Laser-Tissue Interaction

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Core Message

The potential for laser-tissue interaction forms the basis of the usefulness of predictable employment of laser photonic energy as an adjunct to clinical dental and oral therapy. Appreciation of the underlying mechanisms together with acknowledgment of limitations will help the clinician to provide laser therapy and minimize collateral damage.

There is an often-cited belief that in order to obtain benefit from laser photonic energy irradiation of target tissue, there must be absorption of the energy. Such understanding has merit but not the entire truth. Owing to the multi-structural nature of oral hard and soft tissue, the possibility of incident photonic energy reacting in a definite, predictable and exclusive manner with target tissue molecules is flawed through the anisotropic nature of the varying structures. Interaction may be a combination of surface, deeper, scattered and refracted energy distribution; true absorption of power values predicated through laser control panel selection may be impossible to achieve because of the varying interactive phenomena that may occur.

All oral tissues are receptive to laser treatment, but the biophysics governing laser-tissue interaction demands a knowledge of all factors involved in delivery of this modality. Through this knowledge, correct and appropriate treatment can be delivered in a predictable manner.

This chapter looks at the concepts of electromagnetic energy distribution within oral hard and soft tissue and examines the potential for true photonic energy ablation of target molecules. Prime concepts of photothermal action as a pathway to tissue change are explained, and adjunctive spatial and temporal components of the incident beam and the effects of such variance are explored.

The inconsistencies of laser-tissue interaction continue to pose some difficulty for the dental clinician; however, the development of many laser machines, amounting to a facility to produce laser photonic energy at several wavelengths between the visible and far-infrared areas of the electromagnetic spectrum, addresses many of the inconsistencies.

3.1 Introduction

Our understanding of the concepts of color helps to define the interaction of an incident beam of multi-wavelength (λ) electromagnetic (EM) energy—so-called white light—with a target structure. Human interpretation of "light" as a concept is limited to the ability of the retina to respond to this energy and the visual cortex to correlate stimulation in terms of a very limited range of the EM spectrum (λ 350–750 nm), termed the "visible spectrum." White light is seen in nature as the consequence of solar energy that filters through the many layers of the earth's atmosphere, or the emission of a man-made incandescent light source. "Light waves" that arise from such sources are multi-directional (not in phase—incoherent), and through the inverse proportionality relationship of wavelength with photonic energy, of consequent multiple energy values.

The fundamental theories on light of the latter nineteenth and early twentieth centuries—notably the works of Maxwell, Planck, Hertz, Einstein and Bohr—provided a coalescence of the prevailing opinions of light being composed of either particles or waves. Newton, through his "corpuscular" theorem [1], in which light traveled as discrete packets ("corpuscles"), was at variance with earlier work of Huygens. Popular acceptance of a predominant belief in light propagation by waves re-emerged in the early eighteenth-century England with the slit experiments of Thomas Young [2]. The confirmation that light energy was a form of electromagnetic (EM) radiation, capable of causing a photoelectric effect with certain metals, proposed a duality of existence for "packets" of light energy. Einstein is attributed [3] with providing the annotation "photon" (one quantum—smallest unit—of electromagnetic energy is called a photon (origin Greek "φως," meaning "light")) and with others listed before, provided that there is an understanding that photonic energy is a form of energized EM radiation, with each photon traveling at the speed of light (approx. 300×10^6 m/s) in a sinusoidal wave pattern. From this, it is fundamental to our understanding of so-called laser-tissue interaction that EM energy in its many forms is interrelated, and the energy contained therein is capable of conversion (subject to incident power value) to thermal and ablative equivalent within target tissue, through the law of conservation of energy (sic energy cannot be created or destroyed, just transformed from one form to another).

In determining a prescribed level of energy-derived physical change in target oral tissue, it is necessary to appreciate the quantity of incident energy, the degree of positive interaction and the potential for energy conversion. Inherent in every incident laser beam is the photonic energy.

Laser "light" is considered unique in that, unlike other forms of light (sunlight, incandescent, LED irradiation), there are two inherent properties—monochromaticity and coherence. The single wavelength concept is founded in the physics of laser EM propagation, and using the Einstein/Bohr postulations, the delivery of laser irradiation is in sinusoidal waveform with successive waves in phase—so-called wave coherence. Additional man-made configuration of the photonic energy produced can provide high-density beam spatial density over distance—this is termed "collimation." In terms of the benefits of laser-tissue interaction due to these properties, the monochromatic absorption of a chosen laser,

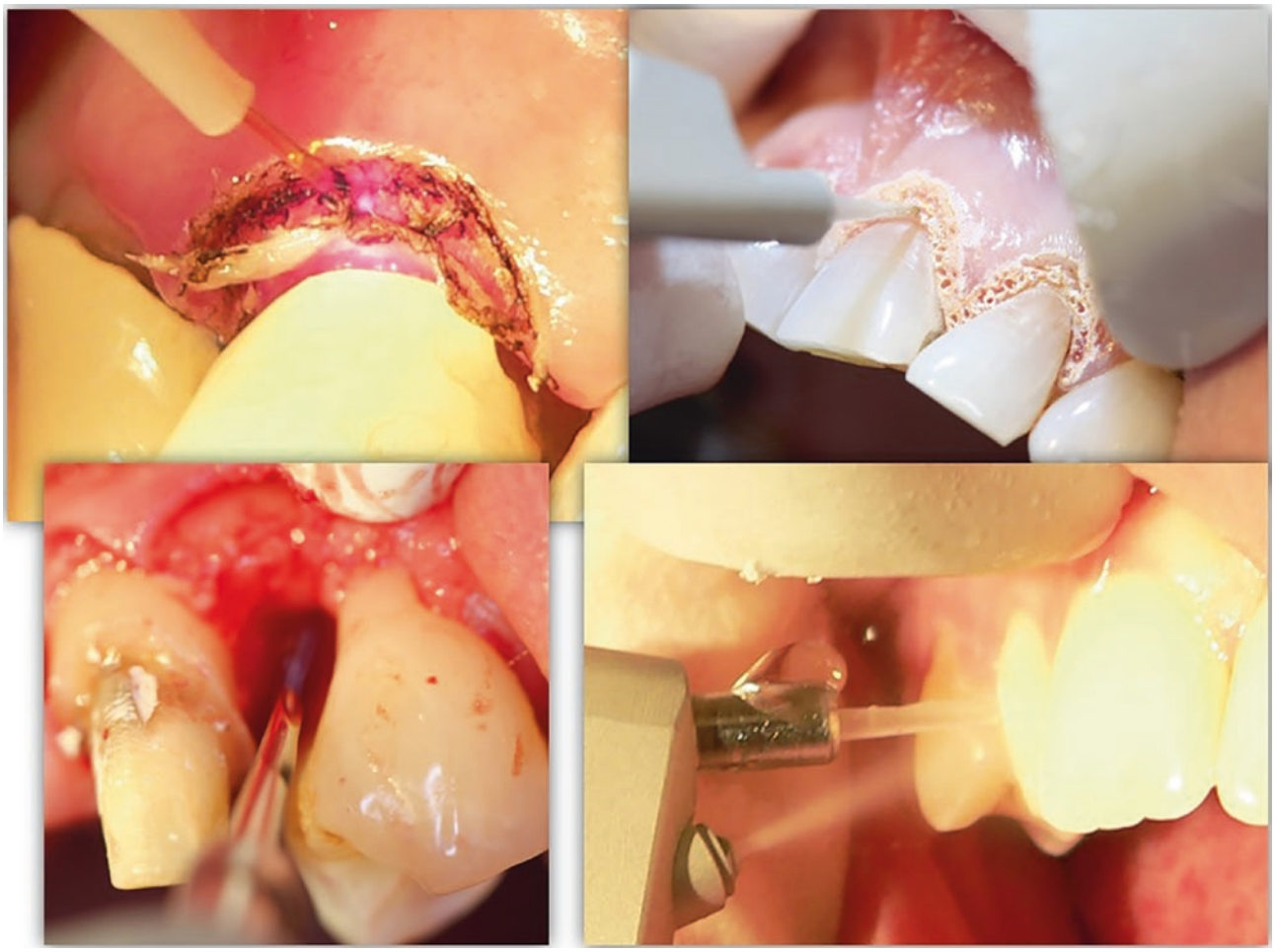


Fig. 3.1 Laser-assisted oral tissue surgery is photothermal in nature. Incident (photonic) energy is absorbed by target tissue elements (chromophores), relative to the laser wavelength. This leads to rapid temperature rise, protein denaturation and water vaporization. This constitutes an example of photoablation

together with the coherence of the beam, will offer selective tissue interaction of high quality; the collimation and ability to focus the beam will define a degree of accuracy and power density.

Figure 3.1 provides clinical examples of predictable laser-tissue interaction.

3.2 Photonic Energy

The emission of a single photon from an atom is the result of a shift in the energy status of that origin. Plank proposed that all matter existed in a state of energy relative to extremes of a lower (ground) state and a higher (energized) state, commensurate with entropic physical form [4]. Boltzmann, through his theories on thermodynamics [5], readily accounted a direct relationship between matter, energy and temperature. Put simply, an electric light filament at room temperature is a dull, inert wire but rapidly heats when energized by an electric circuit. The resistance of the wire leads to thermal conversion of the electrical (EM) energy. At this induced higher energy state of the light filament, the volatility of constituent electrons gives rise to higher thermal energy and emission of such energy as light.

Laser photonic energy assumes the production of high-energy photons from an energized source, whereby each photon scribes an identical waveform, and each photon has identical energy value. Plank and Einstein had established an inverse relation between wavelength and photonic energy, a direct proportional relationship between photonic energy and frequency, and Neils Bohr paved a way for the "quantum" (amount) nature of emitted photons to be calculated; thus, it provided a predictable base for the development of the maser and optical-maser, or laser.

The energy of emitted photons is expressed in Joules, or more conveniently, electron-volt eV (energy derived by acceleration through a PD of 1 V). Since photonic energy is related to wavelength (λ), photons emitted

Table 3.1 Commonly used laser wavelengths associated with dental treatment. Photonic energy and wavelength are inversely proportional. With ascending numerical value of wavelength, the corresponding photonic energy (expressed in electron volt—eV) is reduced

(eV)	Laser	λ (nm)
2.4	KTP	532
2.0	He–Ne	633
1.6	Diode	810
1.2	Nd:YAG	1064
0.4	Er:YAG	2940
0.1	CO_2	10,600

from different sources will have differing energy values. Basic calculation can be derived through:

$$\lambda = hc / E$$

where h = Plank's constant, c = speed of light and E = photon energy in eV.

$c = 300 \times 10^6$ m/s; $h = 6.626068 \times 10^{-34}$ m^2 kg/s represents the proportionality constant between the energy (E) of a photon and the frequency (ν) of its associated electromagnetic wave.

1 eV = 1.602×10^{-19} J, and it is possible to evaluate energy-equivalent values for the many laser wavelengths; for example, photons of wavelength 1240 nm (near infrared) equate to an eV value of 1.0, whereas an eV value of 2.0 is 621 nm (visible red) and eV 0.13 equates to 9600 nm.

Table 3.1 provides an overview of laser wavelengths commonly used in dentistry with corresponding photonic values.

3.3 Photonic Energy and Target Molecular Structures

A simplistic look at one of the many graphic representations of the relationship of target tissue elements, incident laser wavelengths and relative absorption potential would suggest that laser photonic energy is capable of ablative interaction with target tissue elements (chromophores). A chromophore is defined as "a chemical group capable of selective light absorption resulting in the coloration of certain organic compounds" [6]. For those compounds whose color is discernible within the visible spectrum, the definition may be sustained; however, given the concept that the chemical group confers a preferential ability to absorb (to a greater or lesser degree) photonic (EM) energy, the wavelength of that energy may fall within a spectrum of ultraviolet to far infrared, a narrow component of which may be visible to the human eye.

It must be stressed that laser-tissue interaction may occur within one of the two basic scenarios—interaction that is sufficiently powerful to cause direct and irreversible change in the target (usually achieved through thermal rise), a process termed photothermolysis, and a second, less powerful interaction that results in non-ablative, predominately (but not exclusively) stimulatory and biochemically mediated change, termed photobiomodulation (PBM).

According to the first law of thermodynamics, the energy delivered to the tissue must be conserved, and three possible pathways exist to account for what happens to the delivered light energy when laser photonic energy is delivered into tissue (Fig. 3.2):

1. The commonest pathway that occurs when light is absorbed by living tissue is called internal conversion. The energy of the electronically excited state gives rise to an increase in the vibrational modes of the molecule; in other words, the excitation energy is transformed into heat [7]. In many instances, the thermal rise is near instantaneous and substantial and quickly leads to conductive thermal energy into surrounding tissue. In the case of oral soft tissue and visible/near-IR laser wavelengths, the absorption by tissue chromophores gives rise to protein denaturation and secondary vaporization of interstitial water. The result is a visible ablation and vaporization of target tissue [8].

