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Administrative Office:

55 Corporate Drive, 3rd Floor
Trumbull, Connecticut 06611
U.S.A.
Telephone: 203-332-2507
Email: lasers1060@aol.com
www.americanboardoflasersurgery.org

STUDY GUIDE FOR THE BASIC LASER SCIENCE, TISSUE INTERACTION AND LASER SAFETY WRITTEN EXAMINATIONS

2012 Edition

This Study Guide Is Provided to All Candidates for Certification Who Have Qualified to Take the Written Basic Laser Science and Safety Examinations

The American Board of Laser Surgery, Inc.

Major Topics Covered in the Study Guide for
The Basic Laser Science and Laser Safety Written Examinations

Sample Topics and Pages Highlighted in Blue Bold Italics

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INTRODUCTION

Purpose and Importance of the Study Guide and Its Relevance for Clinical Application of Laser and Light-based Therapies

Significance of Becoming a Diplomate

Those who have passed the Board's rigorous examinations and become Diplomates have found that the Certificate of the ABLS is a valuable credential for them in attracting patients as well as demonstrating a greater depth of understanding of lasers and light-based technologies.

In the first decade of the Board's existence, most Diplomates came from the United States and Canada. In recent years, however, many have come from other countries. The Board's Certificate has been awarded to over 500 Diplomates worldwide, including Canada, Europe, South America, Egypt, Saudi Arabia, Iraq, Japan, Korea, Thailand, Malaysia and Australia among others, as well as the United States. Our Diplomates have found that studying for and taking the examinations is a valuable learning experience. In fact, the Cosmetic Physicians Society of Australasia adopted the examinations of the ABLS as their own in the year 2000.

Diplomates of the Board are also often members of other medical societies and organizations such as the ASLMS, AACS, ASPS, ASDS, and others. Only two major medical societies or institutes that we know of currently offer a laser course dealing with fundamentals related to medicine. Furthermore, no other organization worldwide offers board credentialing in laser science, tissue interaction and safety in medicine and surgery at this time. The ABLS is the sole medical specialty Board that offers the rigorous study, and written and oral examinations necessary for full Board certification. The ABLS has led the industry in doing so for over 25 years.

Role of the Study Guide and Related Materials in the Certification

The Study Guide is proprietary and has been developed and updated by members of the Board for the purpose of specifically advancing knowledge of laser science, biophysics and safety in medicine. The Study Guide and the written examinations are truly unique. The minimum passing score for both written examinations is 80%. About 15% of candidates do not pass every portion of the written examination on the first attempt, even with an open-book policy, which attests to the exam's rigor. While 85% pass, less than half earn a total score of 90% or higher, indicating that there is an opportunity for nearly every candidate to expand his or her knowledge.

The Study Guide has a heavy emphasis on laser physics and bio-tissue interactions, and to a more in-depth degree than any other medical publication available that we know of. That said, our candidates must bear in mind the importance of understanding how lasers and other light-based devices work to maximize clinical success and patient safety. If this certification improves the outcome and safety of even a handful of patients, it is well worth it! We believe that it will have a far greater impact on our Diplomate's careers.

Sincerely,

Edward M. Zimmerman, M.D.
Warren B. Seiler III, M.D.
John C. Fisher, Sc.D.

excited dimers can occur at low pumping power, to be suddenly relaxed in a giant pulse of radiant power. The efficiency of the pumping-lasing process is low, often below 1%. Experiments have shown that a XeCl laser can deliver 180 millijoules of radiant energy in a 30-nanosecond pulse with electric-discharge excitation of 150,000 amperes at 48,000 volts.[5] The efficiency of this process is about 0.08%. Thus, if a continuous-wave XeCl laser delivering 10 W were desired, the required electrical input power to the laser tube would be 12,500 watts. Though this is physically possible, it is practically very difficult.[5] For reasons of size, cooling, etc, excimers are confined to pulsed operation.

The Holmium: YAG Laser

The rare-earth element holmium has been used as a dopant in crystals of YAG (yttrium-aluminum-garnet, or $Y_3Al_5O_{12}$) in conjunction with erbium and thulium, which enhance the efficiency of optical pumping of holmium. This material, known as Ho:YAG, emits laser radiation at about 2100 nm. Though Ho:YAG is technically a 4-level material, the lower laser level is so close to ground that the threshold energy per unity volume of material is very high. Consequently, at room temperature, continuous-wave operation is not possible.

The Erbium:YAG Laser

This material emits laser radiation at 2940 nm. It is a 4-level energy system, but the lower laser level has a long lifetime, causing the erbium ions to accumulate in this lower level after emitting radiation. This accumulation interrupts the population inversion and limits the laser to pulsed operation.

The Erbium:YSGG Laser

The erbium:yttrium-scandium-gallium-garnet laser emits radiation at 2790 nm, making it primarily a skin resurfacing laser where water is the chromophore or target. The sealed laser tube that produces this wavelength is in the hand piece of the machine, so no waveguide or articulated arm is employed. It is similar in use and function to the older Er:YAG, but the Er:YSGG laser has a tissue absorption coefficient roughly 5 times that of the CO^2 laser and less than a third that of Er:YAG. Its' ablation threshold is about 3 J/cm², compared to 0.5 J/cm² for Er:YAG (Note that ablation threshold varies somewhat with volumetric power density and pulse width). Therefore, in general, this wavelength causes more ablation of tissue and thermal effect to the surrounding tissue than Er:YAG, and less ablation and surrounding thermal effect than CO^2 lasers.

Important Lasers Used in Medicine and Surgery

Table 1-2 at the conclusion of this chapter shows the important lasers used in medicine and surgery at this writing, with their important operating characteristics and current or future applications. The reader should appreciate that new wavelengths are being explored even as this is written, and that Table 1-2 is not carved in stone. Despite that fact, several lasers have dominated surgical applications (other than ophthalmic) for the past 25 plus years: the carbon dioxide and the Neodymium:YAG. During the past 15 plus years, Nd:YAG has found increasing applications, both at its normal wavelength of 1064 nm and at its frequency-doubled wavelength of 532 nm in the so-called KTP laser (KTP is the acronym for potassium-titanyl-phosphate, a non-linear optical material developed in this country by DuPont.). The KTP laser has largely replaced the argon-ion laser for treating discrete, superficial vascular and pigmented lesions, because of KTP's higher efficiency, greater reliability, and the ability to change from 1064 nm to 532 nm at the flip of a switch. By frequency-tripling, the Nd:YAG laser can deliver a wavelength of 355 nm, in the ultra-violet range now dominated by the xenon-fluoride excimer laser. Shortening the pulse duration to nanoseconds (Q-switching) has allowed the 1064nm and 532nm wavelengths to be used to decrease the appearance of certain colors of tattoos, melasma, congenital nevi and fine vascular netting. Other Nd-YAG wavelengths including 1320nm and 1440nm are being used for non-ablative skin resurfacing and ablative laser lipolysis and sub-dermal skin tightening and collagen induction.

TABLE 1-2a
Important Lasers Used in Medicine and Surgery*

Type of Laser	Major Wavelength (nm)	Temporal Mode	Max Avg (a) or Peak (p) Power (watts)	Delivery of the Beam	Medical and Surgical Applications
ARGON-ION Gas	488 514	Continuous-wave Long pulsed Medium pulsed	1 to 20 (a)	Quartz optical fiber Handpiece Slit lamp and microscope Sapphire tips Micromanipulator	Photocoagulation in the eye Photocoagulation of vascular skin lesions Photocoagulation of ectopic endometriomas Ablation of pigmented skin lesions Laser angioplasty Stapedotomy in ear Pumping dye lasers
CARBON-DIOXIDE Gas CO2, N2, and He Sealed-tube or flowing	10,600	Continuous-wave Long pulsed Medium pulsed	10 to 100 (a)	Direct Articulated arm Handpiece Micromanipulator Hollow reflective waveguide	Precise general surgery Cutting, drilling, and vaporization of tissue everywhere in the body Debridement of necrotic or infected lesions Sutureless anastomosis Skin rejuvenation (fractional or non fractionnal) Warts. Syringoma
DYE Liquid Fluorescent organic dyes (e.g., rhodamines in liquid solutions [e.g., dimethyl sulfoxide])	300 to 1000 Continuously tunable	Continuous-wave Long pulsed Medium pulsed Short pulsed	1 to 15 (a) Up to 50 x 10 to 6th (p)	Direct Quartz optical fiber Handpiece Sapphire tips Interstitial probes	Color-selective treatment of pigmented and vascular skin lesions Excitation of photosensitizers such as dihematoporphyrin ether in cancer therapy Lithotripsy of stones in the ureters Laser angioplasty Acne and scars Some skin rejuvenation
ERBIUM:YAG Solid Erbium ions in crystals of yttrium-aluminum-garnet	2940	Medium or long pulsed	Up to 500,000 (p) or 3 (a)	Direct or articulated arm	Precise vaporization of tissue (skin resurfacing) Still experimental: Cutting of bone Laser angioplasty Incision or excision with very little lateral heating

TABLE 1-2b
Important Lasers Used in Medicine and Surgery*

Type of Laser	Major Wavelength (nm)	Temporal Mode	Max Avg (a) or Peak (p) Power (watts)	Delivery of the Beam	Medical and Surgical Applications
ERBIUM:YSGG Solid Erbium ions in crystals of yttrium-scandium-gallium-garnet	2790	Medium pulsed	Up to 5	Direct or articulated arm	Precise vaporization of tissue (skin resurfacing) Dental debridement
ERBIUM:Glass	1540-50	Medium pulsed	1-200mJ/point	Direct	Non-ablative, fractional skin rejuvenation
EXCIMER Gas: Bi-atomic molecules existing only in the excited state:		Short pulsed	(a) up to:		Sculpturing the cornea of the eye Radial keratotomy Laser angioplasty Cutting of bone Ultra-precise incision or excision Psoriasis
ArF	193		50	Direct	Some skin pigmentation disorders (Vitiligo)
KrCl	222		50	Direct	
KrF	248		100	Quartz fiber	
XeCl	308		150	Quartz fiber	
XeF	351		30	Quartz	
HELIUM-NEON Gas	633	Continuous-wave Long pulsed Medium pulsed	0.001 to 0.025 (a)	Quartz optical fiber Handpiece Articulated arm Micromanipulator	Visible designation for aiming infrared lasers (He-Ne and CO2) Biostimulation: Healing of wounds; Relief of pain
GOLD VAPOR Gas	628	Medium pulsed	Up to 12 (a) or 100,000 (p)	Quartz optical fiber	Excitation of photosensitizer dihematoporphyrin ether in photodynamic therapy for cancer
THULIUM (Tm):YAG Solid	2000	Pulsed	Up to 25 (a)	Quartz optical fiber	Arthroscopic surgery in knee and shoulder Urologic surgery for prostate
HOLMIUM (Ho):YAG	2100	Pulsed 400-500 μsec	Up to 25 (a) 3500 - 10,000 (p)	Quartz optical fiber	Endoscopic sinus surgery Dacryocystorhinostomy Decompression of herniated spinal discs Angioplasty Urologic surgery for prostate, and kidney Stones Other endoscopic applications

