



# Laser Operating Parameters for Hard and Soft Tissue, Surgical and PBM Management

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## Core Message

New technologies introduced into clinical dentistry in recent years have added immeasurably to the quality of care that may be provided. Lasers, dental implants, CAD/CAM, and motorized endodontics have all improved clinical outcomes but require a significant investment in hardware and, most importantly, education to understand concepts and protocols. As with all medical instrumentation, it is not enough to follow basic guidelines or “preset” parameters in approaching each patient situation. A deep understanding of the technology, how it interacts with the patient’s tissues, and what variables are important to consider are necessary for a successful clinical outcome.

Of the four technologies mentioned above, lasers are, perhaps, the easiest to abuse. Without basic education, it is nearly impossible for a clinician to accomplish the first successful case of implant placement, CAM/CAD crown fabrication, or motorized endodontic treatment. However, with a manufacturer-provided user’s manual, a dentist with a new laser may push a button that says “gingivectomy” and attempt a procedure.

The purpose of this chapter is to explore the parameters that are important, and in some cases critical, to successful laser therapy. This discussion then leads to insights into specific lasers and tissues.

As has been presented, various lasers can be used for surgical treatment of both hard and soft tissues. In addition, photonic energy can be injected into tissue to affect cellular metabolism beneficially. In each of these therapies, applied energy density, its wavelength, and the time over which it is applied represent critical factors affecting the outcome.

## 4.1 Intrinsic Properties

A number of parameters are dictated by design decisions during the manufacture of any clinical laser. They cannot be chosen or controlled by the operator and must be accepted with their benefits and shortcomings.

### 4.1.1 The Photon

A photon is a quantum of electromagnetic energy and has been discussed extensively in previous chapters. All photons are certainly not alike. The amount of energy in each photon is determined by the wavelength or, more precisely, the wavelength is determined by the photonic energy released.

The number of photons generated by a laser is mind-boggling. If an 810 nm diode laser is set to deliver 1 W of output power, 4.08 quintillion photons are produced

every second. That is, 4,080,000,000,000,000,000 photons are released each second. On the other hand, an Er:YAG laser at 2940 nm must produce 14.799 quintillion photons in a second [1]. Photons exert their influence on tissue through absorption by cellular elements, and that absorption is highly dependent on wavelength.

### 4.1.2 Wavelength

As an electron drops from one orbital to one of lower energy, a precise amount of photonic energy is released producing a specific wavelength. Every laser is capable of generating and does generate several wavelengths. The resonant cavity is designed to dampen unwanted wavelengths through destructive interference leaving only the desired output to be amplified and emitted.

### 4.1.3 The Laser

Every laser produces only one predominant wavelength of photons. The laser is usually identified by its active medium and host material. The active medium is a material that can absorb photons of energy and then release them when stimulated further by more photons. Many different materials have been shown to produce this unique “stimulated emission.” Lasers are designated therefore as solid state, gas, semiconductor, and liquid, based on the material used.

Solid-state lasers are most commonly trivalent rare earth ions such as neodymium (Nd) and erbium (Er). The host material suspends the active ions and comprises the majority of the laser crystal. They are most commonly grown-crystal structures such as yttrium aluminum garnet abbreviated as YAG [2]. This combination of elements accounts for the naming convention of Er:YAG and Nd:YAG.

Gas lasers consist of a combination of gases. Carbon dioxide lasers are actually a mixture of approximately one part carbon dioxide combined with four parts nitrogen as the active medium distributed in approximately five parts helium as the host material.

Liquid or so called dye lasers use fluorescein, malachite green, coumarin, or rhodamine as the active medium suspended in water, alcohol, or glycol as a host material. They have the great advantage of being “tunable” from approximately 365 to 1000 nm and, because of the inherent cooling provided by the liquid medium, can produce as much as 20 W of continuous wave output and 1.4 kW of pulsed output. Unfortunately, they are bulky, complex, and expensive and are not currently used in dental applications [3].

Finally, semiconductor lasers, represented by the diode laser, are the most common laser available in dentistry today. They are small, simple, and relatively inexpensive making them very attractive. Most diode lasers are based on combining group III and group V compounds from the periodic table. Those fabricated from gallium arsenide and its derivatives typically lase at wavelengths between 660 and 900 nm, while those utilizing indium phosphide-based compounds produce wavelengths between 1300 and 1550 nm [4]. Recent advances in technology allow laser emission at wavelengths ranging from as little as 370 nm to an amazing 15,000 nm. Only a few of these have practical application in dentistry and are limited to approximately 808–1064 nm [5].

All photons produced at a particular wavelength are the same. While a laser manufacturer would like you to believe that their photons are different and better than those of their competitor, it is simply not true. How they deliver those photons to the tissue and how you are able to control that process are unique to each product and could be a factor in your decision to purchase that particular laser.

### Free-Running Pulse Laser

Free-running pulse is a term applied to a laser, meaning that the laser emission lasts as long as the pumping process is sufficient to sustain lasing conditions. Lasers such as Nd:YAG, Er,Cr:YSGG, and Er:YAG release photons as a train of pulses. Most commonly, a flashlamp similar to a photographic flash injects a large number of photons into the active medium. This begins a cascade of events that have been described previously. As long as the flashlamp is injecting photons, lasing will continue. Therefore, the pulse width or length of time that photons will be emitted is controlled by how long the flash lasts. Usually, the duration of the flash in dental lasers and, therefore, the laser output of photons are in the range of 50–1000  $\mu$ s. Complex control electronics usually allow the operator to select this vital parameter through a touch screen.

The amount of energy in each of these pulses is controlled by the intensity of the flashlamp. If the flashlamp injects twice as many photons to stimulate the active medium, approximately twice as many coherent laser photons will be produced in the resonant cavity and be emitted within the same selected time frame or pulse width.

Lasing is not a continuous activity during the duration of the pulse. It takes some time measured in microseconds for the process to build. As energy is released, the number of stimulated photons is depleted and drops below the state of population inversion. As more photons continue to be injected by the flashlamp, the critical level is again exceeded and laser photon production again occurs. This process repeats itself about every 5  $\mu$ s.

Figure 4.1 is a representation of actual laser activity.

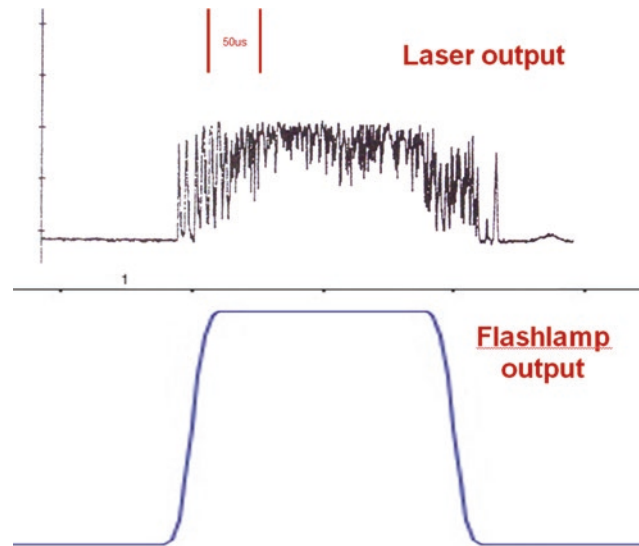


Fig. 4.1 As long as photons are injected into a resonator by a flashlamp or other sources, stimulated coherent photons will be emitted. The amount of lasing activity changes throughout the pulse

### Continuous-Wave Lasers

Lasers such as the KTP, diode, and many CO<sub>2</sub> emit photons on a continuous basis rather than in pulses. The flashlamp is replaced by an electric current, which injects electrical energy instead of photons into the active medium. However, the result is still the release of photons in the familiar stimulated emission mode. As long as the laser is energized, a continuous stream of photons will be emitted. A light-emitting diode (LED) is not capable of laser activity but is a common example of electrical energy being converted to photonic energy. All diode lasers rely on this direct energy conversion.

The CO<sub>2</sub> laser is a bit different. Electrons are passed through a gas starting the lasing process using a high-voltage transformer that can operate continuously. As long as the gas is energized, lasing will occur. Using control circuits, the output can be continuous wave, variable gated continuous wave, pulsed, super-pulsed, gain-switched, or Q-switched.

In super-pulsed mode, the pulse peak power driving the laser discharge can be several times the average continuous-wave power. A super-pulsed CO<sub>2</sub> laser is gated “on” for about 1/3 of the time. This allows it to be driven three times as hard when “on” to produce the same average power while still avoiding overheating. The pulse width is usually in the 5–1000  $\mu$ s range with peak power reaching triple that of its continuous-wave counterpart [6]. Super-pulsed lasers available in dentistry usually operate at a wavelength of 9.3 or 10.6  $\mu$ m.

Gain-switched CO<sub>2</sub> lasers can produce megawatts of peak power, while Q-switched lasers with pulse widths in the nanosecond range can produce peak powers that are several hundred times the average output. These lasers are not currently applied in dentistry.

### Variable-Gated Continuous-Wave Lasers

Continuous-wave lasers may be turned on and off repeatedly by an electronic circuit generating what is commonly called a variable-gated continuous wave. The term “variable” suggests that the ratio of on-time to off-time is not fixed but each parameter can be varied independently. Currently, on-times and off-times can be selected in the range of approximately 10–1000 ms. One laser manufacturer has developed a unique system that allows on-times of as little as 18 μs with pulse repetition rates of up to 20,000 pulses per second.

### Clarification of Pulsed Laser Concept

Confusion continues regarding very important differences in the nature of laser photon emission. Continuous-wave output is inherently different than pulsed output.

Variable-gated continuous-wave lasers are often incorrectly designated as “pulsed lasers.” Continuous-wave lasers produce the same number of photons in each microsecond whether they are gated or not. Gating simply represents an on-off switch. If the on-time is twice as long, twice as many photons will be emitted. Of course, no photons are delivered to the target tissue during the off-time. This concept is illustrated in ■ Fig. 4.8.

The most important characteristic of a pulsed laser is the ability to store and release energy very rapidly. This creates very high peak powers. Pulsed lasers, as described in ► Sect. 4.1.3.1, emit a predetermined amount of energy during the pulse. If a pulse is half as long, the same total number of photons will be emitted but all must be emitted in half the time. Therefore, twice as many photons will be emitted in each microsecond during the pulse. It should be evident that if the pulse width is made very short, very high peak powers will result—a very large number of photons will be emitted in each microsecond. This concept is illustrated in ■ Fig. 4.9.

The best way to illustrate this dramatic difference between pulsed and continuous-wave emission is to compare two lasers that, on the surface, appear to be nearly the same. ■ Table 4.1 shows the large difference in actual output.

#### 4.1.4 The Photon Beam

As photons leave the resonating chamber through the partially reflective mirror, they are gathered by a series of lenses. These lenses collimate the photons making

■ **Table 4.1** Comparison of pulsed Nd:YAG and variable-gated continuous-wave lasers adjusted to what appear to be identical settings. The resulting pulse energy, average power, peak power, and peak power density are calculated to be dramatically different

Laser type	Nd:YAG	Diode
Wavelength	1064 nm	1064 nm
Mode	Pulsed	Variable-gated continuous wave
Delivery fiber diameter	600 μm	600 μm
Power displayed on screen	5 W	5 W
Pulse width	100 μs	100 μs
Pulse repetition rate	50 pps	50 pps
Pulse energy	100 mJ	0.5 mJ
Average power	5 W	0.025 W
Peak power	1000 W	5 W
Peak power density	353,678 W/cm <sup>2</sup>	1768 W/cm <sup>2</sup>

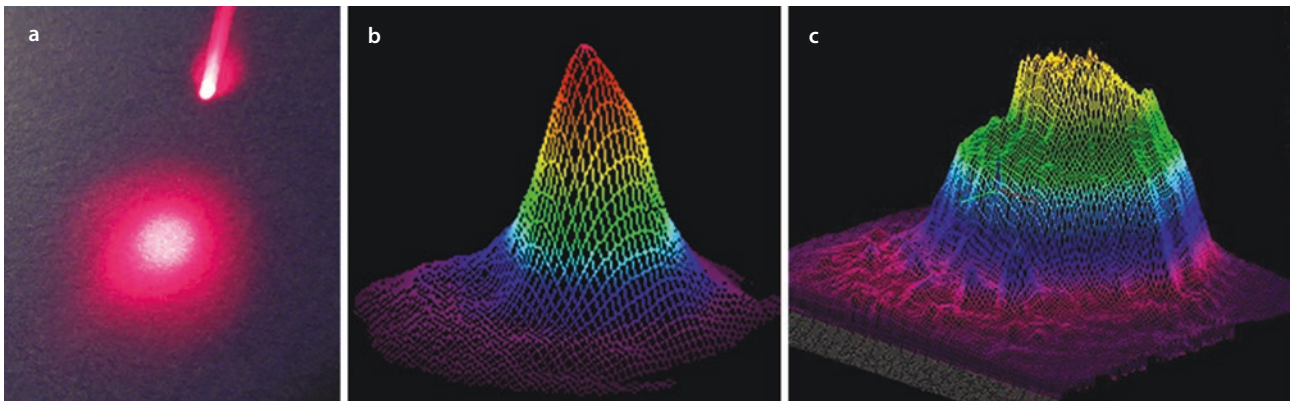
them all travel parallel to each other in a common direction, or they focus the beam to the correct spot size to be fed into the delivery device.

### Beam Profile

The spatial profile of the laser beam is often, erroneously, assumed to be homogeneous with energy output uniform across the entire beam. Most lasers emit in the “fundamental transverse mode” also called the “TEM<sub>00</sub> mode.” This output is Gaussian in cross section as it leaves the resonator as depicted in ■ Fig. 4.2.

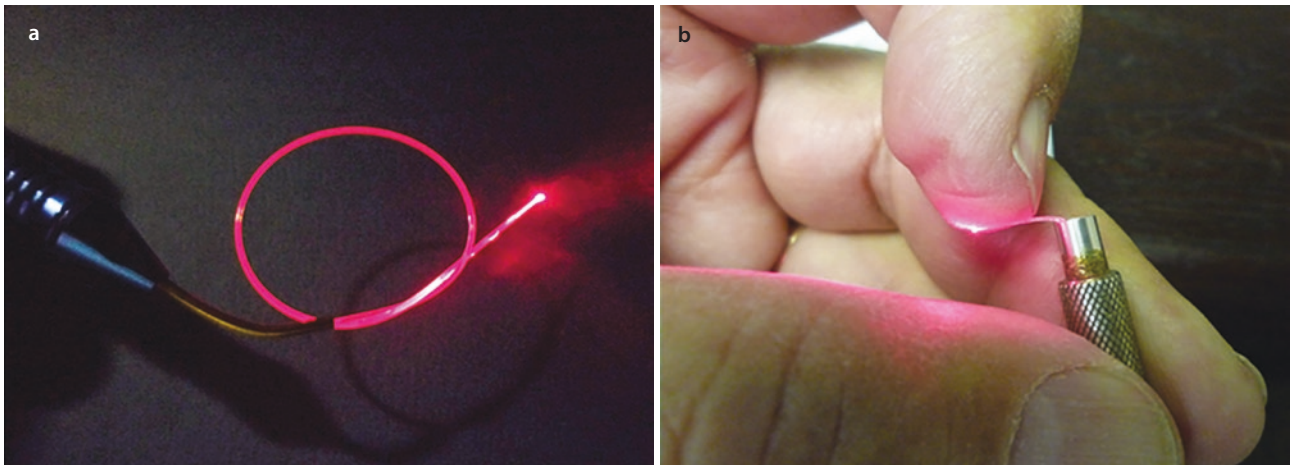
The density of photons is significantly higher at the center of the beam. Cells directly on the beam axis are irradiated at a very high fluence, while those cells on the periphery of the incident beam receive insufficient cellular energy to produce any surgical effect. Laboratory measurement of the actual output from a diode laser reveals that the power density at the beam center is more than twice the average calculated from the applied power while power density on the periphery is as low as 5% of applied power. Significant amounts of sub-ablative energy are deposited into tissue on the fringes causing heating and dehydration instead of the desired ablation.

By the time the light is emitted, imperfections in the laser tip and the fiber-optic bundle can cause further variation from this theoretical output as shown in ■ Fig. 4.2c.



**Fig. 4.2** The output from the laser fiber in **a** is obviously nonuniform. The plot in **b** depicts power density at different points across the beam and is approximately Gaussian in cross section. **c** Depicts

the output profile of an Er:YAG quartz. The vertical axis indicates relative energy density. (Photo courtesy of Frank Yung DDS)



**Fig. 4.3** **a** A 400  $\mu\text{m}$  diameter optical fiber can be tied in a knot and still transmit most of the laser energy. The red color visible in the knot indicates that a small number of photons are escaping. **b** As

long as the protective polyamide coating is present, the same fiber can be bent at nearly a  $90^\circ$  angle without breaking

#### 4.1.5 Delivery System

Laser energy exiting the resonant chamber and passing through collimating and sizing lenses is delivered to one of the three devices for transportation to the tissue: optical fiber, semiflexible waveguide, or articulated arm.

