

Monte Carlo study of skin optical clearing to enhance light penetration in the tissue: implications for photodynamic therapy of acne vulgaris

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ABSTRACT

Result of Monte Carlo simulations of skin optical clearing is presented. The model calculations were carried out with the aim of studying of spectral response of skin under immersion liquids action and calculation of enhancement of light penetration depth. In summary, we have shown that: 1) application of glucose, propylene glycol and glycerol produced significant decrease of light scattering in different skin layers; 2) maximal clearing effect will be obtained in case of optical clearing of skin dermis, however, absorbed light fraction in skin dermis changed insignificantly, independently on clearing agent and place it administration; 3) in contrast to it, the light absorbed fraction in skin adipose layer increased significantly in case of optical clearing of skin dermis. It is very important because it can be used for development of optical methods of obesity treatment; 4) optical clearing of superficial skin layers can be used for decreasing of power of light radiation used for treatment of acne vulgaris.

Keywords: optical clearing, skin, optical properties, glucose, glycerol, propylene glycol, acne vulgaris, Monte Carlo

1. INTRODUCTION

Recent technological advancements in the photonics industry have led to a resurgence of interest in optical imaging technologies and real progress toward the development of non-invasive clinical functional imaging systems. Over the last decade, non-invasive or minimally invasive spectroscopy and imaging techniques have witnessed widespread exciting applications in biomedical diagnostics, for example, optical coherence tomography (OCT)^{1,2}, visible and near-infrared elastic-scattering spectroscopy^{3,4}, fluorescent^{1,3,5} and polarization spectroscopy^{6,7}. Spectroscopic techniques are capable of deep-imaging of tissues that could provide information of blood oxygenation⁸ and detect cutaneous, brain and breast tumors⁹, whereas confocal microscopy¹⁰, OCT^{2,11} and multi-photon excitation imaging^{10,12} have been used to show cellular and sub-cellular details of superficial living tissues. Spectroscopic and OCT techniques are applicable for blood glucose monitoring with diabetic patients¹³⁻¹⁵. Besides diagnostic applications, optical methods are widely used in modern medicine, for example, for photodynamic therapy¹⁶⁻¹⁸ and for laser surgery of different diseases^{19,20}. Interest in using optical methods for physiological-condition monitoring and cancer diagnostics and therapies has been increased due to their simplicity, safety, low cost, contrast and resolution features in contrast to conventional X-ray computed tomography and ultrasound imaging⁹.

The main limitations of the majority of the imaging techniques, including OCT and near-infrared (NIR) spectroscopy deal with the strong light scattering in superficial tissues^{9,21-23} which cause decrease of spatial resolution, low contrast, and small penetration depth. Solution of the problem, i.e. reducing light scattering, and thus improving image quality and precision of spectroscopic information, can be connected with control of tissue optical properties. It is well-known that the major source of scattering in tissues and cell structures is the refractive index mismatch between cell organelles, like mitochondria, and cytoplasm, extracellular media and tissue structural components such as collagen and elastin fibers^{9,11,24,25}. The tissue scattering properties can be significantly changed due to action of osmotically active immersion liquids^{9,23,26-37}. Similar results have been obtained for optical clearing of whole blood³⁸⁻⁴⁰. Administration of the

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immersion liquid having a refractive index higher than that of tissue interstitial fluid induces a partial replacement of the interstitial fluids by immersion substance and hence, matching of refractive indices of tissue scatterers and the interstitial fluid. Osmotic activity of the immersion agents may cause tissue dehydration that also equalizes refractive indices within a tissue. Matching, correspondingly, causes the decrease of scattering. As osmotic immersion liquids, aqueous solutions of glucose and mannitol, propylene glycol, glycerol and other biocompatible chemicals are used.

The possibility of selective translucence of the superficial skin layers is very useful in developing functional imaging and therapeutic techniques. A potential benefit of the optical clearing is the improvement of laser therapeutic techniques that enable sufficient light penetration to a target embedded in tissue. Combination of optical clearing with laser radiation can reduce the laser fluences required for a therapeutic effect. In particular, for treatment of acne vulgaris knowledge of dose of irradiation absorbed in nidus of infection is very important.

Acne vulgaris is the most common skin appendage disease seen in dermatological practice. This is a follicular disorder that affects susceptible pilosebaceous follicles, primarily of the face, neck, and upper trunk and is characterized by both non-inflammatory and inflammatory lesions. Abnormal keratin production with obstruction of the follicular opening, increased production of sebum (lipids secreted by the androgen-sensitive sebaceous glands) and proliferation of *Propionibacterium acnes* leading to inflammation play the main role in the disease development. Within skin the sebaceous glands are located at depths about 200-1000 μm below the stratum corneum, i.e. located in skin dermis. Until now, the main method of acne treatment is antibiotic treatment. However, bacterial resistance is an increasing problem. Therefore, various new therapeutic possibilities using light irradiation were intensively studied. Recently, application of ICG and diode irradiation for acne treatment was described as a new approach based on selective photothermolysis of the sebaceous glands^{16,17}.