TABLE 1-2c
Important Lasers Used in Medicine and Surgery*

Type of Laser	Major Wavelength (nm)	Temporal Mode	Max Avg (a) or Peak (p) Power (watts)	Delivery of the Beam	Medical and Surgical Applications
KTP Solid Neodymium:YAG, frequency-doubled by a crystal of potatssium titanyl phosphate	532	Continuous-wave Long pulsed Medium pulsed	Up to 10 (a)	Same as for Nd:YAG	Vascular indications (telangectasia, couperose, leg red vessels, angioma) Skin pigmentation lesions (lentigo, hyperpigmented scar , burns) Acne Skin rejuvenation Tatoo removal (Qswitched temporal mode) Warts indications (combined to cryotherapy)
KRYPTON-ION Gas	476 521 568 647	Continuous-wave Long pulsed Medium pulsed	Up to 6 (a)	Quartz optical fiber Handpiece Slit lamp and microscope Sapphire tips Micromanipulator	Photocoagulation in the eye Photocoagulation of pigmented skin lesions Pumping of dye lasers
NEODYMIUM:YAG Solid	1064 1320	Continuous-wave Long pulsed	50 to 100 (a)	Quartz optical fiber Handpiece	Voluminous heating of tissues for necrosis or hemostasis Angiology/phlebology indication Long terme hair removal Skin rejuvenation Tatoo removal (Qswitched temporal mode) Incisive and coagulative surgery of soft tissue throughout the body and sub cutaneous fat distruction Angiology/phlebology indication Long terme hair removal Acne and acne scars indications Skin rejuvenation
Neodymium ions in in crystals of yttrium-aluminum-garnet	946	Medium pulsed Short pulsed	1 x 10 to 8th and up (p)	Sapphire tips Articulated arm Micromanipulator	Ultra-precise focal destruction of intra-ocular targets Sutureless anastomosis of skin, vas deferens, etc.

TABLE 1-2d

Important Lasers Used in Medicine and Surgery*

Type of Laser	Major Wavelength (nm)	Temporal Mode	Max Avg (a) or Peak (p) Power (watts)	Delivery of the Beam	Medical and Surgical Applications
RUBY Solid Chromium ions in in a matrix of aluminum oxide	694	Medium pulsed Short pulsed	10 to 8th (p)	Direct Articulated arm	Photocoagulation of pigmented lesions Hair removal (deprecated)
SEMICONDUCTOR Solid Junctions between semiconducting substances containing: aluminum, antimony, arsenic, gallium, indium, phosphorus, and Notably: GaAs, and AlGaAs	590 to 5300 830-920 620-900	Continuous-wave Long pulsed Medium pulsed	0.0005 to 0.5 (a)	Direct Quartz optical fiber	Biostimulation: Acne Healing of wounds Relief of pain Regeneration of neural tissue Neovascularization Ophthalmology
DIODE	800-810	Medium Pulsed	up to 80 (p)	Direct	Long Term Hair Modification/Reduction, some skin rejuvenation
ALEXANDRITE	755	Long pulsed Short pulsed	18,000 (p)	Quartz optical fiber	Long-term hair modification/reduction, some skin rejuvenation Pigmented lesions (lentigo), tatoos - some phlebology indications

*Technical parameters and applications of the important lasers now used in medicine and surgery, both under Pre-Market Approval (P.M.A.) and Investigational Device Exemption (I.D.E.).

controlled by the total power transmitted, which can be up to 20 W with appropriate design.

Terminating Devices for Laser-Beam Delivery Systems

In general, the terminating device for a laser-beam delivery system is used to steer, focus, or otherwise condition to the beam so that it will have the desired surgical effect upon the tissue. Terminating devices are of 6 basic types: (1) the unadorned distal end of the delivery device (optical fiber or hollow light-pipe); (2) a focusing handpiece for an articulated arm; (3) a short, hand-held light-pipe probe attached by an optical coupler to the end of an articulated arm; (4) a micromanipulator for steering and focusing the beam from an articulated arm; (5) an attachable / detachable tissue-contacting probe at the end of an optical fiber; and (6) the specially contoured distal end of an optical fiber to be placed in contact with the target tissue. These are discussed individually in the following sections.

The Distal End of the Delivery Device

A quartz optical fiber is often used to deliver a laser beam directly to the surgical target, with no externally added device at the distal end. In such a case, it is necessary that the distal-end face of the fiber be perpendicular to the axis and polished to a plane surface, so that the emerging beam will be conically diverging and that the transverse distribution of power density will be symmetric in any plane passing through the axis of the fiber, without any local distortions. It is easy to judge whether this is so, at least to a first approximation, by holding the distal end of the fiber perpendicular to a flat, white surface at a distance of about 1 cm, and noting the pattern of reflected light from the aiming beam of the laser, while the surgical beam is turned off. If the distal end of the fiber is properly cleaved and polished, the reflected light will vary in intensity only with radial distance from the center of the spot, having maximum intensity near the center and diminishing gradually toward the outer circumference with no irregular or asymmetric splotches of intensity.

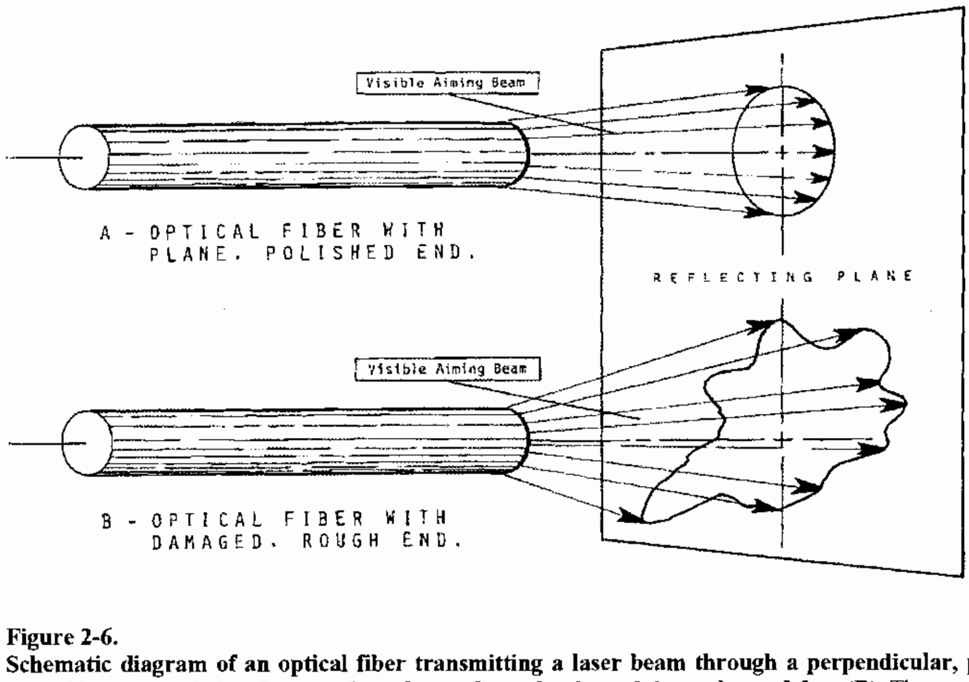

Figure 2-6.
Schematic diagram of an optical fiber transmitting a laser beam through a perpendicular, plane end-face (A), and the same fiber transmitting the same laser beam through a jagged, irregular end-face (B). The rough end-face causes the pattern of reflected light from a smooth, white target to have distorted geometry and irregular intensity.

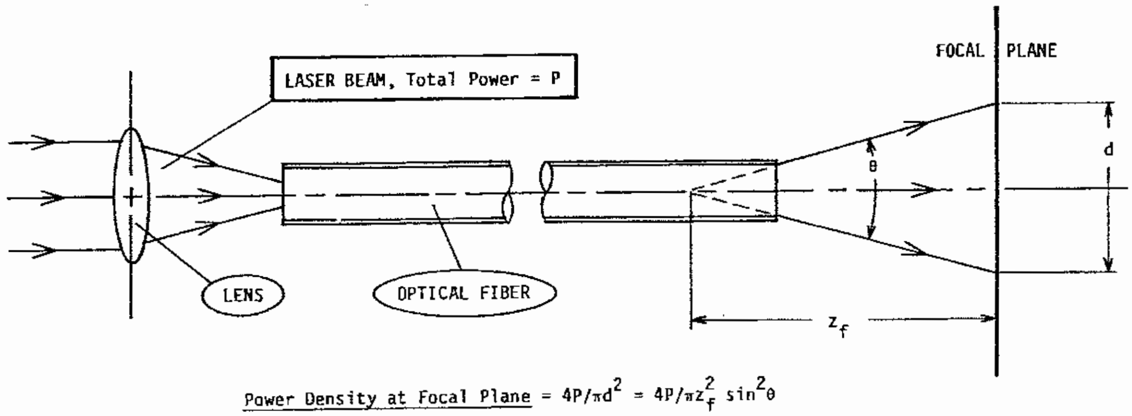

Figure 3–2.
Schematic diagram of an optical fiber with a diverging laser beam emerging from its distal end. The beam entering the proximal end of the fiber is focused to a spot smaller than the core diameter, so that virtually all of the radiant power of the beam is transmitted into the fiber.

used to bore a hole through it. However, because the beams of these lasers are almost always transmitted through slender quartz fibers, there is an easy method for estimating the effective diameter of the beam where it emerges from the distal end of the fiber. The laser beam emerging from the laser itself is always of a diameter larger than that of the fiber core, and so a positive lens is used to focus the beam on the proximal end-face to a spot which is smaller than the core. This means that nearly all of the power of the beam is transmitted through the fiber, except for a small amount lost because of first-surface reflection. For simplicity, we can assume that 100% of the total power of the beam enters the fiber, neglecting the 4% reflectance of dielectrics for near-normal incidence of the rays.