##### Optical Fiber

The propagation of light in fibers depends on the principle of total internal reflection [7]. The core layer of the fiber, made of fused silica, has a larger refractive index than the outer cladding layer. The incident laser energy is reflected off this boundary layer and is trapped inside the core. The core and cladding are coated with a buffer material, such as polyamide which has a refractive index slightly greater than that of the cladding and has the additional advantage of mechanically protecting the fiber.

Optical fibers are exceptionally flexible as shown in **Fig. 4.3a, b**. Even with a severe bend, they continue to reflect photons off the boundary and continue to travel inside the fiber. **Figure 4.3b** illustrates this phenomenon. However, even a slight nick in the fiber will result in an instantaneous fracture. The polyamide plastic coating surrounds the entire fiber to prevent damage from accidental contact.

For dental use, these fibers are produced in diameters from 100  $\mu\text{m}$  to more than 1200  $\mu\text{m}$ . The significance of selecting a particular size will be discussed later in this chapter. These optical fibers are very attractive to use since their flexibility and lightweight make them easy to maneuver when accessing challenging areas in the oral cavity. However, they are only able to be used with short-wavelength lasers such as diode and Nd:YAG. Longer wavelength photons are strongly absorbed in all optical fibers.

## Semirigid Hollow Waveguides

Photons of longer wavelength travel easily through an air medium. Most erbium wavelengths deliver energy through a hollow, semirigid waveguide with inner diameters of typically 300–1000  $\mu\text{m}$ . The inner surface is coated with a silver mirror finish and then with silver halide, creating a very efficient dielectric reflector for infrared wavelengths.

Photons traveling down the lumen of this tube reflect off the sides and continue to the exit portal. However, while bouncing back and forth, photons lose any semblance of collimation and, also, assume a Gaussian distribution. Therefore, lenses are again needed to collimate the output and direct it into the final delivery handpiece. From there, it is delivered to the tissue, using a variety of end attachments, such as a tipless handpiece, a sapphire or quartz tip, and metal or ceramic “guides.”

These waveguides have very limited flexibility and a large diameter making them somewhat unwieldy to maneuver within the oral cavity. There are also some limitations on the power level that may pass through without damaging the silver coating.

## Articulated Arm Waveguides



For even longer wavelengths such as  $\text{CO}_2$ , rigid, hollow waveguides are used. Light is redirected using articulated arms and reflective mirrors. These usually consist of seven segments connected with joints containing carefully aligned mirrors. The mirrors move in such a fashion that the beam is always directed down the center, no matter how the joint is turned. Photons entering each segment are transmitted in a straight line down the center of the tube. When they reach the end, they strike a mirror and are reflected precisely down the center of the next segment.


At the end of the last segment, the photons are presented to mirrors in the handpiece and then on to the delivery tip. In many cases, they are delivered, instead, to a final lens to be focused at some distance away from the handpiece for delivery to the tissue. The distance is usually 6–10 mm to allow vision around the handpiece to the underlying hard or soft tissue. This distance creates some difficulty in precise application of the laser energy although aiming beams provide considerable guidance.


Semirigid hollow waveguides and articulated arm waveguides are used almost exclusively for longer wavelength lasers such as Er:YAG, Er,Cr:YSGG, and  $\text{CO}_2$ . While bulkier and more difficult to maneuver than optical fibers, they overcome the problem of transmission through a solid medium.



### 4.1.6 Emitting Device

The final tip or lens through which the laser energy passes has a very significant impact on interaction with target tissues. An optical fiber has a flat end and will


emit in the pattern shown in  Fig. 4.4a. The beam immediately diverges as will be discussed in  Sect. 4.1.7 below. This pattern also applies when using a removeable flat-ended tip with an Er:YAG or Er,Cr:YSGG laser. These tips are constructed of either quartz or sapphire.

Erbium and  $\text{CO}_2$  lasers using a tipless delivery will emit as shown in  Fig. 4.4b. A lens is the last element in the path of the beam and focuses it. The normal distance between the lens and the focal point is 6–10 mm.

Maximum power density is achieved by positioning the handpiece so that the tissue surface is at the system focal point. In practical terms, the most efficient ablation occurs at this distance. A significant advantage of this system is that, as the tip is moved either nearer or further away than this distance, the beam diverges, spot diameter increases, and power density decreases. This technique is often referred to as “defocusing” the beam and is of value when the aim is to inject energy into the tissue without the consequences of high power density such as ablation. This point is further discussed in  Sect. 4.2.7.

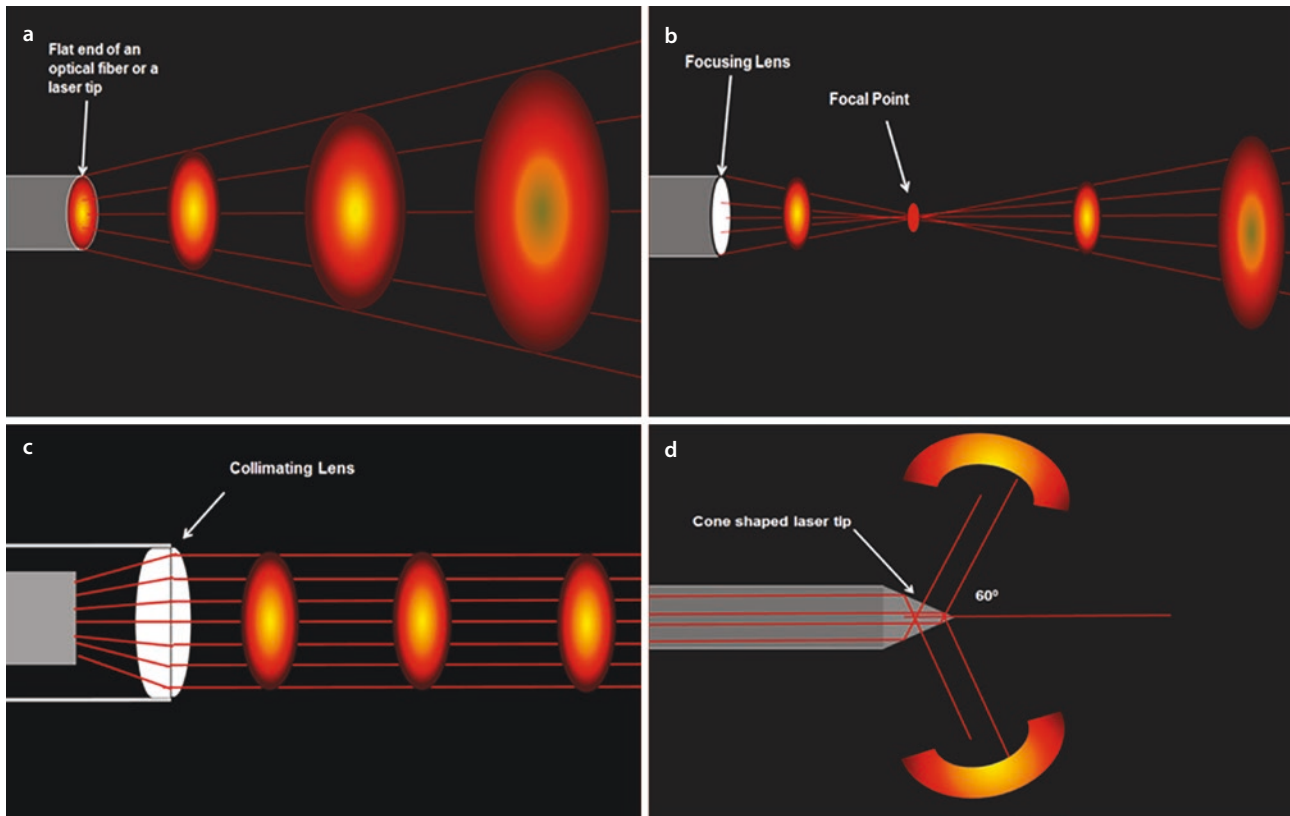
Laser pointers use a collimating lens as shown in  Fig. 4.4c. This allows the beam to have a consistent diameter at a large range of distances. This system is not used in surgical lasers for safety reasons since a misaimed beam could accidentally interact with a distant tissue with the same power density as at the intended target. Also, if the beam were to reflect off a mirror, it could ablate other tissues in random locations. Instead, by using the patterns in  Fig. 4.4a or b, the beam safely loses its power density and, consequently, its surgical ability with distance.

The exception to this rule is seen in the newly developed “flattop” handpiece used in photobiomodulation. In this application, a predictable power density that is independent of tip-to-tissue distance is extremely desirable and important. However, the power levels used are very low, thus mitigating the risk of inadvertent tissue damage.

There are many instances when it is desirable to direct the laser energy laterally. Examples would be when irradiating into dentinal tubules or the threads of a failing implant. A tip has been developed by several manufacturers with a conical terminal end. Energy is internally reflected from this surface and then passes through the opposite side. Practical considerations allow redirection up to approximately  $60^\circ$  from the original forward axis as shown in  Fig. 4.4d.

### 4.1.7 Beam Divergence

As laser energy exits any optical fiber or flat-ended handpiece tip, the beam diverges at a predictable angle. In practical applications, this is usually on the order of  $8\text{--}15^\circ$  per side angle. While this may seem to be a small



■ Fig. 4.4 The output from different laser devices is not the same. **a** Optical fiber or laser tip, **b** tipless delivery system, **c** laser pointer or flattop handpiece, **d** radial firing laser tip

amount, the area on which the delivered photons impact increases rapidly with distance. The influence of this divergence on power density is of great significance and will be discussed in detail later in this chapter.

## 4.2 Adjustable Parameters

The parameters discussed above are all intrinsic to the particular laser being used. They are determined by the laser manufacturer and begin the process of optimally matching the photon delivery to the tissue of interest. The operator has control of a number of parameters that profoundly affect the interaction and, therefore, the outcome of laser therapy. Fluence, irradiance, and peak power are parameters critical to effective laser treatment. However, determining and controlling these parameters are not as straightforward as it would at first seem.

In order to discuss the scientific basis of parameters, it is necessary to have an understanding of definitions:

- Power (measured in W or J/s) is the measure of how much energy is delivered in 1 s of time. One watt is an abbreviation and equivalent of 1 J/s.
- Power density (measured in W/cm<sup>2</sup>) is the power delivered to a unit area of target tissue, usually here 1 cm<sup>2</sup>.

- Irradiance is another term for power density.
- Energy density (measured in J/cm<sup>2</sup>) is calculated as the power density in W/cm<sup>2</sup> multiplied by the total time of illumination.
- Fluence is another term for energy density.

Each laser manufacturer displays a different group of parameters on the control interface. In addition, some manufacturers do not provide direct control of every variable. ■ Figures 4.5, 4.6, and 4.7 are representative of dental lasers currently produced by different manufacturers. Throughout the remainder of this chapter, these three lasers will be used as the basis for discussion and calculation. The parameters available to be directly controlled are different in each case. Therefore, a unique set of equations must be used in each instance to determine the critical parameters. Examples of all appropriate calculations will be provided in Appendix of this book.

### 4.2.1 Average Power

Power and energy are often confused. Energy is, classically, a measure of the ability to do work. It is a “quantity” and is measured in joules. It is important to remember that a joule of energy is produced by a vari-



■ **Fig. 4.5** This 810 nm diode laser has a limited number of variables that may be controlled directly



■ **Fig. 4.6** This 2780 nm Er,Cr:YSGG laser has a choice of two pulse widths. It allows control of average power, but energy per pulse and peak power must be calculated. Levels of auxiliary air and water are set as percentage of arbitrary values determined by supply to the individual dental suite



able number of photons since each photon contains a different amount of energy based on its wavelength.

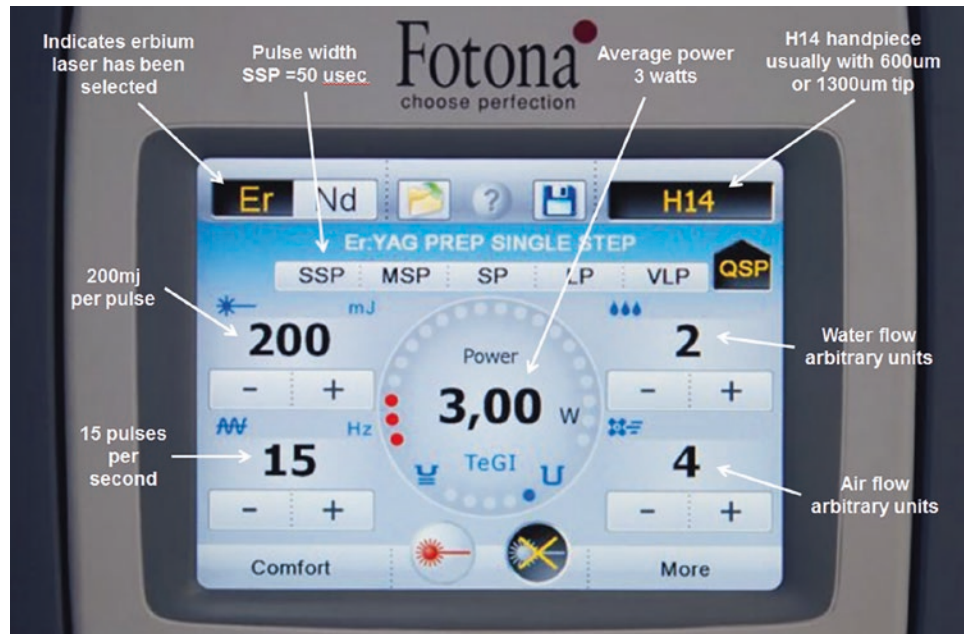
Power is the rate of producing energy. It is the number of joules created in each second of time and is designated in joules per second. A watt is simply an abbreviated word to mean “joule per second.” If 1 J/s or W is produced and continues to be produced for 10 s, then 10 J of total energy will have been produced. If energy continues to be produced at this rate for 60 s, 60 J of total energy will have been produced. If instead the

power (the rate of doing work) is increased to 6 J/s (W) and is applied for just 10 s, the same 60 J of total energy will have produced.

Since average power, peak power, energy per pulse, pulse width, and pulse frequency are so interrelated, they will be discussed as a parameter group interspersed with examples.

Average power represents the total energy produced in 1 s, no matter how uniformly that happens within that length of time. A continuous-wave laser produces a

■ **Fig. 4.7** This laser contains both a 2940 nm Er:YAG laser and a 1064 nm Nd:YAG laser. In erbium mode, it allows selection of six different pulse widths. However, the operator must refer to a manual to find the value. It allows control of pulse energy and pulse frequency. The resulting average power is displayed but not directly selectable



constant number of photons for each time increment that it is activated while a pulsed laser produces photons in short bursts ranging from femtoseconds to milliseconds.

The average power would seem to be provided on two of the three laser control displays in ■ Figs. 4.5, 4.6, and 4.7, but a very important distinction must be made. With a diode laser, the power displayed is usually (but not always) the peak power. The diode laser in ■ Fig. 4.5 allows direct control of the output power, but the power displayed is the “continuous wave” output rather than the average power. Any time that this laser is emitting photons, it is emitting 2 W of peak power.

In *variable-gated mode*, independent selection of off-time and on-time is provided and results in vastly different average power for a given indicated power. If the laser is “on” for 20 ms and “off” for 80 ms, then it is on for 20% of the time:

$$\begin{aligned}\text{Percentage of "on - time"} &= 20 \text{ ms} / (20 \text{ ms} + 80 \text{ ms}) \\ &= 20 \text{ ms} / 100 \text{ ms} = 0.2 = 20\%\end{aligned}$$

Average power is then calculated as:

$$\begin{aligned}\text{Power}_{\text{avg}} &= \text{power}_{\text{indicated}} \times \text{percentage of "on - time"} \\ &= 2 \text{ W} \times 20\% = 0.4 \text{ W}\end{aligned}$$

If the lengths of the on and off periods are changed but stay in the same ratio, average power will be the same. ■ Figure 4.8 illustrates these concepts.

Peak power will always be the same as that indicated on this display, no matter what magnitude or ratio of “on” and “off” times are selected.

While ■ Figs. 4.6 and 4.7 are both erbium lasers, their control features are distinctly different. The Er,Cr:YSGG laser shown in ■ Fig. 4.6 allows control of average power. The pulse width is selected as “H” for hard tissue and represents a pulse width of 60  $\mu\text{s}$ . The laser displays the pulse repetition rate (15 pulses per second). The peak power is adjusted by the laser to provide this output although it cannot be directly controlled. There is no indication of the energy produced in each pulse.

