The American Board of Laser Surgery

Incorporated 1984

Excerpts from the

ABLS STUDY GUIDE

(The complete Study Guide is used as part of the ABLS Certification process. These excerpts are for use in certain seminars and courses on medical lasers that the ABLS may be associated with, but is not a substitute for the actual ABLS Certification process. Information about the Board's Certification is available at the ABLS website: www.americanboardoflasersurgery.org)

Introduction

CHAPTER 1

Fundamentals of Laser Physics, Optics and Operating Characteristics for the Clinician

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CHAPTER 2

Surgical Delivery Systems

CHAPTER 3

Laser Biophysics, Tissue Interaction,

Power Density and Ablative Resurfacing of Human Skin: Essential Foundations for Laser Dermatology and Cosmetic Procedures

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CHAPTER 4

Commentary on Ethics in Cosmetic Laser Surgery

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Safe Use of Lasers in Surgery

CHAPTER 6

Considerations in the Selection of Equipment

Contents and Topics in the Study Materials

The study material consists of the proprietary ABLS Study Guide on fundamental laser science, delivery systems, biophysics/ tissue interaction, ethics, laser safety, and equipment selection. *For cosmetic practitioners only* it also includes several chapters from two excellent books on a broad range of cosmetic laser and light procedures: *Lasers and Lights: Procedures in Cosmetic Dermatology, 2nd Ed. David J. Goldberg, editor*; *and the 3rd Ed. with George Hruza and Matthew Avram, editors. In addition, two journal articles address the science and use of IPL and LED.**

(*Note: these chapters and articles are licensed from the publisher).

The following are the specific contents and topics of the study materials (all practitioners):

The Study Guide (Spiral Bound)

Chapter 1: The Fundamentals of Laser Physics, Optics and Operating Characteristics (the foundations of laser physics and beam delivery that are important for any laser medical discipline).

Chapter 2: Surgical Delivery Systems (an essential foundation for all clinicians in the various methods of beam transmission, delivery and focusing).

Chapter 3: Laser Biophysics, Tissue Interaction, Power Density and Ablative Resurfacing of Human Skin: Essential Foundations for Laser Dermatology and Cosmetic Procedures (focused on is ablative resurfacing which addresses the interaction of lasers and light with human skin, as an essential foundation for dermatology and cosmetics).

Chapter 4: Commentary on Ethics in Cosmetic Laser Surgery (key considerations in providing optimum patient care).

Chapter 5: The Safe Use of Lasers in Surgery (oriented to the needs of the actual practitioner as opposed to support personnel)

Chapter 6: Considerations in the Selection of Equipment.

Two Appendices are also included at the rear of the Study Guide.

Chapters on Cosmetic Laser and Light Procedures and Journal Articles

for Cosmetic Practitioners only

If you are a cosmetic laser and light practitioner, your Study Guide also contains reprints of the several chapters as described below, and also reprints of two journal articles which address intense pulsed light (IPL) and light-emitting diode (LED) technologies.

Lasers and Lights: Procedures in Cosmetic Dermatology, 2nd and 3rd Editions, are comprehensive study books that address several additional and essential cosmetic laser and light specialties including vascular lesions, leg veins, pigmented lesions and tattoos, scars, hair removal, photodynamic therapy, non-ablative skin rejuvenation (including fractional), and skin tightening by means of light-based and other radiofrequency procedures. The focus in the Board's Written Examination will be on the laser science and bio-tissue interaction in these particular areas. Candidates however should also review the examples of clinical applications in each chapter.

Excerpts from

CHAPTER ONE

Fundamentals of Laser Physics, Optics and Operating Characteristics for the Clinician

John C. Fisher, Sc.D.

Edward M. Zimmerman, M.D.

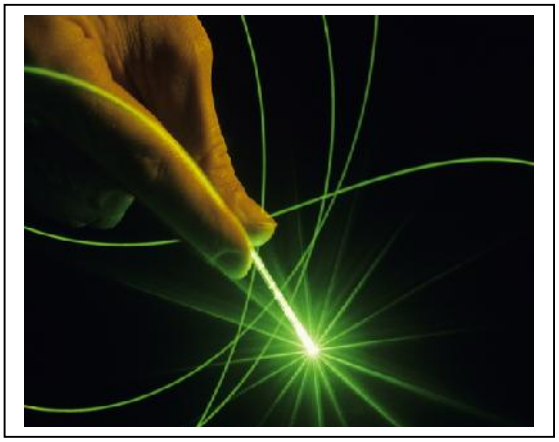
The Nature of Radiation

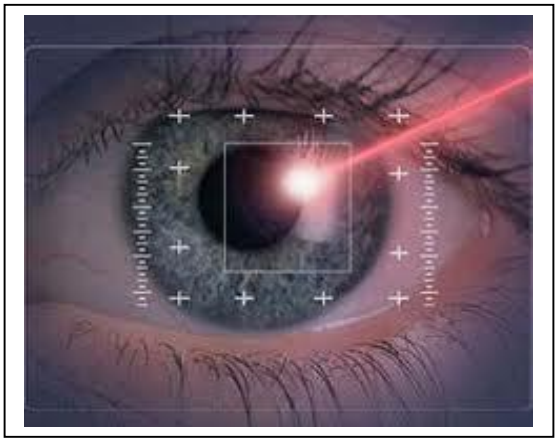
The word *laser* is an acronym composed of the first letters of the words Light Amplification by Stimulated Emission of Radiation. Of these, the most important is *radiation*. The other words describe the means by which lasers generate radiation. Radiation may be defined as the transmission of energy from one point in space to another, with or without an intervening material medium. *Electromagnetic radiation* requires no medium for its transmission: it can travel through free space devoid of any matter whatsoever. It can also propagate through space containing matter in the form of gases, liquids or solids. Upon entering such media from free space, electromagnetic radiation will, in general, be changed in direction and speed of propagation.

Radiation can also be *mechanical*: the transmission of vibrations through a material medium. Sound is an example of this sort of radiation. Unlike the electromagnetic kind, mechanical radiation does require the presence of a material medium for its transmission. However, the medium need not move as a whole; its particles merely oscillate elastically about fixed positions, transmitting energy from one to the next.

Lastly, radiation can be a stream of *material particles*, such as electrons, protons, neutrons or other atomic fragments. This kind of radiation needs no material medium for its transmission, but can pass through various media, usually with some attenuation and/or change of direction. Particulate radiation requires a transfer of *mass*, and the energy transmitted is the *kinetic* energy of the moving particles.

Because electromagnetic radiation is what lasers produce, we shall look only at this kind. There are two basic theories to explain the physical phenomena of electromagnetic radiation: the *wave theory* and the *photon theory*. The older of these is the wave theory, first described by the Scottish physicist James Clerk Maxwell (1831 - 1879) in the year 1864.[1] This theory can adequately explain all the optical phenomena of *light* that have been observed since the dawn of civilization, such as reflections, refraction, diffraction, interference, and polarization. It also accurately describes the 20th century phenomena of radio and radar. However, it cannot adequately explain many of the physical phenomena discovered since the turn of the 20th century, such as the spectral distribution of radiant power from a hot-body source. The German physicist Max Plank (1858 - 1947) early in the 20th century found it necessary to modify the wave theory in order to make the theoretical description of hot-body radiation agree with the empirically observed facts. His *quantum theory* also accounts for such discoveries as the photo-electric effect, light emitting diodes, fluorescence, photochemistry, and lasers.





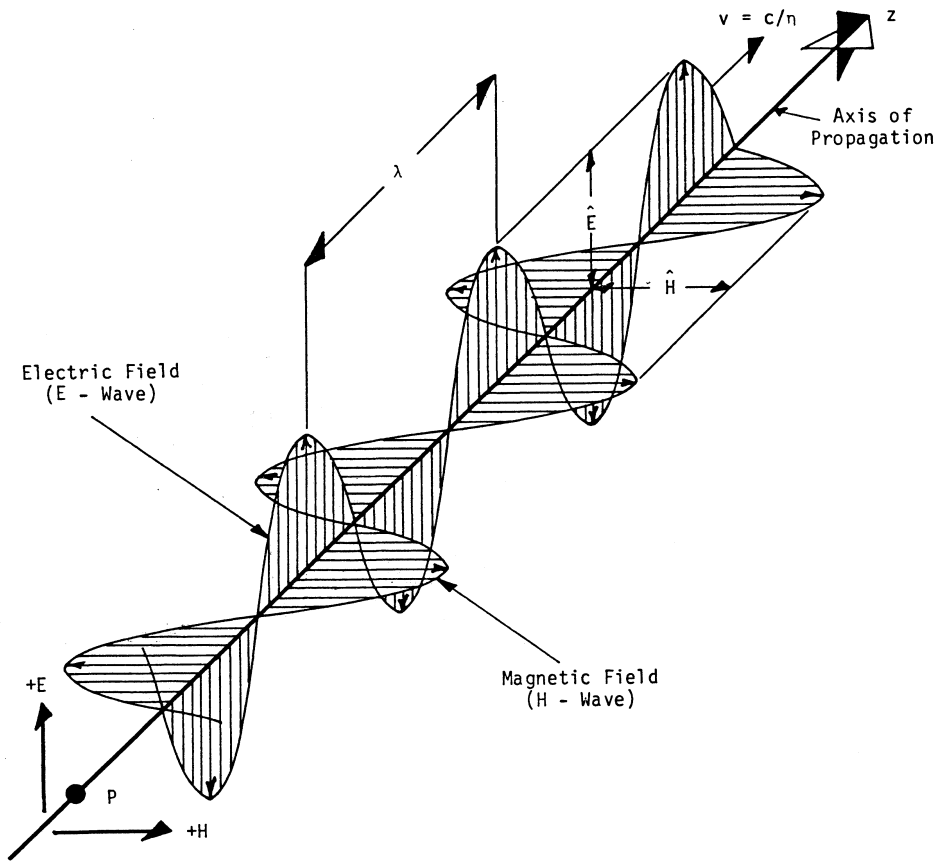


Figure 1-1.

Light depicted as orthogonal waves of electric and magnetic fields. Shown here is a plane-polarized ray of light. Unpolarized light would have the vectors of electric-field intensity and magnetic-field intensity radiating from the axis of propagation in all possible directions, like the spokes of a wheel. 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, 1987:4.

The Wave Theory

This explanation of electromagnetic radiation describes it as traveling waves of electric (E) and magnetic (H) fields that move at high speed through empty space or material media in straight lines. **Figure 1-1** shows a single ray of such radiation. The ray direction is the axis of propagation along which the waves move. The waves are sinusoidal in shape, and the axis-crossings of the electric-field wave coincide with those of the magnetic-field wave. **Figure 1-1** shows a plane-polarized ray: the electric and magnetic fields each exist only in one place. The E-wave and the H-wave are always perpendicular to each other and to the ray direction. A non-polarized ray, the usual kind, would have E-waves radiating outward from the ray direction in all possible planes, like the spokes of a wheel, and for each E-wave there would be a corresponding H-wave angularly displaced from it by 90°.

An electric field may be defined as a region of space within which an electric charge will experience a force parallel to the direction of the field-vector at all points. A magnetic field may be defined as a region of space within which a moving electric charge will experience a force mutually perpendicular to the direction of the field-vector and to the direction of motion of the charge. An electric field may be produced either by the separation of electric charges of opposite polarity or by a changing magnetic field. A magnetic field

may be produced either by an electric current (moving electric charges) or by a changing electric field. Electric and magnetic fields can exist either in empty space or in material media.

The speed of travel of these waves through free space is designated by the symbol **c**:

$$c = 2.998 \times 10^8 \text{ meters/second} \qquad (1\text{-}1)$$

When a ray of electromagnetic radiation travels through a homogenous, isotropic material medium, its speed, v, is reduced

$$v = c / n \qquad (1\text{-}2)$$

where **n** is the *index of refraction* of the medium, a numerical constant greater than or equal to one.

Because **n** > 1 in any medium other than empty space, a ray of light obliquely crossing the interface between free space and a material medium (like a lens) will always be changed in direction, or *refracted*. The same will occur with a ray obliquely crosses the interface between two media of different refractive indices. The *angle of incidence*, *Θ*, between the ray and a line perpendicular to the interface *will always be greater in the medium of lower index*. **Figure 1-2** shows a ray crossing such an interface.

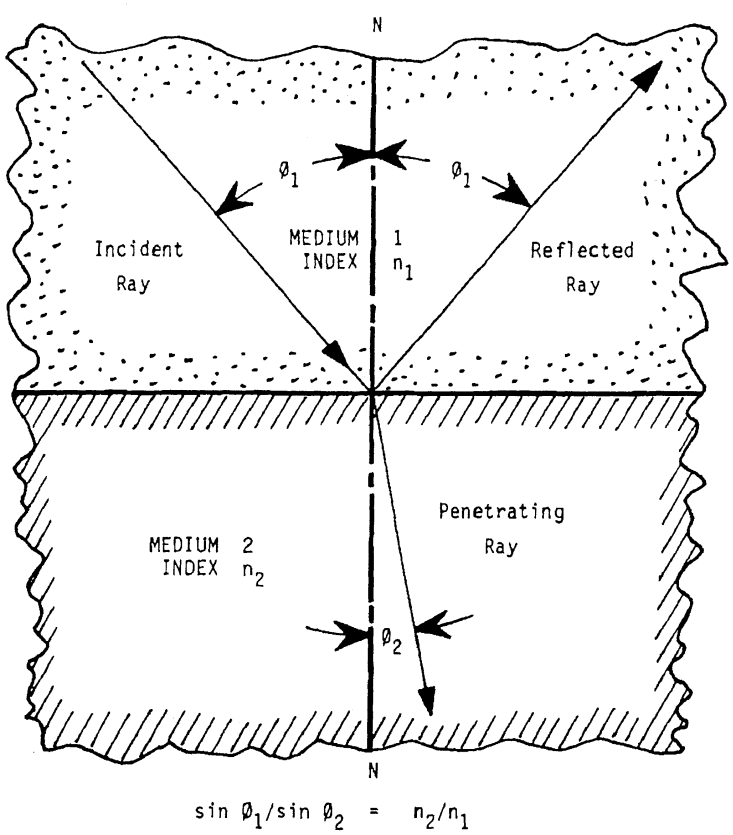


Figure 1-2.
A ray of light crossing a place interface between two transparent media of different indices of refraction. Medium 1 has the lower index: $n_1 < n_2$. Note that the ray direction is closer to the normal in the medium of higher index (medium 2).

The important parameters of the wave theory of electromagnetic radiation are the *wave length*, **λ**; the *frequency*, **f**; and the *speed of propagation*, **v**. These are related by the simple equation

$$v = f \lambda \qquad (1\text{-}3)$$

When a ray of electromagnetic radiation crosses the interface between two regions having different indices of refraction, its speed of travel is changed. However, *the frequency of the wave* (the number of full cycles passing a fixed point in space in a unit of time) is *constant, and so the wavelength changes* proportionally in **Equation 1-3**.

The Photon Theory of Electromagnetic Radiation

In 1905, Max Planck modified the wave theory by postulating that the energy carried by an electromagnetic wave cannot be endlessly subdivided into ever smaller increments, but that radiant *energy consists of small, indivisible units.* Planck named such a unit a *quantum* of energy.

In modern terminology, when speaking of radiant energy, we would call it a *photon*. A photon can be thought of as a *massless particle of radiant energy,* which moves through space at the speed c in straight lines. Although it has no mass, it does have the equivalent of *momentum,* or [MASS] x [VELOCITY], and can exert a force on a material body. A photon can be considered as the equivalent of a wavetrain of finite length in space, or a *wavelet,* as shown in **Figure 1-3**. At very low radiant intensities, such as those received by an astronomical telescope aimed at a distant star, light actually does arrive in discrete quanta that can be individually detected by a photon counter.

One important concept of Max Planck's quantum theory is that *there is a definite value of energy associated with each photon.* This photonic energy is proportional to the frequency of the equivalent wavelet:

$$e_p = hf = \frac{hc}{\mathrm{a}} \qquad (1\text{-}4)$$

In **Equation 1-4**, e_p is the photonic energy, **h** is Planck's constant (**h** = 6.626 x 10^{-34} joule-second), and **f** is the frequency of the wavelet. This fundamental equation of the photon theory of light shows that photonic energy increases directly with frequency, but increases inversely with wavelength. Long-wave radiation is inherently less energetic than short-wave and vice versa.