The repeated total internal reflections of the rays from the interface of core and cladding as they travel to the distal end cause transposition of the rays, loss of spatial coherence, and a flattening of the power-density profile of the distally emerging beam. Because it is no longer gaussian, the beam does not have an "effective" diameter, in the strict sense of the term, and the easiest way to calculate the average power density is to divide the total beam power by the cross-sectional area of the core of the fiber:

$$p_a = P_o/(\pi d_c^2/4) = 4P_o/(\pi d_c^2), \quad (3\text{–}3)$$

where p_a is the average power density within the beam right at the distal end-face, P_o is the total power of the beam, and d_c is the core diameter of the fiber. If the diameter is expressed in millimeters, the power in watts, and the power density in watts per square centimeter, Equation 3–3 becomes

$$p_a = 127(P_o \text{ in watts})/(d_c \text{ in mm})^2. \quad (3\text{–}4)$$

Figure 3–2 shows schematically an optical fiber with a laser beam emerging from its distal end. As noted in Chapter 2, the beam will diverge with an included angle between 5° and 15°. This divergence allows the operator to vary the power density by moving the distal end of the fiber toward or away from the target. A typical operating distance for a non-contacting laser fiber will be one centimeter from distal end to tissue. The power density at the tissue can be calculated approximately by using Equation 3–4 with the diameter set equal to that of the helium-neon aiming beam on the tissue surface. Although the He-Ne beam does not always have the same divergence as the beam of the surgical laser, the approximation is good enough for estimating the power density at the target.

Surgical Importance of Power Density

Mechanisms of Tissue Destruction by Laser Beams

Photochemolysis

Lasers whose wavelengths are in the ultraviolet region of the spectrum, where the photonic energy, e_p, is high enough to break the electronic bonds between atoms in molecules, can destroy tissue by *photochemical breakdown* of complex organic molecules. The excimer lasers are in this category. The destructive process is called *photochemolysis.*

Photothermolysis

For all surgical lasers except the excimers and ultra-short-pulsed Nd:YAG lasers, the major means of destroying tissue is the conversion of laser light to heat and the consequent rise in temperature of the target tissue. This subject will be discussed in greater detail in Chapter 4, but here we shall consider it briefly. If the objective is *coagulation,* the tissue can be heated to temperatures above 60° C but below 100° C (the boiling point of water at atmospheric pressure). The destructive process is then called *photopyrolysis.* If the objective is *cutting or ablation* of the tissue, then the laser beam must be intense enough to cause the histologic temperature to rise quickly to at least 100° C, so that the water in the tissue will be flashed into steam, which ruptures the cells and destroys the histologic architecture. This process is called *photovaporolysis.* Photopyrolysis and photovaporolysis are component processes of *photothermolysis,* the destruction of tissue by heating caused by absorption of radiant energy.

Vaporolysis is the primary mechanism by which most surgical lasers cut or ablate tissue. It is especially effective at any wavelength for which the absorption coefficient in water is 100/cm or more. The surgical wavelengths for which aqueous absorption exceeds 100/cm are all those above 2600 nanometers. In particular, the carbon-dioxide and the erbium:YAG lasers are in this category. The Er:YAG has not been routinely used in surgery, at this writing, largely because it lacks a surgically suitable fiber for transmission of its beam. However, it has great potential as a laser for precise cutting and vaporizing of tissue anywhere in the body. Its beam can be delivered via an articulated arm like those customarily used with the CO_2 laser.

Photoplasmolysis

Lasers like the Nd:YAG, whose wavelength is weakly absorbed in lightly pigmented tissue like the cornea, lens, or lens capsule of the eye, can destroy such tissues by a phenomenon called *optical breakdown,* or, more properly, *photoplasmolysis,* in which the electric field of the light waves is strong enough to *ionize* the atoms in the tissue and form a *plasma,* a hot gas (at temperatures up to 15,000° C) consisting of equal concentrations of free electrons and positive ions. This occurs at power densities above 10,000,000,000 W/cm^2. This plasma is a strong absorber of radiation at all wavelengths, and, once formed, absorbs all of the subsequently incoming radiation and expands rapidly, causing shock waves and mechanical destruction of histologic architecture. Because only ultra-short-pulsed lasers are capable of generating such high power densities, the extent of destruction is controlled by adjusting the energy per pulse, which is typically only a few millijoules.

Thresholds for Photochemolysis

Photochemical breakdown of living tissue is a process of rupturing electronic bonds between atoms in organic molecules by means of energetic short-wave radiation. In the ultraviolet part of the spectrum, this occurs at average power densities (averaged over time and area) below 1 W/cm^2, although it will continue to occur at higher power densities, in conjunction with other effects that occur as power density increases. At some lower limit of power density in tissue, even at energetic wavelengths, there will be no *net* breakdown of organic molecules, because the ruptured inter-atomic bonds will re-form as fast as they are broken. This threshold is dependent on the wavelength of the light and the type of molecule absorbing it. It can be just a small fraction of a watt per square centimeter, as in the case of sunlight falling on dye molecules in fabrics exposed to the sun, whose power density at the earth's surface is seldom more than 100 milliwatts/cm^2. The fading of bright colors in sunlight is a familiar datum of observation to people living in sunny climates. Humans whose skin is exposed to intense sunlight suffer breakdown of the dermal collagen, elastin, and reticulin that give healthy skin its smooth texture.

Thresholds of Destruction for Photothermolysis

For any wavelength that is absorbed in water, there is a *threshold* of power density below which the water in the target cannot be boiled by a laser beam. The threshold value is lowest for wavelengths that are strongly absorbed, and highest for those that are poorly absorbed. The threshold exists because the water in which value is that for which the electric-field strength of the light wave is equal to the strength of the electric field which binds an outer-shell electron of an atom to its nucleus. The radiant power density corresponding to this field strength is about 1.2×10^{16} W/cm^2 for hydrogen, a very abundant constituent of living tissue.

Surgical Significance of Destructive Thresholds of Power Density

As will be seen in greater detail in Chapter 4, there is a range of power density of a laser beam striking living tissue within which certain physical effects take place in that tissue. If the surgeon wishes to have one effect predominate over all the others, the power density in the beam of the surgical laser must exceed the threshold at which that effect begins, but not by so much that other effects occurring at higher power densities set in because the next threshold has been exceeded. These thresholds, as already explained, are wavelength-dependent (Fig. 3–3).

As was stated earlier in this chapter, the important mechanisms by which most surgical lasers destroy living tissue are photopyrolysis and photovaporolysis, both of which are included under the more general category of *photothermolysis*. If a surgeon wishes to coagulate tissue without vaporizing it, clearly the power density of the incident beam, whatever the wavelength, must not exceed the threshold for vaporization at that wavelength. For example, in using the Nd:YAG laser to coagulate the endometrial lining of the uterus in a patient having intractable menorrhagia, the surgeon must avoid using a power density high enough to cause instantaneous vaporization of that tissue.

If the surgeon wishes to use a laser like the carbon-dioxide for vaporizing a part of, say, the uterosacral ligament to destroy the nerve transmitting the pain of dysmenorrhea, but not to damage adjacent tissue by heating, then the applied power density must be at least equal to the threshold of vaporization, and preferably well above it, so as to remove the target tissue quickly and minimize the time of exposure of nearby tissue to the heat conducted from the boiling histologic water in the target zone. The great virtue of the CO_2 laser for precise, atraumatic surgery is its ability to vaporize the water of soft tissue, usually at *fixed* temperatures near or slightly above 100° C. *This is so because water boils at a constant temperature, so long as the pressure on it is constant.* At atmospheric pressure, this constant temperature is 100° C. Because the rapid boiling of histologic water where the laser beam impacts the tissue occurs at a fixed temperature, *the total amount of heat energy transferred to adjacent tissue is dependent only upon the total time of exposure,* and not upon the power density of the laser beam, as long as that is well above 100 W/cm^2.

Thus, vaporolysis by flash-boiling of histologic water to form steam has an inherent, automatic thermal safety mechanism. As long as the steam can escape readily from the boiling surface, the pressure on the tissue will be at or near atmospheric, and the tissue adjacent to the vaporized zone is damaged only by thermal conduction of heat from this isothermal region at about 100° C. Because thermal conduction in soft living tissue is largely through non-convecting water within and between cells, it is a relatively slow means of heat transfer. Experiments performed with thermocouples implanted in a dog's tongue while it was impacted at one spot by a CO_2 laser have showed that the fall of temperature away from the laser crater is exponential with distance normal to the crater wall, and that the maximum thermal power density that can be removed by aqueous conduction to adjacent tissue is only about 10 W/cm^2.[1] Therefore, if the applied power density in the CO_2-laser beam is far above that value, the ablated tissue can be quickly removed before the flow of heat to adjacent tissue can cause significant thermal necrosis.