The Er:YAG laser in ■ Fig. 4.7 allows control of energy per pulse and pulse frequency. While average power is indicated on the display, it is not directly controllable. If the amount of energy contained in each pulse and the number of pulses in each second are known, the average power is calculated as the total of all these pulses in 1 s:

$$\text{Power}_{\text{avg}} = \text{Energy per pulse} \times \text{Number of pulses per second}$$

#### 4.2.2 Peak Power

Peak power represents the maximum instantaneous power that the laser produces at any single time. In a continuous-wave laser, the peak power does not change from that selected on the user interface, no matter what other parameters are changed. As has been discussed, it is the value displayed by the diode laser in ■ Fig. 4.5.

With pulsed lasers such as all dental erbium and Nd:YAG lasers, peak power is dramatically affected by the interrelated control of average power, energy per

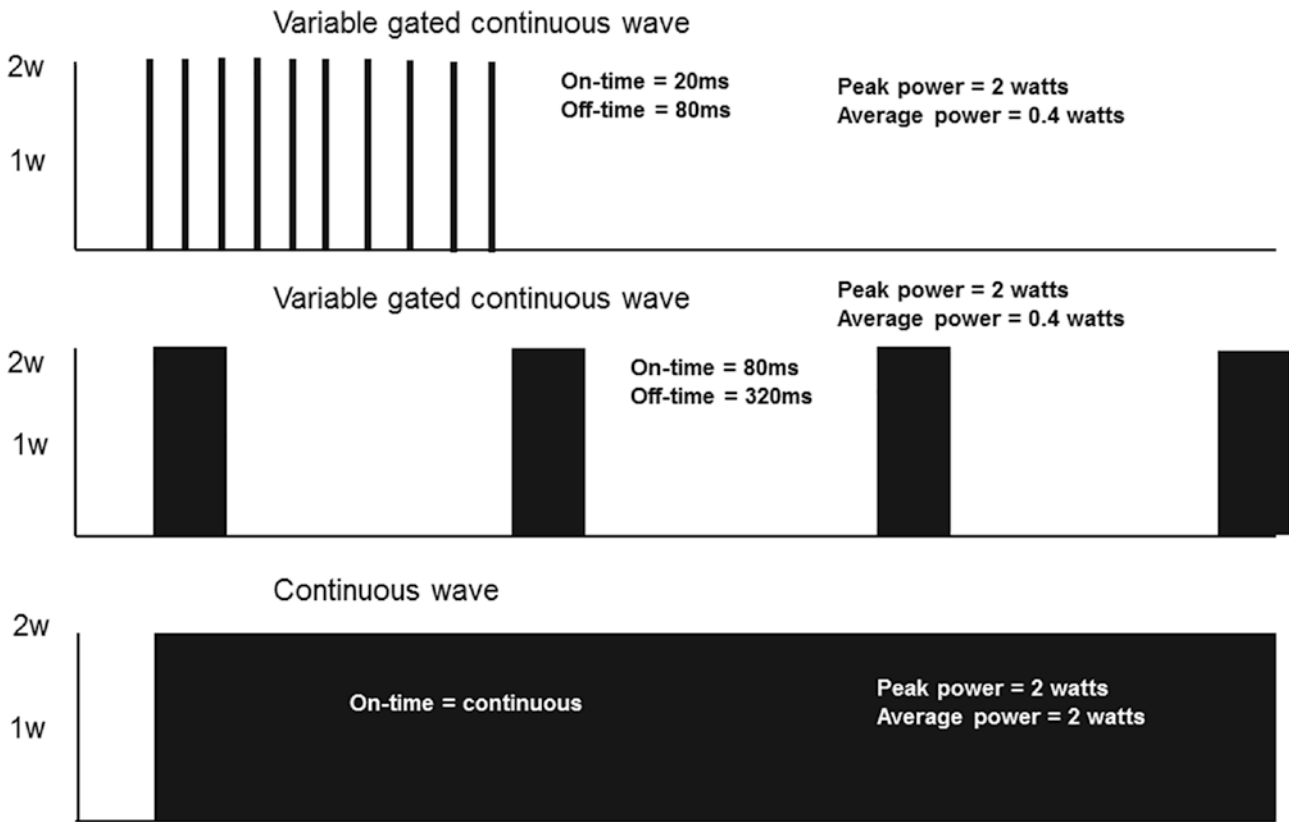


Fig. 4.8 Calculated peak and average power produced at the settings with the diode laser shown in Fig. 4.5

pulse, pulse width, and pulse frequency. If these parameters are not known, they can be calculated.

Peak power can be calculated as:

$$\text{Power}_{\text{peak}} = \text{Energy per pulse} / \text{pulse width}$$

### 4.2.3 Pulse Energy

Some manufacturers allow direct control of the total amount of energy in each pulse instead of controlling the average power. By necessity, if the pulse energy is held constant, altering the pulse width will change the peak power. Figure 4.9 illustrates this concept. It is based on the Er:YAG laser shown in Fig. 4.7. In each case, the energy per pulse is the same. If that energy is delivered in one-half the time, the power or rate of delivery must be doubled.

Some pulsed lasers do not allow direct control of pulse energy but, instead, provide selection of average output power. This output results in vastly different pulse energy and peak power as illustrated in the following example.

In Fig. 4.6, if average power is set to 4 W and there are 15 pulses per second:

$$\text{Energy} / \text{pulse} = \text{Total energy per second} / \text{number of pulses containing that energy}$$

$$\begin{aligned} \text{Energy} / \text{pulse} &= 4 \text{ J} / \text{s} / 15 \text{ pulses per second} \\ &= 0.267 \text{ J} / \text{pulse} = 267 \text{ mJ} / \text{pulse} \end{aligned}$$

At the same time, if average power were to be set to 8 W and there were 30 pulses per second:

$$\begin{aligned} \text{Energy} / \text{pulse} &= 8 \text{ J} / \text{s} / 30 \text{ pulses per second} \\ &= 0.267 \text{ J} / \text{pulse} = 267 \text{ mJ} / \text{pulse} \end{aligned}$$

The energy per pulse is the same even though the average power is doubled.

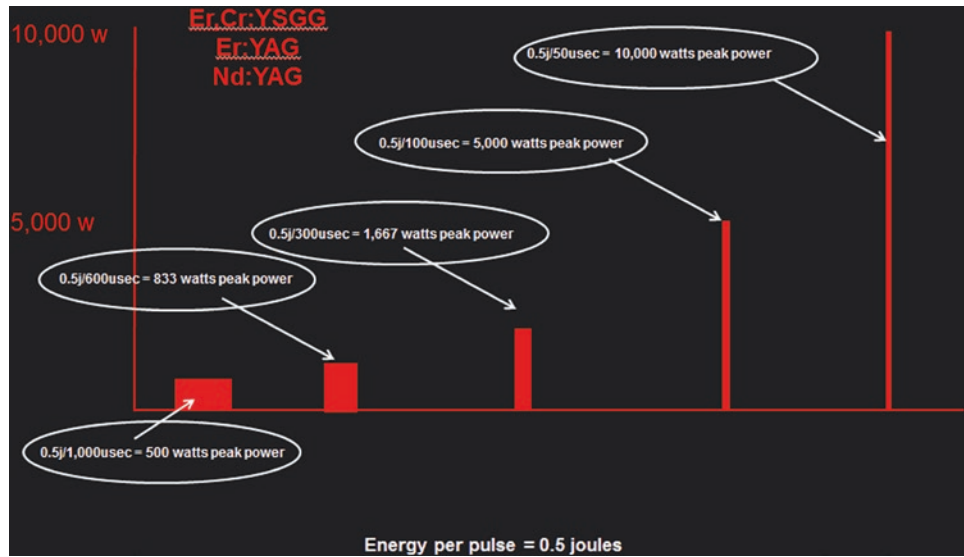
### 4.2.4 Pulse Width

In most pulsed lasers, selection of several different pulse widths is provided. As an example:

For the control panel shown in Fig. 4.6:

- “H” = “hard tissue” = 60 μs
- “S” = “soft tissue” = 700 μs

■ **Fig. 4.9** The peak power emitted by a laser change with changes in pulse width if the energy per pulse is constant. This graph shows examples of free-running pulsed lasers, but some variable-gated pulsed lasers use a similar concept



For the control panel shown in ■ Fig. 4.7:

- “SSP” = “super-short pulse” = 50 μs
- “MSP” = “medium short pulse” = 100 μs
- “SP” = “short pulse” = 300 μs
- “LP” = “long pulse” = 600 μs
- “VLP” = “very long pulse” = 1000 μs
- “QSP” = “quantum square pulse” = a unique pulse train not relevant to this discussion

#### 4.2.5 Pulse Repetition Rate

Doubling the number of pulses that are delivered in a second will double the total amount of energy delivered to the target tissue. If the pulse width and energy per pulse are left unchanged, the peak power will be unchanged. Since there are twice as many pulses applied in a second, the average power will double.

In the Er:YAG laser example of ■ Fig. 4.7:

$$\begin{aligned} \text{Power}_{\text{avg}} &= 0.200 \text{ J / pulse} \times 15 \text{ pulses per second} \\ &= 3.0 \text{ W (J / s)} \end{aligned}$$

If the pulse repetition rate were doubled to 30 pulses per second:

$$\begin{aligned} \text{Power}_{\text{avg}} &= 0.200 \text{ J / pulse} \times 30 \text{ pulses per second} \\ &= 6.0 \text{ W (J / s)} \end{aligned}$$

In each case, the peak power would be 4000 W, if the laser were set to SSP mode.

#### 4.2.6 Diameter of the Final Emitting Device

The number of photons that strike an individual cell and their associated energy is one of the primary determinants of tissue effect. Power density or fluence is a critical factor in determining laser-tissue interaction.

Power density is, by definition, the number of photons passing through a specific area in each second of time. A 400 μm laser fiber has a cross-sectional area of:

$$\text{Area} = \pi \times r^2 = 3.14159 \times (0.02 \text{ cm})^2 = 0.0013 \text{ cm}^2$$

A seemingly small output of 1 W, if passed through this fiber, results in a power density striking an individual cell of 796 W/cm<sup>2</sup>.

$$\begin{aligned} \text{Power density} &= \text{Power} / \text{area} = 1 \text{ W} / 0.0013 \text{ cm}^2 \\ &= 796 \text{ W / cm}^2 \end{aligned}$$

The same 1 W passing through a 200 μm diameter fiber would have a power density of 3185 W/cm<sup>2</sup>.

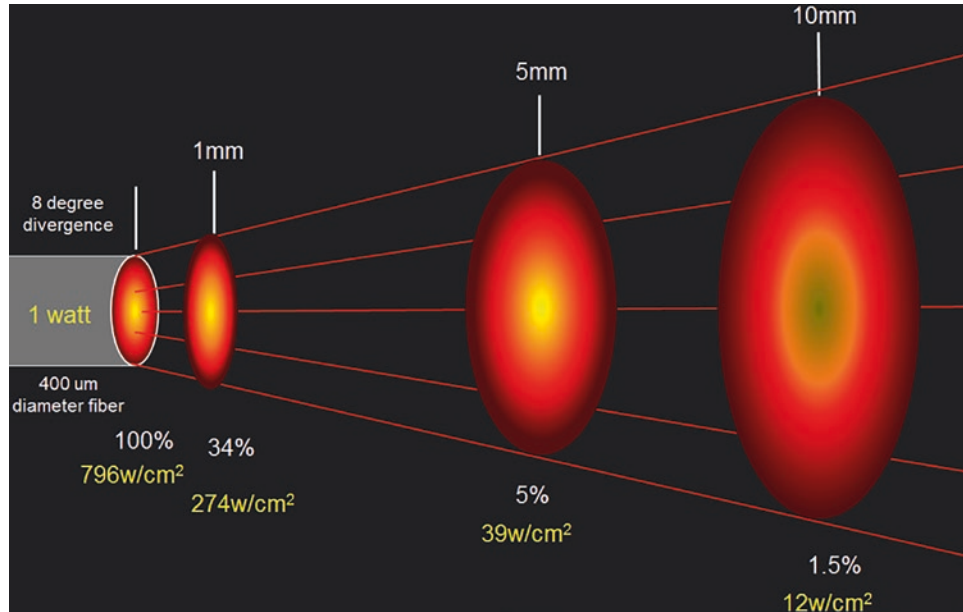
Each of these outputs will have a profoundly different effect on a target cell. ■ Table 4.2 shows the dramatic range of power densities that occur with different optical fiber diameters at average power outputs routinely available in dentistry. It is evident that the choice of fiber diameter has a critical effect on the delivered power density. Simply choosing a 200 μm diameter fiber instead of a 1200 μm fiber makes a 36-fold difference.

The combined influence of pulse width and fiber or tip diameter looms as an overwhelming determinant of

Table 4.2 Average power density in W/cm<sup>2</sup> with different fiber size and power setting

Power density (W/cm <sup>2</sup> )						
Fiber diameter		200 μm	400 μm	800 μm	1200 μm	1.15 cm
Power (W)	0.5	1591	397	99	44	0.5
	1	3185	795	198	88	1
	2	6370	1591	397	176	2

Fig. 4.10 The effect of tip-to-tissue distance on spot diameter and, therefore, on power density. These calculations are based on a beam divergence of 8° per side angle



laser-tissue interaction. In the example of the Er:YAG laser in Fig. 4.7, using an 800 μm diameter tip, producing 1000 mJ of energy per pulse with a pulse width of 50 μs and 50 pps will deposit a mind-boggling peak power density of 4,000,000 W/cm<sup>2</sup> into the target tissue with an average power density of 10,000 W/cm<sup>2</sup>. These calculations are illustrated in Appendix.

### 4.2.7 Tip-to-Tissue or Focus-to-Tissue Distance

Under many circumstances, the optical fiber tip, quartz tip, or sapphire tip is held at some distance away from the tissue. In addition, there are a number of erbium and carbon dioxide lasers that are routinely used without a tip but rely on focused laser energy as discussed in Sect. 4.1.6 above. With a tipless system, the smallest spot diameter and, consequently, highest power density exist at the point of focus. All measurements of distance are referenced to that point.

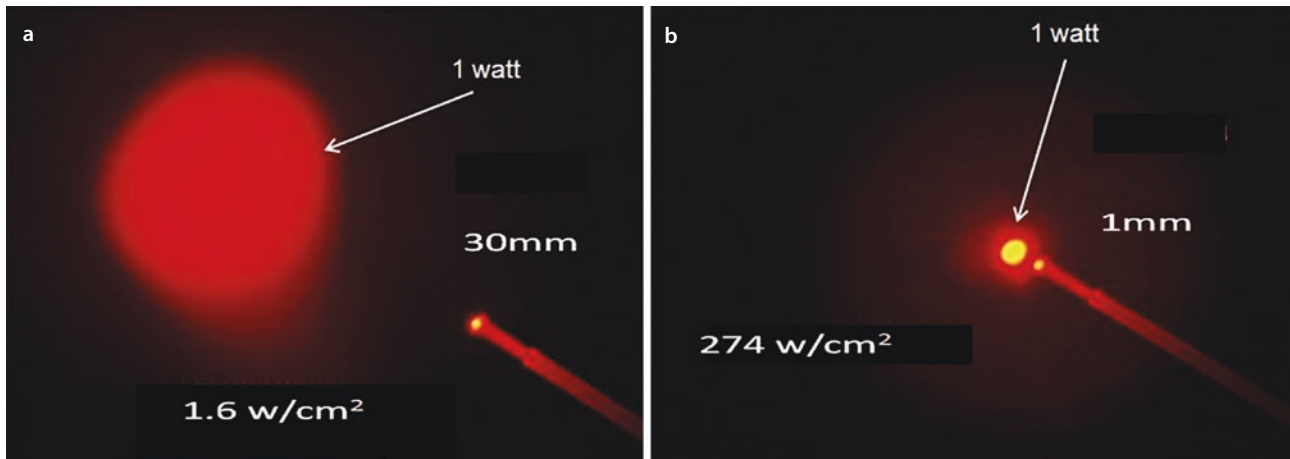
Tip-to-tissue distance, or the comparable focus-to-tissue distance, provides yet another parameter that dramatically affects the density of photons striking an

individual target cell. As laser energy exits an optical fiber or a quartz or sapphire tip, it diverges quite significantly.

The applied energy is distributed over an increasing area as the tip-to-tissue distance increases, dramatically affecting power density at a cellular level as shown in Fig. 4.10. Power density can be diminished by 95% with only 5 mm of tip-to-tissue distance. This is further demonstrated in Fig. 4.11. When using this standard delivery system, the repeatable application of an appropriate energy density is extremely technique and operator sensitive.

