МИНИСТЕРСТВО ЗДРАВООХРАНЕНИЯ РЕСПУБЛИКИ БЕЛАРУСЬ
БЕЛОРУССКИЙ ГОСУДАРСТВЕННЫЙ МЕДИЦИНСКИЙ УНИВЕРСИТЕТ
КАФЕДРА МЕДИЦИНСКОЙ И БИОЛОГИЧЕСКОЙ ФИЗИКИ

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ОПТИЧЕСКИЕ СВОЙСТВА КОЖИ, ТКАНЕЙ ЗУБА И СТОМАТОЛОГИЧЕСКИХ МАТЕРИАЛОВ

OPTICAL PROPERTIES OF SKIN, TOOTH TISSUES AND DENTAL MATERIALS

Учебно-методическое пособие

Минск БГМУ 2013

УДК 53 (811.111)-054.6 (075.8)
ББК 22.3 (81.2 Англ-923)
Н63

Рекомендовано Научно-методическим советом университета в качестве учебно-методического пособия 27.11.2013 г., протокол № 3

Р е ц е н з е н т ы: канд. физ.-мат. наук, доц., декан факультета компьютерного проектирования Белорусского государственного университета информатики и радиоэлектроники С. К. Дик; канд. физ.-мат. наук, доц. кафедры медицинской и биологический физики Белорусского государственного медицинского университета Н. И. Инсарова

Никоненко, Н. А.

Н63 Оптические свойства кожи, тканей зуба и стоматологических материалов : учеб-метод. пособие / Н. А. Никоненко. – Минск : БГМУ, 2013. – 35 с.

ISBN 978-985-528-930-3.

Издание содержит сведения о процессах, связанных с распространением света в биологических тканях, особое внимание уделено рассмотрению оптических свойств кожи, тканей зуба и композиционных материалов для создания эстетических реставраций в стоматологии. Знание оптических характеристик биологических тканей необходимо для понимания физических основ применения методов оптической диагностики и терапии в стоматологии.

Предназначено для иностранных студентов-стоматологов, изучающих медицинскую и биологическую физику на английском языке.

УДК 577.3 (075.8)
ББК 28.707 (81.2 Англ-923)

ISBN 978-985-528-930-3

INTRODUCTION

Currently a great interest is being expressed to the application of optical techniques for medical imaging, diagnosis, therapy and surgery [1–9]. It is motivated by the design of novel lasers, growing development of fiber optic techniques and other associated technologies.

In its turn the development of optical methods in modern medicine has stimulated the investigation of optical properties of various biological tissues, since most applications in medical optics (for example, in therapy), are based on the mechanism of light interaction with biological tissues and spatial distribution of light in the target tissue.

Determination of light propagation mechanism in biological medium is a rather complicated task because it is inhomogeneous and the presence of microscopic inhomogenities (macromolecules, cell organelles, organized cell structures, etc.) makes it turbid. Multiple scattering within a turbid medium leads to loss of directionality. Therefore, the majority of the laws, valid for the homogenous, nonscattering medium optics, are not applicable to light propagation in tissues.

Modeling light propagation and energy distribution inside the tissues requires accurate knowledge of their optical properties. This textbook describes processes related to light propagation in human tissues (especially, the absorption and scattering) with the concentration on the optical characteristics of skin, hard tooth tissues and modern dental materials for esthetic restorations in dentistry.

PART 1
ELECTROMAGNETIC NATURE OF LIGHT

1.1. ELECTROMAGNETIC WAVE CHARACTERISTICS

In physics, electromagnetic radiation is often described from either "classic" or "quantum" viewpoints.

In classical physics (wave optics), electromagnetic radiation is considered to be an *electromagnetic wave* which represents electric and magnetic fields that simultaneously oscillate in planes mutually perpendicular to each other and to the direction of propagation through space (fig. 1.1).

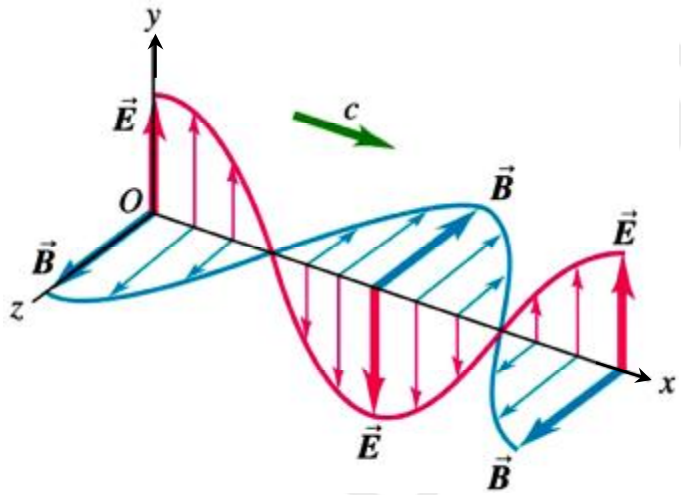


Fig. 1.1. A schematic view of an electromagnetic plane wave propagating along the *x*-axis

It is worth noting that wave optics explains such phenomena as the propagation of light, light polarization, reflection, refraction, diffraction, and interference effects.

Mathematical description of the electromagnetic wave propagating in *x*-axis direction is given by the following expressions (called *the electromagnetic wave equation*):

$$E = E_0 \sin \omega (t - \frac{x}{\upsilon}), \;\; B = B_0 \sin \omega (t - \frac{x}{\upsilon}), \qquad (1.1)$$

where $\omega = 2\pi/T = 2\pi\nu$ (rad/s) is an angular (circular) frequency, $\nu = 1/T$ is the wave frequency, T is the wave period, E_0 and B_0 are the maximum values of the electric ($\vec{E}$) and magnetic ($\vec{B}$) field vectors (the amplitudes); υ is the speed of electromagnetic wave, t and x are the time and the coordinate corresponding to the direction of wave propagation.

Electromagnetic wave is characterized by its frequency, wavelength, amplitude and speed.

Frequency (ν) is defined as the number of complete cycles of oscillation per second, expressed in s^{-1} = Hz (hertz).

Wavelength (λ) is the distance between two consecutive peaks or troughs in a wave (fig. 1.2).

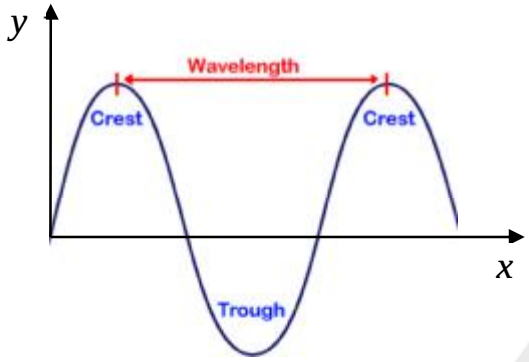


Fig. 1.2. A schematic wavelength representation

The relation between the wavelength and the frequency of wave is $\lambda = \upsilon \cdot T = \upsilon/\nu$, where the υ is the *speed of electromagnetic wave*.

In electromagnetic theory the speed of electromagnetic waves in vacuum (denoted *c*) is expressed as follows:

$$c = \frac{1}{\sqrt{\varepsilon_0 \mu_0}} = 2.98 \cdot 10^8 \text{ m/s} \approx 3 \cdot 10^8 \text{ m/s}, \qquad (1.2)$$

where $\varepsilon_0 = 8.85 \cdot 10^{-12}$ F/m and $\mu_0 = 1.43 \cdot 10^{-7}$ H/m are the electric and magnetic constants.

Theoretically calculated value of electromagnetic wave speed in vacuum (1.2) appeared to be approximately equal to the experimentally determined speed of visible light in air. Thus the wave nature of light has been established.

In the quantum model, electromagnetic radiation is viewed as a stream of particles called *photons* (i. e., "quanta"). Each photon has the energy proportional to the frequency of the electromagnetic wave: $E = hn = hc/\lambda$, where *h* is Planck's constant ($h = 6.6262 \cdot 10^{-34}$ J·s). Light interaction with matter such as the photoelectric effect is described solely by the particle nature.

Thus light has a dual nature. Light behaves either as a wave, or, as a particle. This phenomenon is known as the wave-particle duality.

The electromagnetic spectrum represents the distribution of electromagnetic radiation according to wavelengths and photon energies (fig. 1.3).

The selected electromagnetic radiation regions and the corresponding approximate wavelengths are the following:

- Radio waves, the wavelength $\lambda > 1$ mm;
- Infrared (IR) radiation, 0.78 μm $< \lambda <$ 1 mm;
- Visible light, 400 nm $< \lambda <$ 780 nm;
- Ultraviolet radiation (UV), 80 nm $< \lambda <$ 400 nm;
- X-rays, 10^{-5} nm $< \lambda <$ 80 nm;
- Gamma rays $\lambda < 10^{-5}$ nm.