2. SKIN STRUCTURE AND OPTICAL MODEL

Skin presents a complex heterogeneous tissue where blood and pigment content are spatially distributed variably in depth⁴¹⁻⁴³. Skin consists of three main layers from the surface: epidermis (100 μm thick, the blood-free layer), dermis (1-2 mm thick, vascularized layer), and subcutaneous adipose tissue (from 1 to 6 mm thick, in dependence of the body site). Typically, the optical properties of the layers are characterized by absorption μ_a and scattering μ_s coefficients which are equal to the average number of absorption and scattering events per unit path length of photon travelling in the tissue, and the anisotropy factor g which represents the average cosine of the scattering angles.

The randomly inhomogeneous distribution of blood and various chromophores and pigments in skin produces variations of average optical properties of skin layers. Nonetheless, it is possible to define the areas in skin, where the gradient of skin cells structure, chromophores or blood amounts changing with a depth equals roughly zero⁴¹. This allows subdividing these layers into sublayers regarding the physiological nature, physical and optical properties of their cells and pigment content. The epidermis can be subdivided into two sublayers: non-living and living epidermis. Non-living epidermis or stratum corneum (about 20 μm thick) consists of only dead squamous cells, which are highly keratinized with high lipid and protein content, and has a relatively low water content^{41,42}. Living epidermis (100 μm thick) contains most of the skin pigmentation, mainly melanin, which is produced the melanocytes⁴⁴. Large melanin particles such as melanosomes (> 300 nm in diameter) exhibit mainly forward scattering. Whereas, melanin dust, whose particles are small (< 30 nm in diameter), has the isotropy in the scattering profile, and optical properties of the melanin particles (30-300 nm in diameter) may be predicted by the Mie theory. Basal layer (15 μm thick) lies between epidermal and dermal layers of skin.

Dermis is a vascularized layer and the main absorbers in the visible spectral range are the blood hemoglobin, carotene and bilirubin. In the IR spectral range, absorption properties of skin dermis are determined by absorption of water. Following the distribution of blood vessels⁴³ skin dermis can be subdivided into four layers: the papillary dermis (100-150 μm thick), the upper blood net plexus (100 μm thick), the dermis (1-2 mm thick) and the deep blood net plexus (100 μm thick). The scattering properties of the dermal layers are mainly defined by the fibrous structure of the tissue, where collagen fibrils are packed in collagen bundles and have lamellae structure. Light scatters on both single fibrils and scattering centers, which are formed by the interlacement of the collagen fibrils and bundles. It is important, that the average scattering properties of the skin are defined by the scattering properties of the dermis because of relatively big thickness of the layer (up to 2 mm) and comparable scattering coefficients of the epidermis and the dermis. Absorption of hemoglobin and water of skin dermis and lipids of skin epidermis defines the absorption properties of the whole skin.

In accordance with the structural and morphological properties of skin, in this study, we presented planar seven-layer optical model of skin (see, Fig. 1). The particular layers included in the model are the stratum corneum, the living epidermis, the basal layer of epidermis, the three layers of dermis (such as the reticular dermis with upper vessel plexus, the dermis and the lower vessel plexus) and the subcutaneous adipose tissue layer. Thickness of the layers as well as blood and water content, refractive indices of the layers and mean vessel diameters are presented in Table 1.

Table 1. The parameters of skin layers used in the simulation^{35,36,45}

Skin layers	Thickness, μm	Refractive index	Water content (% vol/vol)	Blood content (% vol/vol)	Scattering coefficient at 577 nm (bloodless), cm^{-1}	Mean vessel diameter, μm
Stratum corneum	20	1.5	0	0	300	-
Living epidermis	100	1.4	60	0	300	-
Basal layer	15	1.4	60	0	300	-
Reticular dermis & upper vessel plexus	200	1.38	75	1.7	120	6
Dermis	1500	1.35	75	1.4	120	15
Lower vessel plexus	100	1.38	75	1.7	120	6
Subcutaneous adipose tissue	3000	1.44	5	0	130	-

In the visible and NIR spectral ranges, the absorption coefficient of each layer includes contributions from the three basic chromophores: blood, melanin and water. The corresponding expression⁴⁵:

$$\mu a_k = B_k C_k \mu a_{\text{blood}}(\lambda) + (1 - B_k - W_k) \mu a_{\text{bg}} + M_k \mu a_{\text{mel}}(\lambda) + W_k \mu a_{\text{water}}(\lambda), \quad (1)$$

