

Lasers in Medicine and Surgery

2021 Edition



Laser and Fiber Optics Educational Series

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About LASER-TEC

LASER-TEC is a not-for-profit National Center for Laser-Photonics and Fiber Optics Education. It is funded by the National Science Foundation to develop a sustainable pipeline of qualified laser and fiber optics technicians to meet the industry demand across the United States.

To learn more about LASER-TEC, visit www.laser-tec.org.

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Preface

About the LASER-TEC Laser and Fiber Optics Educational Series

This series was created for use in engineering technology programs such as electronics, photonics, laser electro-optics, and related programs. This series of publications has three goals in mind: 1) to create educational materials for areas of laser electro-optics technology in which no materials exist, 2) work with the industry to use, adapt, and enhance available industry-created material, and 3) ensure that these materials are available to technicians at no cost to them which will in turn make education in these areas more accessible to everyone. The Laser and Fiber Optics Educational Series is available for free online at www.laser-tec.org.

About Lasers in Medicine and Surgery

Lasers in Medicine and Surgery was created to provide a fundamental background for technicians on the theory of laser-tissue interactions as well as to describe commonly used lasers in medicine, optical delivery systems, and accessories frequently used with lasers. Numerous medical fields have been impacted by the use of lasers, including cosmetic dermatology, ophthalmology, dentistry, urology, and many others. Lasers are used for thermal coagulation of tissues as well as ablation and vaporization of both soft and hard tissues.

To the Student

This book is written at the technician level and can be used in post-secondary electronics engineering technology or related programs. Medical laser technology is used for minimally invasive surgical applications in a variety of medical fields. The book contains all the modern pedagogy. This includes the following sections: introduction-motivation, learning outcomes, self-test questions for each section, summary, glossary, bibliography, and colorful illustrations.

To the Instructor

This book is intended for certificate or associate degree program use in electronics engineering technology. This will not only update the course content but will also provide the student with the latest skills that the industry expects. A PowerPoint presentation and a test bank are available by sending a request using an official college email to info@laser-tec.org.

Acknowledgments

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Lasers in Medicine and Surgery

Introduction

Since the invention of the first laser in 1960, lasers have moved from the realm of the science fiction “death ray” to an everyday and often indispensable part of the medical/surgical armamentarium in practically every medical specialty. The unique qualities of laser energy are being utilized in an ever-growing number of applications. Lasers have been the driving force behind innovations in many surgical specialties. Competition from lasers has also led to significant improvements in many competing, nonlaser devices.

Lasers are now available with output wavelengths ranging from the far ultraviolet to the mid-infrared. A growing number of these laser wavelengths are being applied in medicine and surgery. In some cases, new procedures are being developed around the unique characteristics of laser energy. In other cases, a new wavelength or laser may be replacing less efficient or more expensive systems. Absorption characteristics of the many available wavelengths vary across the laser spectrum. For a laser to have a desired effect, its energy must be absorbed in the targeted tissue. As such, lasers are selected for specific applications based on the absorption characteristics of their wavelength(s) in tissue and the means available to get the laser energy to the treatment site. This chapter will provide a general overview of the role lasers play in medical and surgical applications. The various effects that can be achieved, the types of lasers generally used, the common delivery systems, and the different accessories will all be included.

Learning Outcomes

Upon completion of this module, you should be able to do the following:

- Describe the process of refraction, reflection, absorption, scattering, and transmission of laser light in human tissue. The description should include identification of the approximate wavelength ranges at which absorption is relatively low and relatively high as well as the effect on skin pigmentation.
- For optical properties of a tissue at a specific wavelength, calculate optical penetration depth.
- Describe how the optical properties of tissue change during thermal coagulation or carbonization.
- For a given absorption coefficient and tissue depth, calculate irradiance of light using Beer's law.
- For a particular laser and tissue type, determine the temperature rise resulting from a laser exposure. State whether this exposure is likely to cause tissue destruction.
- Provide the absorption coefficient of an organic tissue at several laser wavelengths.
- Name the most widely used lasers in medicine along with the output wavelength(s) for each and the type of applications where they are most suited.
- For given values of beam power and target spot size, calculate irradiance of laser beam on a target.
- Describe the use of fiber optics in medicine.
- Calculate the losses in an optical fiber due to attenuation and at the ends due to Fresnel reflection.
- For given values of refractive indices, calculate the critical angle for total internal reflection in a fiber and the numerical aperture.
- Understand accessories associated with medical laser systems. This includes articulated arms, hollow waveguides, silica optical fibers, specialty optical fibers, micromanipulators, handpieces, endoscopes, and scanning systems.

1. Properties of Laser Light

It is laser light's unique characteristics that give it much of its utility. Normal light is incoherent. Consider the average light bulb. The photons emitted cover a broad range of wavelengths from the infrared through the visible and ultraviolet. These photons diverge, or spread out, randomly in all directions. The light bulb is only useful for illuminating broad areas. Light from a laser is very different.

Laser light is *monochromatic*, *coherent*, and *directed* (Figure 1). While many laser mediums are capable of emission at several different wavelengths, generally lasers are designed to use the most efficient wavelength(s), or in other words, the most intense lines.

While lasers can be designed to produce multiple wavelengths simultaneously, each laser line is *monochromatic*—of a single, narrow wavelength. Therefore, laser light can be monochromatic even when multiple lines are lasing at the same time. Each wavelength can be separated from the others by passing the beam through a prism or a grating, producing a beam of a single wavelength.

Another characteristic of laser light is its *coherence*. Because of the way the emission of photons is stimulated within the medium, the photons are in phase, or in step, with each other. Comparing incoherent light to coherent light would be like comparing a rain shower to an ocean wave. Raindrops fall randomly in no particular sequence, therefore their impact is diffused over space and time. By comparison, an ocean wave of the same volume, with all of its “drops” in phase with one another, hits the same place at the same time. The impact of those “drops” all together is magnified by their coherence.

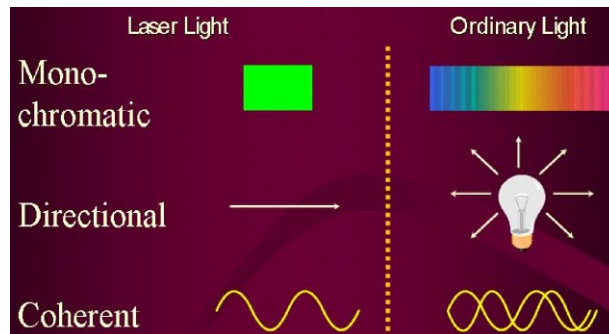


Figure 1. Comparison of monochromatic, directional, and coherent light from a laser versus polychromatic, diverging, and incoherent ordinary light from a light bulb.

Table 1. Summary of properties of laser light versus light bulb.

Properties	Laser Light	Light Bulb
Color	monochromatic	polychromatic
Direction	directional	diffuse
Coherence	coherent	incoherent
Average Power	high or low	high or low
Peak Power	very high	low
Pulse Duration	cw to fs	cw or long pulse
Intensity	very high	low

The third unique characteristic of laser light is *directional*. Directed light consists of photons all traveling in the same direction with minimal divergence. Laser light is emitted in a narrow, well-defined beam that does not spread rapidly. This characteristic makes it easier to focus the laser's output to a very small, well-defined spot (Figure 2). The spot diameter, D , can be approximated based on the divergence angle, θ , the Range, R , and the size of the laser aperture, a , as given in Equation 1:

$$(1) \quad D \sim [2 (\sqrt{2}) R] + a$$

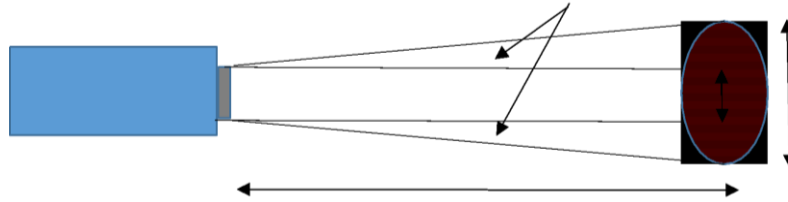


Figure 2. Spot diameter (D) of a laser beam at a target depends on beam divergence angle, θ (in radians), and distance from laser to target. The beam diameter as it exits from the laser aperture is given by a .

The laser power density, commonly referred to as irradiance, I (W/cm^2), can be determined by dividing the laser output beam power, P (W), by the area of the incident laser beam, A (cm^2). This is shown in Equation 2. By convention, the irradiance in laser medicine is provided in units of W/cm^2 rather than W/m^2 .

$$(2) \quad I = P / A$$

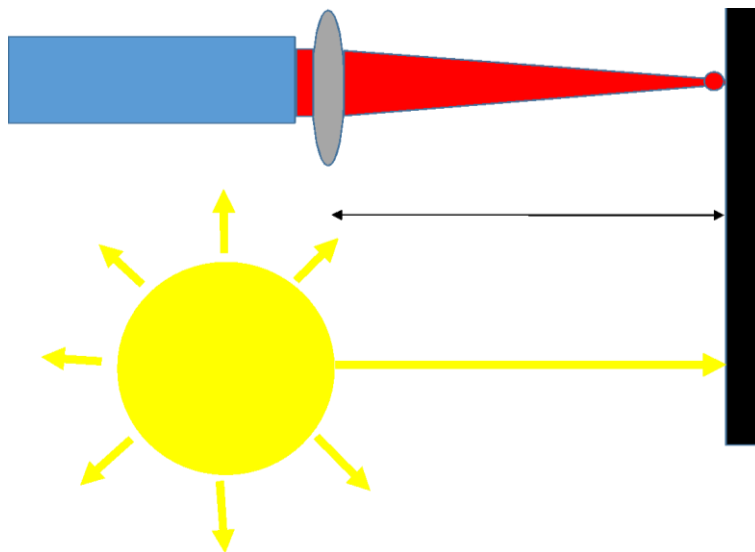


Figure 3. Comparison of low-power (1 mW) laser pointer beam (with biconvex lens providing focusing) and a standard 100-Watt light bulb.

Example 1: In Figure 3, consider a laser pointer with power output of 1 mW (1×10^{-3} W). A lens focuses the output beam to a spot diameter of 10 μm (1×10^{-5} m) at a distance of 10 m. Calculate the power density or irradiance (W/cm^2), then do the same for a light ray from a 100 W light bulb at a distance of 10 m.

Solution: Irradiance is given by power divided by area or $I = P / A$.

Laser: The area of a circle is given by $A = \pi r^2$ and $r = d/2$. Substituting these values gives,

$$I = P / A = (0.001 \text{ W}) / [(3.14) (5 \times 10^{-4} \text{ m})^2] = 1,274 \text{ W}/\text{cm}^2 = 1.3 \text{ kW}/\text{cm}^2.$$

Light Bulb: The surface area of a circle is $4 \pi r^2$, where r is equal to the distance of 10 m. Therefore,

$$I = P / A = (100 \text{ W}) / [4 (3.14) (10 \text{ m})^2] = 0.08 \text{ W}/\text{m}^2 \text{ (m / 100 cm) (m / 100 cm)} = 8 \mu\text{W}/\text{cm}^2.$$

The laser pointer produces an irradiance over one-hundred million times higher than the light bulb. This is because the directional laser beam is focused into a small spot as opposed to divergent light rays from the light bulb.

Self-Test

1. Which of the following sources does not produce electromagnetic waves?

(a) Laser (b) Ultrasound (c) Sun (d) Light bulb

2. Which of the following is a monochromatic light source?

(a) Sun (b) Flashlight (c) Laser (d) Light-emitting diode (LED)

3. Which of the following is an example of a highly diverging light source?

(a) Light bulb (b) Laser (c) Flashlight (d) All of the above

2. Laser-Tissue Interactions

2.1 Introduction

When laser energy is incident on tissue, five things happen. Some of the light will be *reflected*, some of the light will be *refracted* at the interface, some will be *absorbed* at the treatment site, some will be *scattered* in different directions within the tissue, and some will be *transmitted* into tissues beyond the treatment site (Figure 4). For the laser to be effective, the light must be absorbed by the targeted tissues. The degree to which each occurs is a function of the laser wavelength and how the laser energy interacts with the tissue being irradiated based on the optical properties of the specific tissue. If the laser wavelength has been selected properly, the majority of energy is absorbed into the targeted tissue. Reflection of some of the laser energy reduces the laser's efficiency. Any energy not reflected, absorbed, or scattered is transmitted through and eventually dissipated in the underlying tissues.

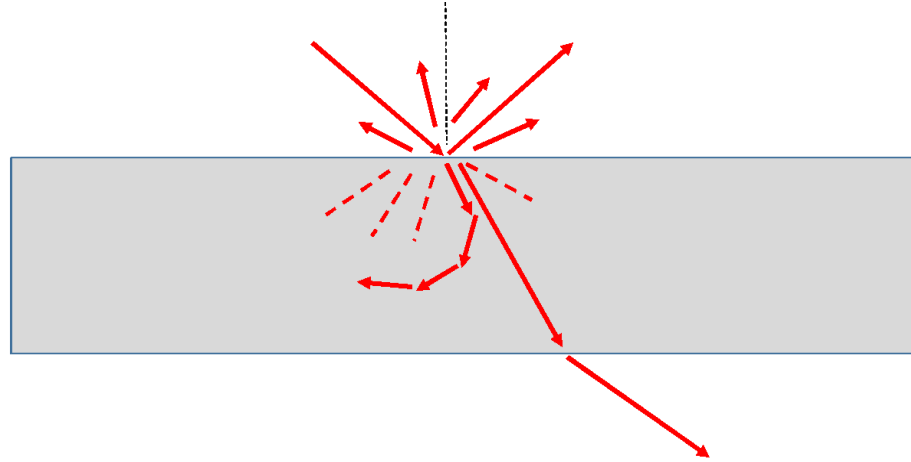


Figure 4. Light interactions with tissue showing reflected, refracted, absorbed, scattered, and transmitted light.

Reflection can be categorized as *specular* if the light is reflected from a smooth surface or *diffuse* if the light is scattered from a rough surface. In specular reflection, the angle of incidence is equal to the angle of reflection, while in diffuse reflection, the light rays are reflected in multiple directions.

This section will primarily focus on what happens to light traveling inside the tissue in terms of both light *absorption* and *scattering*.

The optical properties of any tissue for a specific wavelength, λ , can be completely described by four parameters: *refractive index*, *absorption coefficient*, *scattering coefficient*, and *anisotropy factor* (or direction of scattering), as summarized in Table 1. If these values are all known, then it is possible to predict how deep the light will go into tissue. If the laser wavelength is changed, then all parameters will also change as well. We will describe each of these parameters in more detail.

Table 1. Complete optical parameters describing a given tissue for a specific wavelength (λ).

Parameter	Symbol	Units
Refractive Index	n	None
Absorption Coefficient	μ_a	cm^{-1}
Scattering Coefficient	μ_s	cm^{-1}
Anisotropy Factor (Direction of Scattering)	g	None

2.2 Refractive Index

The *refractive index*, n , is defined as the ratio of the speed of light in a vacuum, c , given by 3×10^8 m/s, divided by the speed of light in a medium, v (Equation 3).

$$(3) \quad n = c / v$$

The refractive indices of materials will be used to determine how much light is lost due to Fresnel reflection at the interface between two different media as well as the refraction of light through an interface between two media. The refractive index values of several relevant media for laser medicine are provided in Table 2.

Table 2. Refractive index (n) of several relevant media.*

Medium	Application	n
Air	free laser beam traveling in air	1.00
Water	fluid medium inside the body	1.33
Soft Tissue	target tissue	1.39
Glass	silica optical fiber delivery system	1.45

* Refractive index is also a function of wavelength, temperature, and pressure.

2.3 Absorption coefficient

The *absorption coefficient*, μ_a , is related to the probability that a photon (packet of light) is absorbed in the tissue per a unit path length (with units of cm^{-1}). A higher value for the absorption coefficient means that the photon has a higher probability of being absorbed over a shorter distance in the tissue. A lower value means the opposite—that the photons will travel deeper into the tissue before being absorbed. Once the photon is absorbed, it disappears and its energy is converted into heat (a temperature rise in tissue) for therapy. For example, thermal coagulation refers to “cooking” of tissue or ablation refers to “vaporization” of tissue. The photon could alternatively be re-emitted at a different wavelength (color) which is described as fluorescence and may be used for diagnostic applications.

It is important to note that the absorption coefficient (and penetration depth of light) in tissue is strongly dependent on the wavelength (λ) of light used as well as major tissue components such as proteins, hemoglobin, melanin, and water (Figure 5). For example, in the ultraviolet spectrum ($\lambda < 400$ nm), both absorption and scattering are very high, and hence light has a shallow penetration in tissue (1-100 nm). In the blue (roughly 400-500 nm) and green (roughly 500-600 nm) parts of the visible spectrum, light is selectively absorbed by hemoglobin (blood), so light penetration is also relatively short (50-800 nm). In the red (roughly 600-700 nm) and near-IR (700-1300 nm) part of the spectrum, referred to as the “optical window,” protein, blood, and water absorption are all relatively low, so light penetrates the deepest (1-5 mm); however, its penetration is ultimately limited by multiple scattering in opaque tissues (most tissues except for the eye, which is transparent). Intuitively, we know that water absorption must be low in the visible spectrum because we can see through the atmosphere and while submerged underwater. In the infrared spectrum, there are several progressively larger water absorption peaks in tissue (at 1440, 1940, and 2940 nm), which result in stronger light absorption and correspondingly shallower light penetration.

If the goal is to thermally coagulate tissue for removal of a tumor, removal of an abnormal growth, or for optical imaging, then operating in the “optical window” at near-IR laser wavelengths between 800-1300 nm provides the deepest laser heating and treatment volume. On the other hand, for precise laser tissue ablation (vaporization), an ultraviolet wavelength (e.g. 193 nm for LASIK eye surgery) or a mid-IR wavelength (2790 nm for vaporization of tooth decay in dentistry) is typically chosen.

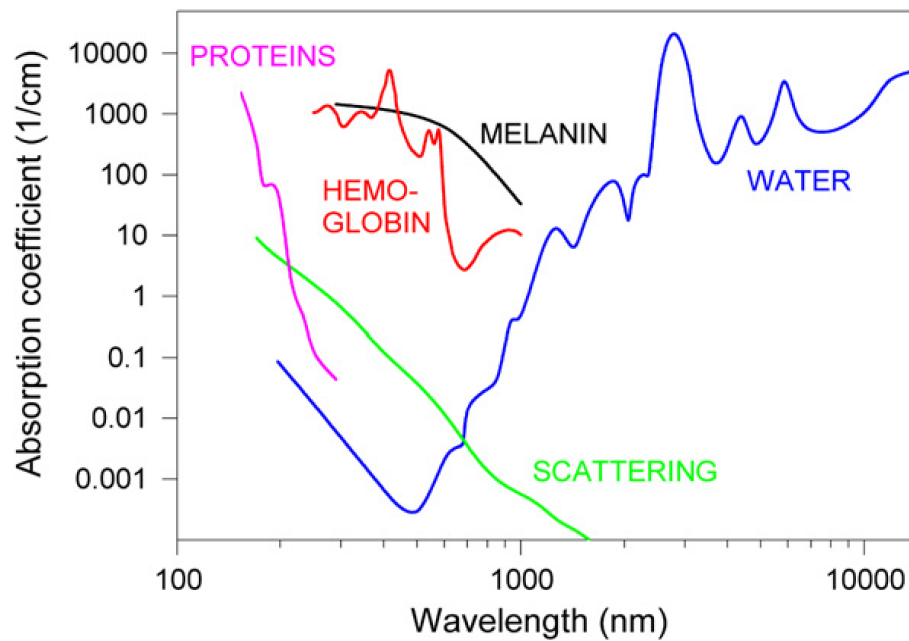


Figure 5. Absorption coefficient values plotted on the vertical axis, as a function of laser wavelength on the horizontal axis, for several common components in soft tissues. This includes proteins, melanin, hemoglobin (blood), and water. Light scattering is also shown as a reference.

Much of laser medicine may simply be reduced to choosing a laser wavelength that is either highly absorbed by a major tissue “chromophore” (e.g. water or blood) or avoiding these absorption peaks for deep light penetration. Water typically makes up about 60-80% of all soft tissues and 10-30% of all hard tissues, and so it is a major component in all tissues in the body and absorbs significantly in the UV and very strongly in the mid-IR spectrum. Figure 6 shows different IR laser wavelengths that closely match major water absorption peaks for tissue ablation.

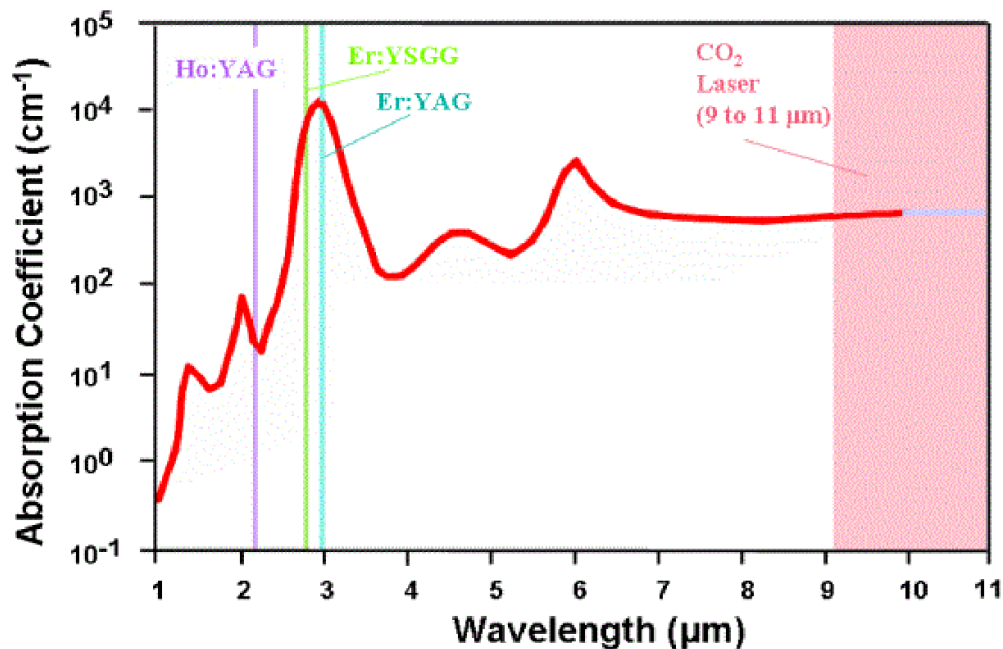


Figure 6. Absorption coefficient of water plotted in infrared spectrum from 1000-11,000 nm with common laser systems labeled, some of which match major water absorption peaks at 1440, 1940, 2940, and 6000 nm.

Oxygenated and deoxygenated hemoglobin (blood) is also a major absorber of light in the visible spectrum. Vascular tissues may have about 2% blood content, but this value can increase to 5-10% for inflamed tissues, so blood absorption can also dominate laser-tissue interactions in the blue (roughly 400-500 nm) and green (roughly 500-600 nm) parts of the visible spectrum, as shown in Figure 7.

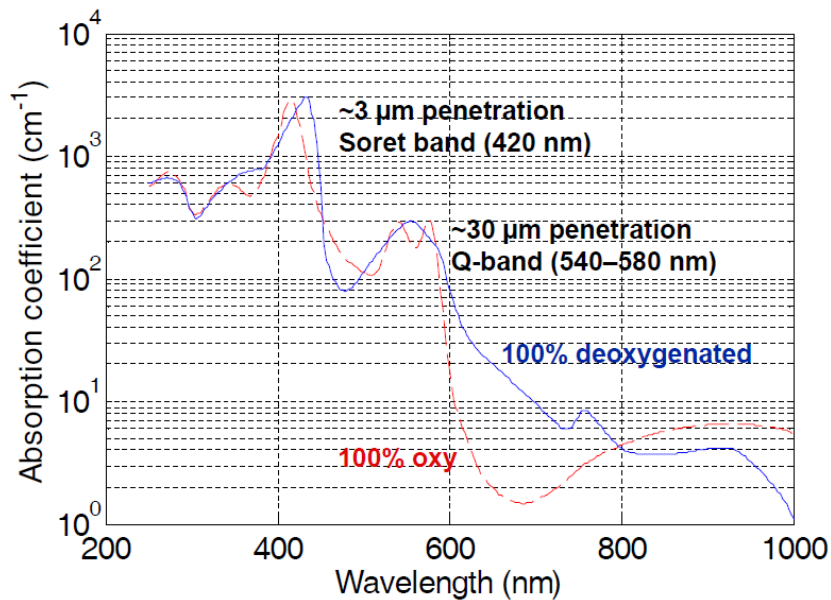


Figure 7. Absorption coefficient of blood plotted from 200-1000 nm. Note the major blood absorption peaks near 420 nm (blue) and 540-580 nm (green) parts of the visible spectrum.

2.4 Scattering coefficient

The *scattering coefficient*, μ_s , is related to the probability that a photon is scattered in tissue per a unit path length. A higher value for the scattering coefficient means that the photon has a higher probability of being scattered over a shorter distance in the tissue, while a lower value means the opposite—that the photons will travel deeper into the tissue before experiencing a scattering event.

Scattering in tissues originates from light interaction with biological structures. This ranges from cell membranes (roughly 10 nm) to whole cells (roughly 10 μ m). Light is strongly scattered when the size of the biological structure matches the laser wavelength and when the refractive index of the biological structure does not match the surrounding medium. Cell mitochondria and nuclei are on the order of 0.5 – 5 μ m, and so they strongly scatter visible (0.4 – 0.7 μ m) and IR light (> 0.7 μ m). It is multiple scattering of the light that ultimately limits the penetration depth of light in tissues, not absorption. Although light scattering decreases at longer laser wavelengths, increasing light absorption due to water dominates, and light penetration depth is therefore still limited (Figure 5).