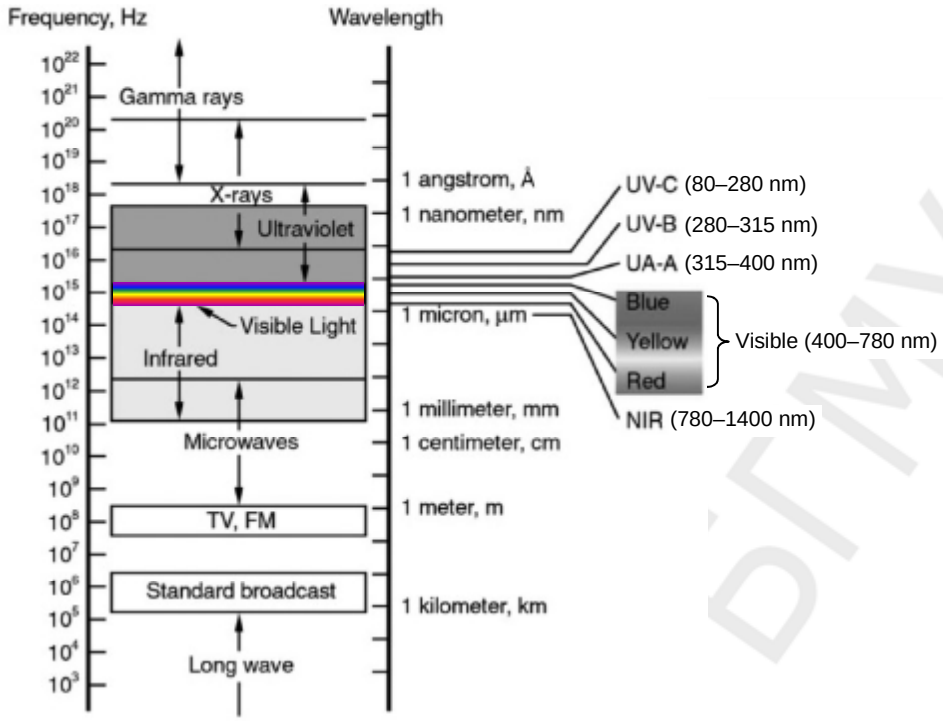


Fig. 1.3. Graphical representation of the electromagnetic spectrum

The term *light* is frequently used to refer to the portion of the electromagnetic spectrum with wavelengths ranged from 1 mm to 80 nm (*optical wavelength region*). This spectral range includes the infrared, the visible, and the ultraviolet radiation. In our further consideration the term light will also be referred to the radiation in the given wavelength range.

1.2. LIGHT REFRACTION AND REFLECTION

Refraction. The index of refraction is a fundamental property of media. A*bsolute refractive index* of the medium (n) is defined by the expression: $n = \frac{c}{\upsilon}$, where c is the speed of light in vacuum and υ is the light speed in the medium. Thus the value of n is equal to the factor by which the velocity of light in the given medium is less than that in vacuum.

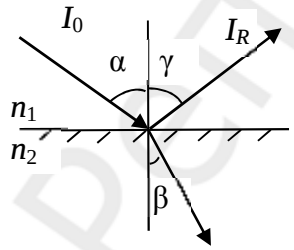


Fig. 1.4. Light reflection and refraction phenomena

The absolute refractive index is dependent on electrical and magnetic properties of the medium: $n = \sqrt{\varepsilon\mu}$, where ε and μ are the electric permittivity and the magnetic permeability, correspondently.

When a light wave propagating in a material with a given refractive index (n_1) falls onto a planar boundary of another material with a different index (n_2), the path of the light is redirected (fig. 1.4). This phenomenon is referred to as *refraction*.

The relation between the angle of incidence (α) and the angle of refraction (β) for the transmitted light is expressed by Snell's law:

$$\frac{\sin\alpha}{\sin\beta} = \frac{n_2}{n_1} = n_{21}, \qquad (1.3)$$

where α and β are the angles between normal to the boundary and the incident and transmitted light, correspondently (fig. 1.4); n_1 and n_2 are the absolute refractive indices of the two respective media, n_{21} is the relative refractive index. From Snell's law it follows that the light propagation direction is closer to the surface normal in the denser (higher refractive index) medium (in case of $n_1 < n_2$ or $n_{21} > 1$).

The relative refractive index is defined by the ratio: $n_{21} = \frac{n_2}{n_1} = \frac{\upsilon_1}{\upsilon_2}$, where υ_1 and υ_2 are the light speed magnitudes in the two respective media. Thus the relative refractive index is equal to the factor by which the velocity of light in the given medium is less (or larger) than that in the other one (the reference medium).

Important properties of the refractive index are as follows:

– $n \geq 1$; for example $n = 1$ for air, $n = 1.33$ for water, and $n = 1.4$–1.7 for many glasses;

– $n = n(\lambda)$; the refractive index is dependent on wavelength (*the phenomenon is called dispersion*);

– the refractive index is independent on the direction (or *isotropic*) for many amorphous materials (e. g. for water, amorphous glass), while for many crystals it depends on the wave propagation direction (e. g. $n_x \neq n_y \neq n_z$) and such materials are called optically *anisotropic*. The corresponding phenomenon is referred to as *birefringence (or double refraction).*

Total internal reflection. If the light is propagating from the denser medium to the less dense medium ($n_1 > n_2$), then there exists a critical maximum angle of incidence (α_{max}) when the angle of refraction $\beta = 90^\circ$. According to the equation (1.3) this angle α_{max} is given by:

$$\frac{\sin\alpha_{max}}{\sin 90^\circ} = \sin\alpha_{max} = \frac{n_2}{n_1} = n_{21}, \text{ for } n_1 > n_2. \qquad (1.4)$$

For incidence angles $\alpha > \alpha_{max}$ all light will be reflected back into the denser medium. This phenomenon is called *total internal reflection*.

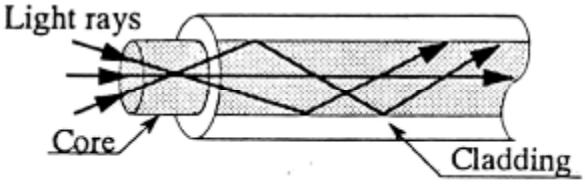
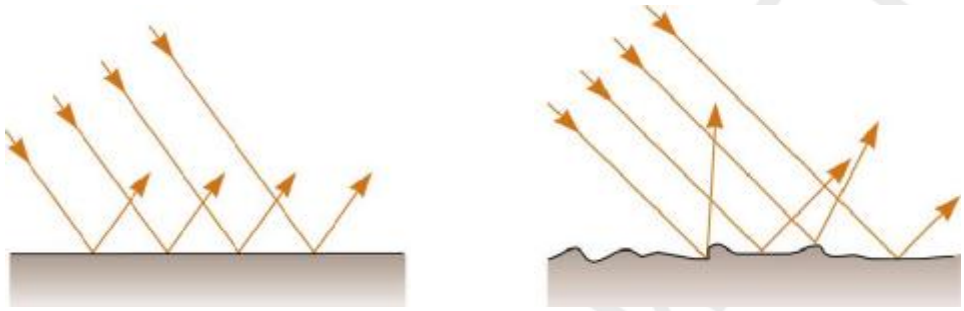


Fig. 1.5. Scheme of light propagation in optical fiber

The total internal reflection phenomenon is used in optical fibers which are widely applied in biomedical optics (fig. 1.5). The optical fibers are usually composed of a single optically transparent transmission element consisting of a cylindrical core with

cladding on the outside. The refractive index of the core is higher than the cladding to confine light in the core. Light travels through the fiber core, bouncing back and forth off the boundary between the core and cladding. Because the light must strike the boundary with an angle greater than the critical angle, only light that enters the fiber within a certain range of angles can travel along the fiber without leaking out. Optical fibers are used to transmit electromagnetic radiation over a long distance with minimum attenuation and disturbance.

Reflection. In optics, two main types of reflection are defined (*specular and diffuse reflection*) (fig. 1.6).



a ***b***

Fig. 1.6. Specular (*a*) and diffuse (*b*) light reflection

In case of *specular reflection*, the incident and reflected rays and the surface normal lie in the same plane. The angle of reflection (γ) is equal to the angle of incidence (α): $\gamma = \alpha$ (fig. 1.4). The amount of light reflected by a boundary depends on the refractive indices of the two materials and the angle of incidence.

For normal incidence onto a planar interface, the *specular reflection coefficient (or specular reflectance) R,* defining the fraction of light reflected (I_R) at the interface of two media, is given by the equation:

$$R = \frac{I_R}{I_0} = [(n_1 - n_2)/(n_1 + n_2)]^2, \qquad (1.5)$$

where n_1 and n_2 are the absolute refractive indices of the two media.

Diffuse reflection occurs when light strikes the rough surface of a material and bounces off in all directions due to multiple reflections. It arises from light scattering by the microscopic irregularities, the size of which is of the same order as the size of incident light wavelength (scattering phenomenon is discussed further in Part 3).

PART 2
LIGHT ABSORPTION

Light absorption is the phenomenon of the decrease in the intensity of light passing through the medium due to the partial energy transformation into other kinds of energy (such as heat) [1–3, 10, 11].

An intensity of light transmitted by an absorbing medium (I) is given by the Bouguer law (fig. 2.1):

$$I = I_0 e^{-\mu_a x}, \qquad (2.1)$$

where I_0 is the incident light intensity, I is the transmitted light intensity, x is the thickness of the absorbing medium, and μ_a is the absorption coefficient.

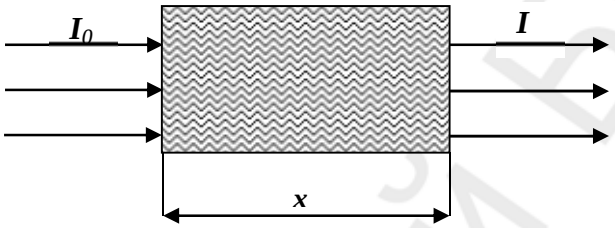


Fig. 2.1. Light passing through the absorbing medium

It is clear that the absorption coefficient characterizes the substance ability to transmit light. The larger the absorption coefficient the less is the ratio I/I_0 at the given absorbing layer thickness x.

The value of absorption coefficient depends on the substance properties and the incident light wavelength. In SI system the unit for μ_a is expressed in m^{-1}, but commonly in cm^{-1}.

The dependence of μ_a on the wavelength ($(\mu_a(\lambda))$ is called the absorption spectrum of the substance. Spectra are always individual. It allows use them for identification of substances (fig. 2.2). The difference of the spectra of aromatic amino acids shown in fig. 2.2 is evident. Moreover the absorption band maxima have their own individual values for every compound.