   With longer laser wavelengths, mid-IR and far IR, the prime absorptive tissue element in both soft and hard oral tissue is water. Ablation of tissue is achieved through the near-instantaneous vaporization of interstitial water, leading to an explosive fragmentation of tissue structure. With hard oral/dental tissue, this interaction can be quite dramatic [8].
2. With incident laser photonic energy values that fall below target tissue ablation, a second pathway can occur as fluorescence. Fluorescence is a luminescence or re-emission of light in which the molecular absorption of a photon triggers the emission of another photon with a longer wavelength. Such action provides the basis for optical scanning techniques used in caries detection in enamel and dentine and tomographic techniques in the scanning of soft tissue for neoplastic change.
3. The third pathway is broadly termed photochemistry [9]. Because of the energy of the photons involved, covalent bonds cannot be broken. However, the energy is sufficient for the first excited singlet state to be formed, and this can undergo intersystem crossing to the long-lived triplet state of the chromophore. The long life of this species allows reactions to occur, such as energy transfer to ground-state molecular oxygen to form the reactive species, singlet oxygen. Singlet or nascent oxygen is an ultrashort-lived form of the parent molecule that can

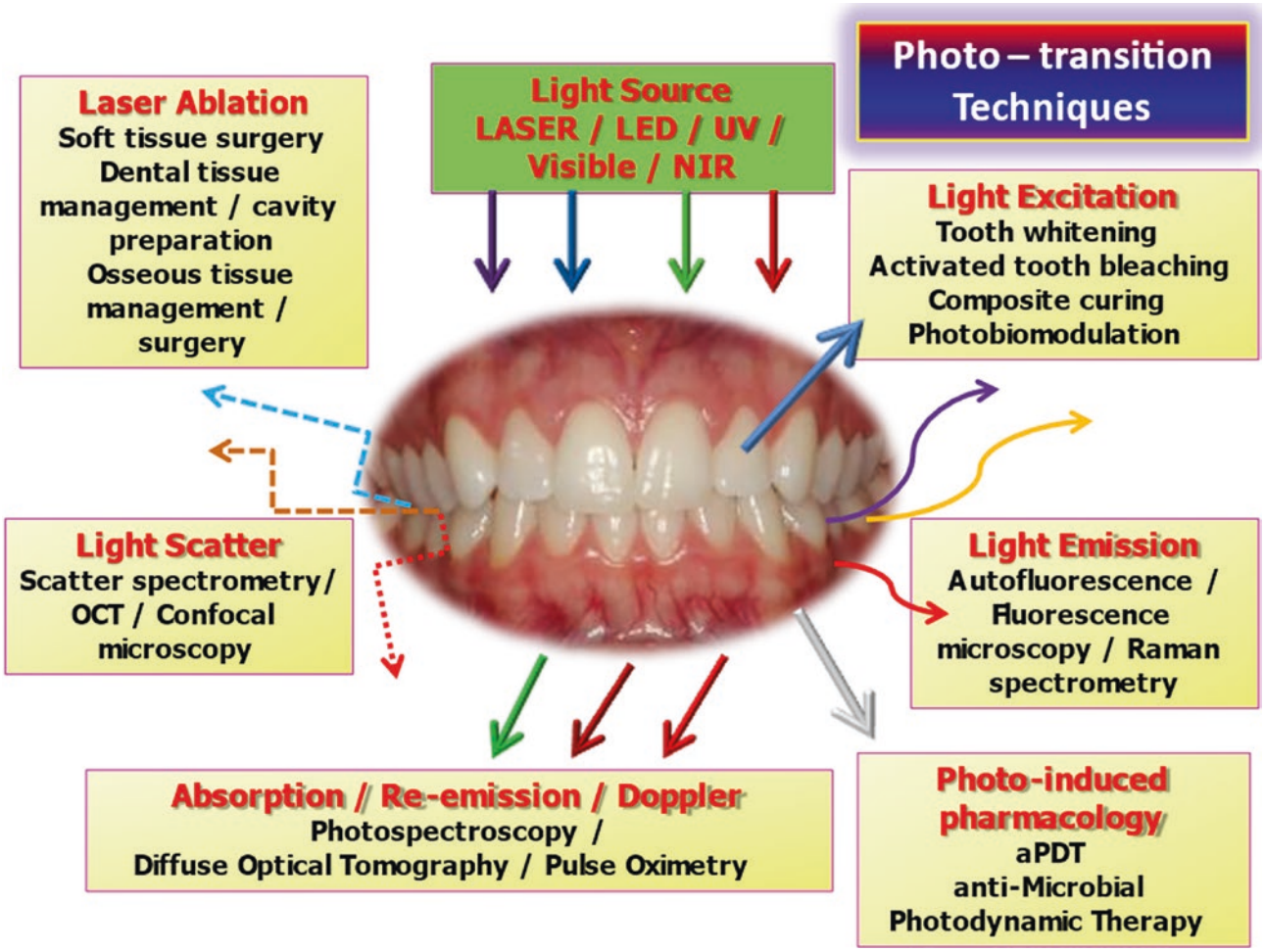


Fig. 3.2 An overview of the manipulation of incident photonic energy, such as laser light as an adjunct to screening, diagnostic and therapeutic clinical activity

cause cell apoptosis through oxidative stress. Such action can be commonly seen in photodynamic therapies where an intermediary chemical—photosensitiser—is employed to direct energy transfer to target tissue sites [10, 11].

Electron transfer reactions are highly important in the host cell mitochondrial respiratory chain [12], where the principal chromophores involved in laser therapy are thought to be situated. An additional photochemistry pathway that can occur after the absorption of a red or NIR photon within a host cell is the dissociation of a non-covalent bound ligand from a binding site on a metal containing co-factor in an enzyme. The most likely candidate for this pathway is the binding of nitric oxide to the iron-containing and copper-containing redox centers in unit IV of the mitochondrial respiratory chain, known as cytochrome c oxidase. Such action may induce an increase in cell pH and production of ATP and has been cited as basic cellular theory in photobiomodulation with low-level lasers.

3.4 Basics of Photothermolysis

Incident photonic irradiation directed onto target tissue will behave in one of the four main ways: transmission, reflection, scatter and absorption. The defining criteria can be simply summarized as dependent on the nature of the target tissue and wavelength of the incident beam (hence the predictability of absorption or transmission), nature of the tissue and its heterogeneity (hence the scope for scatter) and angle of the beam incident to the tissue surface (incident beam angle < total reflective angle) wherein reflection may have predominant effects (Fig. 3.3).

Oral hard and soft tissue is complex and heterogenous, anisotropic and of varying degrees of thickness, commensurate with structural anatomy. Within such tissue, component elements may be found that represent key molecules capable of selective absorption of photonic energy; within the visible EM spectrum, such molecules are termed chromophores, and with longer (IR) spectra, a terminology of absorptive tissue element may

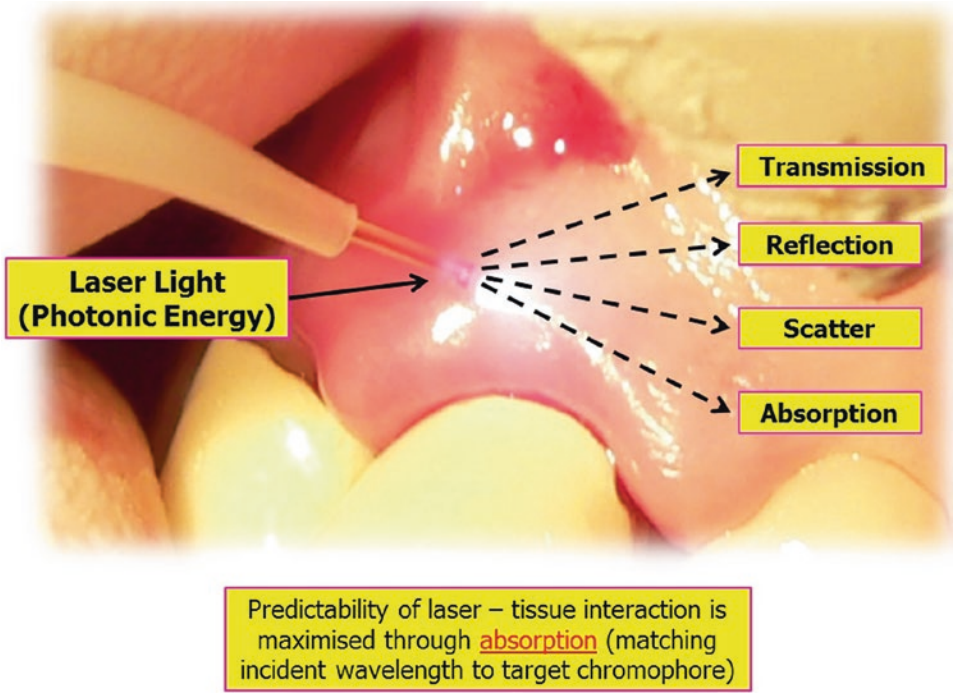


Fig. 3.3 Summary of the basic interactive phenomena of incident laser energy and target tissue. Such is the variance in tissue structure and heterogeneity commonly found in oral hard and soft tissue that there may be multiple and complex degrees of each interactive phenomenon

be adopted. Examples are protein/amino acid-based molecular groups, such as collagen, keratin and non-structural proteins such as melanin, hemoglobin in both its oxygenated and non-oxygenated (HbO_2 and Hb) forms. Dental and osseous tissues are based upon the calcium phosphate complex referred to as hydroxyapatite (HA) found as a structural crystal in bone and the carbonated lattice crystal as a mineral component of enamel and dentine (CHA). Water—as the intra-cellular medium base of cell cytoplasm, a component of circulating blood and plasma, a free molecule in interstitial tissue structure or as a hydroxyl (OH^-) radical as part of the hydroxyapatite molecule—represents a major ubiquitous molecular group of varying degrees of absorptive potential relative to incident photonic wavelength. Figure 3.4 demonstrates a graphical interpretation of the interaction of ascending photonic wavelengths with each of the major tissue chromophores that are found in oral and dental tissues. The term adopted as a measurement of the level of energy absorption by a chromophore is absorption coefficient—considered as a measure of the rate of decrease in the intensity of electromagnetic radiation (as light) as it passes through a given substance. The optical properties of tissue will determine the penetration into tissue of the radiant energy from a laser source. Absorption coefficient is inversely proportional to transmittance, and the depth of penetration of photons within a given chromophore will reduce as the absorption coefficient increases.

Each chromosome has molecular structure and for each there is a "ground state" which defines the structure, atomic configuration and interatomic binding energy at body temperature [13]. If external energy is applied, a point may be reached when molecular vibration is sufficient to overcome the forces binding atoms or molecules together. Examples include protein dissociation and water vaporization. True photonic ablation of a target molecule therefore represents incident energy sufficient to break interatomic binding forces and is termed dissociation energy. Table 3.2 provides examples of commonly found chromophore molecules and the dissociation energy value required to break the interatomic bond.

As is evident from data in Table 3.1, almost none of the popular laser photonic energies is capable of direct intra-molecular bond cleavage and one may be

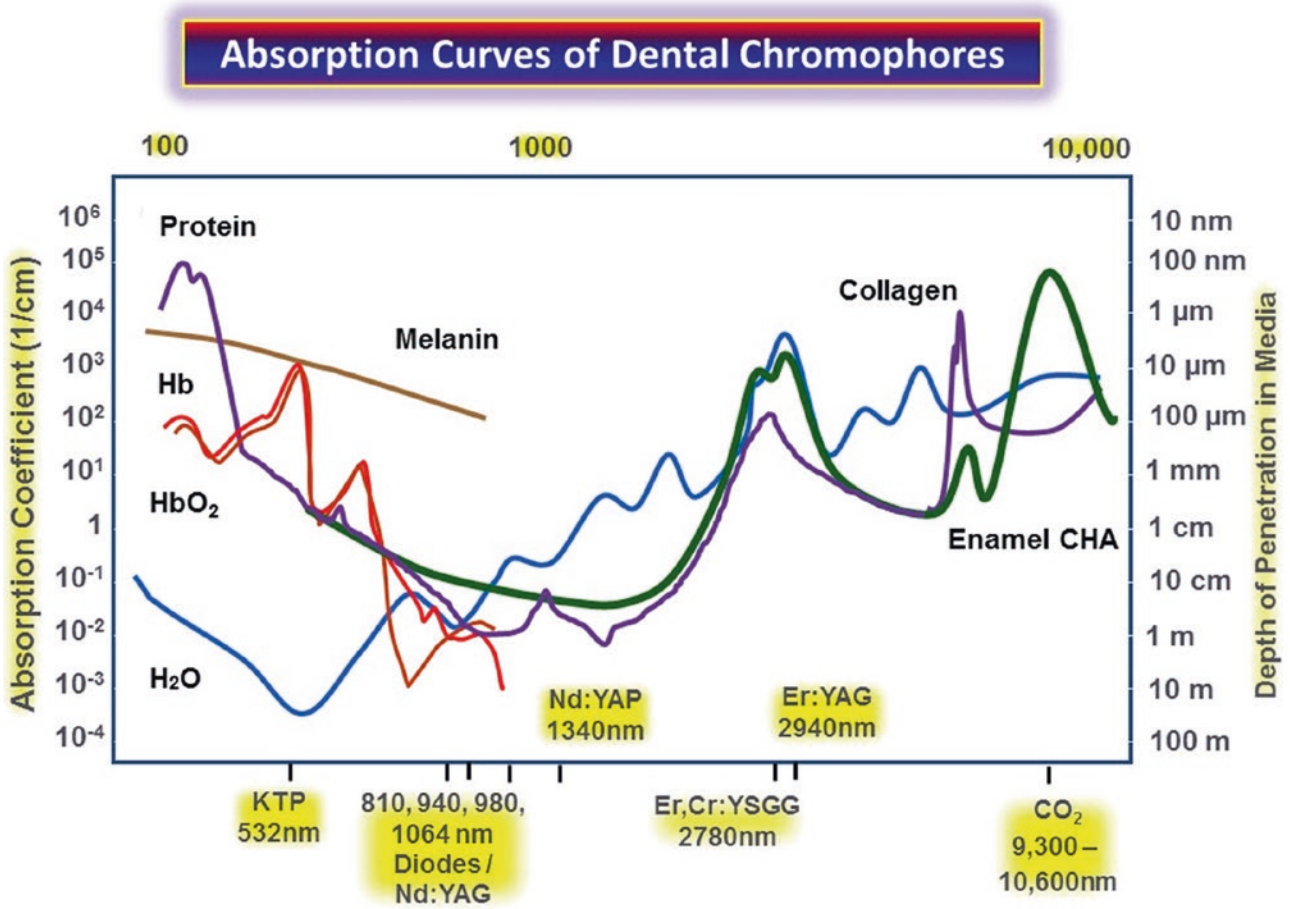


Fig. 3.4 Absorption coefficient curves for commonly found chromophores, relative to incident photonic wavelength

Table 3.2 Provides examples of commonly found chromophore molecules and the dissociation energy value required to break the interatomic bond

General concepts	
Dissociation energy of selected chemical bonds[a]	
Type of bond	**Dissociation energy (eV)**
C=O	7.1
C=C	6.4
O–H	4.8
N–H	4.1
C–C	3.6
C–N	3.0
C–S	2.7
Fe–OH	0.35
HA lattice	310

Dissociation energy, expressed in eV values, required to break the bonds (covalent, ionic, etc.) that bind atoms of common chromophores. Examples represent component molecules within tissue water, protein, blood and ionic forces within the crystal lattice of hydroxyapatite

Data taken from Mó O, Yáñez M, et al. J Phys Chem A. 2005;109(19):4359–65 [13]

forgiven for concluding that dental lasers cannot ablate target oral tissue through the use of empirical state photonic energy. Certainly, when the binding (ionic) lattice energies of crystalline carbonated hydroxyapatite are exposed to the mid-IR laser wavelengths (Er,Cr:YSGG, Er:YAG), the photonic energy value is pitiful compared to the dissociation energy of hard dental tissue [14].