For precise lasers like the CO_2, whose wavelength is strongly absorbed in water, we can define *a temperature-transfer time,* which is a measure of how much time is required for the temperature of adjacent tissue a specified distance from the laser-impact site to rise a given amount above normal tissue temperature. This time depends upon the specific heat, the thermal conductivity, the mass density of the tissue and the allowable temperature-rise, as well as the distance from the laser-impact site to the healthy tissue. For adjacent soft tissue distant 0.2 mm from the impact site of a CO_2-laser beam, the time required for a temperature increase of 5° C is about 90 milliseconds when the power density of a 1.8-mm beam is about 700 W/cm^2.[1] If this beam were allowed to ir-

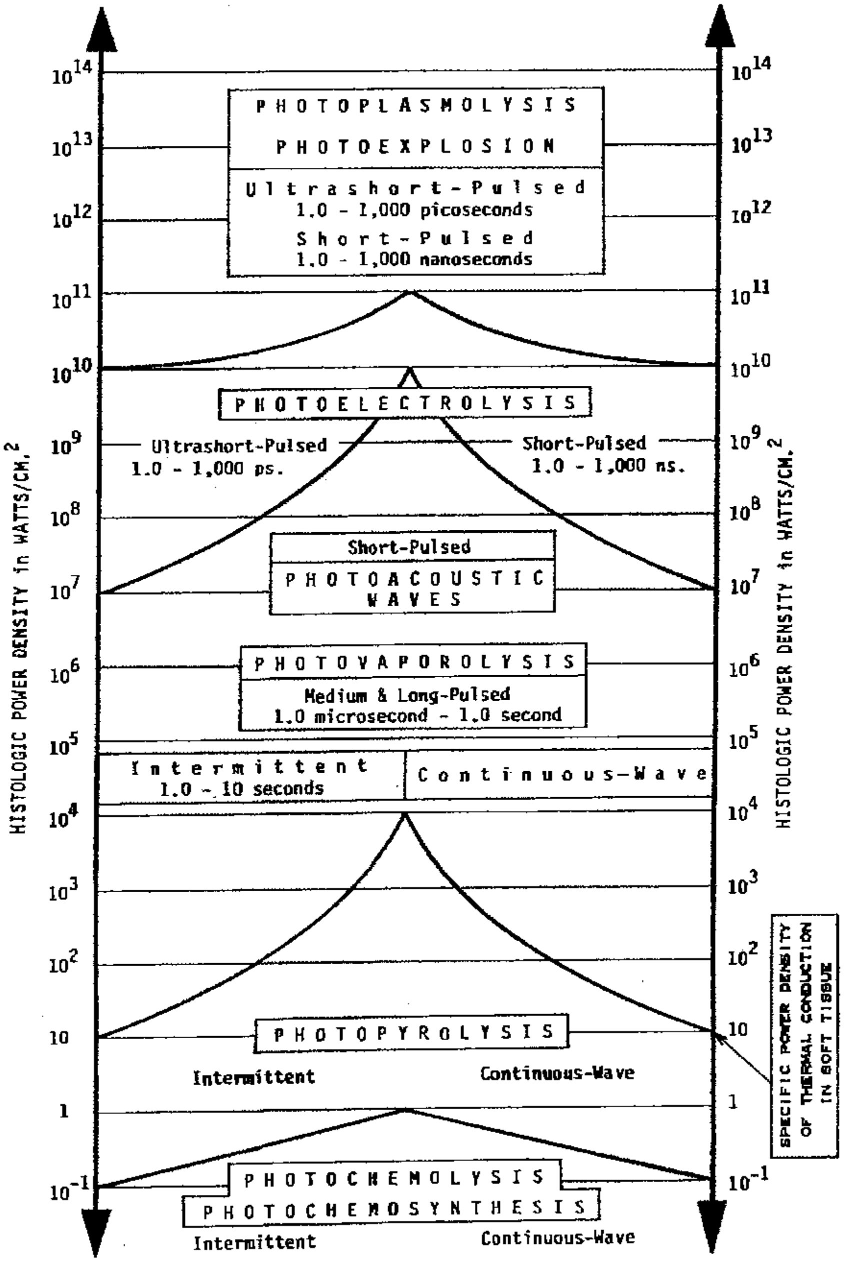

Figure 3–3.
Biologic effects of laser radiation between 100 nm and 10,600 nm as functions of average (over time and space) power density in soft tissue. The sloping boundary lines between regions of different effects denote the fact that absorption coefficient and < > or photon energy vary with wavelength. The peak of each of the three lower boundary lines corresponds to the wavelength having lowest absorption coefficient in a particular tissue and the shoulders of those lines to the wavelength having the highest absorption coefficient. The uppermost boundary is nearly wavelength-independent, and depicts the variation of threshold power density for optical breakdown as a function of pulse duration and focal geometry of the laser beam. Reprinted from Fisher JC. Basic laser physics and interaction of laser light with soft tissue. In: Shapshay SM, ed. Endoscopic laser surgery handbook. New York: Marcel Dekker, 1987: Fig. 29, p. 109.

radiate the tissue only for 1 millisecond, the temperature rise at 0.2 mm would be negligible.

By suitably reducing the time during which tissue is exposed to the beam of a CO_2 laser, the adjacent zone of thermal damage can be made arbitrarily small. This can be done in one of two ways: (1) by raising the power density to such a high level that the target tissue is destroyed in an arbitrarily short time, or (2) by using a high, but controllable, power density applied to the

Technical Parameters of Superpulsing

Pulse Duration

In order to be minimally traumatic to adjacent tissue, a superpulsed CO_2 laser must operate with a pulse duration less than 1 millisecond:

$$\tau_p \leqq 1 \times 10^{-3} \text{ second.} \qquad (3\text{–}5)$$

This statement is made without physical proof, which rests upon a calculation of the *thermal diffusivity,* a parameter equal to the thermal conductivity of soft tissue divided by the product of mass density and specific heat of the tissue, and specification of the diameter of a small volume of tissue selectively absorbing the incident radiation. Anderson and Parrish, who did an analysis of transient heating of blood vessels by laser radiation, concluded that the time-duration of pulses of laser irradiation at a wavelength strongly absorbed in blood (577 nm) should be 1 millisecond or less in order to achieve selective damage to blood vessels in the range of 10 to 200 micrometers in diameter.[2] However, such analyses are based upon several simplifying assumptions that are but crudely fulfilled in actual living tissue, which is very heterogeneous and seldom isotropic.

Despite the uncertainties of mathematical calculations of temperature distributions in living tissue heated by laser beams, empirical measurements can be made in particular situations by using small thermocouples implanted in the tissue at various distances from the impacted region. Such measurements verify, in a general way, the concept that a laser-pulse duration of less than 1 millisecond will produce a damaged zone adjacent to the crater of a CO_2 laser that is less than 0.1 mm wide. The actual width of the damaged zone will decrease as the laser-pulse duration becomes shorter, *provided that the power density of the laser beam is not so high as to cause shock waves or explosions in the tissue.* For CO_2 lasers, the threshold of photoacoustic damage is about 1,000,000 W/cm².

Pulse-duration is explicit and easy to measure when the pulse is rectangular in time-shape, like that shown in Figure 3–5B for an

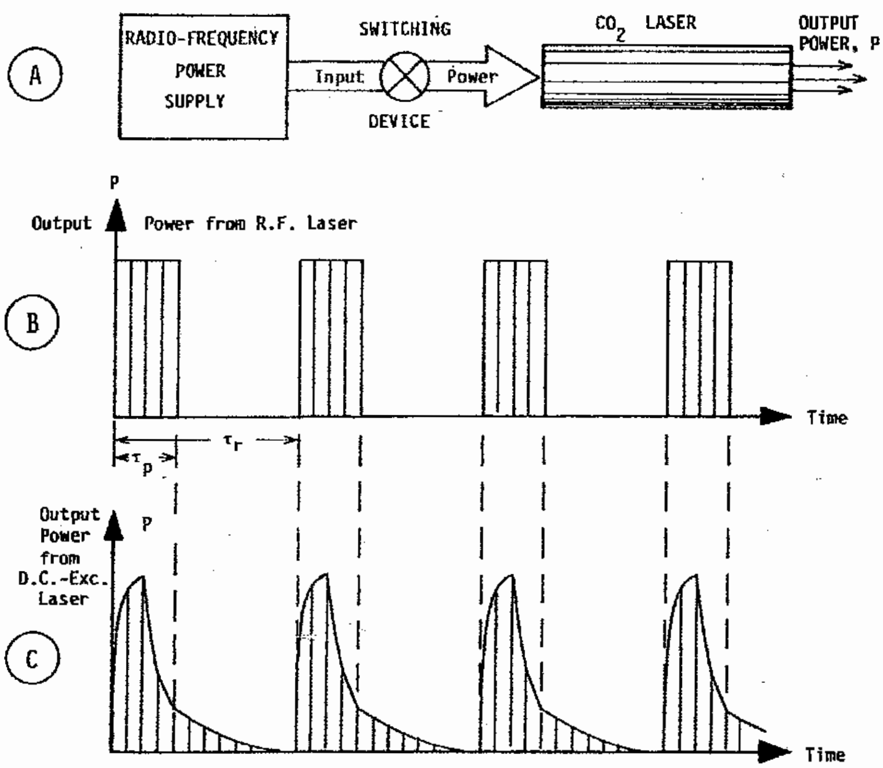

Figure 3–5.
A: Schematic diagram of a radio-frequency-excited CO_2 laser in superpulsed operation. B: Rectangular pulses of radiant output power. C: Pulses of radiant output power from a typical d.c.-excited CO_2 laser having a waveform of driving voltage identical to the rectangular time-envelope of voltage applied to the radio frequency-excited laser.