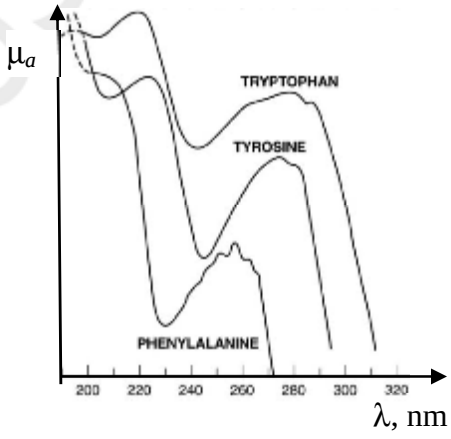


Fig. 2.2. Absorption spectra of aromatic amino acids: phenylalanine, tyrosine, and tryptophan

For solutions the Beer's law is applied. According to it the absorption coefficient of a compound μ_a is directly proportional to its concentration C in a non-absorbing medium, so that:

$$\mu_a = \alpha \cdot C, \tag{2.2}$$

where α is known as the molar absorption coefficient (specific absorption coefficient), usually expressed in $L \cdot cm^{-1} \cdot mol^{-1}$ (the units for concentration are mol/L). The molar absorption coefficient is dependent on the wavelength and the solute substance properties.

The Beer's law is valid only for dilute solutions; deviations from the law occur in concentrated solutions because of interactions between molecules of the solvent and the dissolved substance.

Substitution of μ_a in the Bouguer law gives the Beer–Lambert law:

$$I = I_0 e^{-\alpha \cdot C \cdot x}. \tag{2.3}$$

The Beer–Lambert law states that when a collimated monochromatic light beam passes through a solution in a transparent cuvette, the intensity of transmitted light depends on concentration of absorbing solution and the path length of absorbing medium.

Below the light absorption characteristics which can be measured using spectral instruments are given.

Transmittance (T) is the ratio of the light intensity transmitted by an absorbing medium to the incident light intensity:

$$T = \frac{I}{I_0}. \tag{2.4}$$

Absorbance (A) is the logarithm ratio of the incident light intensity to the transmitted light intensity:

$$A = \lg (I_0/I) = -\lg (T). \tag{2.5}$$

Further consider the relationship of these magnitudes with the solution concentration. Substitution of equation (2.3) into expressions (2.4) and (2.5), gives the dependence of the solution transmittance (T) and absorbance (A) on the concentration (C) of the dissolved substance:

$$T = \frac{I}{I_0} = e^{-\alpha \cdot C \cdot x}; \tag{2.6}$$

$$A = \varepsilon \cdot C \cdot x, \tag{2.7}$$

where ε is the molar extinction coefficient ($\varepsilon = \alpha \cdot \lg e = 0.43 \cdot \alpha$).

The molar extinction and absorption coefficients are conceptually the same, differing only by the logarithm base used in the Bouguer expression (2.1).

It is evident, that *transmittance decreases exponentially with the increase of the solution concentration C, while absorbance linearly depends on this parameter*. Linear dependence of absorbance (A) on C is applied to determine

unknown concentration by measuring absorbance. The corresponding method is called *photoelectric colorimetry* and the devices are termed as photoelectric colorimeters.

From the Bouguer law it follows that $A = k \cdot x$, where k is the extinction coefficient ($k = 0.43 \cdot \mu_a$). According to this expression the dependence of absorbance on wavelength ($A(\lambda)$), when the path length is kept constant, (x = const) is similar to the dependence $\mu_a(\lambda)$. Therefore the absorption spectra typically measured using spectral instruments represent the dependence $A(\lambda)$ defined by the equation (2.5).

PART 3
LIGHT SCATTERING

Light scattering, along with the reflection and absorption, is one of the most important physical processes determining visual perception of most objects. It influences greatly on the light propagation in biological tissues [1–3, 11, 12].

Light scattering is the deflection of a light beam from a straight path in different directions (fig. 3.1). It results from a spatial variation in the refractive index, i. e. inhomogeneities in the refractive index.

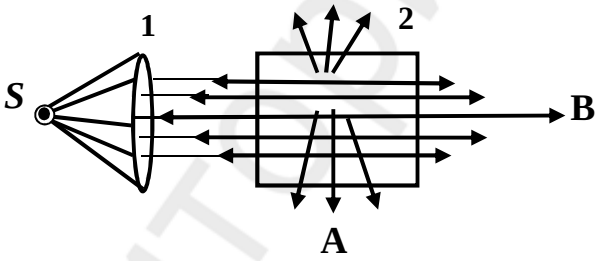


Fig. 3.1. Scheme of the light scattering observation in turbid medium:
S is the white (nonmonochromatic) light source; 1 — lens redirecting light into a beam of parallel raws; 2 — cuvette with scattering medium

Strong light-scattering media are called turbid (or cloudy). For example, smoke (solid particles in gas), fog (liquid droplets in gas), suspensions, emulsions can be referred to such kinds of media. For the first time, light scattering in turbid media has been experimentally investigated by the physicist John Tyndall and it has been called “Tyndall effect” (after his name).

In practice, scattering is classified into two main categories defined by the size of the scattering object relative to the wavelength (λ):

– *Rayleigh scattering (after the English physicist Lord Rayleigh)* refers to scattering by particles with a size (r) much smaller than λ ($r < 0.1\lambda$);

– *Mie scattering* (*after the German physicist Gustav Mie*) refers to scattering by particles with a size on the order of or larger than λ ($r > 0.1\lambda$).

Both above scattering mechanisms have a great impact on the optical properties of biological tissues.

The most important characteristic feature of *Rayleigh scattering* is strong dependence of the scattered light intensity on the incident light wavelength. The *Rayleigh law* states that scattered light intensity is inversely proportional to the fourth degree of wavelength (fig. 3.2, curve 1):

$$I \sim \frac{1}{\lambda^4}. \qquad (3.1)$$

According to the Rayleigh law blue beams (with shorter wavelengths) scatter more intensively than the red ones. Thus in the experiment (fig. 3.1) in case of lateral position (direction A), scattered light has a bluish shade while the light passing through the cuvette (direction B) has a reddish shade.

With the increase in the ratio of r/λ the wavelength dependence of the scattered light intensity gradually weakens. The ultimate case is the *Mie scattering* by large particles with the size of $0.8\lambda < r < 8\lambda$. It is characterized by weak dependence of the scattered light intensity (I) on the incident light wavelength (λ) (fig. 3.2, curve 2):

$$I \sim \frac{1}{\lambda^{0.37}}. \qquad (3.2)$$

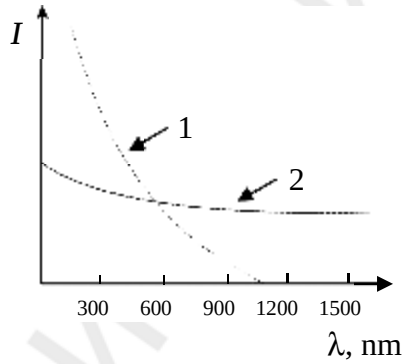


Fig. 3.2. Dependence of the intensity of Rayleigh (curve 1) and Mie (curve 2) scattering on the wavelength of incident light

Another important characteristic feature of the phenomenon under consideration is angular dependence of the scattered light intensity representing the graphic drawing of the dependence of the light intensity I on the scattering angle (θ) (scattering diagram) (fig. 3.3). The scattering angle is formed between the directions of the incident (commonly monochromatic) and scattered light. In case of Rayleigh scattering of unpolarized monochromatic light, the scattering diagram has the shape as shown in fig. 3.3 (curve 1). In such a case angular dependence of the scattered light intensity is expressed as follows:

$$I \sim 1 + \cos^2\theta. \qquad (3.3)$$

It can be seen that the scattering diagram has a symmetric shape both to the primary beam axis and to the line perpendicular to it (fig. 3.3, curve 1).

The intensity of forward and backward scattering is equal and twice as high as the scattered light intensity in the direction perpendicular to the primary beam axis.

Thus the Rayleigh scattered light is equally distributed in the forward and backward directions.

With the increase of the ratio r/λ, the asymmetry of the forward and backward scattering appears, and the forward scattering prevails in the direction of the incident light (fig. 3.3, curve 2). In this case, the scattering diagram retains the symmetry relative only to the primary beam direction.

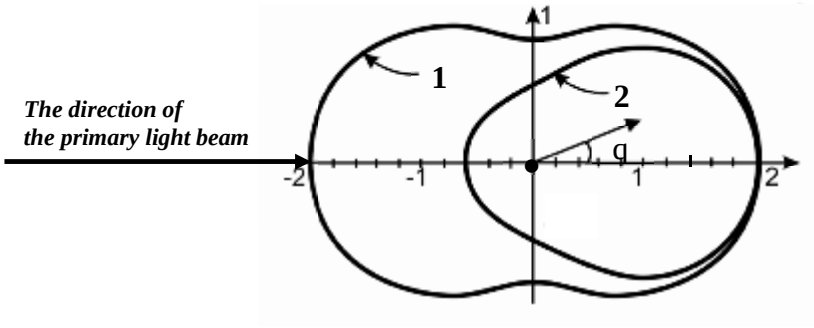


Fig. 3.3. Scattering diagram for the particles with size $r < 0.1\lambda$ (Rayleigh scattering) (curve 1) and $r \sim 0.25\lambda$ (Mie scattering) (curve 2). Dark circle indicates the position of the scattering particles with different sizes

With the further increase in the particle size, scattering asymmetry (mostly forward) increases and angular distribution of scattered light becomes more complicated. If the linear sizes of scattering particles larger than the wavelength ($r > \lambda$) secondary maxima appear on the scattering diagram, the form of which depends on the size and shape of the scattering particles and the nature of their environment (fig. 3.4).

