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REVIEW

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Abstract. A survey of the literature is presented that provides an analysis of the optical properties of human skin, with particular regard to their applications in medicine. Included is a description of the primary interactions of light with skin and how these are commonly estimated using radiative transfer theory (RTT). This is followed by analysis of measured RTT coefficients available in the literature. Orders of magnitude differences are found within published absorption and reduced-scattering coefficients. Causes for these discrepancies are discussed in detail, including contrasts between data acquired *in vitro* and *in vivo*. An analysis of the phase functions applied in skin optics, along with the remaining optical coefficients (anisotropy factors and refractive indices) is also included. The survey concludes that further work in the field is necessary to establish a definitive range of realistic coefficients for clinically normal skin. [DOI: 10.1117/1.JBO.17.9.090901]

Keywords: skin optics; radiative transfer theory; absorption; scatter; melanin; hemoglobin.

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1 Introduction

The color of human skin has long been used as a subjective adjunct to the detection and diagnosis of disease. More recently, the introduction of skin color measurements has extended this to include the potential for objective determination of skin features,[1] including melanin and hemoglobin concentrations,[2–6] the depth and diameter of blood vessels,[7–9] the depth of pigmented skin lesions,[10,11] the maturity and depth of bruises,[12,13] and keratin fiber arrangements.[14]

Such advances have proved invaluable for the advancement of skin laser treatments[15] and photodynamic therapy,[16–18] and have contributed to further advances in the diagnosis of cancerous and noncancerous skin lesions.[10,19–21]

However, the success of these methods depends entirely upon adequate knowledge of the behavior of light as it impinges upon, and travels through, the skin. This article presents a description of the major interactions of visible light with skin and the principal skin features that contribute to these. This is followed by an analysis of published optical coefficients used in simulations of light transport through skin.

2 Background

2.1 *Absorption*

Absorption describes a reduction in light energy. Within the visible region, there are two substances generally considered to dominate the absorption of light in skin: hemoglobin and melanin.

Hemoglobin is the dominant absorber of light in the dermis. Normal adult hemoglobin (Hb A) is a protein consisting four polypeptide chains, each of which is bound to a heme.[22] The heme in Hb A is named iron-protoporphyrin IX[23,24] and is responsible for the majority of light absorption in blood. The free-electron molecular-orbital model describes this absorption as an excitation of loosely bound "unsaturation electrons" or "π-electrons" of the heme.[25] Within the visible region, Hb A contains three distinctive peaks. The dominant peak is in the blue region of the spectrum and is referred to as the Soret peak or Soret band. Two further peaks can be distinguished in the green-yellow region, between 500 and 600 nm, that in combination with the Soret band cause Hb A to appear red. These are known as the α and β bands, or collectively as the Q-band, and have intensities of around 1% to 2% of the Soret band.[26] The excitation levels of π-electrons vary, and therefore the positions and intensities of these bands vary with the ligand state of the heme (Fig. 1).

Melanins are ordinarily contained within the epidermis and produce an absorption spectrum that gradually decreases from the ultraviolet (UV) to the infrared (IR) regions. In contrast to hemoglobin, the variation and complexity of melanins means that their detailed structures are not yet fully understood, despite intense research over the last five decades, and this broadband absorbance spectrum is still a topic of scientific debate.[5,28,29] At present, the scientific consensus appears to gravitate towards a chemical disorder model.[5,28–33] This model proposes that melanins consist of a collection of oligomers or polymers in various forms arranged in a disordered manner. This results in a number of absorption peaks that combine to create a broadband absorbance effect[28,30] (Fig. 2).

Further absorption of light may be attributed to chromophores, such as bilirubin and carotene,[34] lipids,[35] and other structures, including cell nuclei and filamentous proteins[36,37] Although the individual contributions from these secondary chromophores may be considered separately,[13,38,39] most simulations group them into a single overarching value.[40,41]

Despite its abundance in all tissues, water is not a significant absorber of light in the visible region, although its contribution has been considered when simulating skin color.[42]

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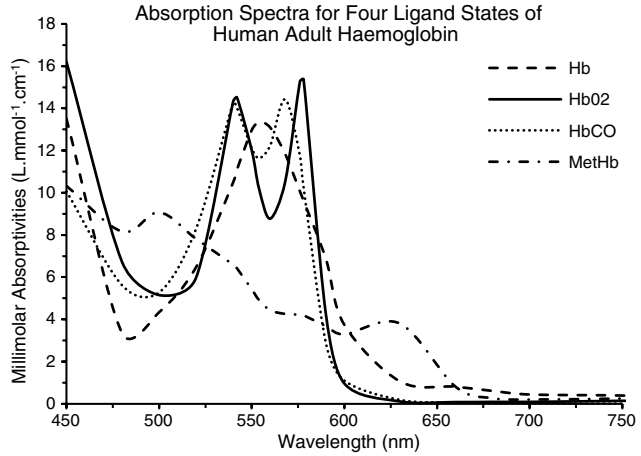

Fig. 1 Absorption spectra of deoxyhemoglobin (Hb), oxyhemoglobin (HbO_2), carboxyhemoglobin (HbCO), and methemoglobin (MetHb) in the visible region, from Ref. 27.