The *total attenuation coefficient* is simply the sum of the absorption and scattering coefficients. While the absorption and scattering coefficients do not have any inherent physical meaning, it is useful to think of their inverse or reciprocal values (units of cm), which correspond to the *mean-free path* or distance that a photon is likely to travel before undergoing either an absorption or scattering event.

When a photon experiences a scattering event, it is important to know in what direction the photon is likely to be scattered. This property is quantified by the *anisotropy factor*, or direction of scattering, given by the symbol, g . The value of g is based on a cosine trigonometric function, and hence theoretically can have any value between $-1 \leq g \leq 1$ (Figure 8). A value of $g = -1$ is interpreted as the photon being back-scattered towards where it originally came from. A value of $g = 0$ refers to isotropic scattering, meaning that the photon has an equal probability of scattering in any direction (think of light emitted equally in all directions from a light bulb). A value of $g = +1$ means that the photon continues exactly in the forward direction after being scattered and without any change in its route of travel (as if the photon was never scattered in the first place). Most tissues have a high positive g value, around $g = 0.9$, meaning that the photons continue in a narrow cone of possible trajectories in the forward direction.

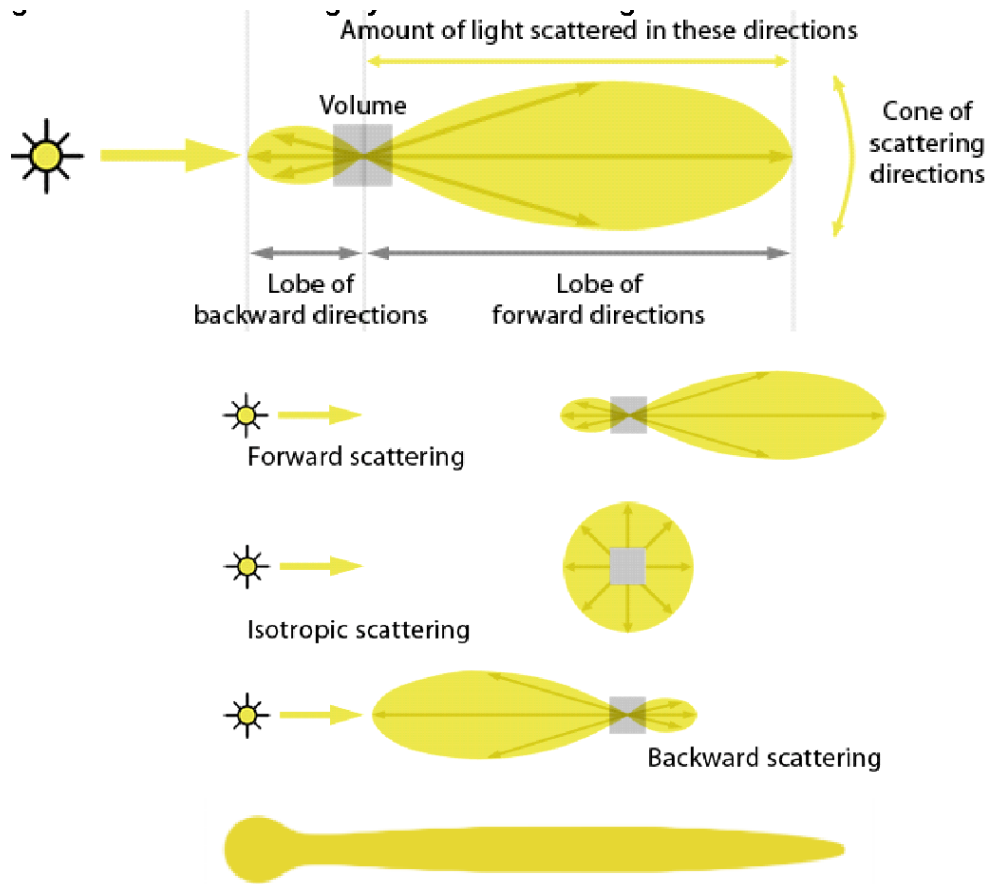


Figure 8. Anisotropy factor (directional scattering coefficient) for a photon in a scattering event in tissue. Values of g can be from -1 to 1 , but for most tissues, g is typically about 0.9 , (scattered in a narrow cone in the forward direction).

The *reduced scattering coefficient*, μ_s' , takes into account both the scattering coefficient and anisotropy factor, g , to provide a more accurate description of the true effect of a light scattering event. This is useful because if a photon is scattered but continues to travel in the forward direction, then the net effect of the scattering event is limited, and the photon appears to continue roughly along its current path. In such a case, μ_s' would be a significantly lower value than μ_s . The reduced scattering coefficient does not appear in Table 1 because it is not an independent parameter. The reduced scattering coefficient is given by the formula (Equation 4)

$$(4) \quad \mu_s' = \mu_s (1-g)$$

2.5 Albedo

The amount of diffuse light *reflected* off of the tissue surface can be quantified by the *albedo*. Lighter skin has a higher albedo because it reflects a higher percentage of the light, while darker skin has a lower albedo because it absorbs a higher percentage of light. Albedo (a) is calculated using the formula (Equation 5):

$$(5) \quad a = \mu_s / (\mu_a + \mu_s)$$

Throughout the visible spectrum there are also appreciable differences in the amount of light reflected, depending on the pigmentation level of the skin. This means that, for a frequency doubled Nd:YAG (532 nm) green or ruby (694 nm) red laser, much more light will be reflected from the skin of a fair-skinned person than from a darker-skinned person (Figure 9). Note that in the near-infrared spectrum the two curves come together. In the mid-infrared spectrum and beyond, the reflectance drops to a low value and is not strongly dependent on the pigmentation level of skin. Therefore, for the Er:YAG laser at 2940 nm or the CO₂ laser at 10,600 nm, very little radiation will be reflected, regardless of the pigmentation. This means that absorption can be high. The reflectance value is important because reflected light is lost and is not available for heating or cutting the tissue. For example, with a ruby laser incident on a fair-skinned person, 0.6 or 60% of the incident energy may be reflected. However, sufficiently high concentrations of laser energy will damage exposed tissue. Once tissue is damaged, its characteristics are altered, which can affect the ratios of reflected, absorbed, and transmitted energy.

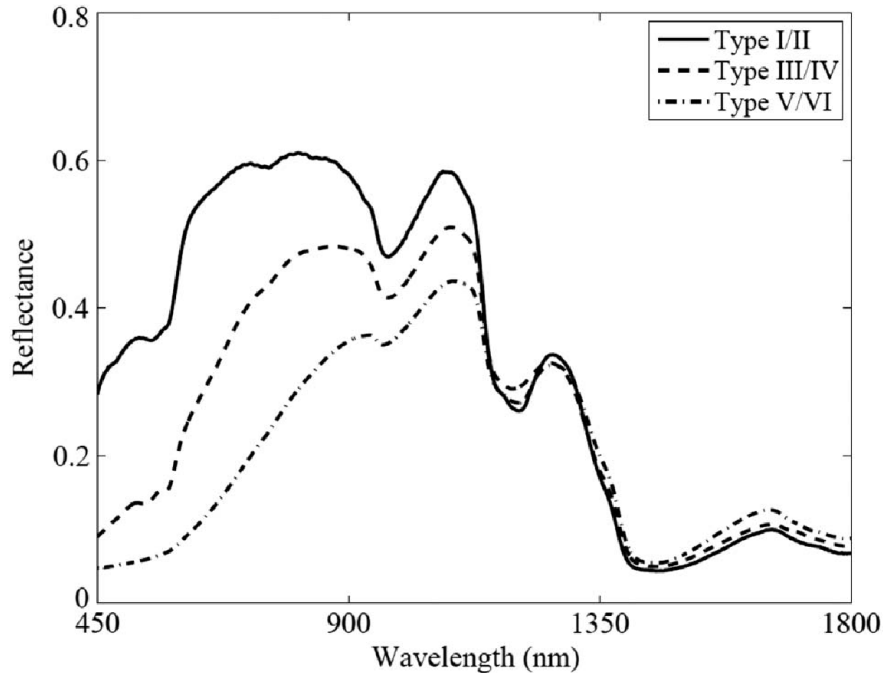


Figure 9. Reflectance of skin as a function of both wavelength and skin type. The skin types I/II are fairest, types III/IV are moderately darker, and types V/VI are the darkest or most heavily pigmented. The visible region (VIS) extends from 400 – 700 nm.

It is also important to note that the optical properties of a given tissue are dynamic, not static. In other words, they change as the laser energy is absorbed by the tissue and the tissue heats up. For example, during laser photocoagulation of a tumor during laser interstitial thermal therapy (LITT), the tissue is blanched or turns white. This results of this are a significant rise in the scattering coefficient, μ_s , of the tissue. Based on the formula above, both the albedo, a , and the attenuation coefficient will increase, leading to a corresponding decrease in the penetration depth of light through the tissue. In general, the optical parameters, g and μ_a , do not change much with blanching.

If the tissue becomes desiccated or dehydrated and then starts to carbonize or char (typically at temperatures above 300 °C) the optical properties also change. As the tissue turns black in color, its absorption coefficient, μ_a , increases significantly. Surgeons typically try to avoid charring the tissue because not only does it severely limit optical penetration depth, but the tissue carbonization can also lead to delayed wound healing.

Transmitted energy raises some interesting issues. It will tend to scatter multiple times as it passes through tissue. It eventually penetrates the surrounding tissue where it is absorbed. If enough laser energy is absorbed in the surrounding tissues, it can cause undesirable *collateral thermal damage*. This is a major safety concern with some laser wavelengths. Choosing the correct wavelength (where the light is selectively and strongly absorbed by the target tissue but not surrounding tissue) as well as delivering the laser energy in a pulsed mode sufficiently short to prevent thermal conduction of the heat to surrounding tissues during the laser pulse will minimize collateral thermal damage. Many cosmetic dermatology applications (e.g. removal of wrinkles, hair, port-wine stains, and tattoos) as well as other laser ablation procedures (e.g. LASIK eye surgery and dentistry) exploit pulsed lasers for selective laser surgery.

In certain cases, a laser wavelength may be selected so that it will transmit through surface tissues to reach underlying target tissues where it will be absorbed. Lightly pigmented tissues will reflect or transmit visible and near-visible wavelengths that heavily pigmented tissues would absorb. Since hemoglobin and oxyhemoglobin appear as red, they will tend to reflect red light and absorb its complement, green light. For this reason, lasers emitting green or yellow wavelengths are often used to treat vascular lesions such as port-wine stains. The energy is transmitted through the skin to the underlying lesion where it is absorbed by the hemoglobin, thereby destroying the vascular structure that makes up the lesion. While this is easily accomplished on a fair-skinned person, it becomes more complicated as the concentration of melanin in the skin increases. Melanin and hemoglobin have similar absorption (Figure 5). Depending on the laser wavelength being applied, the energy may be absorbed in the skin's melanin before it reaches the lesion. This can lead to a bleaching effect over the treated area. Applied cooling of the skin surface in the form of a noncontact cold air or cryogen spray or contact cooling with a gel or sapphire window is frequently integrated into dermatological laser systems to help minimize skin reddening and irritation due to absorption in upper skin layers.

Returning to Figure 9, it is shown that where reflection is high, absorption will be correspondingly low. Comparing both curves in Figure 9, one sees that tissue destruction occurs more rapidly for laser wavelengths that are more strongly absorbed (less reflected). Thus, wavelengths above 1000 nm are strongly absorbed and more prone to cause tissue damage more quickly for both fair-skinned and dark-skinned persons.

2.6 Optical penetration depth

The total attenuation (or loss) of light in tissue is given by adding the effects of both absorption and scattering. The *effective attenuation coefficient*, μ_{eff} , is given by the formula (Equation 6):

$$(6) \quad \mu_{\text{eff}} = [3\mu_a (\mu_a + \mu_s')]^{1/2}$$

This formula is derived from optical diffusion theory, which is beyond the scope of this text. Again, this value does not have any physical meaning. However, the *optical penetration depth*, δ , or the value for the depth that the light penetrates through tissue before the intensity drops to $1/e$, or 37% of its initial value, does have physical meaning, and is given by the inverse of μ_{eff} , or (Equation 7):

$$(7) \quad \tau^{\text{TM}} = 1 / \mu_{\text{eff}}$$

The formula for optical penetration depth, δ , assumes a “wide” collimated beam or large laser spot diameter (about 4 mm or greater).

The optical penetration depth is an important value. For example, as mentioned earlier, laser medicine applications requiring precise vaporization (ablation) of tissue typically use ultraviolet (LASIK for corneal shaping and vision correction) or mid-IR wavelengths (cosmetic skin resurfacing and dentistry) because these wavelengths are strongly absorbed by water and/or proteins in the tissue and the optical penetration depth is small. On the contrary, near-IR wavelengths are used for imaging deep through tissue or for deep volumetric heating and thermal destruction (coagulation) of cancerous tumors and benign growths, and the optical penetration depth is large. Figure 10 provides estimates of the optical penetration depth in generic vascular soft tissues for some common lasers used in medicine.

Figure 10

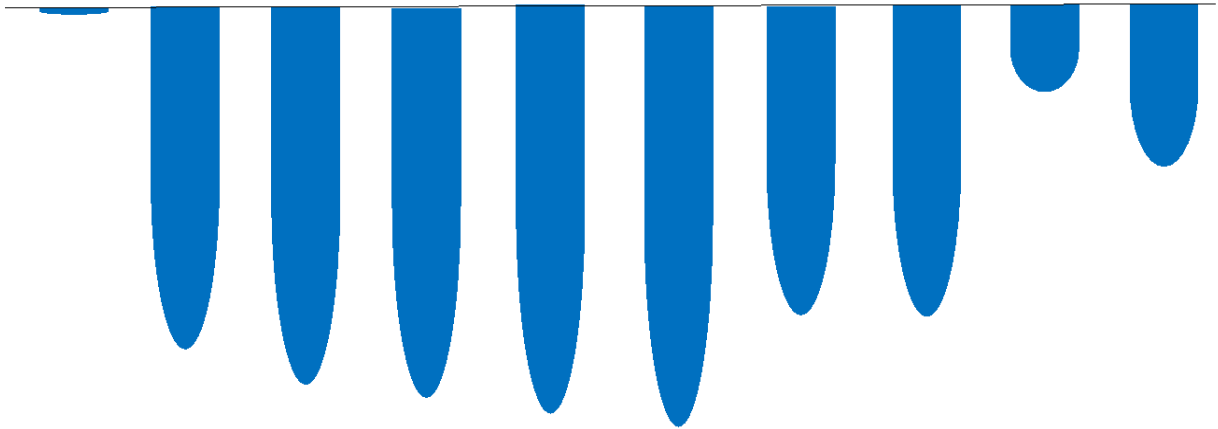


Figure 10. Optical penetration depth (δ), plotted on a logarithmic scale in micrometers, for some common laser wavelengths (λ) used in medicine.

Example 2. If an fluoride (ArF) excimer laser at 193 nm is used for vaporizing cornea tissue, using Figure 10, estimate effective attenuation coefficient (cm^{-1}) due to both absorption and scattering.

Solution: From Figure 10, the optical penetration depth in soft tissue at 193 nm is given by $\tau^{\text{TM}} = 0.001$ mm. The effective attenuation coefficient is related to the optical penetration depth by $\tau^{\text{TM}} = 1 / \mu_{\text{eff}}$. Therefore, the attenuation coefficient is just $\mu_{\text{eff}} = 1 / \tau^{\text{TM}} = 1 / 0.001 \text{ mm} = 1 \times 10^3 \text{ mm}^{-1} = 1 \times 10^4 \text{ cm}^{-1} = 10,000 \text{ cm}^{-1}$. This strong attenuation due to both absorption and scattering in the deep UV spectrum allows ultraprecise corneal shaping for vision correction.

Example 3. Calculate the optical penetration depth, δ^{TM} , of Nd:YAG laser light in skin at $\lambda = 1064 \text{ nm}$, if $\mu_a = 0.5 \text{ cm}^{-1}$, $\mu_s = 200 \text{ cm}^{-1}$, and $g = 0.97$.

Solution:

First, solve for the reduced scattering coefficient, μ_s' .

$$\mu_s' = \mu_s (1-g) = (200 \text{ cm}^{-1}) (1 - 0.97) = (200 \text{ cm}^{-1})(0.03) = 6 \text{ cm}^{-1}$$

Now, substitute this value into the equation for the effective attenuation coefficient,

$$\mu_{\text{eff}} = [3\mu_a (\mu_a + \mu_s')]^{1/2} = [3 (0.5 \text{ cm}^{-1})(0.5 \text{ cm}^{-1} + 6 \text{ cm}^{-1})]^{1/2} = [9.75]^{1/2} = 3.12 \text{ cm}^{-1}$$

Finally, the optical penetration depth is related to the effective attenuation coefficient by:

$$\delta^{\text{TM}} = 1 / \mu_{\text{eff}} = 1 / (3.12 \text{ cm}^{-1}) = 0.32 \text{ cm} = 3.2 \text{ mm}$$

This value closely matches the value in Figure 10. Again, note that in the near-IR spectrum, scattering dominates absorption in tissue, and light penetrates the deepest, but only a few millimeters.

Table 3. Optical Properties of Liver Tissue at 1064 nm.

Tissue State	Absorption Coefficient (μ_a)	Scattering Coefficient (μ_s)	Anisotropy Factor (g)
Normal	0.3 cm^{-1}	150 cm^{-1}	0.93
Coagulated	0.3 cm^{-1}	430 cm^{-1}	0.95
Carbonized	80 cm^{-1}	150 cm^{-1}	0.93

Example 4. The Nd:YAG laser is used for thermal coagulation of liver tumors. Using Table 3, calculate the optical penetration depth of 1064 nm light in liver tissue for normal, coagulated, and carbonized tissue.

Solution: First, solve for the reduced scattering coefficient, μ_s' , for each case, using $\mu_s' = \mu_s (1-g)$:

Normal: $\mu_s' = (150 \text{ cm}^{-1}) (1 - 0.93) = (150 \text{ cm}^{-1})(0.07) = 10.5 \text{ cm}^{-1}$

Coagulated: $\mu_s' = (430 \text{ cm}^{-1}) (1 - 0.95) = (430 \text{ cm}^{-1})(0.05) = 21.5 \text{ cm}^{-1}$

Carbonized: $\mu_s' = (150 \text{ cm}^{-1}) (1 - 0.93) = (150 \text{ cm}^{-1})(0.07) = 10.5 \text{ cm}^{-1}$

Now, substitute these values into equation for the effective attenuation coefficient, $\mu_{\text{eff}} = [3\mu_a (\mu_a + \mu_s')]^{1/2}$:

Normal: $\mu_{\text{eff}} = [3 (0.3 \text{ cm}^{-1})(0.3 \text{ cm}^{-1} + 10.5 \text{ cm}^{-1})]^{1/2} = [9.72]^{1/2} = 3.1 \text{ cm}^{-1}$

Coagulated: $\mu_{\text{eff}} = [3 (0.3 \text{ cm}^{-1})(0.3 \text{ cm}^{-1} + 21.5 \text{ cm}^{-1})]^{1/2} = [19.62]^{1/2} = 4.4 \text{ cm}^{-1}$

Carbonized: $\mu_{\text{eff}} = [3 (80 \text{ cm}^{-1})(80 \text{ cm}^{-1} + 10.5 \text{ cm}^{-1})]^{1/2} = [21,720]^{1/2} = 147.3 \text{ cm}^{-1}$

Finally, optical penetration depth is related to the effective attenuation coefficient by $\delta = 1 / \mu_{\text{eff}}$:

Normal: $\delta = 1 / 3.1 \text{ cm}^{-1} = 0.32 \text{ cm} = 3.2 \text{ mm}$

Coagulated: $\delta = 1 / 4.4 \text{ cm}^{-1} = 0.23 \text{ cm} = 2.3 \text{ mm}$

Carbonized: $\delta = 1 / 147.3 \text{ cm}^{-1} = 0.007 \text{ cm} = 70 \text{ }\mu\text{m}$

In summary, thermal coagulation and blanching of the tissue white reduces the penetration depth significantly through increased scattering. Carbonization of tissue, turning its color black, reduces the penetration depth greatly through increased absorption (70 μm is only the thickness of a human hair).

2.7 Transmission (Beer's law)

As previously introduced in Equation 1, irradiance (I) of a circular laser beam is given by power (P) divided by laser spot area (A). By convention, the irradiance is typically provided in units of W/cm^2 . The area can be calculated as $A = \pi r^2$, where r is the radius of the laser spot. The spot diameter is sometimes measured using a full width half maximum (FWHM) or 50% value. This is more commonly the $1/e^2$ or 13.5% value from the peak (where $e = 2.7$).

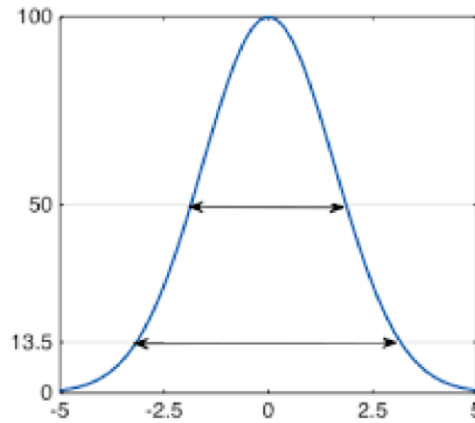


Figure 11. Definition of full width half maximum (FWHM) and $1/e^2$ laser spot diameters.

Irradiance is dependent on both power and laser spot size. Low irradiance laser settings can be used for Low-Level Laser Therapy (LLLT) such as laser acupuncture, accelerated wound healing, and infrared nerve stimulation. Medium irradiance settings can be used for thermal coagulation of tissue for removal of tumors, benign growths, retinal photocoagulation, vascular birthmarks, and endovenous therapy. High irradiance laser settings are used for ablation or vaporization of tissues in applications such as LASIK eye surgery, removal of dental caries, removal of tattoos, fragmentation of kidney stones, etc.

An obvious way to change irradiance is to change the laser power setting. However, surgeons also frequently keep laser power fixed but will vary laser spot size instead. This can be done simply by changing the working distance between the optical fiber delivery system and the target tissue. Since optical fibers do not have any focusing capability and the output beam diverges as it exists the optical fiber, changing the working distance effectively changes the irradiance as well. The resulting effect is that a shorter working distance increases irradiance and provides higher temperatures as well as an ablative effect for tissue vaporization, while a longer working distance decreases the irradiance and yields lower tissue temperatures for heating and thermal coagulation of tissue. For example, 1 W concentrated on one square centimeter ($1 \text{ W}/\text{cm}^2$) is enough power to burn paper, 1 W concentrated on an area one tenth of a square centimeter ($10 \text{ W}/\text{cm}^2$) will burn a wooden door, and 1 W concentrated on an area one-hundredth of a square centimeter ($100 \text{ W}/\text{cm}^2$) will begin to work its way through metal (Figure 12).

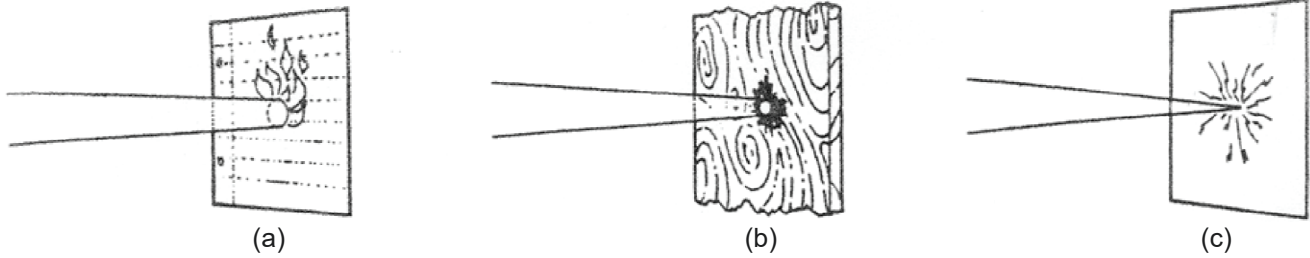


Figure 12. Effect of laser beam spot size on different materials for a fixed power of 1 W: (a) paper, (b) wood, and (c) metal.

Example 5. If the incident laser power on the tissue surface is 1 W, calculate the irradiance of the laser spot (in W/cm²) for spot sizes of 1, 2, and 4 mm.

Solution: Use the formula, $I = P / A = P / \pi r^2$

1 mm spot: $d = 1 \text{ mm} = 0.1 \text{ cm}$; $r = d/2 = 0.05 \text{ cm}$, so $I = (1 \text{ W}) / [(3.14) (0.05 \text{ cm})^2] = 127 \text{ W/cm}^2$

2 mm spot: $d = 2 \text{ mm} = 0.2 \text{ cm}$; $r = d/2 = 0.1 \text{ cm}$, so $I = (1 \text{ W}) / [(3.14) (0.1 \text{ cm})^2] = 32 \text{ W/cm}^2$

4 mm spot: $d = 4 \text{ mm} = 0.4 \text{ cm}$; $r = d/2 = 0.2 \text{ cm}$, so $I = (1 \text{ W}) / [(3.14) (0.2 \text{ cm})^2] = 8 \text{ W/cm}^2$

Note that the irradiance decreases by a factor of four when the laser spot is doubled.