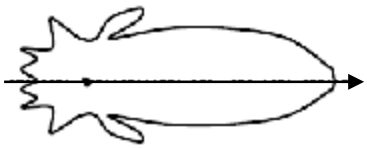


Fig. 3.4. Scattering diagram by particles with size $r > \lambda$

Thus Mie scattering is highly forward directed. The larger the particle, the more intense is the light scattered in the forward direction.

The light propagation in scattering media is often accompanied by absorption. The process of light propagation in such a case can be quantitatively described by definition of the following optical constants — *the absorption* μ_a (see Part II) and *scattering* μ_s *coefficients*.

Scattering coefficient μ_s determines the transmitted light intensity change caused by scattering (usually the unit for μ_s is expressed in cm^{-1}).

The intensity I_0 of a parallel (collimated) light beam normal to the surface of turbid medium is attenuated due to the absorption and scattering according to the Bouguer law:

$$I = I_0 e^{-\mu_t x}, \qquad (3.4)$$

where I_0 and I are the intensities of incident and transmitted light, respectively; $\mu_t = \mu_a + \mu_s$ is the attenuation coefficient; x is the medium layer thickness.

The process of light absorption (transmission) through the biological tissues is also characterized by the magnitude (l) inverse to the attenuation coefficient. It is called *the penetration depth:*

$$l = \mu_t^{-1}, \tag{3.5}$$

where μ_t is the attenuation coefficient.

From the equation (3.5) it follows that the penetration depth is equal to the distance over which the transmitted light intensity is reduced to approximately 37 % (or e^{-1}) of its initial value.

In fact, substitution for $x = l = \mu_t^{-1}$ in the equation (3.4) gives

$$I = I_0 e^{-\mu_t \cdot \frac{1}{\mu_t}} = \frac{I_0}{e} \approx 0.37 \cdot I_0 \ (e \approx 2.72).$$

Taking into account the specular reflection, the intensity of the transmitted light (I) is defined according to the Bouguer law:

$$I = I_0(1-R)e^{-\mu_t x}, \tag{3.6}$$

where I_0 is the intensity of incident light; μ_t is the attenuation coefficient; x is the medium layer thickness; R is the specular reflection coefficient (see Section 1.2). In case of normal direction of the incident beam the coefficient R can be expressed by the equation (1.5).

To determine experimentally the light absorption and scattering characteristics the absorption and reflection spectra are commonly used. They represent dependence of the absorption and reflection coefficients on the wavelength.

The above consideration is related to the cases of the so-called *single light scattering* which occurs when the scattering particles do not interact with the light waves scattered by the adjacent particles, and no re-emission occurs.

However, for many real scattering media, especially for biological tissues containing a large number of particles per unit volume, it is necessary to take into account the effects of *multiple light scattering*.

Study of the scattered light characteristics allows extract information about the structure of heterogeneous, (e. g. biological) media, the size of macromolecules and their aggregates, particles in colloidal solutions, etc. The method of optically inhomogeneous medium investigation on the basis of the scattered light intensity measurement is called *nephelometry.*

PART 4
OPTICAL PROPERTIES OF BIOLOGICAL TISSUES AND DENTAL MATERIALS

4.1. ABSORPTION AND SCATTERING PROCESSES IN HUMAN TISSUES

While propagating in tissues, electromagnetic radiation can undergo a number of processes, including reflection, absorption, scattering and fluorescence (fig. 4.1).

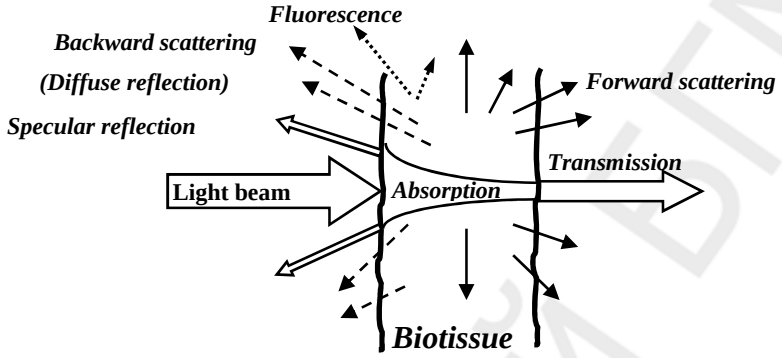


Fig. 4.1. Types of the electromagnetic radiation interactions with biological tissue

Fluorescence involves the re-emission of a photon of lower energy, i. e. larger wavelength than that of incident light.

In biomedical optics, *absorption processes are important for medical diagnosis and therapy*.

– *Diagnostic applications*: absorption spectra of biological tissues allows identify their structural components (e. g. see Part 2, fig. 2.2).

– *Therapeutic applications*: absorption of energy is the primary mechanism that allows laser light produce physical effects on tissues for treatment purposes.

There are many compounds in a biological tissue which absorb light radiation. These tissue components are known as *chromophores.* They have their own unique spectrum. However, the overall absorption of tissues does not display a simple relation to wavelength. Approximating tissue as a homogeneous mixture of compounds, the overall light absorption in tissues at a given wavelength, depends on the type and concentration of chromophores present.

In the optical wavelength region typical absorbers include DNA, aromatic amino acids (tryptophan and tyrosine) (fig. 2.2), proteins, hemoglobin, melanin, bilirubin, collagen and elastin.

In the IR spectral range (2–10 μm), the absorption spectra of biological tissues are essentially determined by the absorption of water (a fact easily understood since cells can consist of up to 85 % of water). It is known, that

water content varies with tissue type, age and is gender-dependent. It is worth noting that even within the so-called water transmission window (UV region to 930 nm) where the absorption of water is rather low, water still contributes significantly to the overall light attenuation in tissue because its concentration is very high in biological tissue.

In biological tissues, scattering processes are often dominant mechanisms affecting light propagation. Therefore light scattering is of special interest in both diagnostic and therapeutic applications.

– *Diagnostic applications*: scattering properties depend on the size, morphology, and structure of the tissue components (e. g., lipid membranes, nuclei, collagen fibers, enamel or dentin mineralization level, status of hydration in a tissue, etc.). Variations in these components due to a disease would affect scattering properties, thus providing a means for diagnostic purposes, especially in imaging applications.

– *Therapeutic applications*: scattering signals can be used to determine optimal light dosimetry (e. g., during laser-based treatment procedures) and provide feedback during therapy. For example, during laser coagulation of tissues, the onset of scattering is an observable end point that correlates with a desired therapeutic goal.

Light scattering in biological media is considered to be originated from spatial variations in the refractive indices and density of various structural components.

The refractive indices for the majority of biological tissues fall in the range of 1.335 to 1.620. For example, the index of refraction for melanin particles is approximately equal to 1.60. The refractive index of tooth enamel is around 1.62 and that of dentin is equal to 1.55.

In terms of scattering properties, human tissues can be categorized into two large classes:

1) *strongly scattering or turbid media* (i. e. “cloudy” or opaque) such as skin, brain tissue, dentin, vascular walls, blood, sclera. Turbid tissues represent heterogeneous structures and correspondingly have spatial variations in their optical properties;

2) *weakly scattering or transparent media* such as the cornea, lens and enamel.

The light scattering mechanism is determined by the ultrastructure of the tissue. A tissue ultrastructure extends from membranes to membrane aggregates, then to collagen fibers, nuclei and finally to cells (fig. 4.2).

Scatterers in biological tissues have dimensions in the range from several angstroms (10^{-10} m) to 20 μm, i.e. they can be larger, comparable or smaller than the wavelength of light. Accordingly, there are different types of scattering (ranging from Rayleigh to Mie scattering) for different tissue components.

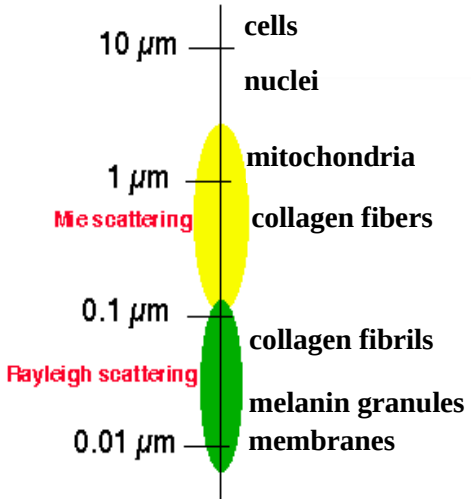


Fig. 4.2 Tissue ultrastructure and possible mechanisms of light scattering

Rayleigh scattering. Structures which yield Rayleigh scattering include cellular components such as membranes and cell subcompart-ments, and extracellular components such as the banded ultrastructure of collagen fibrils (fig. 4.2).

Mie scattering. Structures resulting in Mie scattering include various cellular structures, such as mitochondria and nuclei (fig. 4.2), and extracellular components like collagen fibers having sizes on the order of hundreds of nanometers to a few microns, and comparable in dimension to the wavelengths generally used in biomedical applications (0.6 to 1.3 μm). Being scatterers these structures fall in the Mie regime, exhibiting highly forward directed scattering.

The strongest scattering structures in biological tissues are, for example, red blood cells in blood; melanin granules and collagen fibers in skin; hydroxyapatite crystals in enamel; dentin tubules in dentin.

Anisotropy of optical properties. Biological tissues which have an aligned microstructure (muscle, ligament, tendon, skin, dentin, enamel, etc.) show pronounced anisotropy of optical properties. Here, *the term "anisotropy" states that the light propagation in the tissue is different for different directions relative to the aligned microstructure.*

The optical properties of biological tissues can be effectively controlled by various methods, for example, using various physical and chemical effects such as compression, stretching, dehydration, coagulation, UV irradiation, exposure to low temperature, and impregnation by chemical solutions, gels, and oils [2, 3].

4.2. Skin Optical Properties

Knowledge of the optical properties of the skin allows obtaining information about molecular structure, spatial distribution, concentration and structural configuration of various biological components in skin. It is of great importance for the diagnosis of different skin diseases, determination of the influence of different environmental factors (chemical, UV radiation, temperature, etc.) on the skin state, and the evaluation of the treatment effectiveness. Understanding the ways of light absorption and propagation in skin tissues can assist in the design of protective lotions from harmful solar radiation, and superior cosmetics.