Something else must be happening.

Figure 3.5 offers a summary of the interaction between a photon and target chromophore molecule, through successive stages of absorption, excitation and dissociation. Such predictive events might account for why certain laser wavelengths interact (are absorbed) with certain oral tissues.

Although individual photons possess insufficient energy to break apart target molecules, with each successive photon absorbed, the energy causes increasing molecular vibration up to a point where sufficiently high power density (energy density within ultrashort time) drives molecular fragmentation, or—more commonly seen with current dental lasers—molecular vibration converted into thermal rise leads to protein denaturation and water vaporization.

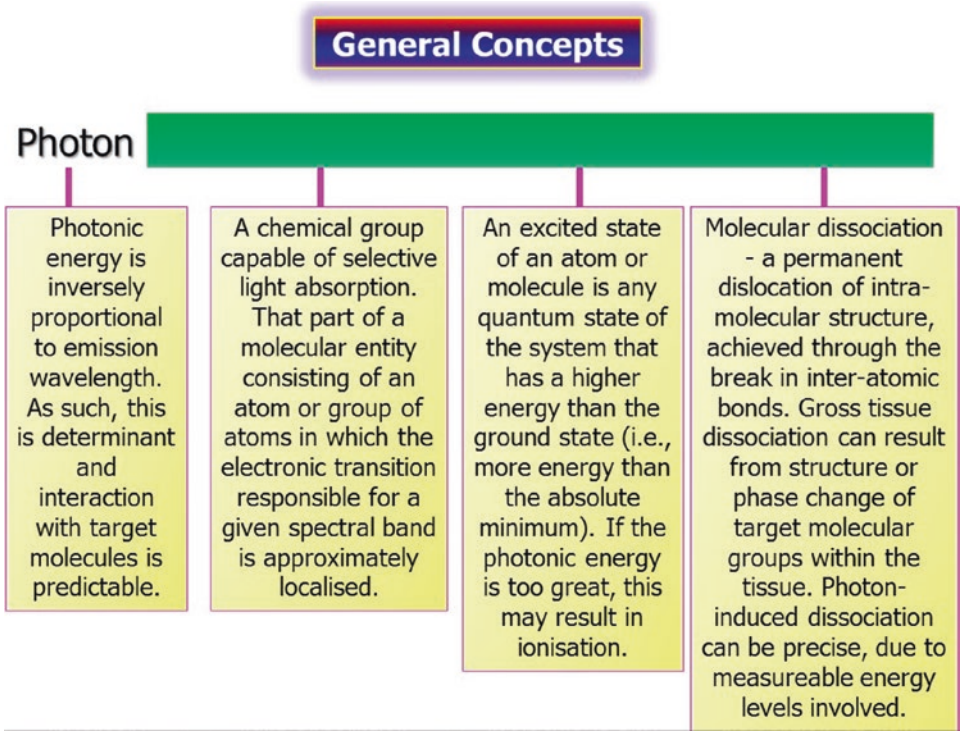


Fig. 3.5 Photonic energy interaction with target chromophore molecules

3.5 Problems Associated with Delivery of Photonic Radiation vs. Laser Wavelength

Considering a clinical application of high-intensity lasers, parameters such as wavelength, energy density, intensity, peak power, average power, repetition rate and pulse length are extremely important to heat generation due to irradiation on any biological tissue. The amount of heat inside the tissue is highly dependent on its optical properties, such as absorption and scattering coefficients [15].

The complex nature of oral soft tissue structure can pose some problems in delivering predictive laser-tissue interaction. For some wavelengths in the visible and near-infrared regions of the EM spectrum, the prime pigmented chromophores may be at some depth within the oral epithelium and covered with a thick keratinised layer. During photothermolysis, photonic energy may be delivered to the target and theoretically transferred (and undergo conversion) in one of the three [16] ways:

- *Radiation*: Non-contact laser waves are emitted from the delivery tip and absorbed by the target. Energy conversion occurs. This is referred to as the "real" laser effect. High-photonic-energy wavelengths such as the KTP (532 nm) may be delivered through a "non-contact" technique and direct photoablation may occur.
- *Conduction*: At slightly longer wavelengths (810–1064 nm), the need arises to both hold the delivery fiber in contact with the tissue and "initiate" the fiber tip with suitable absorbent material. A proportion of the incident laser energy is absorbed and gives rise to a "hot-tip" effect, whereby thermal energy is conducted to the tissue and aids the ablation of the tissue.

  Some concern is expressed as to possible disadvantages of this technique in that the effect of the hot tip on tissue is independent of the wavelength of laser radiation and that the heated fiber (sic) transmits only thermal energy, with no direct radiation energy [17].
- *Convection*: Moving heat within large volumes of liquid or gas either toward or away from the target. A similar effect of cooling may be seen through the effect of using a water spray, high-volume suction or air.

3.6 Concepts of "Power Density"

Figure 3.6 provides an example of simple manipulation of sunlight, using a magnification device such as a simple lens. Multi-wavelength cosmic radiation from the sun, although powerful and capable of tissue damage over time, can be brought to a focal area, and with the

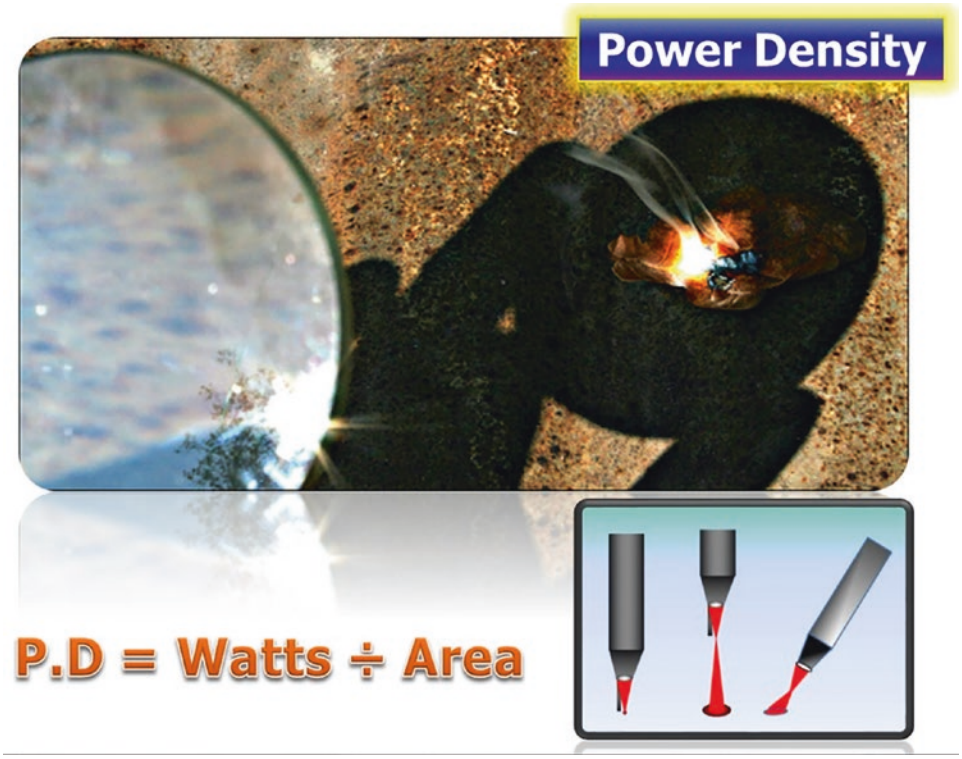


Fig. 3.6 Manipulation of incident beam power over area to enable the power density effect to be a major factor in delivering predictable and powerful laser-tissue interaction

power in the beam concentrated to a small spot, the effects are much more dramatic. The use of a magnifying glass to concentrate the sun's rays is a simple example of power density. Power (energy per second) is an expression of a laser's ability to do work, and when measured over the area exposed to the beam, it will be readily acknowledged that the greater the concentration of photons, the greater the level of potential interaction. Consequently, for any given laser delivery system, reduce the spot size of the beam and one can expect to speed up the interaction—assuming that all other parameters remain constant [18].

"Power density" (PD) is the delivery of energy through time divided by the area of the exposed tissue; it is expressed in W/cm^2.

The output of any laser over time is expressed as average power and equates to the total number of Watts delivered per second. For a continuous wave emission laser, the average power will equal the maximum output; for a micro-pulsed free-running emission, the average power output may be of the order of a few Watts, but due to the active photonic emission only lasting a possible 20% of each second, there will be peaks of energy. A typical free-running emission laser, such as Nd:YAG or Er:YAG, may deliver an average power value of 3.0 W, but due to the pulse width of 150 μs, there will be peak power bursts of 1000+ W [19].

As has been seen elsewhere, the predominant temporal emission mode is a Gaussian distribution, and this lends itself readily to being brought to a focal spot. With control over the area of irradiation, the concept of power density as a prime factor in laser-tissue interaction becomes valid. Even in those conditions where there is little if any direct absorption, the concentration of laser power in ever-higher values over ever-shorter time periods gives rise to power density of such magnitude that photodisruption and photoionization of target molecules can occur.

Technology has much to deliver in terms of future developments, but already it is possible to see predictable laser-tissue interaction involving PD of values of 10^6+ W/cm^2 for microsecond periods and even shorter, enabling photovaporization of interstitial water in tooth tissue and consequent disruption of the crystalline solids.

Nowhere within this discussion does the influence of laser wavelength occur. Of course, the harmonization of incident wavelength, its inherent energy value and selective absorption within a suitable target chromosome will remain as empirical in our understanding of interaction concepts; but as we have already seen, the continuous bombardment of a target tissue with photons of a suitably absorbed wavelength will lead to thermal rise and eventually sufficient heat to effect physical change in the

tissue. Such processes take time, and the risk of collateral thermal damage becomes an ever-present threat. To deliver sufficient energy in a form of concentration of both area of exposure and time must be seen as a distinct advantage.

As our understanding develops, there evolves the interaction of three components to predictive laser-tissue interaction: the absorptive potential of the target tissue, relative to the incident laser wavelength; secondly, the transfer of energy—from initial photonic energy and delivery (emission) mode of the laser, together with power density and time of exposure; and thirdly, the availability of thermal relaxation to enable the target tissue to avoid progressive overheating. Thermal relaxation is inherent with free-running pulsed emission modes and impossible to deliver with continuous-wave emission modes. The clinician would therefore need to be aware of the potential for thermal damage and allow sufficient time and respite periods to enable the tissue to recover.

In Fig. 3.7, the relationship between PD and exposure time is represented with reference to an ascending physical change in the target tissue. Very low irradiance over extended periods may give rise to subtle stimulation of biochemical pathways associated with tissue heath and reparative capabilities and is the core of understanding of photobiomodulation. With ever-shortening time and an ascending level of power density, not only do irreversible physical changes occur in tissue, but with exposure times of micro- and nanoseconds, even with relatively low average power values, the effects on the target can also be spectacularly rapid and without leaving a "thermal thumbprint."

For any given laser-tissue interaction, assuming that absorption can occur through the equation of power density with exposure time may enable the clinician to influence the type of interaction that occurs. It has already been established that at everyday levels of power delivery in dentistry, the predominant effect is tissue ablation through thermal rise—photothermolysis. By reducing the exposure time to milliseconds and microseconds, successively higher peak power density above 10^8 W/cm^2 can be obtained. At such powerful levels, the intensity of energy is so great that electromagnetic fields developed around the interaction are sufficient to tear target molecules apart—photoplasmolysis [20]. Reference to the work of Boulnois and the graphic representation of laser-tissue interaction can be seen as an ascending phenomenon and product of ultrashort exposure time and megawatt peak power [21].

