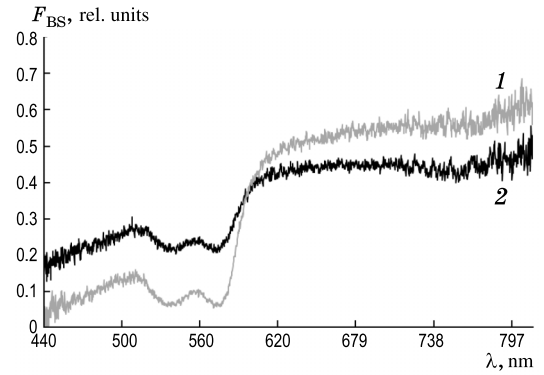

FIG. 3. Experimental white-light reflection spectra from the mucous membrane (*1*, $V_b \approx 0.25$) and the skin (*2*, $V_b \approx 0.05$) of a healthy person.

The concept of amplitude modulation of the backscattered radiation—F_{am}—was introduced in order to estimate the given power variations of the recorded flux when fluctuations occur in the hyperemia level. It is defined as the modulus of the difference between F_{BS} for two different values of V_{b1} and V_{b2}:

$$F_{am}(\Delta V_b) = |F_{BS}(V_{b1}) - F_{BS}(V_{b2})|. \quad (13)$$

For example, the values of $F_{am}(\Delta V_b)$ at the wavelength of 810 nm at which the LAKK-02 laser Doppler flowmeter (NPP Lazma, RF) operates were computed from Eq. (13) in the case of different hyperemia increments ΔV_b in the range 0.02–0.1 rel. units, as well as for different $\mu_{\rho t}$ and $\mu_{\rho b}$. These results are summarized in Table 1.

The power of the Doppler component of flux F_d backscattered from the erythrocytes was computed from Eqs. (11) and (12) as the increment in the summed recorded flux $F_{BS,single}$ caused by the increase of the mean density of erythrocytes $\mu_{\rho b}$ in the second layer of tissue, which was varied in the range from 100 to 1000 cm^{-1} (100, 500, 1000 cm^{-1}). The intensities of the Doppler component at each hyperemia level ($\Delta V_b = 0$, 0.1, 0.2, 0.3, 0.4, and 0.5) were then calculated, and graphs were constructed of the $F_d(\Delta V_b)$ dependence (Fig. 4). As can be seen, the $F_d(\Delta V_b)$ dependence has a linear character for small values of $\mu_{\rho t}$ [Fig. 4(a)] and becomes more nonlinear (close to quadratic) as $\mu_{\rho t}$ increases [Figs. 4(b) and 4(c)]. Thus, the more scatterers there are in a medium, the greater the nonlinearity they introduce into the recorded signal.

Figure 5 shows how the ratio of the power of the Doppler component to the total power of backscattered radiation depends on the hyperemia of the medium when $\mu_{\rho t} = 300$ cm^{-1} and $\mu_{\rho b} = 200$ cm^{-1} (these values, computed from the materials of [16], are typical of normal human skin). As can be seen, the contribution of the Doppler component to the total power lies in the range 1%–5% for the normal physiological conditions $0.05 < V_b < 0.3$. It can also be seen that this dependence on V_b has a quadratic character, and this allowed it to be approximated by a second-degree polynomial (the dashed curve in Fig. 5),

TABLE 1. Amplitude Modulation $F_{am}(\Delta V_b)$ of the Backscattered Flux for Different Values of $\mu_{\rho t}$, $\mu_{\rho b}$, and ΔV_b ($\lambda = 810$ nm)

	F_{am}, rel. units		
$\mu_{\rho t} = \mu_{\rho b}$, cm^{-1}	$F_{am}(\Delta V_b = 0.1)$	$F_{am}(\Delta V_b = 0.05)$	$F_{am}(\Delta V_b = 0.02)$
100	0.056	0.028	0.011
500	0.037	0.018	0.007
1000	0.028	0.014	0.006