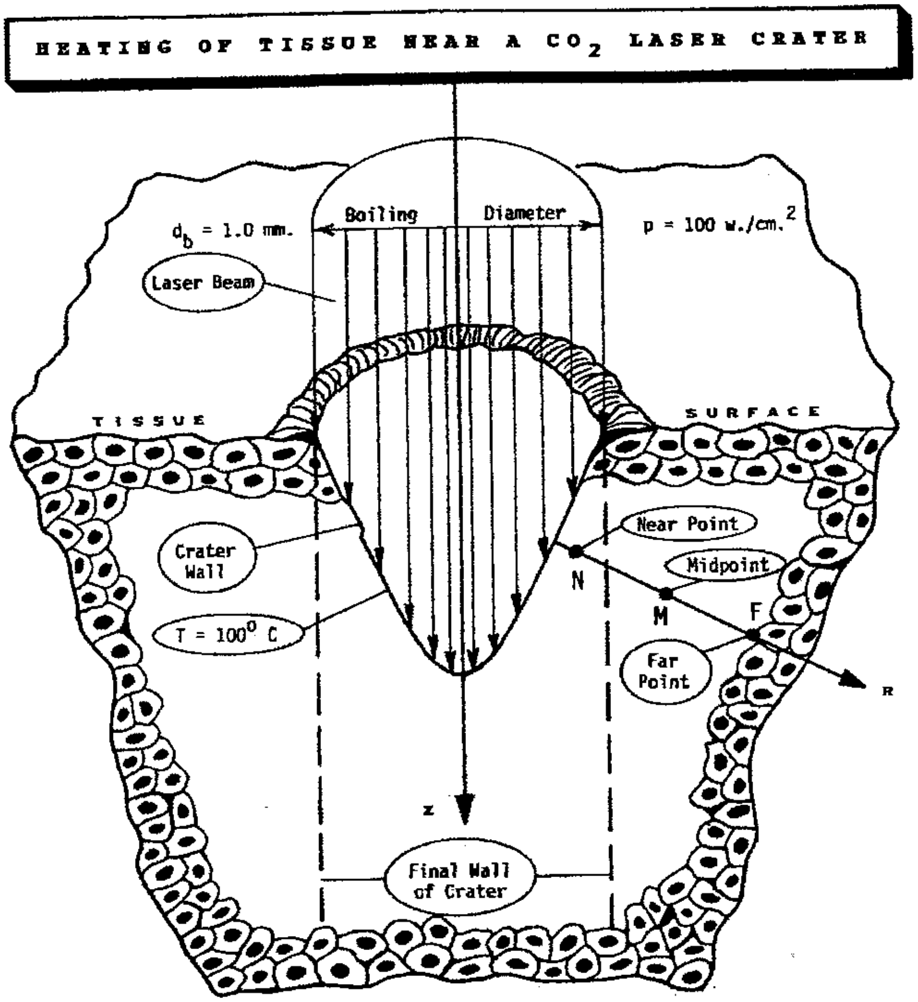

Figure 3–6.
Schematic diagram of the midplane longitudinal cross-section of a crater produced in soft tissue by a gaussian CO_2 laser beam that is stationary on the target. Points N, M, and F lie within 1 mm of the crater wall at the instant shown.

ing tissue are such that it is far from homogeneous and isotropic, and so the solution of the complicated differential equations of biologic heat transfer, even by computerized methods, is difficult without some simplifying assumptions. These often make the numerical results very inexact in their application to real situations.

For the average medical or surgical practitioner, a much more useful method of studying the dynamics of heat transfer in living tissue is the empiric method of measuring temperatures by implanting small thermocouples or thermistors in irradiated living tissue, or by viewing its surface with an infrared-sensitive thermocamera. Thermal models of real surgical situations that are based upon such measurements can yield much more immediately useful and interpretable information for the practicing physician than struggling through pages of esoteric equations.

The pioneering study of Mihashi, Jako, et al with a CO_2 laser in 1975 yielded some very basic insights into the thermodynamics of laser ablation of tissue.[1] Figure 3–6 shows schematically the midplane cross-section of a crater produced in soft tissue by a gaussian laser beam. The micro-mechanics of ablation of the tissue in such a crater will be examined in greater detail in Chapter 4. Here we are interested in the fact that tissue is ablated at the wall of this crater by conversion of laser light at 10,600 nm to heat and the sudden boiling of histologic water to form steam, which expands rapidly and ruptures the cells and intercellular bonds.

At power densities between 100 and 10,000

CHAPTER FOUR

John C. Fisher

Qualitative and Quantitative Tissular Effects of Light from Important Surgical Lasers: Optimal Surgical Principles

Interaction of Laser Light with Living Tissue

When a beam of laser light strikes the surface of living tissue, four basic physical phenomena occur. These can be measured in terms of the power density at various points along a ray of light within the beam as it passes from the open air outside the tissue into the depths of that tissue:

1. *Reflection* and *backscattering* from the first surface;
2. *Transmission* into or through the tissue;
3. *Scattering* within, and perhaps out of, the tissue;
4. *Absorption* by the tissue between scattering points.

The relative and absolute magnitudes of these phenomena are functions of wavelength and the physical properties of the tissue. Because the organic compounds within tissue can be broken down or altered by irradiation, these properties may be changed during exposure to the laser light, either on a short or on a long scale of time. Time-variation of the physical properties of tissue makes it impossible to do a purely analytic study of the interaction of laser light with living tissue, although numerical solutions to physical equations having variable coefficients are possible with the aid of computers. However, for purposes of understanding the basic phenomena and deriving some quantitative estimates of their magnitudes, it is sufficient to begin by assuming constancy of physical properties during irradiation, and to adjust the numerical values of the inconstant parameters to suitable average figures over intervals of time or space within which they are not changing rapidly.

An example of such an inconstant parameter is the thermal conductivity of water in living tissue. It is a function of temperature, which changes from 37° C to 100° C or more during irradiation by wavelengths at which light is converted largely to histologic heat. Another is the value of the scattering coefficient in tissue which is undergoing thermal necrosis: e.g., liver being irradiated by a Nd:YAG laser beam.

The errors introduced into our computations of important histologic variables by the assumption of constant physical parameters are of lesser importance than those caused by the assumption that those same parameters do not change from point to point within the tissue. Even on a *macroscopic* scale (i.e., over distances large by comparison with one cell-diameter), tissue is neither homogeneous nor isotropic. On a *microscopic* scale, the inhomogeneities within cells are even greater. In view of these facts, we shall avoid the elaborate high-order partial differential equations that delight the souls of computer-wielding scientists and engineers, and rely upon models derived from experiments to understand and quantify the complex interaction of laser light with living tissue.

Reflection

What appears to be reflection from the first tissue surface struck by a ray of laser light, if viewed by the eye or by a spectrophotometer, may actually be a composite of true optical reflection, as first described by Augustin-Jean Fresnel (1788–1827), and *backscattering* from the superficial layers of the tissue. Fresnel defined a numerical parameter known as *reflectance,* which is expressible as the ratio of the power density of the reflected portion of a ray of incident light to the power density of the incoming ray before it strikes the reflecting surface:

$$R = \frac{\text{(reflected power density)}}{\text{(incident power density)}}. \quad (4\text{–}1)$$

In general, when an unpolarized ray of light strikes the surface of a reflecting medium, the electric field of this ray parallel to the surface will be reflected with a different amplitude than that of the field perpendicular to the surface (i.e., in the plane defined by the incident and reflected rays). The net effect of this is some degree of polarization of the reflected ray. Another way of stating this is to say that if the incident ray is polarized parallel to the surface of the reflecting medium, its reflectance will be different than for a ray polarized perpendicular to the surface.

However, since we are not concerned about polarization in surgical applications of lasers, we shall assume that the ray of light impinging on the surface of tissue (1) strikes the tissue from a gaseous medium (usually air), (2) has an angle of incidence that is either 0° (i.e., perpendicular to the surface) or 90° (i.e., parallel to the surface), and (3) is unpolarized. Thus, for either a medium that does not conduct electricity (a dielectric) or a medium that is a good electrical conductor (a metal), the reflectance will be single-valued, both parallel and normal to the surface. Living tissue falls somewhere between dielectrics and good conductors, usually having a low-to-moderate electrical conductivity.

Under these assumed conditions, the reflectance at 90° incidence (parallel) is 100%, and at 0° incidence (normal) ranges from 4% to about 63%, depending upon wavelength of the light and the type of tissue. Figure 4–1 shows a plot of reflectance for normal incidence (0°) of monochromatic light on human skin. Note the steep peaks and valleys of both curves in the range of 400 to 1500 nanometers. Note also the pronounced differences between light and dark skin in this same range of the spectrum. Figure 4–2 shows the spectral variation of reflectance at normal incidence for human blood in vivo, over the visible range (400 to 700 nm).

In general, the reflectance of all living tissues at normal incidence will show pronounced variations within the spectral range from 400 to 1500 nm. The shape and maximum height of the curve for each kind of tissue will be strongly dependent on the pigments present in that tissue. However, in the ranges of 100 to 300 nm and of 2200 to 40,000 nm, reflectance is "colorblind".

The reader can easily see that as much as 50% of the radiant power of the beam of a Nd:YAG laser can be wasted by first-surface reflection from lightly pigmented tissue. This is costly, when one remembers that Nd:YAG lasers for surgery cost about $1500 per watt of output.

The most significant effect of reflection of laser light from living tissues is *the reduction of power density in the rays that penetrate into those tissues.* Figure 4–3 shows, schematically, a ray of laser light being partially reflected from the surface of first incidence on a mass of tissue. Table 4–1 gives values of reflectance for representative types of tissue within four ranges of the spectrum of laser radiation.

Absorption

Absorption of radiant energy occurs at the level of atoms, ions, molecules, and radicals (combinations of atoms that pass unchanged through

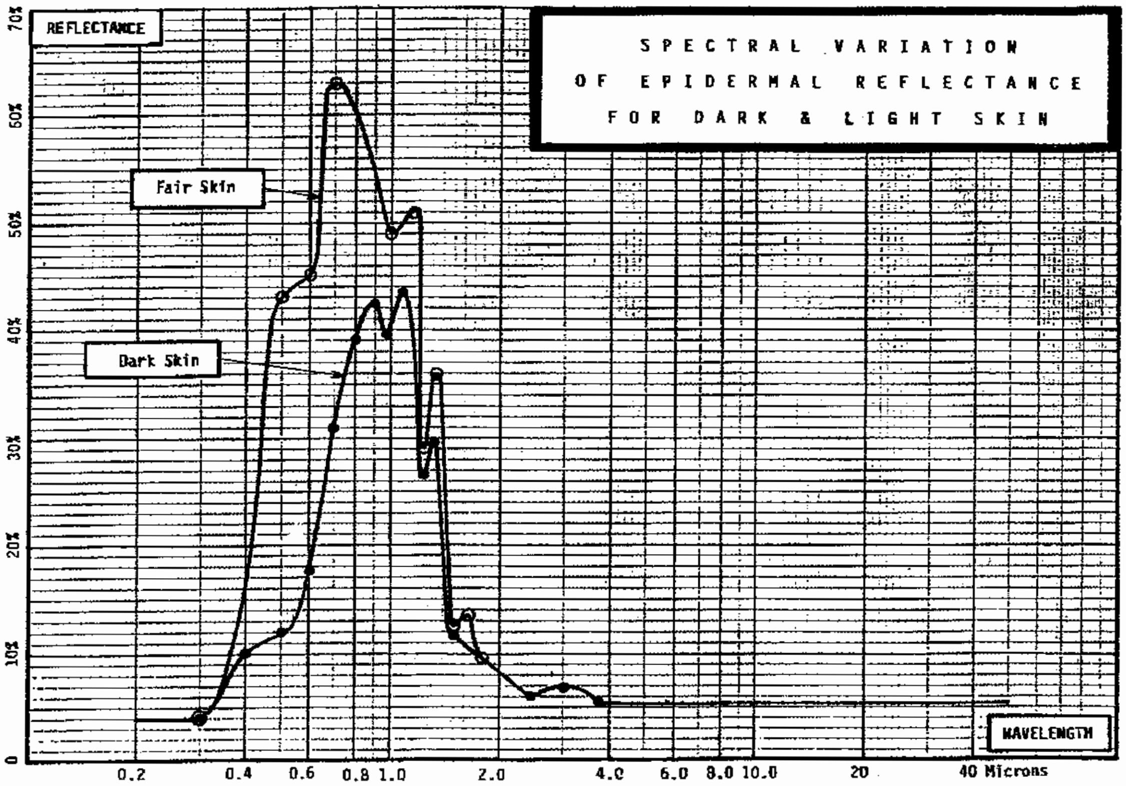

Figure 4–1.
Variation of epidermal reflectance for fair skin and dark skin with wavelength from 0.2 to 45 μm. Note that, below 0.3 and above 4.0 μm, reflectance is low, constant, and independent of wavelength. Reprinted from Fisher JC. Basic laser physics and interaction of laser light with soft tissue. In: Shapshay SM, ed. Endoscopic laser surgery handbook. New York: Marcel Dekker, 1987:94.