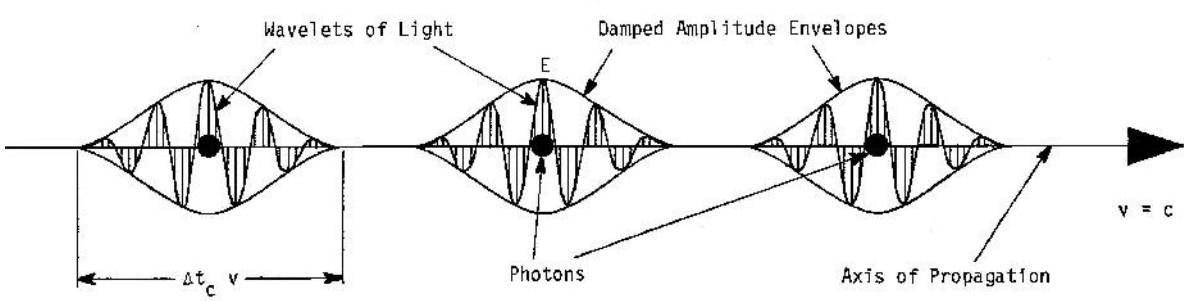


Figure 1-3.

A ray of light depicted as a stream of photons. A photon is a quantum of radiant energy, equivalent to a wavelet; a wavetrain of finite length in space. Only a few cycles of the E-wave in each wavelet are shown, for clarity. Actual wavelets would have thousands or millions of such cycles. Note that each wavelet has a damped amplitude envelope. 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, 1987:20.

Unique Properties of Laser Light

The light produced by laser has three special characteristics not found in light from any other source: (1) collimation, (2) coherence, and (3)

monochromaticity. We shall describe these properties in the following sections. Later, we shall see that they are not all equally important for surgery with lasers.

Collimation

Figure 1-9 shows four rays of light emanating from a laser (at the left-hand side) and traveling to the right at the speed c. Collimation means simply that these rays are all parallel to each other. This property of laser light makes it possible to capture all the light emitted by a laser, because it emerges in a beam of small diameter that has no divergence or convergence, unless a lens or mirror is placed in the path of the beam.

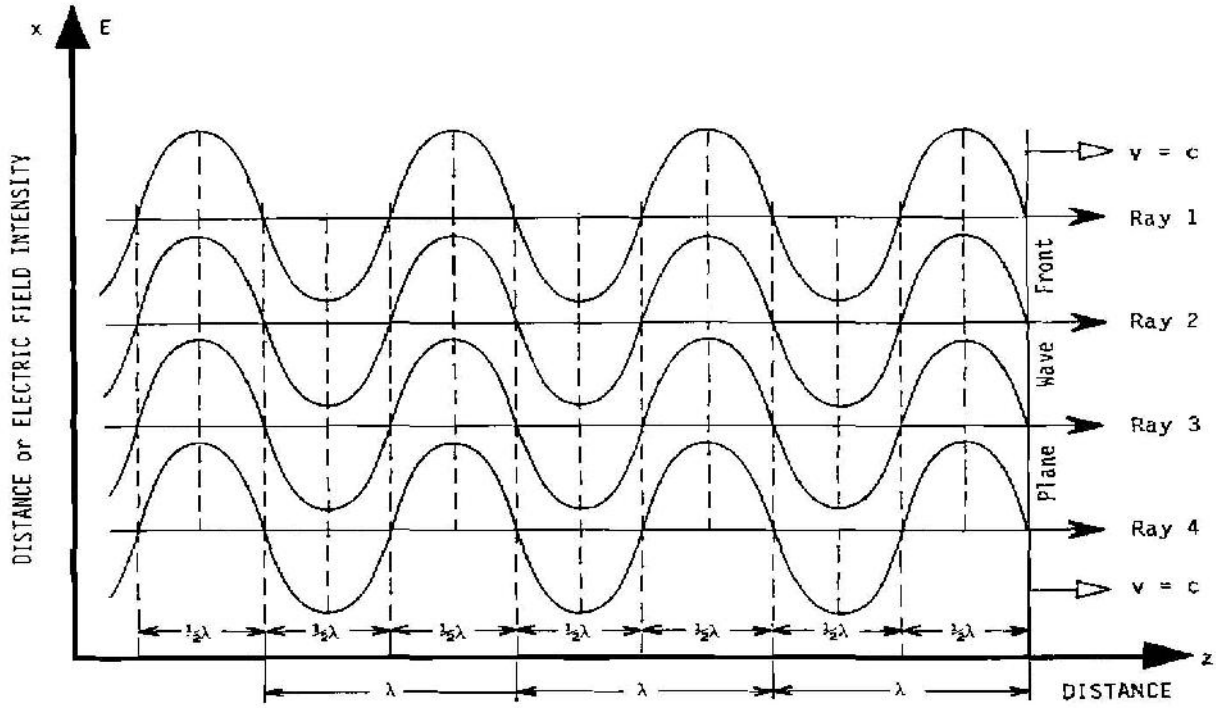


Figure1-9.
Schematic diagram showing four collimated (parallel), plane-polarized rays of light emanating from a laser on the left (not shown). Spatial coherence is evident from the coincidence of the crests and valleys of the E-waves along lines perpendicular to the axes of the rays. Temporal coherence is evident from the fact that all of the rays have the same frequency, wavelength, and speed of propagation. Monochromaticity is evident from the fact that all of the rays have the same 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, 1987:72.

Coherence

Coherence means that the E-waves of the light rays in **Figure 1-9** are in phase with each other in both space and time. Spatial coherence means that the crests and troughs of all the waves coincide along lines perpendicular to the rays. Temporal coherence means that the frequency, wavelength, and speed of travel are all constant, so that the value of electric-field intensity at any point along the axis can be predicted for any future instant of time by knowing what it is now at some other point.

Monochromaticity

Monochromaticity means that the light rays shown in **Figure 1-9** have just one wavelength, which is constant. In the light of actual lasers there is always some small spread of wavelength, as previously discussed, but this is so small in most lasers that it is less than 0.007% of the central wavelength. Gas lasers, like the carbon-dioxide and the helium-neon, have the smallest spread in wavelength, because the energy levels of atoms or molecules in gases are sharp lines, not broadened by the proximity of other individuals, except at high pressures. The wavelength spread of such lasers results from the limited time required by an individual to make the downward energy transition that causes emission of laser light. Only a transition occurring over a very long time (continuously) would produce a wavelet of light having just one wavelength. However, a typical transition time is in the order of $1x10^{-8}$ second, and the corresponding bandwidth of the light from a CO_2 laser is only 0.0375 nm. Lasers provide the highest spectral purity of any known light sources.

Temporal Operating Modes of Lasers

If a laser delivers radiation continuously, it is said to operate in the *continuous-wave* mode. Most lasers are capable of continuous-wave (c.w.) operation. However, some, like the ruby and neodymium:glass lasers, can be operated only in a *pulsed* mode. In the ruby laser, c.w. operation is prevented by the problems of creating a continuous population inversion. In the Nd:glass laser, it is prohibited by the low thermal conductivity of glass. In surgery with lasers, there are situations which require that the laser light be delivered in pulsed fashion. Several means are available for achieving pulsed output from a c.w. laser. These are called *mode locking, Q-switching, cavity-dumping, and pump-pulsing*. It is possible also to produce intermittent output from a laser by cyclically opening and closing the shutter that is provided on all medical lasers to cut off the beam when it is not in use. The first three techniques can produce very short pulses, from picoseconds (1 ps = 1×10^{-12} second) to a microsecond (1×10^{-6} second). Pump-pulsing can produce output pulses ranging from one microsecond to a large fraction of a second. Cyclic actuation of the shutter can produce pulses from about 10 milliseconds (1 millisecond = 1×10^{-3} second) to a half-second or more.

Mode-locking, Q-switching, cavity-dumping, and pump-pulsing can produce pulses whose peak power is much higher than the average power available from the same laser when it is operated in the continuous-wave mode.

Mode-Locking

Mode-Locking is a method of clipping the avalanche of wavelets, reflected back and forth between the laser's mirrors, in synchronism with the reciprocating travel of these wavelets in the optical cavity, so that only those wavelets whose intensity is above a certain threshold are transmitted. It produces laser output in pulses of picoseconds' duration closely spaced in time under an exponential amplitude envelope of nanoseconds' duration. The highest pulses of the train reach many millions of watts in peak power, although the energy per pulse is only a few millijoules. These pulses of laser light have very high spectral purity.

Q- Switching

This is a technique of cyclically or intermittently spoiling the resonance of the optical cavity by some electro-optical switching device, while a large population inversion is maintained by strong pumping. While the spoiler holds the cavity in a non-resonant condition, the laser produces no output. However, when the spoiler allows a resonance suddenly to develop, a short, powerful burst of light emerges from the laser through the partially transmitting mirror.

Cavity-Dumping

As the name implies, this method creates a large population inversion and a condition of strong resonance in the optical cavity, but does not allow any of the coherent light to escape from the resonator except when an electro-optic switch is activated. This light then emerges from the laser in a pulse of short duration and high intensity.

Pump-Pulsing

As the name suggests, this is a method of cyclically or intermittently interrupting the flow of power from the pumping source into the laser resonator, by a mechanical, electric, electronic, or

electro-optical switching device, according to the form of energy used from pumping the active laser medium. It produces output pulses that range from 10 to 100 times as high as the maximum c.w. power obtainable from the same laser. This kind of pulsing is the most commonly used in surgical lasers

Table 1-1 shows the range of pulse-durations achievable by each of the foregoing means of producing pulsed output from lasers that can also operate in the continuous-wave temporal mode.

Table 1–1

Temporal Operating Modes*

← Pulsed →			Continuous-Wave	
Short	***Medium***	***Long***	***Bursts***	***Steady***
1×10^{-12} to 1×10^{-7} second	1×10^{-7} to 1×10^{-2} second	1×10^{-2} to 1×0.5 second	0.5 to 5 seconds	5 to ∞ seconds
Mode-locking Q-switching Cavity-dumping	Q-switching Pump-pulsing	Pump-pulsing Auto shutter timing	Operator control by shutter and footswitch	

*The range of pulse-durations achievable by the various commonly used means of pulsing lasers.

Reprinted from Fisher JC. Basic laser physics and interaction of laser light with soft tissue. In: Shapshay SM. Endoscopic laser surgery handbook. New York: Marcel Dekker, 1987: 62.

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Excerpts from

CHAPTER TWO

Surgical Delivery Systems

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Maged Rizkallah, M.D.

Peter Vitruk, Ph.D.

Introduction to Delivery Systems

The size and weight of typical surgical laser systems are such that the laser cannot be held in the surgeon's hand like a scalpel. It is relatively immobile. Therefore some flexible, lightweight device must be provided to transmit the radiant power from the laser to the surgical target.

The transmitting device must be capable of carrying up to 150 W of continous-wave radiant power, or as much as millions of watts from some pulsed lasers. It must be relatively efficient, so as not to attenuate the laser beam too severely. Finally, it must not grossly distort the geometry of the laser beam. Unfortunately, not all wavelengths of laser light can be transmitted efficiently through the most flexible and convenient device of all: *a slender quartz optical fiber*. Wavelengths in the far ultraviolet (100 to 300 nanometers) and in the mid-to-far infrared (2500 to 20,000 nm) ranges of the spectrum must be transmitted via a series of mirrors, or else by direct line of sight, from laser to target.

Finally, the transmitting system must usually be terminated in a device that focuses the beam to a suitable diameter, within which the power density is adequate for the intended surgical purpose. This terminating device can be a detachable handpiece containing a lens or system of lenses, a detachable tissue-contacting probe that focuses the beam, or a properly contoured distal end of the optical fiber itself that is put in contact with the tissue. When a laser is used in conjunction with a surgical microscope, colposcope, or ophthalmoscope, the transmitting system may be terminated in a device called a *micromanipulator*, which allows the surgeon to steer the beam and to choose both the focal length and the focal diameter of the directed beam.

Practical Delivery Systems

Optical Fibers

Technology of Optical Fibers

For those lasers whose wavelengths lie in the range from 300 to 2100 nm, the delivery system used in surgery to the virtual exclusion of all others is the quartz optical fiber. This is a slender monofilament of crystalline silicon dioxide, ranging from 4 to 6 meters in length, which is coated with an adherent, thin layer of another material, called the *cladding*, having a lower index of refraction than that of the quartz core. Fibers intended for freehand use in general surgery also have a loose-fitting outer jacket, or sheath, with a small annular space between it and the cladding of the fiber to allow the transmission of gas or liquid for cooling the fiber and its terminating devices. Both the cladding and the jacket may be made of suitable polymeric materials. Fibers designed for use inside obstructed arteries will usually have the sheath omitted, in order to minimize the overall outside diameter.

One of the best material combinations for surgical fibers is a core of high-purity quartz and a cladding of the co-polymer tetrafluoroethylene-hexafluoropropylene known by the trade name Teflon FEP. This co-polymer has the lowest index of refraction (in the order of 1.35) of any readily available substance that can be used as a cladding.

Many scientists are working on developing more delivery systems. For example, a team of scientists led by John Badding, a professor of chemistry at Penn State University, has developed the very first optical fiber made with a core of zinc selenide, a light-yellow compound that can be used as a semiconductor. The new class of optical fiber, which allows for a more effective and liberal manipulation of light, may well open the door to more versatile laser technology, which could lead to improved surgical and medical lasers.

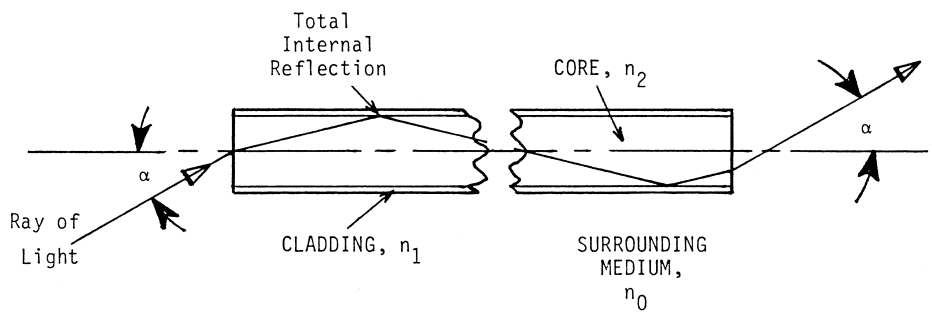

Figure 2-1.
Schematic diagram of an optical fiber, showing core and cladding. Rays of light entering the proximal end within the acceptance angle (*a*) will be totally reflected internally at each incidence on the core-cladding interface. In any plane containing the axis of the fiber (any diametric plane), the angle of incidence of the light ray on the core-cladding interface (i.e., the angle between the ray and the radius to the point of incidence) must always be greater than the *critical* angle if total internal reflection is to occur. This critical angle is given by the formula $\sin \Theta_c = n_1 \, I \, n_2$. This is always true for rays entering a *straight* fiber within the acceptance angle. However, when a ray entering exactly at the acceptance angle strikes the core-cladding interface within a *bend* in the fiber, the angle of incidence will be less than critical, and some of the intensity of that ray will be lost by transmission through the interface. Thus, the outermost rays of the cone of light entering the proximal end of the fiber wilt be attenuated by bends along the length of the fiber. Note that $n_0 < n_1 < n_2$.

Figure 2-1 shows schematically a longitudinal axial-plane section of a cylindrical optical fiber with a thin cladding closely bonded to the core. The diameter of the core in surgical fibers will be between 0.1 millimeter (mm) and 0.8 mm, and the radial thickness of the cladding will be a small fraction of the core diameter. If the indices of refraction of the surrounding medium, the cladding, and the core are n_0, n_1, and n_2, respectively ($n_0 < n_1 < n_2$), then all the rays of a conically converging beam of laser light focused at the center of the proximal end-face of the fiber will be *totally reflected internally* each time a ray strikes the interface between the core and cladding, provided that the half-angle of convergence of the conical entering beam is equal to or less than ***a***, the *acceptance angle* of the fiber, defined by…

(2-1) $$\sin a = (n_2^2 - n_1^2)^{1/2} / n_0$$

A ray of light striking the core-cladding interface will be totally reflected at every such impingement, and finally emerge from the distal end with an angle of departure equal to the proximal-end angle of incidence, so long as that angle does not exceed ***a***, as given by **Equation (2-1)**, and provided that the core is a perfect cylinder, the cladding is in intimate contact with the core at all points, and the fiber is straight over its entire length.

If the fiber has several bends of short radius, however, it is evident that the impingement of the outermost rays of the conical entering beam on the core-cladding interface within these bends will be more nearly perpendicular to that interface than it is for the inner rays, and so the outer rays will suffer some attenuation because of partial transmission through the cladding at the bends. Even with a perfectly straight fiber, there is always some scattering of the rays within the core, and these scattered rays may impinge on the core-cladding interface at angles such that the reflection is less than total. The internally scattered light escaping externally from the cladding of the fiber can be seen clearly with the eye for visible wavelengths. Even if scattering were absent, there would be some leakage across the core-cladding interface because of irregularities in the geometry of the outer surface of the core,

and imperfect contact (gaps) between the cladding and the core.

Because the indices of refraction of the core and cladding decrease with increasing wavelength, neither the acceptance angle, ***a***, nor the critical angle of incidence of rays at the core-cladding interface is constant, but varies with wavelength. Hence, the overall transmittance of a clad fiber will change with wavelength.