where $k = 1, \dots, 7$ is the layer number, B_k and W_k are the volume fractions of blood and water in the layer, factor M_k is a unity for the melanin containing layers including living epidermis, and basal layer and M_k is zero for the other layers, C_k is the correction factor, μa_{blood} , μa_{mel} , μa_{water} and μa_{bg} are the absorption coefficients of blood, melanin, water and background tissue absorption, respectively. The latter absorption coefficient is suggested to be wavelength independent and equals to 0.15 cm^{-1} . The correction factors are the number from zero through unity, taking into account the fact that blood is confined to the vessels rather than being distributed homogeneously in the tissue bulk⁴⁵. Actually, if the vessels are thick enough, the light cannot penetrate to their inner part and, therefore, the interior of the vessel does not work as an absorber. If this is the case, the correction factor is appreciably smaller than a unity. Conversely, for very thin vessels the correction factor is close to a unity. It follows from this discussion that correction factor depends on the mean vessel diameter and the blood absorption coefficient at the particular wavelength. To evaluate these factors the numeric data from⁴⁶ have been approximated by the relation: $C_k = 1 / \left(1 + a \left(0.5 \mu a_{\text{blood}} d_k^{\text{vessels}} \right)^b \right)$, where d_k^{vessels} is blood vessels diameter, cm. In case of collimated illumination of the vessels the coefficients a and b have values $a = 1.007$ and $b = 1.228$. In the case of diffuse illumination of the vessels the coefficients a and b have values $a = 1.482$ and $b = 1.151$. Optical (the absorption and the scattering) properties of blood have been calculated by us earlier⁴⁷. In this study, we use typical values 0.4 for the hematocrit and 0.8 for the oxygen saturation, the later being the average value for the venous (0.7) and arterial (0.9) blood. The absorption spectrum of water in the visible and NIR may be found in Refs. 48-50.

The total scattering coefficient can be defined as⁴⁵:

$$\mu s_k(\lambda) = B_k C_k \mu s_{\text{blood}}(\lambda) + (1 - B_k) \mu s T_k(\lambda). \quad (2)$$

The total scattering coefficient of the bloodless tissue, $\mu s T_k$, falls with the increase in the wavelength. In this study, the following relation has been used⁴⁵:

$$\mu s T_k(\lambda) = \mu s 0_k \left(\frac{577 \text{ nm}}{\lambda} \right), \quad (3)$$

where $\mu s 0_k$ are the scattering coefficients at the reference wavelength 577 nm listed in Table 1.

The expression for the anisotropy of scattering may be constructed to include the contribution from blood⁴⁵:

$$g_k(\lambda) = \frac{B_k C_k \mu_{S_{blood}}(\lambda) g_{blood} + (1 - B_k) \mu_{S_{T_k}}(\lambda) gT(\lambda)}{\mu_{S_k}(\lambda)} \quad (4)$$

where $gT(\lambda)$ is the anisotropy factor of the bloodless tissue and

$$gT(\lambda) = 0.7645 + 0.2355 \left[1 - \exp\left(-\frac{\lambda - 500 \text{ nm}}{729.1 \text{ nm}}\right) \right]. \quad (5)$$

Melanin is confined entirely to the epidermis with its total concentration depending on the skin type. In the context of our seven-layer model there are two layers containing melanin: the living epidermis and basal layers. The partitioning of melanin between the two layers depends on the skin type as well. For light skin, melanin is confined mainly to the basal layer, whereas for dark skin, the distribution of melanin in the epidermis is somewhat more homogeneous. For skin type III (in accordance with Fitzpatrick classification) investigated in this study, we use the following relationship, which describes the melanin absorption coefficient:

$$\mu_{a_{mel}}(\lambda) = A \exp\left(-\frac{\lambda - 800 \text{ nm}}{182 \text{ nm}}\right), \quad (6)$$

where parameter A is relative optical density of the pigmented layer (living epidermis or basal layer) to their thickness. In accordance with the data presented in Ref. 45 the parameter is equal to 0.87 cm^{-1} for living epidermis and 13.5 cm^{-1} for basal layer. The optical properties (the absorption and scattering coefficients and anisotropy factor) of the skin layers have been calculated in accordance with the presented model, and the result presented in Figs. 2-4.

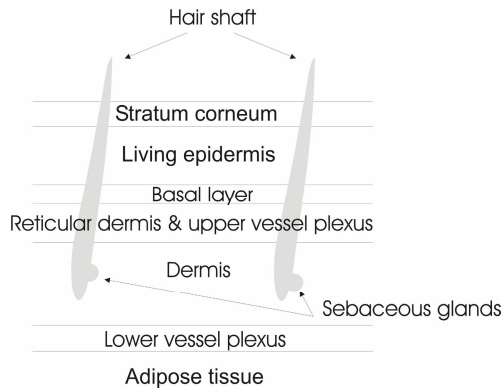


Fig. 1. The seven-layer optical model of skin.

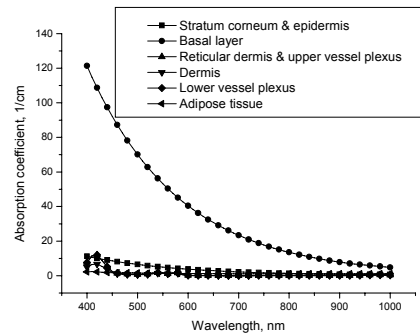


Fig. 2. The absorption spectra of different skin layers calculated in accordance with the presented optical model.