chemical reactions, but may be incapable of existing alone). It is a process of conversion of the radiant energy into other forms of energy:

1. Excited levels of atoms, ions, molecules, and radicals.
2. Chemical energy stored in new compounds.
3. Re-radiated energy at other wavelengths, usually longer than that of the absorbed radiation (fluorescence).
4. Thermal energy (heat): the kinetic energy of random motions of atoms or molecules in liquids and gases, or the vibrations of atoms and ions in solids, always at temperatures above absolute zero.
5. Kinetic energy of free electrons or ions in plasmas, or of molecular fragments produced by photochemolysis.

For all of the lasers whose wavelengths are greater than 319 nm, the conversion of light to heat is the major means by which tissue is destroyed. At 319 nm, the photonic energy is equal to the first ionization potential of the element cesium, 3.89 electron volts. Cesium has the lowest first ionization potential of all the elements. Therefore, since photonic energy increases with decreasing wavelength, all lasers having wavelengths shorter than 319 nm are capable of producing photochemolysis at relatively low power densities. The only lasers currently available in that range of the spectrum are the excimers argon-fluoride (193 nm), krypton-chloride (222 nm), krypton-fluoride (248 nm), and xenon-chloride (308 nm).

Photochemolysis can occur at any wavelength for which the photonic energy is equal to or greater than the bonding energy between two linked atoms in a molecule. This bonding energy may be lower than the first ionization potential of cesium. However, a convenient dividing wavelength between the spectral range in which photochemolysis predominates and that in which thermolysis predominates is 319 nm.

Even at these short ionizing wavelengths, if the average power density of the beam far exceeds the threshold for photochemolysis, the excess will be converted to heat in the tissue, and then thermolysis will occur. However,

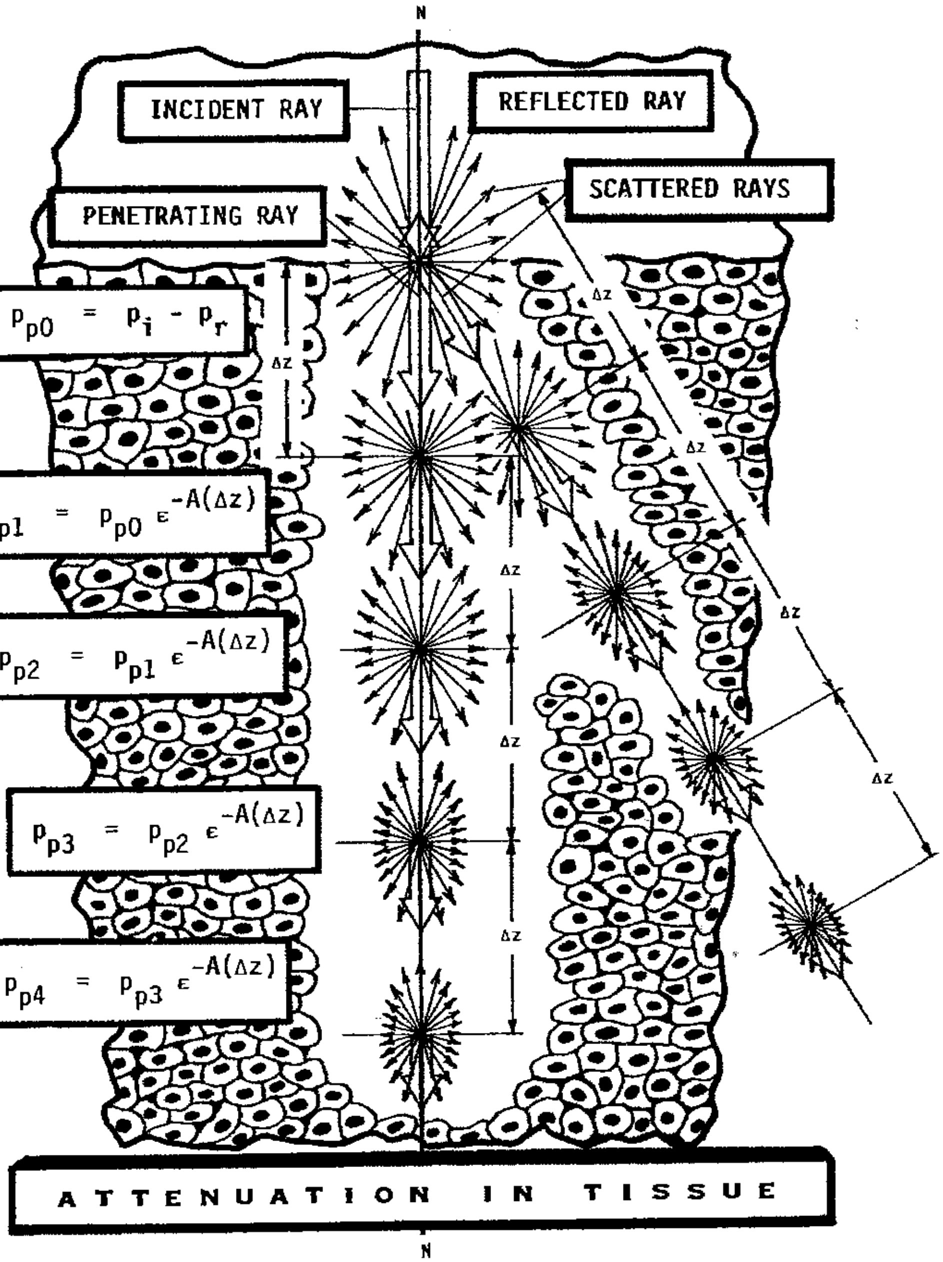

Figure 4–5.
Schematic diagram of the attenuation of a ray of monochromatic light by absorption and scattering within living tissue. This process is exponential: each penetrating ray loses a constant fraction of its intensity in the direction of propagation in each unit distance (Δz) of forward travel. The porcupine figures depict omnidirectional scattering.

base of the natural logarithms (2.71828-----), and A is the *attenuation coefficient*. Equation 4–3 is the mathematical expression of *Bouguer's law*, named after the French scientist Pierre Bouguer (1698–1758). Equation 4–3 has been attributed both to Bouguer and to the German physicist Johann Lambert (1728–1777). It has erroneously been called Beer's Law, which actually states that the absorption coefficient of a medium is directly proportional to the concentration of the absorbing element in that medium.

Figure 4–5 shows, schematically, the attenuation of a ray of laser light as it penetrates into living tissue. The porcupine-like figures depict omnidirectional scattering, which is generally means of an electrosurgical device (often called a "bovie"): *hemostasis and minimization of adjacent thermal trauma are inherently antagonistic objectives.* Whatever is done to enhance one will diminish the other. The final result will always be a compromise. In many cases this compromise is not critical. At the critical end of the surgical spectrum, extreme care must be exercised to do precise removal of tissue (e.g., in stapedectomy and footplate fenestration in the ear). At the other end, as in excising a plantar wart on the foot, removing some excess tissue may actually be helpful, and simultaneous hemostasis may be a desirable goal to prevent the spread of the causative virus.

When a laser surgeon is dealing with a lesion that must be meticulously removed (e.g., a tumor on the optic nerve), as opposed to one that can be less carefully destroyed (e.g., a small uterine fibroid in a post-menopausal woman), the first objective should be precise removal of the lesion with minimum thermal trauma. Hemostasis may still be a secondary concern, but that can be achieved in ways other than coagulation of blood vessels by a laser beam. In this situation, the first principle of laser surgery should be observed: *the surgeon should always use the highest power density in the laser beam that is compatible with his or her level of eye–mind–hand coordination, without unintended removal of surrounding tissue.* The total time of exposure to the laser beam should be minimized.

To enhance achievement of this objective, the laser should be either an inherently pulsed type or a continuous-wave laser used in a pulsed mode. Superpulsing, as defined in Chapter 3, is presently ~~available only on~~ certain CO_2 lasers, but could be adapted to other c.w. lasers as well.

When the surgical target is a lesion that can be less accurately destroyed, such as a cavernous hemangioma on a buttock, where control of bleeding may be the primary problem, then thermal necrosis may be appropriate. The laser of choice in this case is the Nd:YAG, used in continuous-wave fashion with non-contacting delivery of the beam. When the objective is massive thermal necrosis, then the proper technique is to *apply low power density for long periods of time, while watching the lesion for blanching and shrinking without vaporization.* The surgeon should not be impatient and try to cause total necrosis immediately. Lesions that are Nd:YAG-irradiated continue to undergo necrosis for minutes, hours, or even days after the exposure.