The irradiance of light at a specific depth in tissue, or through a tissue layer thickness, can be calculated using a simple formula. *Beer's law* states that irradiance decays at an exponential rate in a medium (Figure 13) based on the tissue depth (x) and the total attenuation coefficient, which is in turn dependent on both the absorption and scattering coefficients. For the ultraviolet ($< 400 \text{ nm}$) and mid-IR ($> 1900 \text{ nm}$) spectrum, absorption dominates scattering ($\mu_a \gg \mu_s$); scattering can be neglected and Beer's law provides an accurate measurement of irradiance with tissue depth (Equation 8):

$$(8) \quad I = I_0 \exp(-\mu_a x),$$

where I (W/cm²) is final irradiance, I_0 (W/cm²) is initial irradiance, μ_a is absorption coefficient (cm⁻¹), and x is tissue depth (cm).

If Beer's law is rewritten in terms of I / I_0 , then this ratio is defined as the fraction of collimated light that is transmitted through the tissue at a given depth. If the value is multiplied by 100, then a percentage of *transmitted* light is obtained (Equation 9):

$$(9) \quad T = I / I_0 = \exp(-\mu_a x)$$

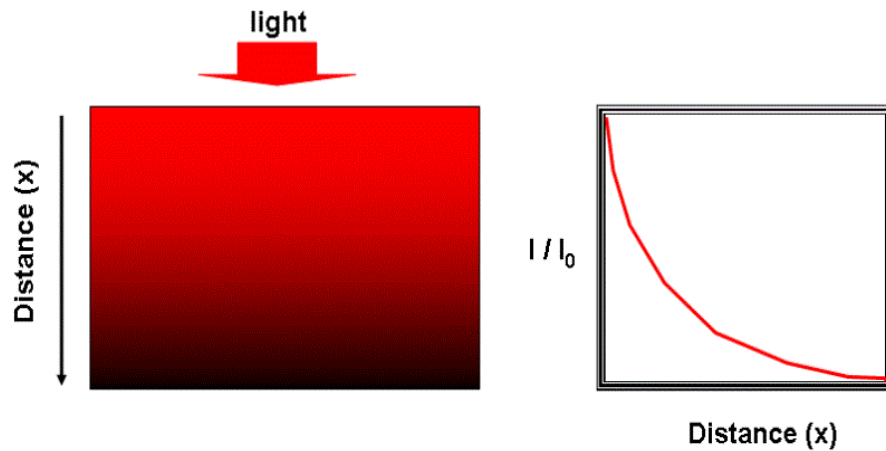


Figure 13. Beer's law can be used to calculate the exponential decay in irradiance with depth in the tissue, and is a valid approach when absorption dominates scattering at ultraviolet and mid-IR wavelengths.

Example 6. As a continuation of Example 2, calculate the percentage of 193 nm light transmitted through a 400 μm thick human cornea. Compare this value with that for a Ho:YAG laser at 2120 nm ($\mu_a = 25 \text{ cm}^{-1}$). Explain the difference.

Solution: Using Beer's law, collimated Transmission through cornea for thickness of $x = 0.04 \text{ cm}$ is:

$$T = I / I_0 = \exp(-\mu_a x)$$

For an excimer laser at 193 nm, $T = \exp [(-10,000 \text{ cm}^{-1})(0.04 \text{ cm})] = 0\%$.

For a holmium laser at 2120 nm: $T = \exp [(-25 \text{ cm}^{-1})(0.04 \text{ cm})] = 0.37 = 37\%$.

The excimer laser is used for cornea ablation due to its ultraprecise removal of tissue. On the other hand, more than 1/3 of the holmium laser light would penetrate the entire cornea, so it is incapable of precise tissue removal for eye surgery.

There is a large matrix of laser parameters (Table 4). For a specific medical laser application, only a handful of these parameters may be relevant. Typically, the wavelength is the most important parameter because all of the tissue optical properties are dependent on wavelength. The pulse energy, power, pulse duration, pulse rate, spot size, and irradiation time are also typically considered.

Table 4. Summary of important laser parameters.

Parameter	Units	Definition
Wavelength (λ)	nanometers (nm)	color of light
Pulse Energy (E)	Joules (J)	energy in a single laser pulse
Average Power (P_{ave})	Watts (W)	amount of energy over 1 s time period
Peak Power (P_{peak})	Watts (W)	energy divided by pulse duration

Spot Size (d)	millimeters (mm)	diameter of laser spot
Radiant Exposure	J/cm ²	energy density
Fluence (F)	J/cm ²	total energy deposited in a spot area
Spatial Beam Profile	none	shape of single laser pulse in space
Irradiance (I)	W/cm ²	power density
Pulse Duration (t)	seconds (s)	length of single laser pulse in time
Pulse Repetition Rate (R)	Hertz (Hz) or 1/s	number of laser pulses delivered in 1 s
Temporal Beam Profile	none	shape of laser pulse in time
Duty Cycle	none	fraction of a period in which laser is on
Irradiation Time	seconds (s)	total time that laser is on

Now we will look at some of these parameters in more detail.

2.8 Operation mode

Lasers may be operated in continuous-wave (CW) or pulsed mode. When a laser is operated in CW mode, the power, average power, and peak power are all the same. However, when a laser is operated in pulsed mode, then the average and peak powers can be very different. *Average power* refers to amount of energy delivered over a 1 s time period and can be calculated by multiplying the pulse energy, E, in Joules, times the pulse repetition rate, R, in Hertz (Equation 10):

$$(10) \quad P_{\text{ave}} = E \times R$$

The *peak power* refers to the energy, E, in Joules, contained in a single pulse divided by the duration of the pulse, t, in seconds (Equation 11):

$$(11) \quad P_{\text{peak}} = E / t$$

As Figure 14 shows, one advantage of lasers over light bulbs is that although the average power may be low, the peak power can be extremely high since lasers are capable of emitting light with extremely short pulse durations as short as femtoseconds (10^{-15} s). These high-peak powers can be used, for example, to heat tissue up to very high temperatures for coagulation and vaporization during therapeutic laser procedures.

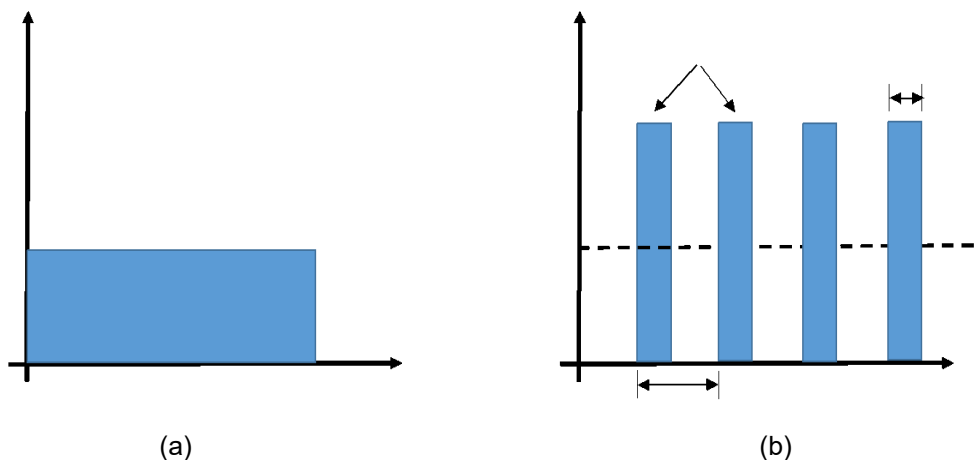


Figure 14. Comparison of (a) continuous-wave and (b) pulsed operation modes for lasers.

Example 7. Calculate the average and peak power for a Ho:YAG laser delivering energy in an optical fiber inserted through a flexible endoscope for fragmenting kidney stones. The laser is operated with a pulse energy of 500 mJ (0.5 J) pulse duration of 200 μs (2×10^{-4} s), and pulse repetition rate of 20 Hz.

Solution:

Average power is given by:

$$P_{\text{ave}} = E \times R = (0.5 \text{ J}) (20 \text{ Hz}) = 10 \text{ W}$$

Peak power is given by:

$$P_{\text{peak}} = E / t = (0.5 \text{ J}) / (2 \times 10^{-4} \text{ s}) = 2.5 \times 10^3 \text{ W} = 2,500 \text{ W}$$

Note that there is a large difference between peak power and average power in this example.

When a laser is operated in pulsed mode, the time duration of the laser pulses may vary considerably depending on the specific laser system and mode of operation (Table 5). Lasers can also sometimes be operated at short pulse durations and high pulse rates, effectively producing a “quasi-CW” mode for heating tissue, since the duty cycle is high with minimal tissue cooling in between delivery of successive pulses.

Table 5. Laser Modes of Operation.

Mode	Pulse Length	Time Period
Continuous-wave (CW)	always on	seconds (s)
Modulated, chopped, or gated	milliseconds	10^{-3} s
Long pulse (free-running)	microseconds	10^{-6} s
Short pulse (Q-switched)	nanoseconds	10^{-9} s
Ultrashort pulse (mode-locked)	picoseconds-femtoseconds	10^{-12} s - 10^{-15} s

2.9 Laser spot size

When a laser is operated at a wavelength in which the photons are not highly absorbed in the tissue, then multiple light scattering events dominate the laser-tissue interaction process ($\mu_s \gg \mu_a$). This is typical for wavelengths in the range of 600 – 1300 nm, the red to near-infrared spectrum, referred to as the “optical window.” For this regime, the laser spot diameter is important in determining the irradiance of light at a given depth in tissue. For example, the irradiance of a narrow beam (0.2 mm) decays much more rapidly in tissue than for a wide beam (5 mm) due to higher losses from light scattering along the periphery of the laser beam (Figure 15). One way to think of this effect is to take the ratio of the volume to surface area of a cylindrical laser beam in tissue, $\pi r^2 l / 2\pi r l$, where r is the radius and l is the length of the cylinder. A wider laser beam has a larger volume to surface area ratio (scales roughly as r) and hence loss of light intensity due to scattering out of the periphery of the beam is relatively lower.

The implications are widespread. For example, many successful cosmetic laser therapy applications (such as tattoo removal, skin resurfacing, hair removal, treatment of vascular birthmarks, and others) use a large

diameter beam to get more uniform delivery of light to the target depth (e.g. for laser removal of ink particles in a tattoo located below the skin surface).

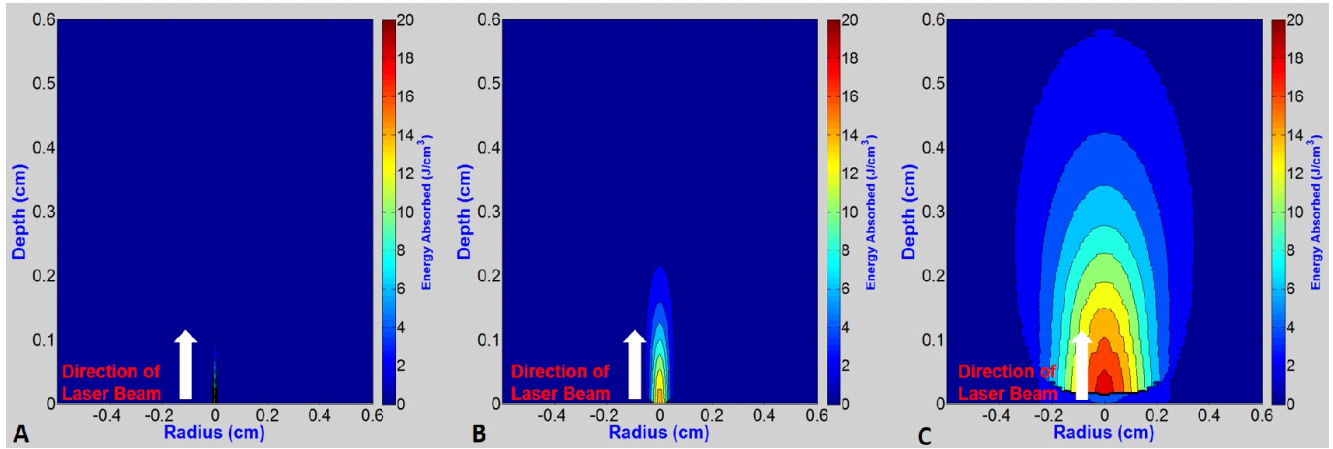


Figure 15. Effect of laser spot size on intensity of light with depth in highly scattering tissues for a wavelength of 1064 nm. The three simulations (a,b,c) are normalized for the same intensity, but the larger spot size in (c) produces much deeper light penetration. Laser beam diameters measure (a) 0.2 mm, (b) 1 mm, and (c) 5 mm.

2.10 Spatial beam profile

The shape of the laser beam in space, referred to as the *spatial beam profile*, is also important, because it shows how uniformly laser power is distributed (Figure 16). Single-mode laser beams produce a Gaussian beam profile that can be focused down to a smaller spot for coupling of higher laser power into smaller optical fibers for surgery. Multimode laser beams are nonuniform and cannot be focused to a small spot as easily.

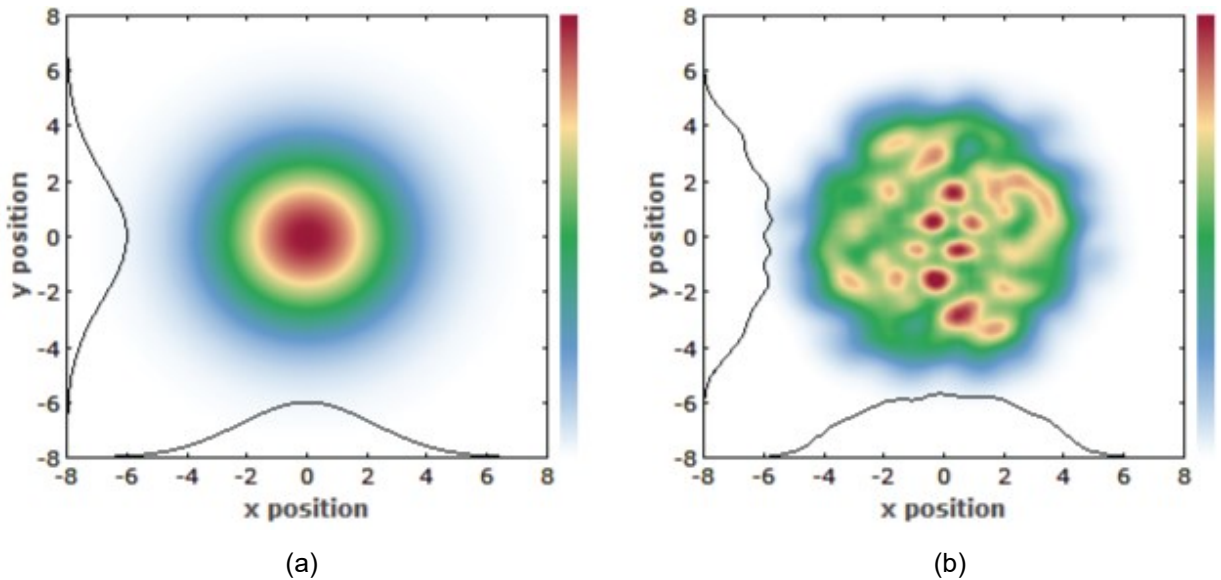


Figure 16. (a) Lasers emitting a single mode beam produce a uniform, symmetrical, Gaussian-shaped beam profile. (b) Lasers emitting a multimode beam produce a nonuniform, asymmetrical beam profile.

2.11 Temporal beam profile

The shape of the laser beam in time is referred to as the *temporal beam profile*. This parameter characterizes how the energy is distributed during a single laser pulse. Flashlamp-pumped, solid-state lasers such as the Ho:YAG and Er:YAG, as well as CO₂ lasers, typically produce a “front-loaded,” asymmetrical, or “spiky,” temporal beam profile, characterized by an initial spike, followed by a decay in intensity. Diode lasers and diode-pumped lasers typically produce a flat-top beam profile where the laser energy is more uniformly distributed (Figure 17).

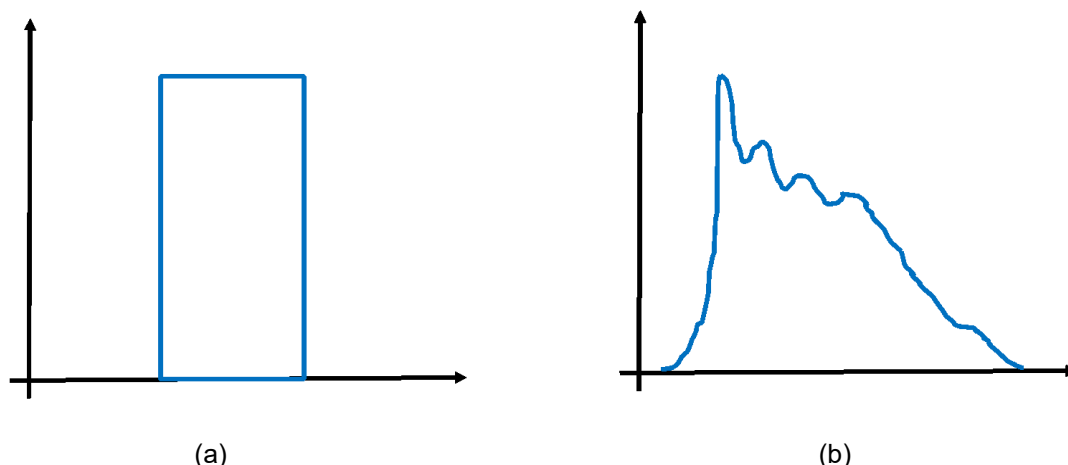


Figure 17. (a) Diode lasers emit a uniform, flat-top temporal beam profile. (b) Flashlamp-pumped, solid-state lasers and CO₂ lasers emit a nonuniform beam profile where energy may be “front-loaded.”

2.12 Thermal interactions

In surgical applications the goal is generally to destroy tissue. How that is accomplished depends on the wavelength used and how it is applied to the tissue. A major consideration when using lasers in a surgical application is the degree of collateral damage that results. With some wavelengths, the amount of transmitted light and its effect on underlying tissues is a concern. The interaction of laser radiation with tissue produces a thermal response. If the absorption of the thermal energy raises the tissue to a high enough temperature for a long enough time, cells will be destroyed. Figure 18 shows the amount of heating needed to cause tissue destruction; it illustrates how long a cell can survive at elevated temperatures before cell viability is compromised sufficiently enough to render it nonviable. Normal body temperature is 37 °C (98.6 °F). From the curve in Figure 18, cell death should not occur even after six hours at 45 °C (113 °F). However, if the tissue temperature is raised between 55 and 60 °C (130–140 °F), cell death can occur in as little as a few seconds. When underlying tissue is sufficiently heated by transmitted laser energy, it will be destroyed. As a result, a layer of necrotic (dead) tissue will remain afterward. The depth of the dead tissue is called “the zone of thermal necrosis.” The depth of this zone is a function of the wavelength and how it is applied. Which wavelength is chosen and how the energy will be applied often depends on the nature of the application.

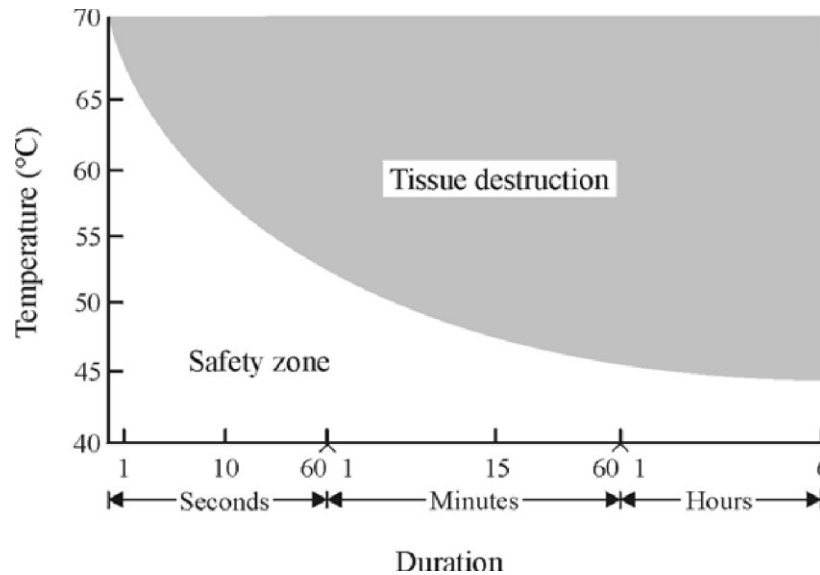


Figure 18. Relationship of temperature and time for tissue destruction to occur (normal body temperature is 37 °C). In the region below the shaded area in the figure, tissue destruction and cell death should not occur.

Self-Test

4. Which is NOT one of the four optical parameters that characterizes optical properties of a tissue?
 (a) μ_a (b) μ_s (c) g (d) n (e) I
5. Which of the following situations always results in light penetrating deeper into tissue?
 (a) Higher μ_a and μ_s (b) Higher μ_a and lower μ_s (c) Lower μ_a and higher μ_s (d) Lower μ_a and μ_s
6. What criteria is necessary for precise vaporization of tissue without collateral thermal damage?
 (a) High g value (b) $\mu_a > \mu_s$ (c) $\mu_s > \mu_a$ (d) Low g value
7. If you shine a flashlight through your hand, only the red visible light is transmitted, but not blue or green light. Why?
 (a) Red light is more strongly absorbed by water than blue and green light
 (b) Blue and green light are more strongly absorbed by hemoglobin and scattered than red light
 (c) Red light is more strongly scattered by melanin in your skin than blue and green light
 (d) Blue and green light are more strongly absorbed by water than red light
8. What approximate percentage of soft tissues is water?
 (a) 1-2% (b) 5-10% (c) 20-25% (d) 40-50% (e) 60-80%
9. What approximate percentage of vascular soft tissues is blood?
 (a) 1-2% (b) 5-10% (c) 20-25% (d) 40-50% (e) 60-80%
10. What component of tissue is the main absorber of laser radiation in the mid-IR spectrum?
 (a) Hemoglobin (b) Melanin (c) Water (d) DNA

11. What component of tissue is the main absorber of green laser radiation in the visible spectrum?
(a) Hemoglobin (b) Collagen (c) Water (d) DNA
12. In general, what laser wavelength provides the deepest penetration in tissue?
(a) Ultraviolet (b) Visible (c) Near-infrared (d) Mid-infrared
13. For $\mu_a = 2000 \text{ cm}^{-1}$, assuming ($\mu_a > \mu_s$), how far does light penetrate in tissue?
(a) $5 \mu\text{m}$ (b) $50 \mu\text{m}$ (c) $500 \mu\text{m}$ (d) 5 mm
14. Does white or dark skin have a higher albedo in the visible spectrum?
(a) White skin (b) Dark skin (c) White and dark skin have same albedo
15. Which value for the anisotropy factor, g , indicates strong, forward light scattering?
(a) -0.9 (b) -0.5 (c) 0 (d) $+0.5$ (e) $+0.9$
16. Which clinical laser application requires deep light penetration in tissues?
(a) LASIK (b) Wrinkle removal (c) LITT of tumors (d) Tattoo removal (e) Dental decay removal
17. Which laser provides a logical replacement for a Nd:YAG laser for deep heating of tissues?
(a) Excimer (b) Frequency doubled Nd:YAG (c) Infrared Diode (d) Ho:YAG (e) Er:YAG
18. Beer's law is accurate for calculating light transmission through a tissue layer in which spectrum?
(a) UV (b) Visible (c) Near-IR (d) Mid-IR (e) UV and mid-IR
19. A Q-switched laser operates in what mode?
(a) CW (b) pulsed, ms (c) pulsed, μs (d) pulsed, ns (e) pulsed, fs
20. What is the average power of a laser operating at 1 J and 20 Hz?
(a) 0.2 J (b) 0.2 W (c) 20 J (d) 20 W
21. What is the peak power of a laser operating at 1 J and 50 ms?
(a) 0.2 J (b) 0.2 W (c) 20 J (d) 20 W
22. In highly scattering tissues, which laser spot size provides the deepest penetration of light?
(a) 0.1 mm (b) 0.5 mm (c) 1 mm (d) 2 mm (e) 5 mm.
23. What is normal body temperature?
(a) 37°C (b) 50°C (c) 100°C (d) 150°C (e) 300°C
24. What is the lowest approximate temperature that causes thermal damage to tissue?
(a) 37°C (b) 50°C (c) 100°C (d) 150°C (e) 300°C
25. At what temperature does the water in the tissue start to boil and vaporize?
(a) 37°C (b) 50°C (c) 100°C (d) 150°C (e) 300°C
26. At what temperature does desiccated tissue start to char or carbonize?

- (a) 37 °C (b) 50 °C (c) 100 °C (d) 150 °C (e) 300 °C

27. When tissue is thermally coagulated (turns white), which parameter changes the most?

- (a) μ_a , increases (b) μ_a , decreases (c) μ_s , increases (d) μ_s , decreases

28. When tissue is thermally coagulated, how is the optical penetration depth affected?

- (a) Increases (b) Stays the same (c) Decreases

29. When tissue is carbonized (turns black), which parameter changes the most?

- (a) μ_a increases (b) μ_a decreases (c) μ_s increases (d) μ_s decreases

30. When tissue is carbonized, how is the optical penetration depth affected?

- (a) Increases (b) Stays the same (c) Decreases

31. What can be done to prevent tissue carbonization?

- (a) Cool tissue surface (b) Keep fiber tip clean (c) Monitor tissue temperature (d) All of the above

3. Commonly Used Lasers

3.1 Introduction

The heart of the laser is the active laser medium, a material that will absorb and emit radiation when excited with the appropriate type and intensity of excitation energy. Each type of laser wavelength's characteristics are determined by a particular medium. It is customary to refer to a laser by the wavelength-producing component of its active medium. In some cases, this is only part of the active medium.