Table 4.3 illustrates how significantly the power density changes with just minor changes in the tip-to-tissue distance. It is also apparent that the effect of beam divergence is far greater with smaller diameter emitting devices. A collimated beam eliminates this destructive divergence but is only possible to produce if a lens is the final element in the path of the laser energy as it is emitted

All concepts associated with beam divergence from a flat-ended tip apply to tipless systems with distances being measured from the focal point as demonstrated in Fig. 4.12.



■ Fig. 4.11 Power density at tip-to-tissue distances of 30 mm **a** and 1 mm **b** with an identical 1 W of applied diode laser power

■ **Table 4.3** Average power density at 1 W laser output for different tip diameters and tip-to-tissue distances assuming an 8° per side beam divergence

Power density (W/cm <sup>2</sup> )						
T-T distance		Contact	1 mm	2 mm	5 mm	10 mm
Tip diameter	400 μm	796	275	138	39	12
	800 μm	199	109	69	26	10
	8 mm	2	1.9	1.7	1.4	1.1
	1.15 cm	1	0.92	0.88	0.76	0.62
	1.15 cm (collimated)	1	1	1	1	1

#### 4.2.8 Auxiliary Water

All erbium lasers, together with the 9300 nm CO<sub>2</sub> laser, have the ability to spray water on the tissue during treatment. In the case of hard tissue, this spray is used to cool the tooth or bone and remove debris from the ablation site. Without extrinsic water, the tooth will overheat under repeated pulses and ablation may cease.

Er:YAG laser energy is absorbed in approximately 0.8 μm of water [8–10]. Logic suggests that, if auxiliary water is used, all energy will be absorbed by overlying water before it can reach the tissue surface and no ablation will occur. On the other hand, without extrinsic water, charring will rapidly occur and ablation will cease with significant damage to the tooth.

In practice, other principles resolve this issue [11]. When laser energy strikes the surface of a water film, it is rapidly absorbed causing vaporization. This rapid phase change occurring in a few microseconds creates a shock wave of very significant magnitude generating pressures of a few hundred atmospheres. This shock wave displaces water at the point of contact creating an open channel to the underlying tooth surface [12] as shown in ■ Fig. 4.13.

If the laser tip is completely submerged, water is evacuated from the laser path by a slightly different phenomenon. When laser energy strikes the water directly in front of the tip, it once again causes vaporization. An air bubble is created, as shown in ■ Fig. 4.14, which pushes the water out of the way allowing energy from the remainder of the pulse to strike the tooth surface and cause ablation.

Nahen and Vogel [13] showed that such a bubble forms in the first 10–20 μs of the laser pulse. The balance of the pulse energy is then able to ablate as though no water is even present. Nearly all of the laser energy strikes the tooth surface unimpeded allowing excellent ablation. As the bubble collapses between each pulse, the newly ablated surface is bathed in water, thus cooling, cleaning, and rehydrating the structure.

Within limits, the thickness of the water layer is of little consequence. The bubble formed has a finite size estimated by the author from ultra-speed photographs, such as in ■ Fig. 4.15, to be approximately 3 mm in length.

When only a water mist is used, any droplets in the path of the laser energy are rapidly vaporized allowing passage of subsequent energy.

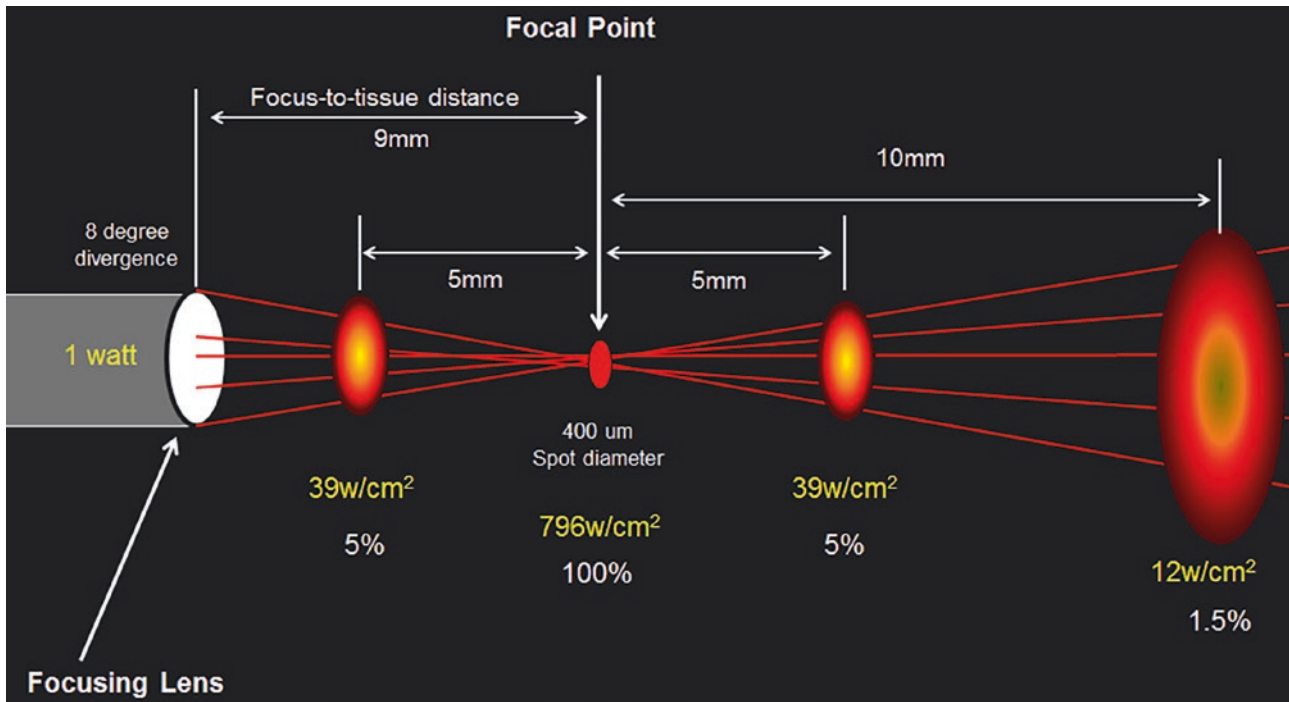


Fig. 4.12 The effect of focus-to-tissue distance on spot diameter and, therefore, on power density. These calculations are based on a beam divergence of 8° per side angle. It is evident that all of the dis-

tances from the focus point are the same as those from the tip in Fig. 4.10. Consequently, the calculated values are identical

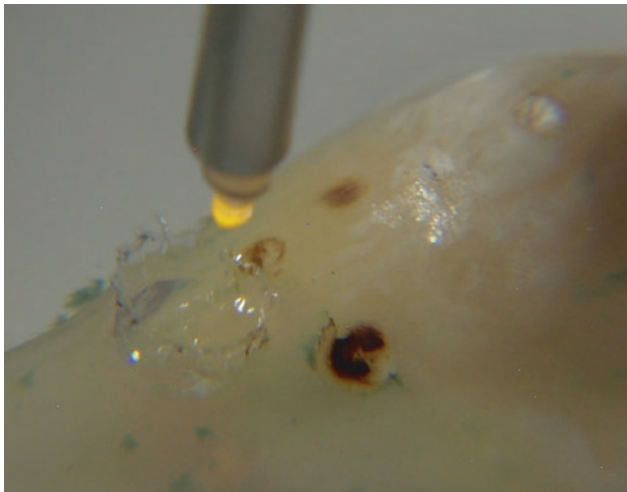


Fig. 4.13 Ultra-speed macrophotograph of Er:YAG laser energy interaction with a superficial water layer. The shock wave generated by the laser pulse displaces the water on the tooth surface



Fig. 4.14 Ultra-speed macrophotography of Er:YAG laser energy interaction when the tip is completely submerged in water. Vaporization of the water in front of the tip creates a bubble forming a channel for ablation

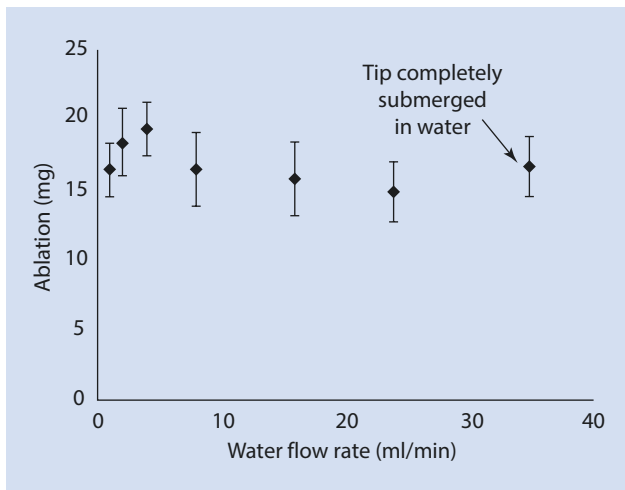
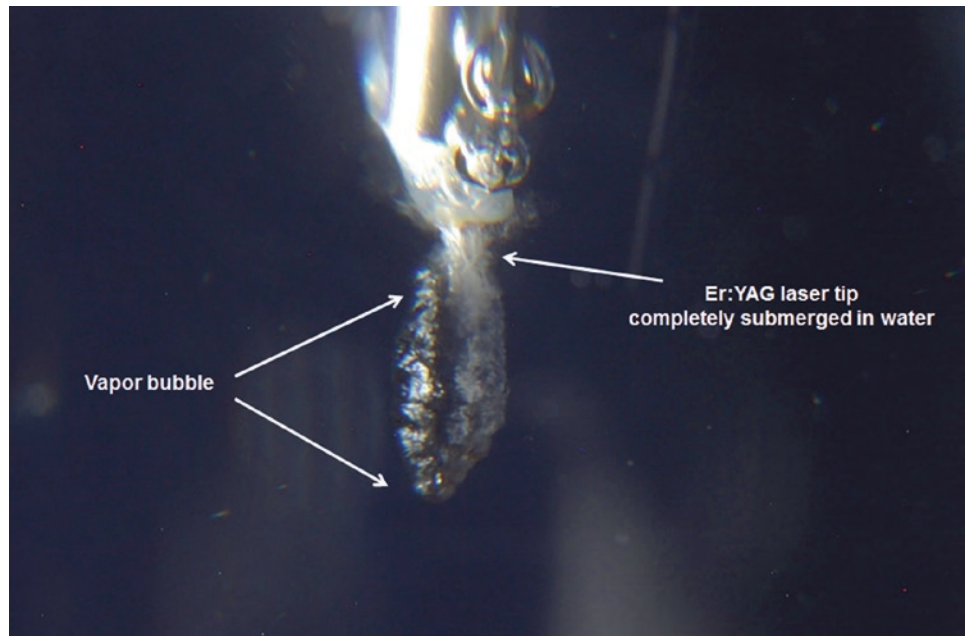
In an experiment to verify the theoretical concepts discussed above, the flow rate of irrigation water was controlled at 1, 2, 4, 8, 16, and 24 mL/min while the tip-to-tissue distance was held at 0.5 mm. Other samples were ablated completely submerged in water at the same controlled distance [11].

Figure 4.16 shows that Er:YAG ablation efficiency is essentially independent of irrigation water flow rate.

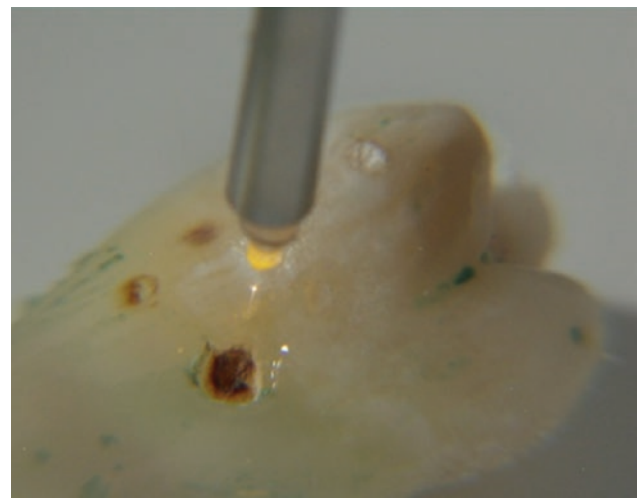
In addition, applying energy with the tip completely submerged in water has no detrimental effects on ablation efficiency as long as the laser tip is within about 2 mm of the tissue surface.

At low irrigation water flow rates, the possibility of dehydration and charring is significant. Without water coolant, residual heat quickly dehydrates the enamel making further ablation inefficient or impossible. At the

**Fig. 4.15** Ultra-speed macrophotography of the bubble formed surrounding an Er:YAG laser tip when it is completely submerged in water. The vaporization bubble in this picture has a diameter of approximately 1.5 mm and a length of more than 3 mm beyond the tip



**Fig. 4.16** Effect of irrigation water flow rate on Er:YAG laser ablation of enamel. Tip-to-tissue distance was constant at 0.5 mm. The volume of irrigation water has a minimal effect on ablation efficiency



**Fig. 4.17** Effect of irrigation water flow rate on Er:YAG laser ablation of enamel. Without irrigation, hydroxyapatite absorbs laser energy causing progressive desiccation, overheating, and carbonization

same time, a layer of exploded hydroxyapatite particles covers the enamel surface and absorbs some of the energy of the next energy pulse, heating this unwanted layer significantly.

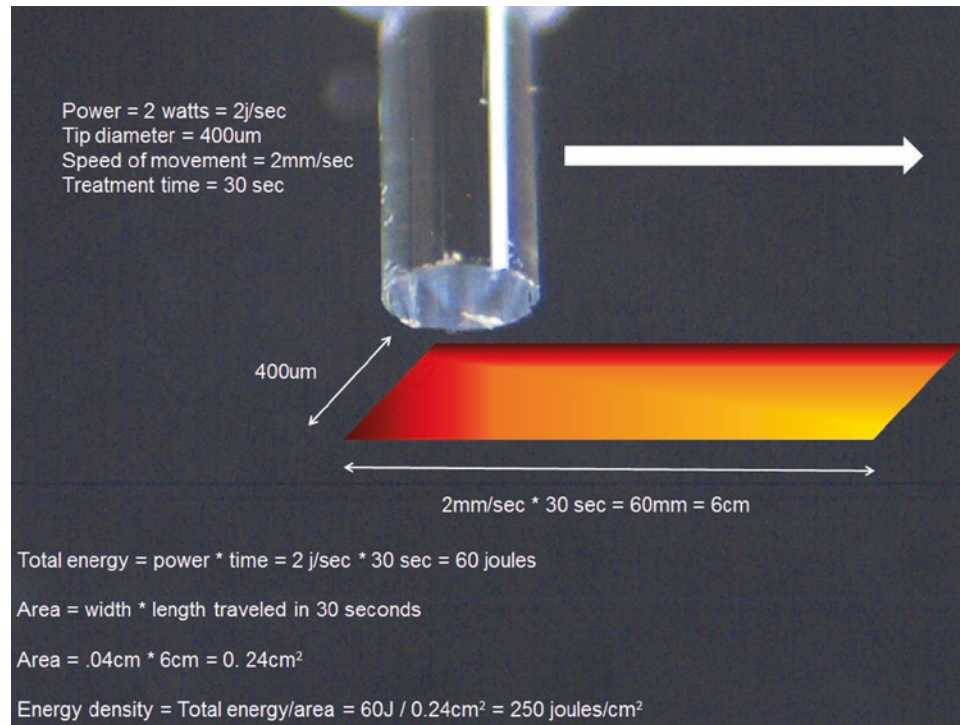
Figure 4.17 shows enamel ablated without water. Thermal damage results in carbonization, cracking, and a loss of laser ablation effectiveness. Water spray both cools the tooth and washes away the debris. Irrigation also serves to enhance ablation rate and efficiency, improve surface morphology, alter chemical

composition, and enhance adhesion to restorative materials.

A flow of at least 8 mL/min is needed to minimize adverse effects. Since flows of up to 24 mL/min have been shown to have no negative effect on laser ablation efficiency, it is prudent to use 8–24 mL/min to ensure adequate hydration. In fact, it can be argued that using the maximum amount of water possible is an appropriate strategy. However, ultimately, site visibility by the dentist as well as patient tolerance for



■ **Fig. 4.18** The energy density striking individual cells is calculated by dividing the total energy delivered by the total area irradiated



large water volumes will dictate the water flow rate used.

Super-pulsed CO<sub>2</sub> lasers also use auxiliary water to cool the tooth or bone and remove debris from the ablation site, and it is vital to effective use of this modality. Water is not used with continuous-wave CO<sub>2</sub> lasers since it would absorb all the energy in the scenario discussed in the second paragraph of this section.

#### 4.2.9 Auxiliary Air

Auxiliary air is also provided by all erbium lasers. However, its value and effect are not as well understood. Certainly, air may help dissipate accumulated thermal energy, but its primary function is to remove debris from the path of the laser energy.