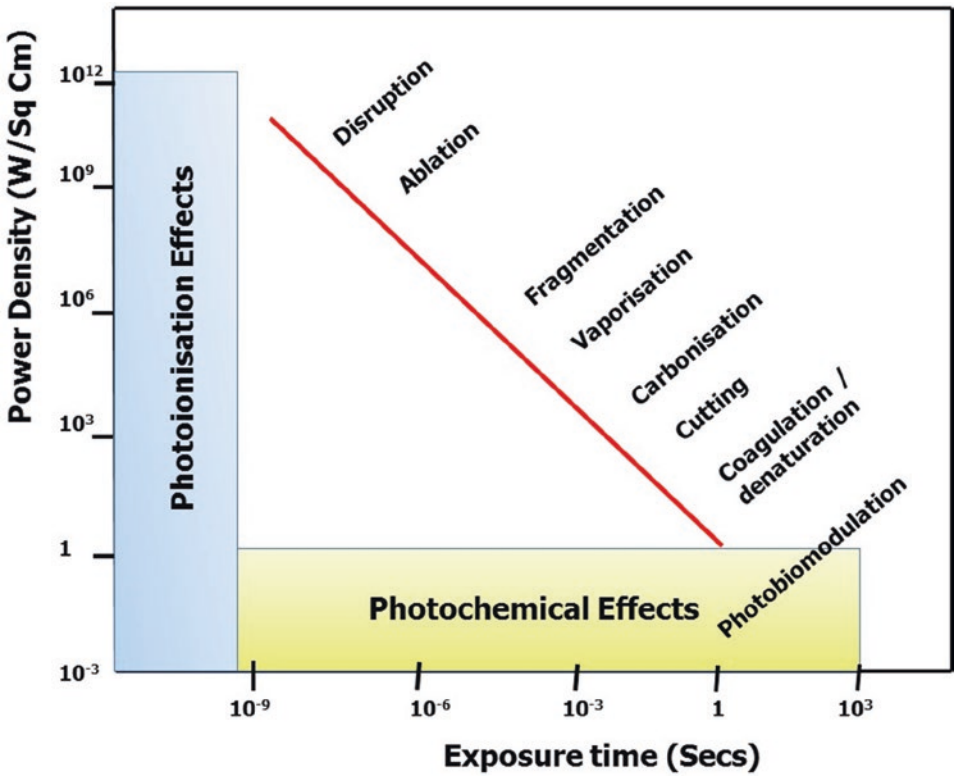


Fig. 3.7 Relationship of incident photonic power density and exposure time. (Source: Boulnois J-L. Laser Med. Sci. 1986;(1):47–66 [21])

3.7 Thermal Rise and Thermal Relaxation

In considering the broadest concepts of photothermal action and regardless of the laser system used for a soft tissue surgical application, the effects may be broadly classified as follows:

- *Tissue heating*: To a non-destructive level, the warming of tissue may be a desired total effect (as part of a biomodulation therapy) or may occur at some distance from an ablation site, along a thermal gradient within the tissue. This latter example is covered in greater detail later.
- *Tissue coagulation*: Over a period of time, the temperature rise in soft tissue at or above 45 °C will constitute "destructive heating," i.e., amounting to progressive irreversible change. At about 50 °C, bacteria can be demonstrated to achieve a state of deactivation, with tissue protein denaturation occurring at around 60 °C. Within this zone of thermal rise, the walls of small-diameter vessels (arterioles, venules and lymphatics) within the irradiated area will undergo structural change of vessel walls and lead to progressive blood and lymph coagulation. Dependant on the wavelength of the laser used and concentration of chromophores specific to that wavelength, a concept of "selective photothermolysis" can be considered, such as in the ablation of melanin in diode laser-assisted gingival de-pigmentation.
- *Vaporization*: At normal atmospheric pressure (1 Bar), vaporization of water occurs at 100 °C. Within soft tissue, the vaporization of water will accompany existing protein denaturation that may have occurred at a lower temperature. The phenomenon of water vaporization is accompanied by volumetric change and expansion in the ratio of 1:1600 as the liquid is vaporized to steam. With short wavelengths (visible and near IR), the structural change in soft tissue collagen scaffolding would allow water vaporization to occur as part of the overall ablation process. With mid-IR erbium family wavelengths, the scenario is often very different owing to the poor absorption of these wavelengths in pigmented tissue, the limitation of thermal rise in the tissue when a co-axial water spray is being used and the FRP emission mode of the lasers with associated high peak power values, and with the extremely high absorption of these wavelengths in water, the vaporization is often more dynamic and accompanied by audible "popping."

  When erbium lasers are used on soft tissue without water spray, or when CO_2 laser wavelengths are used (often without water spray), the vaporization is reflective of a more "thermal" exchange, where the heating of the tissue as well as the absorption of the photons in water occurs. Often, the result of such vaporization leads to visible signs of tissue desiccation, structural shrinkage and predisposition to rapid heating and carbonization.
- *Carbonization*: As laser-assisted surgery proceeds, the risk remains to be the potential for tissue heating that leads to the production of end-stage molecular destruction and residual carbon. It is generally considered to be at temperatures around 200 °C, although the actual rise may be considerably higher. Carbonization would only occur as a result of either an inappropriate high dose, relative to the parameters consistent with a desired surgical outcome, or the application of laser photonic energy over excessive time, resulting in opportunity for destructive collateral effects. The characteristic visual sign is the development of black residue associated with the soft tissue incision.

  By far, the consequence of such development is the preferential absorption in carbon residue of (in effect) all incident EM photonic wavelengths. This is the basis of the "black body" concepts of preferential absorption and characterized by the re-emission of multi-wavelength, incandescent near-IR thermal radiation; in consequence, the carbonized tissue continues to absorb incident laser energy and becomes the source of thermal conductive energy that significantly contributes to collateral damage during soft tissue surgery.
- *Photoacoustic phenomena*: As has been seen already, incident coherent photonic energy can be subject to conversion into other forms of energy, notably thermal in the dominant effect of photothermolysis. With the instantaneous phase change of water from stable liquid to vapor, the volume change can give rise to a cavitation phenomenon and consequent shock wave. Additionally, the energy may be changed to sound, and this may be witnessed with mid-IR interaction with tissue and the "popping" sound often heard. True photoacoustic effects are used elsewhere in medicine and surgery in procedures such as lithotripsy, where kidney and gallstones are fragmented using indirect shock waves; in maxilla-facial surgery, a similar approach has been reported to assist in the fragmentation and subsequent safe passage of sialoliths within the submandibular gland [22, 23].

Given the current limitations of laser emission parameters and the consequence that by far the greater consequence of laser-tissue interaction is photothermal shift and temperature rise, the ablation of target tissue can be severely compromised by excessive thermal rise and a build-up of the ablation residue that may rapidly overheat.

The effects of thermal rise can be both subtle and dramatic—depending on the rate of warming. Figure 3.8 provides a tabulated outline of the effects that temperature rise may have, relative to both the visual change and the biological change (the latter as may be applied to soft tissue). In addition, the varying stages of thermal rise have been investigated and provide opportunity to influence the structural changes in the tissue and the effect of heat on associated bacterial cells. Irradiated tissue should not be regarded as sterile, although there will be significant pathogen reduction at the site of maximum laser-tissue interaction.

Two concepts of ablation may be considered: a zone of tissue removal/permanent change preceded by an "ablation" front and a second advancing line denoting the permanent effect of change rendered by thermal rise—a "thermal" front. In an ideal situation, the "ablation" front will denote the predicted volume of ablated tissue and through the correct management of heat rise and debridement, the risk of unwanted thermal damage can be avoided. In hard tissue management, the concepts of ablation and thermal zones will be discussed with specific reference to tooth cavity preparation.

Thermal relaxation time can be deduced mathematically [24] as the time taken for the irradiated tissue to dissipate about 63% of the incident thermal energy. It is additionally related to the area of the irradiated tissue and thermal diffusivity and bulk of the tissue.

Thermal damage time is the time required, for the entire target, including the primary chromophore (e.g., melanin) and the surrounding target (e.g., gingiva), to cool by about 63%. It includes cooling of the primary chromophore as well as the entire target.

Extinction length is the thickness of material necessary to absorb 98% of incident energy.

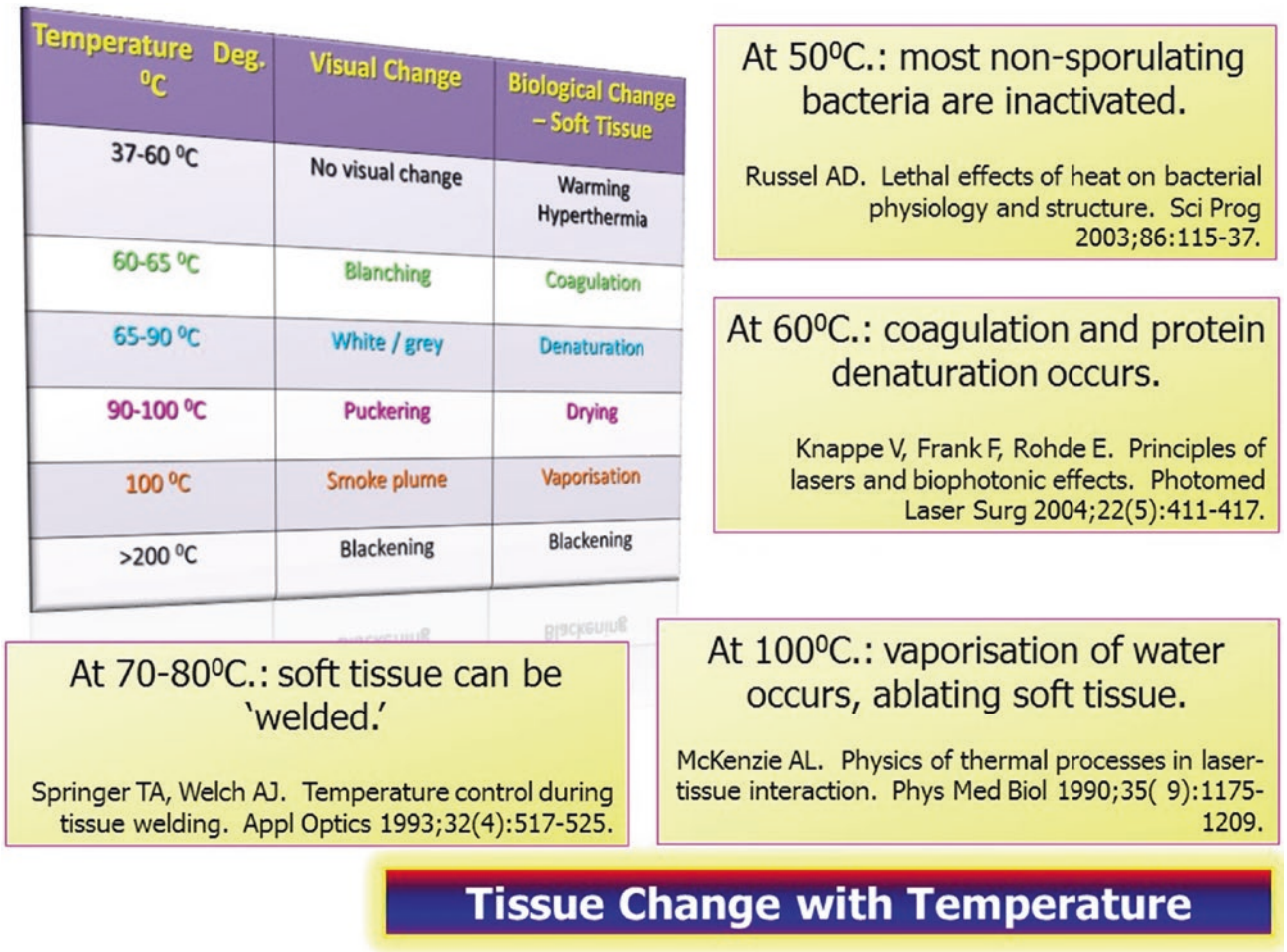


Fig. 3.8 Effects of thermal rise on (soft) tissue

3.8 Laser Photonic Energy and Target Soft Tissue

Various laser wavelengths are available for clinical use with target oral soft tissue and span the visible (blue) EM spectrum through to the far infrared. Examples of laser wavelength currently available to the clinician are shown in ◘ Fig. 3.9.

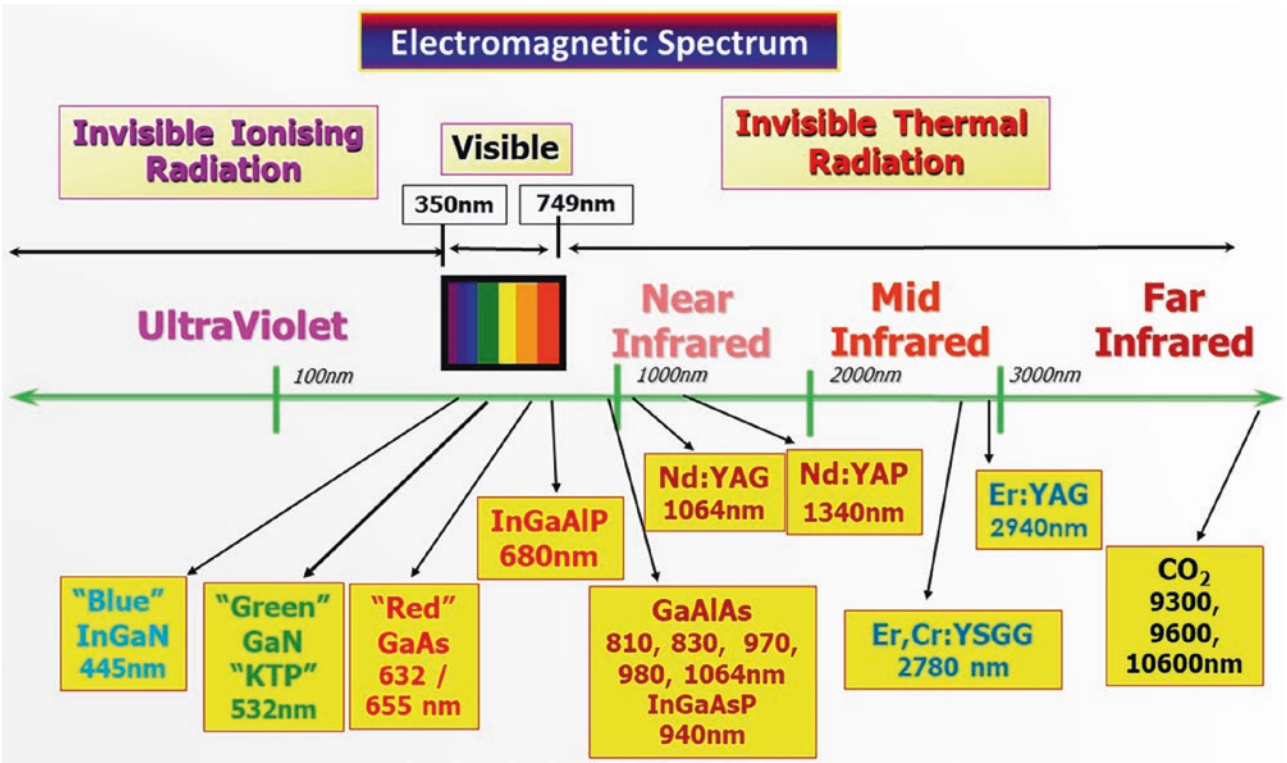
With current configurations of emission modes, power limits and commercial technology application, all soft tissue ablation achievable in clinical dentistry is primarily and almost exclusively due to photothermolysis [25]; in general, chromophore absorption is by pigmented molecules (haem, melanin) with short wavelengths (532–1064 nm), whereas longer wavelengths experience greater interaction with tissue water components (H_2O and OH^- radicals), with peak absorption occurring at approximately 3000 and 10,600 nm. The emergence of commercially available laser units with wavelength emissions at 450–490 nm offers the opportunity of utilizing absorption in water with a diode-source active medium. Protein as a structural component of oral soft tissue appears to have moderate absorption of ultraviolet wavelengths, together with peaks at 3.0 and 7.0 μm. Visible and near-IR wavelengths have limited absorption in protein, but as has been discussed above, secondary thermal rise consequent upon time-related photonic energy exposure will give rise to conductive heat changes in proteinaceous material.