Another cause of attenuation in quartz optical fibers is absorption of the light by the material of the core. This is also a function of wavelength. For quartz, it is high in the far-ultraviolet, moderate in the visible and near-infared, and high again in the mid- and far-infared.

All of the foregoing factors contribute to attenuation of the transmitted laser beam. In general, the rays which enter the proximal end of the fiber at angles of incidence near the acceptance angle will be more severely attenuated than those having small angles of incidence.

Throughout the range from 300 nm to 1200 nm, the transmittance of modern quartz surgical fibers is in the range of 50% to 80% in lengths of a few meters.

For wavelengths shorter than 300 nm and longer than 2200 nm, quartz fibers, even with air cladding (bare fibers), have unacceptable high attenuation of the transmitted light. Because of the precise surgery that can be performed with the carbon dioxide laser (wavelength: 10,600 nm), attempts have been made in various countries to develop a suitable optical fiber to transmit this wavelength. Until recently, attempts to produce commercially available fibers having the required parameters of small outside diameter, acceptably low attenuation, small bend radius, long flex life, and low toxicity to living tissues have not met with success. However in 2007, OmniGuide, Inc. (www.omni-guide.com) announced the commercial availability of its new OtoBeam flexible CO_2 hollow core waveguide laser fiber and intuitive handpiece product line for use in otology procedures. (Note: Omniguide "fiber" is a hollow waveguide, not a solid core fiber).

In 2009, Samuel R. Browed et al described their initial experience with a CO_2 laser fiber system in tethered spinal cord surgery. They used a flexible fiber to conduct CO_2 laser energy to perform accurate micro-neurosurgical dissection. They described the Beam-Path-Neurofiber as a hollow-core fiber with dielectric mirror lining.[2]

LuxarCare Corporation manufactures a flexible hollow waveguide fiber designed as a single anti-reflective dielectric coating over a single highly reflective metal layer inside the elongated flexible hollow tube. The metallic surface is silver and the dialectric layer is silver halide.

Optimization of the Er:YAG laser for precise incision has been tried in many medical fields. In 2010, Jörg Meister (Department of Conservative Dentistry, Periodontology and Preventive Dentistry, Medical Faculty, RWTH Aachen University, Aachen, Germany) described the first clinical application of a liquid core light guide connected to an Er:YAG laser for oral treatment of leukoplakia.

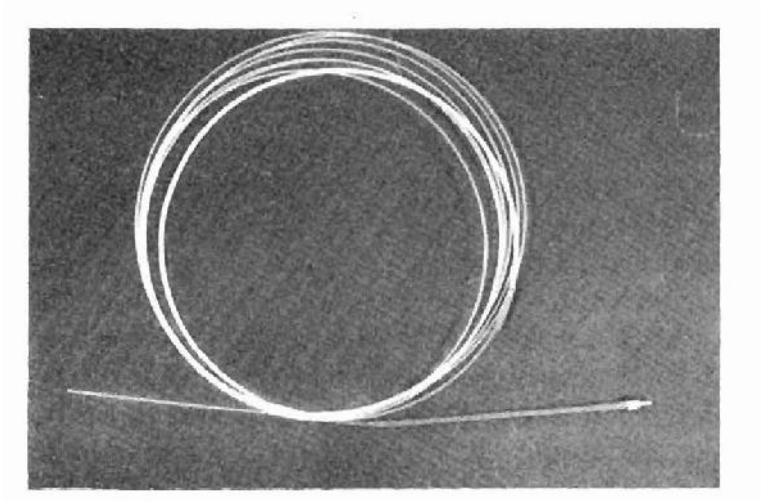

Figure 2-2.

A typical surgical optical fiber is shown. Note the special coupling at the proximal end. This is necessary to ensure proper optical alignment of the input end of the fiber with the lens system which focuses the incident laser beam. Correct coupling of the fiber to the laser is critical.

Transmission Systems Using Sequential Mirrors

A laser beam of almost any wavelength can be successfully transmitted from the exit aperture of a laser to target by means of a sequence of plane mirrors, each positioned so that it reflects the beam onto the center of the next mirror. A collimated beam is easiest to transmit in this way, because the size of the mirrors is the same for all. **Figure 2-3** shows schematically such a system. If the reflectance of each mirror is **R**, and the number of mirrors is $\mathbf{n_m}$, then the ratio of the reflected radiant power at the distal end of the mirror-sequence to that which enters the proximal end is…

(2-2) $$P_o/P_i = R^n m$$

The importance of high reflectance can be seen from **Equation (2-2)**. If the reflectance of each mirror is 0.90, and the number of mirrors is 7, then $\mathbf{P_o/P_i}$ is only 47.8%! However, if the reflectance per mirror is 0.99, the transmissive efficiency rises to 93.2%.

The mirror-sequence may be permanently fixed in position, as it is in short-pulsed Nd:YAG lasers used for ophthalmic surgery, or it may be mounted in a multiply jointed elbow-and-tube structure having an over-all flexibility comparable to that of a human arm. Such a transmissive system is called an *articulated arm*. **Figure 2-4** shows schematically the essential elements of a typical articulated arm. This assembly consists of seven rigid, metallic, 90º elbows, each with a plane mirror at its apex, set at 45º to the axes of the elbow's stubs, and two long, straight rigid tubes. The first four elbows are connected in two close-coupled pairs, and the last three are connected as a close coupled triplet. Each elbow is free to rotate a full 360º relative to the elbow just proximal to it, while always maintaining coaxial alignment of the

stubs that face one another. The proximal straight tube connects the first elbow-pair to the second, and the distal straight tube connects the second elbow-pair to the elbow triplet at the output end of the arm. The rotational freedom of each segment of the arm is indicated by a corresponding circular arrow in **Figure 2-4**.

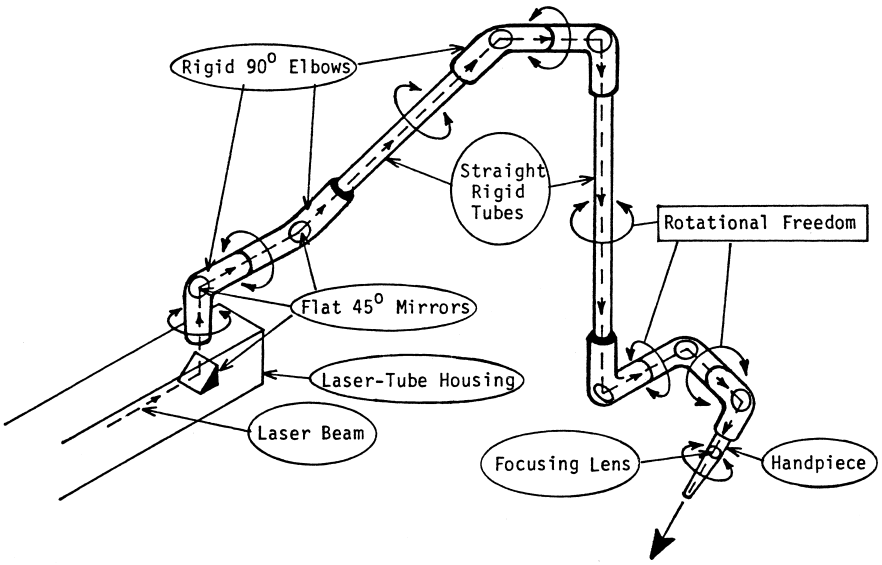

Figure 2-4.

Schematic drawing of a typical articulated arm used for transmitting the beam of a CO_2 laser. 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: 52.

An articulated arm can transmit either a collimated laser beam, or one that is slightly converging to a focus well beyond the distal end of the arm. Exact alignment of each mirror is critical to proper transmission of the beam, and the mirrors are usually provided with adjusting screws to allow alignment of the whole arm after assembly. The mirrors have enhanced reflective coatings vacuum-deposited on thick substrates of either fused silica or copper. The ball-bearings that provide rotational freedom must prevent axial or radical play of the elbows and tubular segments, yet allow rotation with minimum frictional torque. The alignment of an articulated arm is delicate. Bumping the arm against hard objects must be carefully avoided to prevent jarring the mirrors out of alignment. Routine system maintenance is necessary to ensure proper alignment of the mirrors so that the laser beam is fired properly.

At the distal end of the arm, a focusing hand-piece may be attached for free-hand surgery, the arm may be coupled to a *micromanipulator* for use with a surgical microscope, or the arm may be connected to a rigid endoscope (laparoscope, bronchoscope, arthroscope, etc.).

The major disadvantages of an articulated arm are its sensitivity to impacts with hard objects and its relatively limited flexibility as compared with that of an optical fiber. Its major advantages are high efficiency of transmission of laser beams over a broad band of wavelengths, preservation of the

coherence and TEM of the beam, and the ability to transmit radiant power up to millions of watts in pulses or hundreds of watts continuously, at safe power densities on the mirrors. The power density of the collimated beam within the articulated arm can be controlled by appropriate choice of the diameter of the beam, which is at the discretion of the designer.

Hollow Waveguides

In the latter half of the 1980s, several companies introduced hollow tubes for the transmission of light from the CO_2 laser. Such tubular waveguides may be made of either a metal-like stainless steel or aluminum with the interior surface highly polished, or of a metallic outer sheath lined with a close-fitting dielectric material. The cross-section of such waveguides is usually circular.

Several metals, notably aluminum, polished until all superficial micro-irregularities are much smaller than the wavelength of the light, have high reflectance over the sub-spectrum from mid-ultraviolet to mid-infrared for rays impinging normal to the surface. Their reflectance rises toward 100% for rays impinging on the surface at grazing incidence ($\theta \rightarrow 90^{o}$). Therefore, a slender, hollow, cylindrical tube of metal with proper dielectric coating can transmit a beam of light with an efficiency in excess of 90%, if the following conditions are fulfilled:

1. The beam is either collimated or slightly converging as it enters the tube, and the beam diameter is slightly smaller than the inside diameter of the tube;
2. The cross-section of the beam is of the same geometry as that of the tube (e.g. circular);
3. The length of the tube is 10m or less.

Such a hollow waveguide transmits a convergent laser beam by multiple grazing reflections of the rays from its inner surface. The emerging beam at the distal end will always have a small conical divergence, even if the entering beam is perfectly collimated, because of diffraction. For a convergent entering beam, the emergent beam will diverge because of the multiple glancing reflections of the outer rays within the tube. The divergence is typically between 4° and 10°.

A straight light-pipe has the highest efficiency of transmission for a given design. If it is even slightly curved, its efficiency declines sharply, because the number of internal reflections increases for each ray, and the reflectance at each impingement decreases steeply for angles of incidence slightly less than 90°. In the plane of curvature of a light-pipe, every angle of incidence is reduced by an amount that is inversely proportional to the radius of curvature. In commercially available hollow waveguides (or hollow waveguide fibers) the bending losses are controlled to less than 10% attenuation relative to straight orientation.

For a fixed geometry, a waveguide will exhibit an attenuation of laser-beam intensity that is exponential with length, but the attenuation factor is much higher than that of a true optical fiber of the same length and core diameter. Unfortunately, some of the companies offering light-pipes for sale refer to them as “fibers”. A more acceptable and widely used terminology is “hollow waveguide fiber” to reflect on both the hollow-core nature and high flexibility of such devices.

The leader in the development of hollow waveguides has been Luxar. The only company presently offering a surgical CO_2 laser with 1.0m and 1.5m-long flexible waveguide fiber in place of an articulated arm for a widest range of FDA-cleared indications is Luxarcare of Woodinville, WA. Luxar Corp patents and technology have been acquired

and improved upon by a new and different company: LuxarCare.

A Focusing Handpiece

Such a handpiece is usually available from the manufacturer of the laser in focal lengths of 75, 125, and 150 millimeters, corresponding to respective focal-spot effective diameters of about 0.17, 0.28, and 0.33 mm. A typical surgical handpiece has a single positive lens, made of zinc selenide, internally mounted near is proximal end in a cylindrical, anodized aluminum housing, connected permanently or detachably to a conic distal portion that may be made of anodized aluminum or a rigid polymer. At the distal end of this cone, a paraxial, offset tip may be fitted, so that when the end of the tip touches the target tissue, this tissue is at the focal plane of the lens. A small metal tube enters the handpiece just below the focusing lens, and a small, flexible hose connects to this tube to provide a flow of carbon dioxide gas across the distal face of the lens, for the purpose of cooling the lens and keeping it free of backstreaming tissue debris from the target. These details are shown schematically in **Figure 2-8**.

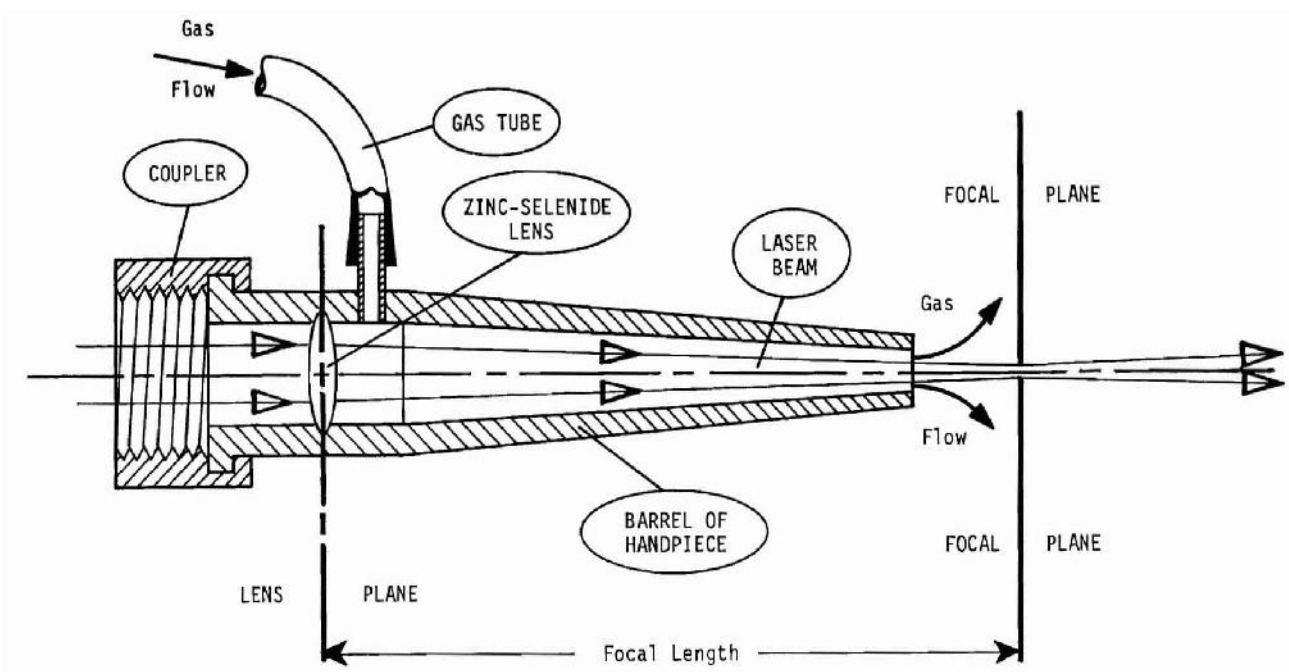

Figure 2-8.

Schematic diagram of the details of a typical surgical handpiece for a CO_2 laser. The flexible tube attached at the upper end of the handpiece carries a flow of CO_2 gas to keep tissue debris from splattering on the distal surface of the focusing lens.

As stated previously, the focusing handpiece was historically used only with the CO_2 laser , however nowadays Er:YAG and Nd:YAG lasers are also associated with the focusing handpiece. This is customarily screwed onto the distal end of the articulated arm of the

laser. Such a handpiece is usually available from the manufacturer of the laser in focal lengths from 0.5 to 12mm, and some handpieces, as those offered by Asclepion, are available with integrated smoke evacuation.

As an earlier example of a short, flexible waveguide tip for terminating an articulated arm, Luxar Corporation offered this under the trade name Flexiguide. This transmitted a maximum power of 30 W at 10,600 nm through the 0.9 mm bore with an efficiency of 70% per meter if straight, or 60% per meter if curved to a radius of 4 inches. The transmissive efficiency at 633 nm (He-Ne laser) was only 10% per meter, straight, and 5% per meter, curved. The full included angle of divergence of the emerging CO_2 laser beam was 8°. The outside diameter of the Flexiguide was 1.2 mm. It is shown in **Figure 2-12**. More recently, LuxarCare introduced the next generation of re-usable, flexible, hand-held fiber waveguide technology for CO_2 laser surgery under the LightScalpel™ brand name. This is shown in **Figure 2-13**.