Human skin is an inhomogeneous multilayered medium with strong scattering and absorbing properties.

It is usually described as consisting of three main layers. The outer layer is the epidermis of a 0.027–0.15 mm thick structure. Below it there is the dermis, the thickness of which is equal to 0.6–3 mm. It is made up of skin appendages such as hair follicles and sweat glands, surrounded by fibrous supporting tissue and collagen. Dermis contains blood vessels, the volume fraction of blood in it can vary, roughly in the 0.2–7 % range.

Finally, the underlying layer, called the subcutaneous tissue (hyperdermis or subcutis), contains fat-producing cells and fibrous tissue. The outermost part of the epidermis is the corneal layer.

Melanocytes containing in epidermis produce melanin, the protective pigment responsible for the skin colour. Melanocytes are stimulated by sunlight and therefore produce melanin to protect the skin against harmful UV radiation.

Processes of light interaction with skin is shown schematically in fig. 4.3.

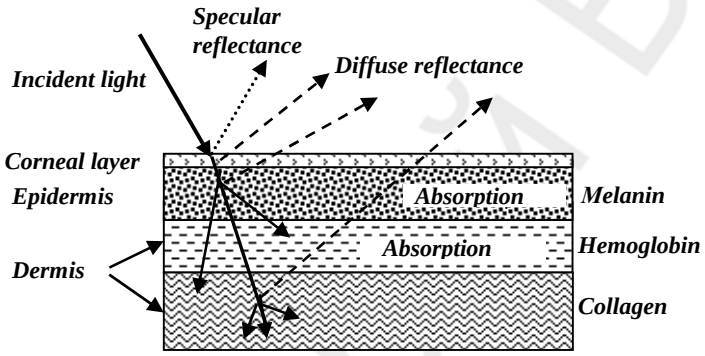


Fig. 4.3. Processes of light interaction with skin

Absorption properties of the skin. The dominant chromophores within epidermis and dermis are melanin and hemoglobin, correspondently (fig. 4.4).

Melanin exhibits a broad strong absorption in the visible range, which decreases at longer wavelengths.

As it was mentioned above, melanin is produced by cells called melanocytes found in melanosomes. The melanin absorption level depends on the melanosome density in the epidermis. Typically, the volume fraction of the epidermis occupied by melanosomes varies from 1.3 % (light pigmented specimens) to 43 % (dark pigmented specimens)

Hemoglobin molecules within the red blood cells (erythrocytes) carry 97 % of oxygen in blood, while the remaining 3 % is dissolved in the plasma. Each hemoglobin molecule contains four iron-containing hemo groups as well as the protein globin. Oxygen atoms easily bind to the iron, causing the hemoglobin to assume different structures. Hemoglobin in the oxygenated state is referred to as *oxyhemoglobin* (HbO_2); in the reduced state, it is called *deoxyhemoglobin* (Hb). In the arteries, 90–95 % of hemoglobin is oxygenated, while in the veins, more than 47 % of the hemoglobin is oxygenated. These two types of hemoglobin, both oxygenated and deoxygenated, have slightly

different absorption spectra, particularly in the red and near infrared regions (fig. 4.4). Oxygenated hemoglobin is a strong absorber (up to 600 nm); then its absorption drops off very steeply, by almost two orders of magnitude, and remains low. The absorption of deoxygenated hemoglobin, however, does not drop dramatically; it stays relatively high, although it decreases with increasing wavelengths. The isobestic point where the two spectra intersect occurs at about 800 nm.

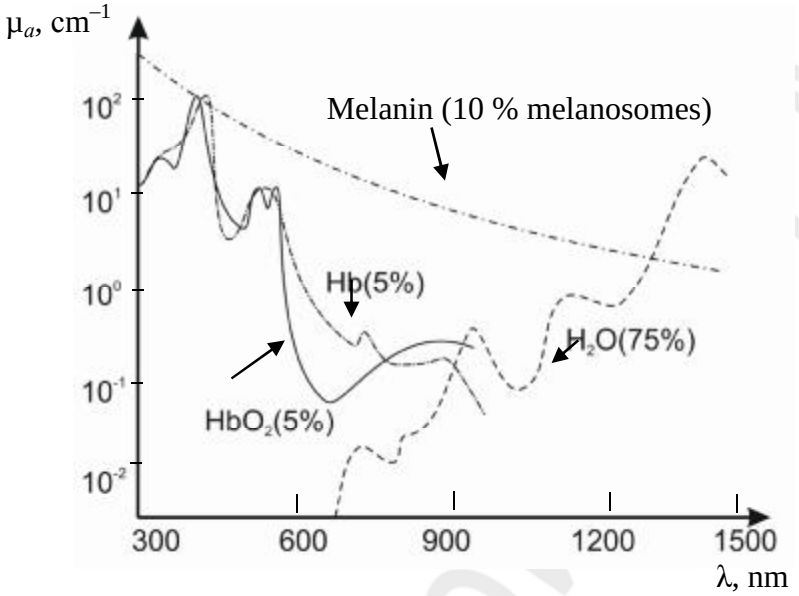


Fig. 4.4. Absorption spectra of the dominant chromophores in human skin (melanin, oxyhemoglobin (HbO_2), deoxyhemoglobin (Hb), and water (H_2O). The volume fraction of chromophore content in the skin sample in percents

The above difference in the absorption of hemoglobin and oxyhemoglobin can be used for determination of the hemoglobin transformation into the oxygenated form and vice versa.

Hemoglobin oxygen saturation measurements in biological tissues are also based on the significant difference in the absorption coefficients of hemoglobin and oxyhemoglobin, particularly, in the 600–700 nm range.

The electromagnetic radiation in the IR region ($\lambda > 2$ μm) is practically entirely absorbed by water containing in skin.

Scattering properties of the skin. The skin is characterized by strong light scattering properties. The main scattering structures in epidermis are melanin granules approximately from 10 to 30 nm in size. In dermis, collagen fibers (approximately 2.8 μm in the diameter and cylindrical by the form) are responsible for Mie scattering, while smaller scale collagen fibers (fibrils) and other microstructures are responsible for Rayleigh scattering (fig. 4.2). Because epidermis is less thick than dermis, scattering in the epidermis is not so much as dermal scattering. Dermal tissue is responsible, to a great extent for the majority of light scattering that takes place in skin.

Light gets scattered multiple times inside the dermis before it is either propagated to another layer or absorbed (fig. 4.3). This means that the spatial distribution of the light scattered within the dermis quickly becomes diffuse. Backscattered light from the dermis is found to be diffuse. It is worth noting that with the increase of the light penetration depth in skin, the fraction of the backscattered light increases due to multiple scattering within the skin.

The above processes of light absorption and scattering by the skin structural elements determine the experimentally measured *diffuse reflectance spectra* which represent the dependence of the reflection coefficient on the wavelength.

Optical diagnostic techniques of the skin are based on the reflectance coefficient measurements in certain spectral intervals. Typical applications of the reflectance spectra include quantitative evaluation of erythema[1] and melanin pigmentation of skin, determination of hemoglobin concentration and the blood oxygenation, evaluation of the skin colour changes, investigation of the photo protective and cosmetic remedies application effectiveness, etc.

Diffuse reflectance spectra. The ratio between absorption and scattering of light in the epidermis and dermis determine the shape of the reflectance spectra (fig. 4.5). The contribution of these processes into the formation of the diffuse reflectance spectra is different for various spectral regions.

Typical diffuse reflectance spectra of human skin are shown in fig 4.5. It is evident that they are individual for different types of skin.

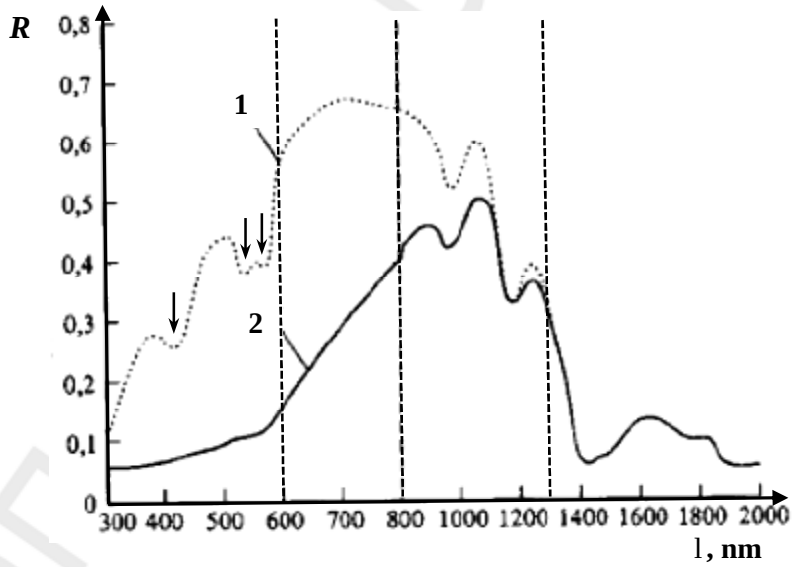


Fig. 4.5. Diffuse reflectance spectra of human skin in vivo:
1 — white skinned; 2 — black skinned [2]

[1] **Erythema** (from the Greek *erythros*, meaning red) means redness of the skin caused by hyperemia of the capillaries in the lower layers of the skin.

Below characteristic features of a white skin reflectance spectrum (fig. 4.5, curve 1) are analyzed.

In the UV range (300–400 nm), the shape of the reflectance spectrum is determined by strong absorption of epidermal chromophores. In such a case a contribution of scattering processes is comparatively small.