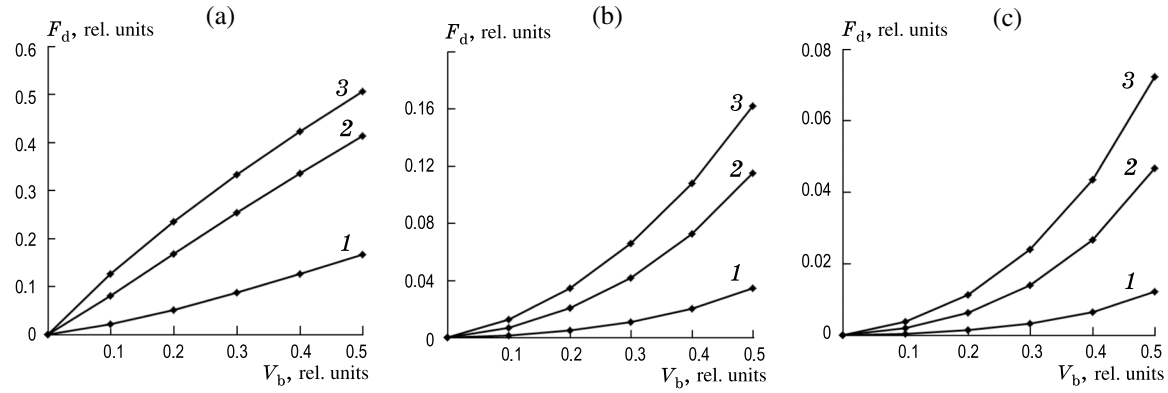

FIG. 4. Intensity F_d of the Doppler component versus hyperemia V_b with (a) $\mu_{\rho t} = 100$, (b) 500, and (c) 1000 cm^{-1}, with $\mu_{\rho b} = 100$ (*1*), 500 (*2*), and 1000 (*3*).

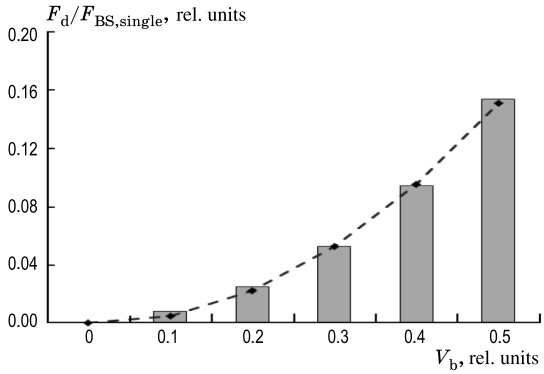

FIG. 5. Ratio of the power F_d of the Doppler component of radiation to the total radiation intensity $F_{BS,single}$ vs. the hyperemia level V_b of the medium with $\mu_{\rho t} = 300$ cm^{-1} and $\mu_{\rho b} = 200$ cm^{-1}. The dashed curve denotes the approximating curve.

$$F_d/F_{BS,single}(V_b) = 0.636V_b^2 - 0.016V_b. \qquad (14)$$

Additionally, based on the data of Table 1, the possible ratio of the Doppler component to the amplitude modulation of the backscattered flux was estimated for different values of $\mu_{\rho t}$, $\mu_{\rho b}$, and ΔV_b. These results are shown in Fig. 6. The ratio F_d/F_{am} corresponds to the SNR in LDF devices and, as can be seen from Fig. 6, for hyperemia of 0.02–0.1 rel. units, its value lies in the range 0.25–3.5, depending on the density of scatterers inside the medium. That is, the SNR in LDF, for example, can even fall below unity under normal physiological conditions of the hyperemia of the skin.

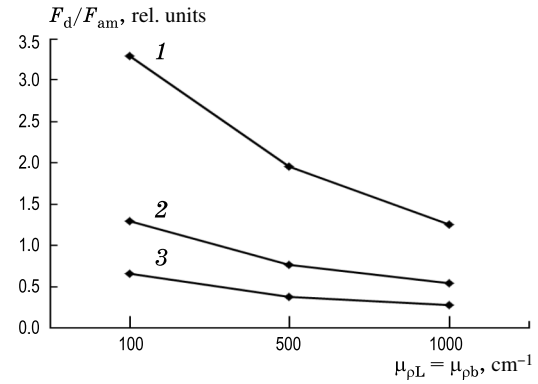

FIG. 6. Ratio of the power F_d of the Doppler component of radiation to the amplitude modulation of the backscattered flux F_{am} vs. the scatterer density for different hyperemia increments $\Delta V_b = 0.02$ (*1*), 0.05 (*2*), and 0.1 (*3*).

Now, using the relationship $F \sim A^2$, where F is the power of the incident flux and A is the field amplitude, it is possible to estimate the amplitude ratio of the Doppler component of the field at the photodetector to the amplitude of the modulated component of the field, which has the form

$$A_d/A_{am} = 2A_d/\kappa A_S \sim \sqrt{F_d/F_{am}}. \qquad (15)$$

The given amplitude ratio can thus lie in the range of values 0.5–1.9.