Micromanipulators

As the name suggests, a micromanipulator allows the surgeon to steer the beam of a laser with a high resolution of movement while watching the surgical target through a microscope. In this context, high resolution of movement means that the minutest controllable displacement of the focal spot on the target is considerably smaller than the diameter of the focal spot. Focal-spot diameters of modern micromanipulators are adjustable, as is the focal length of the device. Since binocular surgical microscopes, such as those made by Zeiss and Wild, have focal lengths ranging from about 200 mm to about 500 mm, nearly all makers of CO_2 surgical lasers offer focal-length adjustability, either in steps or continuously, covering most of that range. The total range of focal-spot diameter available (not always in the same micromanipulator) is about 0.2 mm to 4.0 mm or so. In general, the smaller focal spots are available only at the shorter focal lengths:

A micromanipulator made by Laser Engineering, Inc. is shown in **Figure 2-15**. The distal end of the articulated arm attaches to the upper end of the micromanipulator by means of a fine-pitch threaded connector, and the unit has clamping screws to hold it to the face of the surgical microscope.

Tissue-Contacting Probes

Because the beam emerging from the distal end of an optical fiber is divergent, with an included angle between 5° and 15°, it has maximum power density right at the distal end-face. The divergence gives the surgeon a means of reducing the power density on the tissue, simply by moving the tip of the fiber away from the tissue. However, there is no means of increasing the power density at the tissue except by raising the total power of the transmitted beam. If the surgeon wishes to cut or vaporize tissue with a fiber-delivered beam, it is necessary to apply a power density above a threshold value which depends upon the absorption coefficient of the tissue at that wavelength, and upon the thermal conductivity of the tissue. In general, this threshold value is lowest for wavelengths that are strongly absorbed and for tissues that are poor conductors of heat, like epidermis and collagen. For wavelengths between 400 nm and 1200 nm, aimed at lightly pigmented soft tissue having a high-water content, the threshold of power density for vaporization can be very high. It is highest at 1064 nm, the principal wavelength of the Nd:YAG laser, because scattering is strong (σ = 5/cm ↔ 15/cm), absorption in pigments like hemoglobin and melanin is the lowest in the whole laser spectrum (a = 1.0/cm ↔ 3.5/cm), and absorption in water is weak (a = 0.2/cm). Furthermore, the reflectance of most tissues at 1064 nm is between 40% and 50%.

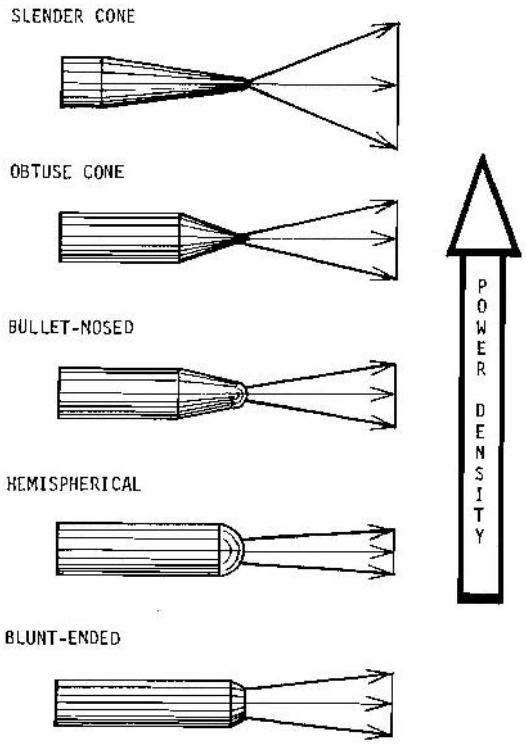

Figure 2-18.

Schematic comparison of sapphire probes of various geometries, in terms of power density and beam divergence. Note that distal power density and beam divergence are greatest for the conical tips of smallest included angle and distal·tip radius.

The sapphire stylus cuts soft tissue largely by virtue of rapid boiling of the extra- and intra- cellular water to form steam. This is the same mechanism by which the CO_2 laser vaporizes soft tissue. However, the power density required at the tip of the stylus at 1064 nm is about 4,000 times as high as would be needed in the beam of a CO_2 laser. At such high power densities (40,000 W / cm^2 and up), there is significant heating of the distal end of the sapphire stylus itself, because sapphire has appreciable absorption at 1064 nm. Temperatures of the distal tips of sapphire styli have been measured as high as 350° C with continuous-wave irradiation while in contact with soft tissue. Therefore, part of the vaporizing effect of a sapphire stylus is caused by thermal conduction of heat from the stylus to the tissue.

Sculptured Quartz Fibers

Excerpts from

CHAPTER THREE

Laser Biophysics, Tissue Interaction, Power Density and Ablative Resurfacing of Human Skin: Essential Foundations for Laser Dermatology and Cosmetic Procedures

John C. Fisher, Sc.D.

Edward M. Zimmerman, M.D.

Peter Vitruk, Ph.D.

INTRODUCTION

The biophysical analysis of facial, and more recently neck, chest and body resurfacing with lasers and nitrogen plasma, as presented in this chapter, is based upon experimental and theoretic observations garnered by the authors, both from long personal experience and from the papers published in the literature by many others. The emphasis here is on giving the reader a rational, coherent, comprehensive explanation of first-order phenomena, i.e. those that are of primary importance in understanding what will be seen clinically, or read in peer-review and lay-press publications. Higher-order effects are discussed but not emphasized, so as not to distract the candidate for certification, but also allow growth for more advanced clinicians.

Novices, as well as those more sophisticated readers, who have perused some of the growing body of articles on laser resurfacing of skin and dermatology, may find apparent discrepancies between what is presented here, and the published results of fragmentary, slice-of-life experiments conducted by investigators who are supposedly well versed in biophysical effects of laser light in living tissue. However, most of these abbreviated empirical studies do not account for all the minute details of their experiments, some of which may be of great importance in skewing the observed results so that they appear to contradict those of other studies.

If any reader of this chapter should be puzzled by apparent inconsistencies between what is presented here and the "evidence" from partial experiments reported elsewhere, he or she is welcome to email or call the Board and discuss them. The Board and its authors have gone through a long process of evolving the theoretical framework given in this chapter, and reconciling it with apparent contradictions in peer-review periodicals. The totality of this subject is complex and must deal with biophysical phenomena of higher orders up to the fourth or fifth, but that is not appropriate for a physician who must learn the fundamentals before being able to understand the nuances of the global subject. In similar fashion, one cannot deduce the laws of gravity by emptying a bag of feathers from the top of a tower in a windstorm.

This chapter was originally written by John C. Fisher, ScD, in the mid 1990's. Dr. Fisher, a capable laser physicist and medical scientist who mentored many physicians during the early use of lasers in medicine, was one of the founding members of the ABLS and its President for many years. While laser companies have come and gone and technologies have become more complex, the physics describing basic laser function are as accurate and appropriate as when Dr. Fisher originally described them. To that end, much of the original formulas have been left, gratefully, intact. This chapter, increasing in its complexity and breadth, has and will continue to be enhanced by current members of the Board, for the benefit of future members of the ABLS and their patients.

1. FUNDAMENTALS OF THE INTERACTION OF LASER LIGHT WITH LIVING TISSUE

This chapter examines in detail the biophysical phenomena which are involved in the removal of the outer cutaneous layers by lasers. It is essential for every practitioner of laser resurfacing of skin to understand these phenomena. In particular, it is critically important that everyone performing laser resurfacing recognizes the intrinsic factors which make a laser suitable or unsuitable for this purpose.

a. Fundamental Biophysical Processes by Which Laser Light Destroys Living Tissue

There are three fundamental biophysical processes by which laser light causes histologic destruction. Power density was introduced in **Chapter 1** and each of the biophysical processes has a wavelength-dependent threshold of power density below which it will not occur. The prevalence of one process over another is determined by the range of power density in the tissue.

(1) Photochemolysis

This is the rupture of inter-atomic (electronic) bonds in complex organic molecules by the photonic energy of light at all wavelengths shorter than 319 nanometers. When the intensity of such light exceeds a threshold level at which the rate of bond rupture just equals the rate of spontaneous bond repair, progressive disintegration of molecular and histologic structure occurs, with atoms, ions, and radicals being ejected from the irradiated site. This occurs at average power densities (averaged over time and area) below 1 W/cm^2. At irradiances well above the threshold, the velocities of the ejecta are high enough that the process can resemble thermal vaporization. Low-level photochemolysis is the major cause of actinic damage to the skin of persons who regularly expose themselves to the ultraviolet rays of the sun.

Unless the intensity of continuous-wave irradiation exceeds the maximum level that can be absorbed by organic molecules solely for rupture of chemical bonds, in the order of 10 watts/cm^2, photochemolysis is a non-thermal process. At higher intensities of incident radiation, it can cause heating of the irradiated substance. When radiant energy at 193 nm is delivered in short pulses, the fluence (the time-integral of power density) can go as high as 6 joules per square millimeter without thermal damage to nearby unablated tissue [5].

(2) Photothermolysis

This is the basic mechanism by which most surgical lasers destroy tissue. It is the absorption of light by target materials (chromophores) and conversion of this radiant energy into thermal energy, i.e. heat. Heat raises the histologic temperature above its normal value. If the resultant temperature is between 50° C and 100° C, the destructive effect on tissue is called photopyrolysis: thermally induced necrosis. As a very general rule, significant photopyrolysis of soft tissue (water content 75% or higher) occurs at power densities of 1 W /cm^2 to 100 W /cm^2.

When the temperature reaches 100° C, at atmospheric pressure, and the energy is kept being delivered, the intra- and extra-cellular water is rapidly boiled to form steam, which ruptures cells and destroys histologic architecture. This process is called photovaporolysis. It is the mechanism by which monopolar electrosurgical instruments cut tissue. Photovaporolic thresholds for strongly absorbed wavelengths are between 100 W /cm^2 to 1,000,000 W /cm^2. Photovaporolysis is the process by which lasers are suitable for facial resurfacing remove the outer layers of the skin.

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 the beam is absorbed can transfer the absorbed energy, converted into the form of heat, by thermal conduction and /or convection to adjacent masses of water not directly impacted by the laser beam. When the rate of radiant energy input per

unit volume of water is below the maximum possible rate of thermal-energy removal per unit volume, the water will be only warmed by the absorbed radiation, but not to the boiling temperature. At wavelengths for which there is also significant *scattering* of the light within the water, either by solutes or by suspended, unabsorbing particulate matter, the power density of the laser light within the water will be less than that of the incident beam, making *elevation of temperature* in the depths even more difficult in terms of required power density in the incident laser beam.

If the pressure on the tissue at the impact spot of the laser beam exceeds 760 torr (atmospheric pressure), the temperature of the boiling tissular water can rise above 100° C, and at high levels of irradiance the photovaporization can cause shock waves and other explosive effects, which are largely undesirable for surgery.

(3) Photoplasmolysis

This is the destruction of histologic architecture by the photonic formation of a plasma, a gas-like fourth state of matter in which there are approximately equal concentrations of free electrons and positive ions, having temperatures of several thousand degrees C. It occurs only above radiant intensities in the order of 10 billion watts/cm^2 or above. At such intensities, the electric field of the light wave is strong enough to pull outer-shell electrons out of their atomic orbits, this causing ionization and structural disintegration of any material substance.

(4) Lasers and Biophysical Processes

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. 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.

Visible and infrared lasers can produce chemolysis, but only at elevated temperatures where the inter-atomic bonds in organic compounds are ruptured by molecular vibrations and rotations.

The ultra-short-pulsed lasers used to produce photoplasmolysis (chiefly the Nd:YAG) also cause total destruction of

molecular architecture in all compounds, because of the near-total ionization of atoms throughout the material and the high temperatures attained in plasmas (>15,000°C).

b. Unique Properties of Laser Light: Definition of a Laser

The distinguishing characteristics of laser light are monochromaticity, coherence, and collimation. *Monochromaticity* is the property of having just one wavelength. Actually, no light source produces just a single wavelength, but the bandwidth of light from a surgical laser is less than 0.1 nanometer. *Coherence* is manifested in two ways: spatial and temporal. *Spatial coherence* is the alignment of the crests and troughs of the electric-field waves of the light rays in a laser beam on lines perpendicular to the rays. *Temporal coherence* is the constancy of the frequency, wavelength, and speed of propagation of the light waves. *Collimation* is the lack of divergence or convergence of the rays of light in a laser beam. They are all parallel to one another in the primary beam emerging from the laser and continue on in that fashion.

For the purposes of this discussion, a laser may be defined as a source of radiant energy having these unique properties. There are hundreds of physical materials that can be used to produce laser light, including gases, liquids, and solids. The physical details of how **L**ight **A**mplification by **S**timulated **E**mission of **R**adiation, or LASER action, occurs within a laser are not critically important to the cosmetic surgeon. *What is vitally important is an understanding of how laser light interacts with living tissue.*

c. Basic Optical Phenomena of Laser Light in Living Tissue

When a ray of laser light strikes the surface of living tissue, four fundamental optical phenomena occur. These can be quantified in terms of the intensity per unit area (power density) at various points along a single ray of light, as it passes from the air above into the depths of the tissue. They are:

1) Reflection and backscattering of the beam by the surface of first incidence
2) Transmission into, or through, the tissue
3) Scattering within, and perhaps out of, the tissue
4) Absorption by the tissue between scattering points

Reflection is measured in terms of reflectance: the ratio of the intensity of the reflected fraction of a ray of light to the intensity of the incident ray of light. Reflectance is independent of wavelength and tissue color for wavelengths shorter than 300 nm and longer than 4,000 nm. Between these limits, it is dependent upon both wavelength and tissue pigmentation.

Figure 3-1 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. 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 2,200 to 40,000 nm, reflectance is "colorblind"

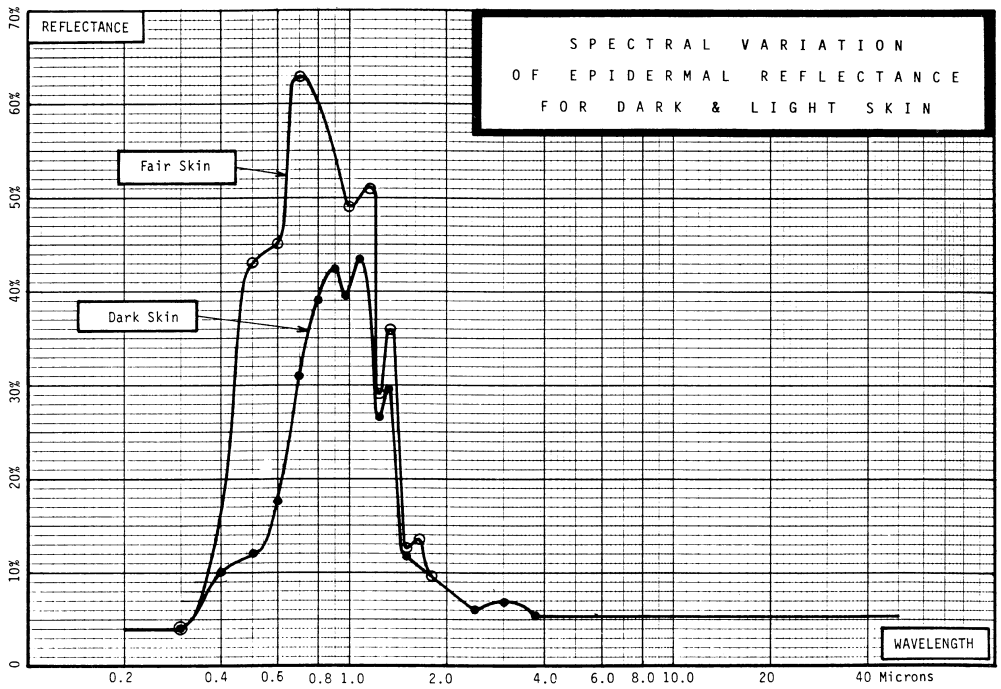

Figure 3-1.

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.

Similarly, ***transmission*** is measured in terms of transmittance: the ratio of the intensity of the transmitted ray as it emerges distally from the tissue to that of the same ray just after entering the tissue. ***Scattering*** is actually a composite of several distinct optical phenomena, but for the purposes of laser surgery, it is defined as a change in the direction of a ray of light without a change in its wavelength. ***Absorption*** is defined as the conversion, within tissue, of radiant energy into other forms, such as heat.

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 3-2** shows, schematically, a ray of laser light being partially reflected from the surface of first incidence on a mass of tissue.

Attenuation is a process of diminishing the intensity of laser light as it travels deeper into a medium that does not totally reflect the radiation at its first surface. In particular, we are interested in the attenuation in living tissue.