A CO₂ laser uses a combination of three or more gases to achieve the laser output. It is the CO₂ gas that produces the wavelength, so it is called a CO₂ laser. In the Nd:YAG laser, it is the neodymium (Nd) which emits, producing the output wavelength. The mineral YAG (yttrium aluminum garnet) is simply a host crystal for the neodymium. One aspect that can be confusing is that YAG is used as a host crystal for a variety of emitting elements designated as Er:YAG (erbium), Ho:YAG (holmium), and Tm:YAG (thulium). In medical circles, the term "YAG laser" is generally used to refer to the Nd:YAG. The others are referred to as either the "erbium," "holmium," or "thulium" lasers.

The active laser medium may be solid, liquid, or gas. The excitation energy may (in the case of medical lasers) be in the form of an electric current, intense light, radio frequency, or electromagnetic (EM) radiation—which one of the three that is appropriate depends on the nature of the medium. Laser mediums may be crystals (solid), dyes (liquid), gases, or semi-conductors. The crystals and dyes require exposure to intense light to stimulate emission. In some cases, another laser may be used as the light source. Semi-conductor or diode lasers require an electric current to stimulate them. Gas mediums may be excited using either an electrical discharge or radio frequency radiation depending on the type of gas(es) being used.

A relatively new type of laser, the "fiber laser," uses diode lasers to pump and optically excite a chemically-doped optical fiber, such that the active medium is actually the optical fiber itself. This results in high-power delivery in a small laser spot from the fiber. Common infrared fiber lasers include ytterbium, erbium, and thulium-doped silica fibers.

The resulting emission from the energized medium produces the characteristic wavelength(s) of the medium, the laser output wavelength(s).

This section will provide an overview of some of the different therapeutic lasers. In some cases, examples of certain procedures will be described for illustrative purposes related to wavelength or tissue effects. A complete listing of the procedures that can be performed using each laser would be too exhaustive for this text.

3.2 Medical laser systems

There are several common lasers used in medicine (Figure 19). Table 6 summarizes their wavelengths, laser-tissue interaction features, clinical applications, and optical delivery systems. We will consider some of these lasers in more detail below.

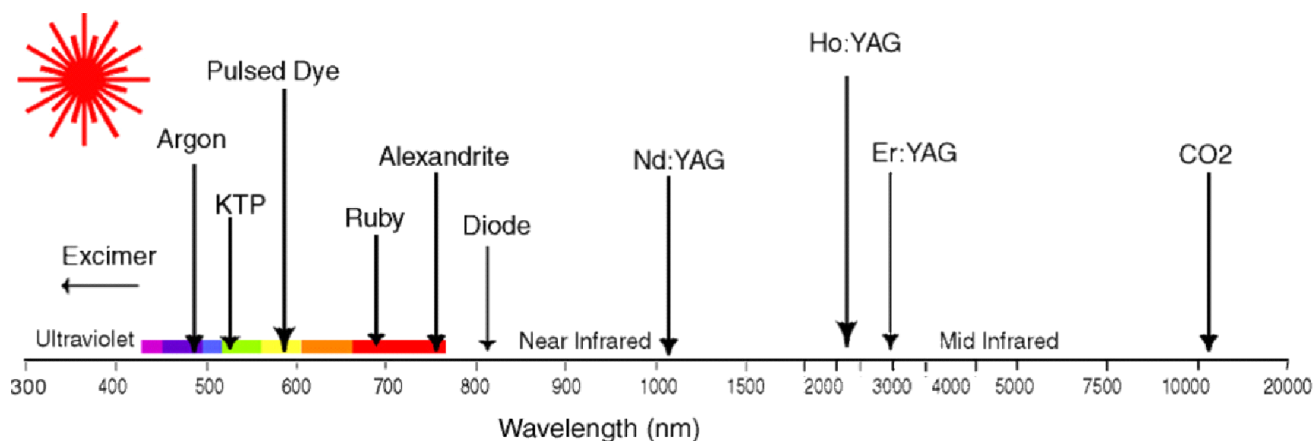


Figure 19. Some lasers commonly used in medicine.

Table 6. Laser Types Established in Medical Applications.

Laser	λ (nm)	Features	Applications	Beam Delivery Type
Excimer	193	Very high absorption in tissue for precise vaporization	Ophthalmology (LASIK eye surgery)	None
Excimer	308	High absorption in tissue	Cardiology (angioplasty), Dermatology (psoriasis)	Silica
KTP	532	High absorption in blood	Ophthalmology (retina), Urology (BPH)	Silica
Diode	635	Wavelength matches cancer drug absorption peak	Cancer (light-based chemotherapy, PDT)	Silica
Alexandrite	755	Pulsed laser with deep penetration	Dermatology (hair and tattoo removal)	Silica
Diode	808, 980	Low absorption, deep penetration	Cancer (LITT), Dermatology (hair removal)	Silica
Nd:YAG	1064	Low absorption, deep penetration	Cancer (LITT), Dermatology (hair and tattoo removal)	Silica
Diode	1470	Similar depth to Ho:YAG	Endovenous therapy	Silica
Th:YAG	2010	Intermediate penetration depth for vaporization and coagulation	Gastroenterology, Urology (BPH)	Silica

Ho:YAG	2120	Intermediate penetration depth for vaporization and coagulation	Gastroenterology, Urology (kidney stones and BPH)	Silica
Er:YSGG	2790	Very high tissue absorption for precise tissue vaporization	Dentistry (remove tooth decay)	mid-IR fiber
Er:YAG	2940	Very high tissue absorption for precise tissue vaporization	Dermatology (skin resurfacing), Dentistry (remove tooth decay)	Articulated arm, hollow waveguide, mid-IR fiber, scanner
CO ₂	9,600 10,600	High absorption in tissue (water and mineral)	General Surgery, Dentistry (remove tooth decay)	Articulated arm, hollow waveguide, scanner

The excimer laser—The ArF excimer laser operates at a wavelength of 193 nm in the ultraviolet spectrum. This laser has the most precision of any laser used in surgery. This is due to its very high absorption by both tissue proteins and blood as well as high scattering of light at short wavelengths. Due to its extreme precision, the excimer is used for vaporization of small amounts of corneal tissue during vision correction in LASIK/PRK procedures (Figure 20). The UV-C wavelength cannot be delivered through any optical fibers. However, this is not required for LASIK. Instead, the free laser beam is delivered directly to the cornea. The laser only operates in short pulse mode with pulse durations on the order of tens of nanoseconds. Overall, the laser is large, relatively expensive, and not user-friendly, so it is typically not used for any other procedures. In general, the UV spectrum falls into three categories based on wavelength of light; these are categorized as UV-A (320-400 nm), UV-B (280-320 nm), and UV-C (< 280 nm).

Summary

Laser Type:	Gas laser
Wavelength:	193, 248, 308, 351 nm
Spectrum:	Ultraviolet
Operation Mode:	Short pulse (nanoseconds)
Laser-Tissue Interactions:	Strong absorption by water and proteins, carcinogenic potential
Technical Comments:	Very user-unfriendly, large, expensive laser systems
Laser Parameters:	Short pulse operation (nanosecond pulses)
Delivery Systems:	None at 193 nm, silica fibers at longer wavelengths
Clinical Applications:	Ophthalmology (LASIK/PRK) at 193 nm Cardiology (angioplasty) at 308 nm Dermatology (psoriasis) at 308 nm



(a)

(b)

Figure 20. Ultraviolet lasers. (a) Argon fluoride excimer laser at 193 nm, used for LASIK corneal shaping and vision correction procedures. (b) Xenon chloride excimer laser at 308 nm, used for treatment of psoriasis.

The frequency-doubled Nd:YAG laser—A potassium titanyl phosphate (KTP) crystal placed in front of the laser output from an Nd:YAG laser at 1064 nm results in the generation of harmonics or integral multiples of the frequency. Since the wavelength (λ) is inversely related to the frequency (f), shorter wavelengths in the visible and ultraviolet spectrum can be obtained (e.g. 1st harmonic: $1f = 1064$ nm, 2nd harmonic: $2f = 532$ nm, 3rd harmonic: 355 nm, etc). The second harmonic of a Nd:YAG laser, at 532 nm (green), has higher potential output powers and similar tissue effects to the argon gas laser, but it is more user friendly and has lower repair costs. The frequency-doubled Nd:YAG laser has essentially replaced the argon laser as a photocoagulator for all applications that require high blood absorption and hemostasis during surgery. On such application is in ophthalmology, where small, retinal blood vessels need to be sealed; another such application is in urology for the treatment of an enlarged prostate gland (known as benign prostatic hyperplasia or BPH), where benign tissue needs to be removed through vaporization but hemostasis is also required to limit blood loss. The green 532 nm laser wavelength can also be delivered through a fluid environment with minimal absorption of the energy as opposed to mid-IR lasers like Ho:YAG and Er:YAG.

Summary

Laser Type:	Solid-state medium (laser rod), KTP crystal for doubling frequency
Wavelength:	532 nm
Spectrum:	Visible (green)
Operation Mode:	Continuous-wave (CW), long pulse, or short pulse (Q-switched)
Laser-Tissue Interactions:	Strong hemoglobin absorption for soft tissue vaporization/hemostasis
Technical Comments:	Compact laser—has replaced the argon ion laser for photocoagulation
Delivery Systems:	Standard silica optical fibers
Clinical Applications:	Ophthalmology (coagulation of abnormal retinal blood vessels) Urology (BPH)

Diode lasers—Diode lasers are compact, relatively inexpensive, solid-state, semiconductor lasers that primarily operate in the visible and near-IR spectrum. These lasers have high wall-plug electrical efficiency, allowing them to be air cooled (eliminating large bulky water-cooling systems) and operated from a standard 110-V electrical outlet. Low-power red and green visible diode lasers are commonly integrated into infrared medical laser systems to be used as aiming beams for aligning the invisible infrared laser beam on the tissue during the medical procedure. Higher power diode lasers can also be as efficient pump sources for solid-state and fiber laser systems. The 635 nm red diode laser has replaced the 633 nm helium-neon gas laser for many applications, and it is used in photodynamic therapy (PDT) for targeting the absorption peak of a common photosensitizer drug during treatment of cancer. The 808 and 980 nm near-IR lasers were initially developed as pump sources for solid-state laser systems but have since been used in laser medicine due to the deep penetration of these laser wavelengths in tissues. Uses for this include laser interstitial thermal therapy (LITT) of tumors and dental soft tissue applications. Their compact footprint has made them an attractive alternative to solid-state Nd:YAG lasers. The most recent development of the 1470 nm diode laser enables it to provide a CW laser alternative to the pulsed, solid-state, Ho:YAG laser, since both laser wavelengths are similarly absorbed by water in tissue, yielding an optical penetration depth of about 400 μ m. Another attractive feature of diode and diode-pumped lasers is their high electrical efficiency which enables them to be air-cooled (with fans). This makes them very quiet for use in the operating room as opposed to water-cooled, flashlamp-pumped, solid-state lasers (e.g. Ho:YAG and Er:YAG lasers).

Summary

Laser Type:	Solid-state medium (semiconductor laser)
Wavelength:	635, 808, 980, 1470 nm

Spectrum:	Red (635 nm) and near-infrared (808, 980, 1470 nm)
Operation Mode:	Continuous-wave (CW) and long pulse (ms)
Laser-Tissue Interactions:	Deep penetration of light into tissue due to weak absorption
Technical Comments:	Compact and inexpensive lasers, 635 nm diode replaces helium-neon laser 808/980 nm diodes replace Nd:YAG laser CW 1470 nm diode has similar penetration to pulsed Ho:YAG laser
Delivery Systems:	Standard silica optical fibers
Clinical Applications:	Photodynamic therapy (PDT) for light-based cancer treatment at 635 nm Laser interstitial thermal therapy of tumors and growths at 808/980 nm Endovenous laser therapy at 1470 nm

The Nd:YAG laser—Though its applications have been broadened considerably with the advent of contact tips and fibers, the Nd:YAG (neodymium: yttrium aluminum garnet) laser was originally designated as a “photo-coagulator” because of its ability to coagulate blood and control bleeding. The Nd:YAG laser has a principal wavelength in the near-IR spectrum at 1064 nm. It also has a seldom-used secondary wavelength at 1320 nm. The 1064 nm wavelength is primarily absorbed by pigmented components such as melanin, hemoglobin, and proteins. It transmits well through standard, silica optical fibers and was originally used for controlling bleeding in the bowel, debulking gastro-intestinal (GI) and pulmonary obstructions (tumors), and treating a number of urinary tract conditions. It has since been abandoned in urology since its deep penetration caused complications during early treatment of benign prostatic hyperplasia (BPH) and been replaced by the holmium:YAG laser which has a much shallower penetration depth.

The Nd:YAG laser is frequently used in laser interstitial thermal therapy (LITT) for treatment of tumors because its laser wavelength enables deep, thermal coagulation and destruction of large tumors up to about 2 cm diameter when thermal conduction of the energy is also considered. Most recently, with advances in MRI imaging platforms and real-time temperature measurement, MRI-guided laser coagulation of tumors for brain cancer treatment and destruction of epileptic centers in patients resistant to drug and radiation therapy has gained popularity. Unlike other heating sources (e.g. ultrasound, microwaves, radio frequency, etc), the optical fiber delivery system is MRI-compatible.

Q-switched Nd:YAG lasers with nanosecond pulse durations are used for delicate work in ophthalmology. Short, high-energy pulses are used to drill precise holes in structures internal to the eye. A very common application is treatment of secondary cataracts. Short pulse Nd:YAG lasers are also used for selective removal of tattoos.

In summary, the major characteristic of the Nd:YAG laser is that its wavelength of 1064 nm penetrates most deeply into tissue, making it an ideal laser for deep heating, thermal coagulation, and destruction of large tumors or benign growths in tissue. Melanin, blood, and water are absorbers at this wavelength, but absorption is still very low. Instead, light scattering dominates at this near-IR wavelength. The Nd:YAG laser’s main limitation is that it is slowly being replaced by more compact and less expensive diode lasers operating at other near-infrared wavelengths, such as 808 and 980 nm, which provide almost the same penetration depth.

Summary

Laser Type:	Solid-state medium (laser rod)
Wavelength:	1064 nm
Spectrum:	Near-infrared
Operation Mode:	Continuous-wave (CW), long pulse, or short pulse (Q-switched)
Laser-Tissue Interactions:	Deepest light penetration due to weak absorption by tissue components
Technical Comments:	Smaller, more compact diode lasers provide competition
Delivery Systems:	Standard silica optical fibers
Clinical Applications:	Laser interstitial thermal therapy-coagulation of tumors/benign growths Neurosurgery (MRI-guided laser coagulation of brain tumors, epilepsy) Dermatology (Q-switched pulses for hair and tattoo removal)

Ophthalmology (Q-switched pulses for cataract removal)

The Ho:YAG laser—The holmium:YAG (Ho:YAG) laser has been used more frequently over the past few decades. This is due in part to its ability to both vaporize and coagulate soft tissues as well as fragment kidney stones through a standard silica optical fiber. It is neither precise like the excimer, Er:YAG, or CO₂ lasers, nor does it provide deep thermal coagulation like diode and Nd:YAG lasers. Instead, the Ho:YAG laser is essentially a compromise, which allows the surgeon to use a single laser platform for a wide variety of applications, including soft tissue ablation and coagulation as well as hard tissue fragmentation. Some clinical applications include use inside the body during laparoscopic and endoscopic surgeries in gastroenterology and urology. The holmium laser is currently the dominant laser in urology because it can be used for incision and coagulation of the prostate gland (BPH) as well as fragmentation of kidney stones during lithotripsy, two of the most common procedures in urology.

The Ho:YAG laser is capable of high energy output of several Joules per pulse and is typically operated in long-pulse mode with laser pulse durations of several hundred microseconds in duration. The laser energy can be delivered through optical fibers with core diameters as small as 200 μm . The Ho:YAG laser is a water-cooled, flashlamp-pumped, solid-state laser. While this laser architecture is inefficient, it is also relatively inexpensive, which makes the laser attractive for use in medicine. A major limitation of the Ho:YAG laser is that thermal effects in the laser rod limit its operation to pulse rates less than about 30 Hz, unlike the CO₂ laser, which can be operated at 1000 Hz.

Summary

Laser Type:	Solid-state medium (laser rod)
Wavelength:	2120 nm
Spectrum:	Mid-infrared
Operation Mode:	Long pulse (200-1500 μs), flashlamp-pumped version limited to low pulse rates (< 30 Hz)
Laser-Tissue Interactions:	Capable of both vaporization and coagulation of tissue
Technical Comments:	High-power lasers up to 140 W available
Delivery Systems:	Standard silica optical fibers
Clinical Applications:	Endoscopic surgery in Gastroenterology and Urology (kidney stones and BPH)

The Tm:YAG laser—The thulium:YAG (Tm:YAG) laser has recently been introduced as an alternative to the Ho:YAG laser for endoscopic applications. For example, in urology, the Tm:YAG laser is used for vaporization and hemostasis during incision of the prostate gland for treatment of BPH. The laser can be either flashlamp-pumped for pulsed operation or diode-pumped for CW operation. The Tm:YAG wavelength of 2010 nm has an optical penetration depth in water of about 200 μm , about one half that of the Ho:YAG laser.

Summary

Laser Type:	Solid-state medium (laser rod)
Wavelength:	2010 nm
Spectrum:	Mid-infrared
Operation Mode:	Diode-pumped, Long pulse (200-1500 μs) or Continuous-wave
Laser-Tissue Interactions:	Capable of both vaporization and coagulation of tissue
Technical Comments:	High-power lasers up to 200 W available
Delivery Systems:	Standard silica optical fibers
Clinical Applications:	Endoscopic surgery in Gastroenterology and Urology (BPH)

The Er:YAG and Er:YSGG lasers—The erbium:YAG (Er:YAG) and erbium:YSGG (Er:YSGG) lasers operate at mid-IR wavelengths of 2790 and 2940 nm, respectively. The Erbium lasers closely match a large water-

absorption peak in tissue, hence the light is strongly absorbed within a tissue layer of only 5-10 μm . After the excimer laser, erbium lasers are the most precise lasers used for soft tissue applications. Unlike the excimer laser, erbium lasers are smaller, less expensive, and more user-friendly. They are used for precise ablation of tissue with only minimal collateral thermal damage. For example, erbium lasers are used in dermatology for cosmetic skin resurfacing as well as in dentistry for precise removal of dental caries (tooth decay). The pulp of the tooth is extremely sensitive to any temperature rise, so long-pulsed erbium lasers provide precise removal of decayed tissue without damaging the pulp when the lasers are used with a water spray. Although erbium lasers are more expensive than the dental drill, lasers provide a number of advantages over the drill, including reduced noise, high aspect ratio for removing decayed tissue without damaging healthy tissue, and less pain due to elimination of vibration from the drill. The erbium lasers, similar to the holmium laser, are typically water-cooled, flashlamp-pumped, solid-state lasers, so they are relatively small and inexpensive compared to excimer lasers. However, they are limited to pulse rates of about 30 Hz or less.

Summary

Laser Type:	Solid-state medium (laser rod)
Wavelength:	2790 nm (Er:YSGG) and 2940 nm (Er:YAG)
Spectrum:	Mid-infrared
Operation Mode:	Long pulse (200-1500 μs), flashlamp-pumped version limited to low pulse rates (< 30 Hz)
Laser-Tissue Interactions:	Very high absorption in water, precise vaporization of tissues
Technical Comments:	Penetration depth varies with water temperature
Delivery Systems:	Cannot be used with standard silica optical fibers; instead, an articulated arm, hollow waveguide, or specialty mid-infrared fiber is needed
Clinical Applications:	Cosmetic dermatology (skin resurfacing); Dentistry (caries removal)

The CO₂ laser—The CO₂ is an all-purpose laser. Due to the transmission characteristics of its wavelengths in the mid-IR, it is limited to use with an *articulated arm*, *hollow waveguide*, or *specialty fiber*. Its most common wavelength is 10,600 nm for soft tissues, although 9,600 nm lasers have recently been introduced for dental hard tissue applications as well due to high mineral absorption at that wavelength ($\mu_a = 8,000 \text{ cm}^{-1}$). Both wavelengths are strongly absorbed by water and the *optical penetration depth* of the 10,600 nm CO₂ laser in water is about 20 μm . Used to cut or ablate (vaporize) tissue, the CO₂ laser boils the water inside and between the cells causing them to explode. All remaining solid components will be carbonized. Because the CO₂ laser energy only penetrates 20 μm deep, the thermal necrosis zone can be very thin. While CO₂ lasers can seal off small capillaries, thereby providing a relatively bloodless field due largely to its shallow penetration, it is not applicable for controlling bleeding in larger blood vessels. At medium power levels, a focused beam can cut through tissue rapidly. At high power levels, a relatively diffuse beam can efficiently ablate large areas.

The major advantages of the CO₂ laser are that it can be used as a general surgical workhorse laser for both soft and hard tissues. It can also be operated at pulse rates of up to about 1000 Hz, for rapid tissue removal. Its main limitation is that the long laser wavelength in the mid-IR spectrum prevents the laser's use with standard silica (glass) optical fibers. Instead, relatively bulky articulated arms, hollow waveguides, or more expensive specialty mid-IR fibers are needed to deliver light from the laser to the tissue. Since it is a gas laser, the laser medium decays over time and needs to be replaced periodically as well.

Summary

Laser Type:	Gas tube
Wavelength:	9,600 and 10,600 nm
Spectrum:	Mid-infrared
Operation Mode:	Continuous-wave (CW) or long pulse (20-200 μs), capable of operating at high pulse rates of up to 1000 Hz
Laser-Tissue Interactions:	Strong absorption by water in soft tissue and minerals in hard tissues Provides tissue vaporization and limited hemostasis

Technical Comments:	Compact and inexpensive technology
Delivery Systems:	Cannot be used with standard silica optical fibers; instead, an articulated arm, hollow waveguide, or specialty mid-infrared fiber is needed.
Clinical Applications:	General surgical laser used for both soft and hard tissue surgery Endoscopic and laparoscopic surgery, dermatology, dentistry

Figure 21 shows some of the representative diode, solid-state, and gas lasers described above. Note that diode lasers are extremely compact, tabletop systems, and are ideal for integration into endoscopy carts in the operating room. The KTP, 532 nm laser (“Greenlight”) is ideal for removing vascular tissues while providing hemostasis. The mid-IR Ho:YAG laser provides both vaporization and coagulation of tissues through a standard silica optical fiber delivery system. The longer wavelength Er:YAG and CO₂ lasers are typically used with articulated arms, integrating focused handpieces or scanners. Many of these lasers are activated with a foot pedal for convenience, so the surgeon can keep their hands free for the surgical procedure.



Figure 21. (a) Small, tablet-size 980 nm diode laser used in soft tissue dental procedures. (b) High-power “Greenlight” KTP visible laser at 532 nm used in urology for vaporization of the prostate gland. (c) High-power Ho:YAG infrared laser at 2120 nm with flexible fiber-optic delivery system for endoscopic procedures in gastroenterology and urology. (d) High-power Er:YAG mid-IR laser at 2940 nm with an articulated arm and scanning system, used in dermatology for cosmetic skin resurfacing. (e) High-power CO₂ mid-IR laser at 9,600 nm with an articulated arm, used in dentistry for removal of dental decay.

3.3 New laser systems

Ultrashort pulse lasers—The ultrashort pulse laser systems deliver laser pulses on picosecond or femtosecond time scales. The intensity is so high that it produces a plasma (ionized gas) in the tissue. Plasma-mediated tissue ablation is attractive because the plasma absorbs incoming laser energy independent of laser wavelength, providing an artificially enhanced tissue absorption coefficient. In other words, unlike other laser systems discussed so far, the laser wavelength used does not ultimately determine the absorption coefficient. Tight focusing of the laser beam produces the high intensities necessary for plasma-mediated ablation with ultraprecise removal of tissue.

However, there are several major limitations of plasma-mediated ablation. First, the laser systems are expensive. Second, the high-intensity laser pulses, due to the short pulse duration, prevent delivery of the laser energy through standard silica optical fibers. Third, although precise, the laser tissue removal rates are low,

making removal of bulk tissues (such as large tumors) impractical. Ultrashort pulse lasers have been limited to use in replacing the microtome (ultrafine knife) blade for preparation of the corneal flap during LASIK procedures as well as the LASIK procedure itself, experimentally in neurosurgical procedures, and for intracellular surgery in the laboratory.

Summary

Laser Type:	Diode laser, pumped solid-state laser
Wavelength:	800-1000 nm (titanium sapphire laser), 1053 nm (Nd:YLF laser)
Spectrum:	Near-infrared
Operation Mode:	Short pulse, mode-locked, picoseconds to femtoseconds (10^{-12} – 10^{-15} s)
Laser-Tissue Interactions:	Absorption due to plasma formation, independent of wavelength
Technical Comments:	Expensive, precise but very slow tissue removal due to plasma shielding High peak powers prevent use with standard optical fibers
Delivery Systems:	Cannot be used with standard silica optical fibers
Clinical Applications:	Ophthalmology (LASIK and cataract removal) Neurosurgery and intracellular surgery

Fiber lasers—The newest series of lasers to be used in laser medicine are fiber lasers. These lasers utilize a chemically-doped silica optical fiber as their active medium, which enables high-power delivery in a small laser spot for high brightness or high-intensity applications. The lasers are diode-pumped, and although not as efficient as direct diode lasers, they are still more efficient and compact than conventional flashlamp-pumped, solid-state lasers (such as the Ho:YAG and Er:YAG lasers). Several IR fiber lasers are available, including ytterbium-doped silica (1070 nm), erbium-doped silica (1550 nm), and thulium-doped silica (1940 nm). Erbium and thulium fiber lasers have been used in cosmetic dermatology applications, including fractional skin resurfacing for wrinkle removal. This is due in part to their excellent, single-mode, Gaussian spatial beam profile output. Most recently, thulium fiber lasers have been introduced in urology for a more efficient endoscopic laser ablation of kidney stones than the conventional Ho:YAG laser (Figure 22).