In the visible spectral range (400–600 nm) both absorption and scattering determine the formation of the reflectance spectrum. It has the characteristic minima in the 415–430 nm and 540–580 nm regions caused by the hemoglobin absorption (fig. 4.5). The extent of their appearance is increased in skin erythema. This fact allows carry on the quantitative evaluation of the erythema stage using the reflectance spectra.

In the 600–1300 nm range, scattering dominates absorption, the reflectance coefficient may run up to 70 % ($R \approx 0{,}7$). The spectrum in the given range is significantly affected by melanin absorption. Thus, by comparison of the reflectance spectra in fig. 4.5 (curves 1 and 2) it is evident that in case of heavily pigmented skin, a considerable decrease in the skin reflectance coefficient is observed due to the increase in melanin absorption. The methods of skin melanin pigmentation indices determination are based on this evidence.

Decrease in the reflectance coefficient at $\lambda > 1300$ nm is due to the significant increase in absorption by water containing in skin.

For the diagnostic and therapeutic applications it is of great importance to know the light depth penetration into the skin (fig. 4.6).

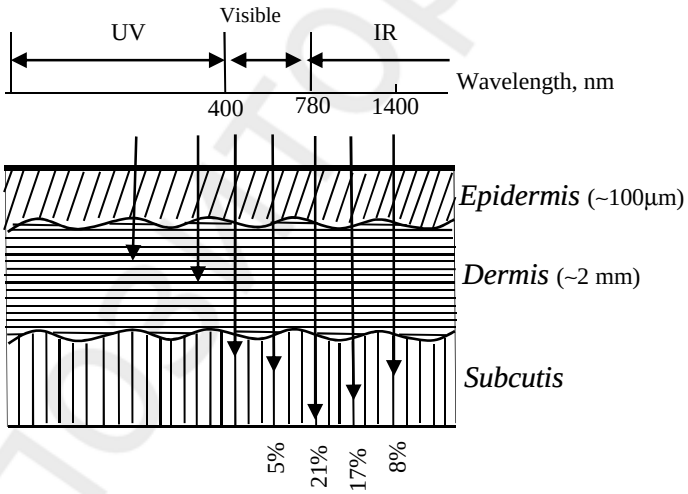


Fig. 4.6. Wavelength dependence of light transmission by human skin

The light penetration depth. The ability of light to penetrate into tissues depends on their absorption and scattering properties. From the above consideration it follows that in the UV region (at $\lambda < 300$ nm), the dominant process is the light absorption by melanin and in the IR region (at $\lambda > 2$ μm), the penetration depth is limited by the water absorption properties. As a result, the optical penetration depth in these regions is estimated to be less than 100 μm.

As the melanin and hemoglobin absorption is decreased, the penetration depth rises with the wavelength increase (fig. 4.6). Tissue absorption continues to decrease up to about 1300 nm, and then it rapidly increases again, mainly as a result of light absorption by water. The maximum penetration depth into the skin is observed at the wavelength of $\lambda \approx 800$ nm.

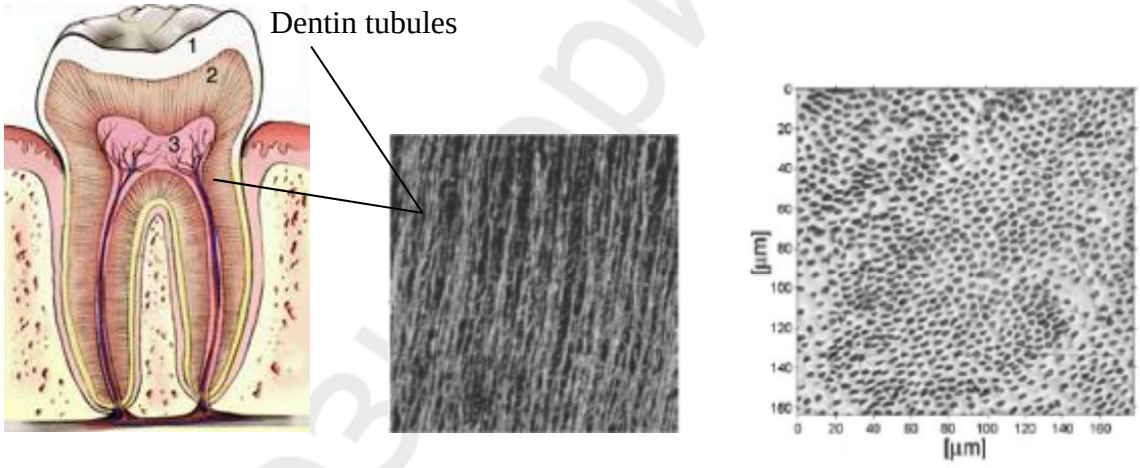
Within the spectral region known as *the therapeutic window (or diagnostic window)* the most significant penetration of light into skin is observed. This window extends from 600 to 1300 nm (fig. 4.6). It is worth noting that in the given wavelength region scattering is dominant over absorption, and so the propagating light becomes diffuse.

The electromagnetic radiation within the therapeutic window is applied in the photodynamic therapy and optical tomography techniques [2] [1–7].

4.3. OPTICAL CHARACTERISTICS OF HARD TOOTH TISSUES

In dentistry, understanding of light propagation in teeth is important not only for therapeutic laser applications or diagnosis, but also for the precise matching of the visible appearance of restorative materials and teeth.

Optical properties of teeth are related to their main structural elements (enamel and dentin) (fig. 4.7).



a ***b*** ***c***

Fig. 4.7. Tooth structure including dental tubules (*a*), 1 — enamel, 2 — dentin; 3 — pulp; laser scanning (*b*) and optical (*c*) microscope picture of the tubules (extensions of the picture $\approx 90 \times 90$ μm) (*c*) [13]

The outer part of a tooth crown consists of enamel. The enamel layer (thickness of 1–3 mm) is the only part of the sound tooth that is in direct contact with the oral cavity.

Enamel represents an ordered array of rods (or prisms) of inorganic apatite-like crystals (carbonated hydroxyapatite), which are roughly

[2] Optical tomography is a biomedical imaging technique that uses light as the probing radiation.

perpendicular to the tooth surface. These prisms are 4 to 6 μm wide and extend from the enamel-dentin junction to the outer surface of the tooth. The crystals are approximately 15 to 40 nm in the diameter and can be as long as 20 μm. Prisms are surrounded by a protein/lipid/water matrix.

Enamel contains minerals (85 %), water (12 %), and organic compounds (protein and lipid) (3 %).

Dentin can be described as a conglomerate of several compartments. It contains long tubules surrounded by the peritubular dentin. Between the tubules with their peritubular dentin lies intertubular dentin.

Intertubular dentin, in its turn, is divided into collagen fibrils and interfibrillar compartments. Except for the tubules all compartments contain mineral crystals of hydroxyapatite, which are needle shaped with an ~ 5 nm thickness and an ~ 20 nm length. The tubules have the diameter of 1 to 5 μm, and its density is 15 000–75 000 tubules per mm^2 [13]. They are uniformly oriented from the enamel-dentin junction to the pulp, and so in a small sample they lie more or less parallel (fig. 4.7).

Dentin is a bone-like material containing minerals (46 %), organic compounds (mostly collagen) (32 %) and water (22 %).

Process of light propagation in hard tooth tissues can be represented by the following model (fig. 4.8).

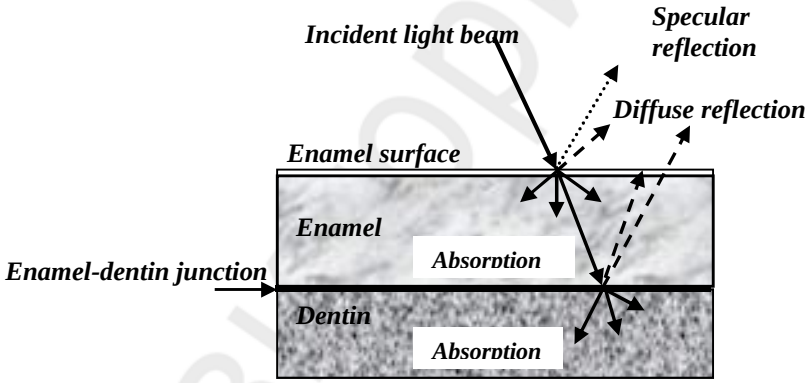


Fig. 4.8. Light propagation model in hard tooth tissues

Light is transmitted through enamel and dentin to the pulpal surface. Enamel and dentin together collect and distribute light within the tooth.

The mechanism of light interaction with hard dental tissues is inherently complex and varies markedly with the radiation wavelength and the nature of the primary absorbing and scattering structural elements in the tissue (e. g. water, proteins, or minerals, etc.).

The incident light is partly reflected by the enamel surface (specular reflection), while the remaining part of the light beam passes through the surface layer deep into enamel (fig. 4.8). Inside enamel the light is weakly absorbed and scattered forming diffusely reflected light flux. The remaining portion of light is absorbed and scattered inside dentin. Thus attenuation of

light within the tooth is caused by several different factors, but primarily by scattering and absorption.

Light absorption in enamel and dentin. *Enamel.* In enamel, the organic (protein) components are responsible for the light absorption in UV and visible spectral regions. Enamel absorption is very weak in the visible range ($\mu_a \sim 1\ cm^{-1}$) and increases in the ultraviolet range ($\mu_a \sim 10\ cm^{-1}$ at λ = 200–240 nm) (fig. 4.9). Owing to this fact teeth naturally fluoresce upon irradiation with UV light.