#### 4.2.10 Speed of Movement

Energy is being dispensed as long as the laser is activated. The speed with which the tip is moved across the target tissue will dictate how much energy strikes each individual cell. Moving too quickly will not allow sufficient energy absorption to initiate ablation, and the tissue will be detrimentally heated. Moving too slowly or maintaining the tip in one position will ablate the tissue but also causing overheating. A method of calculating energy density with movement is illustrated in ■ Fig. 4.18.

### 4.3 Further Calculated Parameters

Having adjusted laser parameters in accordance with an established protocol, it is instructive to understand the nature of the resulting energy applied to the target. Average power, peak power, fluence (energy density), and irradiance (power density) all have a very significant effect on the target tissue. Further parameters are also important.

#### 4.3.1 Spot Area at the Tissue

The spot area at the tissue surface must be determined before fluence and irradiance can be calculated. Incorporating the effect of beam divergence and tip-to-tissue distance can often make this a confusing calculation. Further confusion arises regarding even the definition of spot size.

In physics and engineering, spot size refers very specifically to the radius of the laser beam [14], while in many medical publications, it is used to describe the beam diameter or even the area. While the exact definition is of minor importance, knowledge of the author's intent is critical. It is preferable to use the more specific terms of "spot diameter at tissue" and "spot area at tissue."

#### 4.3.2 Average Power Density

Average power density is determined by dividing the average power by the spot area at the tissue surface.

Since spot area is generally very small as shown in ► Sect. 4.2.6 above, average power densities can be surprisingly large.

### 4.3.3 Peak Power Density

Peak power density is simply determined by dividing the peak power by the spot area at the tissue surface. Once again with a small spot area, it is possible to get a stunning 4,000,000 W/cm<sup>2</sup> as shown in ► Sect. 4.2.6.

### 4.3.4 Total Energy Applied

The total energy applied to the tissue is simply the average power multiplied by the total number of seconds of treatment.

The discussion in this chapter highlights the importance of understanding and controlling the exact nature of the parameters selected in any clinical protocol in order to effectively interact with either hard or soft tissue. While the necessary calculations are not complicated, they need to be precise. Appendix (Sect. 4.7) assembles all relevant information and provides calculations based on the settings shown in ■ Figs. 4.5, 4.6, and 4.7.

## 4.4 Hard Tissue Considerations

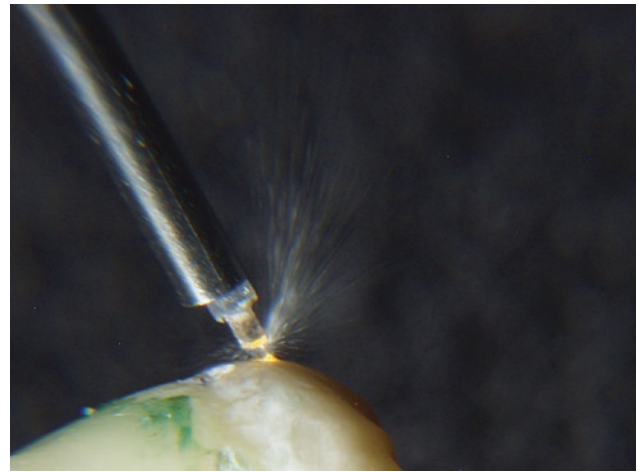
### 4.4.1 Most Appropriate Laser for Hard Tissue Procedures

In the past, erbium lasers were the only devices appropriate for ablation of hard tissues. In recent years, super-pulsed carbon dioxide lasers have been developed that are capable of ablating dentin, enamel, and bone effectively and safely. In addition, research has shown that almost any wavelength of laser energy can be used if the pulse width is short enough and the peak power high enough. The next decade will, no doubt, see the development of practical nanosecond and femtosecond pulsed devices.

### 4.4.2 Mechanism of Hard Tissue Ablation

#### Mechanism of Ablation for Erbium Lasers

While a hydrokinetic theory was originally proposed to explain tissue interaction with Er,Cr:YSGG laser energy, a different mechanism is now most widely accepted [15–17]. All erbium lasers interact with tissue in a similar fashion. Ablation of enamel, dentin, and bone occurs through the explosive removal of tissue in a thermome-



■ Fig. 4.19 Ultra-speed microphotography of Er:YAG laser energy interaction with an enamel surface

chanical event. Because laser energy at erbium wavelengths is very highly absorbed by water, molecules in the target tissue are rapidly superheated [18]. When the steam pressure within the tissue exceeds the structural strength of the overlying material, micro-explosions occur, ejecting particles of fractured material as shown in ■ Fig. 4.19. This explosive phenomenon is initiated long before melting of the carbonated hydroxyapatite occurs.

Considerable pressure is needed to fracture enamel and dentin, so temperatures much higher than the intuitive 100 °C must be generated. A study suggests that ablation actually occurs at superheated temperatures of approximately 600 °C for enamel and 500 °C for dentin [19].

#### Mechanism of Ablation for Carbon Dioxide Lasers

Ablation at CO<sub>2</sub> wavelengths is predominantly a photothermal phenomenon [19]. Laser energy is directly absorbed by the carbonated hydroxyapatite. Melting of the mineral is nearly instantaneous, occurring in less than a picosecond, and is followed rapidly, with increased temperature, by vaporization and expulsion of molten mineral droplets [20]. This liquid mineral quickly resolidifies into characteristic globules routinely seen in SEMs. Lesser absorption by water and protein at these wavelengths only has a minor effect on the ablation process.

If laser energy at CO<sub>2</sub> wavelengths is applied in a continuous-wave mode, the principal interaction consists of melting of carbonated hydroxyapatite and massive transfer of thermal energy to the surrounding tooth structure. Charring and pulp death are the inevitable outcomes. However, applying this energy in super-pulsed mode with very short pulses, ablation as described above

still occurs, but most of the thermal energy is ejected with the molten carbonated hydroxyapatite. Water spray directly removes further amounts of diffused energy.

### Mechanism of Ablation for Femtosecond Lasers

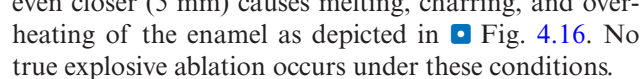
Femtosecond lasers emit energy in extremely short pulses with durations of less than a nanosecond. Even with pulse energy as low as 20  $\mu\text{J}/\text{pulse}$ , a 400 fs pulse width produces peak power of 50 million W. This is sufficient to ionize carbonated hydroxyapatite directly to plasma. The plasma then quickly dissipates, carrying virtually all energy away and transferring almost none to surrounding tissues. Residual thermal energy and collateral damage are almost nonexistent. The wavelength of the applied laser energy is of little consequence as a study of efficient enamel ablation using a 1025 nm laser shows [21]. Ablation simply relies on plasma formation. It should be noted that this wavelength is nearly identical to that of the Nd:YAG laser (1064 nm). Peak ablation efficiency occurs at approximately 5  $\text{J}/\text{cm}^2$ . Clinical femtosecond lasers are not currently available in dentistry but show great promise.

#### 4.4.3 Role of Hydroxyapatite in Ablation

To further clarify the discussion above, carbonated hydroxyapatite is a major absorber of carbon dioxide laser energy and is the primary vehicle of ablation. It absorbs energy, injecting heat into the enamel or dentin, promoting dehydration and structural change. If sufficient energy is absorbed to raise the temperature to about 1280  $^{\circ}\text{C}$ , melting followed by vaporization of the enamel will occur. Melting is a short-lived change in physical state, and there is a new solid created upon cooling. This new solid has a modified structure and composition and will interact with subsequent laser energy differently.

This is a completely different phenomenon than that seen with erbium lasers where water is the primary vehicle of ablation. With these lasers, ablation is the explosive removal of tissue using thermal energy to vaporize internal water. In either case, if the enamel temperature is sufficiently elevated, some of the thermal energy will diffuse into the interior of the tooth compromising pulp vitality. Femtosecond lasers have the potential to nearly eliminate this problem.

A simple experiment illustrates this point. Directing erbium laser energy at a dried extracted tooth with no water spray using 240 mJ at 25 pps (commonly recommended parameters for enamel ablation) from a distance of 10 mm results in sub-ablative irradiation. After about 10 s, the tooth becomes too hot to hold having reached about 200  $^{\circ}\text{C}$ . No ablation has occurred, but a signifi-

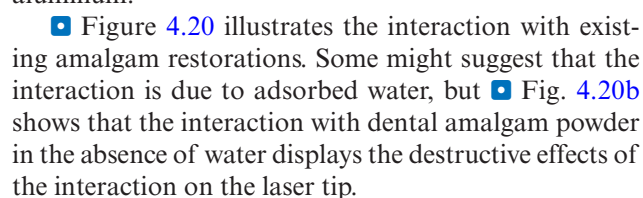
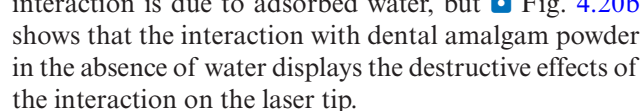
cant amount of energy has been absorbed [18]. Moving even closer (5 mm) causes melting, charring, and overheating of the enamel as depicted in  Fig. 4.16. No true explosive ablation occurs under these conditions.

#### 4.4.4 Cautionary Considerations

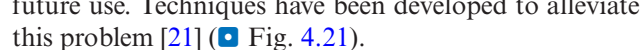
##### Interaction with Non-biological Materials in the Oral Cavity

In the course of applying clinical lasers to intraoral structures, a number of non-biological restorative materials will be encountered. Each of these may lead to reflection, absorption, scattering, refraction, or transmission of energy. Research on the potential negative results of accidental or intentional interaction has been sparse.

Many publications have asserted that laser energy is reflected off the surface of metal restorations and so is of little consequence except to use caution with reflective surfaces. In fact, laser energy of most wavelengths is absorbed by metals. Lasers are routinely used in industrial situations to both cut and weld steel, copper, and aluminum.

 Figure 4.20 illustrates the interaction with existing amalgam restorations. Some might suggest that the interaction is due to adsorbed water, but  Fig. 4.20b shows that the interaction with dental amalgam powder in the absence of water displays the destructive effects of the interaction on the laser tip.

All of these photographs represent the effect of commonly used laser parameters. This suggests that these negative side effects will likely occur any time restorative materials are inadvertently exposed to laser irradiation.

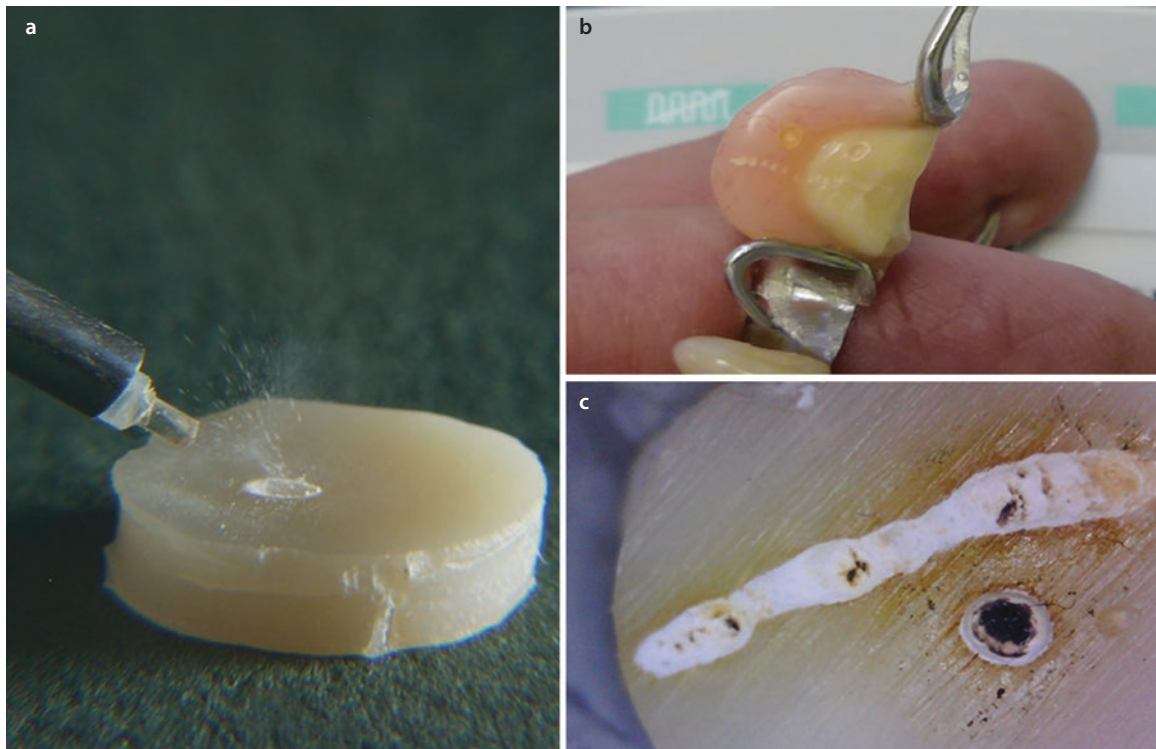
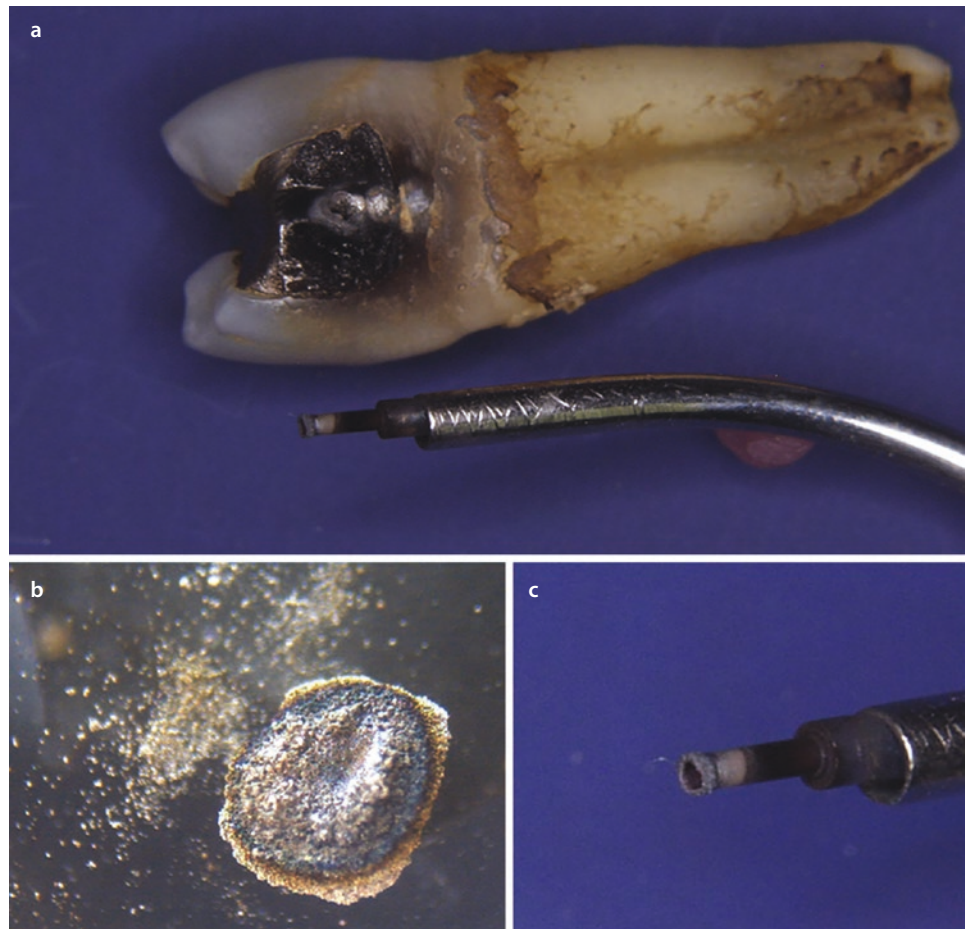
Laser tips and fibers can become coated with exploded or melted debris making them ineffective for future use. Techniques have been developed to alleviate this problem [21] ( Fig. 4.21).

##### Other Considerations

As discussed earlier in this chapter, applying erbium energy to enamel, dentin, or bone without adequate irrigation will almost always be disastrous. The significant thermal energy absorbed by the target will cause irreversible changes. A minimum of 8 mL/min is the recommended level.

Majaron [22] showed that maintaining a gap of 0.3–0.7 mm between the tip and tissue increased enamel ablation significantly. Theoretically, this would allow more water in to flush accumulated debris, allow debris to exit the gap, and remove thermal energy that had been absorbed by the hydroxyapatite. As the distance exceeds 1 mm, decreases in energy density as well as absorption and refraction from interposed water will minimize these advantages.