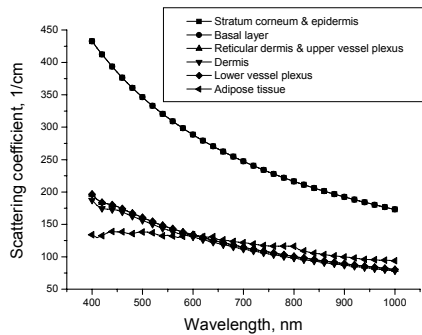


Fig. 3. The scattering spectra of different skin layers calculated in accordance with the presented optical model.

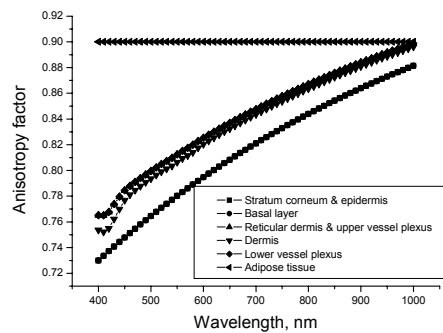


Fig. 4. The wavelength dependence of anisotropy factor of different skin layers calculated in accordance with the presented optical model.

3. MATERIALS AND METHODS

In this study, we used Monte Carlo algorithm developed by L. Wang and presented in Ref. 51. The stochastic numerical MC method is widely used to model optical radiation propagation in complex randomly inhomogeneous highly scattering and absorbing media such as biological tissues^{9,26,35,36}. Basic MC modeling of an individual photon packets trajectory consists of the sequence of the elementary simulations⁵¹: photon path length generation, scattering and absorption events, reflection or/and refraction on the medium boundaries. The initial and final states of the photons are entirely determined by the source and detector geometry. For simplicity, the incident light is assumed to be a pencil beam and the photons packets are launched normally to the tissue surface. The specular reflection from the air–tissue surface is taking into account in the simulations. At the scattering site a new photon packet direction is determined

according to the Henyey–Greenstein scattering phase function: $f_{\text{HG}}(\theta) = \frac{1}{4\pi} \frac{1-g^2}{(1+g^2-2g\cos\theta)^{3/2}}$, where θ is the

polar scattering angle. The distribution over the azimuthal scattering angle was assumed as uniform. MC technique requires values of absorption and scattering coefficients and anisotropy factor of each skin layer, its thickness and refractive indices, and the required data have been presented in Figs. 2-4 and Table 1.

For modeling of skin optical clearing we calculated the scattering properties (see Figs. 2-4) using Mie theory⁵². The theory required knowledge of refractive indices of skin scatterers and interstitial fluid and sizes of the skin scatterers. In the calculations, for approximation of scattering properties of stratum corneum, living epidermis and basal layers we used spherical particles, and for approximation of scattering properties of dermal layers we used cylindrical particles. Since the function of distribution of skin scatterers and packing factor are unknown now, we used for the calculations monodisperse, so-called Mie-equivalent, particles. In this case, for the upper layers, scattering coefficients can be

calculated as $\mu_s(\lambda) = \frac{3}{4} \frac{\varphi}{\pi a^3} \pi a^2 Q_s(a, n_s, n_l) F(\lambda)$, where a is scatterers radius (spheres), $Q_s(a, n_s, n_l)$ is

scattering efficacy factor, $F(\lambda)$ is packing factor, n_s is refractive index of scatters and n_l is refractive index of interstitial fluid, i.e. surrounding medium, and φ is volume fraction of scatterers for each layer. For dermal layers the

scattering coefficient can be calculated with the similar equation $\mu_s(\lambda) = \frac{\varphi}{\pi a^2} 2a Q_s(a, n_s, n_l) F(\lambda)$, but in this

case, a is the scatterers radius (cylinders), and $Q_s(a, n_s, n_l)$ will be calculated for cylindrical particles, too. In this study, we calculated the effective scattering size and packing factor for each wavelength by minimization of the target function

$$\text{TF}(a(\lambda), F(\lambda)) = \left(\mu_s^{\text{model}} - \mu_s^{\text{Mie}} \right)^2 + \left(g^{\text{model}} - g^{\text{Mie}} \right)^2. \quad (7)$$

Here μ_s^{model} is data presented in Fig. 3 and g^{model} is data presented in Fig. 4 for each layer. μ_s^{Mie} is scattering coefficient calculated for each layer with formulas presented above. g^{Mie} is anisotropy factor calculated to find effective scatterers size using Mie theory. To minimize the target function the Levenberg-Marquardt nonlinear least-squares-fitting algorithm described in detail by Press et al.⁵³ has been used. Iteration procedure repeats until experimental and calculated data are matched.