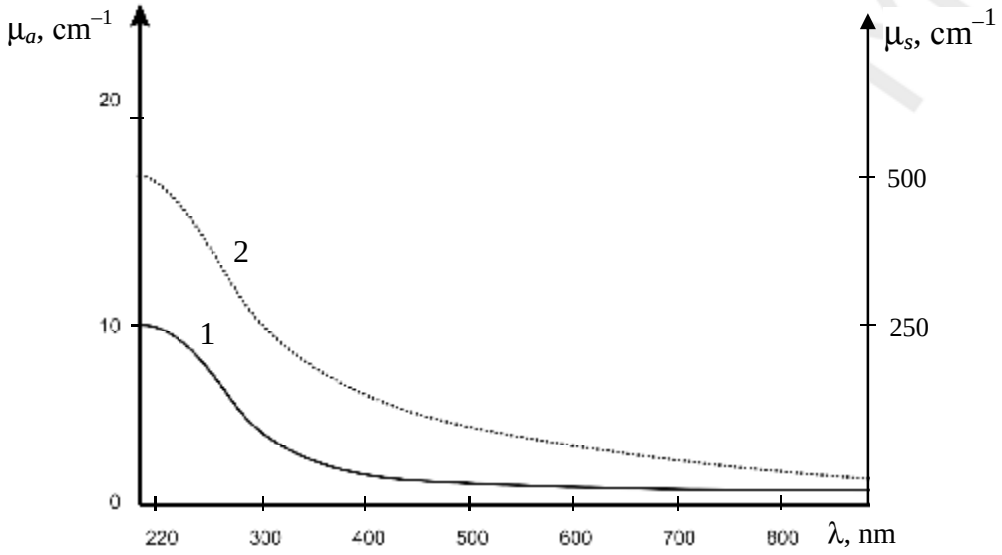


Fig. 4.9. The absorption coefficient (μ_a) (curve 1) and scattering coefficient (μ_s) (curve 2) of dental enamel in dependence on light wavelength [14]

The absorption coefficient is decreased with the wavelength increase (fig. 4.9). Therefore yellow and red light (560–760 nm) is highly transmitted by normal enamel, and thus can be reflected from the enamel-dentin junction and transmitted back through enamel to the surface, greatly effecting the tooth colour. The scientific foundations of the modern tooth lightening concept rest largely on reducing transmission of yellow light by enamel.

Dentin. Collagen fibers are the dominant absorption structures in dentin. In the visible wavelength range, the dentin absorption is essentially wavelength independent, with a value of absorption coefficient which is several times higher than that of enamel ($\mu_a \sim 4.0–6.0\ cm^{-1}$).

It is worth noting that the magnitude of the absorption coefficients for dental hard tissues is high in the mid-infrared range due to resonant absorption by molecular groups containing in water, protein, and mineral. Thus, absorption coefficients corresponding to erbium (Er:YAG) laser line (λ = 2.94 μm) were determined to be $\mu_a = 800\ cm^{-1}$ (enamel) and $\mu_a = 2200\ cm^{-1}$ (dentin) [14]. These high absorption coefficients indicate that in this case the laser light will penetrate into hard tooth tissues to a very small depth, allowing to apply laser ablation of dental hard tissues.

Light scattering in enamel and dentin. *Enamel.* Enamel does not contain pigments, therefore diffusely reflects all incident light wavelengths and has white colour.

As in case of Rayleigh scattering, strong wavelength dependence of the scattered light intensity is specific to enamel. According to the law $I \sim \lambda^{-3}$ scattering in enamel is strong in the near-UV and decreases with the increase of wavelength (fig. 4.9, curve 2). Therefore enamel scatters shorter wavelengths of visible light (blue and violet) much better than longer wavelengths (yellows and reds) and sound teeth have a bluish shade. Rayleigh scattering of visible light by enamel is quite high. Therefore enamel plays a significant role in modifying tooth colour by scattering the shorter wavelengths in the visible blue range, while colour of teeth is determined mainly by dentin.

It is also necessary to take into account that an internal play of colours (the so-called opalescence[3]) is observed in enamel that is due to the strong light scattering caused by the enamel liquid filling micropores and microscratches.

According to the above absorption and scattering properties, enamel looks bluish when observed in reflection and yellowish when observed in transmission.

In enamel, it is the hydroxyapatite crystals themselves which contribute significantly to scattering, rather than the prism structure. For this reason, dissolution of crystallites or their incomplete formation will affect the light scattering and consequently, the tooth colour. The more the enamel scatters blue light, the lighter it appears. The more porous the enamel, the less it scatters short (blue) wavelengths of light, and as a result the porous enamel looks opaque.

For example, roughness of enamel surface (presence of voids, scratches etc.) leads to the increase in diffuse light scattering, and the tooth looks more opaque and whiter.

In carious enamel the differences in refractive indices of the structures present are increased, resulting in much higher scattering coefficients than in sound enamel. Therefore sound enamel is rather transparent, while carious enamel is opaque. Increased backscattering from the demineralized region of early caries lesions is the basis for the visual appearance of white spot lesions.

Increased porosity of the lesion leads to increased scattering at the lesion surface and higher scattering in the body of the lesion, producing an increase in the magnitude of the diffusely reflected light flux. This evidence can be basis for the caries diagnosis optical techniques [8].

The influence of water present in enamel. The water affects the ability of enamel to both absorb and scatter light. For early carious lesions and other enamel conditions with increased subsurface water, longer wavelengths of

[3] Opalescence is the phenomenon of great growing light scattering by pure substances, and also by solutions of liquids or gases due to the large density fluctuations.

yellow and red are mainly reflected back to the surface. This radiance change is mainly determined by the lack of mineral and the increased water content in these lesions. Even without the white spot lesions development or altered enamel formation, microscopic subsurface water voids are common as enamel is never perfectly mineralized. These water-rich areas can undergo subsurface remineralization by repeated topical application of some special paste. Restoring mineral should, therefore, change the scatterring of short wavelengths of light and thereby alter the appearance and colour of the enamel surface. Micropolishing the enamel surface makes tooth shade lighter and less yellow, and this effect is enhanced upon air-drying, which removes subsurface moisture. This provides a rationale for micropolishing in combination with subsurface remineralization.

As it is pointed out above the light penetration depth is determined by the magnitude which is inversely proportional to the sum of the absorption and scattering coefficients (3.5). The estimation of the light penetration depth to enamel indicates that the dominant portion of the visible light penetrates into dentin.

Dentin. In daily light dentin has yellow colour. The ability of selective light reflection by dentin is due to the pigments, containing in its structure.

Structural inhomogeneity and complex composition of dentin cause considerably higher values of scattering coefficients in comparison to enamel.

In contrast to enamel, scattering in dentin is high throughout the near-UV to near infrared range (μ_s ~ 250–280 cm^{-1}) [14]. It is scattering light in dentin that determines the tooth opacity.

The main scatterers in dentin are the tubules the diameter of which is approximately 1000 times higher than hydroxyapatite crystal transverse sizes. The collagen fibrils play only a minor role, and the mineral crystals do not contribute significantly to the scattering process. This is confirmed by the results demonstrating a rather small wavelength dependence of the scattering, which is not much higher at shorter wavelengths as compared with longer wavelengths (fig. 3.2, curve 2). Such kind of wavelength dependence points at large scatterers (Mie scattering). Also, near the enamel-dentin junction the scattering is low (low tubule density and small tubule size).

As the tubules sizes larger than the incident light wavelength, then light is scattered in dentin mostly in forward direction (along with the dentine tubule direction oriented perpendicular to the tooth surface) (fig. 4.7, *a*), i. e. from the enamel-dentin junction to pulp.

Demineralization leads to the tubule transverse cross-section increase which causes the change in the dentin scattering properties.

As in case of enamel, in carious dentin light scattering from demineralized region is increased. Carious lesions in their earliest stage can be observed as white spots. The increased whiteness of these spots is caused, as it has been

mentioned above, by increased light scattering. This evidence can be applied in dentin caries diagnosis techniques.

Dentin and enamel are optically anisotropic, i. e. their optical properties depend on the light propagation direction.

In case of dentin, it has long been known that anisotropic light propagation is especially pronounced. It was shown that disks of dentin, taken from extracted human teeth, demonstrated the optical property of the image magnification/reduction or that light is predominately conducted from the enamel-dentin junction to the pulp [15]. As it follows from the above consideration, the light guiding effect in dentin is caused by scattering on its aligned (cylindrical) microstructure (fig. 4.7).

A light beam bends, when it moves through dentin, and light propagation in human dentin exhibits a strong directional dependence. Thus the scattering distributions depend on tissue orientation relative to the irradiating light beam in addition to the polarization of the incident light. This phenomenon is demonstrated in figure 4.10.

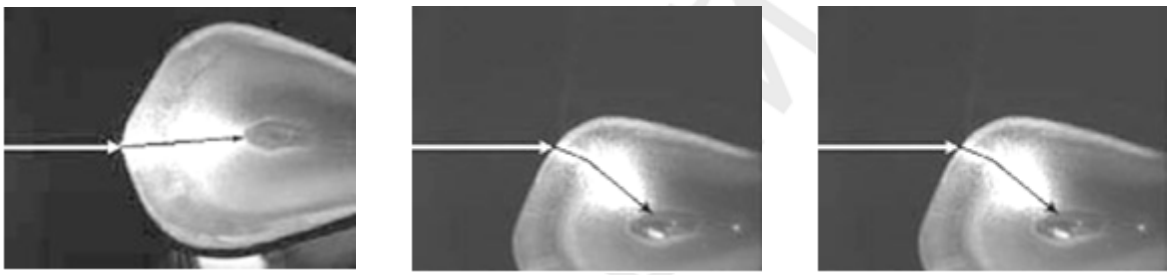


Fig. 4.10. Light propagation in hard tooth tissues in dependence on the He-Ne laser beam orientation (white arrow) with respect to the tubules direction (dark arrow) [16]

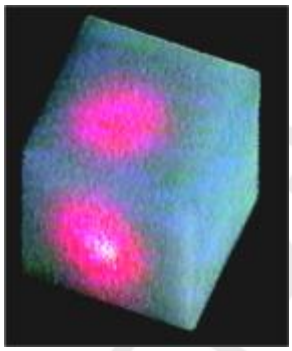


Fig. 4.11. Image of a cube of dentin illuminated by He-Ne laser [17]

Other effect of pronounced anisotropic light propagation in dentin is illustrated in fig. 4.11. When a cube of dentin is illuminated perpendicularly at the center of a certain plane (the front plane in the picture), almost all light is transmitted from one of the lateral planes (the upper plane in the picture) or re-emitted from the incident plane. Much less light is transmitted from the other planes.