COMPARING THE MODELING RESULTS WITH OBSERVATION

Calculations to determine the contribution of the Doppler component to the total backscattered radiation intensity but based on a numerical Monte Carlo method were recently given in [16]. The author of that paper also used a two-layer model of a medium in which the first layer has no blood. The introduction of a packet of (10^8–10^9) photons into the medium was modeled in that paper, and their propagation in this medium was calculated. As a result of a numerical experiment with the given optical characteristics of the epidermis ($\mu_{\rho 1} = 300$ cm^{-1}, $\mu_{\rho 2} = 200$ cm^{-1}), an approximate dependence was obtained for the volume hyperemia of the microvasculature on the measured experimental ratio of the intensity of the Doppler component to the total intensity of the recorded radiation,

$$V_b^* = 104.618(I_D/I)^2 - 21.3(I_D/I) + 1.513, \qquad (16)$$

where $V_b^* = V_b/0.14$ is the relative level of the volume hyperemia (with respect to the normal hyperemia of the tissue of 0.14 rel. units, [16]), I_D is the intensity of the Doppler component corresponding to F_d from Eq. (12), and I is the total intensity of backscattered radiation corresponding to $F_{BS,single}$ from Eq. (11).

The solution of Eq. (16) for I_D/I can be written in the form

$$(I_D/I)_{1.2}(V_b) = \frac{21.3 \pm \sqrt{2989.086V_b - 179.458}}{209.236}. \quad (17)$$

When $V_b = 0$, Eq. (17) has no real solutions. Let us substitute the values $V_b = 0.1, \ldots, 1$ into Eqs. (14) and (17) and compare the resulting solutions from the model proposed in this model and from the Starukhin model [16]. The graphs of the $F_d/F_{\mathrm{BS,single}}(V_b)$ and $I_D/I(V_b)$ dependences from these two models are shown in Fig. 7.

As can be seen from Fig. 7, these results of the calculation of the ratio of the power of the Doppler component to the total power of the backscattered radiation as a function of the hyperemia level of the medium qualitatively agree with the model [16] but indicate that the quantitative agreement of the results is not correct in this case, since different approaches are used in modeling light propagation in a turbid medium. This paper uses the one-dimensional two-flow Kubelka–Munk model, whereas [16] uses the method of numerical Monte Carlo modeling, based on the implementation of probability processes. It seems doubtful that Eq. (16) is valid, since the equation has a nonzero solution in the absence of a Doppler signal ($I_D = 0$) with hyperemia V_b, and this does not physically correspond to the actual situation. Moreover, Eq. (16) was obtained for a wavelength of 633 nm, while Eq. (14) in this paper was obtained for 810 nm. Therefore, additional studies are required to substantiate the choice of the parameters in one way or another when the results of the calculations are compared.

Nevertheless, the results of the two independent approaches were in qualitative agreement, from which it can be concluded that there may be added noise in the non-steady-state reference signal recorded in the LDF that results from the absorption of light by blood, along with hyperemia fluctuations in the biological tissue as a consequence of various non-steady-state physiological processes in the microcirculatory system. The SNR in the LDF can fall below unity in this case. Based on these results, it can be assumed in accordance with Eq. (5) that the input optical signal of the LDF devices contains much amplitude-modulated noise. This can cause errors in the computation of the blood flow in LDF, especially when various blood-flow fluctuations are computed (microhemodynamics rhythms), which are often used as an additional diagnostic parameter when the microcirculatory system of blood is studied in the clinic. The problem of amplitude-modulated noise in LDF thus needs to be studied in detail.

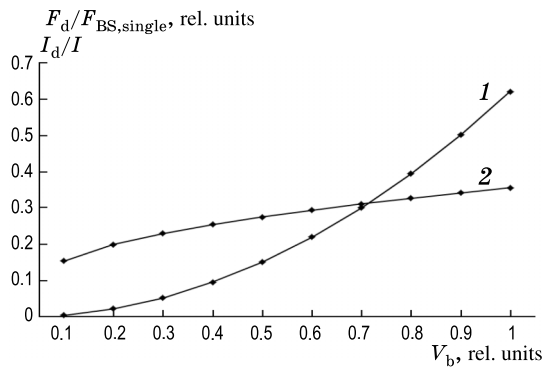

FIG. 7. Dependence on the hyperemia level V_b of the intensity ratio of the Doppler component to the total radiation intensity, $F_d/F_{\mathrm{BS,single}}$, using the model of authors of this paper (*1*) and I_D/I using the Starukhin model [16] (*2*).