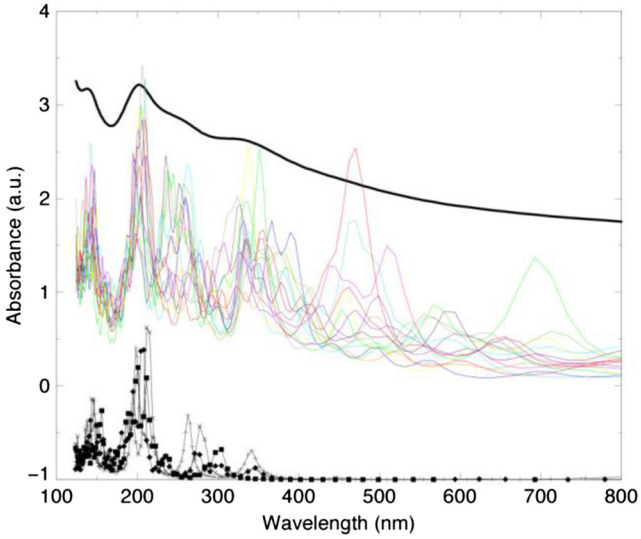

Fig. 2 Colored lines show individual absorption spectra of tetramer subunits within melanin extracted from human epidermis. The average absorption spectrum of these is shown by thick black line, shifted up 1.5 units for clarity. Thin black lines shifted down by one unit represent absorption spectra from monomer subunits. a.u. = arbitrary units. Reprinted figure with permission from Ref. 30.

2.2 *Scattering*

As well as absorption, scattering contributes significantly to the appearance of skin. Scattering describes a change in the direction, polarization or phase of light and is commonly portrayed as either a surface effect (such as reflection or refraction) or as an interaction with a small region whose optical properties differ from its surroundings (particulate scatter).

It has been estimated that 4% to 7% of visible light is reflected from the surface of the skin, independent of wavelength and skin color.[43,44] The remaining light is refracted as it passes from air into the skin.

The primary sources of particulate scatter within the skin are filamentous proteins. Keratins are the filamentous proteins of the epidermis and form this layer's major constituent, whereas collagen is the principal filamentous protein of the dermis and occupies approximately 18% to 30% of its volume.[45] Further scatter is attributed to melanosomes in the epidermis, cell nuclei, cell walls and many other structures in the skin that occur in smaller numbers.[46]

Scatter from filamentous proteins has been approximated using a Mie solution to Maxwell's equations applied to data from *in vitro* skin samples.[47,48] This approach provides an increase in simulated scattering probability with increasing fiber diameter and decreasing wavelength. The dependence of scatter on fiber diameter suggests that the protein structures of the dermis, which may be 10 times as large as those in the epidermis,[45,49] possess a greater scattering cross-section. This in part compensates for the lower number densities of filamentous proteins in the dermis. The scattering events that occur are mainly in the forward direction, meaning that on average, light that returns to the surface undergoes a large number of scattering events.[50] One implication of the wavelength dependence of scatter is that blue and green light that has returned to the surface of the skin will have, on average, travelled less deeply than red light. This is considered the primary reason why blood vessels and pigmented nevi that are situated deeper within the skin are only able to absorb light from the red end of the spectrum and therefore appear bluer than their superficial equivalents.[51,52]

The volume fraction of melanosomes in the epidermis varies typically from 1% in pale skin to 5% in darker skin,[53] although one group has suggested greater values.[54,55] However, despite their low numbers relative to keratins, melanosomes are approximately 10 times the diameter of the largest keratin structures in the epidermis[56] and possess a greater refractive index[57] (and therefore a greater difference in refractive index at their interface with skin). Melanin has been shown to contribute significantly to the degree of scatter within the epidermis.[58,59] As well as the volume fraction, the distribution and size of melanin structures in the epidermis also vary with skin type. Thus, the total amount of scatter that occurs as a result of melanin in the epidermis can vary substantially between individuals,[60,61] although this is not always taken into account when simulating the effects of varying melanin concentration on skin color,[42,62] or when simulating laser treatments,[15,63] for example.