**Fig. 4.20** **a** The erbium laser energy produces a significant interaction with amalgam restorations. **b** Interaction with amalgam powder suggests that it is not interaction with adsorbed water. Any such interaction will severely damage the laser tip. **c** An erbium laser was applied at 240 mJ and 25 pulses per second



**Fig. 4.21** Energy from both erbium lasers **a** and **b** as well as diode lasers **c** interacts with composite resins and dental acrylic materials at commonly used power levels

## 4.5 Soft Tissue Surgery Considerations

### 4.5.1 Most Appropriate Laser for Soft Tissue Procedures

All of the existing dental lasers can surgically incise soft tissue.

The diode laser is the most popular dental laser with thousands in use in dental offices around the world. They are attractive because they are small, lightweight, low maintenance, durable, and relatively inexpensive.

While a valuable tool, these devices are plagued by a very significant shortcoming. They are very poorly absorbed in soft tissue. With diode lasers ranging from approximately 808 to 1064 nm, melanin, hemoglobin, and protein are the primary chromophores. However, water which makes up more than 70% of most tissues is not a chromophore and produces almost no interaction at diode wavelengths. This suggests that the diode is a poor laser for incising soft tissue.

Since the laser energy spreads through a large area of tissue, the risk of significant collateral damage is high. On the other hand, this energy is highly absorbed in blood so causes excellent coagulation and hemostasis.

The Nd:YAG laser at 1064 nm suffers from the same poor absorption as all diode lasers, again suggesting that this is not a good laser for incising soft tissue. Its advantage lies in its very high peak power with short pulse width providing thermal containment and moderate ablation with very good hemostasis. Nd:YAG lasers are more complex and considerably more expensive than diode lasers.

Erbium lasers have a primary chromophore of water, which is the most abundant constituent of biological tissue making them an excellent choice to perform surgical

soft tissue procedures. Both the Er:YAG and the Er,Cr:YSGG lasers are absorbed efficiently in a few microns of tissue. Since little energy is transferred to the surrounding area, little collateral tissue damage occurs. Because of this same property, these lasers do not produce significant hemostasis from thermal absorption at the incision margins.

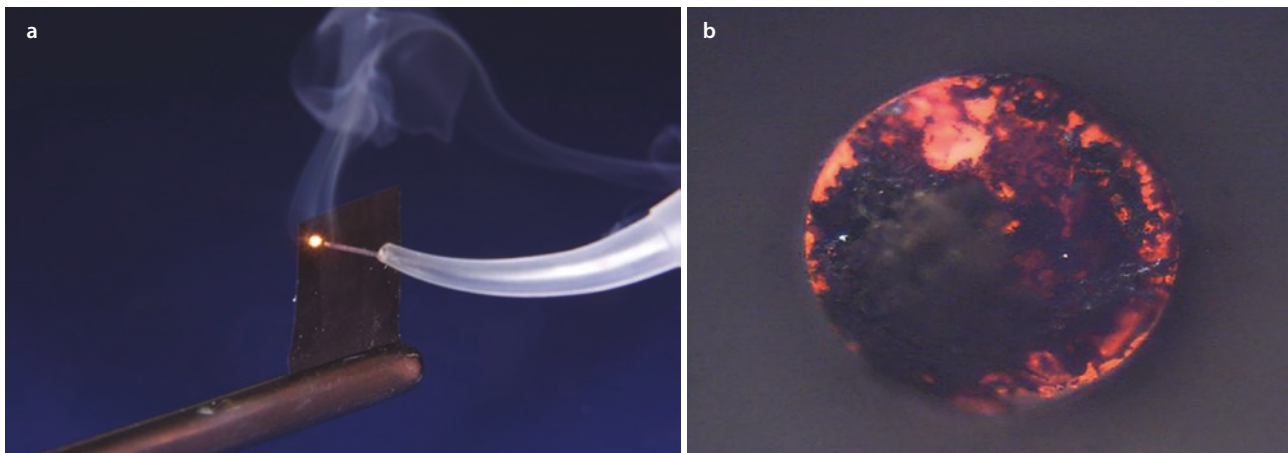
Finally, the CO<sub>2</sub> laser has primary chromophores of water and collagen, making it a useful soft tissue laser. Its action is very superficial but produces melting of collagen as well as explosion of tissue through steam generation. Since this energy is delivered via a noncontact handpiece, precision is more difficult to achieve.

### 4.5.2 Fiber Initiation Consideration

While diode lasers are poor soft tissue lasers, they are small and relatively inexpensive. The optical fiber delivery system is flexible and very maneuverable allowing access to hard-to-reach areas of the oral cavity. Fiber initiation provides a technique to circumvent this laser's shortcomings.

An energy-absorbing material is applied to the end of the fiber as shown in [Fig. 4.22](#) in a process called initiation or activation. Material is transferred to the fiber end creating a thin layer that absorbs subsequent laser energy.

When the laser is activated, this thin layer absorbs the emitted photons and is rapidly heated to several hundred degrees. This "hot tip" is used to melt proteins, thus separating the tissue. The argument is often made that this technique is similar to electrosurgery or applying a heated instrument. However, because only the thin layer of absorbent material is heated, the thermal energy is concentrated and there is significantly less collateral thermal damage than with these other modalities.



**Fig. 4.22** The technical process of initiation is shown in **a** while the resulting coat on the fiber end is shown in **b**

While each laser manufacturer has suggestions for proper fiber initiation, the following technique has been shown to provide superior results:

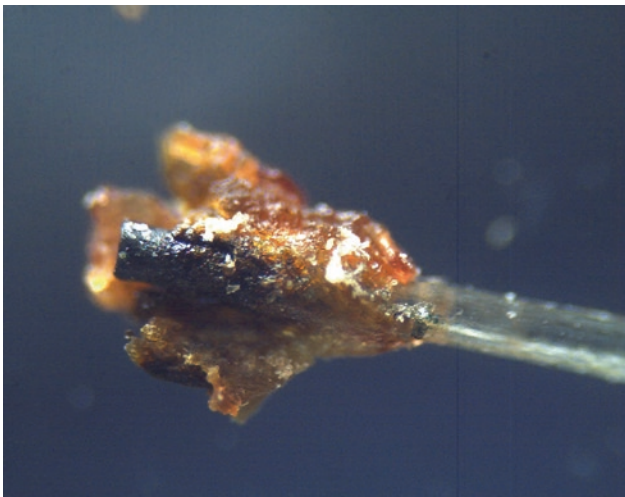
- Adjust the laser to a very low power setting of less than 0.5 W.
- Touch the fiber tip to articulating paper without activating the laser.
- Activate the laser momentarily until a perforation is seen in the articulating paper.
- Repeat at least eight times.
- Observe the aiming beam against a surface. If it is still clearly visible, repeat more times.

#### 4.5.3 Effect of Fiber Size

As has been discussed in ▶ Sect. 4.2.6 above and ■ Table 4.2, smaller fiber diameters significantly increase power density and positively affect the ability of the laser to incise tissue. However, fibers with a diameter smaller than 400  $\mu\text{m}$  are very flexible affecting the ability to cut precisely and are at higher risk of spontaneous fracture.

#### 4.5.4 Effect of Debris Accumulation

In the course of clinical treatment, cellular debris adheres to the warm fiber tip as shown in ■ Fig. 4.23. This material absorbs laser energy and is progressively heated, increasing the risk of collateral thermal damage. It is prudent to remove this debris repeatedly during a surgical procedure. This is best accomplished with wet gauze.



■ Fig. 4.23 During clinical application, tissue adheres to the fiber tip. It in turn is heated causing collateral tissue damage

#### 4.5.5 Precooling of Tissue

Collateral damage occurs through the prolonged heating of cells. A study by Simanovskii et al. [23] asserted that mammalian cells can survive a temperature increase to 42–47 °C for prolonged periods of time. Above this point, the length of time at higher temperatures becomes a critical variable. Cells can only survive at 70 °C for about 1 s and 130 °C for 300  $\mu\text{s}$ . Since optical fiber tip temperature can exceed 800 °C, it becomes apparent that this is the source of damage.

The margins of a surgical excision will be rapidly heated although not to this level. It has been shown that approximately 350  $\mu\text{m}$  of collateral tissue death may occur [24]. Precooling the tissue with an air/water spray can reduce, but certainly not eliminate, collateral damage. Frequent cooling during procedures is also considered to be of benefit. However, using a constant flow of water during surgery is not recommended. Vision will be compromised, and, more importantly, thermal energy will be diffused over the tissue surface scalding a superficial layer as shown in ■ Fig. 4.24.

#### 4.5.6 Cautionary Considerations

The preceding discussions illustrate the need for careful attention to tissue response during even the most rudimentary procedure.



■ Fig. 4.24 Three cuts all performed with a 940 nm diode laser at 1 W, CW. The bottom cut done with no precooling and air or water irrigation. Note the significant collateral damage. The middle cut was performed with air directed over the surface. Speed of cut was slower but as effective with much less collateral damage. Applying an air/water spray during treatment in the top cut caused significant surface cell death. (Courtesy of Dr. Mark Cronshaw)

## 4.6 Photobiomodulation Considerations

It is well established that, in general, photobiomodulation can have a positive effect on tissue [25–27]. However, photonic energy must reach target cells at the appropriate intensity to be effective. This energy density is widely accepted to be between 3 and 10 J/cm<sup>2</sup>, relative to the individual cell surface, in accordance with the Arndt-Schultz law [28, 29]. At less than 3 J/cm<sup>2</sup>, there is insufficient energy absorbed by the cell to increase its metabolism, while densities of more than 10 J/cm<sup>2</sup> appear to inhibit cellular function.

While energy is applied on a macroscopic level, the interaction with tissue occurs on a subcellular level [30]. Ideally, each individual cell in the path of the applied energy should be illuminated with the same number of photons within this range. The positive effects of photobiomodulation are very strongly associated with the density of photons striking the cells. Unfortunately, a method of delivering photons to a group of individual cells, often deep within a tissue mass, in a uniform and predictable manner has been lacking.

### 4.6.1 Most Appropriate Laser for Photobiomodulation

Between approximately 600 and 1200 nm, very few absorbing chromophores exist allowing laser energy to penetrate deeply into the tissue mass. This range of wavelengths has been labeled the “optical window” [31] and is exploited in most photobiomodulation protocols. In general, diode lasers are the most appropriate laser for this application.

### 4.6.2 Recommended Parameters

Further chapters will explore the intricacies of dosage in photobiomodulation. The energy density at the cellular level is a key determinant of biostimulation and appears to be between 3 and 10 J/cm<sup>2</sup> depending on the cell type and should be applied at a rate of approximately 100 mW/cm<sup>2</sup>.

For lesions on the surface of tissue, calculation of the appropriate dose is rudimentary. However, as energy enters tissue, it is moderately absorbed and highly scattered reducing the effective dose.

### 4.6.3 Distribution of Applied Energy in a Tissue Volume

As energy penetrates a tissue mass, it is absorbed and widely scattered. Bashkatov [32] asserted that photons are 100 times more likely to be scattered than to be

absorbed. When treating a muscle or a temporomandibular joint, energy reaching the target cells is predictably and drastically reduced. ■ Figure 4.25 is a presentation of energy reaching cells deep within the tissue.

### 4.6.4 Effect of Lesion Depth and Tissue Type on Parameter Calculations

Many of the anatomic structures that are purported to benefit from low-level laser therapy such as the TMJ or joint bursa are located at some distance within the surrounding tissue. ■ Figure 4.26 suggests that, at 5 mm into representative tissue ranging from beef muscle (similar to human muscle) to beef liver (similar to highly vascularized tissue), energy density has decreased to about 10% of its surface value.

### 4.6.5 Cautionary Considerations

Arany et al. [33] state that a skin temperature increase to greater than 45 °C is phototoxic. While it is tempting to apply large amounts of energy in a short period of time in order to stimulate deep structures and to speed treatment, very negative unintended consequences may occur in overlying tissues.

Of all the protocols for laser use in dentistry, photobiomodulation is at the greatest risk of misunderstood technique and unintended consequences. Unfortunately, there is no direct visual feedback of effectiveness leaving the clinician to rely on an understanding of parameters.

While not settled science, the currently accepted guidelines recommend applying a fluence of 100 mW/cm<sup>2</sup> and a total cellular dose of 3–10 J/cm<sup>2</sup>.

There are several unique devices available for applying therapeutic doses including large-diameter (8–10 mm) optical fibers, rectangular therapy/bleaching handpieces, and flattop handpieces. Many clinicians use the expedient of a surgical optical fiber held at a distance from the tissue surface as a perceived equivalent. However, these devices can produce very different tissue exposure.

If 1 W of power is applied through a 400 μm fiber from a distance of 10 mm for 30 s as it is moved over a tissue area of 2 cm<sup>2</sup>, the result is:

$$\text{Fluence} = 1 \text{ W} / 2 \text{ cm}^2 = 0.5 \text{ W} / \text{cm}^2 = 500 \text{ mW} / \text{cm}^2$$

$$\text{Irradiance} = 0.5 \text{ W} / \text{cm}^2 \times 30 \text{ s} = 15 \text{ J} / \text{cm}^2$$

Applying the same parameters through an 8 mm diameter PBM handpiece in contact without movement results in:

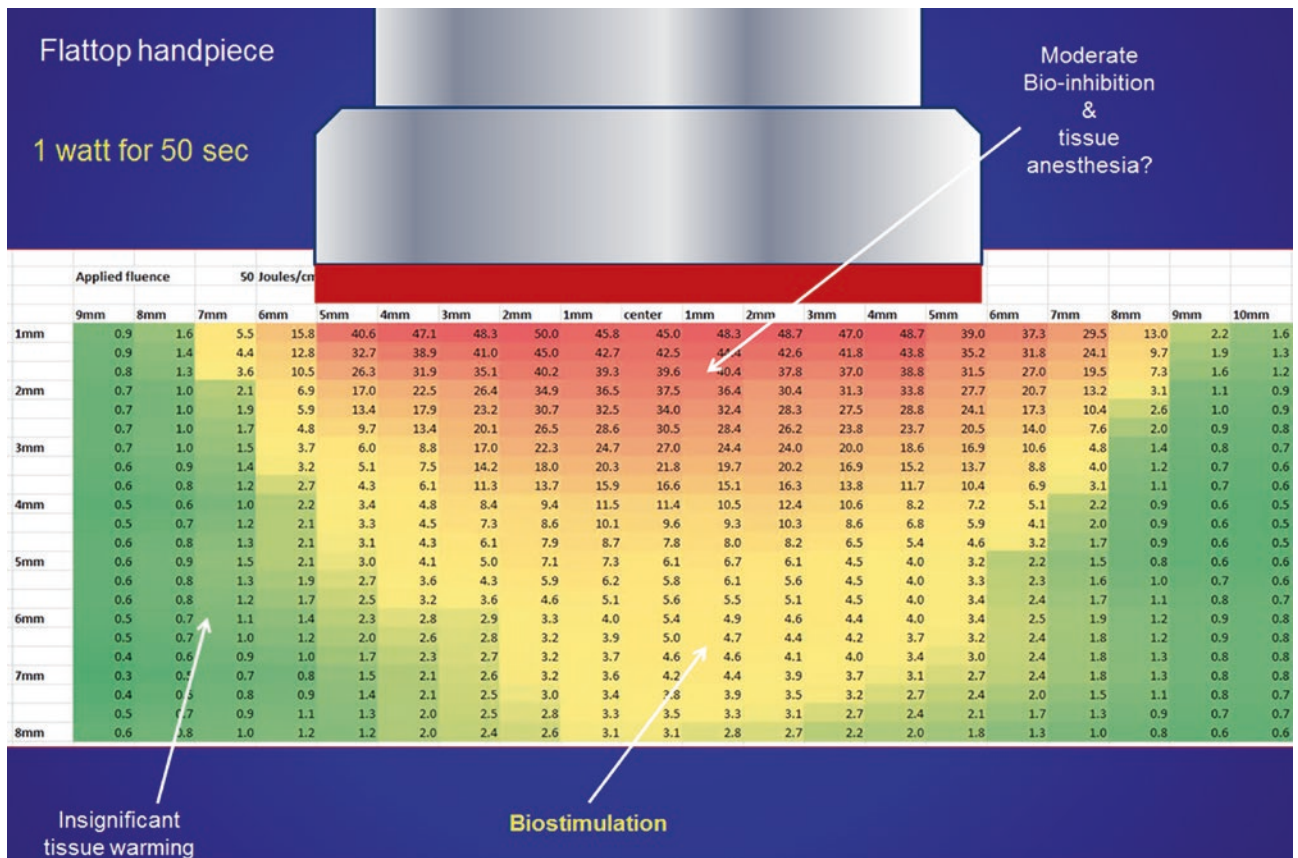
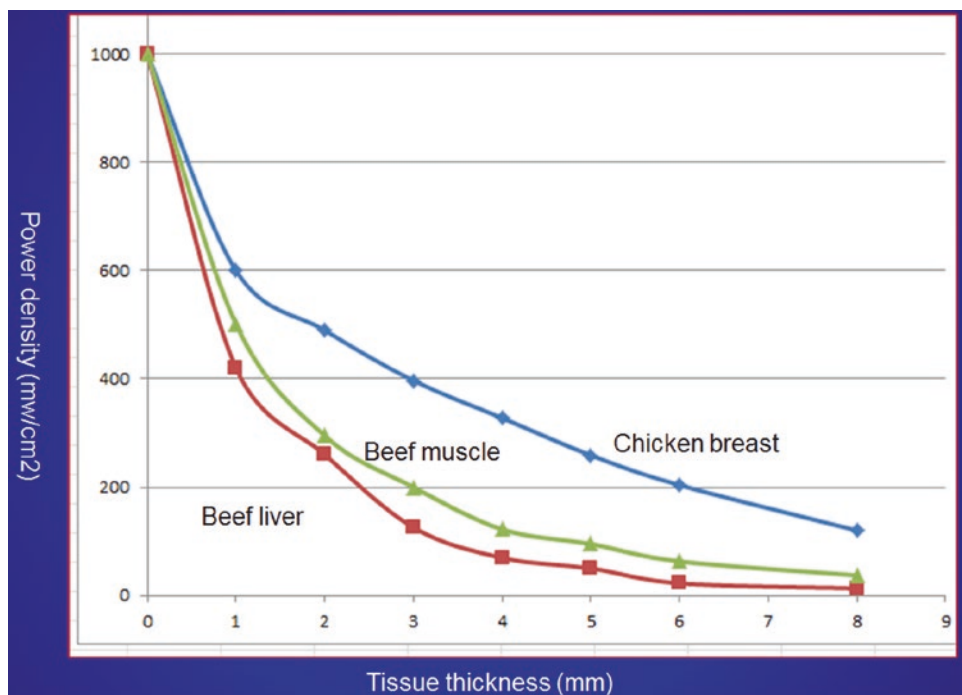


Fig. 4.25 All tissue areas in red exceed the optimal range and experience bioinhibition and perhaps even irreversible cellular damage. This zone extends about 3.5 mm into the tissue. Biostimulation would occur from 10 J/cm<sup>2</sup> to about 3 J/cm<sup>2</sup> as depicted in yellow.