CONCLUSION

The process of light propagation in two-layer biological tissue with variable hyperemia has been theoretically modeled in this paper, as applied to problems of modeling the input optical signal of a medical optical laser Doppler flowmeter. The influence of variable hyperemia on the recorded optical signal has been investigated. The detected wavelength dependence of the backscattered light flux at different hyperemia levels of the medium confirms that a small change of the volume hyperemia in the skin can cause amplitude modulation of the recorded backscattered radiation in the LDF, and this is not taken into account in the classical approach to LDF [4]. As shown in this paper, the ratio of the power of the Doppler component of the radiation to the total power of the backscattered radiation can vary in the range 1%–5% with normal hyperemia of the medium. With normal hyperemia, the ratio of the Doppler component to the amplitude modulation of the backscattered flux, which corresponds to the SNR in the LDF, lies in the range 0.5–3.5, depending on the mean hyperemia level and the density of scatterers inside the medium. This indicates that, as a consequence of amplitude modulation of the recorded reference flux in the LDF, the noise can in certain cases be equal to or even greater than the power of the useful Doppler signal.

1. A. V. Dunaev, E. A. Zherebtsov, D. A. Rogatkin, N. A. Stewart, and E. U. Sokolovski, "Substantiation of medical and technical requirements for noninvasive spectrophotometric diagnostic devices," J. Biomed. Opt. **18**(10), 107009 (2013).
2. D. G. Lapitan and D. A. Rogatkin, "Evaluation of the Doppler component contribution in the total backscattered flux for noninvasive medical spectroscopy," Proc. SPIE **9129**, 91292X (2014).
3. V. Rajan, B. Varghese, T. G. Leeuwen, and W. Steenbergen, "Review of methodological developments in laser Doppler flowmetry," Lasers Med. Sci. **24**, 269–283 (2009).
4. R. F. Bonner and R. Nossal, "Model for laser Doppler measurements of blood flow in tissue," Appl. Opt. **20**, 2097–2107 (1981).
5. V. V. Tuchin, *Handbook of Optical Biomedical Diagnostics* (SPIE Press, Bellingham, Wash., 2002).
6. D. A. Rogatkin, "Features in the determination of the optical properties of turbid biological tissues and media in calculational problems of medical noninvasive spectrophotometry," Med. Tekh. (Moscow) (2), 10–16 (2007).
7. A. Ishimaru, *Wave Propagation and Scattering in Random Media: Multiple Scattering, Turbulence, Rough Surfaces, and Remote Sensing* (Academic Press, New York, 1978; Mir, Moscow, 1981), vol. 1.
8. D. A. Rogatkin, "Physical principles of optical oximetry," Med. Fiz. (2), 97–114 (2012).
9. M. A. Dmitriev, M. V. Feducova, and D. A. Rogatkin, "On one simple backscattering task of the general light scattering theory," Proc. SPIE **5475**, 115–122 (2004).

10. I. S. Saidi, "Transcutaneous optical measurement of hyperbilirubinemia in neonates," Doctor of Philosophy dissertation (Rice University, Houston, 1992).
11. E. Kreps, *Oxyhemometry* (Meditsina, Moscow, 1978).
12. D. A. Rogatkin, L. G. Lapaeva, E. N. Petritskaya, V. V. Sidorov, and V. I. Shumskiy, "Multifunctional laser noninvasive spectroscopic system for medical diagnostics and metrological provisions for that," Proc. SPIE **7368**, 73681Y (2009).
13. I. V. Meglinski and S. J. Matcher, "Quantitative assessment of skin layers absorption and skin reflectance spectra simulation in the visible and near-infrared spectral regions," Physiol. Meas. **23**(4), 741–753 (2002).
14. A. V. Dunaev, V. V. Sidorov, N. A. Stewart, S. G. Sokolovski, and E. U. Rafailov, "Laser reflectance oximetry and Doppler flowmetry in assessment of complex physiological parameters of cutaneous blood microcirculation," Proc. SPIE **8572**, 857205 (2013).
15. D. Rogatkin, V. Shumskiy, S. Tereshenko, and P. Polyakov, "Laser-based non-invasive spectrophotometry—an overview of possible medical application," Photonics Lasers Med. **2**(3), 225–240 (2013).
16. P. Yu. Starukhin and Yu. V. Klinaev, "Modeling the Doppler broadening of the spectrum of scattered laser radiation and the diagnostics of blood flow in biological tissues," Vest. Sarat. Gos. Tekh. Univ. **4**(1), 28–35 (2010).