Enamel and dentin exhibit double refraction or birefringence. Birefringence in hard dental tissues originates from mineralized structures formed by the hydroxyapatite crystals, i.e. an ordered array of enamel prisms and the uniformly oriented cylindrical dentin tubules.

To summarize everything up, two main optical properties of enamel and dentin, namely absorption and scattering together determine the visual appearance, i. e. the tooth crown colour.

Under daily conditions the tooth colour is mainly determined by dentin. But enamel plays a role in the total appearance of tooth. This appearance changes with age.

When teeth become older their colour will change, teeth in elderly people are on average more yellow than those in younger ones. Firstly, it is believed that dyes from food, drinks, and tobacco diffuse into enamel causing some discolouration. Secondly, during its maturation enamel may become more homogeneous, leading to less light scattering and an increased transparency of the enamel. This is observed as a relative higher absorption (a more saturated color of the enamel). The increased transparency of enamel may also result in larger light reflection by the more coloured dentin. Thus the relationship between the scattering, absorption characteristics and age should be taken into account.

4.4. OPTICAL PROPERTIES OF MODERN RESTORATIVE MATERIALS IN DENTISTRY

One of the biggest challenges in restorative dentistry is matching between restorative material and the surrounding dental tissues. Nowadays restorations perfectly matching the tooth structure have been designed.

The ability of a composite to reflect, absorb and scatter light similarly to the natural tooth makes it possible to achieve a perfect matching the surrounding tooth structure. An excellent so-called “chameleon effect” can thus be achieved, resulting in aesthetically invisible restorations. The term ”chameleon effect” refers to the perception that dental materials and hard dental tissues practically coincide not just in colour but also in shades.

Such materials allow imitate even such “thin” optical effects as optical anisotropy, opalescence, fluorescence, etc.

This excellent ability of dental materials is related to their extremely diverse structural composition, which results in reproducing the optical properties of natural teeth.

Dental composites typically consist of a resin matrix, filler materials and a coupling agent. The amount of each filler, its size and distribution within the composite is carefully chosen so that being combined they reproduce optical characteristics of natural teeth.

Nowadays nanohybrid composites with a chameleon effect are becoming popular among dentists because they have superior esthetic, mechanical and strength characteristics. Typical example of such kind of nanohybrid composites is G-ænial. The types of the fillers of the given composite and their size are shown in fig. 4.12.

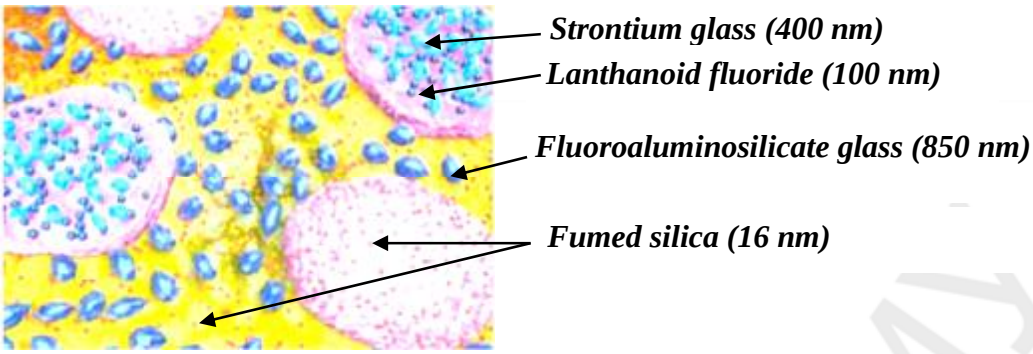


Fig. 4.12. Structural drawing of the G-ænial filler system [18]

Depending on different factors, including age, smoking, food type etc., optical properties of teeth exhibit different changes. However, through improvements in formulations, optimization of physical and optical properties, and the development of new placement techniques, today's composite resins even in given cases match rather perfectly optical properties of natural teeth and enable reliable and predictable realization of esthetic outcomes.

CONCLUSION

The intent of the author was to offer the information about the influence of the human tissue microstructure on light propagation as it is essential for the effective light applications in medical diagnosis and therapy.

From the therapeutic point of view, in case of photodynamic therapy and laser-induced thermotherapy, optical biopsy, etc. the optical properties in combination with light propagation models are necessary for dosimetry and prediction of treatment outcome. In laser-induced fluorescence diagnostics, they provide information concerning the nature of the fluorescence signal.

Knowledge of the hard dental tissue optical properties and their origin is also of special interest for the development of new optical methods in caries diagnosis and design of composites for esthetic restorations in dentistry.

Author would like to thank Associate Professors N. I. Insarova, PhD., A. A. Ivanov, PhD., Dr. A. N. Korolevich for helpful discussions and valuable suggestions and I. I. Tikhonovich for technical assistance in medical English.

ISSUES FOR DISCUSSION

Part 1. Electromagnetic nature of light

1. The nature of electromagnetic radiation. Electromagnetic wave characteristics and the relationship between them.
2. Electromagnetic spectrum wavelength regions.
3. Light reflection and refraction phenomena.

Part 2. Light absorption

1. The light absorption phenomenon definition.
2. The physical sense of the Bouguer and Beer-Lambert laws.
3. Transmittance and absorbance, their relationship with the solution concentration.
4. Physical basis for photoelectric colorimetry.

Part 3. Light scattering

1. The light scattering phenomenon definition and its mechanisms.
2. The scattered light intensity dependence on the wavelength.
3. The scattering diagram shape variations in dependence on the scattering particle size.
4. Physical basis for nephelometry.

Part 4. Optical properties of biological tissues and dental materials

1. The main optical characteristics applied for quantitative description of light propagation in biological tissues.
2. The quantitative description of transmitted light intensity attenuation in biological tissues. The physical sense of the penetration depth.
3. The typical chromophores in biological tissues.
4. The main scattering mechanisms in biological media and structural components determining them.
5. The dominant chromophores in skin and their absorption spectra.
6. The main scattering structural elements in skin.
7. The reflectance spectra of skin. The therapeutic window and its meaning for light application in medicine.
8. Light absorption characteristics of enamel and dentin.
9. The light scattering mechanisms in enamel and dentin and the main structural components determining them.
10. The sense of the term “chameleon effect” in creating esthetic restorations in dentistry.

PROBLEMS

1. Visible light has a wavelength range of about 400–780 nm. a) Calculate the corresponding frequency range. b) Calculate the corresponding energy range in Joules.

2. Consider a monochromatic light beam propagating from air to glass. If the speed of the beam in glass is $\upsilon = 1.97 \cdot 10^8$ m/s, and the angle of incidence is $\alpha = 60^o$, calculate the angle of refraction (β). What is the refractive index of glass for this monochromatic light?

3. The typical value of the refractive index for the optical fiber cladding is 1.52. The refractive index for core is typically 1.62. Find the critical angle of incidence for the optical fiber. What is the speed of light propagation in the optical fiber?

4. Absorbance of solution is decreased from value A_2 to A_1 so that $\Delta A = A_2 - A_1 = 2$. How many times is the transmitted light intensity changed?

5. Express the absorbance value A = 2 in terms of percent transmittance (T %).

6. The intensity of light beam is reduced by 4 times while passing through the solution. Calculate the solution transmittance (T %) and absorbance (A) values.

7. Light beam travels through the glass plate. Find the glass plate thickness providing the transmitted light intensity is reduced twice. Assume the absorption coefficient of a glass plate equal to 0.5 cm^{-1}.

8. If the transmittance is 50 % in a 1,00 cm cell, what is the absorbance in a 3.00 cm cell?

9. Solution of tyrosine is placed in a 1.00 cm cuvette. a) What is the concentration of the solution if the solution absorbance is found to be 0.850 at 280 nm. b) What will be the absorbance if the path length of the original solution is increased to 2.00 cm? Assume the value of molar extinction coefficient (ε) for tyrosine equal to about 1280 $L \cdot mol^{-1} \cdot cm^{-1}$ at 280 nm.

10. How many times is the scattered intensity of blue light (λ = 480 nm) by enamel larger than that of red one (λ = 780 nm)?

11. An erbium laser beam at $\lambda = 2.94$ μm is applied for ablation of hard tooth tissues. Calculate the thickness of enamel and the dentin layer over which the intensity of an erbium laser beam is attenuated twice due the absorption. Assume the absorption coefficients of hard tooth tissues at the given wavelength equal to $\mu_a = 800$ cm^{-1} (enamel) and $\mu_a = 2200$ cm^{-1} (dentin).

12. Monochromatic yellow light beam travels through a tooth. Calculate the fraction of the incident light beam which passes through the enamel layer with the 0.1 mm thickness. Assume the attenuation coefficient of enamel for yellow light (λ = 560 nm) equal to about 80 cm^{-1}.

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CONTENTS

Учебное издание

Никоненко Наталия Анатольевна

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Компьютерная верстка Н. М. Федорцовой

Подписано в печать 28.11.13. Формат 60×84/16. Бумага писчая «Снегурочка».
Ризография. Гарнитура «Times».
Усл. печ. л. 2,09. Уч.-изд. л. 1,81. Тираж 50 экз. Заказ 743.

Издатель и полиграфическое исполнение:
учреждение образования «Белорусский государственный медицинский университет».
ЛИ № 02330/0494330 от 16.03.2009.
Ул. Ленинградская, 6, 220006, Минск.