Summary

Laser Type:	Diode-pumped, chemically-doped silica fiber (light originates in fiber)
Wavelength:	1070 nm (ytterbium), 1550 nm (erbium), 1940 nm (thulium)
Spectrum:	Near- and mid-infrared
Operation Mode:	Continuous-wave (CW) or short pulse (Q-switched, nanoseconds)
Laser-Tissue Interactions:	Strong absorption by water
Technical Comments:	Single-mode spatial beam profile enables high power in a small spot
Delivery Systems:	Standard silica optical fibers
Clinical Applications:	Laser interstitial thermal therapy (LITT) at 1070 nm Dermatology (skin resurfacing) and Urology (kidney stones) at 1940 nm



Figure 22. (a) Dual 1550/1927 nm, erbium/thulium fiber laser used in dermatology for fractional skin resurfacing (wrinkle removal). (b) Thulium fiber laser at 1940 nm used in urology for treatment of kidney stones.

Self-Test

32. What advantage does a CO₂ laser have compared to a flashlamp-pumped Er:YAG laser?
 - (a) More precise soft tissue ablation
 - (b) Operation at higher pulse rates
 - (c) Solid-state design
 - (d) All of the above
33. What advantages do diode lasers have over traditional gas and solid-state lasers?
 - (a) Inexpensive
 - (b) Compact
 - (c) Air-cooled and quiet
 - (d) Efficient
 - (e) All of the above
34. What is the active medium of a fiber laser?
 - (a) Gas
 - (b) Semiconductor
 - (c) Doped optical fiber
 - (d) Crystal
 - (e) Plasma
35. What advantage does a fiber laser offer over a conventional solid-state laser?
 - (a) High efficiency
 - (b) Small spot
 - (c) High intensity
 - (c) Quiet operation
 - (e) All of the above
36. For which laser is the tissue ablation process not dependent on the laser wavelength?
 - (a) Excimer laser
 - (b) Er:YAG laser
 - (c) Ultrashort pulse laser
 - (d) CO₂ laser

4. Delivery Systems

4.1 Introduction

Delivery systems for laser energy present a number of challenges depending on the application. To be effective, the appropriate levels of laser energy must be delivered to treatment sites both outside and inside of

the body. For some applications, it is important to maintain all of the output characteristics of laser energy. In others, one or more of the unique qualities of laser light can be, and often are, sacrificed. While some procedural sites are easily accessed, others require a high degree of flexibility (e.g. endoscope applications inside the body in gastroenterology and urology). Primary means of delivery include a *free-beam* from the laser resonator, *articulated arms*, *fiber optics* (some of which are highly specialized), and *hollow waveguides*.

In many cases, the distal (far) end of fiber optics and hollow waveguides are used to deliver the energy directly to the target tissue. However, in other applications, as with articulated arms, a secondary delivery device must be attached. These secondary delivery devices can vary from a specially-designed sapphire tip for fiber optics (e.g. laser dentistry) used in direct contact with tissue to a single- or multiple-lensed device that collects and focuses the energy for *noncontact* applications. Common lensed devices include *surgical handpieces*, *microscope adaptors*, and *scanners*.

4.2 Free-beam delivery

The least complicated way to deliver laser energy is for the clinician to simply hold the laser in their hand and direct the output onto the treatment site—*free-beam* delivery. A number of systems, from the CO₂ to diode lasers, use this type of delivery. This severely limits the size and potential output power available from such systems. In some cases, this means a fairly bulky handpiece.

For free-beam delivery systems, focusing optics can be included within the laser tube or the optics may be attached at the laser aperture. Special tips may be available with different angled mirrors to help deliver the laser energy to less accessible areas (e.g. in side-firing fibers used with the frequency-doubled Nd:YAG laser for ablative removal of the prostate gland during treatment of benign prostatic hyperplasia). Free-beam delivery systems are generally limited to fields such as dermatology, podiatry, gynecology (external), otolaryngology, and dental/oral surgery, where the tissue is readily accessible. One limitation of this delivery system is that it generally lacks a visible aiming beam. Delivery tips with small *guide tips* are provided for some systems to aid the clinician in determining the proper focal distance and some indication of where the beam will impact tissue. It is not uncommon for a clinician to fire a CO₂ laser onto a tongue depressor prior to surgery to determine the relative location of the beam to the special tip. While useful for a wide-range of procedures, these lasers are not intended for surgeries that require a very high degree of accuracy.

One system that uses free-beam delivery of laser energy is one that is designed for hair removal. The handpiece contains an array of IR laser diodes and a chilled tip. This delivery tip is pressed against the skin. When fired as a brief pulse, the laser energy is sufficient to kill the hair follicles. The chiller serves to lower the skin temperature at the treatment site. This helps to reduce pain and limit collateral thermal damage to tissue surrounding the follicles. The chiller window typically uses flowed water and a sapphire window, since sapphire is both optically transparent and thermally conductive.

4.3 Articulated arms

Articulated arms can be designed for use with any laser beam. All that is required is the presence of a correct reflective coating on the mirrors and a laser beam with a low-divergence angle. For some wavelengths, such as in the mid-IR and UV, articulated arms are the primary option. This is the only other delivery system that maintains all of the unique characteristics of the laser beam.

An articulating arm is essentially two long sections of straight tubing that are joined together by a series of precision bearings, each connected at a 90° angle. Each 90° joint or knuckle has an adjustable front surface mirror that reflects the laser beam down the center of the next section of tubing. Between each pair of mirrors there is a precision bearing that allows a free 360° rotation of each successive mirror and arm segment. A typical articulated arm has eight bearings and seven mirrors. In some designs, an eighth mirror is used to align or launch the beam into the arm while other systems simply align the arm to the laser by adjusting the centering

and tilt angle of the arm to the laser beam. The first section of the arm is relatively short and is mounted vertically to allow the arm to be rotated completely around the laser. The second section is horizontal and allows for a range of vertical movements. At the end of the second, long section of the arm, the last three mirrors and four bearings are designed to allow a full range of motion for the laser accessory (Figure 23).

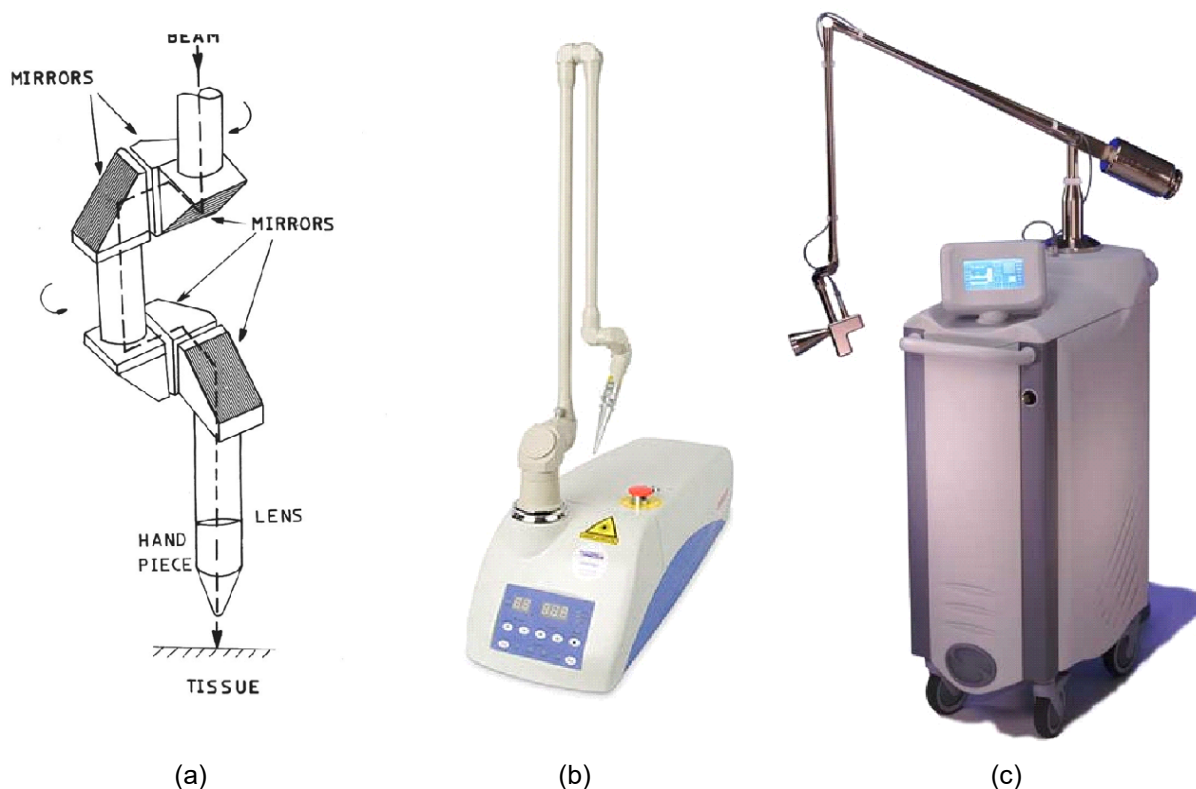


Figure 23. Articulated Arms. A series of hollow tubes with mirrors mounted at each joint provides several degrees of freedom for light delivery. Focusing optics for precise delivery, or automated scanning systems for treating large surface areas (e.g. cosmetic surgery), can be attached to the articulated arm. Such delivery systems are used with mid-IR laser wavelengths that are not transmitted through standard silica optical fibers (e.g. Er:YAG and CO₂ lasers). (a) Diagram of articulated arm. (b) Articulated arm attached to low-power CO₂ laser for use in veterinary surgery. (c) Er:YAG laser attached to an articulated arm with scanner on distal end.

Early articulated arms were made of heavy, stainless steel tubing with large diameter bearings. Due to their weight, a heavy, adjustable, counter-balancing weight was used to make the arm seem weightless to the clinician. Even with their robust construction, these arms were prone to alignment problems. Newer arms are constructed of aluminum alloys or carbon fiber-reinforced composites. Combined with high-precision bearings, these improvements have enhanced the flexibility and reliability of articulated arms. While some arms still rely on a counterweight, others incorporate a spring balance device. In either case the counterbalances are adjustable to compensate for various accessories such as handpieces or laparoscopes that will be used with the laser. The idea is to make the arm and accessory seem relatively weightless to the clinician.

Generally, all articulated arms are susceptible to alignment problems. Arm alignment problems were a major factor in the promotion of many laser systems in the 1980s. Today, misalignment of the arm is an ever-present problem since most lasers that use articulating arms, such as the CO₂ laser, have output wavelengths that are not visible. These systems usually incorporate a low-power visible diode laser as a marker or aiming beam to show where the working laser target spot will impact tissue. The output of these low-power "guide" lasers is combined with the surgical beam so that they follow the same path through the arm and accessory. If the

working laser beam is off center in the arm, diffraction and a resulting separation of the beams will occur as the two lasers pass through the focusing optics of the accessory. As the beams pass through the lens off center, the prismatic effect causes the beams to separate. Beam coincidence at the treatment site will be affected. This is especially true for devices using longer focal lengths, such as microscopes and laparoscopes. The relationship between the guiding and cutting beams is critical for accuracy in surgical procedures.

Sometimes trying to situate a laser in a crowded surgical field with microscopes, operating room lights, and other accoutrements results in flexural stresses to the arm. If a section of the arm is stressed too much, an alignment problem may occur. These problems are generally resolved when the stress is removed.

Even arms that are promoted as “permanently aligned” have some means by which they can be adjusted. There are two aspects to aligning an articulated arm. The first, and most obvious, is the angular adjustment of the mirrors. If the angle of a mirror is off, the beam will tend to wander at the distal end of the arm as the arm is moved. In bad cases, the beam may be clipped by the arm or even lost inside the arm.

The second aspect, and one that can be affected by improperly adjusting the angular aspect, is mirror location. If a mirror is at the proper location in the joint, its center is aligned with the center of both adjacent arm sections. If the mirror is too deep in the joint or too shallow, an alignment problem can occur. This type of alignment error is most problematic with narrow lumen endoscopes such as the laparoscope. While this type of error can appear to be adjusted through appropriate angular movements, the problem may persist. Beyond a mirror that is “off depth,” the laser beam will travel through each successive arm section at an angle. As a result, while the beam appears to be centered at the distal end of the arm, it is exiting the arm at an angle to the center line. Rotation of the joints of the arm will cause the beam to wander, as it describes a cone, inside the laparoscope lumen, to the point where it can clip the aperture or reflect off the inner wall of the lumen. A special lensed tube must be attached to the arm to evaluate and correct this type of problem.

Some arms use a four-point adjusting system with two hold-down screws and two standoff screws. The mirror assembly is mounted in the joint and the beam is observed as the adjustments are made. The screws are worked in pairs to adjust the angle of the mirror in an x-y pattern. If one screw is to be tightened to correct an alignment error, the opposing screw must first be loosened the same amount. By adjusting the x- and y-axes separately, and by maintaining proper screw tension, mirror depth can be maintained during angular alignment.

A more common mirror adjustment design uses three adjusting screws. As three points describe a plane, any one screw can be adjusted to move the beam. In this system, the mirror is mounted in the joint cover and must be removed to access the screws. Some sort of tensioning device, such as a wave washer or disk spring, is mounted behind the mirror to hold it firmly against the adjusting screws. Generally, one of the three screws is offset from the others. The offset screw is only adjusted when it is necessary to correct a mirror depth problem. All angular corrections are made using the other two screws. The mirror is often offset in its cap to match the offset of the adjusting screws. It is important that the mirror and cap be reinstalled in its correct orientation.

The actual techniques for evaluating and properly aligning an articulated arm are beyond the scope of this text. In brief, both *evaluation* and *alignment* must begin at the laser aperture. Coincidence between the surgical beam and any visible guiding beam device (helium neon or diode laser) at two different points (near and far) must first be ascertained. Once beam coincidence at the two locations is verified with the arms bearings immobilized, each bearing is rotated individually to check for errors. The checks proceed from the laser head to the end of the arm. Any adjustment needed is made on the mirror immediately before the bearing being rotated.

4.4 Fiber optics

Fiber optics play an important role in many surgical applications, not all of which involve lasers. Fiber-optic bundles are used to transmit light through both rigid and flexible endoscopes in order to illuminate surgical sites inside the body. In flexible endoscopes, a second fiber-optic bundle transmits a view of the surgical site back to the clinician. In laser procedures, a single fiber is used to deliver the energy to cut, coagulate, and ablate tissue. This section addresses the application of fibers in delivering laser energy to the surgical site.

There are several different types of fiber-optic tip configurations used in medicine (Figure 24). In a standard optical fiber, the light is delivered axially out of the flat, distal end. This is the most common configuration used in laser medicine. Ball tip fibers are used during Ho:YAG laser lithotripsy. The smooth, rounded tip is designed to eliminate damage to the inner lining of the working channel in the ureterscope during insertion of the fiber. Tapered tips can be used at either of the proximal ends of the fibers to enable coupling of a large laser beam into the fiber. Side-firing fiber tips are assembled by integrating an angled mirror into the fiber optics applicator to reflect the light at 90° at the distal end. These fibers are used in urology during endoscopic frequency-doubled Nd:YAG (“Greenlight”) laser ablation of enlarged prostates for BPH treatment. Frequently, during endoscopic procedures in confined spaces (e.g. the urethra), there is insufficient space to bend the entire optical fiber so a side-firing tip provides flexible light delivery options. A diffusing fiber-optic tip enables cylindrical delivery of light at 360° in all directions. The diffuser can be created by etching into the fiber cladding to allow leakage of light, embedding scattering particles into the fiber core, adding a frosted cap to enhance light scattering, and/or placing of a mirror at the end of the fiber. Clinical applications of diffusing fibers include low-power, Photodynamic Therapy (PDT) or light-based chemotherapy to destroy tumors, or higher power laser interstitial thermal therapy (LITT) for thermal coagulation and destruction of tumors and benign growths. In such cases, the fiber-optic applicator can be integrated into a needle and placed in the center of the tumor for maximum irradiation volume or placed within a balloon and inflated for anchoring the device within a tubular structure (e.g. for treatment of esophageal cancer).

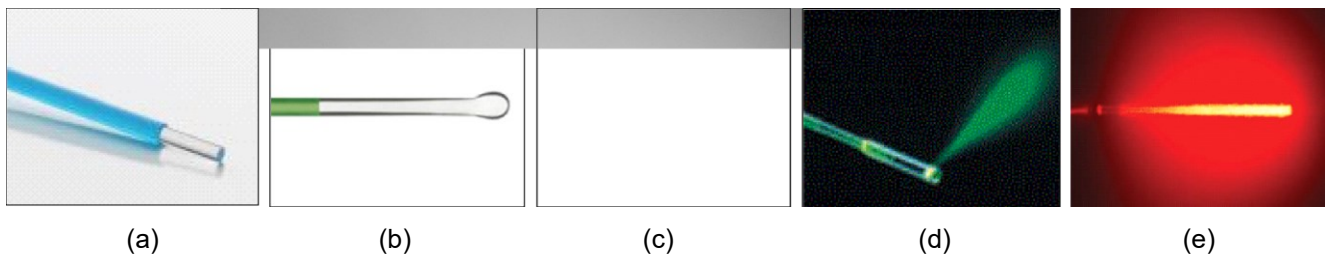


Figure 24. Some fiber-optic tip configurations used in medicine. (a) Standard optical fiber where light is delivered axially out of the flat distal end. (b) Ball-tip fiber. (c) Tapered fiber-optic tip. (d) Side-firing fiber-optic tip. (e) Diffusing fiber-optic tip.

Optical fibers provide a flexible means of delivering laser energy to otherwise inaccessible areas of the interior body. While fibers provide great utility, in most surgical applications, some of the unique characteristics of laser light can be lost as the energy passes through the fiber. While laser light retains its monochromatic characteristic after passing through an optical fiber, it may no longer be collimated. Light from a fiber can *diverge* or spread out rapidly, depending on the numerical aperture (NA) of the fiber, that is, on the *effective light-gathering cone* at the ends of the fiber.

While fibers can be made of many types of exotic materials for special applications, the vast majority of fibers are made of silica (glass). Most fibers consist of three distinct layers: *core*, *cladding*, and *buffer*. The core transmits the laser energy. The cladding, which has a lower index of refraction, is applied over the core and acts to keep the energy in the core from leaking out through a process called “total internal reflection.” The buffer is generally a tough plastic and acts as a protective covering to prevent damage to the fiber as it is handled and passed through endoscopes or similar secondary delivery devices.

The two primary types of fibers used are *silica/silica*, where both the core and cladding are made of glass, and *plastic clad silica* (PCS). In either case, the cladding material is selected so that it has a lower index of refraction than the core. Total internal reflection (TIR) is a function of the difference between the higher index of refraction from the core material and lower index of refraction from the cladding material. The difference in the indices of refraction creates a mirrorlike surface at the core-cladding interface. At this interface, a light ray is incident at an

angle θ . If θ is greater than a certain *critical angle*, θ_c , *all of the incident light* is reflected back into the fiber core. This reflection obeys the ordinary law of reflection, with the reflected angle equaling the incident angle treating the core-cladding interface as a *mirror*. If the incident angle θ is less than the critical angle θ_{crit} , the incident ray partially penetrates the interface in agreement with Snell's law, thereby losing photons to the cladding. As usual, according to Fresnel's equation, some of the light is also reflected, but not much, so the loss of laser light in the core is significant.

The critical angle can be derived from Snell's law of refraction ($n_1 \sin \theta_1 = n_2 \sin \theta_2$) for the case when the refracted angle, $\theta_2 = 90^\circ$, corresponds to light rays being reflected at the core-cladding interface instead of being refracted. Substituting $\theta_2 = 90^\circ$ results in $\sin 90^\circ = 1$, and then solving for θ_1 yields the definition for the critical angle. In this formula, for the specific case of optical fibers, θ_1 is renamed θ_c , n_1 becomes the n_{core} , and n_2 becomes the n_{clad} (Equation 12):

$$(12) \quad \theta_c = \sin^{-1} (n_{clad} / n_{core})$$

Again, light rays traveling inside the core of the optical fiber and incident on the core-cladding interface at angles greater than the critical angle (glancing to surface) will continue to be totally internally reflected, while light rays incident at a smaller angle (steeper to surface) will not be reflected and will instead be refracted through the interface and lost into the cladding (Figure 25).

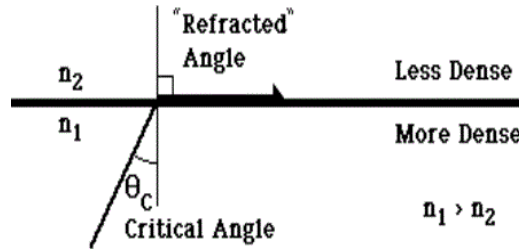


Figure 25. Definition of the critical angle θ_c . For $n_{core} > n_{clad}$, or $n_1 > n_2$, a condition exists at which the refracted angle, $\theta_2 = 90^\circ$. For this case, the incident angle, θ_1 , is then defined as the critical angle, θ_c . Light rays incident on the interface between the fiber-optic core and cladding at angles greater than θ_c will be reflected, while light rays incident at a smaller angle than θ_c will be refracted and lost through the core-cladding interface.

Note that Total Internal Reflection (TIR) of the light can only occur if the core material has a higher refractive index than the cladding material ($n_{core} > n_{clad}$), the primary condition for TIR.

Severe bending of the fiber will change the angle of incidence and can affect the amount of power lost. Also, the fiber material itself may result in attenuation due to both absorption and scattering losses in the glass. Losses in optical fibers are rated in *decibels per kilometer* (dB/km) in other fields, such as telecommunications. Since most surgical fibers are only a few meters in length, this rating means very little. The attenuation of light traveling down the fiber, A (dB/m), can be calculated from the following formula (Equation 13):

$$(13) \quad A = -10 \log (P_{out} / P_{in})$$

Alternatively, if the attenuation (dB/m) of the material is known, then the fraction of power emitted through the fiber, or Transmission ($T = P_{out} / P_{in}$), can also be calculated by rearranging the equation (Equation 14):

$$(14) \quad T = 10^{(-A/10)}$$

Losses due to reflection occur at the fiber ends, and reflections occurring at the air-core interface account for most of the normal losses. For example, Fresnel reflection losses at each fiber end can be calculated, if refractive indices of each medium are known for the interface (Equation 15):

$$(15) \quad R = [(n_1 - n_2) / (n_1 + n_2)]^2$$

For an air-glass interface, $n_1 = 1$ and $n_2 = 1.45$, yielding reflection losses of 0.034 or 3.4% at each of the proximal and distal fiber tips. Thus, 100% of the laser power can never be transmitted down an optical fiber even if absorption losses in the fiber are negligible. Antireflection-coated optical fibers have recently become available, which reduce losses from 3.5% to less than 0.3%, but are limited to operation at low powers. Most recently, structured fiber tips have been used for high-power applications to reduce reflection losses from 3.4% to less than 0.01%. It is important to note that the reflection losses change with laser wavelength used, since refractive index values for materials are dependent on wavelength.

Example 8. Germanium oxide mid-IR fibers are used in Er:YAG laser dental applications. If the refractive index of germanium oxide is $n = 1.84$ and the material attenuation is 0.5 dB/m, calculate the amount of light transmitted through a 1 meter length of the fiber. Be sure to include losses due to end reflections and attenuation within the fiber. Assume 1 Watt of power from the laser.

Solution: First calculate the percent loss due to end reflections for air ($n = 1$):

$$R = [(n_1 - n_2) / (n_1 + n_2)]^2 = [(1 - 1.84) / (1 + 1.84)]^2 = [(-0.84) / (2.84)]^2 = (-0.296)^2 = 0.088 = 8.8\%$$

Then calculate the fraction of light transmitted through the fiber:

$$T = 10^{(-A/10)} = 10^{(-0.5/10)} = 0.89 = 89\%$$

Now add up the losses as the light travels into, through, and out of the fiber:

Initial Power = 1 W

Power into fiber tip after air-fiber interface = (1 W) (100% - 8.8%) = (1 W) (0.912) = 0.912 W

Power down 1 meter long fiber = (0.912 W) (89%) = (0.912 W) (0.89) = 0.812 W

Power out fiber tip after fiber-air interface = (0.812 W) (100% - 8.8%) = (0.812 W) (0.912) = 0.741 W

So total power out of the mid-IR specialty fiber is 0.741 W or 74.1% of initial power from the laser.

Note that for comparison, low-loss standard silica optical fibers with lower refractive index and attenuation transmit visible or near-IR light with transmission rates closer to 90%.

There is also a phenomenon called cladding modes where light that leaks from the core becomes trapped between the cladding and buffer. In longer fibers this light usually leaks out of the cladding but in short surgical fibers it can be transmitted to the surgical site. This can sometimes be seen as a halo effect when viewing the transmitted energy.