Using the Right Laser for the Procedure

There is no single laser (wavelength) that will perform every kind of surgery optimally. Surgeons in small hospitals that have only one or two lasers may be tempted to use them for almost any procedure. While this may be done without getting into trouble, it is not likely to produce the best results. The two mainstays of general laser surgery are still the CO_2 and the Nd:YAG, the latter with both contact and non-contact delivery of the beam. The SYCUTE lasers (visible, Ho:YAG, and Tm:YAG) may be the appropriate choice in special applications like joint surgery, because of their fiber-deliverability. For safe, precise surgery the WYSIWYG lasers (excimers, CO_2, and Er:YAG) are best. For coagulative surgery or photodynamic therapy, the WYDSCHY lasers (Nd:YAG, He-Ne, gold-vapor, etc.) are most appropriate.

Whenever a laser surgeon is either trying to avoid thermal injury to tissue or attempting to cause thermal necrosis, he or she should be aware that there is a transitional zone between death and survival of biologic systems which are heated above their normal temperatures. This is best expressed on a plot of temperature and time, like that shown in Figure 4–13. This plot shows that any combination of temperature and time corresponding to a point lying above the transition zone will cause death of the tissue, and that any combination lying below this zone will allow the tissue to recover. The center line of this transition zone can be approximated by the equation

$$T_c = 37 + 31.5t^{-0.164}\ ^{\circ}C, \qquad (4\text{–}10)$$

where T_c is the temperature threshold at which irreversible necrosis begins, and t is the time in seconds from the onset of heating, in the range 1.0 second $\leqq t \leqq$ 1000 seconds. Note that Figure 4–13 does not apply exactly to every living tissue, but it indicates the general regions of survival and death in terms of time and temperature. It was plotted by the author from the data of Henriques.[7]

Evolution of Applications in Surgical Lasers

Over the years, new applications of surgical lasers have evolved. Some

3-mm laser spot at successive positions on a flat surface in a regular geometric pattern, within a second or less. The relative location of the centers of adjacent circular spots are the apexes of equilateral triangles, juxtaposed to create circles, squares, rectangles, triangles, hexagons, parallelograms, lines, and / or doughnuts. By varying the center-to-center distance of each equilateral triangle in this lattice, the diametric overlap of adjacent circular spots can be varied from -20% to +50%. The pattern size can be varied from one spot to a maximum of 20 mm x 20 mm, and the effective diameter of each spot can be varied from 1.5 mm to 3.0 mm. Pattern generators increase the speed and relative precision of "painting" most treatment areas, but elevated lesions, lesions with deeper areas of pigmentation, and deeper rhytids still require additional attention with different settings and/or hand pieces.

The outline of the preset pattern is visually delineated by the red helium-neon aiming beam, which sweeps around the perimeter of the pattern before the operator fires the laser. When the foot switch is depressed, the CPG® places the beam, in step-wise fashion, once at every position in the lattice of the pre-selected pattern, and the laser fires a single shot at each position. The system can be set to fire one complete pattern, or a pre-selected number of patterns, on the same area. The obvious advantage of such a system is that it eliminates from the epidermal-ablation process any dependence of the results upon the surgeon, except for the initial choice of operating parameters.

In 1996, Clinicon Corporation introduced is SureScan® pattern generator, which has since been adapted to the lasers of several manufacturers, both CO_2 and Er:YAG. In that year, Sharplan and Heraeus introduced pattern generators for their CO_2 lasers. However, pattern generators can negate one of the significant advantages of that wavelength, as will be discussed later. Also, increasing spot sizes and repetition rates in newer lasers have decreased the need for pattern generators, added more precision, and decreased the rate of scanner footprints (hypo-pigmentation from overlapping scans with fluence too high for the melanocytes in the treatment area to recover from) to this procedure.

h. Inevitable Thermal Damage to Tissue Below the Ablated Layer: Mesa-Mode Beams

The previous sections have discussed ways in which pyrolytic damage to underlying tissue, resulting from thermal <u>conduction</u>, can be minimized. There is another cause of pyrolytic damage, however, than cannot be mitigated by pulsing the laser beam, or by using high fluence and power density. This is absorption of laser light by the underlying tissue <u>at intensities below the threshold of boiling</u> the histologic water.

Refer now to **Figure 5-8**, which depicts an ideal WYSIWYG laser beam of square cross-section and width much larger than the extinction depth, having a perfect mesa-mode profile of power density and / or fluence, perpendicularly striking the surface of soft tissue. The thermal effects produced by a mesa-mode beam are simpler to analyze than those of a gaussian beam, which will be discussed later.

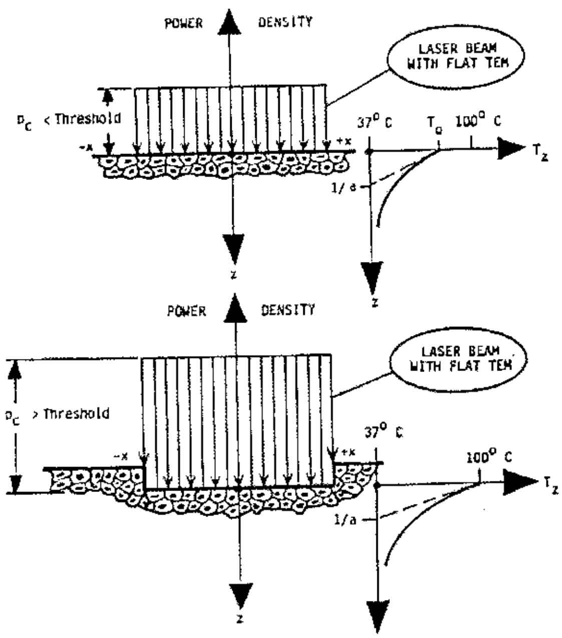

Figure 5-8.
Schematic diagram of a short-pulsed, ideal mesa-mode laser beam irradiating soft tissue. When the power density is below the threshold of ablation (upper diagram), the laser rays penetrate into the tissue below the first surface, causing instantaneous heating. The temperature is highest at the surface, but below the boiling point, and declines exponentially with depth, as shown at the right-hand side of the figure. If the power density is raised above the threshold of boiling (lower diagram), ablation of tissue begins at the surface, which moves downward, the incoming rays are attenuated at the boiling surface to the threshold level, and they still penetrate into the tissue below. The initial slope of the curve of temperature vs. depth is such that the tangent to the curve at the surface intersects the z-axis (for 37° C) at a depth **1/a** in each case. Inevitable thermal necrosis of subsurface tissue will occur to a depth at which the temperature is equal to the necrotic value for short exposure.

Reprinted from Fisher J.C.; Basic biophysical principles of resurfacing human skin by means of the carbon dioxide laser; Journal of Clinical Laser Medicine and Surgery, 1996: 4:198.

Assume that this beam is pulsed with a duration short enough to make thermal conduction out of the heated volume insignificant by comparison with the energy absorbed by the heated volume during the pulse. Assume also that the fluence delivered by a single pulse of the beam is below the threshold of boiling, as given by **Equation (5-7)**, and that the temperature of the tissue prior to this single pulse is 37° C.

Under the foregoing conditions, the surface of the tissue will not reach the boiling temperature during the single pulse, and the power density and fluence within the tissue will be exponentially attenuated by absorption with insignificant scattering, as quantitatively described by **Equation (5-1)**, is **s** is set equal to zero. Because the histologic absorption is instantaneous (i.e., it occurs within a fraction of a microsecond) and there is no loss of heat from the absorbing volume of tissue, the temperature of the tissue at all points within the extinction depth will instantly rise above 37° C. The highest temperature will be at the irradiated surface of the tissue, and the histologic temperature will decline exponentially with depth below that surface,

$$\text{(5-21)} \qquad \Delta T_z = T_z - 37^\circ = (\Delta T)_{max}\, \varepsilon^{-az} \qquad {}^\circ\text{Celsius}$$

where $\mathbf{T_z}$ is the tissue temperature at depth **z**, and $(\Delta \mathbf{T})_{max}$ is the temperature rise at the surface of the tissue. The surface temperature rise is given by,

$$\text{(5-22)} \qquad (\Delta T)_{max} = a\,(f_o / h_s)$$

where $\mathbf{h_s}$ is the specific heat of the histologic water at its prevailing concentration, [ENERGY] / ([VOLUME) x [DEGREE ΔT]), and $\mathbf{f_o}$ is the

c. Heat in Laser Resurfacing: Detrimental vs. Beneficial Effects

Inevitable thermal necrosis is essentially a second-degree burn, whether it extends only to the basal layer of the epidermis or actually invades the papillary dermis. This is a deleterious effect of heat. The cosmetic surgeon should do everything possible to minimize residual thermal necrosis below the ablated layers of skin, e.g. make as few laser passes as possible to exfoliate to the papillary dermis, and he/she may be well advised to cool the tissue pre-operatively with either contact cooling, or cooled tumescent anesthesia, if aggressive resurfacing is planned.

Many practitioners of resurfacing skin by means of lasers advocate removal of the epidermis by several sequential passes of the laser beam, at low to moderate fluence, completely over the same area of skin, with mechanical removal of desiccated epidermal debris between one pass and the next. When the CO_2 laser is used, as was pointed out in a previous section of this chapter, such dehydrated residue must be removed because it strongly absorbs the laser light at 10,600 nm but has no water remaining to hold the temperature at 100^{o} C. However, those who do such mechanical debridement should do it gently, not vigorously, because vigorous scrubbing of the face (practiced by some cosmetic surgeons) is nothing more than manual dermabrasion, which probably does not add anything of value to a laser resurfacing procedure, and could just as well be performed without a laser.

As has been noted in prior portions of this chapter, it is not necessary to perform mechanical debridement of desiccated residue between passes of the Er:YAG laser, unless multiple passes are made, or a build-up of eschar is noted, because the absorption at 2,940 nm in this cutaneous debris is much lower at that wavelength that at 10,600 nm.

The beneficial role of heat, generated by absorption of laser light in the avascular epidermis, is simply to create a field of miniature, localized explosions caused by the ultra-rapid flashing of histologic water into steam, which strip away the epidermis precisely and efficiently. If a laser with a water-absorption coefficient of 1,000,000/cm were available, it would perform that task with a vanishingly small zone of inevitable thermal necrosis under the ablated layer.

d. Thermal Shrinkage of Collagen During Facial Resurfacing by Lasers

It has been presented at cosmetic surgery meetings that the heat generated in the skin by absorption of laser light causes shrinkage of old, damaged collagen fibers, and that visible skin shrinkage is an important end-point of the resurfacing process. While it is true that infrared light is absorbed by collagen, and that collagen fibers shrink when heated, this biophysical effect is not likely to be a significant contributor to resurfacing.