As such, oral soft tissue, high in water and protein with varying degrees of pigment and blood perfusion, remains a straightforward target tissue wherein low-dose irradiation can be configured to deliver predictable laser-tissue interaction with limited collateral damage.

Of practical interest to the clinician, the following factors (◘ Table 3.3) will each and collectively affect the absorption of laser light by a chosen target tissue [26]:

Shorter wavelengths tend to penetrate soft tissue to depths of 2–6 mm [27], and scatter is a significant event, both back-scatter of photons and forward scatter into the tissue. Longer wavelengths are attenuated at or near the tissue surface, due to water content of cellular tissue. As tissue ablation proceeds, short-wavelength photonic energy causes protein denaturation and conductive effects as the tissue is heated. A typical soft tissue zone of near-IR laser ablation is surrounded by a zone of reversible edema and little evidence of acute inflammatory response. Classically, the progression of near-IR laser ablation of soft tissue is through a crater-shaped zone where depth and volume removed appear proportional [28].

In ◘ Fig. 3.10, the interaction and progression of near-IR irradiation in soft tissue are graphically represented. In an ideal fashion, the zone of ablation and



◘ **Fig. 3.9** Laser wavelengths commonly available for use in dentistry

Table 3.3 Individual factors associated with laser-tissue interaction that may affect the predictability of clinical use of a chosen laser

Factor	Comment
Laser wavelength	Individual wavelengths (visible extending to far infrared non-ionizing radiation) and inversely proportional to the photonic energy
Laser emission mode	Inherently continuous wave (CW) or free-running pulsed (FRP), due to the excitation source or additionally modified by the manufacturer to deliver gated CW and mode-locked CW or modification in pulse width (>10 μs) with FRP
Laser power value	With increasing power delivery, there is potential for thermal rise. Below the ablation threshold, this may be reversible (tissue warming/PBM)
Exposure time	Together with laser power, "spot size" and emission mode, this will affect power density and thermal relaxation
Tissue-type composition	All oral tissue is heterogenous, and the proportions of common chromophore content will alter the potential for individual laser wavelength absorption
Tissue thickness	Thicker tissue will take longer to incise/ablate. Additional factors may be thermal diffusivity and longer thermal relaxation times
Tissue surface wetness	Due to water or saliva—of note with longer wavelengths > approx. 1500 nm. Wetness will affect tissue reflection (below)
Incident angle of the laser beam	Incident angle of beam to tissue of 90° will define maximum potential for interaction. As angle approaches the reflection limit (TIR), this reduces the potential for interaction to zero
Contact vs. non-contact modes	Employed between laser delivery tip and tissue. With visible and near-IR wavelengths, contact technique may be essential to allow a "hot-tip" technique. Non-contact may have a focussed beam, and distance of tip/handpiece to tissue may be crucial to maximize laser-tissue interaction
Thermal relaxation factors	Exogenous (water spray, tissue pre-cooling, high-speed suction, pulsing/gating laser emission) Endogenous (tissue type and density, blood supply)

conductive temperature spread occurs over time. The predominant scatter phenomenon of these wavelengths gives rise to a complex pattern of photon penetration, wherein there may be indeterminate tissue effects, giving rise to the acronym WYDSCHY—"What You Don't See Can Harm You," coined by Fisher in his 1993 paper [29]. In essence, visible and near-IR laser wavelengths have deeper penetrating effects on oral soft tissue and demand that optimal and non-excessive operating parameters are used in order to avoid unwanted tissue damage.

Clinically, such interaction can be seen in Fig. 3.11, during the removal of a fibroma from the lateral tongue, using an 810 nm diode laser:

Certainly, even incisions will have a "U"-shaped cross-sectional appearance, and this is due in part to progression of photonic energy through scatter as well as some direct conductive thermal spread [29]. A simple in vitro example using pig mucosa with short and longer wavelengths provides an excellent example of the structure of the incision and difference in laser-tissue interaction between the two wavelengths (Fig. 3.12).

With longer laser wavelengths (mid-IR approx. 3.0 μm—Er,Cr:YSGG, Er:YAG—and far IR approx. 10 μm—CO_2), a more "V"-shaped cross-sectional appearance prevails (Figs. 3.13 and 3.14). The bulk of laser-tissue interaction occurs at or within the confines of the tissue surface, and as an incision is developed, the majority of excess energy (thermal) is released through the escape of vaporized tissue water [29]. This predominant effect reduces the conductive thermal rise into adjacent tissue. A risk exists with soft tissue in that desiccation of target tissue can predispose to the formation of carbonized tissue elements (surface char—termed eschar) and the preferential absorption of this material leading to very high temperatures that might cause conductive collateral tissue damage and post-operative pain. Various techniques have been developed to address this risk; the eschar is loosely adherent and can be easily wiped away with a damp gauze to allow fresh tissue exposure (this technique forms part of the so-called laser peel techniques associated with surgical treatment of surface pathology). Parameter manipulation may include the choice of short-gated CW or pulsed laser emission modes or coaxial water spray that may enhance tissue thermal relaxation.

Broad consensus would suggest that although laser soft tissue incisions do not heal any faster than scalpel, there is evidence that with appropriate operating parameters, these wounds appear to heal less eventually [30–32].

In terms of laser-tissue interaction and disregarding any reduction in bacterial contamination, there will be a point at some distance from the wound where both temperature and photon scatter are reduced to a point of containment within the tissue. By this, the temperature gradient reduces to a level of tissue stimulation, tissue molecular energizing and increased local blood flow [33]. In addition, a scatter gradient exists where the energy delivered is reduced to that point where biomodulation effects predominate [34]. For these reasons, it

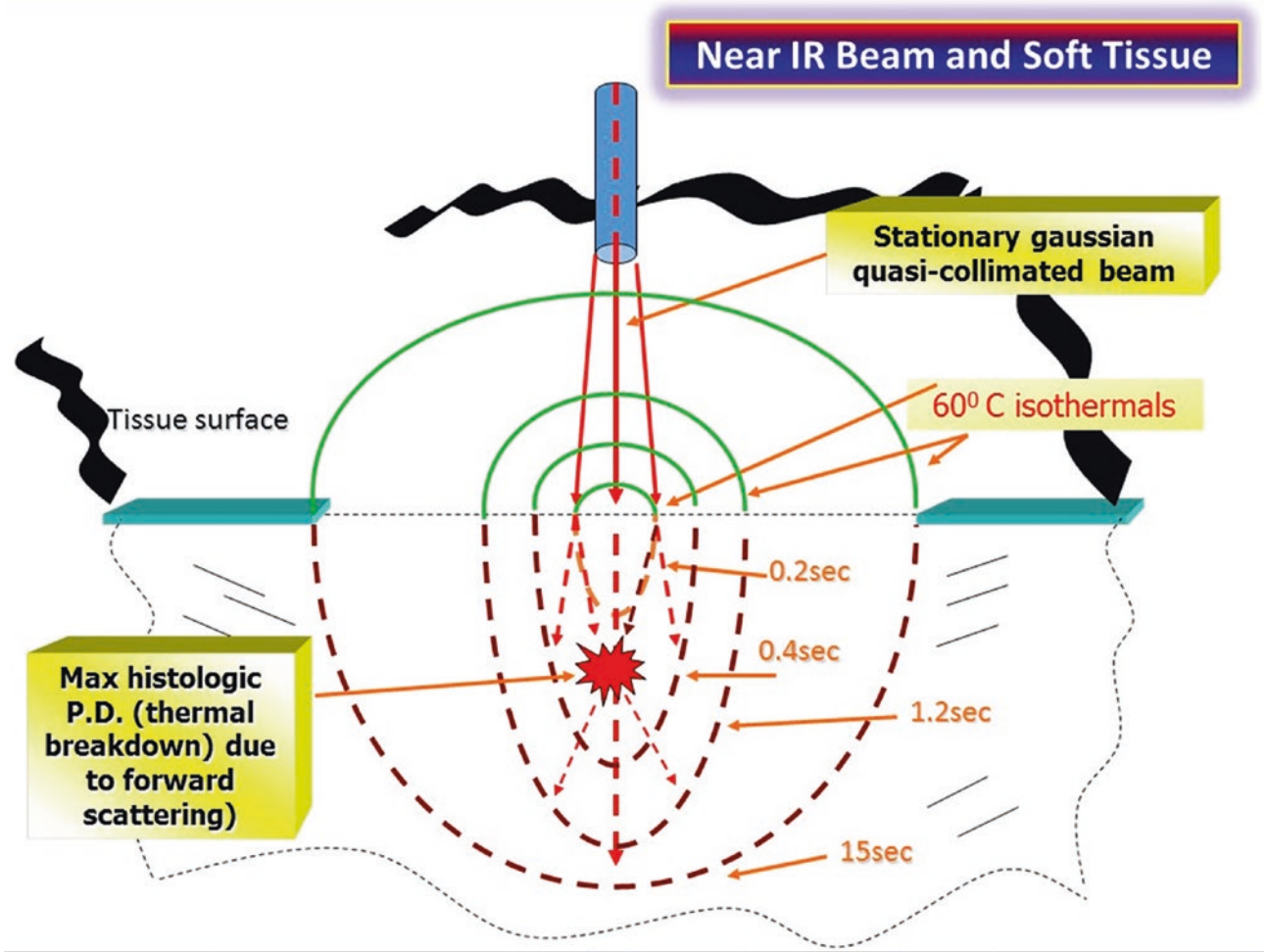


Fig. 3.10 Graphic representation of visible and near-IR laser photonic energy interaction with oral soft tissue. (Graphics: S. Parker after Fisher J.C. 1993)

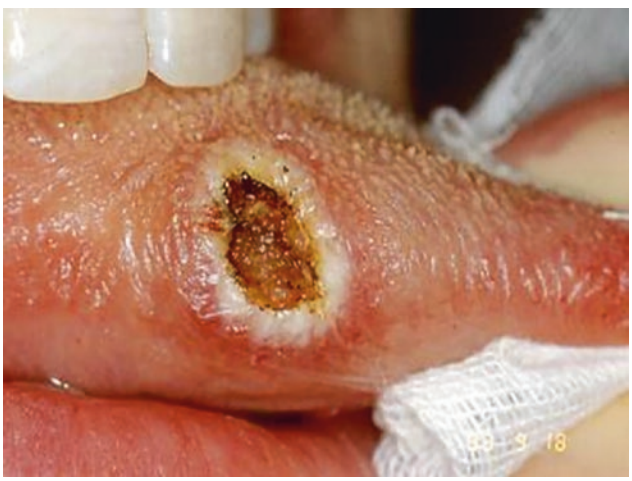


Fig. 3.11 Following removal of a fibroma lateral tongue, using an 810 nm diode laser. The central ablation zone is surrounded by an area of edema. The lack of carbon residue indicates a correct choice of laser power parameters

may be seen that laser-assisted surgical wounds respond in a positive and supportive framework that delivers less eventful healing (Fig. 3.15).

Fisher [35] defines a comprehensive understanding of photon scattering into deeper soft tissue areas that is seen with the use of visible and near-IR lasers. With successive interaction and as photons are absorbed, the possibility exists for a scenario whereby the ablation threshold of the host tissue at deeper sites is greater than the photonic energy. This "energy gradient" phenomenon might provide explanation as to how distant effects of (surgical) laser use may mimic essentially low-level (photobiomodulation) stimulation of cells and host tissue. Standard textbooks [36] provide authoritative and evidence-based explanations of how the host tissue may respond positively to low-level photonic energy, and the reader is directed to such references for further information.

Positive healing effects following laser surgery: One of the often-cited side effects of laser-assisted surgery is

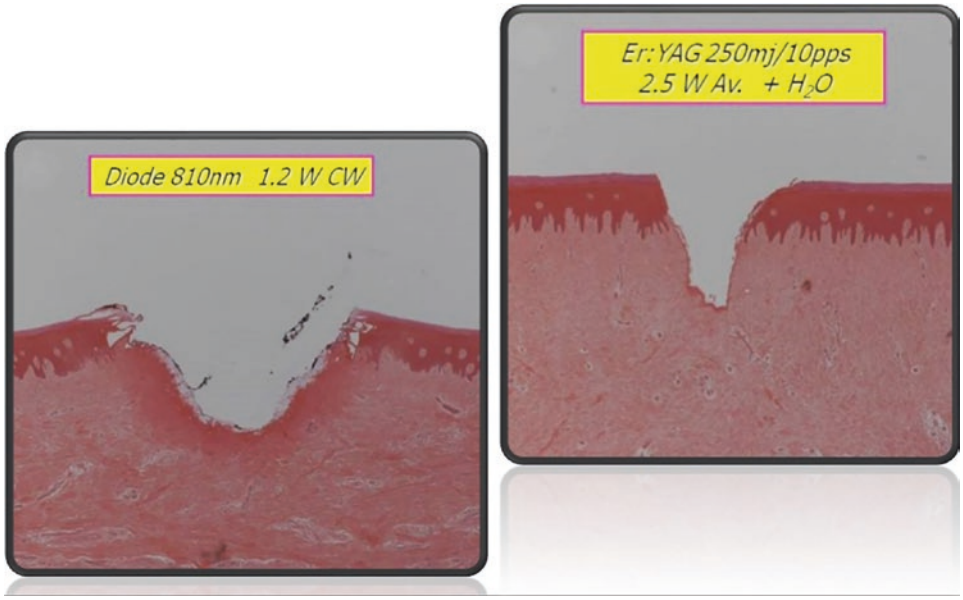


Fig. 3.12 Histological representation of two laser wavelengths (diode 810 nm and Er:YAG 2940 nm interaction with pig mucosa in vitro). This demonstrates the progressive crater-shaped incision with shorter wavelengths and a "V"-shaped incision with longer wavelengths