This semicircular band is about 4 mm wide directly in front of the handpiece tapering down to about 1 mm wide on the periphery. Beyond that band, the tissue cells would be perhaps warmed a little bit with no discernable photobiomodulation effect

Fig. 4.26 Power density diminishes rapidly as the laser energy penetrates the tissue. Beef muscle and liver are representative of the range of tissues seen in the oral cavity





$$\text{Fluence} = 1 \text{ W} / 0.5 \text{ cm}^2 = 2 \text{ W} / \text{cm}^2 = 2000 \text{ mW} / \text{cm}^2$$

$$\text{Irradiance} = 2 \text{ W} / \text{cm}^2 \times 30 \text{ s} = 60 \text{ J} / \text{cm}^2$$

Finally, applying the same parameters through a rectangular therapy/bleaching handpiece with dimensions of 8 mm by 35 mm results in:

$$\text{Fluence} = 1 \text{ W} / 2.8 \text{ cm}^2 = 0.36 \text{ W} / \text{cm}^2 = 360 \text{ mW} / \text{cm}^2$$

$$\text{Irradiance} = 0.36 \text{ W} / \text{cm}^2 \times 30 \text{ s} = 10.7 \text{ J} / \text{cm}^2$$

It is easy to understand that, while using identical laser parameters, each of these common delivery systems will have a vastly different effect on the target tissue.

Whichever means of energy delivery is chosen, it is incumbent on the clinician to carefully calculate the tissue dose and to adjust parameters appropriately. To achieve the recommended protocol:

- The 400  $\mu\text{m}$  fiber should be applied at 0.2 W for 50 s from a distance of 10 mm.
- The PBM handpiece should be applied at 0.05 W (an impossibility) for 100 s.
- The therapy handpiece should be applied at 0.3 W for 90 s.

## Summary

Knowledge of actual laser energy being applied to the target tissue with any laser is vital to achieving positive results. This chapter has focused on understanding the meaning and the effect of each of the variable parameters available to the clinician when making a decision to apply laser technology for the benefit of a patient.

■ **Fig. 4.27** This 810 nm diode laser has limited variables that may be controlled directly. It is used here with a 400  $\mu\text{m}$  diameter optical fiber



## 4.7 Appendix – Laser Photonic Energy – Mathematical Quantification and Calculation

Calculation of parameters is provided for each of the lasers shown below. While some readers may find this rudimentary, others will appreciate the review of physics and mathematical calculations.

Please note that, for calculations to be valid, it is necessary to convert all dimensions to centimeters, all times to seconds, all energies to joules, and all powers to watts before computation. This standardized system allows the statement of parameters as  $\text{W}/\text{cm}^2$ , etc.

Because of basic differences in variable-gated continuous-wave and true pulsed lasers, each needs a different set of equations to calculate the same set of variables (■ Fig. 4.27).

### 4.7.1 Variable Gated Continuous Wave Laser (■ Fig. 4.27)

#### Emission Cycle

If the laser is “on” for 20 ms and “off” for 80 ms, then it is on for 20% of the time:

$$\begin{aligned} \text{Percentage of "on - time"} &= 20 \text{ ms} / (20 \text{ ms} + 80 \text{ ms}) \\ &= 20 \text{ ms} / 100 \text{ ms} = 0.2 = 20\% \end{aligned}$$

### Average Power

This diode laser allows direct control of the output power, but the power displayed is the “continuous wave” output rather than the average power. Any time that this laser is emitting photons, it is emitting 2 W. Average power is calculated as:

$$\begin{aligned} \text{Power}_{\text{avg}} &= \text{power}_{\text{indicated}} \times \text{percentage of "on - time"} \\ &= 2 \text{ W} \times (20 \text{ ms} / (20 \text{ ms} + 80 \text{ ms})) = 0.4 \text{ W} \end{aligned}$$

If the lengths of the on and off periods are changed but stay in the same ratio, average power will be the same.

$$\begin{aligned} \text{Power}_{\text{avg}} &= \text{power}_{\text{indicated}} \times \text{percentage of "on - time"} \\ &= 2 \text{ W} \times (80 \text{ ms} / (80 \text{ ms} + 320 \text{ ms})) = 0.4 \text{ W} \end{aligned}$$

### Peak Power

For this diode laser, peak power will always be the same as that indicated on the display, no matter what magnitude or ratio of “on” and “off” times is selected.

$$\text{Power}_{\text{peak}} = \text{power}_{\text{indicated}} = 2 \text{ W}$$

### Pulse Width

Pulse width, in this case, is simply the “on” time:

$$\text{Pulse width} = 20 \text{ ms}$$

### Pulse Repetition Rate

One repetition is the same as one emission cycle or the sum of one “on” time and one “off” time. So:

$$\text{Emission cycle} = 20 \text{ ms ("on")} + 80 \text{ ms ("off")} = 100 \text{ ms}$$

Then, the number of these events that occur in 1 s represents the pulse repetition rate.

$$\text{Pulse repetition rate} = 1 \text{ s} / 100 \text{ ms} = 10 \text{ pulses per second}$$

### Beam Divergence

Beam divergence is a function of the fiber construction and numerical aperture. For optical fibers used with diode and Nd:YAG lasers, the numerical aperture is commonly 0.22. This results in a divergence of 12.7° per side angle.

### Tip Area

Tip area is the radius squared of the tip multiplied by  $\pi$ :

$$\begin{aligned} \text{Tip area} &= 200 \mu\text{m} \times 200 \mu\text{m} \times 3.14159 \\ &= 0.02 \text{ cm} \times 0.02 \text{ cm} \times 3.14159 = 0.00126 \text{ cm}^2 \end{aligned}$$

### Spot Diameter at Tissue

Diode lasers are used in contact for surgical procedures. Therefore, the spot diameter at the tissue surface is the same as the tip diameter:

$$\text{Spot diameter at tissue} = 400 \mu\text{m} = 0.04 \text{ cm}$$

### Spot Area at Tissue

Similarly, the spot area at the tissue surface is the same as the area of the optical fiber tip.

$$\text{Spot area at tissue} = 0.00126 \text{ cm}^2$$

### Power Density

Power density is the total energy in joules passing through a specific area in 1 s of time. With the attached 400  $\mu\text{m}$  laser fiber, this diode laser produces:

$$\begin{aligned} \text{Power density}_{\text{avg}} &= \text{power}_{\text{avg}} / \text{area} = 0.4 \text{ W} / 0.00126 \text{ cm}^2 \\ &= 318 \text{ W} / \text{cm}^2 \end{aligned}$$

$$\begin{aligned} \text{Power density}_{\text{peak}} &= \text{power}_{\text{peak}} / \text{area} = 2 \text{ W} / 0.00126 \text{ cm}^2 \\ &= 1587 \text{ W} / \text{cm}^2 \end{aligned}$$

### Total Energy

The total amount of energy delivered to the tissue during a 30-s treatment is:

$$\text{Total energy} = \text{power}_{\text{avg}} \times \text{time} = 0.4 \text{ W} \times 30 \text{ s} = 12 \text{ J}$$

### Energy Density

If the fiber tip is held stationary in one location, the energy density is calculated as:

$$\begin{aligned} \text{Energy density}_{\text{avg}} &= \text{power}_{\text{avg}} \times \text{time} / \text{area} \\ &= 0.4 \text{ W} \times 30 \text{ s} / 0.00126 \text{ cm}^2 \\ &= 9540 \text{ J} / \text{cm}^2 \end{aligned}$$

$$\begin{aligned} \text{Energy density}_{\text{peak}} &= \text{power}_{\text{peak}} \times \text{time} / \text{area} \\ &= 2 \text{ W} \times 30 \text{ s} / 0.00126 \text{ cm}^2 \\ &= 47,610 \text{ J} / \text{cm}^2 \end{aligned}$$

### Movement of the Tip

If the fiber tip is moved through the tissue at a speed of 2 mm/s, the applied energy is distributed over a much larger area affecting energy density. This may be calculated as shown below (■ Fig. 4.28):

Energy density with movement:

$$\text{Area} = \text{width} \times \text{length traveled in 30s}$$

$$\text{Area} = 0.04 \text{ cm} \times 6 \text{ cm} = 0.24 \text{ cm}^2$$

$$\text{Energy density} = 0.4 \text{ W} \times 30 \text{ s} / 0.24 \text{ cm}^2 = 50 \text{ J} / \text{cm}^2$$

#### 4.7.2 True Pulsed Laser with Control of Average Power (■ Fig. 4.30)

##### Average Power

This erbium laser allows direct control of the average output power. In this case, it has been adjusted to 4 W.

##### Pulse Width

Pulse width is established by choosing one of the two values available—"H" = "hard tissue" = 60  $\mu\text{s}$  or "S" = "soft tissue" = 700  $\mu\text{s}$ . In this case, 60  $\mu\text{s}$  has been selected.

##### Pulse Repetition Rate

Pulse repetition rate is selected as 15 pulses per second (pps) ■ Fig. 4.29

##### Emission Cycle

The emission cycle indicates the percentage of time that the laser is actually emitting energy. With any true pulsed laser, this is a very small amount. It is calcu-

lated as the length of time that the laser is emitting during each second or the sum of the width of all pulses:

$$\text{Emission cycle} = \left( \frac{\text{number of pulses per second}}{\times \text{width of each pulse} / 1 \text{ s}} \right) \times 100\%$$

$$\begin{aligned} \text{Emission cycle} &= 15 \times 60 \mu\text{s} = 15 \times 0.0006 \text{ s} \\ &= 0.009 \times 100 = 0.9\% \end{aligned}$$

##### Energy Per Pulse

If 4 W is produced in 1 s, then 4 J/s is produced. If this energy is divided equally between 15 pulses, then:

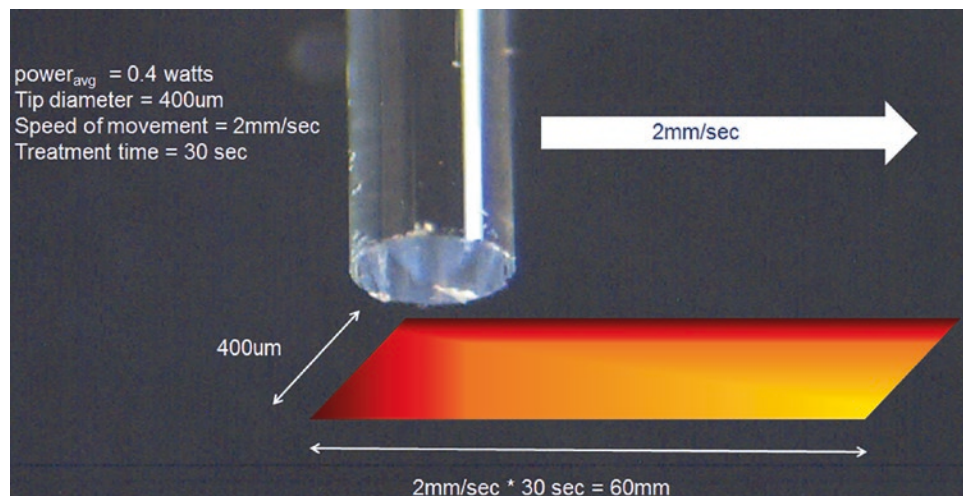
$$\text{Energy per pulse} = \frac{\text{energy per second}}{\text{number of pulses per second}}$$

$$\begin{aligned} \text{Energy per pulse} &= 4 \text{ J} / \text{s} / 15 \text{ pps} \\ &= 0.267 \text{ J} / \text{pulse} \\ &= 267 \text{ mJ} / \text{pulse} \end{aligned}$$

A key point must be considered here. With any pulsed laser that allows control of average power, the energy per pulse changes if the pulse repetition rate changes. As an example, if the laser in ■ Fig. 4.30 has the pulse repetition rate changed to 5 pulses per second, while the average power is unchanged, the energy per pulse changes to:

$$\begin{aligned} \text{Energy per pulse} &= 4 \text{ J} / \text{s} / 5 \text{ pps} \\ &= 0.8 \text{ J} / \text{pulse} \\ &= 800 \text{ mJ} / \text{pulse} \end{aligned}$$

■ Fig. 4.28 When the optical fiber or emitting tip is moved across the tissue during treatment, the energy produced is distributed over a larger area based on the speed of movement. When determining energy density received by an individual group of cells, it is necessary to calculate the entire area that is irradiated



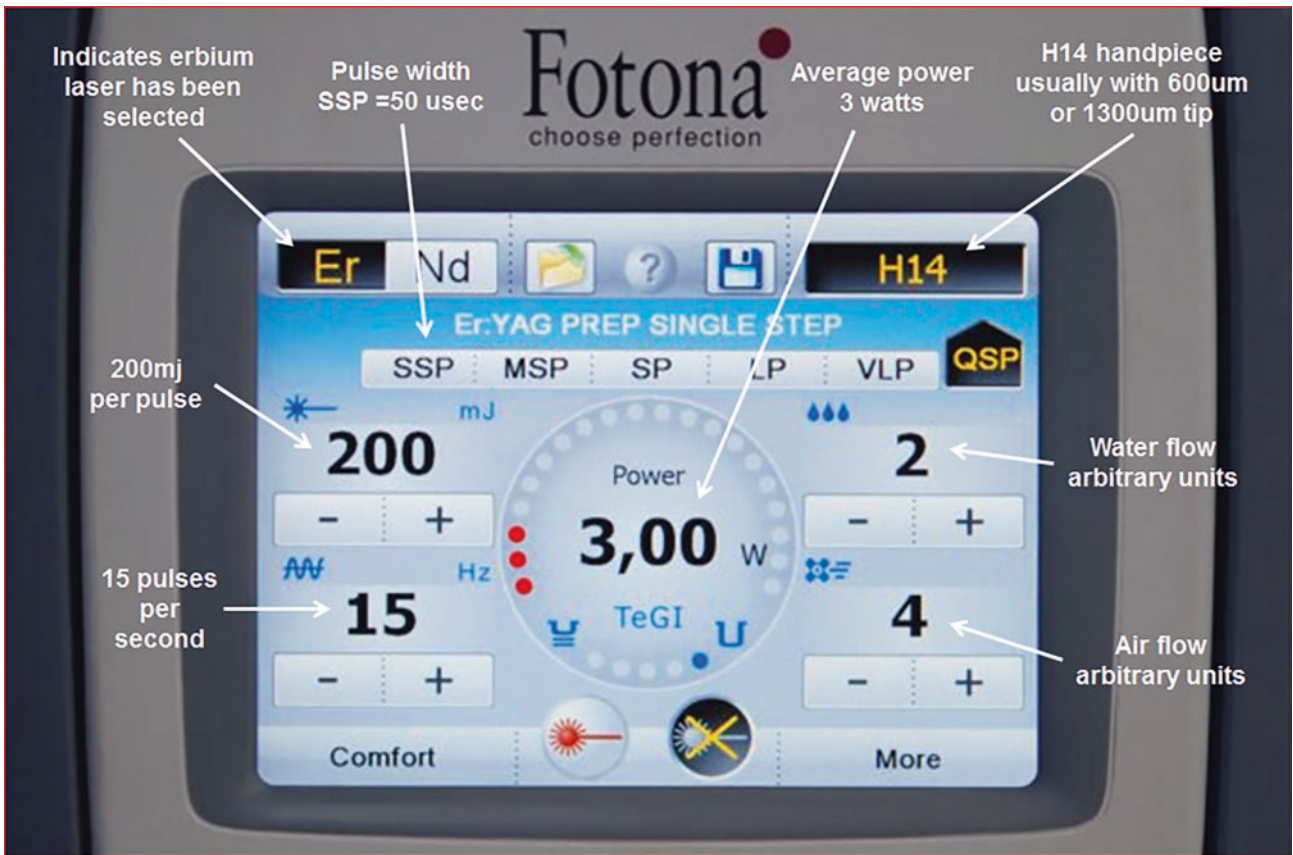


Fig. 4.29 This laser contains both a 2940 nm Er:YAG laser and a 1064 nm Nd:YAG laser. In erbium mode, it allows selection of six different pulse widths. However, the operator must refer to a manual

to find the value. It allows control of pulse energy and pulse frequency. The resulting average power is displayed but not directly selectable

Fig. 4.30 This 2780 nm Er,Cr:YSGG laser has a choice of two pulse widths. It allows control of average power, but energy per pulse and peak power must be calculated. Levels of auxiliary air and water are set as percentage of an arbitrary value



This concept applies to any Er:YAG, Er,Cr:YSGG, or Nd:YAG laser that provides control of average power instead of energy per pulse.