Figure 3-2 also shows schematically the attenuation (diminution of intensity) that occurs as

a ray of laser light penetrates into living tissue. Both absorption and scattering contribute to the process of attenuation. In a homogeneous, isotropic medium, such as hydrated gelatin, the attenuation is exponential: the ray loses a constant fraction of its intensity in the direction of propagation in every unit distance of forward travel. In living tissue, which is neither homogeneous nor isotropic, the attenuating process can be described approximately as exponential:

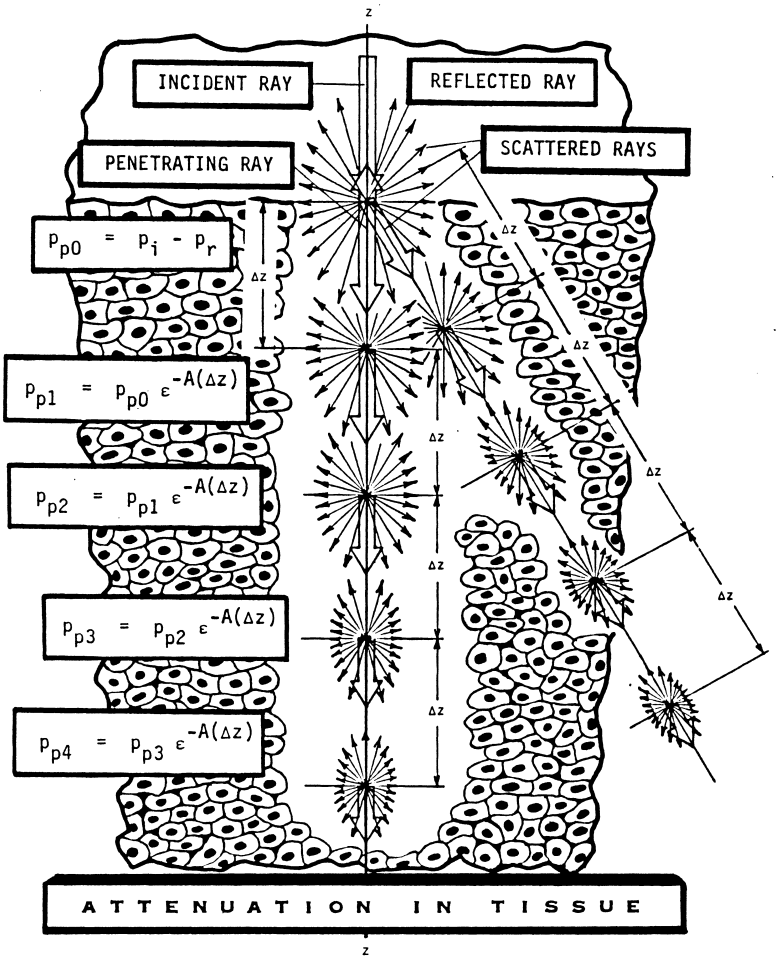

Figure 3-2.

Schematic diagram of the attenuation of a ray of laser 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 within each unit distance, **Δz**, of forward travel. The porcupine figures depict omni-directional scattering.

Reprinted from Fisher, J.C., Qualitative and quantitative tissular effects of light from important surgical lasers: optimal surgical principles. In: Wright, VC. and Fisher, JC., eds. Laser surgery in gynecology: a clinical guide. Philadelphia: W. B. Saunders, 1993: 65.

d. Suitability of a Laser for a Particular Surgical Application

As stated previously, suitability is determined for thermolytic laser types by the absolute and relative magnitudes of the absorption and scattering coefficients, **a** and **s** (as in **Equation 3-2**). The choice of such a laser for a specific surgical purpose may be influenced by secondary factors, such as transmissibility of its beam via optical fiber, hollow mirrored wave guide or articulated arm, the spot size(s) generated, maximum energy available, and the size and cost of the laser. **However, if the choice is made objectively and scientifically, only the coefficients of absorption and scattering are important.**

In the most general sense, the choice of laser type should be made first on the basis of the preferred mode of tissue destruction: photochemolysis, photothermolysis, or photoplasmolysis. However, in the use of lasers for resurfacing of skin, photothermolysis is currently the preferred process. Therefore, selection of the laser type in this case is based upon the magnitudes of **a**, **s**, and the ratio of **a/s**.

Using these factors, all types of surgical lasers can be assigned to one of three categories:

WYSIWYG, the acronym for **W**hat **Y**ou **S**ee **I**s **W**hat **Y**ou **G**et;

SYCUTE, for **S**ometimes **Y**ou **C**an **U**se **T**hem **E**ffectively; and

WYDSCHY, for **W**hat **Y**ou **D**on't **S**ee **C**an **H**urt **Y**ou.

These categories are defined as follows:

WYSIWYG: $a > 100/cm$; $a/s > 10$

SYCUTE: $1 < a < 100/cm$; $0.1 < a/s < 10$

WYDSCHY: $a < 1.0/cm$; $a/s < 0.1$

WYSIWYG lasers are suitable for precise surgery with minimum thermal damage to adjacent tissue. They are generally fair to poor coagulators. Examples are CO_2 at 10,600 nm, Holmium:YAG at 2,100 nm and erbium:YAG at 2,940 nm, and the argon-fluoride (excimer) at 193 nm.

SYCUTE lasers are useful for color-selective thermolytic destruction of pigmented tissue. The wavelength must be chosen for strong absorption in the pigment of the target tissue (chromophores). These lasers have wavelengths in the visible and near-infrared regions of the electromagnetic spectrum. Examples are KTP, pulsed dye, ruby, alexandrite and diode lasers.

WYDSCHY lasers are well suited to causing thermal necrosis for coagulation of bleeding vessels or destruction of malignant tumors. They are useless for precise cutting or

ablation with minimum thermal damage to nearby tissue. The outstanding example of this type is the continuous-wave Neodymium:YAG at 1,064 nm. All of these lasers have wavelengths in the near-infrared part of the spectrum. Their rays are strongly scattered and weakly absorbed in most tissues, unless free carbon from prolonged thermal necrosis is present. Carbon strongly absorbs all wavelengths, and causes any thermolytic laser to cut like a WYSIWYG, but not without thermal damage, which has already occurred by the time that free carbon is present during laser irradiation of living tissue.

e. Absorption & Scattering Coefficients for Various Constituents of Tissue

(1) Absorption: There are several major absorbers of light in living tissue, among the more important of which are:

1. Water, which constitutes from 75% to 85% of soft tissue;
2. Pigments, such as bilirubin, melanin, hemoglobin, and xanthophyll, especially important at visible wavelengths;
3. Fat and lipids, especially at ultraviolet and mid- to far-infrared wavelengths;
4. Other complex organic molecules, especially at ultraviolet and mid-to-far-infrared wavelengths;
5. Carbon, an abundant constituent of all living tissue, which is an end-stage breakdown product of pyrolysis, and is a strong absorber of light *at all wavelengths.*

As said previously, the major constituent of living tissue, in both plants and animals, is water. It is also a very strong absorber of light at wavelengths greater than 2,500 nm: a > 100/cm. In pure water or normal saline, scattering is negligible by comparison with absorption in this spectral range. However, when water contains even a small fraction of particulate matter, it becomes a scattering medium. Blood is a good example. **Figure 3-4** shows the spectral variation of absorption coefficient for water, Normal saline in the human body contains only 0.9% sodium chloride, but its absorption coefficient is not significantly different from that of water over the spectrum from 200 to 10,000 nm. Note that the absorption coefficient for water varies through at least 8 orders of magnitude (factors of 10) from ultraviolet through visible to far-infrared wavelengths.

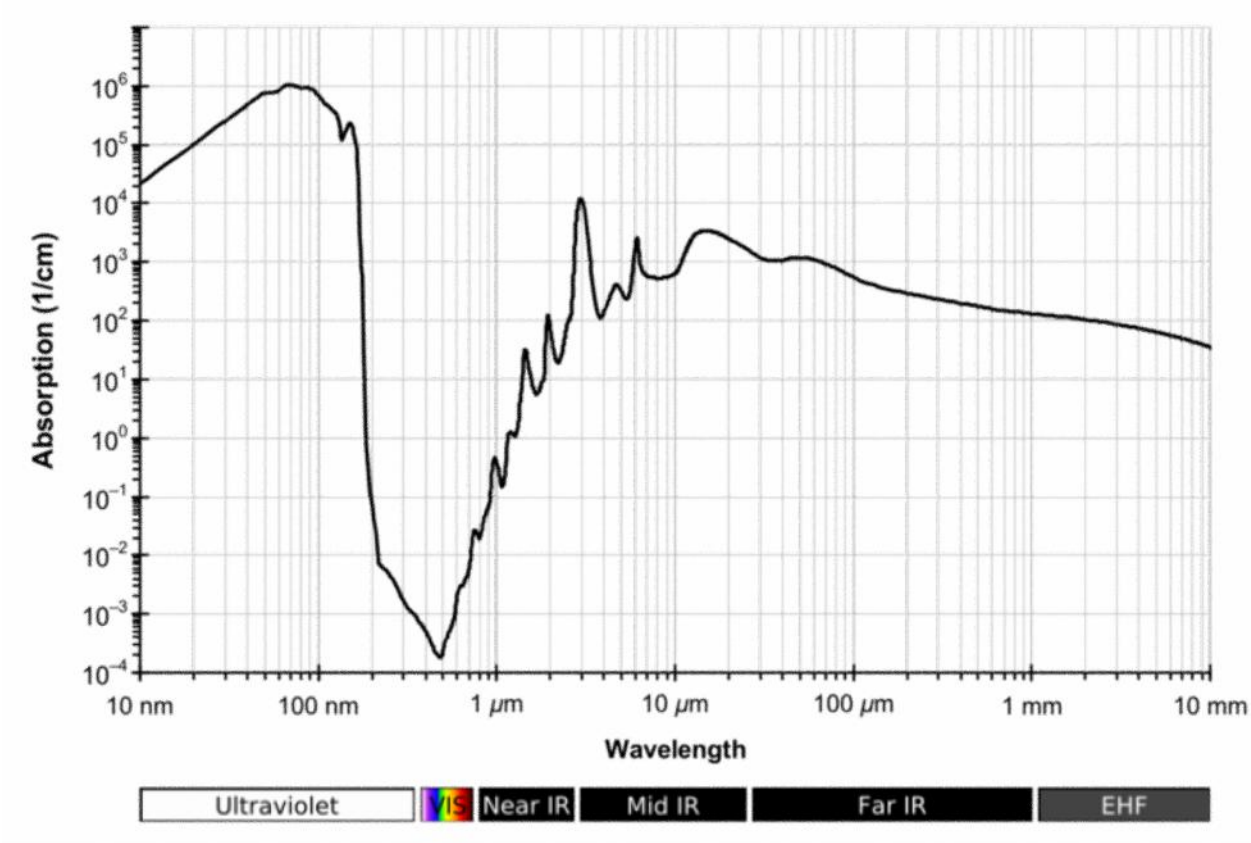

Figure 3-4.

Spectral variation of absorption coefficient for water. Note that the vertical axis shows variation over at least 8 orders of magnitude. Physiologic saline is a major absorber of radiation in living tissue from 2 to 11 micrometers. The absorption coefficient of water is not markedly different from that of normal saline.

Source: Absorption of electromagnetic radiation by water, http://en.wikipedia.org/wiki/Electromagnetic_absorption_by_water, Wikipedia, 2012.

Water is a very beneficial absorber for laser light in the human body, because it boils at a constant temperature dependent only upon the pressure at its surface. That temperature is 100° C when the pressure is 760 torr. The basic process by which a thermolytic laser ablates tissue is flash boiling of histologic water to form expanding steam. While that water is boiling at constant pressure, the impact surface of the laser beam on the tissue is isothermal. Therefore the temperatures at points within the adjacent tissue remain at or below the boiling temperature of the water, irrespective of the power density of the laser beam as long as liquid water is present in the tissue.

There are numerous other absorbers, often called chromophores, in living tissue that absorb light at various wavelengths. Notable examples are pigments, such as melanin, hemoglobin, xanthophyll, and bilirubin. At wavelengths shorter than 319 nanometers, complex organic molecules of many varieties are significant absorbers: collagen, fat, proteins and carbohydrates are examples.

At wavelengths where pigments are the major absorbers, however, and water is relatively transparent, the absorbing chromophores must transmit their heat to the aqueous histologic matrix by thermal conduction, which requires a temperature difference between the absorbing particles and the surrounding liquid. **Therefore, even though the water still boils at a constant temperature, the absorbers must be higher in temperature than 100° C.**

The spatial distribution, as well as the concentration of the absorbers, or chromophores, play important role in how the absorbed laser light impacts the photo-thermal laser-tissue interaction. While the concentration of water varies more or less smoothly throughout most of the soft tissue, the hemoglobin distribution is limited to the whole blood inside the blood vessels, and the melanin in the skin is confined to the melanosomes in the epidermis. These most significant absorbers in the soft tissue in the Visible and Infrared spectral ranges, i.e. water, melanin, hemoglobin and oxy-hemoglobin, were depicted in **Figures 3-3** and **3-4** at their respective maxima:

- liquid water, i.e. water at 100%;
- Hb in whole blood, i.e. at 150g/L;
- HbO2 in whole blood, i.e. at 150g/L;
- Melanin at 100% of volume fraction of melanosomes in the epidermis (Jacques SL. Origins of tissue optical properties in the UVA, visible, and NIR regions. In: Alfano RR, Fujimoto JG, ed. *OSA TOPS on Advances in Optical Imaging Photon Migration*. Optical Society of America 1996;2:364–69).

However, typical water content in the soft tissue is not at 100%, which needs to be reflected in the absorption coefficient. For instance, scaling the pure water absorption by 75% mimics a typical soft tissue with 75% water content – see **Figure 3-4a**. A highly dehydrated tissue, e.g. tooth enamel with 4% water content, will exhibit a 4% scaled absorption coefficient of 100% water.

The hemoglobin (Hb) and oxyhemoglobin (HbO2) are present at their maximum concentration 150g/L only in a whole blood (either de-oxygenated or oxygenated) inside the blood vessel capillaries. Photons encounter the full strong absorption of whole blood, presented in **Figure 3-3**, only when they strike blood vessels. In other words, the local absorption properties (in **Figure 3-3**) govern light-tissue interactions. However, the average Hb and HbO2 concentration in the soft tissue is significantly lower, because the volume fraction of blood is only a few percent in tissues. Accordingly, the average Hb and HbO2 absorption coefficient that affects light transport is relatively low, as shown in **Figure 3-4a** for 10% average blood presence in the soft tissue (assuming 5 L of whole blood in the average 70 kg human body: Alberts B, Johnson A, Lewis J, Raff M, Roberts K, Walter P. *Molecular Biology of the Cell*. 5th ed. New York, NY: Garland Science; 2007:Table 23-1).

Similarly to the hemoglobin, the melanin (inside the melanosomes in the epidermis) is a very strong absorber of light, as seen in **Figure 3-3**, and the local interaction of light with the melanin is quite strong. However, the volume fraction (f_v) of melanosomes in the epidermis varies between differently pigmented types of skin colors (or epithelium colors of gingiva in the oral cavity). Accordingly, the average melanin absorption coefficient that affects light transport is relatively low as shown in **Figure 3-4a**. In other words, the local interaction of light with the melanin is strong, but the epidermis' light transport properties are only weakly affected by melanin absorption depending on volume fraction of melanosomes. In skin, the volume fraction (f_v) of melanosomes is estimated to vary as 1-3% for light Caucasians skin, 11-16% for well-tanned Caucasian and Mediterranean skin, and 18-43% for darkly pigmented African skin (Jacques SL. Origins of tissue optical properties in the UVA, visible, and NIR regions. In: Alfano RR, Fujimoto JG, ed. *OSA TOPS on Advances in Optical Imaging Photon Migration*. Optical Society of America 1996;2:364–69). The concentration of melanin within melanosomes is quite variable, however, the melanosome absorption spectrum for skin is well approximated as (f_v x 1.70 x 10^{12} $nm^{-3.48}$ [cm^{-1}]), and is presented in **Figure 3-4a** for volume fraction of melanosomes f_v = 2, 13, 30, and 100%, and "nm" refers to the wavelength expressed in nanometers (Jacques SL, McAuliffe DJ. The melanosome: threshold temperature for explosive vaporization and internal absorption coefficient during pulsed laser irradiation. *Photochem. Photobiol.* 1991;53:769-775,. Jacques SL, Glickman RD, Schwartz JA. Internal absorption coefficient and threshold for pulsed laser disruption of melanosomes isolated from retinal pigment epithelium. *SPIE Proc* 1996; 2681:468-477. Jacques SL. Optical properties of biological tissues: a review. *Phys Med Biol.* 2013;58(11):R37-61).

In all of the foregoing figures, the extreme range of the value of **a** is from about 0.0001/ cm to about 9000/ cm, or at least 8 orders of magnitude, for wavelength varying from 180 to 11,000 nm. This huge range emphasizes the need to choose wavelength (i.e., the *type* of laser) properly for the tissue to be treated. **Table 3-2** characterizes the absorption of four components of tissue at six discrete wavelengths.