Fiber-optic core diameters can vary from a few micrometers to over a millimeter. Most surgical fibers are at the larger end of the scale (200-600 μm). The larger sizes are used for a number of reasons. Larger fibers, while more rigid, are also more robust and less likely to fracture when passed through a secondary device. Many fibers are used in contact with tissue and, as such, need to be fairly rigid. Also, most higher-power lasers have large-diameter beams. The fibers selected for use with a particular laser must be large enough to accept the spot size to which the beam can be focused and have a numerical aperture (NA). This determines the entrance "cone angle of acceptance." The entrance cone angle is shown as angle θ_{acc} in Figure 26. The minimum spot

size to which a laser beam can be focused is a function of the wavelength and the angle of divergence of the unfocused beam. For the typical air medium between the laser and fiber, the numerical aperture (NA) is given by (Equation 16):

$$(16) \quad NA = \sin \theta_{acc} = (n_{core}^2 - n_{clad}^2)^{1/2}$$

Changing the refractive index of the core or cladding can be achieved by doping the glass with other materials. Typical NA values for optical fibers used in medicine vary from 0.22 – 0.66. A large value accepts a wide cone of light on the input end, but also results in a wide cone of diverging light on the output end, which is important to consider when working in noncontact mode with a certain working distance and desired intensity.

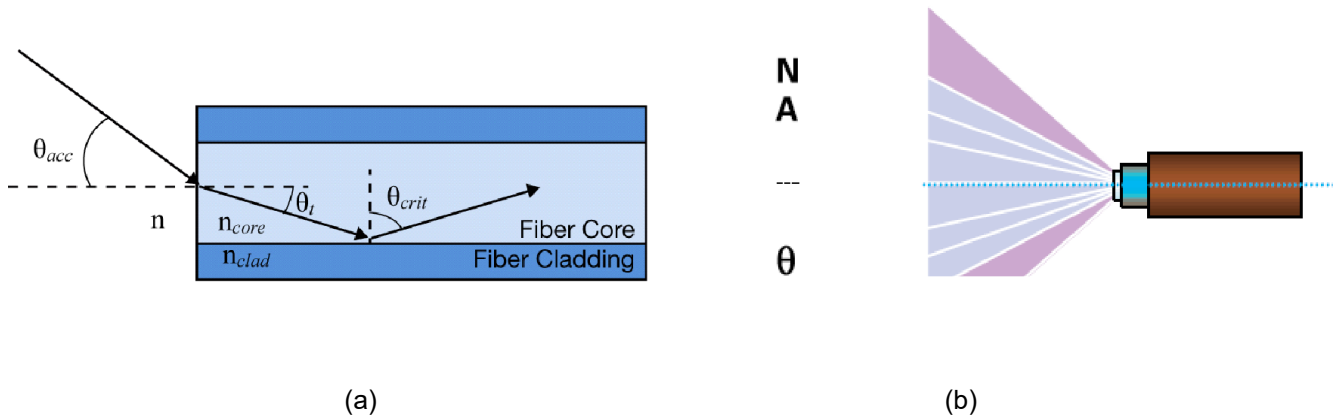


Figure 26. Numerical Aperture (NA). (a) The NA measures extent of acceptance cone of light for an optical fiber and is dependent on refractive indices of core and cladding materials. (b) Each NA value corresponds to the half angle (θ) in degrees of the acceptance cone of light into the fiber.

Example 9. An optical fiber has refractive indices of 1.50 and 1.45 for the core and cladding layers, respectively. Calculate the critical angle, numerical aperture, and acceptance half angle for the fiber.

Solution:

Critical Angle: $\theta_{crit} = \sin^{-1} (n_{clad} / n_{core}) = \sin^{-1} (1.45 / 1.50) = \sin^{-1} (0.97) = 75.2^\circ$

Numerical Aperture: $NA = (n_{core}^2 - n_{clad}^2)^{1/2} = [(1.50)^2 - (1.45)^2]^{1/2} = (2.25 - 2.10)^{1/2} = 0.39$

Acceptance Half Angle: $\theta_{acc} = \sin^{-1}(NA) = \sin^{-1}(0.39) = 23^\circ$

Once the laser energy has been transmitted through the fiber, it can be used in two ways. It can be applied in either a *free-beam* or a *contact* mode. Using just the fiber in the free-beam mode, the energy can be directed onto the target tissue without making contact. The spot size applied to the tissue can be varied by adjusting the distance from the fiber tip to the tissue. Varying the diameter of the spot size dramatically affects the irradiance of the incident energy, as we have already seen (Example 5). Alternately, the light diverging from the fiber can be captured by a lens, or a series of lenses, to recollimate or focus the light for use with a handpiece, microscope, slit lamp, or other secondary device.

In contact mode, the fiber itself or a special tip can be used in contact with tissue. In essence, when used in the contact mode, the laser becomes nothing more than a very expensive heat source. The tip of a bare fiber can be used directly in contact with tissue for cutting or ablation. Alternatively, a special sapphire or diamond tip can be attached to the ferrule of a jacketed fiber. In either case, the tip will become coated with carbonized matter. Many tips are manufactured with a frosted or precarbonized surface. Generally, these tips are carbonized prior to use on a patient. The laser energy is absorbed by the carbonized matter, thereby heating the tip to a very high temperature. Once carbonized, very little of the laser wavelength escapes the tip or fiber to interact with the surrounding tissue. For this reason, when used in a contact mode, there is no appreciable difference in tissue effects between lasers of different wavelengths. There are a variety of special tip shapes available for contact applications including hemispherical, cylindrical, conical, flat, and chisel shapes.

4.5 Hollow waveguides

A *hollow waveguide* is essentially a special case of the optical fiber. Used primarily with the CO₂ and Er:YAG lasers, a waveguide is basically a fiber without a core, or a core of air. Since the mid-IR wavelength of the CO₂ laser does not transmit through glass or other materials suitable for making a flexible fiber, one viable option is to use a flexible hollow tube with a reflective lining. These waveguides have high losses compared to glass fibers and are limited in length, with most being under two meters. An air purge is used to help cool the waveguide, prevent damage to the coating, and prevent debris from collecting on and eventually clogging the surgical tip. Since the CO₂ laser is always used in a noncontact mode, this type of delivery system is never used in contact with tissue. The Er:YAG laser, using a hollow waveguide, may have a sapphire tip at the distal end to allow contact applications such as soft tissue cutting or cavity removal in dentistry.

Table 7 provides a summary of the major delivery systems.

Table 7. Types of Delivery Systems.

System	Wavelength Range (nm)	Strengths	Limitations	Comments
Articulated Arm	UV – Mid-IR	High-power delivery	Cannot be used inside body Large and not flexible	Used with erbium and CO ₂ lasers, scanners
Silica Fiber	240 – 2500	Low cost, flexible, biocompatible, low-end reflection losses (n=1.45)	Cannot be used with UV and mid-IR lasers	Visible and Near-IR lasers
Sapphire Fiber	500 – 3100	High melting temperature, strong material, biocompatible	Rigid, high-end reflection losses (n=1.71), high cost, high transmission losses	Erbium laser dental tips, windows for contact cooling
Germanium Oxide Fiber	1000 – 3200	High-power delivery Flexible fiber Low bending losses	High-end reflection losses (n=1.84), low melting temperature, limited biocompatibility	Trunk fiber for erbium dental lasers
Zirconium Fluoride Fiber	285 – 5500	Low transmission losses Flexible	Hygroscopic, degrades in fluids, not biocompatible	Erbium lasers
Chalcogenide Fiber	1100 – 6500	Low transmission losses	High-end reflection losses (n=2.9), low melting temperature	Mid-IR lasers for low power and sensing applications
Hollow Silica Waveguide	900 – 25,000	High-power delivery, no end reflection losses	High bending losses, need to be sealed for use in body	Erbium and CO ₂ lasers for high-power delivery
Photonic Crystal Fiber	800 – 2000	High-peak power delivery, low transmission losses, no end reflection losses	High cost, need to be sealed for use in body	Ultrashort pulse lasers

Plastic fiber	500 - 1300	Inexpensive, flexible, easy-to-cleave ends	Low-power transmission	Sensing, illumination, data links
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Self-Test

37. Why is silica glass a good material to use for optical fibers?
 (a) Wide transmission range (b) High melting temperature (c) Inexpensive (d) Flexible
 (e) All of the above
38. What property needs to be satisfied for total internal reflection in an optical fiber to occur?
 (a) $n_{\text{core}} > n_{\text{clad}}$ (b) $n_{\text{clad}} > n_{\text{core}}$ (c) $n_{\text{core}} = n_{\text{clad}}$
39. Which of the following contributes to light losses for an optical fiber?
 (a) End reflections (b) Absorption due to glass impurities (c) Tight fiber bending (d) All of the above
40. What percentage of light is typically reflected at an air-glass interface for a silica fiber?
 (a) 0.5% (b) 4% (c) 10% (d) 20% (e) 50%
41. What can be done to decrease back-reflection losses at the surfaces of optical fibers?
 (a) Add an antireflection coating (b) Etch glass surface
 (c) Use an index-matching fluid or gel (d) All of the above
42. What is an articulated arm?
 (a) Series of hollow tubes, joints, and mirrors (b) Flexible, hollow tube
 (c) Glass rod (d) Long, thin, flexible glass tube
43. What advantages does a hollow waveguide provide over a standard silica optical fiber?
 (a) No end reflection losses (b) High-power delivery (c) Mid-IR light delivery (d) All of the above
44. What limitations does a hollow waveguide have?
 (a) Not very flexible (b) Needs to be sealed for use in fluid environment
 (c) High bending losses (d) All of the above
45. What is a typical value for the numerical aperture of an optical fiber?
 (a) 0.05 (b) 0.22 (c) 0.95 (d) 1.0 (e) 5
46. If the working distance between the optical fiber tip and the tissue is doubled, how does that effect the irradiance (I) at the tissue surface if the power remains the same?
 (a) I increases by factor of 4 (b) I increases by factor of 2 (c) I is unchanged
 (d) I is halved (e) I decreases by factor of 4.
47. Light exiting the distal end of an optical fiber is _____.
 (a) Collimated (b) Converging (c) Diverging
48. Light exiting the distal end of an optical fiber has a spatial beam profile with _____ shape.
 (a) Square (b) Rectangular (c) Circular (d) Hexagonal (e) Linear

49. Which light delivery system can NOT be used with the CO₂ laser?
 (a) Articulated arm (b) Free-beam (c) Silica fiber (d) Hollow waveguide
50. Which light delivery system is the most flexible for endoscopic applications inside the body?
 (a) Sapphire fiber (b) Articulated arm (c) Hollow waveguide (d) Silica fiber
51. How can a continuous-wave laser mimic a pulsed laser during operation?
 (a) Scan the beam to reduce dwell time (b) Split single beam into multiple beams
 (c) Cool the tissue during operation (d) All of the above

5. Laser Accessories

5.1 Introduction

Laser accessories are secondary delivery devices. They connect to the primary delivery device, such as an articulated arm or optical fiber, or they provide a delivery mechanism for an optical fiber. These devices include handpieces, microscopes or adaptors for microscopes, rigid and flexible endoscopes, scanners, and catheters. It is these devices that make feasible such a broad range of applications for laser energy. Whether it is a handheld scanner for removing wrinkles or unwanted hair or a special catheter for opening blocked coronary arteries, it is the accessory that gets the energy where it is needed and may even control the exposure.

5.2 Handpieces

The simplest of these secondary delivery devices is the *handpiece*. A handpiece may attach to the distal end of an articulated arm or it may be designed for use with an optical fiber. Many handpieces contain one or more lenses that can focus and/or defocus the beam. The optics in a handpiece, or any other accessory, must be designed for use with the particular laser's wavelength. The simplest handpiece provides a means of holding on to an optical fiber. The most sophisticated handpieces actually scan the laser output over a treatment site.

The simplest forms of the handpiece are used with optical fibers. While focusing handpieces are available, contact and nonfocusing handpieces are far more common. For contact surgery, the handpiece is simply a stylus, generally with a thin metal tube protruding from its distal end. The metal tubes can range from as short as 10 or 20 mm to as long as 300 or 400 mm, depending on their intended application. An unjacketed surgical fiber is inserted through the handpiece so that its tip extends beyond the tubular metal tip. A single-use fiber may have the handpiece permanently attached. In other cases, there is a clamping device on the proximal end of the handpiece, thereby allowing adjustment of the fiber length and a tightening to hold the fiber in place. Handpieces with short tips are used for a variety of surgical applications. Those with long (200–400 mm) tips are generally used with rigid endoscopes such as in laparoscopy.

A noncontact fiber-optic handpiece may not contain any optics, but simply provide a means to hold the fiber for free-beam delivery of the diverging laser energy. The handpiece is usually a straight tube with a clear plastic tip. As with the contact surgical handpiece, the fiber passes through a clamping device. The tip of the fiber is recessed inside of the handpiece a distance sufficient to provide the desired spot size at the end of the handpiece tube. The handpiece is generally pressed against the treatment site and the laser is fired. The handpiece is then moved to a different site and the laser fired again. These are commonly used with a variety of wavelengths for treating superficial conditions such as vascular lesions, age spots, or hair removal.

More complex fiber-optic handpieces use lenses to either collimate or focus the fiber's output. By collimating the output of the laser, a consistent spot size can be achieved without undue concern for the distance of the handpiece from the treatment site. This design is unnecessarily complex for the majority of unfocused,

noncontact fiber-optic lasers applications. While not a surgical use, this type of handpiece was used with the first blue argon lasers, developed for curing light-activated resins used in dentistry. By refocusing the beam, fairly small spot sizes can be achieved. This type of handpiece is designed for strictly noncontact use in cutting or ablating tissue. Since many of the more common fiber-delivered wavelengths can penetrate a significant distance into tissue, the surgical technique with these handpieces is very important.

Unlike the more common, noncontact applications, more of the laser's energy is absorbed during the incision/ablation due primarily to absorption in the plume and carbonized tissue. However, the more defocused the beam becomes, the more likely collateral damage occurs. Because of this, focusing handpieces for fiber-delivered wavelengths are not widely used.

More commonly used are focusing handpieces for articulated-arm-delivered lasers, primarily the CO₂ laser. In its simplest form this handpiece is a tubular housing with a single lens that focuses the laser beam. A conical tube or handpiece barrel somewhat shorter than the focal length of the lens is attached to the lens housing. This provides a handhold and places the focus of the beam at a comfortable distance for the clinician. Usually, a removable tip is provided to help the clinician judge the distance for optimal focus of the beam. The size of the spot applied to tissue is controlled by adjusting the distance the handpiece is held at from the target tissue. For procedures requiring a sterile field, the lens housing and arm can be covered with a sterile drape and the barrel and tip can be sterilized.

There are various focal lengths available for these handpieces. For CO₂ lasers, these measure 50 mm, 125 mm, and 200 mm. Some adjustable handpieces use a two-lens system to provide a variable spot size. The short 50 mm handpiece, while it will provide the smallest spot size, is generally used for working with the beam defocused. The short focal length allows the clinician to defocus the beam by withdrawing the handpiece a short distance from the treatment site. This is also the principal application for adjustable-focus handpieces.

The *125-mm handpiece* is the workhorse for freehand surgery. It is ideal for incisional and excisional cutting and can be moderately defocused by withdrawing the handpiece a reasonable distance. The longer 200 mm handpiece is fairly rare, even though it provides a little better reach in some applications. One problem shared by all focusing handpieces is the presence of a laser plume. Due to the density of the laser energy and the explosive action of tissue being vaporized, the laser plume will invariably travel back along the beam path and up the handpiece barrel where the debris will tend to collect on the focusing lens. If the plume debris collects on the focusing lens and absorbs laser energy, the antireflective coatings on the lens and the lens itself can be damaged. This problem is overcome by using a purge gas.

All focusing laser handpieces have a fitting for a tube which will supply a purge gas just beyond the focusing lens. The flow of the purge gas creates a positive pressure within the handpiece barrel, preventing the laser plume from entering. At higher flow rates, the purge gas displaces the plume from the surgical field. However, such use scatters the plume, making it more difficult to capture it with suction or evacuation equipment.

5.3 Microscopes

Surgical microscopes are widely used in a variety of medical specialties (Figure 27). They provide the surgeon with an illuminated, magnified view of the treatment site. One of the first applications of lasers in medicine was in *ophthalmology*. For this, a laser adapted to a slit lamp is used to deliver the laser energy to the treatment site within the eye. To visualize the treatment site, an ophthalmologist uses a microscope with a slit lamp to illuminate the retina or other internal structures. Since the microscope cannot be manipulated on three axes, a mirror connected to a small joystick allows movement of the beam throughout the microscope's field of view. A focusing mechanism allows adjustment of the focal spot within the eye. In ophthalmology, the focusing power of the eye's own lens must also be considered. Today, Nd:YAG lasers mounted to slit lamps are commonplace.

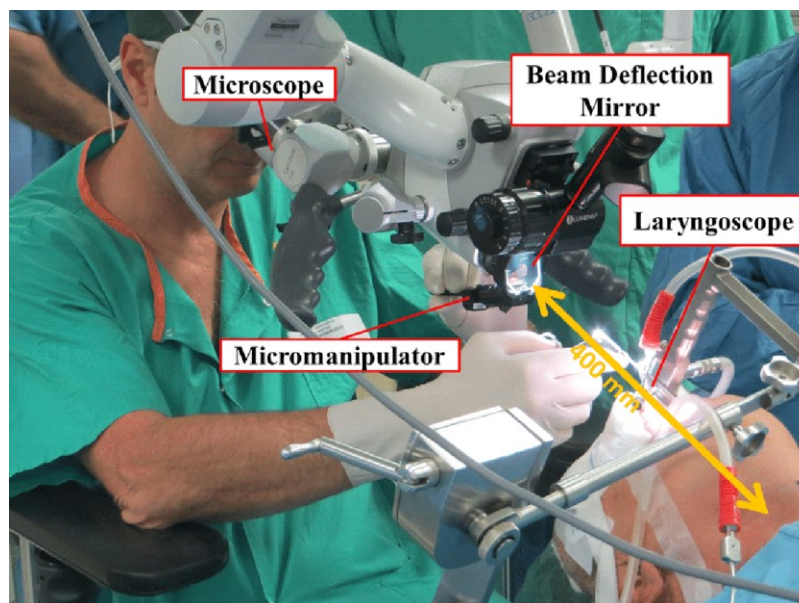


Figure 27. A modern CO₂ laser micromanipulator use in microsurgery.

Surgical microscopes provide a stable platform for precise application of laser energy. The most common surgical applications for microscope-mounted lasers are in otolaryngology, neurosurgery, and gynecology. Often referred to as a *microman* (short for *micromanipulator*) or *microslad* (short for *microscope laser adaptor device*), these devices are integrated with a microscope on an “as-needed” basis. They are available for both fiber- and arm-delivered wavelengths. The optic train consists of two lenses mounted in a telescopelike focusing device. One lens is moved relative to the other to focus or defocus the laser beam at the surgical site.

Ideally, the longest focal length is matched to the microscope’s objective lens and movement of the micromanipulator’s focusing telescope moves the focus closer to the microscope and away from the surgical site. The beam is then reflected by a movable mirror connected to a joystick used to direct the laser beam.

To be used with a microscope, an adaptor for the micromanipulator must be installed on the microscope. Generally, this is attached around the objective lens of the microscope. In most cases, the microscope’s objective lens is removed and an adaptor ring threaded in its place. The objective lens is then screwed into the adaptor ring and the micromanipulator clamps to the adaptor ring.

Surgical microscopes and micromanipulators are available with a variety of focal length lenses—200, 300, or 400 mm focal lengths being the most common. It is important that the focal length of the microscope’s objective lens be matched to that of the micromanipulator. It is also important that the laser/microscope be set at the proper distance from the surgical site for everything to work properly.

When first starting to use a laser with a microscope, it is not uncommon to find that the sharpest focus of the microscope is not set at the distance of the objective lens. In nonlaser surgery, it is common for the microscope to be set at an approximate distance from the patient and the oculars then adjusted so the clinician sees clearly. This simply does not work when the focus of the microscope and the optimal focus of the laser must coincide. If the oculars are set incorrectly, the microscope may be set too far away to achieve the optimal spot size for cutting. If too close, the optimal focus may be midway through the range of the focus-defocus adjustment.

Microscope-adapted lasers may be used as an adjunct in conventional surgery or they may be the principal tool. They are often used with a speculum for gynecological or rectal surgery. In otolaryngology, they can be used through a suspension laryngoscope or nasal speculum. Since these accessories are generally shining metal, accidental reflections may occur, especially when working in the airway.

5.4 Endoscopes

An *endoscope* is a rigid or flexible viewing instrument for insertion through a natural opening or orifice inside the body (like the nose, mouth, ears, rectum, and urethra). They may or may not have a hollow working channel, or *lumen*, through which a surgical instrument, or in this case a laser fiber, may be passed. Those discussed here do have such a lumen. Some endoscopes can be adapted for use with either an articulated arm or a fiber-optic delivery system. Flexible endoscopes and some rigid ones will be limited to use with fiber-optic lasers.

Rigid endoscopes—Rigid endoscopes are inserted either through a small puncture or through a natural bodily orifice. They are conventional surgical instruments that can be used with or adapted for use with a laser. All operating scopes have an open channel through which surgical instruments are passed. They also have a fiber-optic bundle that passes light to the distal end of the scope to illuminate the treatment site (Figure 28). If used with fiber optics, no modification or coupler is generally necessary. If they are used with a CO₂ or other free-beam laser, a laser coupler must be used to adapt the endoscope to the laser. Depending on the type of application, a CO₂ laser coupler may be little more than a focusing lens situated in a housing. In other cases, the coupler may be a combination surgical microscope and micromanipulator used to direct the laser beam down the endoscope lumen.

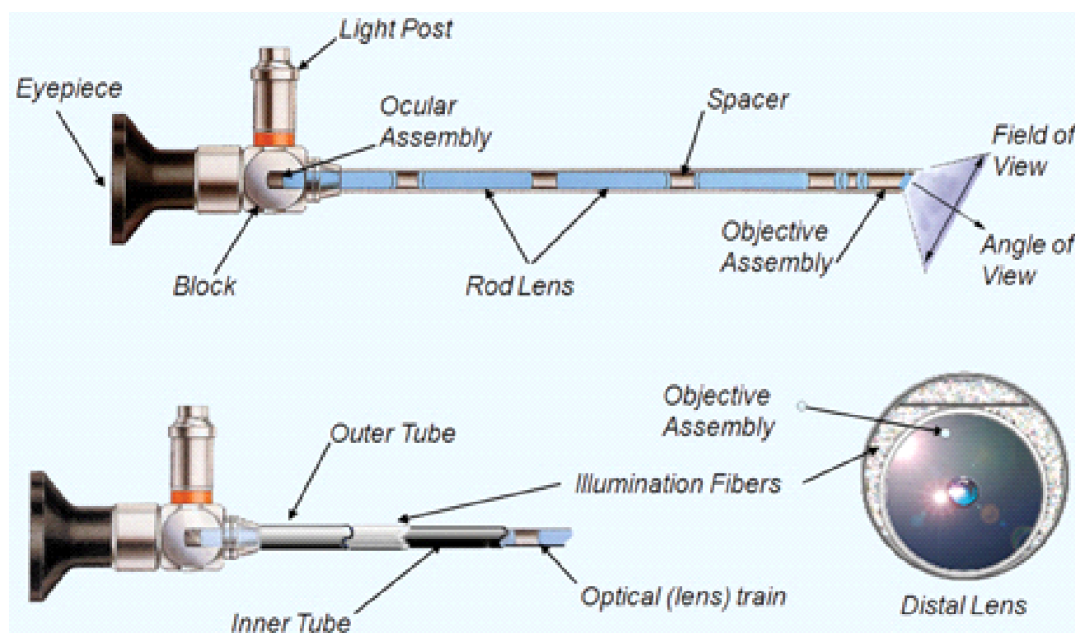


Figure 28. Basic components of a rigid endoscope.

Flexible endoscopes—Like rigid endoscopes, *flexible endoscopes* have a fiber bundle for delivering light for illumination to the treatment site. They also have one or more operating lumens through which instruments or a laser fiber may be passed (Figure 29). Unlike the rigid endoscopes, they use a bundle of optical fibers to capture and transmit an image to the clinician, and of course, they are semi-flexible. They must be stiff enough to be pushed through the internal passages of the body and flexible enough to be maneuvered. Some have a tip that can be flexed or deflected on two axes to facilitate viewing and to help in guiding the scope around bends as it passes through the bowel, lungs, etc. The most flexible bundles can be turned to look back on themselves.