### Peak Power

Peak power represents the maximum instantaneous power that the laser produces.

Peak power can be calculated as:

$$\text{Power}_{\text{peak}} = \text{energy per pulse} / \text{pulse width}$$

$$\text{Power}_{\text{peak}} = 0.267 \text{ J} / 0.00006 \text{ s} = 4444 \text{ W}$$

### Tip Area

Commonly available tips for this laser are cylindrical, tapered, and chisel shaped. While calculating the area, divergence and spot area for chisel tips are beyond this discussion, and the diameter of the output end of a tapered tip can be treated the same as a cylindrical tip as calculated below.

Assuming the use of a 600  $\mu\text{m}$  diameter cylindrical tip allows the following calculations:

Tip area is the radius squared of the tip multiplied by  $\pi$ :

$$\begin{aligned} \text{Tip area} &= 300 \mu\text{m} \times 300 \mu\text{m} \times 3.14159 \\ &= 0.03 \text{ cm} \times 0.03 \text{ cm} \times 3.14159 \\ &= 0.00283 \text{ cm}^2 \end{aligned}$$

A tip that tapers from 1200 to 600  $\mu\text{m}$  at the output end would have the same tip area of 0.00283  $\text{cm}^2$ .

### Spot Diameter at Tissue

Erbium lasers are usually used in contact for soft tissue surgical procedures. Therefore, the spot diameter at the tissue surface is the same as the tip diameter:

$$\text{Spot diameter at tissue} = 600 \mu\text{m} = 0.06 \text{ cm}$$

### Spot Area at Tissue

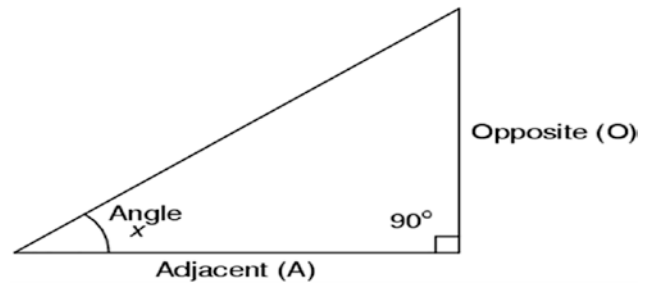
For this scenario, the spot area at the tissue surface is the same as the area of the optical fiber tip.

$$\text{Spot area at tissue} = 0.00126 \text{ cm}^2$$

### Beam Divergence

When this or any laser is used out of contact, the effect of beam divergence has a critical effect on spot diameter at the tissue surface and therefore energy and power density.

Beam divergence is a function of the particular tip chosen. For tips used with this laser, it is commonly 12° per side angle. For a tip-to-tissue distance



■ Fig. 4.31 Basic geometry establishes the relationship between different variables in a right triangle

(O) of 2 mm and an angle (x), from geometry (■ Fig. 4.31):

$$O = A \times \tan(x)$$

$$O = 2 \text{ mm} \times \tan(12^\circ)$$

$$O = 2 \text{ mm} \times 0.2126 = 0.425 \text{ mm}$$

$$\begin{aligned} \text{Spot radius at tissue} &= \text{tip radius} + \text{opposite} \\ &= 0.03 + 0.0425 = 0.0725 \text{ cm} \end{aligned}$$

$$\text{Spot diameter at tissue} = 0.145 \text{ cm}$$

As shown in ■ Fig. 4.32, at a tip-to-tissue distance of 2 mm with a 12° beam divergence per side angle:

$$\begin{aligned} \text{Spot area at tissue} &= \pi \times r^2 \\ &= 3.14159 \times 0.0725 \text{ cm} \times 0.0725 \text{ cm} \\ &= 0.0165 \text{ cm}^2 \end{aligned}$$

The effect of divergence on parameters is similar to that with an optical fiber.

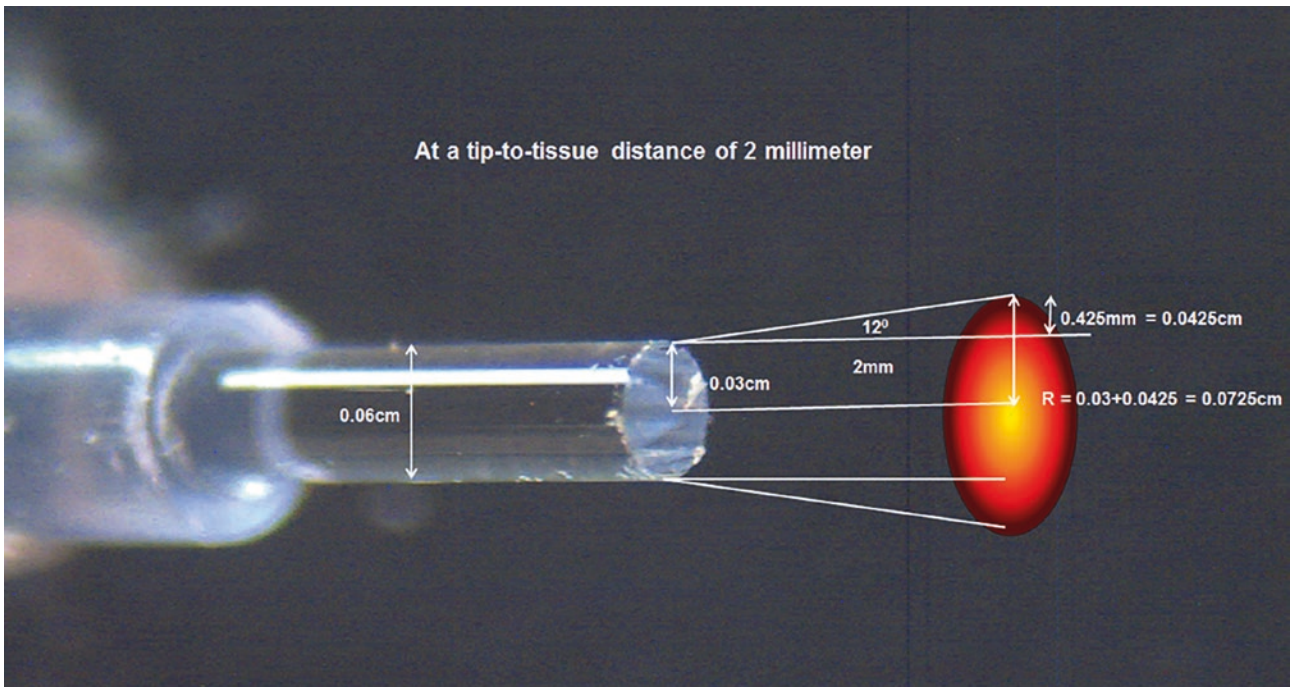
### Power Density

Power density is the total energy in joules passing through a specific area in 1 s of time. With the attached 600  $\mu\text{m}$  laser fiber and a tip-to-tissue distance of 2 mm, this erbium laser produces:

$$\begin{aligned} \text{Power density}_{\text{avg}} &= \text{power}_{\text{avg}} / \text{area} \\ &= 4 \text{ W} / 0.0165 \text{ cm}^2 = 242 \text{ W} / \text{cm}^2 \end{aligned}$$

$$\begin{aligned} \text{Power density}_{\text{peak}} &= \text{power}_{\text{peak}} / \text{area} \\ &= 4444 \text{ W} / 0.0165 \text{ cm}^2 \\ &= 269,333 \text{ W} / \text{cm}^2 \end{aligned}$$

This peak power calculation should make it clear how a pulsed laser can ablate enamel.



■ Fig. 4.32 This is a depiction of the various values calculated to determine the spot area at the tissue with a divergent beam

### Total Energy

The total amount of energy delivered to the tissue during a 30-s treatment is:

$$\text{Total energy} = \text{power}_{\text{avg}} \times \text{time} = 4 \text{ W} \times 30 \text{ s} = 120 \text{ J}$$

### Energy Density

If the fiber tip is held stationary in one location, the energy density is calculated as:

$$\begin{aligned} \text{Energy density}_{\text{avg}} &= \text{power}_{\text{avg}} \times \text{time} / \text{area} \\ &= 4 \text{ W} \times 30 \text{ s} / 0.0165 \text{ cm}^2 \\ &= 7273 \text{ J} / \text{cm}^2 \end{aligned}$$

$$\begin{aligned} \text{Energy density}_{\text{peak}} &= \text{power}_{\text{peak}} \times \text{time} / \text{area} \\ &= 4444 \text{ W} \times 30 \text{ s} / 0.0165 \text{ cm}^2 \\ &= 8,080,000 \text{ J} / \text{cm}^2 \end{aligned}$$

### 4.7.3 True Pulsed Laser with Control of Pulse Energy (■ Fig. 4.29)

#### Pulse Width

Pulse width is established by choosing one of the six values available—choosing the SSP mode selects a pulse width of 50  $\mu\text{s}$ .

### Pulse Repetition Rate

Pulse repetition rate is selected as 15 pulses per second (pps).

### Emission Cycle

The emission cycle indicates the percentage of time that the laser is actually emitting energy. With any true pulsed laser, this is a very small amount. It is calculated as the length of time that the laser is emitting during each second or the sum of the width of all pulses:

$$\text{Emission cycle} = \left( \frac{\text{number of pulses per second}}{\times \text{width of each pulse} / 1 \text{ s}} \right) \times 100\%$$

$$\begin{aligned} \text{Emission cycle} &= 15 \times 50 \mu\text{s} = 15 \times 0.0005 \text{ s} \\ &= 0.0075 \times 100 = 0.75\% \end{aligned}$$

### Energy Per Pulse

The energy per pulse is directly selected as 200 mJ.

### Average Power

This erbium laser indicates an average output power of 3 W. In this case, it is not directly adjustable but is internally calculated by the laser. It can be calculated as shown below:

$$\begin{aligned} \text{Power}_{\text{avg}} &= \text{energy per pulse} \\ &\quad \times \text{number of pulses per second} \end{aligned}$$

$$\text{Power}_{\text{avg}} = 0.2 \text{ J} / \text{pulse} \times 15 \text{ pulses per second} = 3.0 \text{ W}$$

## Peak Power

Peak power represents the maximum instantaneous power that the laser produces.

Peak power can be calculated as:

$$\text{Power}_{\text{peak}} = \text{energy per pulse} / \text{pulse width}$$

$$\text{Power}_{\text{peak}} = 0.200 \text{ J} / 0.00005 \text{ s} = 4000 \text{ W}$$

## Tip Area

As with the Er,Cr:YSGG laser discussed above, commonly available tips for this laser are cylindrical, tapered, and chisel shaped. All calculations below are based on the use of a 600  $\mu\text{m}$  diameter tip at a tip-to-tissue distance of 2 mm. The resulting calculations will be identical to those above.

Assuming the use of a 600  $\mu\text{m}$  diameter cylindrical tip allows the following calculations:

Tip area is the radius squared of the tip multiplied by  $\pi$ :

$$\begin{aligned} \text{Tip area} &= 300 \mu\text{m} \times 300 \mu\text{m} \times 3.14159 \\ &= 0.03 \text{ cm} \times 0.03 \text{ cm} \times 3.14159 = 0.00283 \text{ cm}^2 \end{aligned}$$

A tip that tapers from 1200 to 600  $\mu\text{m}$  at the output end would have the same tip area of 0.00283  $\text{cm}^2$ .

As before, at a tip-to-tissue distance of 2 mm with a 12° beam divergence per side angle:

$$\begin{aligned} \text{Spot area at tissue} &= \pi \times r^2 \\ &= 3.14159 \times 0.0725 \text{ cm} \times 0.0725 \text{ cm} \\ &= 0.0165 \text{ cm}^2 \end{aligned}$$

## Power Density

Power density is the total energy in joules passing through a specific area in 1 s of time. With the attached 600  $\mu\text{m}$  laser fiber and a tip-to-tissue distance of 12°, this erbium laser produces:

$$\begin{aligned} \text{Power density}_{\text{avg}} &= \text{power}_{\text{avg}} / \text{area} \\ &= 3 \text{ W} / 0.0165 \text{ cm}^2 = 182 \text{ W} / \text{cm}^2 \end{aligned}$$

$$\begin{aligned} \text{Power density}_{\text{peak}} &= \text{power}_{\text{peak}} / \text{area} \\ &= 4000 \text{ W} / 0.0165 \text{ cm}^2 \\ &= 242,424 \text{ W} / \text{cm}^2 \end{aligned}$$

## Total Energy

The total amount of energy delivered to the tissue during a 30-s treatment is:

$$\text{Total energy} = \text{power}_{\text{avg}} \times \text{time} = 3 \text{ W} \times 30 \text{ s} = 90 \text{ J}$$

## Energy Density

If the fiber tip is held stationary in one location, the energy density is calculated as:

$$\begin{aligned} \text{Energy density}_{\text{avg}} &= \text{power}_{\text{avg}} \times \text{time} / \text{area} \\ &= 3 \text{ W} \times 30 \text{ s} / 0.0165 \text{ cm}^2 \\ &= 5455 \text{ J} / \text{cm}^2 \end{aligned}$$

$$\begin{aligned} \text{Energy density}_{\text{peak}} &= \text{power}_{\text{peak}} \times \text{time} / \text{area} \\ &= 4000 \text{ W} \times 30 \text{ s} / 0.0165 \text{ cm}^2 \\ &= 8,080,000 \text{ J} / \text{cm}^2 \end{aligned}$$

## Assumptions

Several assumptions are made in these calculations:

- Output is assumed to be uniform over the radiated area. In practice, this is rarely true since a Gaussian output is normally produced.
- Output is assumed to be uniform over time. That is, each pulse is assumed to rise instantaneously from zero to the peak power being produced. Again, this is rarely true.
- Output is assumed to be constant over the duration of the pulse. In fact, as shown in this chapter, [Fig. 4.27](#) of this text, the true output varies significantly and must be averaged for these calculations.
- The output is assumed to be the same as that displayed on the interface screen. Laser manufacturers are careful to point out in their technical specifications that most parameters are only guaranteed to be within about 20% of the stated value.
- Actual spot diameter is as calculated. Because of the Gaussian nature of the beam cross section, a portion of the photons will strike tissue outside of the calculated area.

## Summary

The calculations presented here are based on representative values as shown on actual dental lasers for all parameters. By substituting actual known values for the laser being used into the equations, it is possible to determine appropriate parameters. In general, all continuous-wave lasers can be assessed using the set of calculations in Sect. 4.7.1, while true pulsed lasers can be analyzed using calculations in Sects. 4.7.2 and 4.7.3 depending on which parameters are displayed or can be selected.

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