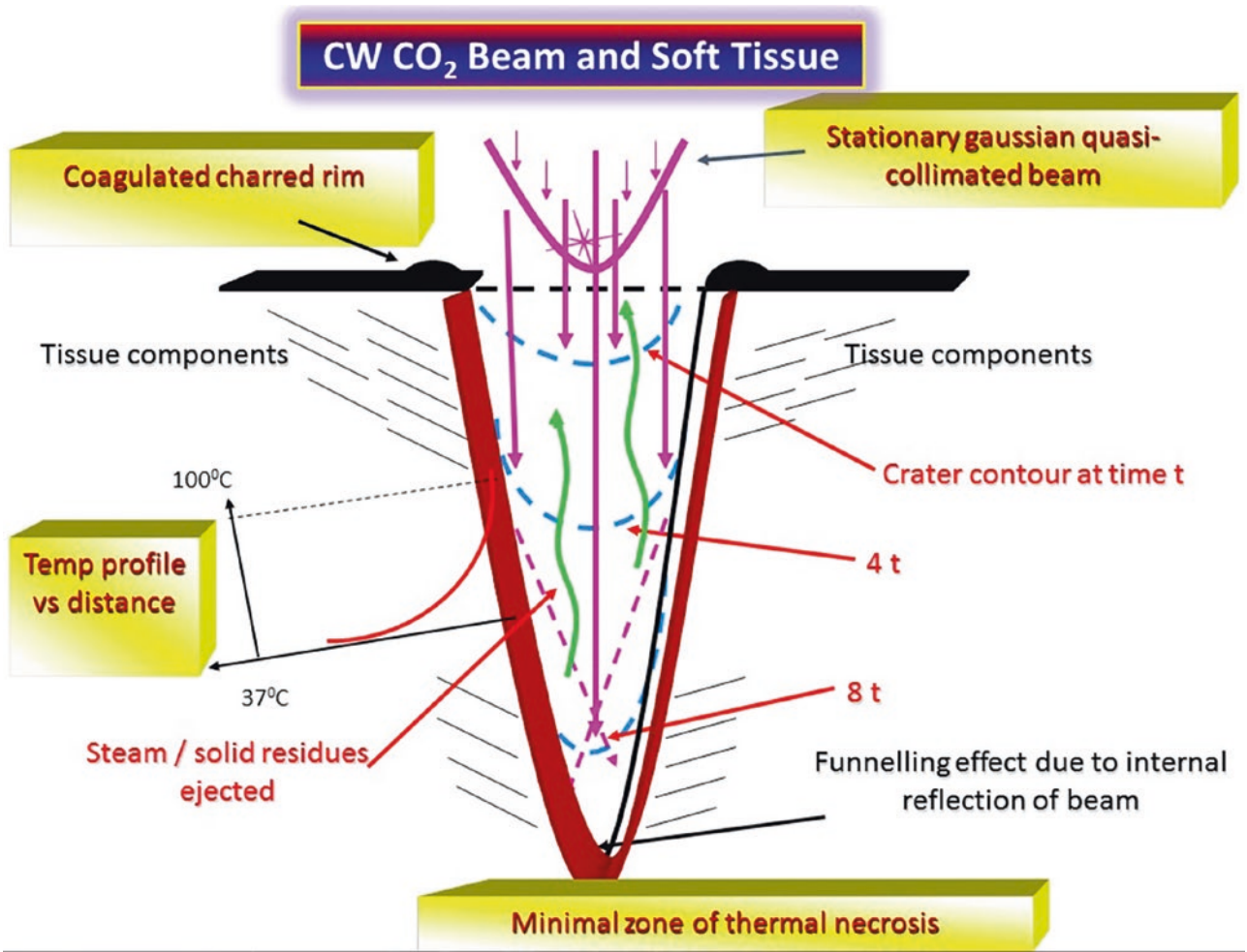


Fig. 3.13 Graphic representation of far-IR (and potentially mid-IR) laser photonic energy interaction with oral soft tissue. (Graphics: S. Parker after Fisher J.C. 1993)

the lack of post-operative inflammation and uneventful healing. Inasmuch as many claims are anecdotal, often if not always, the need for dressings or sutures can be avoided, and irrespective of the laser wavelength employed, all soft tissue healing will be by secondary intention in that it will be impossible to oppose the cut tissue edges to their original alignment. Of note, however, is the phenomenon of lack of post-incisional contamination by bacteria, due to a possible sterility of the cut surface [37] but certainly through the protective layer of coagulum of plasma and blood products—a tenacious film that allows early healing to take place underneath [38]. Additionally, studies with longer wavelengths show that there is a lack of fibroblast alignment associated with the incision line and consequent reduced tissue shrinkage through scarring [39]. Such findings are often borne out in the clinical setting.

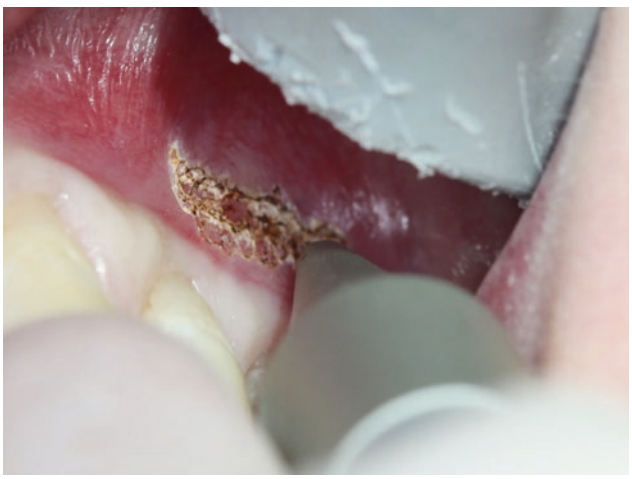


Fig. 3.14 Clinical example of a mucosal incision using a CO_2 laser. In the absence of a water spray, note the build-up of eschar, which can be easily removed with damp gauze to minimize thermal damage

3.9 Laser Photonic Energy and Target Oral Hard Tissue

Oral hard tissue includes cortical and trabecular (cancellous) bone and components of deciduous and permanent teeth (enamel, dentine, cementum). Within this group, it acknowledges the association of dental caries, being the predominant reason why teeth are subject to surgical intervention.

In common with laser-tissue interaction mechanisms described above, the current limitations of operating parameters of those lasers that are commercially available in dentistry center on the targeting of chromophores within the host tissue.

From the development of early lasers to ablate dental hard tissue, the predominant chromophore has been water—both interstitial "whole molecular" water and OH^- radicals forming part of the carbonated hydroxyapatite molecule ($[Ca_{10}(PO_4)_{6-Y}(CO_3)_Z(OH)_2] + H_2O$). Prime tissue groupings of oral hard tissue are listed by percentage of constituent structural elements [40, 41] in Table 3.4.

Absorption curves for both water and carbonated hydroxyapatite between wavelengths of approximately 3.0 and 10 μm are graphically listed in Fig. 3.16. The

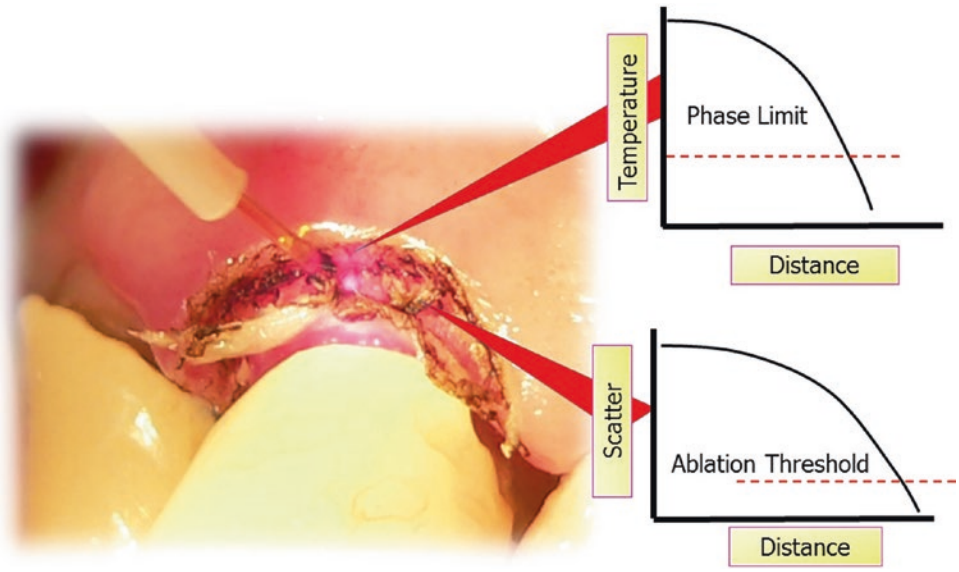


Fig. 3.15 At the point of surgical ablation of tissue, two intratissue gradients predominate. One is a thermal gradient, and with distance, a reduction in temperature will define a point where the temperature provides tissue stimulation. The other gradient—scatter—can produce a similar point-at-distance stimulation wherein biomodulation effects predominate

Table 3.4 Oral and dental hard tissues have structural components that may be viewed as potential chromophores—mineral, protein and water. For each tissue, the percentage of each of these chromophores will differ and define a level of laser-tissue interaction with a suitable laser wavelength

Tissue	Component chromophore as percentage		
	Mineral HA/CHA (%)	Protein collagens I and II (%)	Water (%)
Cortical bone	65	25	10
Cancellous (trabecular bone)	55	28	17
Tooth enamel	85–90	1–3	4–12
Tooth dentine	47	33	20
Tooth cementum	50	40	10
Dental caries	>5	70	25

carbonated hydroxyapatite molecule (CHA) is a relatively complex inorganic molecule with a parent calcium chain supporting radicals of phosphate, carbonate and hydroxyl subgroups. Additionally, within clinical specimens, there is whole-molecule free water; each radical is capable of preferential absorption, and peaks occur to indicate that laser interaction is possible, assuming that correct spatial and temporal operating parameters are used [42].

As was seen in the earlier table (Table 3.2), there is insufficient energy associated with an incident photon of erbium YAG (2940 nm) whose value is 0.42 eV, to break the atomic bond of a hydroxyl radical (value—4.8 eV). By the same measure, the dissociation energy within the crystal lattice of hydroxyapatite at 310 eV is two orders of magnitude greater [14]. Given that successive photons within a stream of irradiance will lead to progressive molecular vibration in the target structure, it follows that the chief goal of laser ablation of oral hard tissue would be the induced phase transition of water to vapor (steam), leading to the dislocation and explosive

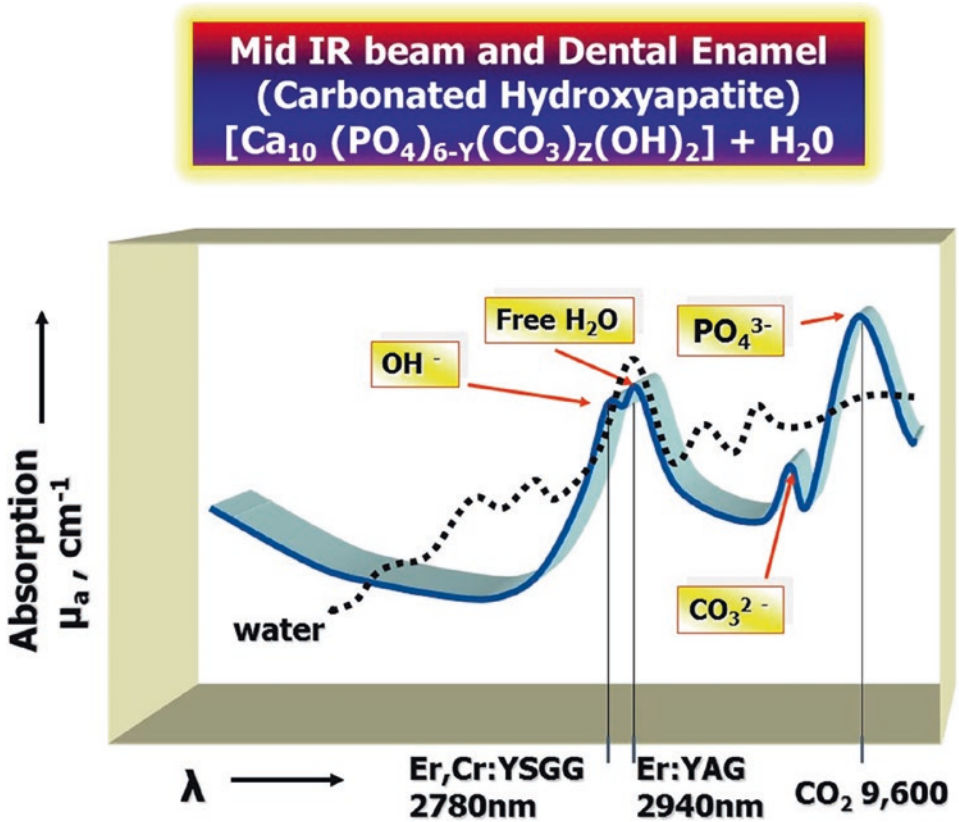


Fig. 3.16 Absorption peaks for water and CHA exist coincidentally for both Er,Cr:YSGG and Er:YAG wavelengths. High absorption exists in the phosphate group of CHA, coincident with CO_2 at 9300 and 9600 nm. (Source: Parker S. BDJ 2007;202(8);445–454)