Thermal shrinkage of collagen occurs between 55^{o} C and 58^{o} C, and involves the rupture of hydrogen bonds. Melting or denaturing of collagen occurs between 60^{o} C and 70^{o} C. These ranges of temperature are so close that any deliberate attempt to attain the shrinking range without reaching the melting range is virtually impossible with a thermolytic WYSIWYG laser applied by a cosmetic surgeon who can pre-select only the values of energy per pulse and pulse-repetition frequency. It is often stated by speakers at meetings on cosmetic surgery that the surgeon can see collagen shrinkage when a laser beam strikes skin. What is actually observed, however, is shrinkage caused by desiccation, i.e. when the water content of epidermis is reduced from 70% to lower values, the histologic structural matrix contracts.

The time-waveform of the power during each pulse, which always has a duration of about 1.0 millisecond, is nearly rectangular, so that it delivers maximum energy during each pulse. At this writing, the UltraPulse® laser system is the most versatile, but also the most expensive, available for surgery of any type that can be performed effectively by a CO_2 laser.

The absorption coefficient for the CO_2 laser is 770/cm in liquid water, and is comparable in nearly all biological substances; its scattering coefficient in any histologic material is negligible by comparison, making it an ideal WYSIWYG laser. It is well suited to incisive procedures like blepharoplasty, because of its small minimum focal-spot diameter, and its ability to achieve hemostasis in vessels of 0.5 mm or smaller.

b. Erbium:YAG Lasers for Resurfacing of Skin

Figure 5-12 shows the NaturaLase® erbium:YAG laser, manufactured in the United States by Focus Medical. It is one of a number of such lasers offered for sale in the United States by various manufacturers.

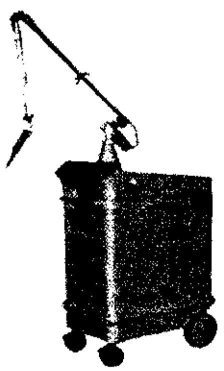

Figure 5-12.
Photograph of the NaturaLase® erbium:YAG laser for resurfacing of skin. Source: Focus Medical, Bethel, CT.

Generically, the erbium:YAG laser is similar to its cousins, the neodymium:YAG and the holmium:YAG. "YAG" is an acronym for Yttrium Aluminum Garnet, a crystalline material having high transparency in the visible, near-infrared, and mid-infrared portions of the spectrum. It has the additional advantages of good thermal conductivity, high mechanical strength, ability to operate at high temperatures, and resistance to thermal shock.

YAG's physical properties are as follows: density, 4.55 gm/cm^3; melting point, 1,970° C; thermal conductivity, 0.13 watt/cm/ °C; coefficient of thermal expansion 6.9 x 10^{-6}/ °C; and refractive index, 1.823. Its crystalline structure is cubic, and it can be doped with a number of rare-earth metals. The concentration of dopant is in the order of 1% in most cases. The primary wavelength emitted during laser action is determined by the dopant. For erbium, it is 2,940 nm.

Unlike the Nd:YAG, the Er:YAG laser cannot be operated in continuous-wave mode at room temperature, because the lifetime of the lower laser state is too long to allow the Er ions to relax quickly back to the ground state. Since its wavelength, 2,940 nm, is beyond the transmission range of quartz fibers, it is relegated to the articulated arm for delivery of the beam. The usual means of pumping this laser is a pulsed, high-intensity xenon or krypton flashlamp, which can deliver hundreds of watts of radiant power, but at efficiencies limited to 2% or so, thus necessitating copious liquid cooling to prevent thermal damage to the flashtube.

An Nd:YAG crystal can deliver 200 watts of continuous-wave radiant power from a rod one cm in diameter and 10 cm in length. Because the rod is short, sturdy, and dimensionally stable, the resonator is simple and rugged. A typical flashlamp-pumped YAG resonator is shown in and pulse durations are adjusted precisely to limit the effected zone of treatment and adjacent damage.

Definition of Risk

Risk is an important concept in the study of safety. Risk can be high if a possible hazardous event has a large probability of occurring, even though the consequences of that event are not very morbid, and never fatal. It can be high also if the probability of occurrence is low, but the results are always very morbid or fatal. Therefore, we shall define the risk of an accident as follows:

risk = (probability of occurrence) X (severity of consequences)

Laser-Specific Hazards/Risks

- Burns from laser-ignited combustion
- Accidental laser trauma to unintended targets
- Inappropriate or unskilled clinical use of lasers
- Adverse sequelae of laser surgery or therapy
- Malfunction of lasers and related equipment

In the subsequent sections, these five categories of mishaps shall be examined in detail. Not all of them are relevant to every discipline of laser surgery, but a complete discussion of laser safety must include them.

1. Burns from Laser-Ignited Combustion

Early estimates by ECRI Institute suggested that there were approximately 100 surgical fires per year. They have been investigating surgical fires and publishing on their prevention for more than 35 years. However, in 2007 the Pennsylvania Patient Safety Authority published the first hard data on the incidence of surgical fires in Pennsylvania facilities. According to the statistics, the chances of a surgical fire in Pennsylvania are 1 in 87,646 operations—an average of 28 surgical fires per year in Pennsylvania alone. ECRI Institute scaled the Pennsylvania statistics to the United States in two ways: based on population and on the number of surgical procudures. From that analysis, the ECRI Institute estimates that the number of surgical fires in the United States each year ranges from 550 to 650 per year. Of these, about 20-30 are serious, with disfiguring or disabling injuries. One or two fatal fires occur each year, most of which are airway fires. 75% of the fires are caused by O_2 enrichment and combustion under surgical drapes. 4% are due to surgical scrub preps containing alcohol (Duraprep, Chloraprep, etc.), hence the need to let them completely dry or simply use non-alcohol containing preps. Ignition sources are reported as follows: 70% due to Bovie cautery; 20% due to hot wire cautery pencil, defibrillation pads, fiberoptic light sources, and sparks from surgical burs; 10% due to lasers. The ECRI Institute judges that the frequency of surgical fires is generally comparable to that of other rare surgical misadventures, such as retained instruments or wrong-site/patient surgery.

More recent cases of fires due to lasers involve tracheostomy and bronchoscopy. The sheath or fiber may be inappropriately stripped or O_2 causes ignition of the end of the bronchoscope because the laser is fired too closely to the end of the scope. Two reported cases (by the ECRI Institute) of fire outside of the body are mentioned here for example. First, there is a report of a surgeon's clog shoe slipping and activating the foot pedal of a laser which fired through the legging drapes and ignited the inner surgical towels on the perineum and burned for 30 seconds. A second accidental case occurred when two sources for cautery and laser were being used in a neurosurgical case. Both foot pedals were pressed when only one was thought to be pressed and the surgical drapes caught fire causing an 18% BSA burn to the patient. It is obvious in both of these cases, and likely in most other cases, that the fires were and are caused by error on the part of the device user. Finally, the ECRI has on file 5 fires due to lasers since 2002 (one per two years) and the FDA MAUDE system reports 21 since 1991.

The ND:YAG is also used to remove darker, deeper hair especially in darker skin.

There are multiple devices for facial resurfacing, skin tightening, texture smoothening, and photorejuvenation. These include CO_2 (ablative and fractional ablative), Er:YAG, infrared, radio frequency, intense pulsed light (IPL), and others. The practitioner must know the theory and physics of each device and its uses to determine which device to use for each condition.

Cosmetic vascular conditions including rosacea, telangiectasia, angiomas, port wine stains, and others are commonly treated with pulsed-dye lasers, ND:YAG, IPL, RF, and others.

Flat, pigmentary lesions such as solar lentigines, hyperpigmentation, minimally elevated nevi, keratoses, melasma, and others can be treated with many devices including Q switched Nd:YAG, IPL, CO_2, and others.

Different lesions can be effectively treated with various modalities. It is the responsibility of the laser practitioner to make educated selections of appropriate wavelengths and parameters, seek equipment-specific training, and become knowledgeable of the conditions and the treatment devices.

Inappropriate or Unskilled User of Lasers

One of the most common causes of complications with lasers is the unskilled or under-skilled laser practitioner. In the authors' opinion, this is currently much more common in the cosmetic laser realm, but is not unheard of in the surgical realm. Simply put, one must seek advanced knowledge of the condition to be treated and advanced training in the procedure to be performed. Currently, there is a wide variation of state medical board requirements for required training, and one can easily find this information on the individual state board of medical examiner's website or by calling the state medical board. Although one must meet the requirements of the state to perform certain procedures, the practitioner often times must seek more training than simply that which is "required."

One of the most common problems with laser procedures in the cosmetic realm is that many practitioners performing the "lower end" cosmetic procedures (laser hair removal, laser spider vein treatment, intense pulsed light, and ablative and non-ablative facial resurfacing) simply do not have the knowledge of laser theory and physics, the medical/cosmetic conditions being treated, or the lasers themselves, and are not properly supervised by a trained physician. Many of these procedures are unfortunately perfomed in "money-driven" facilities and not "patient driven" facilities. These procedures are often not taken seriously but can obviously cause major complications. Burning, scarring, pigmentary complications, and lack of expected result are usually not life-threatening, but can still be serious and cause mental and physical anguish. In addition, it is unfortunately common that a non-physician laser practitioner is not properly supervised by the responsible physician because the procedure is not thought to be dangerous or serious. Therefore, all laser practitioners must seek the proper training and comply with state and federal regulations regarding these procedures and supervision.

4. Adverse Sequelae of Laser Surgery or Therapy

Smoke and Vapor from the Surgical Target

Combustion of dehydrated organic residues produces smoke, which is usually very odorous and can be harmful to the respiratory tract of anyone who inhales it.

Mechanism of Smoke Generation

The most copious generators of smoke are monopolar electrosurgical instruments, hot-wire-cautery devices, and CO_2 and Er:YAG lasers. Among the other lasers now used in surgery, the production of smoke depends heavily on the absorption coefficient of the laser's wavelength in the type of tissue being treated. Tissue-wavelength combinations having high absorption coefficients produce the most