For investigating the influence of clearing agents on the optical parameters of skin (the diffuse reflectance, total transmittance and the light absorbed fraction), the skin interstitial fluid (water) has been replaced for clearing agents in series for different skin layers. Sizes of effective skin scatterers, packing factor and refractive index of skin scatterers are constant during all calculations. Skin layers thickness is constant, too. As clearing agents, the glycerol (refractive index 1.47), the propylene glycol (refractive index 1.43), and 40%-aqueous solution of glucose (refractive index 1.39) have been used. Refs. 16 and 17 have been introduced that for photodynamic therapy of acne vulgaris, the glycerol-ethanol-propylene glycol-water-ICG solution with ICG concentration 1 mg/ml and refractive index 1.39 can be used and the solution should be embedded in skin dermis.

4. RESULTS AND DISCUSSION

Figures 5 and 6 demonstrated optical clearing of skin after immersion of the skin by glucose solution. Fig. 5 shows reflectance spectra calculated in the framework of the presented model in the spectral range from 400 to 1000 nm. From the figure it is seen that immersion of upper skin layers (the stratum corneum, living epidermis, basal layer and reticular dermis with upper vessel plexus) changed the skin reflectance insignificantly. However, optical clearing of dermis produced significant decreasing of skin reflectance. Similar result has been observed for skin transmittance. Figure 6 shows transmittance spectra of skin calculated in the framework of the presented model in the spectral range from 400 to 1000 nm at immersion of different skin layers by glucose solution. From the figure it is seen that immersion of upper skin layers (the stratum corneum, living epidermis, basal layer and reticular dermis with upper vessel plexus) changed the skin transmittance insignificantly. However, optical clearing of skin dermis produced significant increasing of skin transmittance. It is connected with decreasing of the skin scattering properties due to the partial replacement of skin dermis interstitial fluid by glucose solution. The significant decreasing of skin reflectance and corresponding increasing of skin transmittance are connected with bigger thickness of skin dermis in comparison with thickness of other skin layers. In Figs. 5-19 legends (1-5) correspond to: 1- not cleared skin; 2 - stratum corneum, living epidermis & basal layer immersed; 3 - stratum corneum, living epidermis, basal layer & reticular dermis immersed; 4 - stratum corneum, living epidermis, basal layer, reticular dermis & dermis immersed; 5 - stratum corneum, living epidermis, basal layer, reticular dermis, dermis & lower vessel plexus immersed.

In the spectral range 400-900 nm, the form of the presented spectra is defined by the absorption bands of blood of the upper and deep blood net plexus of skin with maxima at 415, 540 and 575 nm and the spectral dependence of the scattering properties of the skin dermis. Absorption of water in this spectral range is negligible^{48,49}. The main scatterers of the dermal layer are collagen fibrils packed in collagen bundles and scattering centers, which are formed by the interlacement of the collagen fibrils and bundles. In the spectral range from 900 to 1000 nm the form of the presented spectra is defined by the absorption band of water with maximum at 975 nm^{49,50}, and the spectral dependence of the scattering properties of the skin dermis. The influence of skin melanin, which localized in upper skin layers, appear as decreasing both reflectance and transmittance of skin, especially in the short-wavelength spectral range.

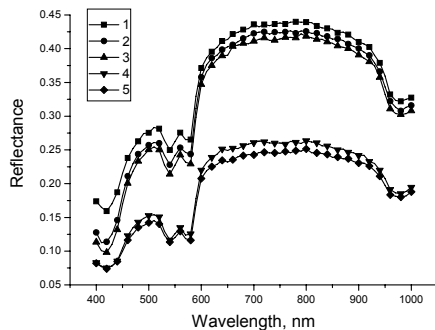


Fig. 5. The reflectance spectra of skin before and after the skin immersion by glucose solution.

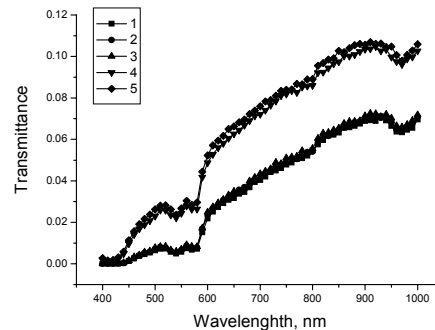


Fig. 6. The transmittance spectra of skin before and after the skin immersion by glucose solution.

Figures 7-9 demonstrated evolution of light absorbed fraction in different skin layers at immersion of the layers by glucose solution. Fig. 7 shows wavelength dependence of light absorbed fraction in upper skin layers at immersion of the layers by glucose solution. From the figure it is seen that immersion of upper skin layers (the stratum corneum, living epidermis, basal layer and reticular dermis with upper vessel plexus) changed of the light absorbed fraction insignificantly. However, optical clearing of dermis produced significant decreasing of the light absorbed fraction. It is connected with increase of light penetration depth into skin, and corresponding decreasing of backscattering flux from lower skin layers. Opposite result has been obtained for skin adipose layer (see, Fig. 9). From the figure it is seen that optical clearing of skin dermis produced significant increase of light penetration depth into skin and corresponding increase of the light absorbed fraction in subcutaneous adipose tissue. It is a very interesting result obtained for skin dermis (see, Fig. 8). From the figure it is seen that light absorbed fraction is not changed independently from optical clearing of any skin layers. That can be explained by the fact that fluence rate in the layer is formed by backscattering flux from subcutaneous adipose tissue.