Blood normally occupies around 0.2% to 0.6% of the physical volume of the dermis[2,6,54,64–66] depending upon its anatomical location. The vessel walls surrounding this blood, in addition to the walls of vessels that remain vacant, may occupy a similar volume. Dermal vessels vary in thickness and structure from capillaries of around 10 to 12 μm diameter at the epidermal junction to terminal arterioles and post-capillary venules (approximately 25 μm in diameter) in the papillary dermis and venules (approximately 30 μm) in the mid-dermis.[67] Furthermore, blood vessels occur in higher densities at particular depths, giving rise to so-called blood vessel plexi.[67] The contribution to light scatter by these structures, inclusive of refraction effects, may be significant[68–70] and varies with location and depth, as well as between individuals.* Larger, deeper vessels may also contribute to the color of skin.

*Assuming a reduced scattering coefficient of 0.5 mm^{-1} for blood and 40 mm^{-1} for vessel walls at 633 nm, a 0.5% volume fraction of each contributes approximately 0.2 mm^{-1} to the dermal reduced scattering coefficient, measured at around 1-5 mm^{-1} (Fig. 9). The contribution will be larger within blood vessel plexi.

Scattering from the remaining structures of the skin, including cell walls, nuclei, and organelles,[36] hairs and glands, is rarely of central interest to a study of skin optics. As a result, the contributions from these structures to the total measured scattering coefficients are not routinely considered separately.[71]

3 Simulating Light Transport Through Skin

Optical simulations involving mathematical models of healthy human skin generally approximate the surface as perfectly smooth, although some computer graphics models have applied calculations of directional reflectance from rough surfaces.[72] Surface scattering effects (reflection and refraction) can be calculated for smooth surfaces using Fresnel's equations and Snell's law, respectively:

$$R = \frac{1}{2}\frac{(a-c)^2}{(a+c)^2}\left\{1+\frac{[c(a+c)-1]^2}{[c(a-c)+1]^2}\right\}. \quad (1)$$

Fresnel reflection (R) of unpolarized light from air to skin, where $c = \cos(\theta_i)$, θ_i is the angle of incidence, $a = n^2 + c^2 - 1$ and n is the refractive index of skin.

$$\theta_t = \arcsin\left(\frac{1}{n}\sin\,\theta_i\right). \quad (2)$$

Angle of refraction (θ_t) at the skin's surface calculated using Snell's law.

Within the skin, both absorption and scatter must be considered simultaneously. These may be described in the classical approach by Maxwell's equations, which consider the interactions between the electric and magnetic fields of light with matter. However, an exact solution to Maxwell's equations requires precise knowledge of each structure within the medium and becomes prohibitively complex for the case of human skin.

The most commonly used approximation to Maxwell's equations in the field of skin optics is radiative transfer theory (RTT).[73] This considers the transport of light in straight lines (beams),with absorption simulated as a reduction in the radiance of a beam and dependent upon the absorption coefficient (μ_a). The degree of scattering is described by the scattering coefficient (μ_s), which considers both a loss of radiance in the direction of the beam and a gain from beams in other directions, and the phase function (p), the probability that an individual beam will scatter in any particular direction. The reduced-scattering coefficient (μ_s') combines these variables, i.e., $\mu_s' = \mu_s(1-g)$, where g is the anisotropy factor, the average cosine of the scattering angle θ:

$$g = \int_{4\pi} p(\cos\,\theta)\cos\,\theta \mathrm{d}\omega, \quad (3)$$

where $\mathrm{d}\omega$ is a differential solid angle.

In order for RTT to be valid, it must be assumed that any cause for increasing or decreasing the radiance of a beam other than that described by the absorption and scattering coefficients, including inelastic scatter (fluorescence or phosphorescence) and interactions between beams (interference), is negligible. The skin model must also consist of volumes that are homogeneous with regards to μ_s, μ_a and p, and that do not change over time.

4 Optical Coefficients of Skin

A considerable amount of work has been carried out to determine appropriate values of the RTT coefficients. Cheong et al.[74] described both direct (*in vitro*) and indirect (*in vivo*) methods of measuring absorption and scatter. A comprehensive analysis of the literature involving each method is presented here.