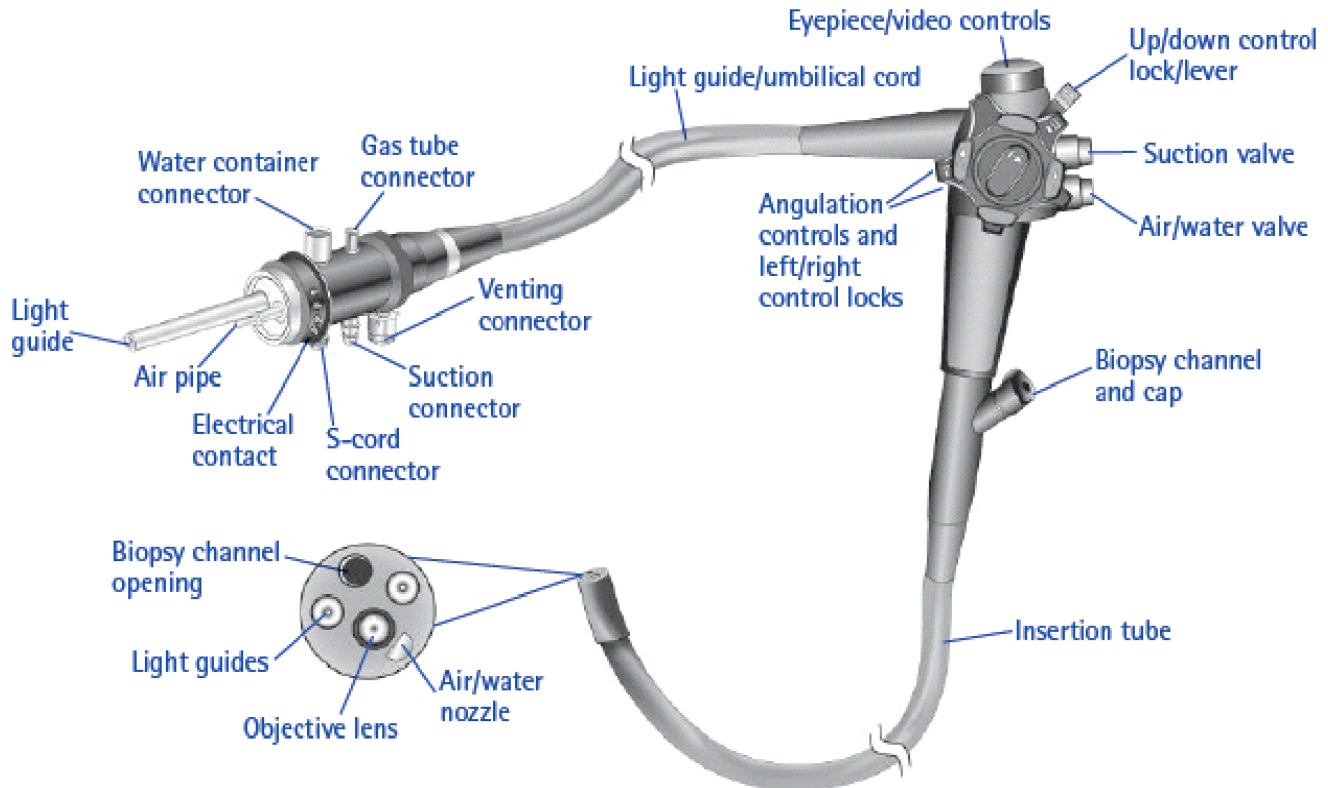


Figure 29. Basic components of a flexible endoscope.

The primary difference between the different semi-flexible endoscopes is size. They come in various lengths and diameters, with the longest being the *colonoscopes*, some two meters in length and about 15 mm in diameter. The *ureterscope* is small enough to be passed through the bladder and up the urethra to the kidneys. *Gastrosopes*, typically about a meter in length and 12 to 14 mm in diameter, are used in the upper digestive tract, including the esophagus, stomach, and duodenum. *Bronchoscopes* generally have a working length of about a 500–600 mm. Their diameters range from less than 3 mm to 5 mm. The smaller diameters are small enough for insertion into the peripheral bronchi.

Digital endoscopes—The latest developments in endoscopes include the use of digital endoscopes. The endoscopes replace the broadband lamp and fiber-optic bundle for illumination with light-emitting diodes (LEDs) at the distal tip of the endoscope. This eliminates white balancing issues and potential loss of lighting from individual broken fibers due to wear and tear with reuse. The fiber-optic bundle used for detection is also replaced with a CMOS chip on the distal end of the endoscope as well. This “chip-on-a-tip” design provides superior image quality and eliminates the segmented images produced by fiber-optic bundles. It also eliminates the loss of image quality when individual fibers break. The major advantage of the digital endoscope is the superior lighting and image quality. However, the CMOS chips are currently still larger than fiber-optic bundles, limiting flexibility.

Disposable endoscopes—Most recently, reusable endoscopes have been replaced with single-use endoscopes for some procedures. Such disposable endoscopes provide several advantages, including reduced capital cost, elimination of high repair costs, and elimination of resterilization and with it the risk of antibiotic-resistant hospital infections. Surgeons may have several disposable endoscopes with different bending and viewing angles available to choose from during a given procedure. However, it is unclear if they create more biohazard waste from an environmental standpoint.

Cystoscopes—The *cystoscope* is a simple adaptation of a laser endoscope for surgery. Used to access the bladder or prostate, the cystoscope sheath is passed through the urethra. This sheath has a fitting for pressurizing and thus distending the bladder to effect visualization. This is accomplished with an irrigating fluid. Usually all that is needed is a bag of saline fluid hanging from an IV pole. An inflatable cuff may be fitted over the bag to provide additional pressure. Due to the fluid-filled field, cystoscopes are not suited for use with the CO₂ laser or other free-beam lasers. Unjacketed optical fibers or fibers fitted with contact tips can be passed through the operating channel to treat areas in the bladder or the prostate. Cystoscopes have a flexible device called an Albarran bridge that allows the clinician to deflect a flexible surgical instrument or an optical fiber to treat areas not directly in line with the cystoscope's lumen. A variety of lasers may be used in a noncontact mode with the cystoscope. This includes diode, Nd:YAG, Ho:YAG, and Th:YAG lasers.

Ureteroscopes—An *ureteroscope* is basically a longer, thinner version of a cystoscope. It is passed through the urethra into the bladder and then into the ureter to reach the kidney. Ureteroscopes are typically about 65 cm long. The principal use for laser applications is destruction and removal of kidney stones. Larger kidney stones may become stuck in the ureter just after leaving the kidney. The clinician may push the stone back up into the kidney. Once the stone is back in the kidney, the stone may be broken up. Conventional approaches involve the use of percutaneous nephrolithotomy (PCNL) or extracorporeal shock wave lithotripsy (ESWL) to crush the stones. Ho:YAG and thulium fiber laser lithotripters provide a means of fragmenting stones using a photothermal mechanism. The laser fiber is passed through the lumen and brought into contact with the stone. When the laser is fired, the laser pulses chip away at the stone. The sandlike residue then passes out through the urinary tract. Several laser operating modes are used, depending on the size of the stone and its location. "Fragmentation" mode utilizes high-pulse energies and low-pulse rates to fragment bladder stones into large chunks that are then removed from the urinary tract using a stone basket. "Dusting" mode utilizes low-pulse energy and high-pulse rates to reduce stone trapped in the ureter to dustlike particles capable of being flushed out of the urinary tract using saline irrigation. "Popcorning" utilizes intermediate laser pulse energies and pulse rates when multiple stones are present in the kidney. The laser is activated and the energy is delivered into a saline medium, creating turbulent flow, and the kidney stones are fragmented into many small pieces when they pass by the fiber tip.

Bronchoscopes—As their name implies, *bronchoscopes* are used to access the bronchial tubes, primarily the main bronchi. As mentioned earlier, these are conventional instruments that can be used with fiber-optic lasers or adapted to a free-beam laser such as the CO₂ by means of a laser coupler. Since mobility of the scope is limited by the fairly narrow and rigid structure, a laser coupler with a microscope and joystick-manipulated mirror is used to direct the CO₂ laser beam down the operating channel to the treatment site.

Rectoscopes—*Rectoscopes* are used to access areas in the rectum. While there is somewhat greater mobility in the rectum, these endoscopes may also use a coupler similar to the bronchoscope for the CO₂ laser.

Laparoscopes—The *laparoscope* is perhaps the most widely used of all the rigid endoscopes. Once used mainly in gynecology, they are now being used widely in other specialties as well. Both fiber-optic and free-beam lasers can be used with the laparoscope. Unlike the previously discussed endoscopes, laparoscopies involve at least one incision, or more accurately a puncture, made in the abdomen. This is usually, if not always, made through the umbilicus or navel, the naturally occurring opening in the musculature of the abdominal wall. More involved laparoscopies often require use of multiple instruments passed through additional punctures made through the muscles of the abdomen.

Laparoscopies, like cystoscopies, require distension to allow visualization of the organs and structures within the abdomen. This is most commonly achieved through the use of a pressurized gas, normally CO₂, though N₂O has also been used. To maintain intra-abdominal pressure, a seal must be maintained between the laparoscope and the abdominal wall. This is achieved by making a puncture rather than cutting into the

abdomen. The abdomen is first inflated by passing a hollow insufflation needle through the umbilicus. The needle is connected to a device called an insufflator which supplies the pressurized gas necessary to distend the abdomen. Regulation of the pressure is very important since too much pressure can inhibit respiration. Once the abdomen is adequately inflated, the needle is removed and a *trocar* (a large, sharp nail) is pushed through the umbilicus. The trocar is then withdrawn, leaving behind the trocar sheath and a hollow tube with a trumpet-type valve through which the laparoscope will be passed. The gas line from the insufflator would be switched to a fitting on the trocar sheath to maintain insufflation.

In most simple cases, an instrument or the laser can be passed through the operating lumen of the laparoscope. Very often two or more additional punctures are made to permit introduction of laser fibers, graspers, scissors, probes, or other instruments. Sometimes a laparoscope without a lumen is used. In that case, the laser or another instrument would be used through one of the additional punctures. For laser surgery, a surgical fiber may be passed through the operating lumen of the laparoscope or through one of the secondary punctures. Surgical fibers or contact tips are used exclusively in the contact mode during these procedures.

Use of a CO₂ laser with the laparoscope has presented a number of challenges. One of the early problems involved proper alignment of the beam to the laparoscope. Most older laparoscopes were not intended for use with lasers. Consequently, the proximal end to which the laser coupler was attached was not necessarily machined perpendicular to the lumen. This required some adjustment to ensure that the beam would pass through the scope. A second problem involved the alignment of the laser's articulated arm. Due to the length and small diameter of most laparoscope lumens, any irregularity in the arm's alignment would result in losing the beam as the laparoscope and arm were manipulated. A third problem had to do with interaction of the CO₂ wavelength and the insufflating gas.

If a CO₂ laser is to be used, a sealed coupler with the appropriate focal length lens is connected to the laparoscope. There are a number of designs available. Some of these designs feature a joystick which allow adjustment of the laser beam within the lumen; others are designed to automatically align the laser beam with the lumen. Provided the arm is properly aligned, any of these will work.

Historically, once the laparoscope was inserted into the trocar sheath, the insufflator was moved from the trocar sheath to a fitting on the laparoscope lumen. This was done to prevent the laser plume from entering the lumen and diffusing the laser energy or collecting on the optics.

However, this presented a second problem. As surgeons became more proficient in the use of the laser with the laparoscope, they began increasing the laser power. Rather than noticing an increase in cutting efficiency, they noticed an increase in the laser spot size and lower efficiency. The problem was that the gas being used to inflate the abdomen was the same gas being used to generate the laser wavelength, CO₂. As the laser beam passed down the lumen, a portion of the energy was absorbed by the insufflation gas. Since the beam irradiance profile of most CO₂ lasers is *Gaussian* (bell-shaped), the gas near the center of the beam was heated more than the gas near the edges. As the gas was heated, it expanded. Being partially trapped by the walls of the lumen, the gas became denser at the edges of the beam and less dense near its center. In effect, the laser beam passing through the CO₂ gas in the lumen created a gas lens that defocused the laser beam. The degree of defocus was dependent on the laser's power level.

To address this problem, a CO₂ laser was developed that used the C¹³ isotope of the CO₂ gas (along with any other isotope of CO₂ that could be used for a laser). The conventional CO₂ laser uses the C¹² isotope and produces a principal wavelength at 10,600 nm. The C¹³ isotope produces a wavelength at 11,100 nm, which is outside the absorption range of the insufflating gas. Another approach was taken using an inert, nonabsorbing gas such as argon at a very low flow rate as a purge gas in the laparoscope lumen. In this case, the insufflator remained connected to the trocar sheath. This approach worked well with the conventional CO₂ lasers and was far more economical.

Arthroscopes—Like cystoscopes and laparoscopes, *arthroscopes* require distension for visualization. They are used to visualize and treat problems within joints, primarily the shoulder and knee. These joints can be

distended with either gas or liquid. In most cases fluid is used, which means that the CO₂ laser cannot be used. Contact fibers or contact tips are most commonly used for surgery. As with the laparoscope, a puncture must be made to gain entry to the joint and a trocar sheath is used to pass the arthroscope into the joint.

5.5 Scanners

Scanners provide automated delivery of laser energy to tissue. They allow uniform application of laser energy over large areas that would be difficult, if not impossible, to achieve manually. They also provide repeatability in tissue effects. Generally, scanners are integrated to some extent with the laser's control circuitry. When the laser's on-off footswitch is depressed, the scanner will pass the beam over the treatment site. Movement of the beam is accomplished through use of one or two electronically controlled mirrors. Scanners may be hand-held, free standing, or microscope-mounted.

The simplest scanners are hand-held designs. They may scan in either a spiral or an x-y pattern. A scanner doesn't really have to be integrated with the laser if the duration of the laser exposure can be set to coincide with a single scan of the mirror, or if the scanning handpiece itself has a shutter that controls exposure. CO₂ and Er:YAG lasers are commonly used with hand-held scanners for these applications. Argon, frequency doubled Nd:YAG, copper, and gold vapor lasers have been used with this type of scanner for treating vascular lesions and, in some cases, tattoo removal. Figure 30 shows a galvanometer scanner, which used a series of small motorized mirror mounts for laser beam steering and scanning applications with high precision and accuracy. This is standard equipment for cosmetic laser applications such as skin resurfacing/wrinkle removal.

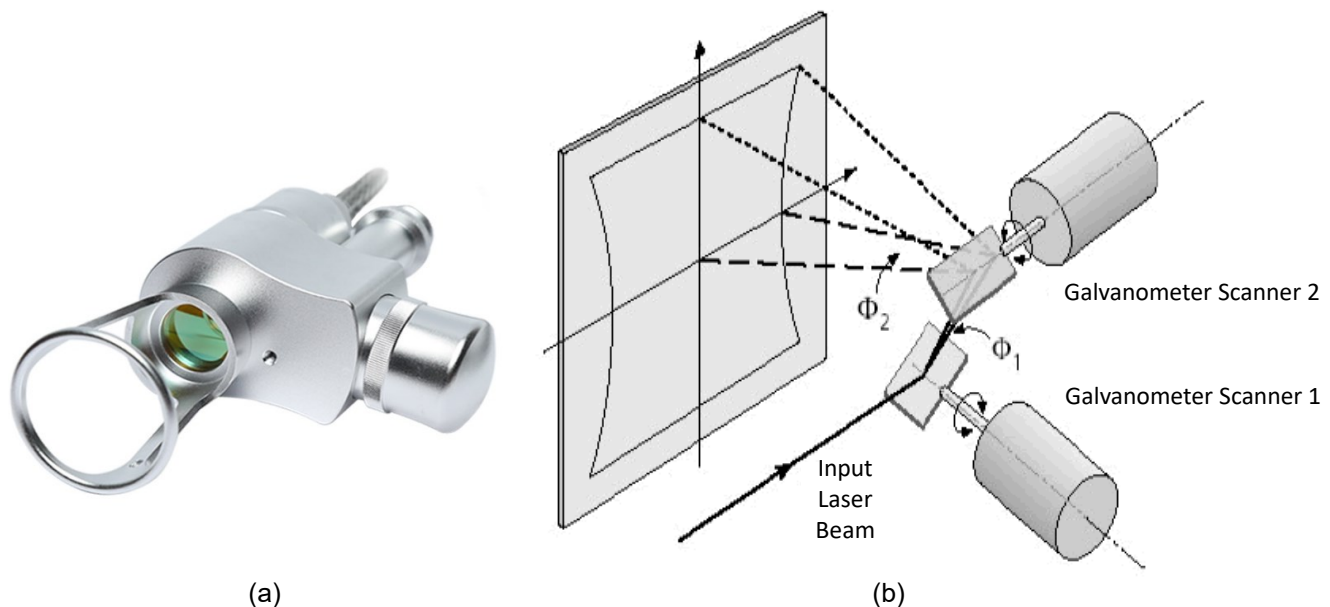


Figure 30. (a) Photograph and (b) diagram of a galvanometer scanner.

More sophisticated scanners are designed for use with the operating microscope. Microscope mounted scanners are generally programmable to some extent. These scanners may allow the operator to outline an irregular shape and designate separate areas within the larger area. When the footswitch is depressed, the laser fires as the scanner makes incremental passes over the area. The laser turns off, or the shutter closes, as the beam passes over the smaller, outlined areas, thereby avoiding application of laser energy to these sites.

One aspect of laser surgery that can be a problem is involuntary movement of the target tissue during noncontact laser application. This is most noticeable during procedures where patient respiration causes

movement of the target tissue. Attempts have been made to design a motion tracking system to increase the precision of laser incisions by tracking and predicting tissue movement.

Perhaps the most sophisticated scanners in common use today are those used for the laser-assisted in situ keratomileusis (LASIK) procedure. This application requires complete integration and control of the laser by a sophisticated computer. To achieve a simple spherical correction of the corneal surface, the laser must remove a precise amount of tissue from the cornea. The amount of tissue to be removed changes as the laser scans across the cornea. To perform *photorefractive keratectomy* (PRK) for myopia, where a spherical correction is required, more tissue is removed from the center of the cornea than is removed from the periphery. To address more complex abnormalities such as cylindrical corrections for astigmatism, the treatment becomes much more complex. A typical LASIK procedure may take twenty minutes to perform. The majority of that time is used to prepare the patient and program the laser equipment. The actual laser may fire for as few as thirty seconds or for as long as two minutes.

5.6 Laser catheters

Catheters provide a means of delivering laser energy to areas of the body that would otherwise be inaccessible by any other means short of major surgery. These catheters are generally introduced into an artery in the groin and directed to the blockage site (for example, the heart) under fluoroscopy. A successful example includes the introduction of laser optical fiber catheters into peripheral blood vessels for thermal coagulation and treatment of endovenous malformations such as varicose veins. Laser catheters have also been tested for the past few decades for laser angioplasty applications to ablate plaque in arteries or break up clots, but they have largely been unsuccessful due in part to complications like perforation of arteries and restenosis (where the artery becomes blocked again).

5.7 Connectors

Many laser manufacturers use accessory connections that are unique and proprietary to their systems. While one may use a standard thread to connect an accessory to an articulated arm, another may use a spring clip design, a twist lock, or a multistart thread. Fiber-optic lasers have different pin arrangements on their connectors to “tell” the laser which type of fiber is attached.

Some companies specialize in manufacturing laser accessories and can provide adaptors that will permit use of their product(s) on the most common lasers. Additionally, adaptors can usually be obtained that allow the use of one manufacturer's accessory with a laser from a different manufacturer. For this reason, it is important to ensure the compatibility of the accessory with the laser to be used. Mixing an accessory designed for one wavelength with a laser of a different wavelength may result in little to no effect on tissue as well as destruction of some very expensive optics. As long as there is wavelength compatibility, using accessories from one manufacturer with a laser from another is not typically a problem with simple accessories such as handpieces, microscope adaptors, or endoscopes. However, this could be problematic with more sophisticated devices. For example, if a scanner is designed to be used with a particular laser, it may interface with the laser's control circuitry for trigger, scan speed, and exposure time. Some may require installation of an interface PC card and/or upgrades in system software before the device can be used with a laser.

The most common fiber-optic connector is the SMA905 connector, where SMA stands for SubMiniatureA (Figure 31). These connectors are used for power delivery with large, multimode optical fibers having core diameters equal to or larger than 200 μm . This is a standard, threaded metal connector, where the glass or plastic optical fiber is permanently glued inside the core of the metal ferrule. Two fiber-optic cables can be connected using a mating connector. For very high laser power delivery, thermal effects in the standard SMA905 connector can result in melting the glue adhesive and a failure of the connector. Instead, a high-power SMA905 connector is used. This eliminates the glue adhesive and contains a thermally conductive copper sleeve in the design to remove heat buildup. In a standard SMA905 connector, the fiber can be misaligned and off-center by about 10 μm . This tolerance may be unacceptable for small laser spot sizes being coupled into

smaller fibers. Thus, for small optical fibers (less than 200 μm core diameter) and for low-power laser applications, an FC connector is used where FC stands for ferrule channel. A ceramic sheath is used, and the metal connector has a key insert along with the standard threading. FC connectors are commonly used with single-mode optical fibers (9/125 μm core/cladding outer diameter) in the telecommunications industry and for optical diagnostics in medicine (e.g. optical coherence tomography). FC/APC connectors are also used, where APC stands for angle polished cleave, at an angle of 8° to eliminate back-reflections.

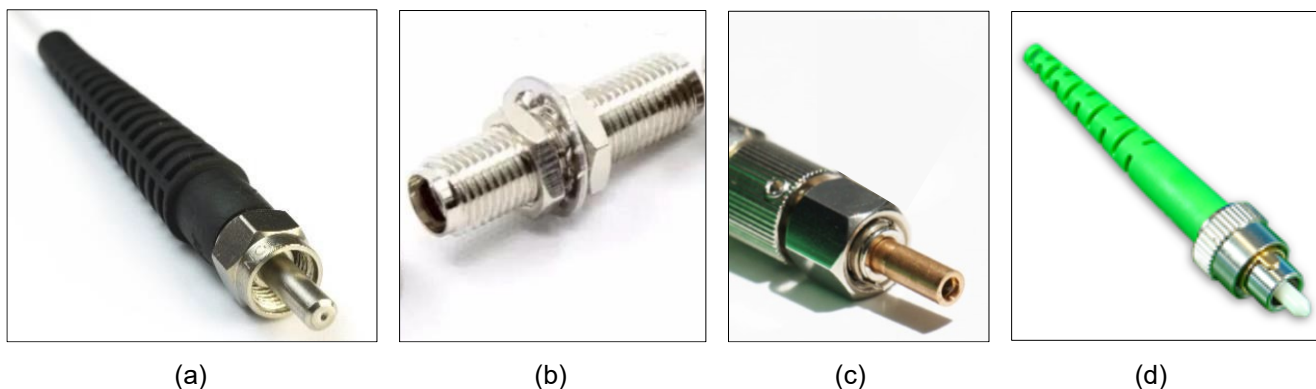


Figure 31. (a) SMA905 connector. (b) Mating connector. (c) High-power SMA905 connector. (d) FC connector.

Self-Test

52. What components of a standard endoscope use a fiber-optic bundle?

- (a) Illumination and detection channels
- (b) Working channel
- (c) Illumination channel
- (d) Detection channel

53. To access the kidney for removal of a kidney stone, which type of scope would be used?

- (a) Resectoscope
- (b) Laparoscope
- (c) Bronchoscope
- (d) Ureteroscope

54. Which of the following is an advantage of a disposable scope over a reusable scope?

- (a) Less expensive
- (b) Less wear and tear
- (c) No resterilization
- (d) Eliminates antibiotic resistant infections
- (e) All of the above

55. Which of the following is an advantage of a digital scope over a fiber-optic scope?

- (a) Superior image quality
- (b) Less expensive
- (c) Smaller scope tip
- (d) All of the above

56. What is the appropriate fiber-optic connector to use for a large, multimode optical fiber?

- (a) FC
- (b) FC/APC
- (c) SMA
- (d) All of these connectors would work

57. Which clinical application may require a scanner?

- (a) LASIK
- (b) Skin resurfacing
- (c) Tattoo removal
- (d) Port-wine stains
- (e) All of the above

Basic Laboratory: Exponential Beer's Law of Absorption

Objective

In this experiment, the student will measure the transmission of three filter materials as a function of filter thickness.

Equipment

Laser pointer or 1–2 mW red diode laser
Plastic filters of equal thickness (6 red, 6 blue, and 6 green)
Power meter with sensor head
Beam expander
Micrometer
Filter holder
Lab jack
Optical bench
Pin carriers
Micrometer

Procedure

Mount the laser on a lab jack near one end of the optical bench. Attach the beam expander to the output aperture (Figure 32).

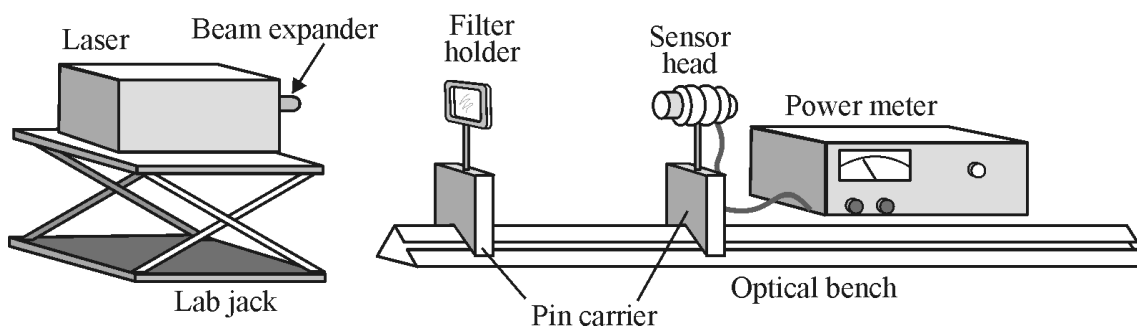


Figure 32. Experimental setup for absorption measurements.

Using a pin carrier, mount the sensor head of the power meter about 15 cm to the right of the laser. Between the laser and the sensor head, use another pin carrier to attach the filter holder to the bench, as illustrated in Figure 32.

With the micrometer, measure the thickness of several filters. Record the average thickness in the data table.

Using the proper safety precautions, plug in the laser and turn it on after the power meter and sensor have been properly placed and adjusted. Adjust the range selector on the power meter to 10 mw full scale, and then turn the power meter on. Position the sensor head in such a manner that the active area of the detector is fully illuminated; however, do not overfill the detector.

Measure and record the power output (P_o) of the laser. Place a blue filter sheet in the holder and measure the power transmitted (P_t) by the filter. Repeat with two, three, four, five, and six sheets of blue filter. Record the results in the data table.

Repeat Step 6 for green and red filters. Note: The range switch on the power meter will have to be adjusted to obtain some of the readings since the power levels to be measured will vary over a range from about 1–2 mW to < 100 μ W.