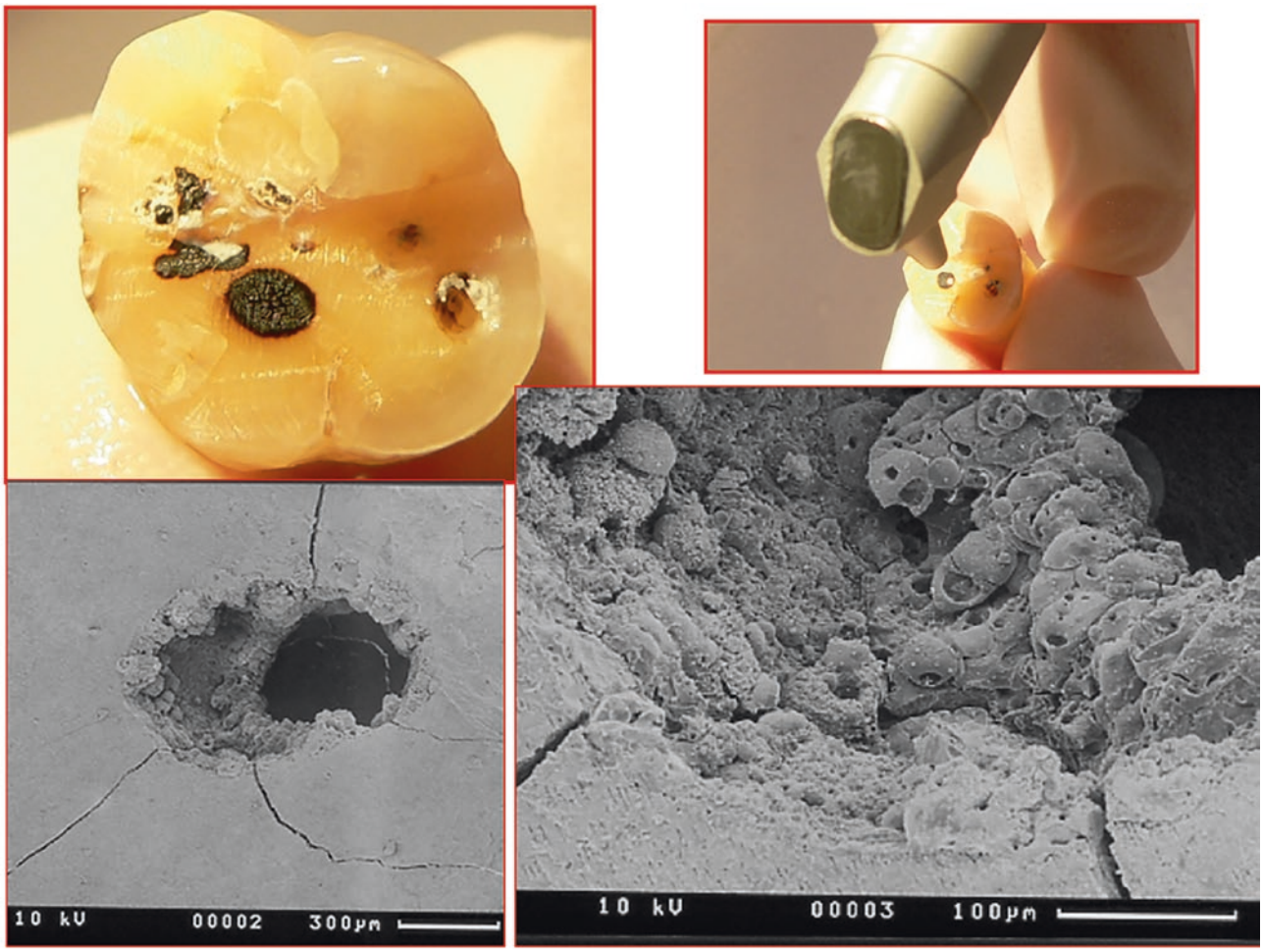


Fig. 3.17 In vitro exposure of molar tooth to CO_2 laser irradiation and SEM examination. Globules of melted and re-solidified (amorphous) hydroxyapatite are present, with large voids and gross disruption of structure. Average power 1.5 W CW with no water spray

derangement of the surrounding crystal lattice. Both erbium (erbium YAG, erbium chromium YSGG) laser wavelengths have free-running pulsed emission modes (pulse width 50–150 µs), which give rise to high peak power levels (>1000 W). Such power levels result in an instantaneous, explosive vaporization of the water content of enamel and dentine, which leads to dissociation of the tissue and ejection of micro-fragments [43].

The increased water content of caries results in rapid and preferential ablation of such material compared to normal enamel and dentine; to some extent, this may allow cavity preparation to be accomplished with a more conservative preservation of intact dental tissue.

Sustained exposure of hydroxyapatite and carbonated hydroxyapatite to laser irradiance will quickly render the structure to overheating, first to drive off any residual water and then to rapidly melt the mineral and produce signs of carbonization (Fig. 3.17). It is evident that sufficient heat may be produced to cause the melting of hydroxyapatite (several hundred degrees Celsius) and associated thermal cracking. Of course, such temperatures would lead to direct pulpal damage through heat conduction.

In consequence, the interactions of high-intensity laser irradiation with bone and dental hard tissues are the result of a photothermal action [44] targeting both molecular and interstitial water.

With an appropriate laser wavelength such as the Er,Cr:YSGG (2780 nm) and Er:YAG (2940 nm) and operating parameters configured to maximize interaction, together with adequate coaxial water spray, the outcome is completely different. With both enamel and dentine, the outcome of the "explosive" vaporization and ejection of tooth fragments results in a clean-cut surface, without smear layer often associated with rotary instrumentation. Due to the outward dissipation of energy, there is minimal thermal rise within the structure of the tooth and conduction to the pulp.

The fragmented appearance of cut enamel (Fig. 3.18) especially was historically thought to enhance the facility for bonding of restorative resins and composites without the need for acid etching. However,

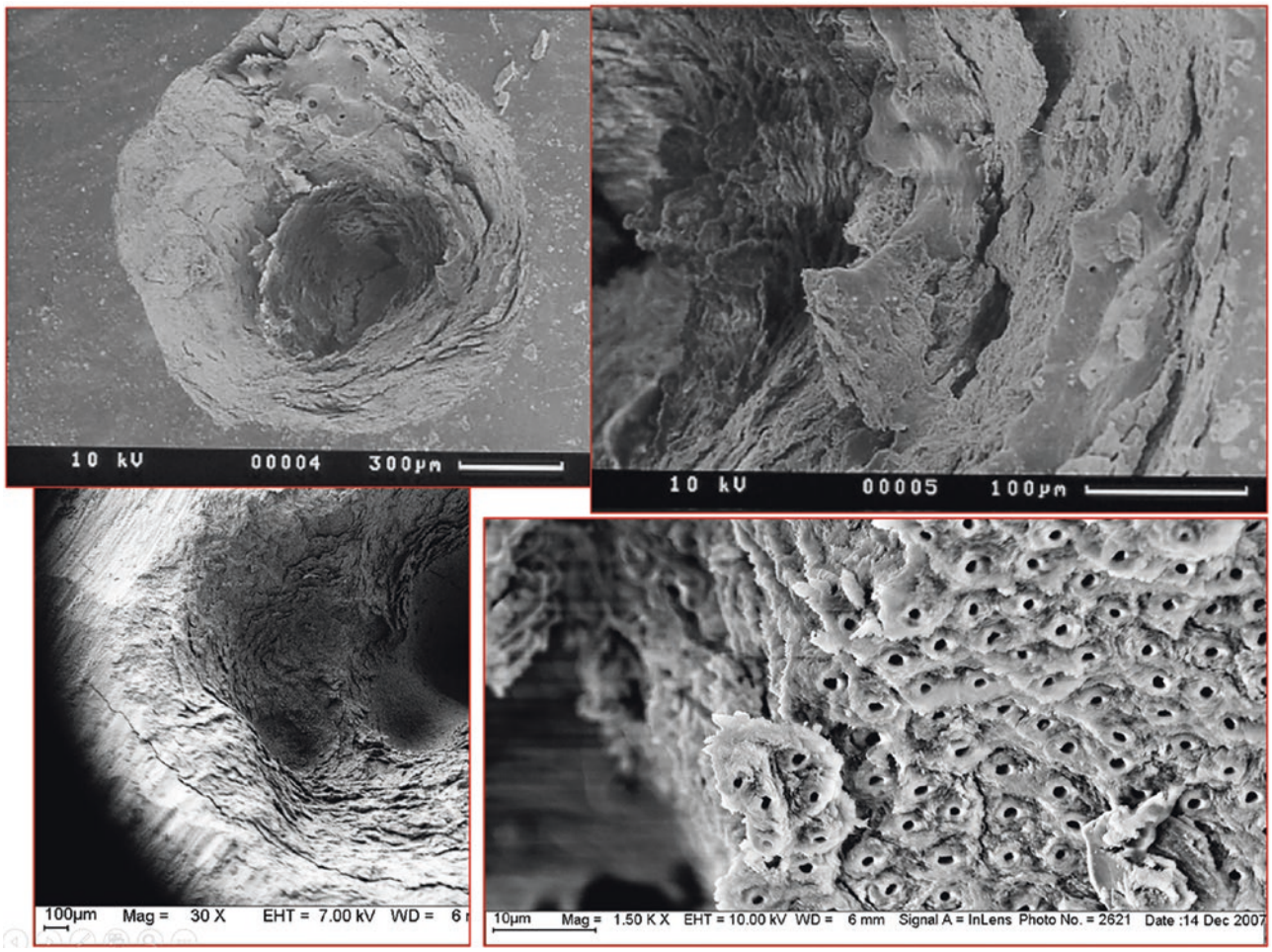


Fig. 3.18 SEM examination of tooth structure exposed to Er:YAG laser irradiation. Top images L and R—enamel structure showing evidence of dislocation and some fragmentation of the cut surface. Lower images L and R—similar cut surface of dentine, showing absence of smear layer and open tubules

many studies have highlighted the fragility of the cut margin in enamel and subsequent failure of the restoration margin as weakened tooth fragments gave way under tensile stress, with resulting failure and secondary caries risk [45, 46].

Mid-IR laser beam interaction with enamel (and to some extent also with dentine and bone) is a combination of temperature and pressure [47]. Both can be seen to rise rapidly during the pulse train of a clinical ablation procedure. The increase in volume as water vaporizes (1:1600) is significant and will give rise to significant rise in pressure just prior to the explosive dislocation of the enamel structure. As pressure rises, the continued vaporization leads to increase in temperature, resulting within the micro-confines of the interaction in "superheating" of the water and temperatures of several hundred degrees Celsius.

Furthermore, there are additional characteristics of laser ablation of hard tissue, surrounding the use of coaxial water, a necessary component to both aid excessive temperature rise and help wash away debris as a result of ablation. In addition, commercial models of both lasers use coaxial water spray to aid dispersal of ablated tissue and to cool the target [48], in a process called "water augmentation" [49]. According to this study, when dental hard tissues are irradiated with Er,Cr:YSGG and Er:YAG emission wavelengths with an additional thin water layer, the cutting efficiency increases at the same time that the pulp temperature decreases. However, the thickness of water layer should be well controlled to avoid a compromise in cutting efficiency and the blurring of the visual field.

As a liquid, water has a moderately high surface tension (72.8 mN/m at 20 °C), and this accounts for the intact film that may surround the tooth surface during water-augmented laser irradiation. In a further study [50], the influence of water thickness was investigated.

During ablation, the stream of photons is emitted in a free-running train of microsecond pulses (>100 µs), and any water between the laser delivery tip and the tar-

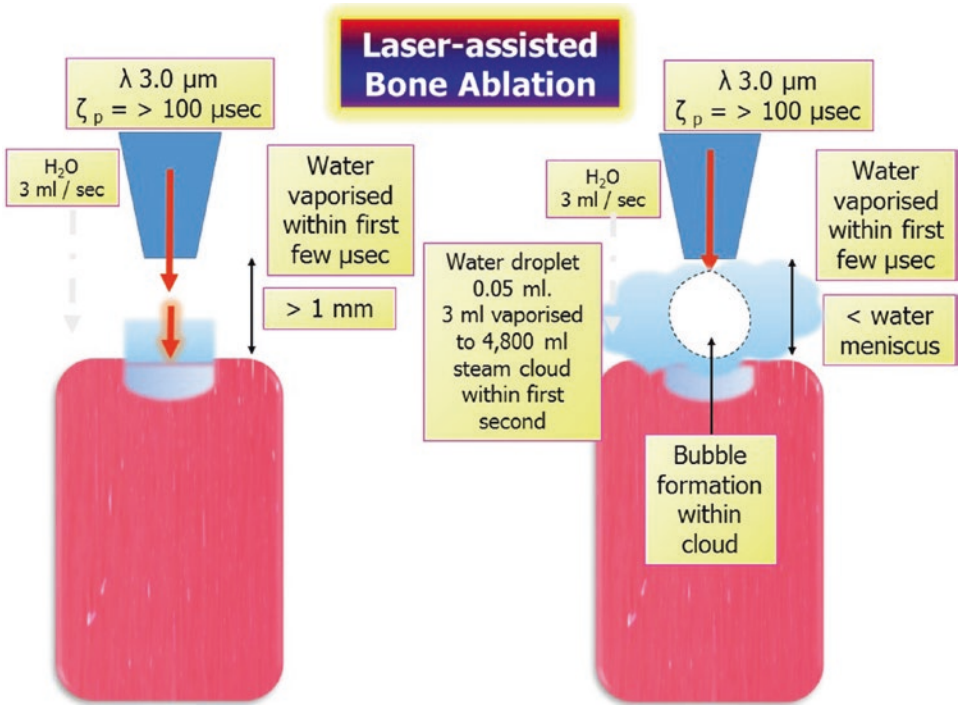


Fig. 3.19 Diagrammatic influence of the tip-to-tissue distance and influence of a contiguous water film, facilitating laser-induced cavitation phenomena. (Adapted S. Parker from: Mir M, Gutknecht N, et al. Lasers Med Sci (2009) 24:365–374 [50])

get would be vaporized during the first >30 µs of each pulse, allowing successive photons to interact with the target. If the tip-to-target distance is greater than the distance wherein the integrity of the water meniscus is maintained, the photon stream will pass through air before interacting with water at the surface of the tissue.

However, in circumstances where the delivery tip is close enough to the tissue surface and due to the surface tension of the water, a continuous envelopment of that distance occurs, the vaporization of the water film happens as before, but the vapor is contained within a rapidly expanding bubble; as it collapses, it gives rise to a cavitation phenomenon, and associated pressure waves may be sufficient (50–100 MPa) to initiate laser-induced "tripsy" (disintegration) of the tooth surface (Fig. 3.19).

This concept is exciting and in common with the similar phenomenon of laser-induced cavitation in water using wavelengths at 3.0 µm that may occur in endodontic and laser-assisted osteotomy procedures. With reference to the study by Mir et al. above [50], the staging can be summarized as follows:

- Energy intensity in the first pulses leads to absorption in the first µm layers of water opposite to the tip.
- Bubble formation with higher output energy density bubble dimensions were not clear, and a cloud-shaped appearance of laser-water interactions was recorded.
- In, for example, 140 µs pulses, after approximately 20–30 µs of the start of the pulse curve, a plume of vapor (comparable with a cloud) covers the tissue surface.
- The suction force exerted by the collapsing bubble and by the impact of the high-velocity jet generated during bubble collapse results in tissue ablation.