4.1 *Absorption Coefficients*

4.1.1 *In vitro absorption coefficients*

Direct measurements have the potential to produce repeat measurements of a predetermined volume or section of skin and, unlike *in vivo* measurements, can include transmission data. However, the processes necessary to extract and prepare a skin sample cannot be carried out without altering its optical properties.

The *in vitro* studies presented in Fig. 3 vary significantly in tissue-processing methodologies, measurement setup and the interpretation of data. For example, Jacques et al.'s work[75] included three methods of tissue preparation. The epidermis was separated from the dermis using a micro-cryotome for one set of skin samples, after mild thermal treatment in a water bath in another set, and was not separated in a third set. The same mild thermal treatment was used to separate the dermis and epidermis in Prahl's work[55] and a micro-cryotome was also applied in Salomatina et al.'s study.[59] No separation of the epidermis was reported by Chan et al.[76] or Simpson et al.[77] Although Salomatina's work shows greater absorption from the *in vitro* epidermis when compared to the dermis, the studies analyzed here do not demonstrate a clear distinction in absorption coefficients reported between the methods of separation described, nor between those that separated the epidermis and those that did not.

The level of hydration is likely to have varied considerably between the studies analyzed. Prahl[55] and Jacques et al.[75] soaked samples in saline for at least 30 min before carrying out measurements, during which the samples were placed in a tank of saline. Salomatina et al.[59] also soaked their skin samples prior to measurement and sealed them between glass slides to maintain hydration. Chan et al.[76] and Simpson et al.[77] did not soak their samples prior to or during measurement. Jacques et al. reported that soaking the sample increases back-scattered reflectance, although the effects on the calculated absorption are not described. Chan et al. commented that dehydration may elevate the measured absorption coefficient. However, the greatest reported absorption coefficients are those from rehydrated tissue samples. From the information available, the effect of tissue hydration on the measured absorption coefficients is not clear.

Data was interpreted using Monte Carlo simulations by Salomatina et al.,[59] Simpson et al.[77] and Graaf et al.,[78] an adding-doubling technique by Prahl et al.,[55] and by direct interpretation in Chan et al.'s[76] and Jacques et al.'s[75] studies. Both the methods described in the Monte Carlo simulations and Prahl's adding-doubling technique are based upon assumptions of optically homogeneous tissue layers, uniform illumination and no time dependence, and both are essentially discrete solutions to the radiative transport equation. The methods described contrast in their approach to internal reflection for beams exiting the skin model and the adding-doubling method relies upon accurate representation of the angular distribution of beams exiting the thin layer upon which the model is built. It is not directly