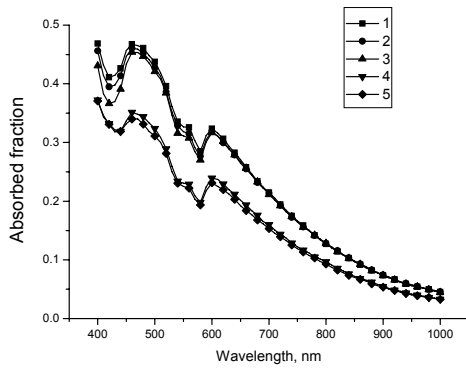


Fig. 7. The absorbed fraction of light in upper (the stratum corneum, epidermis & basal) layers of skin before and after the skin immersion by glucose solution.

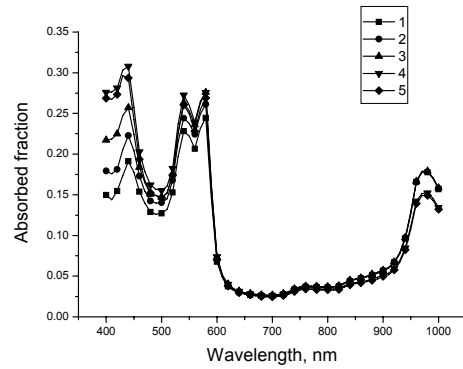


Fig. 8. The absorbed fraction of light in skin dermis before and after the skin immersion by glucose solution.

Figures 10-14 present the results of computer modeling of skin optical clearing by propylene glycol. From these figures it is seen that the obtained results are very close to the results presented in Figures 5-9, which is connected with skin optical clearing by glucose solution. That can be explained by differences between value of refractive index of collagen (the main structural component of biological tissues including skin) and refractive indices of glucose solution and propylene glycol. Mean value of refractive index of collagen is 1.41, value of refractive index of glucose solution is 1.39, and value of refractive index of propylene glycol is 1.43. In both cases the differences between refractive indices of tissue scatterers and surrounding medium is 0.02, and thus the scattering efficacy has to change in the similar manner.

Figures 15-19 presented the results of computer modeling of skin optical clearing by glycerol. From these figures it is seen that the obtained results are different in comparison from the results obtained for skin optical clearing by glucose solution and propylene glycol. From figures 15 and 16 it is seen that glycerol-produced effect is significantly smaller decrease of skin reflectance and increasing of skin transmittance in comparison with propylene glycol and glucose solution. That can be explained by differences between value of refractive index of collagen (the main structural component of biological tissues including skin) and refractive indices of glycerol. Mean value of refractive index of collagen is 1.41, value of refractive index of glycerol is 1.47. In this case replacement of water (interstitial fluid in normal state) by glycerol produced only insignificant reducing of skin scattering.

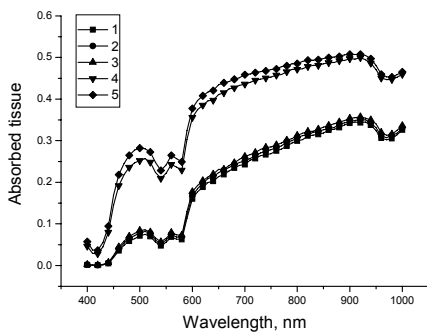


Fig. 9. The absorbed fraction of light in skin adipose layer before and after the skin immersion by glucose solution.

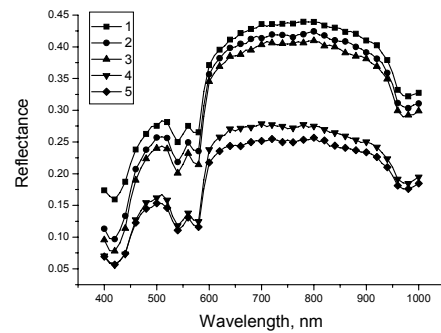


Fig. 10. The reflectance spectra of skin before and after the skin immersion by propylene glycol.

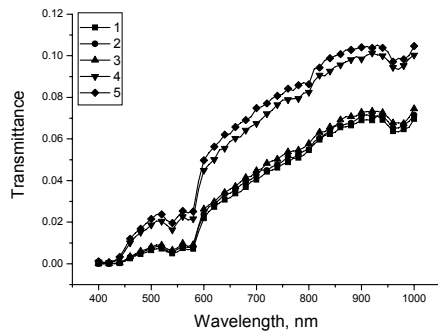


Fig. 11. The transmittance spectra of skin before and after the skin immersion by propylene glycol.