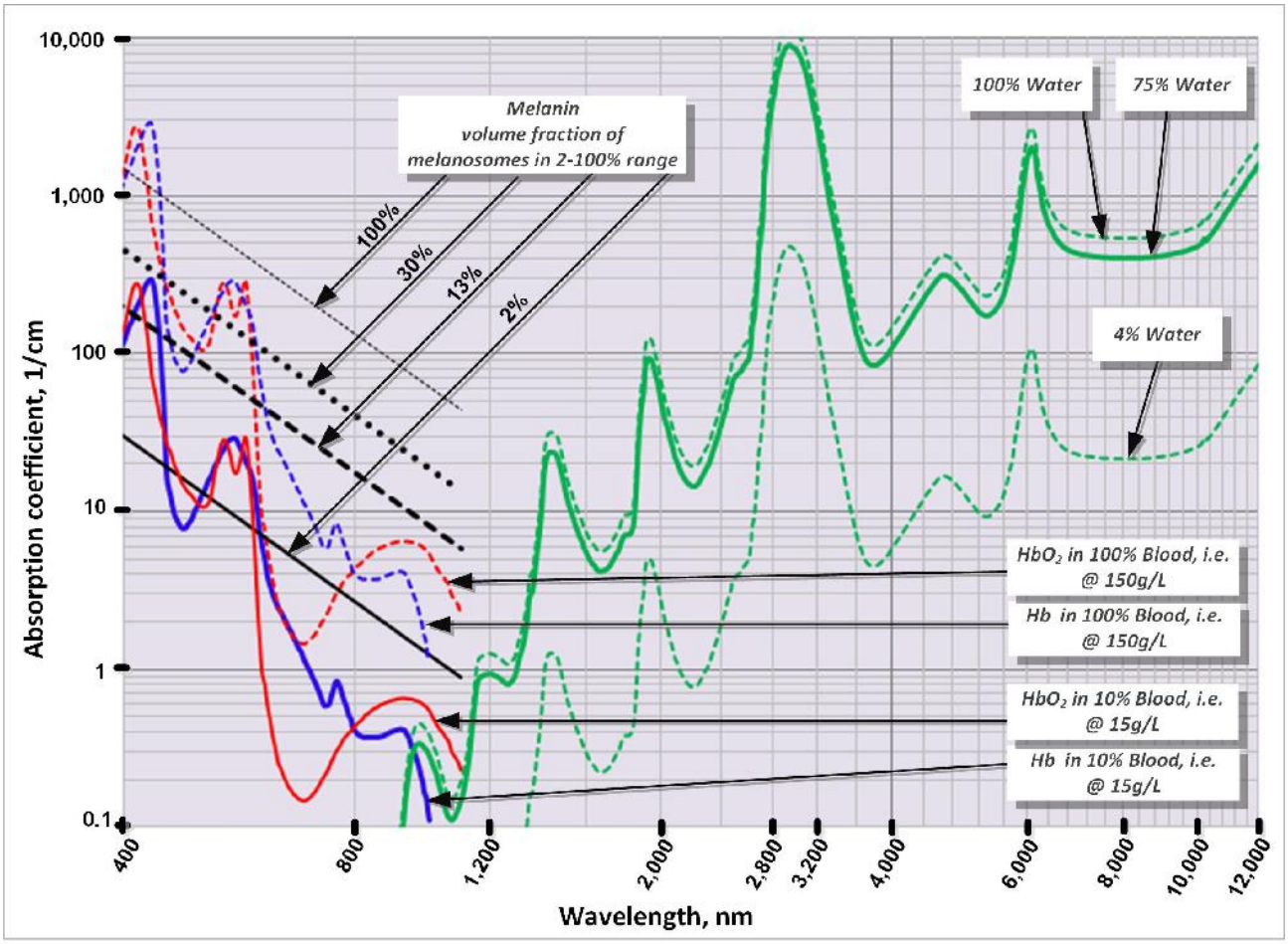

Figure 3-4a. Absorption Coefficient Spectra for: 4%, 75% and 100% Water (green solid and dotted curves); 10% and 100% Whole Blood for HbO_2 (red solid and dotted curves) and Hb (blue solid and dotted curves); 2-100% Melanin (black solid and dotted lines).

Data are derived from: Jacques SL. Origins of tissue optical properties in the UVA, visible, and NIR regions. In: Alfano RR, Fujimoto JG, ed. *OSA TOPS on Advances in Optical Imaging Photon Migration*. Optical Society of America 1996;2:364–69. Jacques SL. Optical properties of biological tissues: a review. *Phys Med Biol* 2013 Jun 7;58(11):R37-61. Fisher JC. Basic laser physics and interaction of laser light with soft tissue. In: Shapshay SM. ed. *Endoscopic laser Surgery Handbook*, New York, NY: Marcel Dekker 1987:101. Fisher JC. Qualitative and quantitative tissue effects of light from important surgical lasers. In: Wright CV, Fisher JC, ed. *Laser surgery in gynecology: a clinical guide*. Philadelphia, PA: Saunders 1993: 68. Vitruk P. Oral Soft Tissue Laser Ablative & Coagulative Efficiencies Spectra. *Implant Practice US*, 2014;7(6):22-27.

(2) Scattering: Scattering of light in living tissue is strongest at short wavelengths, and diminishes with increasing wavelength. For our purposes in surgery, we may define scattering as *a change in direction of a light ray without a change in wavelength.* Scattering, as we observe it in living tissue, is a composite of several distinct phenomena:

1. Diffuse reflection from irregular interfaces between histologic materials having different indices of refraction and physical dimensions much larger than the wavelength.
2. Refraction of light rays at interfaces between histologic materials of different indices (lens effects) and physical dimensions much larger than the wavelength.
3. Reflection and diffraction of light waves by discrete particles in the tissue, ranging in size from organic molecules to cellular inclusions.
4. Resonant absorption of light by atoms and molecules and re-emission at the same wavelength but in different directions.

Scattering by particles much smaller than the wavelength is omni-directional and is called Rayleigh scattering, after the British physicist Lord Rayleigh (1842-1919). It varies in intensity inversely with the fourth power of wavelength. Scattering by particles greater in size than the wavelength is predominantly forward and is named after the German physicist G. Mie. It varies approximately with the inverse square-root of wavelength. The coefficient of combined Rayleigh and Mie scattering in living tissue ranges from a low of about 5/cm to a high of about 50/cm for the types of tissue in the human body, over the range from 10,000 to 100 nm.

Scattering coefficients have been examined in studies of Halldorsson and Langerholc [15]; Gijsbers, Breederveld et al [16]; and van Gemert, Cheong et al [17], among others. By searching the literature, the following general facts can be gleaned:

1. Scattering coefficients, as might be expected, are highest at short wavelengths. This is so for several reasons. First, the indices of refraction of all materials, except near absorption bands, are highest for the shortest wavelengths. Second, as stated before, Rayleigh scattering increases inversely with the fourth power of the wavelength. Third, Mie scattering increases inversely with the 1/2-power of the wavelength.
2. In biologic tissue, Rayleigh scattering is usually less important than Mie scattering and diffuse reflection and refraction at histologic interfaces in changing the direction of light rays.
3. Scattering is most significant in relation to absorption in the range of wavelengths between 600 and 2200 nm. This is so because $\mathbf{s} \geq \mathbf{a}$ for most tissues in this part of the spectrum.

When scattering is much stronger than absorption in living tissue, laser light within that tissue is no longer collimated and spatially coherent, but becomes <u>randomly diffused radiant flux</u> (r.d.r.f.). This is characterized by rays of light traveling with equal probability in all directions and it is the exact antithesis of a laser beam. R.d.r.f. is useless for precise incision or vaporization of tissue, but is very effective for coagulation. It is what the pilot of an aircraft sees when flying in dense fog during daylight: it appears equally bright in every direction.

Conversely, when a laser beam enters a medium in which scattering is insignificant by comparison with absorption, then the beam remains collimated within that medium, and becomes less intense with increasing depth below the first surface. This is what occurs when WYSIWYG lasers are used for surgery.

For surgery, the most important consequence of scattering is *the spatial redistribution of radiant power density,* from what would otherwise be a narrow pencil of light, into a surrounding *volume of irradiated tissue*.

(1) Importance of Power Density: Power density is such an important operating parameter of a surgical laser that it must be understood by the surgeon in order to do laser surgery safely and effectively. The concepts of energy and power were discussed in Chapter 1, and the reader should refer to those portions of Chapter 1 for definitions of these basic entities. The ideal power-density profile of a laser beam for surgery is the Gaussian, or TEM_{oo} transverse mode, discussed in Chapter 1. This is preferred because it can be focused to the smallest effective diameter on a target.

(2) Definition: Power *density* is *defined as the radiant power transmitted per unit area of cross-section of a laser beam, or* radiant *power striking the target of the beam per unit area of target surface illuminated by* the *beam.* In the study of optics, power density is referred to as *intensity.* Power density is proportional to the square of the amplitude of the electric field of a light wave.

(5) Surgical Significance of Destruction of Destructive Thresholds of Power Density: 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. 3-6)**.

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.*

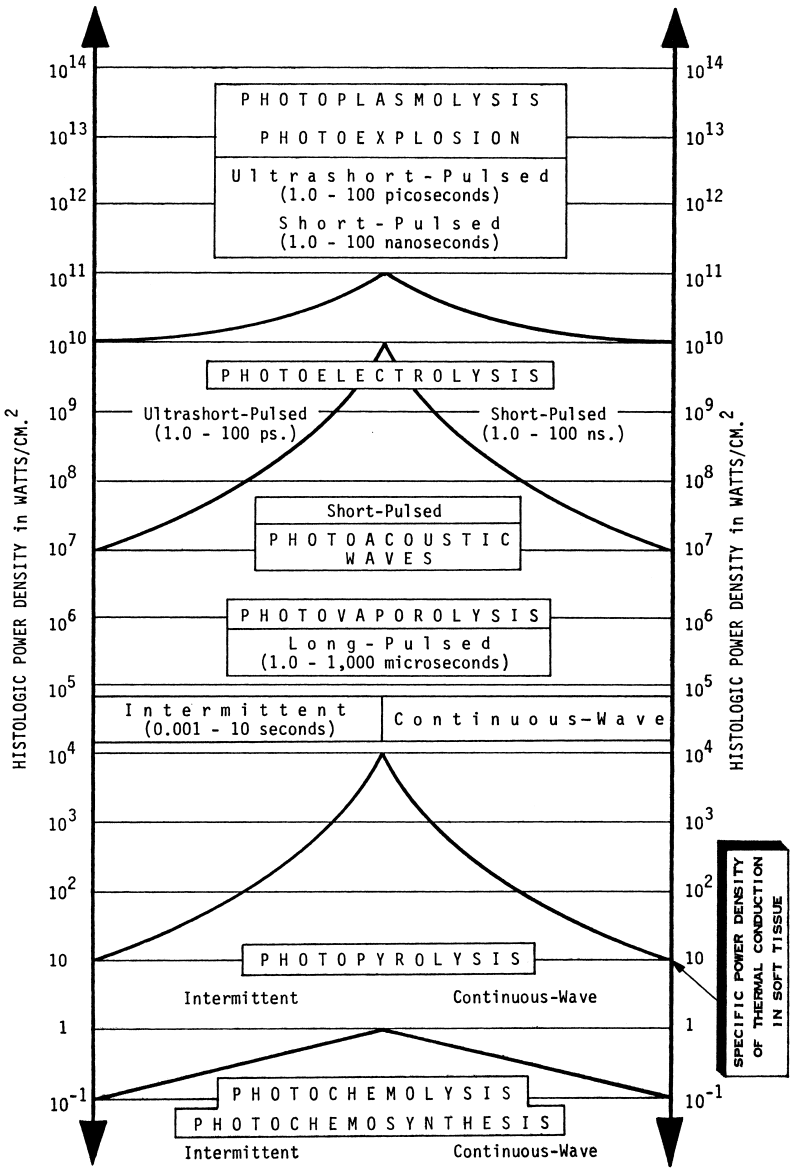

Figure 3-6.
Biologic effects of laser radiation between 100 nm and 10,600 nm as functions of average (over time and space) power density in soft tissue. The sloping boundary lines between regions of different effects denote the fact that absorption coefficient and/or photon energy vary with wavelength. 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.

(3b) Ablation/Vaporization Depth

In the case of a laser beam pulsed with a duration short enough to preclude any significant conductive loss of heat from the irradiated volume of tissue, i.e. shorter than ***TRT***, the depth of tissue ablated during each pulse will be proportional to the fluence in excess of the threshold value at each point within the boiling diameter,

(3-11) $$z_{ap} = (f_p - f_t) / h_v \quad \text{[LENGTH]}$$

where $\mathbf{z_{ap}}$ is the depth of ablation below the original surface during one pulse, $\mathbf{f_p}$ is the fluence at the end of the pulse, $\mathbf{f_t}$ is the threshold fluence, and $\mathbf{h_v}$ is the latent heat of vaporization of water, 2,260 joules/cm^3. For a gaussian TEM, the cross-section of the ablated volume in any plane passing through the axis (the z-axis) of a stationary beam is also gaussian, except for minor deviations caused by variations of water concentration in the tissue from the point to point. The cross-section of ablated tissue for a gaussian beam swept at constant speed across the tissue in a direction (i.e. the x-axis) perpendicular to the beam axis is only quasi-gaussian. The reason for this is that the x-z plane passing through the laser beam is the only one in which the variation of power density and fluence with the x-distance from the beam axis is truly gaussian. In all planes parallel to this central x-z plane, the x-variation of power density and fluence is bell-shaped, but not gaussian.

Minimization of pyrolytic damage from thermal conduction to tissue adjacent to the impact spot of a continuous-wave laser beam can be achieved by sweeping the beam rapidly across the tissue in the x-direction, perpendicular to the axis of the beam (the z-direction). This is illustrated schematically in **Figure 3-9**, which shows a three-dimensional gaussian bell of radiant power density swept at constant speed, **v**, in the x-direction across a tissue surface lying in the x-y plane. This scheme is advantageous when large fluence is desired, because it can use a laser beam of relatively low power, focused to a spot small enough to achieve high power density. Its major disadvantage is that a small spot requires a relatively long time to cover a specified area, if other factors are the same as in the case of a pulsed laser beam having a large spot diameter.

In **Figure 3-9** it should be noted that the fringes of the gaussian bell have been cut off in lateral planes parallel to the x-z plane, because these cross-hatched plane surfaces are equidistant from the x-z plane by the vaporization/ablation radius of this swept beam. The footprint of the gaussian beam, therefore, is a circle except for the two segments which are missing because the lateral fringes have been cut off. The x-dimension of this footprint is taken as $\mathbf{1.5d_e}$, where $\mathbf{d_e}$ is the effective diameter of the beam, because almost 99% of the total radiant power is transmitted within a coaxial circle having this size.

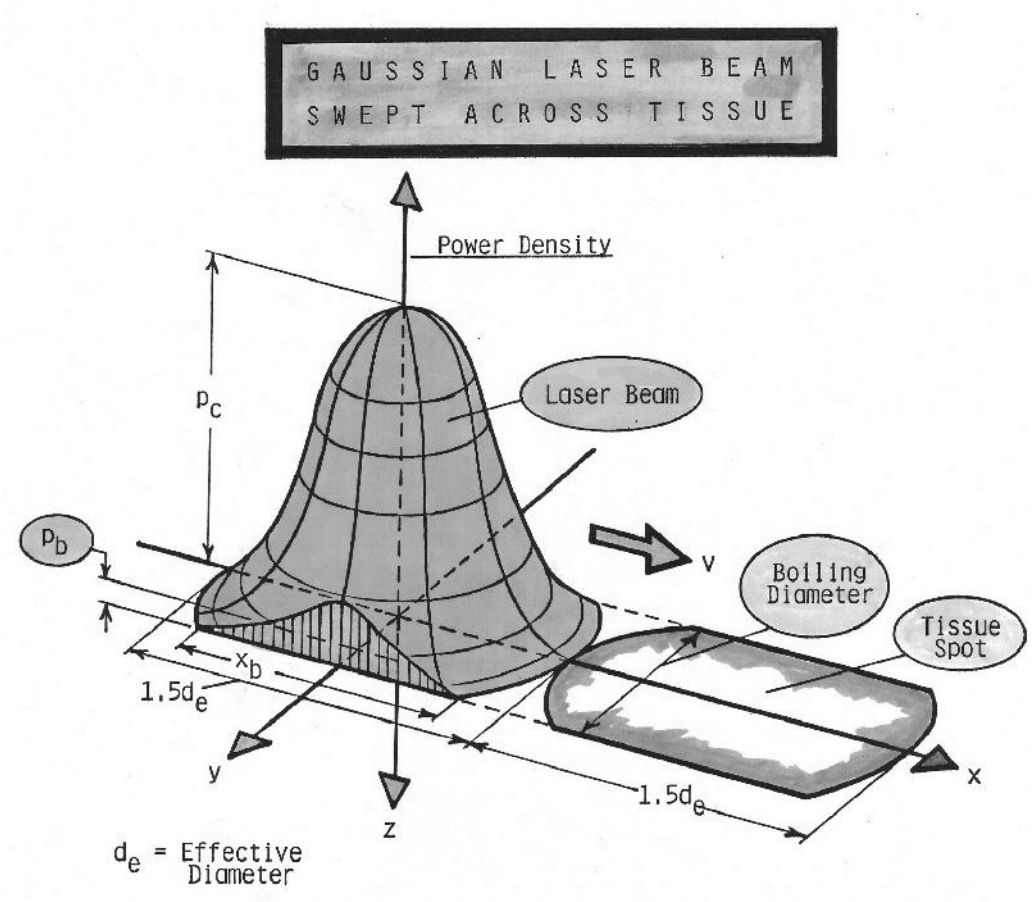

Figure 3-9.