From the data obtained in Step 7, plot a curve of transmitted power versus thickness (number of filters) for each of the three types of color filters. The resultant curves should provide verification of the exponential law of absorption. Your first data point on each curve should be the output power of the laser, P_o , that is, the power when no filters are in the beam path.

Data Table: Exponential Law of Absorption.

Equipment	Manufacture/model number		
Laser optical power meter			
Laser power output, P _o _____mW			
Average filter thickness_____mm			
Number of filters	Transmitted power, P _t		
	Blue	Green	Red
1			
2			
3			
4			
5			
6			

Advanced CO₂ Laser Laboratory

Objective

The purpose of this lab is to become familiar with power settings, spot sizes, and time settings using a CO₂ laser and tongue depressor to represent the effects on tissue.

Equipment

Laser safety glasses for the appropriate laser wavelength(s) CO₂ laser system

125 mm handpiece with focusing guide/tip

Scalpel, gauze pads, cotton swabs

Wet towels and/or a bowl of water

Smoke evacuator or suction device with filter, various colored markers (black, blue, red, yellow)

Target materials: Tongue depressors—some dry, some soaked in water (the tongue depressor is the small wooden blade typically used by doctors in routine examinations of the mouth and throat)

Procedure

The CO₂ laser is used in a strictly free-beam or noncontact mode. This exercise may be performed with either the handpiece or a micromanipulator. While other accessories are available for models equipped with an articulated arm, the effects will be similar.

The CO₂ laser may be focused for incision/excision or defocused for ablation. The end results will depend largely on the power setting, spot size, and time on tissue.

In this lab you will fire the laser on a tongue depressor in lieu of actual tissue specimens (Figure 33). The effects of focused and defocused laser energy at various power levels will be noted. Any differences between tissue effects in lightly-pigmented and darkly-pigmented tissues will also be observed. Also, note the appearance of the aiming laser on the tongue depressor and other specimens. The differences, if any, between continuous and any available enhanced-pulse modes will also be observed.

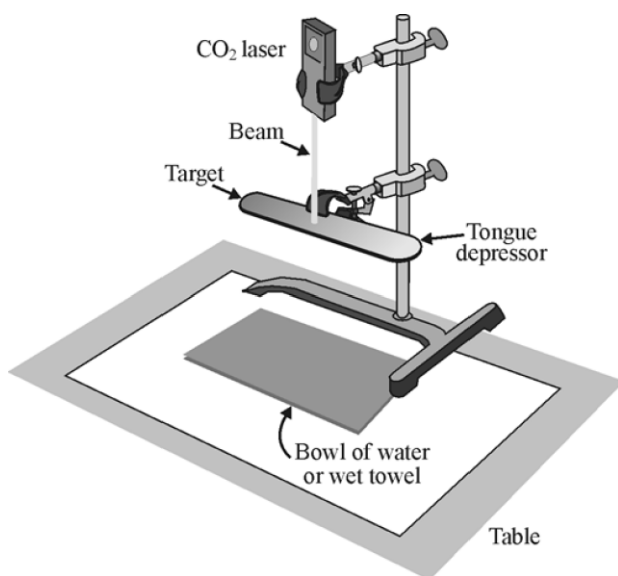


Figure 33. Lab setup.

To ensure safety, approved goggles and a smoke evacuator are required. It is best that at least two people work together during this lab. One person should act as the laser operator setting power, exposure, and mode while another is using the laser. If working in larger groups, each person should take a turn at each position.

Part I

Start with a dry tongue depressor as the target. Since a focused CO₂ laser can burn through a tongue depressor at relatively low powers, work over a wet towel or a container of water to absorb any excess energy. Never point the handpiece at anyone. To help control the odor of burning wood, hold the evacuator wand or suction device as close as possible to the impact point of the laser beam.

To begin, set the laser for 2–3 Watts. Select a single pulse of 0.1 second. Place the handpiece with its focusing guide against the tongue depressor. Depress the footswitch. Now put the laser in *standby* and inspect the results.

Did it burn through the tongue depressor?

Duplicate this with another tongue depressor that has been soaking in water to compare to living tissue. Wipe any excess water from the surface of the wood. How do the burns compare?

Now, hold the handpiece back an inch and fire the laser on both tongue depressors. Note the difference in the size of the burn. Repeat this while holding the handpiece at increasing distances until there is no effect on the tongue depressor. Did the water content affect the results?

Repeat this process for longer pulse widths and higher power settings of 5W, 10W, 20W. Note the differences that the increase in power makes. Switch the laser to repeat pulse, 2–3 Watts.

Depress the footswitch as you move the handpiece along the tongue depressor. Repeat using various powers and spot sizes.

Mark off an area on the tongue depressor of 1 cm² and try to uniformly ablate the surface leaving a uniform char layer.

Part II

Switch the laser to continuous-wave (cw) mode and repeat the previous exercise. Which mode is easier to control?

If SuperPulse or another enhanced-pulse mode is available, compare its effects with the CW mode using the same power level(s) and spot size(s).

Make a series of lines across a dry tongue depressor using colored markers. Use a beam sufficiently defocused so it barely marks the wood as you scan across the tongue depressor. Scan across the marks from lightest color to darkest. Does one color react more strongly than another? What can you deduce from this?

Prepare a detailed report of your results, comparing data and observations for Part I and Part II.

Answers to Self-Tests

1. b	2. c	3. a	4. e	5. d	6. b	7. b
8. e	9. a	10. c	11. a	12. c	13. a	14. a
15. e	16. c	17. c	18. e	19. d	20. d	21. d
22. e	23. a	24. b	25. c	26. e	27. c	28. c
29. a	30. c	31. d	32. b	33. e	34. c	35. e
36. c	37. e	38. a	39. d	40. b	41. d	42. a
43. d	44. d	45. b	46. e	47. c	48. c	49. c
50. d	51. a	52. a	53. d	54. e	55. a	56. c
57. e						

Module Review Questions

1. If a KTP crystal is used to double the frequency of an Nd:YAG laser at 1064 nm, yielding the second harmonic, what is the resulting wavelength?
 - a. 266 nm
 - b. 532 nm
 - c. 1064 nm
 - d. 2132 nm
2. For fixed laser power, what would cause beam intensity to increase by a factor of 4?
 - a. Increase radius of beam by factor of 2
 - b. Increase radius of beam by factor of 4
 - c. Decrease radius of beam by factor of 2
 - d. Decrease radius of beam by factor of 4
3. What is the approximate value for the refractive index, n , of soft tissue?
 - a. 1.0
 - b. 1.3
 - c. 1.4
 - d. 1.5
4. At mid-infrared laser wavelengths, what is the strongest absorber of light in tissue?
 - a. Water
 - b. Melanin
 - c. Hemoglobin
 - d. Proteins
5. What laser wavelength is best used for ultraprecise corneal shaping during LASIK?
 - a. 193 nm
 - b. 532 nm
 - c. 1064 nm
 - d. 10,600 nm
6. What laser wavelength would be best used for both tissue cutting and hemostasis?
 - a. 193 nm
 - b. 532 nm
 - c. 1064 nm
 - d. 10,600 nm
7. In what part of the spectrum does light scattering dominate absorption?
 - a. Ultraviolet
 - b. Visible
 - c. Near-infrared
 - d. Mid-infrared
8. The total attenuation coefficient is given by _____.
 - a. Absorption coefficient
 - b. Scattering coefficient
 - c. Sum of absorption and scattering coefficients
 - d. Absorption minus scattering coefficient
9. Which value for the anisotropy factor represents isotropic scattering?
 - a. 0
 - b. 0.5
 - c. 1

- d. - 1
10. If scattering coefficient is 100 cm^{-1} and $g = 0.9$, the reduced scattering coefficient is ____.
- 1 cm^{-1}
 - 10 cm^{-1}
 - 100 cm^{-1}
 - 1000 cm^{-1}
11. Which of the following surfaces would have the highest value for the albedo?
- Blood on the surface of the skin
 - Carbonized (blackened) tissue
 - Normal, dark-colored skin
 - Blanched (whitened) tissue from thermal coagulation
12. To calculate optical penetration depth in tissue, which parameters are needed?
- Absorption coefficient
 - Scattering coefficient
 - Anisotropy factor
 - All of the above
13. If the effective attenuation coefficient in tissue increases, optical penetration depth ____.
- Decreases
 - Stays the same
 - Increases
 - It depends on other factors
14. Which mid-infrared laser provides the most precise soft tissue ablation?
- Neodymium:YAG
 - Holmium:YAG
 - Erbium:YAG
 - Carbon dioxide
15. Which near-infrared laser provides the deepest penetration in soft tissues?
- 808 nm diode
 - 980 nm diode
 - 1064 nm Nd:YAG
 - 1470 nm diode
16. The intensity of the laser beam in tissue is dependent on what parameter?
- Initial intensity
 - Total attenuation coefficient
 - Depth in tissue
 - All of the above
17. Beer's law of attenuation describes a _____ decay of light intensity in tissue.
- Linear
 - Doubling
 - Exponential
 - None of the above
18. A simple explanation as to why blood appears red is that _____.
- Blood reflects red light and absorbs blue and green light
 - Blood absorbs red light and reflects blue and green light
 - All the colors are reflected equally
 - None of the above

19. Which laser parameter is typically most important in laser medicine?
 - a. Temporal beam profile
 - b. Wavelength
 - c. Spatial beam profile
 - d. Duty cycle
20. Which has a higher peak power, a light bulb or pulsed laser?
 - a. Light bulb
 - b. Pulsed laser
 - c. They are the same
 - d. It depends
21. Which has a higher average power, a light bulb or a laser pointer?
 - a. Light bulb
 - b. Laser pointer
 - c. They are the same
 - d. It depends
22. Which laser parameters need to be known to calculate peak power?
 - a. Spot size and energy
 - b. Pulse rate and spot size
 - c. Energy and pulse duration
 - d. Spot size and pulse duration
23. Which laser parameters need to be known to calculate average power?
 - a. Energy and pulse rate
 - b. Energy and spot size
 - c. Pulse rate and pulse duration
 - d. Spot size and pulse rate
24. What is the pulse duration of a CW laser?
 - a. Milliseconds
 - b. Microseconds
 - c. Nanoseconds
 - d. Always on
25. In highly scattering tissue, use of a larger laser spot size results in _____.
 - a. More shallow penetration of the light
 - b. Deeper penetration of the light
 - c. Does not change the penetration depth of the light
 - d. Stronger absorption of the light
26. What minimum temperature is needed to start thermal coagulation of a tumor?
 - a. 0 °C
 - b. 20 °C
 - c. 37 °C
 - d. 60 °C
27. For most laser medical applications, tissue is destroyed by _____.
 - a. Light converted into heat and a temperature rise in the tissue
 - b. Light converted into a chemical reaction in the tissue
 - c. Light converted into acoustic energy and a mechanical wave in the tissue
 - d. Light converted into a plasma in the tissue
28. Which of the following lasers does not operate in the visible spectrum?
 - a. Ruby

- b. KTP
 - c. Dye lasers
 - d. Excimer
29. Which of the following lasers is a gas laser?
- a. Carbon dioxide laser
 - b. Holmium:YAG laser
 - c. Diode laser
 - d. Fiber laser
30. Which of the following lasers would be the best compact, inexpensive replacement for the CW Nd:YAG laser but would also provide similar laser-tissue interactions?
- a. Excimer laser
 - b. 980 nm diode laser
 - c. Carbon dioxide laser
 - d. Erbium:YAG laser
31. Which laser may provide a CW alternative to the holmium:YAG laser?
- a. 1470 nm diode laser
 - b. 808 nm diode laser
 - c. Nd:YAG laser
 - d. Er:YAG laser
32. Which is an advantage of a diode-pumped laser over a flashlamp-pumped laser?
- a. More efficient
 - b. Smaller laser
 - c. Air-cooled and quiet system
 - d. All of the above
33. Why are some medical laser systems so large?
- a. The size of laser scales with power output needed due to low wall-plug efficiency
 - b. Electrical supply takes up space
 - c. Water-cooling supply takes up space
 - d. All of the above
34. What is the main advantage of a fiber laser over conventional solid-state lasers?
- a. High intensity or brightness due to single-beam profile
 - b. Long pulse output
 - c. Less expensive capital equipment
 - d. All of the above
35. Ultrashort pulse lasers are ideal for applications that require _____.
- a. Deep tissue heating and removal of large bulky tumors
 - b. Rapid removal of a large amount of tissue
 - c. Precise removal of small amounts of tissue without collateral thermal damage
 - d. Use of small optical fibers for endoscopic applications inside the body
36. Articulated arms can be used in situations that require _____.
- a. Scanning of the laser beam across large surface areas such as the face
 - b. Delivery of high power
 - c. Integration into a rigid endoscope for use inside the body
 - d. All of the above
37. Standard silica (glass) optical fibers can be used in which part of the spectrum?
- a. Deep ultraviolet at very short wavelengths
 - b. Visible to near-infrared

- c. Mid-infrared
 - d. Far-infrared
38. Which optical fibers are the most flexible, most biocompatible, and least expensive?
- a. Sapphire
 - b. Germanium oxide
 - c. Silica
 - d. Chalcogenide
39. Which material is best to use as windows, disposable tips, and tissue cooling plates due to its high melting temperature, optical transparency, and thermal conductivity?
- a. Sapphire
 - b. Germanium oxide
 - c. Silica
 - d. Chalcogenide
40. Which delivery system would have the least losses from reflections at its ends?
- a. Silica
 - b. Hollow waveguide
 - c. Sapphire
 - d. Germanium oxide
41. What is the typical transmission percentage of light for a 2-meter long silica optical fiber after subtracting losses due to end reflections and attenuation in the fiber?
- a. 60%
 - b. 70%
 - c. 80%
 - d. 90%
42. Which value for the numerical aperture would correspond to the highest diverging angle of the laser beam out of the optical fiber?
- a. 0.10
 - b. 0.22
 - c. 0.37
 - d. 0.49
43. What happens to light down an optical fiber if the light rays exceed the critical angle?
- a. They experience total internal reflection and continue to travel down the fiber
 - b. They leak into the fiber cladding
 - c. They travel in a straight line down the fiber
 - d. They bounce back out of the input end of the fiber
44. What is the necessary condition for total internal reflection in an optical fiber?
- a. Refractive index of cladding is greater than core
 - b. Refractive index of cladding is equal to core
 - c. Refractive index of cladding is less than core
 - d. Refractive index of cladding is a negative value
45. If a surgeon keeps the laser power output constant but moves the optical fiber tip closer to the tissue surface, what happens to the intensity of the incident beam?
- a. It is lower
 - b. It stays the same
 - c. It is higher
 - d. Not enough information

46. How can a continuous-wave laser beam be used to produce a similar effect on tissue as a long-pulse laser beam?
- Focus the beam down to a very small spot
 - Rapidly scan the beam across the tissue
 - Split the beam into multiple smaller beams
 - All of the above
47. What is the main difference between an endoscope and a laparoscope?
- They are similar but just refer to different parts of the body
 - Laparoscopes are always smaller in size than endoscopes
 - Laparoscopes are more flexible than endoscopes
 - Laparoscopes are inserted through a surgeon-made hole in the skin and endoscopes through a natural opening in the body
48. What are the primary components of a standard flexible endoscope?
- A flexible, incoherent fiber-optic bundle for lighting/illumination of tissue surface
 - A flexible, coherent fiber-optic bundle for detection of reflected light and imaging
 - A hollow, working channel for inserting instruments to irrigate or remove tissue
 - All of the above
49. What is the most common fiber-optic connector for high-power surgical lasers?
- SMA
 - FC
 - FC/APC
 - None of the above
50. Use of a digital endoscope eliminates the need for _____.
- A fiber-optic bundle for lighting and illumination
 - A fiber-optic bundle for detection
 - A lamp as a lighting source
 - All of the above
51. What is the main advantage of a disposable endoscope vs. a reusable endoscope?
- It is more durable
 - It is smaller
 - It eliminates potential for antibiotic-resistant infections
 - All of the above
52. How can a silica optical fiber be made more flexible for use in a flexible endoscope?
- Make the fiber diameter smaller
 - Use sapphire optical fiber instead
 - Use a hollow waveguide instead of a solid glass fiber
 - None of the above
53. If there is no space to bend an optical fiber inside the body (e.g. in a small hollow tubular tissue structure), how can the light output be manipulated instead?
- Use a tapered optical fiber
 - Use a hollow waveguide instead
 - Use a side-firing fiber-optic tip
 - Use a larger optical fiber
54. Which fiber-optic tip provides most uniform illumination of largest volume of tissue (for example, for thermal coagulation inside a large tumor)?
- Diffusing tip
 - Side-firing tip
 - Tapered tip
 - Standard tip with axial delivery

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Glossary

(Listed in Alphabetical Order)

Ablation: The removal of tissue through vaporization.

Absorption coefficient: A measure of how strongly tissue absorbs light.

Anisotropy factor: The average cosine of the angle providing the direction at which a photon is scattered.

Articulated arm: A series of hollow tubes, joints, and mirrors used to reflect light.

Attenuation: The loss of light through absorption and/or scattering.

Benign Prostatic Hyperplasia (BPH): A benign condition with an enlarged prostate gland.

Biocompatible: Not harmful to living biological tissues.

Buffer: The outermost layer of an optical fiber, also sometimes referred to as the jacket, which provides protection and mechanical support for glass fibers under ending conditions as well as insulation from the surrounding environment. The layer is commonly made of a polymer material.

Carbonization: The formation of a charred surface layer on the tissue with continued heating after the tissue has become completely dried out or desiccated.

Cataract: The opacification or clouding of the lens in the eye, which leads to a loss of vision.

Catheter: A flexible tube inserted through a narrow opening into a body cavity to perform a clinical procedure or remove fluid from an organ (for example, the bladder).

Chromophore: A part of a molecule that absorbs light at a specific wavelength and is responsible for its color.

Cladding: The outer layer of an optical fiber that reflects light during total internal reflection.

Coagulation: The removal of tissue through heating and necrosis or cell death where the connective tissue proteins undergo thermal denaturation.

Coherence: A property of light when the photons travel in phase.

Continuous-wave: The property in which light is emitted continuously without interruption.

Core: The innermost layer of an optical fiber in which the light travels.

Cornea: A transparent layer forming the front of the eye and also acts as the outermost lens for focusing light.

Critical angle: Light rays traveling at a greater angle than the critical angle will be totally internally reflected while light rays traveling at a lesser angle will be refracted through the interface.

Cystoscope: An endoscope used to access the lower urinary tract, including the urethra and bladder.

Distal: The far end or output end of a system (for example, of an optical fiber or endoscope).

Endovenous: Inside the vein.

Endoscope: A telescope used to see inside a natural opening in the body.

Extracorporeal shock wave lithotripsy (ESWL): ESWL uses focused ultrasonic waves emitted from outside the body to fragment kidney stones into smaller pieces for spontaneous removal from the urinary tract.

Fluoroscopy: A medical imaging technique that uses continuous x-rays, or an x-ray movie, to track motion within the body (for example, the insertion of medical instruments such as catheters with radiopaque markers).

Gaussian: A bell-shaped curve.

Hemoglobin: A red protein responsible for transporting oxygen in the blood.

Hygroscopic: Tending to absorb water or moisture.

Insufflation: The act of blowing CO₂ gas into body cavities to separate tissues and provide improved workspace, visibility, and safety during laparoscopic surgery.

Irradiance: The intensity of light or power density.

Laparoscope: A telescope inserted through a small incision in the skin to see inside the body.

Laser-assisted in situ keratomileusis (LASIK): LASIK uses an ultraviolet laser to vaporize corneal tissue for vision correction after the top flap of the cornea has been removed.

Laser interstitial thermal therapy (LITT): LITT typically involves the use of a near-infrared laser with deeply-penetrating laser energy delivered through an optical fiber and placed inside a tumor for thermal coagulation and destruction of the tumor.

Lumen: The hollow inside of an object (for example, a blood vessel or working channel of endoscope).

Mean-free path: The distance that a photon is likely to travel before being either absorbed or scattered.

Melanin: A dark brown or black pigment found in hair, skin, and the iris of the eye. It is responsible for the tanning of skin exposed to sunlight.

Mid-infrared spectrum: Part of the electromagnetic spectrum referring to light with wavelengths longer than about 3000 nm, as defined by the medical field.

Mode-locked: Delivery of light in very short pulses on the order of picoseconds to femtoseconds.

Monochromatic: Light that has a single, narrow wavelength.

Multimode: For an optical fiber, multiple or many modes of light propagation—or different paths—are supported by the fiber core.

Near-infrared spectrum: As defined by the medical field, part of the electromagnetic spectrum referring to light with wavelengths greater than 700 nm but less than approximately 3000 nm.

Necrosis: Cell death

Numerical aperture: Sine of the half angle for light either entering or leaving an optical fiber.

Optical penetration depth: The depth at which light penetrates a tissue before being attenuated or lost (due to absorption and scattering), where the irradiance decays to $1/e$ or 37% of its initial value at the tissue surface.

Percutaneous: The medical device is delivered through the skin (e.g. a needle biopsy), typically in a minimally invasive manner.

Percutaneous nephrolithotomy (PCNL): PCNL uses a small incision made in the back or abdomen to remove kidney stones that are too large to be removed using endoscopic methods or shockwave lithotripsy.

Photodynamic therapy (PDT): PDT is a minimally invasive, light-based chemotherapy treatment involving delivery of red light to a photosensitive drug which has selectively accumulated at a tumor site. The presence of light, a drug, and oxygen produces a chemical reaction and singlet oxygen which is toxic to cancer cells.

Photorefractive Keratectomy (PRK): PRK uses an ultraviolet laser to vaporize corneal tissue for vision correction; unlike LASIK, the top part of the cornea is not removed during the procedure.

Polychromatic: Light that consists of a broad band of multiple wavelengths.

Port-wine stains: A birthmark involving the development of abnormal small blood vessels with a dark red color similar to the color of port wine.

Potassium titanyl phosphate: A KTP crystal used with the neodymium:YAG laser to double the frequency and half the wavelength from 1064 nm in the infrared spectrum to 532 nm in the visible (green) spectrum.

Protein: A large molecule that is an essential structural component of tissues.

Proximal: The near end or input end of a system (for example, an optical fiber or endoscope).

Q-switched: Delivery of light in short pulses on the order of nanoseconds.

Reduced scattering coefficient: These parameters take into account the directional scattering or anisotropy factor, g , in providing a more accurate representation of the effects of light scattering.

Refractive index: The ratio of the speed of light in a vacuum divided by the speed of light in a medium.

Retina: A layer at the back of the eye containing cells that are sensitive to light and that trigger nerve impulses that pass through the optic nerve to the brain where a visual image is formed.

Scattering coefficient: A measure of how strongly tissue scatters light.

Single mode: For an optical fiber, only one mode of light propagation, or path, is supported by the fiber core.

Spatial beam profile: The shape or structure of the laser beam that provides information on how the energy is distributed in space across the laser spot.

Temporal beam profile: The shape or structure of the laser pulse that provides information on how the energy is distributed during the duration of a single laser pulse.

Total attenuation coefficient: This parameter takes into account both the absorption and scattering coefficients.

Total internal reflection: The process of light traveling down the core of an optical fiber and being reflected multiple times at the core-cladding interface.

Trocar: An access port to internal organs during laparoscopic surgery. It may consist of a sharp cutting point for piercing the skin, a hollow tube for introducing laparoscopic instruments, and an air- and fluid-tight seal for preventing CO₂ gas and bodily fluids from escaping. Insufflation, the delivery of CO₂ gas into a body cavity, is common during laparoscopic surgery to separate tissues for improved workspace, visibility, and safety.

Ultraviolet spectrum: Part of the electromagnetic spectrum referring to light with wavelengths shorter than 400 nm.

Umbilicus: The hollow opening of a tube.

Ureteroscope: a rigid or flexible endoscope used to access the upper urinary tract, including the ureter and kidneys.

Varicose veins: Swollen, twisted veins caused by weakened or damaged vein walls and located underneath the skin, usually in the legs.

Visible spectrum: Part of the electromagnetic spectrum referring to light with wavelengths in the range of 400-700 nm.

Credits for Graphics

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Figure 5. Adapted from Peng, Quian, Asta Juzeniene, Jiyao Chen, Lars O. Svaasand, Trond Warloe, Karl Erik Giercksky, and Johan Moan. “Lasers in medicine.” *Reports on Progress in Physics* 71, no. 5 (2008).

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Figure 16. Courtesy of RP Photonics Consulting GmbH.

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Figure 27. Acemoglu, Alperon, Nikhil Deshpande, and Leonardo S. Mattos. "Towards a Magnetically-Actuated Laser Scanner for Endoscopic Microsurgeries." *Journal of Medical Robotics Research* 3, no. 1 (November, 2017).

Figure 28. Agoston, Gregg, and Karl Storz. "Redefining endoscope quality in HD world." *Endoscope Maintenance Guide*, 2009.

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- Spectroscopy and Pollution Monitoring
- Spectroscopy and Remote Sensing

Photonics Enabled Technologies Modules: Optoelectronic

- Photonics in Nanotechnology
- Photonics in Nanotechnology Measurement: A study of Atomic Force Microscopy
- Photonics Principles in Photovoltaic Cell Technology

Precision Optics Modules

- Quality Assurance of Precision Optics
- Metrology of Optical Systems

Semiconductor Laser Diodes

Enhanced Spectroscopy

Light Emitting Diodes

Thermoelectric Device Measurement

Solid-State Laser Crystal Manufacturing

Field Service Engineering

High Energy Pulsed Solid-State Laser Design and Testing

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