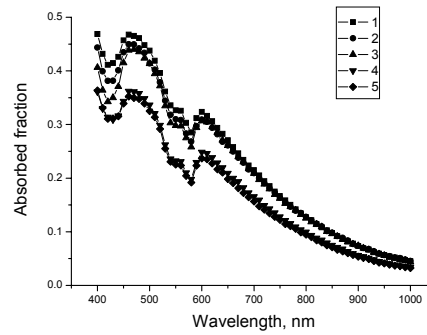


Fig. 12. The absorbed fraction of light in upper (the stratum corneum, epidermis & basal) layers of skin before and after the skin immersion by propylene glycol.

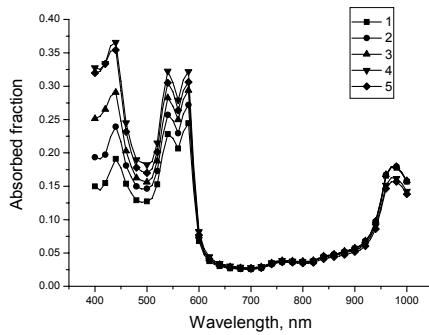


Fig. 13. The absorbed fraction of light in skin dermis before and after the skin immersion by propylene glycol.

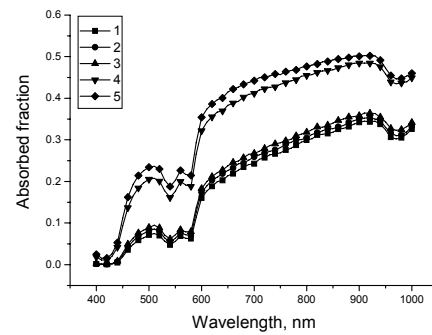


Fig. 14. The absorbed fraction of light in skin adipose layer before and after the skin immersion by propylene glycol.

Figures 17-19 show wavelength dependence of light absorbed fraction in different skin layers at immersion the layers by glycerol. Fig. 17 shows light absorbed fraction in the upper skin layers at immersion of the layers by glycerol. From the figure it is seen that immersion of any skin layers changed the light absorbed fraction in the upper skin layers insignificantly. The similar result has been observed for skin dermis, too (see, Fig. 18).

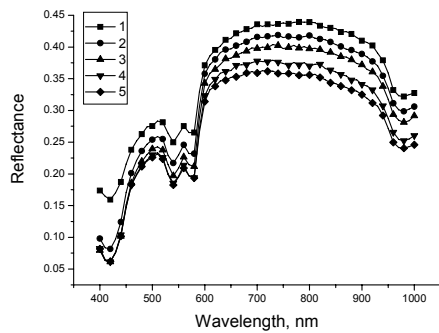


Fig. 15. The reflectance spectra of skin before and after the skin immersion by glycerol.

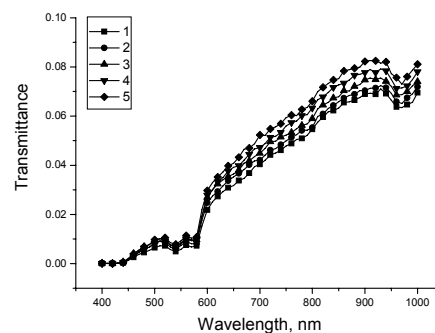


Fig. 16. The transmittance spectra of skin before and after the skin immersion by glycerol.

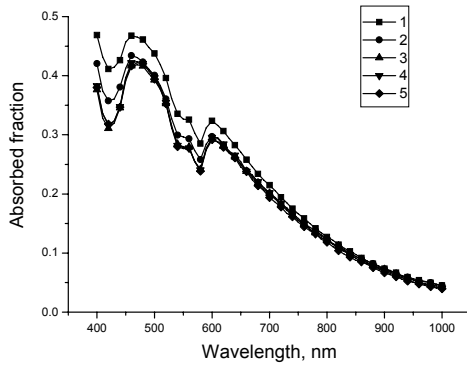


Fig. 17. The absorbed fraction of light in upper (the stratum corneum, epidermis & basal) layers of skin before and after the skin immersion by glycerol.

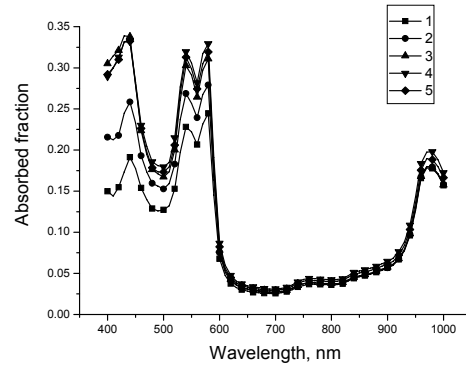


Fig. 18. The absorbed fraction of light in skin dermis before and after the skin immersion by glycerol.