Three-dimensional diagram of a gaussian CO_2 laser beam being swept at constant linear speed, **v**, in the x-direction across a spot on a flat tissue surface lying in the x-y plane, and having the same shape and size as the footprint of the beam. The width of the gaussian bell in the y-direction is equal to the boiling, i.e. ablation/vaporization, diameter of the beam. The length of the bell in the x-direction is 1.5 times the effective diameter, d_e, because within that span almost 99% of the total power of the beam is transmitted.

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; 4:207.

The biophysical mechanism by which a laser beam ablates soft living tissue is sudden boiling and vaporization of histologic water to form steam, which expands rapidly, rupturing individual cells, tearing contiguous cells apart at their interstices, and ripping connective tissue apart. The solid residues of cells and connective tissue are dehydrated, and ejected from the impact zone of the laser beam with velocities up to several meters per second. The cumulative effect on tissue structure is the same as if each cell were implanted with a small explosive charge that is ignited by absorption of the laser light. Without the histologic water there would be no ablation (Latin, meaning "carrying away"), only burning of tissue.

2. SUPERPULSING AND ULTRAPULSING

3. COMPUTERIZED PATTERN GENERATORS

Coherent, Inc. was the first to introduce a computerized pattern generator, trade-named the CPG®, using a galvanometric, collimated-beam-deflecting hand piece, the UltraScan®, to place a 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.

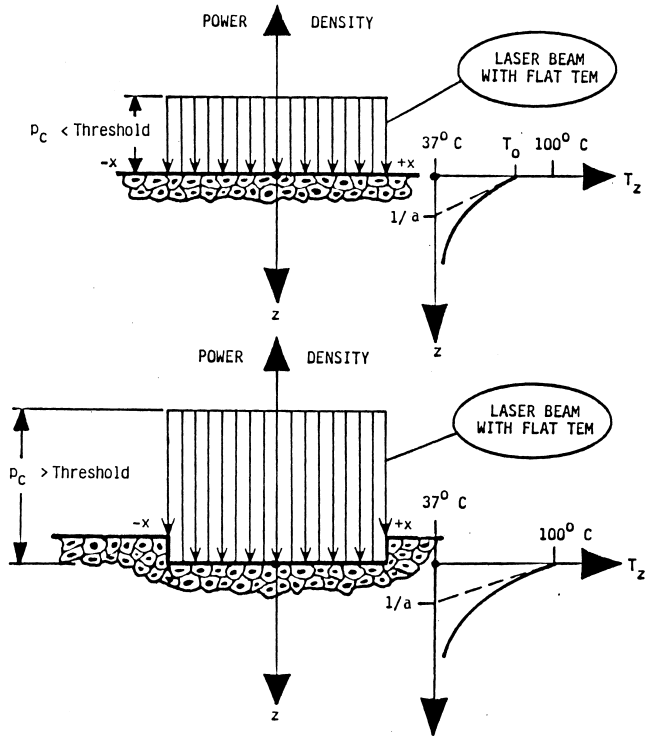

Figure 3-12.

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.

5. INEVITABLE THERMAL DAMAGE TO SUBSURFACE TISSUE: GAUSSIAN LASER BEAMS

When the laser beam is gaussian, and the boiling diameter of the beam is comparable to the effective diameter, then it will produce a crater in the tissue that has a gaussian cross-section in any plane passing through the axis of the beam. This situation is depicted schematically in **Figure 3-13.**

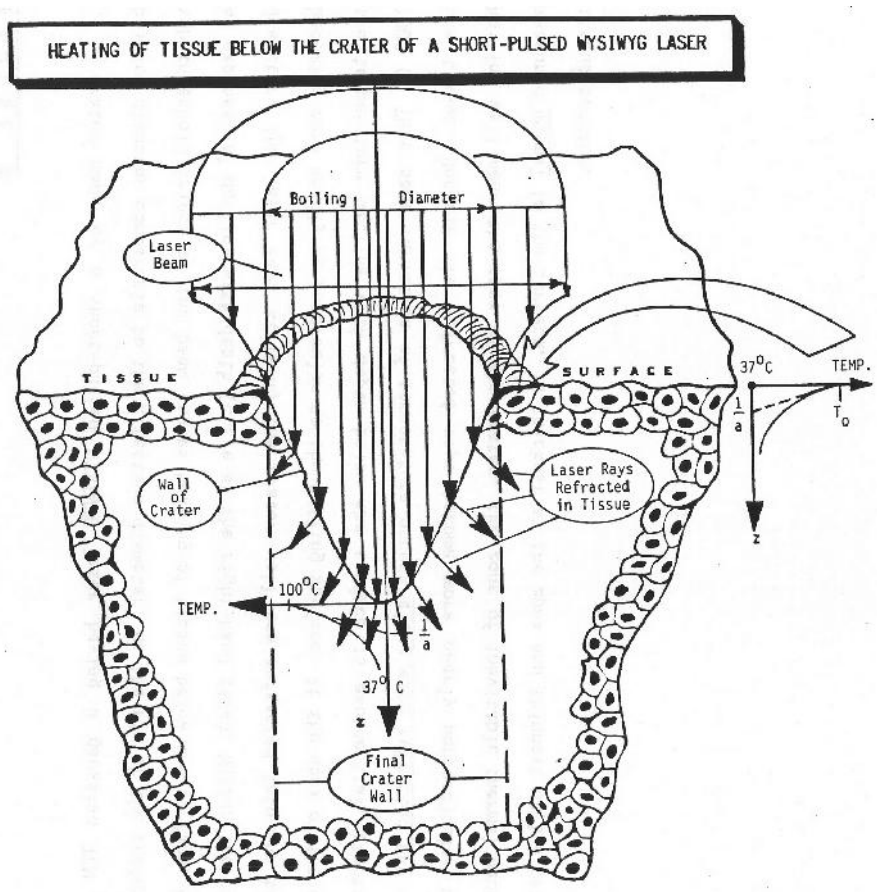

Figure 3-13.

Crater made by a short-pulsed WYSIWYG laser having a gaussian TEM and a boiling diameter comparable to the effective diameter. At the rim of the crater, the sub-threshold fringe of the beam causes heating of tissue below the surface, having an exponential decline with depth (curve at the right-hand side). Within the boiling diameter, the laser rays striking the crater wall are refracted into the tissue, becoming more nearly perpendicular to the boiling surface. At the apex of the crater, the temperature decline with depth below the surface is exponential, starting at 100° C. This same variation of temperature occurs <u>along each refracted ray</u> within the tissue. Because the refracted rays become more nearly perpendicular to the boiling wall near the apex of the crater, the zone of inevitable thermal necrosis, measured <u>normal</u> to the surface, is thickest at the apex and thinnest at the original tissue surface.

The fringe of the beam outside the boiling diameter causes only heating of the tissue to temperatures below 100° C, as shown by the exponential curve to the right of the crater. Within the crater, the laser rays striking the boiling surface are instantly attenuated to the threshold intensity, in a microscopically thin boiling layer, and are refracted as they enter the tissue, so that their direction relative to the surface is more nearly perpendicular. Along each of these refracted rays, the power density diminishes exponentially from the threshold value, according to **Equation (3-1)**, with **A = a** (because **s** is comparatively negligible for the CO_2 or Er:YAG laser). Because the intensity of each refracted ray just below the boiling surface is at threshold level, it is the same for every ray striking the crater wall. Therefore, the zone of inevitable thermal necrosis around the crater is very nearly uniform in thickness perpendicular to the wall, varying only because of ray-to-ray variation in the angle of incidence on that wall.

Consequently, the only significant difference between single-pulse ablation of epidermis by a mesa-mode beam, and by a gaussian-mode beam, is that the latter produces a crater of non-uniform depth. The thickness of the zone of inevitable thermal damage is nearly the same if other factors are equal. The gaussian crater profile is poorly suited to ablation of epidermis to a uniform depth, because substantial overlap is required to produce a crater-bottom that is relatively flat, and the topography of this post-ablation surface is very sensitive to the degree of overlap. This effect is shown in **Figure 3-14**.

D. PHYSIOLOGIC CONSEQUENCES OF INEVITABLE THERMAL DAMAGE TO UNDERLYING TISSUE

1. PERI-OPERATIVE AND POST OPERATIVE PAIN

2. ERYTHEMA AND EDEMA AFTER EPIDERMAL ABLATION

3. HEAT IN LASER RESURFACING: DETRIMENTAL VS. BENEFICIAL EFFECTS

4. THERMAL SHRINKAGE OF COLLAGEN DURING FACIAL SURFACING BY LASERS

5. PATTERN GENERATORS AND SCANNERS FOR ERBIUM:YAG LASERS: EFFECTS ON PAIN

E. TECHNICAL DETAILS OF CARBON DIOXIDE AND ERBIUM:YAG LASERS

1. CARBON-DIOXE LASERS

Figure 3-15 shows the Lumenis UltraPulse® Encore laser in one of its most recent forms. Many minor and major changes of design have been made in this laser since it was first introduced to the surgical market in 1992. It offers per-pulse energies up to 500 millijoules, pulse-repetition frequencies up to 200 hertz, and maximum average power in repetitive-ultrapulse (quasi-continuous-wave) operation of 100 watts. The CPG® computerized pattern generator and the UltraScan® beam-defecting hand-piece offer a wide variety of pattern sizes and shapes, as described

earlier in this chapter, and the ability to choose between full and partial ablation, depth of ablation, and amount of untreated tissue (bridges or normal tissue) left intact.

Figure 3-15.

Photograph of the Lumenis UltraPulse® Encore CO_2 laser with various pattern generator and Scanning hand pieces. Source: Lumenis Inc., Santa Clara, California.

Like most CO_2 lasers, the UltraPulse® requires an articulated arm to deliver the beam to the target, because there are no commercially available optical fibers suitable for transmission of the beam, whose wavelength is 10,600 nm. There is one model of CO_2 laser produced now for the veterinary market that uses a polished, hollow metal tube to transmit the laser beam from the tube to the surgical site. Its internal resonator is of the sealed-off, radio-frequency-excited type, utilizing a patented design referred to as a *slab-laser*, which has been described elsewhere in the literature. Its TEM is a nearly perfect gaussian, which allows focusing of the beam to the smallest possible spot diameter.

The articulated arm shown in **Figure 3-15** is of conventional design, with 7 rigid, 90° elbows, each having a plane mirror of high reflectance set at 45° to the axis of either stub. These elbows are arranged in a close-coupled pair at the proximal (laser) end, separated by a relatively long, rigid, straight tube from a second close-coupled pair, which is in turn separated by another long, rigid, straight tube from a close-coupled triplet of elbows at the distal (hand-piece) end.

Each of these elbows is free to rotate relative to its neighbor, giving the whole assembly a net flexibility comparable to that of a human arm. The articulated arm has three major advantages: it has high transmissive efficiency (86.6% if each mirror has a reflectance of 98%), it preserves the TEM and spatial coherence of the incoming laser beam (unlike an optical fiber), and it can transmit high power density in the beam.

At the distal end of the articulated arm, a hand piece for focusing, collimating, and / or galvanometrically deflecting the beam may be attached. Tissue-spot diameters from 0.2 mm up to 3.0 mm are obtainable.

The input power requirements of the UltraPulse® laser are 110-120 volts, 20 amperes, 50 to 60-hertz, making it operable from a standard baseboard outlet.

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.

2. ERBIUM:YAG LASERS FOR RESURFACING OF SKIN

Figure 3-16 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 3-16.

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^o C; thermal conductivity, 0.13 watt/cm/ ^{o}C; coefficient of thermal expansion 6.9 x 10^{-6}/ ^{o}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 **Figure 3-17**. To maximize absorption in the crystal of the light emitted from the lamp, the axis of the

lamp is located on one focal axis of a cylindrical enclosure having elliptic cross-section, and the crystal axis at the other. The inner surface of the elliptic enclosure is highly reflective. The laser mirrors are mounted outside of the enclosure, which is traversed by a flow of cooling water, to avoid optical distortion of the beam path by turbulence and bubbles in the liquid.

In the so-called free-spiking mode of operation, the Er:YAG laser delivers a sequence of many short, spike-shaped pulses of laser radiation for each pulse of pumping light from the flashlamp. The entire sequence of laser spikes occurs within the time-width of the pumping pulse. Each laser spike is called a micropulse, and the entire sequence of micropulses is referred to as a macropulse. The macropulse duration can be varied within a moderate range (100 to 300 microseconds). Each micropulse is a few microseconds in length. The energy stability of the macropulses is in the order of ±2%. A typical macropulse, with 20 micropulses, is shown in **Figure 3-18**.

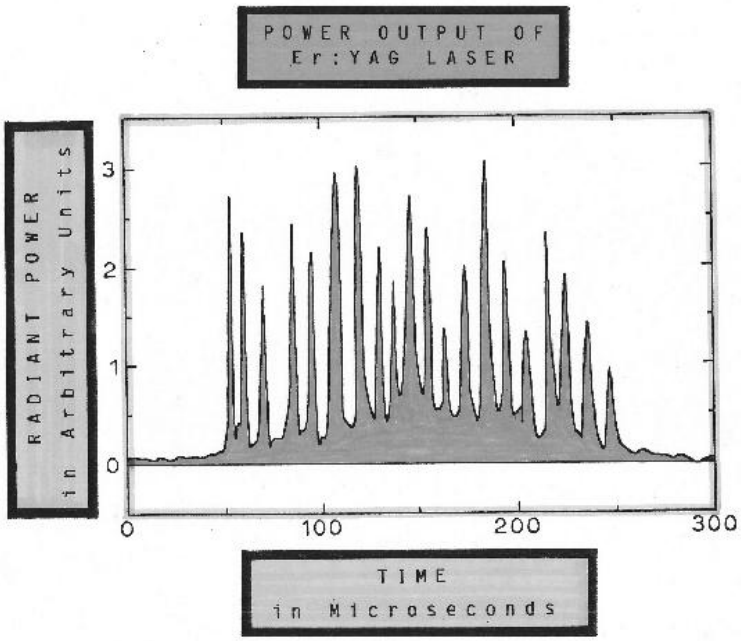

Figure 3-18.

Time-waveform of the radiant power output from an Er:YAG laser made by Spectron Laser Physics, U.K. The duration of the macropulse is 200 microseconds, and it contains 20 micropulses. Reprinted from Rose C.H. Haase K.K., Wehrmann M., & Karsch K.R. Occurrence and magnitude of pressure waves during Er:YAG ablation of atherosclerotic tissue: comparison to XeCl excimer laser ablation. Journal of Lasers in Surgery and Medicine, 1996; 19: 274. Note that this waveform can vary from one manufacturer's laser to another's, and with energy per macropulse is any one laser.

An alternative mode of operation for the Er:YAG laser is Q-switching, in which the resonance of the laser head is spoiled by some optical means while pumping is continued at full power, and then resonance of the head is suddenly re-established. The resulting pulse of output power is continuous for a duration of about 100 nanoseconds, and the TEM of the output beam is gaussian rather than quasi-mesa-mode. The variation of peak power from pulse to pulse in the Q-switched mode can be as large as 50%. The short pulse duration and high peak power can cause photoplasmolysis, especially when bone is ablated. This is undesirable, because the plasma totally absorbs the incoming laser beam at all wavelengths, and effectively shields all objects distal to it from further irradiation.

Photoplasma could be used as a means of ablating epidermis, but the power density needed to produce a photoplasma is above 10 billion watts/cm^2. In order to produce such intensity, it is necessary to focus the beam of a laser to a very small spot, and to have a pulse duration in the order of 100 nanoseconds. To achieve the required fluence over a focal spot of 3 mm with a 100-nanosecond pulse (for a Q-switched laser) would require an energy of more than 70 joules/pulse, corresponding to a peak power exceeding 700 million watts! Consequently, photoplasmolysis is not a feasible tissue-destroying process to use for resurfacing skin.