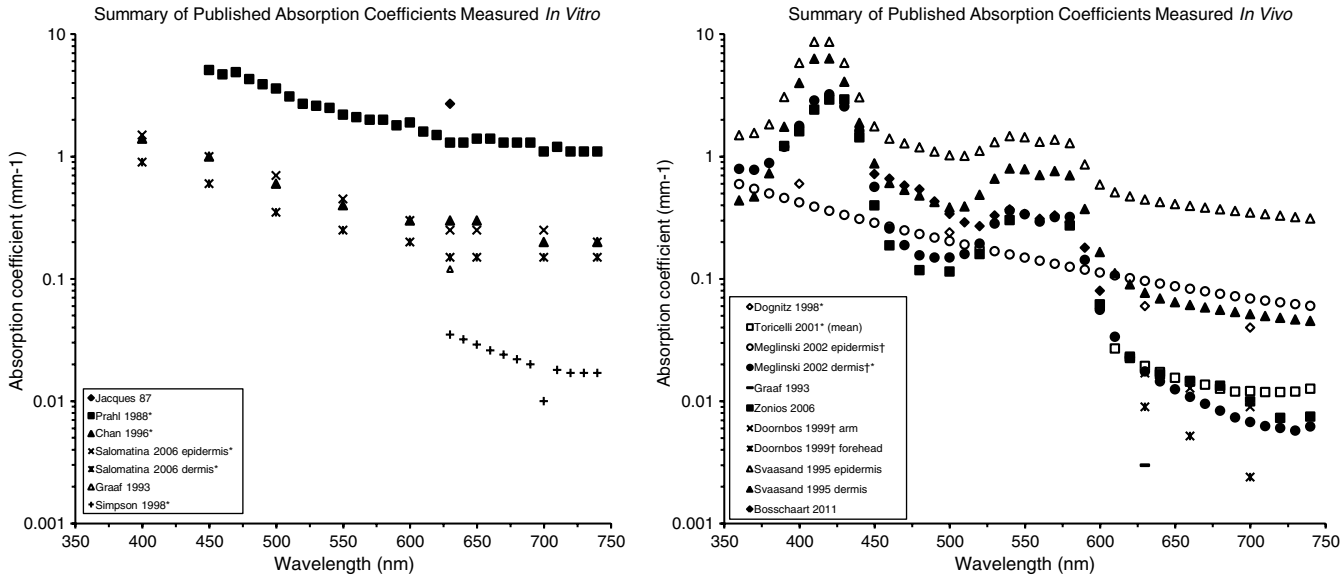

Fig. 3 Summary of absorption coefficients available in the literature. *In vitro* data represents absorption coefficients from exsanguinated skin whereas dermal *in vivo* data is inclusive of blood absorption. *Data obtained from graphical presentation. †Data presented was not complete and required input of hemoglobin or water optical properties obtained from.[35] The raw data is provided in the Appendix (Table 1).

clear if or how these differences may have contributed to the higher absorption coefficients reported by Prahl et al.

It is also of interest that Prahl et al.'s,[55] Chan et al.'s[76] and Salomatina et al.'s[59] studies, whose samples varied in thickness between 60 and 780 μm, did not demonstrate a clear correlation between sample thickness and published absorption coefficients but Simpson et al.'s published absorption coefficients, which are an order of magnitude smaller than the other values analyzed here, involved much thicker samples (1500 to 2000 μm thick). Thus, differences between the published absorption coefficients across these studies may have resulted from variations in the regions of skin investigated or the ability of the simulations to correctly account for boundary effects at the lower boundary.

4.1.2 *In vivo absorption coefficients*

Indirect measurements do not suffer from such changes in the properties of the interrogated skin volume, although care must be taken to consider variations in blood perfusion, for example, which may result from sudden changes in ambient temperature, the use of some drugs and even contact between the skin and the measurement device.[79]

In general, absorption coefficients measured *in vivo* may be expected to be higher than *in vitro* values where the highly absorbing pigments from blood are removed from the samples. This is particularly true in the blue-green regions of the visible spectrum. Assuming a value of 0.5% blood volume in the dermis, this would contribute approximately 8 cm^{-1} at 410 nm (Soret band), 0.6 cm^{-1} at 500 nm and 1.4 cm^{-1} at 560 nm (Q-band), but only around 0.05 cm^{-1} at 700 nm (values calculated from[35]). This contribution is not reflected in the literature. Absorption coefficients obtained from *in vivo* work show greater variation, but are not consistently higher than those obtained from *in vitro* work (Fig. 3).

Absorption coefficients from Svaasand et al.,[80] Zonios et al.,[81] and Meglinski and Matcher[42] clearly demonstrate the effect of blood on the measured absorption coefficients. Each study shows an absorption peak between 400 and 450 nm corresponding to the Soret band and a double peak at approximately 540 and 575 nm corresponding to the α and β bands of oxyhemoglobin (see Fig. 1). There are, however, notable differences between the absorption coefficients produced from the three studies. Meglinski and Matcher and Svaasand et al. considered epidermal absorption coefficients separately to dermal values. The reported values from Svaasand et al. are greater, and show a different spectral curve to those from Meglinski and Matcher. This is a direct result of Svaasand et al.'s inclusion of 0.2% blood by volume in the calculation of epidermal absorption coefficients, representing blood infiltrating the modeled epidermal layer from the papillae. Compared to Meglinski and Matcher's dermal values and Zonios et al.'s absorption coefficients for their skin model consisting a single layer, both of which also included the influence of blood, Svaasand et al.'s reported dermal absorption coefficients were consistently high. This is despite using a dermal blood volume fraction of 2%, compared to an average of 12% from Meglinski and Matcher's study and a value of 2.6% in Zonios et al.'s work. The cause of this discrepancy is the variation in magnitude of the blood absorption coefficients applied across the three studies (Fig. 6, Appendix). Bosschaart et al.[82] employed a diffusion approximation technique to their data collected from neonates, effectively applying a single value of absorption across the skin volume. Their data is in close agreement to Meglinski and Matcher's dermal absorption coefficients in the 530- to 600-nm range, but the contribution of melanin produces a relative increase in Bosschaart et al.'s values at shorter wavelengths.

Data selected for analysis in this work involved "Caucasian" skin types only. Where stated, these studies involved skin types described as Northern European. Where not stated, it was assumed that such skin types were used except for the studies carried out by Zonios et al.[81] and Torricelli et al.[83] that were conducted in Southern Europe. However, the latter two studies did not report higher absorption coefficients, as may be expected from measurements on darker skin types. In Zonios et al.'s

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