Figure 20 shows absorption spectrum of the glycerol-ethanol-propylene glycol-water-ICG solution with ICG concentration 1 mg/ml that is introduced into skin dermis in accordance with method of acne vulgaris treatment described in Ref. 16. Since the knowledge of dose of irradiation absorbed in nidus of infection is very important for light treatment of acne vulgaris, and optical clearing of different skin layers changes absorbed fraction of light in skin dermis (especially in the short-wavelength spectral range) then computer modeling of the absorbed fraction of light in skin dermis at optical clearing of different skin layers can be important.

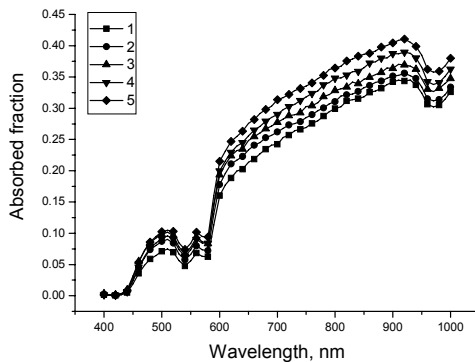


Fig. 19. The absorbed fraction of light in skin adipose layer before and after the skin immersion by glycerol.

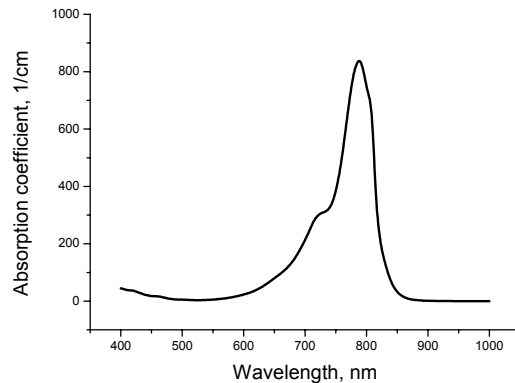


Fig. 20. The absorption coefficient of the glycerol-ethanol-propylene glycol-water-ICG solution with ICG concentration 1 mg/ml¹⁶.

Figures 21-23 demonstrated light absorbed fraction in skin dermis at immersion of different layers of skin by glucose solution, propylene glycol and glycerol at simultaneous introduction in interstitial fluid of dermis the glycerol-ethanol-propylene glycol-water-ICG solution. From the figures it is seen that immersion of upper skin layers (the stratum corneum, living epidermis and basal layer) changes the light absorbed fraction in the red spectral range insignificantly. However, in short-wavelength spectral range optical clearing of the layers produces more significant increase of the light absorbed fraction. From the result one can conclude that when applying the glycerol-ethanol-propylene glycol-water-ICG solution, the effect of using of optical clearing technique is insignificant. However, at when applying blue light sources (and corresponding dyes) practical effect from using of optical clearing of superficial skin layers can be important and significant.

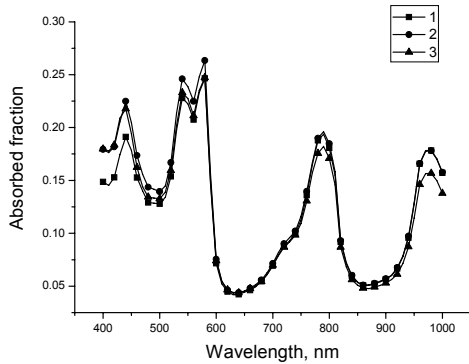


Fig. 21. The absorbed fraction of light in skin dermis before and after the skin immersion by glucose solution.

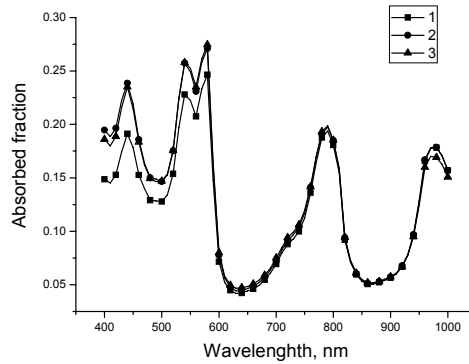


Fig. 22. The absorbed fraction of light in skin dermis before and after the skin immersion by propylene glycol.

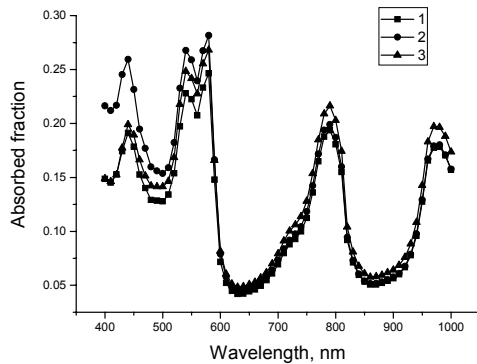


Fig. 23. The absorbed fraction of light in skin dermis before and after the skin immersion by glycerol.

In the Figs. 21-23:

- 1- without clearing;
- 2 – stratum corneum, epidermis & basal layer immersed;
- 3 - dermis immersed

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