3. SPECIAL BIOPHYSICAL EFFECTS OF FREE-SPIKING ERBIUM:YAG LASERS

Walsh and Deutsch [11] reported in 1989 that an Er:YAG laser made by Schwartz Electro-Optics was operated by them in the free-spiking mode at a repetition frequency of 2 macropulses per second, each containing a train of 20 micropulses of one-microsecond duration. They measured ablation rates of this laser in bone, pigskin, and bovine aorta. The TEM of the laser beam was approximately mesa mode. Calculations made using their data for ablation of guinea-pig skin are given in **Table 3-3**.

If the values of excess fluence, $\mathbf{\Delta f}$, in **Table 3-3** are divided by the corresponding values of $\mathbf{z_a}$, the quotients are apparent values of the latent heat of vaporization, $\mathbf{h_{vw}}$, of guinea-pig skin (which included both epidermis and dermis in these experiments). **Table 3-3** shows that the values of $\mathbf{\Delta f/z_a}$ range from a high of 4,396 joules/cm^3 at the lowest excess fluence to a low of 1,805 joules/cm^3 at the highest value of $\mathbf{\Delta f}$. The plot of $\mathbf{f}$ vs. $\mathbf{z_a}$ shown in Figure 3A of their paper exhibits a curvature which is concave upward. Because the actual value of $\mathbf{h_{vw}}$ for their skin specimen could not have varied except for slow dehydration during the experiment, it is apparent that some higher-order physical effects were present. Their discussion does not appear to account for such effects.

When an Er:YAG laser is operated in the free-spiking mode, with a fluence per spike of up to 4 joules/cm^2 and a average single-spike-duration of one microsecond, the corresponding peak power density is 4,000,000 watts/cm^2. Under such transient conditions in each micropulse, the boiling of histologic water is not a steady-state process like those upon which steam tables [12] are based, but one in which parameters such as pressure and temperature can vary from point to point within and above the liquid.

When steam is evolved from the liquid surface at very high rates of volume per unit time, it accumulates in the space above the boiling surface,

and can momentarily increase the pressure on that surface far above the value which prevailed just before boiling commenced. Such an abrupt rise in pressure creates a shock wave which propagates away from the boiling surface at a speed well above that of sound. The distinct popping noise made by an Er:YAG laser when ablating epidermis is evidence that miniature shock waves are being emitted from the irradiated surface. In 1996, Rose, Haase, and associates [13] empirically demonstrated the existence of shock waves produced by Er:YAG ablation of atherosclerotic tissue.

By contrast, when a **WYSIWYG** laser delivers pulses that are hundreds of microseconds long, or continuous-wave radiation at power densities less than 100,000 watts/cm^2, the vapor formed above the boiling liquid has time to diffuse away from the surface, and the pressure there remains at or very near the value which existed before boiling began.

It is evident from **Figure 3-18*** that the amplitude varies widely from one micropulse to the next, even though the total energy per macropulse is stable from one to the next. In order to analyze quantitatively the effects of the short, highly variable micropulses, it is useful to approximate the actual macropulse waveform shown in **Figure 3-18** by an idealized macropulse in which the length of each micropulse is fixed at one microsecond, the height of each pulse in terms of power is constant, the pulse repetition period is uniform at 10 microseconds, and each micropulse is a slender rectangle starting and ending at zero power, and delivering 0.05% of the total energy per macropulse. **(*NOTE: Figure 3-18** does not depict the pulse waveform in Walsh and Deutsch's experiment, which was not shown explicitly in their paper.)

Figure 3-19 shows the variation $\mathbf{z_a}$ with **f** per macropulse of an Er:YAG laser vaporizing skin having an assumed hydration of 70% by volume. It was plotted from the data of Walsh and Deutsch for ablation of guinea-pig skin by an Er:YAG laser operating in the free-spiking mode with 20 micropulses, each having an average duration (full width at half maximum) of one microsecond, and an average inter-pulse period of 10 microseconds, within a macropulse of 200 microseconds in length.

It should be noted here that the plot of ablation depth vs. total fluence per macropulse can vary significantly from one manufacturer's erbium:YAG laser to the next, because of the possible differences among the actual waveforms of the macropulses. A macropulse waveform having more of its energy in the hump and less in the spikes (such as that of the Ho:YAG) will be more effective at ablating tissue than one in which more of the energy contained in the spikes.

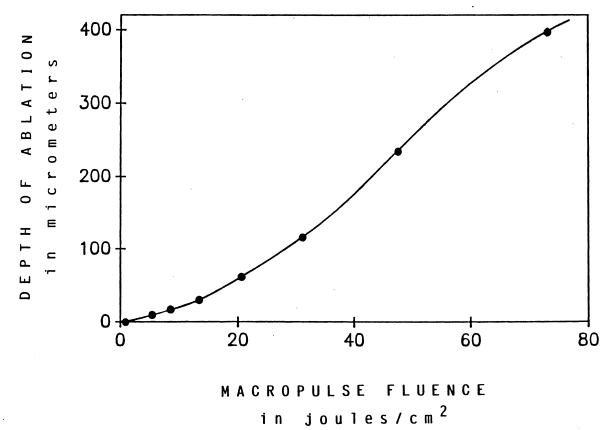

Figure 3-19.

Curve of depth of ablation per macropulse vs. fluence per macropulse from an Er:YAG laser made by Schwartz Electro-Optics, during hole-boring in excised specimens of guinea-pig skin. Drawn by Fisher, J.C. from Figure 3A of Walsh, J.T. &

Deutsch, T.F.; Er:YAG laser ablation of tissue: measurement of ablation rates. Journal of Lasers in Surgery and Medicine, 1989; 9:330.

Spiking during pulsed operation is characteristic of solid lasers other than the Er:YAG ruby and Nd:glass, for example. It is caused by relaxation oscillations in which energy is exchanged between excited states of ions and resonant radiation in the optical cavity. The time required for such oscillations to damp out is approximately equal to the lifetime of the upper laser state, which is hundreds of microseconds in the case of Er^{3+} ions. Limitation of the number of amplifiable modes of the optical cavity of the laser can reduce the spiking.

F. NEW AND EVOLVING TECHNOLOGIES

G. SUMMARY AND CONCLUSIONS

Excerpts from

CHAPTER FOUR

Commentary on Ethics in Cosmetic Laser Surgery

James L. Cromwell, M.D., B.S.C., F.A.C.O.G.

Several ethical norms are expected in our relationship with patients today. They include: *Informed consent* - the willing acceptance of a medical or surgical procedure after understanding the risks, benefits and alternative treatments available, *Honesty* - the exercise of complete and truthful information about the patient's condition, *Confidentiality* - the patient's right to privacy of personal medical information and the right to decide to whom he or she will divulge such history.

IMPORTANT CONSIDERATIONS FOR THE PHYSICIAN USING LASER TECHNOLOGY

Physicians, by virtue of their sense of duty and credentials are caretakers for the ill and misfortuned patients. Our ethical commitments seem clearer when addressing the care of ill patients than when dealing with patients requesting cosmetic, elective, laser procedures.

Certification: Who should be certified to use lasers on patients is a topic around the world at most laser meetings, and there are differing opinions. Some of these opinions are based on secondary gain and are inherently biased, i.e. providers connected to laser companies with possible monetary gain by promoting certain technology to their peers or patients. Other physicians have successful clinics with good reputations in spite of having the physician off - site, while allied health professionals consult, diagnose, treat and follow up patients-a possible ethical dilemma. In an article in *Healthy Aging*: (May/June 2009) "Who Should Fire a Laser" by Marci Landsmann, she quotes Christopher Zachary, MD, MBBS, FRCP, professor and chair of the department of dermatology at the University of California:

"I am biased in favor of safety. There is concern that laser light and cosmetic surgery is being practiced by poorly trained professionals. If this were the gallbladder, there wouldn't be a question that only a doctor should perform the surgery. Using a laser is the practice of medicine."

Non-core physicians and allied professionals play a large role in laser medicine and surgery with some studies showing a somewhat higher complication rate than the core physicians, i.e. dermatologists and plastic surgeons. Other studies showed no difference of complication rate with laser procedures between core, non-core, and allied professionals.[3]

Training: "Cosmetic" surgery is practiced by a diversity of medical specialties that generally provide overall safe, quality skin care with satisfactory results. Each practitioner brings a unique perspective to this complex profession, and each can learn and profit by mutual, unbiased cooperation and education. In reality, many excellent cosmetic surgeons, including plastic surgeons and dermatologists, received further education in the appropriate use of lasers and light therapies after their core residency training. They learned to perform known, and sometimes new procedures using new technologies at weekend and week-long meetings, preceptorships, and mentorships throughout the world. While some laser practitioners have had some training in elective cosmetic procedures during their residency, a 2008 survey of 89 U.S. plastic surgery residencies concluded that "many programs offer inadequate or nonexistent training in cosmetic surgery".[4] Educational programs of many varieties - from on site didactic and clinical programs, to web based CMEs are available to clinicians interested in efficacious and safe treatment using lasers.

Responsibility:

The philosophy of "responsibility to patients by physicians" includes the obligation to treat patients appropriately and to the best of our abilities within the scope of our training (and certification). We should educate patients that Lasers and Light are just tools and not a miracle in and of themselves. We should be aware that as cosmetic laser surgeons, we see a higher percentage of patients suffering from Body Dysmorphic Disorder. Performing procedures on these patients, without psychotherapy is disadvantageous to their care as it is not likely to improve their health, function or self image.

Complications:

The list of potential complications below serves to remind us to pay attention to detail, avoid rushing and distraction, and to follow up on complaints, concerns, and adverse results with compassion, integrity, and second opinions as indicated.

1. Burns
2. Pigmentation
3. Scarring or prolonged healing or redness
4. Ocular injuries
5. Allergic reactions
6. Folliculitis
7. Infections
8. Edema
9. Anxiety or depression out of proportion to the cause
10. Body Dysmorphic Disorder
11. Death

Standards:

Litigation:

1. **Duty** was owed to the patient
2. **Failure** to conform to standard of care
3. **Injury** occurred
4. **Damages** result

SUMMARY

Lasers are complex technologies; best utilized by clinicians who appreciate their potential for safe, efficacious results and their risks for irrefutable harm. Treating cosmetic afflictions may be just as important to the mental and physical health of a patient as treating many other ailments. Laser surgeons should be committed to education, knowing and improving the standards of care, and as our specialty matures, certification in the use of lasers and light technologies throughout the medical community.

From the Medical Student Pledge at the University of Toledo, Adapted from Texas-Houston Medical School "Student Ethical Pledge"

"Knowing my own limitations and those of medicine, I commit myself to a lifelong journey of learning how to cure, relieve and comfort with humility and compassion."

The American Board of Laser Surgery
Incorporated 1984

Excerpts from

CHAPTER FIVE

Safe Use of Lasers in Surgery

Warren B. Seiler III, M.D.

Edward M. Zimmerman, M.D.

Michael D. Swick, DMD

John C. Fisher, Sc. D.

HISTORICAL PERSPECTIVE

When the laser first appeared on the surgical scene, the attention of a small group of imaginative surgeons focused only on its potential to perform surgery in new ways and not on its destructive possibilities. However, with the advent of new wavelengths and the application of lasers in treating more parts of the human body, it became apparent that this new surgical tool had harmful as well as beneficial potential. Undoubtedly, many accidents with surgical lasers occurred before the medical-surgical community at large was aware that this new modality could injure and kill as well as heal! In the authors' opinion, lasers can be just as much of a threat to the safety of patients as can scalpels. Although scalpels can sever critical arteries or nerves in a small fraction of a second, many lasers are powerful enough to severely damage a patient's body, skin, and self-image (i.e. from burn scars).

Prior to August 1, 1976, there were no governmental regulations on the manufacture, sale, or clinical use of surgical lasers. On that date, legislation by Congress established jurisdiction of the Food and Drug Administration over all laser products manufactured here or imported into the U.S. The agency of the FDA currently charged with the regulation of lasers is the National Center for Devices and Radiological Health. The applicable rules are published in the Code of Federal Regulations, 21 CFR 1040. In 1990, there was a movement in some states (i.e. Arizona and New York) to pass laws regulating the use of lasers in medical/surgical applications. This was a departure from past practice when only the manufacturers of lasers were regulated and medical professionals were deemed to automatically have adequate training and judgment to use lasers clinically. However, the presumption that a license to practice medicine conveys the skills to practice laser surgery has increasingly come into question. We can expect to see more states adopt stricter regulations regarding the use of lasers by medical personnel. Currently, it is very easy to find state regulations on the use of lasers, who can use which devices, what levels of training different practitioners must acquire, and what level of involvement by the physician is required. This information can be found on the websites for the state board of medical examiners for each state. There are also state medical board representatives that should be able to advise a practitioner as to this information.

There are a number of books, documents, papers in surgical peer-review journals, and websites addressing the problems of safety associated with lasers. Some of these have been encyclopedic, like the superb volume of Sliney and Wolbarsht, which covers in great detail all the hazards associated with sources of light in general. Some have been narrowly focused on just one aspect of laser safety, such as smoke plumes. Recommendations specifically dealing with the Safe Use of Lasers in Medicine and Surgery are the booklet ANSI Z136.3 (1988 and updated in 2005) and Z126.3 (2005) on the Safe Use of Lasers in Health Care Facilities published by the American National Standards Institute. These are both an outline and an index of laser hazards and a reference for clinicians. There are many websites including state medical websites, www.lasersafety.com, www.aslms.org, and university websites that have very valuable information about laser safety. Many facilities, especially universities, now have a certified laser safety officer. The authors, having contributed chapters and papers on laser safety to various books and periodicals, believe that the encyclopedic texts are of greatest value to the laser safety officer of a hospital, but often contain too much detail for the busy practitioner. A general text on laser safety must deal with every minute aspect of the problem, including many facts that are

less relevant to the operating room or office. Therefore, this chapter discusses those risks of lasers in surgery that are more likely to cause problems for the practitioner.

Misconceptions

Since the early days of lasers in surgery, certain false notions about the hazards of surgical lasers have appeared and survived. Some of the more common ones are:

Are Lasers "Star Wars" Death Rays?

Do Lasers Cause Cancer?

Do Lasers Disseminate Viable Malignant Cells?

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:

Laser-Specific Hazards/Risks

1. Burns from Laser-Ignited Combustion

Fires in Elastomeric Endotracheal Tubes Carrying O_2

Poster.

Figure 5-1. Mallinckrodt Laser Tube. Source: Cardinal Health website, 2011.

Use of Nitrous Oxide and Oxygen in Dental Procedures

Burning of a Flexible Bronchoscope in O_2

Ignition of Rectal Gas

Laproscopic Surgery

Ignition of Sterile Drapes or Pads

Combustion or Vaporization of Surgical or Diagnostic Preparations

2. Accidental Laser Trauma to Untargeted Body Parts

Perforation of Hollow Organs and Vessels

Injury to Nerves, Brain, and Spinal Cord

Injury to Cornea, Sclera, Lens, or Fundus of the Eye

Injury to All Other Parts of the Body, Especially Skin

3. Inappropriate or Unskilled Use of Lasers

Sites of Ocular Damage in Relation to Wavelength

The wavelength of light, from lasers or non-coherent sources (light or energy based devices), determines the site of damage in the eye.

1. MICROWAVES, X-RAYS, and GAMMA RAYS pass through the eye with little absorption, but the radiant dose is cumulative for x-rays and gamma rays, which can damage the entire eye. Microwaves cause near-uniform heating of the whole eye, but the dose is not cumulative from one exposure to the next.
2. FAR ULTRAVIOLET (<300 nm) and FAR INFRARED (>7000 nm) are absorbed at the scleral or corneal surface.
3. NEAR ULTRAVIOLET (300 to 400 nm) is absorbed by the cornea, sclera, aqueous humor, and crystalline lens of the eye. It is an important cause of cataracts in people who spend time outdoors in sunny climates.
4. VISIBLE and NEAR INFRARED (400 to 700 nm and 700 to 1200 nm) are partially absorbed in the anterior structures of the eye, and chiefly by the fundus, notably the retina.

Protection of Eyes from Laser Light

Laser Treatment of Lesions of Unknown Cytology, Histology, or Spatial Extent, or Lesions Not Fully Irradiable

Excess Thermal Necrosis from Low Power Density or Prolonged Exposure

Delayed Fistulae Caused byPhotodynamic Therapy of Mural Tumors in Hollow Organs Such As the Trachea, Esophagus, Bladder, and Bowel

Uncontrollable Bleeding During Laser Surgery

Choice of the Wrong Laser for a Given Procedure

Inappropriate or Unskilled User of Lasers

4. Adverse Sequelae of Laser Surgery or Therapy

Smoke and Vapor from the Surgical Target

Mechanism of Smoke Generation

Effects of Smoke on the Respiratory Tract

Viral Particles in Laser SmokeThe Need for Adequate Evacuation of Smoke

Breakage of Laser Fibers During Surgery

5. Malfunction of Lasers and Related Equipment

U.S. Federal Regulations

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Excerpts from

CHAPTER SIX

Considerations in the Selection of Equipment

John C. Fisher, Sc.D.

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