It remains to be seen to what extent this contributes to the "classic" understanding of 3.0 µm-mediated hard tissue ablation, but it is worthy of note that superheating of the vapor and hyperbaric pressure phenomena play a part.

In consideration of the ever-changing nature of available technology and its incorporation into laser wavelength choice, a precise irradiation parameter must be chosen in order to avoid collateral damage. This has importance no more than in terms of hard dental morphological damage, such as surface carbonization or cracking, which could produce structural and aesthetic damage and post-operative complications such as transient pulpitis. Moreover, the energy densities used must be safe with regard to pulp and periodontal tissue vitality [51]. Studies have indicated that temperature increments above 5.6 °C can be considered potentially threatening to the vitality of the pulp [52], and increments in excess of 16 °C can result in complete pulpal necrosis [53]. In comparison with rotary instrumentation, pulpal temperature

rise is minimal when erbium laser wavelengths are employed in cavity preparation [54].

Until recently, the commercially available CO_2 laser has been predominately a soft tissue ablation tool. The CW and gated CW emission modes of the 10,600 nm wavelength, together with an absence of coaxial water to aid tissue cooling and disperse ablation debris, give rise to rapid overheating of tooth tissue, cracking, carbonization and melting, which has made its use in restorative dentistry impossible [55–57].

However, due to the "four-level" nature of photon generation within the laser cavity, three major wavelength emissions occur—at 9300, 9600 and 10,600 nm. The longer wavelength is easier to manipulate from a technical point of view and has predominated the availability of CO_2 laser in clinical therapy. Absorption in water is a strong feature at this far-IR range, but the shorter 9300 and 9600 wavelengths are also strongly absorbed in the phosphate radical of the hydroxyapatite molecule. Investigation into this laser-tissue interaction has spanned almost 20 years [58], and in consequence, with a shorter CO_2 wavelength and manipulation of the emission to allow microsecond bursts together with a coaxial water spray to minimize heat generation, the interaction is both more positive and clinically acceptable [59].

If pulse durations in the range of 5–20 μs are used, efficient ablation occurs with minimal peripheral thermal damage [60, 61], and this has now resulted in a (for instance) commercially available laser unit emitting at 9300 nm with a specimen pulse duration of 10–15 μs and repetition rate of 300 Hz, demonstrating that enamel and dentine surfaces can be rapidly ablated by such lasers with minimal peripheral thermal and mechanical damage and without excessive heat accumulation [62, 63].

Commercial pressures may dictate the direction and speed of investigation into laser-tissue interaction with dental hard tissue that is based upon concepts of power density and pulse width as predominant factors to minimize collateral thermal damage, as opposed to the pure selection of chromophore-related laser wavelengths. By way of an example, in a most recent published investigation, a diode-pumped, thin-disk femtosecond laser (wavelength 1025 nm, pulse width 400 fs) was used for the ablation of enamel and dentin. Laser fluence, scanning line spacing and ablation depth all significantly affected femtosecond laser ablation efficiency and were predominant in comparison with the intuitively inappropriate choice of a near-IR laser wavelength of 1025 nm [64].

As may be seen with the erbium family of lasers at 3.0 μm, there is also the potential for easy disruption and ablation of composite restorative materials, and this has been the subject of published data [65, 66].

3.10 Laser Interaction with Dental Caries

Key to the efficient and safe laser-assisted removal of caries would be the facility to selectively ablate decayed dental tissue without causing injury to surrounding tooth tissue or pulp. Caries removal should be within the desired outcome of complementary aesthetic restoration of the tooth.

Prior to the development of the erbium family of lasers, the limited laser wavelengths and technology available within dental application meant that only visible and near-IR lasers (diode and Nd:YAG), together with CO_2 10,600 nm wavelength, could be investigated. Due to the rapid heat build-up with the available laser delivery parameters, attempts to remove (pigmented) dental caries received only limited success [67–69].

With the use of 3.0 μm wavelengths, the high absorption in water has transformed interaction. Caries as a demineralised residue of bacteriogenic acid action on enamel and dentine has varied structure but predominately a much higher water content than normal tooth structure. Interaction of laser photonic energy with this material will allow some selective ablation relative to the tooth tissue, and this remains a major advantage of lasers over more conventional rotary instrumentation [70, 71]. Additional positive indications support the development of the new generation of micro-pulsed CO_2 lasers and the ability to utilize the absorption on water at this wavelength. This has been the subject of intense research and investigation by the group at UCSF in San Francisco, USA [72, 73].

3.11 Caries Prevention

Laser-tissue interaction with hard dental tissue may pose difficult challenges relative to the wavelength and operating parameters, and this has been outlined above. Peripheral to the blunt outcome of thermal damage potential has been the careful application of several wavelengths to achieve a thermally mediated change in the carbonated hydroxyapatite structure of enamel, to change the crystal lattice to a more acid-resistant amorphous "glass-like" state. This change has been shown to occur with injudicious laser use on hard tissue, but with care, a number of studies [74–78] have proposed that many laser wavelengths may be manipulated to provide caries resistance in non-diseased teeth.

3.12 Laser-Tissue Interaction with Bone

The structure of osseous tissue of the maxilla and mandible resembles that of dentine in terms of proportional ratios of mineral, protein and water (Table 3.4). Bone

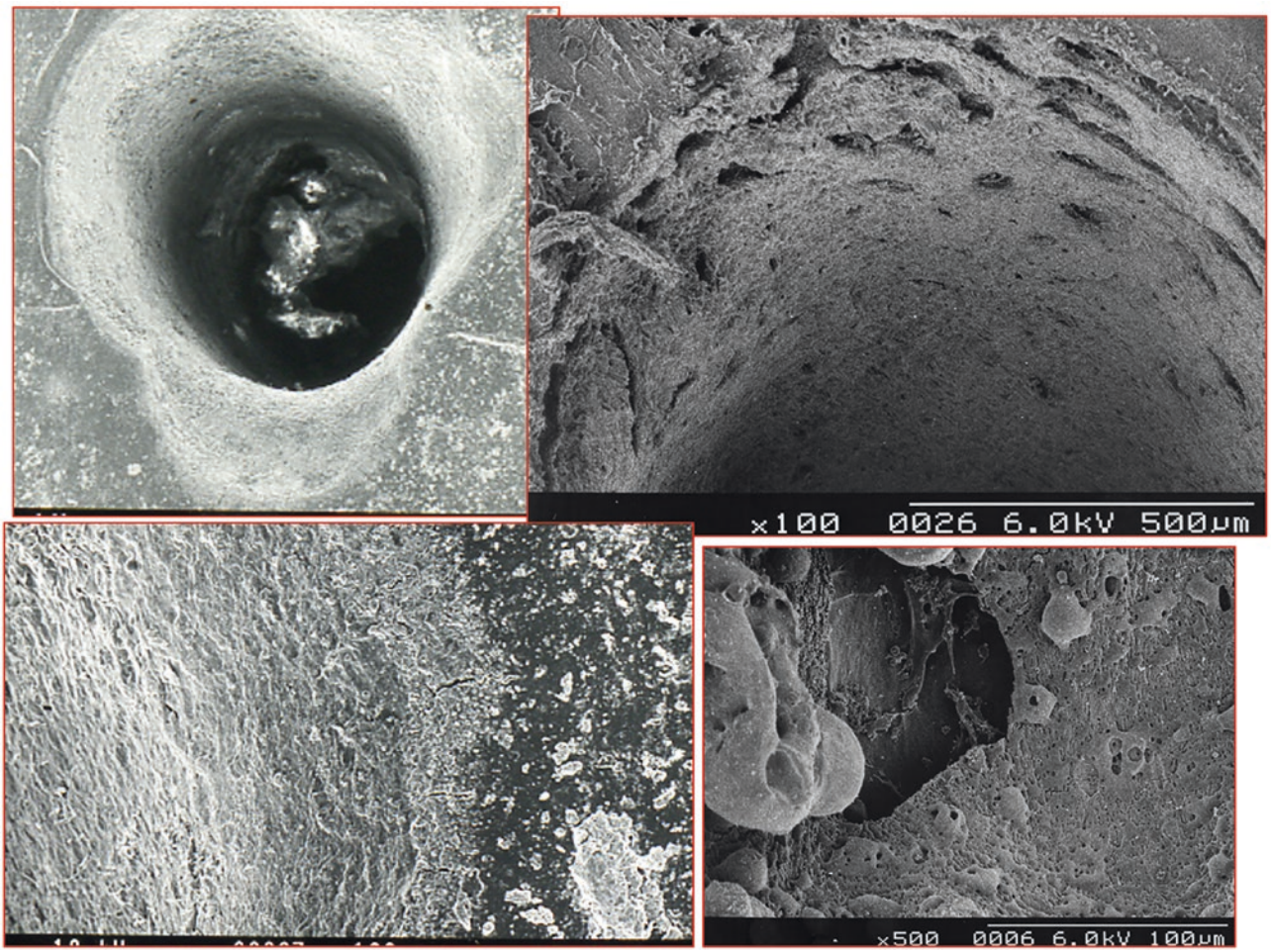


Fig. 3.20 SEM representation of laser interaction with osseous tissue. Top left, right and lower left images relate to ascending magnification of Er:YAG (2940 nm) with porcine rib bone in vitro. The cut is clean with minimal evidence of tissue disruption and thermal damage beyond the cut margin. Compare this to the image at lower right, using Nd:YAG (1064 nm), where apatite melting and large thermally induced voids are visible

is a much more dynamic tissue with reference to cell activity and turnover, compared to tooth tissue, and care must be observed to respect the potential for disruptive consequences of using inappropriate laser parameters. Although early reports of supportive use of CO_2 (10,600 nm) laser wavelength in surgical bone management are recorded [79], the potential for photothermolysis and collateral damage remains high. Laser ablation of bone with erbium laser wavelengths (2780 and 2940 nm) with high absorption in water defines a level of selective ablation through the vaporization of water and tissue structure disruption, and in this way, laser ablation of bone proceeds in a similar fashion to that seen in laser-mediated tooth tissue ablation. The higher water content and lower density of bone compared to enamel allow faster cutting, through dislocation of hydroxyapatite and cleavage of the collagen matrix (Fig. 3.20). This ease of cutting places the use of Er:YAG and Er,Cr:YSGG laser wavelengths as the preferred choice for laser bone ablation when compared to other wavelengths [80].

3.13 Laser-Tissue Photofluorescence

In an earlier section of this chapter, it was defined that fluorescence, as a form of sub-ablative laser-tissue interaction, is a luminescence or re-emission of light in which the molecular absorption of a photon triggers the emission of another photon with a longer wavelength. In absorbing incident photonic energy, some of that energy is expended and the difference between the absorbed and emitted photons ends up as molecular vibrations or heat. With re-emission, that energy loss is seen as a longer wavelength. This event is governed by the biophysical nature of tissue molecules involved (*termed fluorophores*) and as such can be the basis for optical scanning techniques used in caries detection in enamel

Table 3.5 Common fluorophores found within the oral cavity and dentistry. Each fluorophore is capable of excitation at specific light wavelength, and corresponding re-emission measurement can prove helpful in differential diagnosis of tissue change

Autofluorescence wavelength?			
Fluorophore	**Excitation (nm)**	**Fluorescence peak**	**Comments**
Tryptophan	275	350	Protein
Collagen	335	390	Connective tissue (CT)
Elastin	360	410	CT
Keratin	370	505	Surface analysis
Porphyrins	405, 630	590, 625, 635, 705	Cell mitochondria/metallo-, copro-, proto-porphyrins
Healthy enamel	405	533	
Caries	405, 488, 655	580–700	
Inorganic composites	655	Mean fluorescence intensity closely matched to healthy enamel	
GI composites	655	Mean fluorescence intensity closely matched to carious enamel	

Source: Kim A, Roy M, et al. J Biomed Opt. 2010;15(6):066026 [81]

and dentine and tomographic techniques in the scanning of soft tissue for neoplastic change.

The oral cavity provides substantial opportunity for scanning and fluorescence techniques, due to the ease of access of oral structures and the database of the excitation and emission wavelengths of individual tissue elements as well as non-biologic materials that may have use in dentistry [81]. In Table 3.5, it may be seen that through the choice of precise monochromatic laser wavelengths, predominately in the visible spectral range, the resultant re-emission would help to provide analysis of the target composition. A specific example might be that whereas healthy enamel exposed to a blue incident irradiation re-emits as a green color, the presence of porphyrin (pigment component of dental caries) re-emits at a longer red-brown color and would allow differentiation diagnosis of dental caries to be made.

Fluorescent and photodynamic diagnosis may provide screening facility or part of a hierarchical series of tissue investigation and must be regarded as an adjunct to a range of investigations—direct visual, microscopic and histologic examination, and genetic analysis—to provide support to the clinician, especially within the field of soft tissue health screening [82].

In oral soft tissue structure, disease changes the concentration of the fluorophores as well as the light scattering and absorption properties of the tissue, due to changes in blood concentration, collagen content and epithelial thickness. Such effects may be seen as follows:

- Recorded fluorescence signal will be lower in the case of hyperplasia—the epithelial layer shields the strongly fluorescent collagen layer.
- Excessive keratin production by lesions may produce an increase in autofluorescence intensity.
- Cell metabolism may increase with malignant changes, which changes the balance between fluorescent NADH (increase) and non-fluorescent NAD+ (decrease).

Many studies have been performed to investigate photodynamic diagnosis and fluorescence techniques in the oral cavity. These studies may be grouped in an attempt to address specific criteria of relevance in clinical assessment of neoplastic soft tissue change:

1. Whether autofluorescence imaging is capable of providing a higher contrast between a lesion and healthy tissue than white light or tactile and visual inspection? This is certainly the case for flat, early lesions [83, 84].
2. Whether autofluorescence imaging is helpful in differentiating between different lesion types, in particular between benign, dysplastic and malignant lesions? Overall, the specificity of autofluorescence imaging for distinguishing (pre) malignant from benign lesions does not seem to be very promising [85, 86].
3. The detection of unknown lesions and unknown extensions of known lesions, which would be useful for tumor demarcation: Indications have indeed been found that autofluorescence imaging is capable of detecting invisible lesions or invisible tumor extensions [87, 88].

Autofluorescence imaging might be appropriate as an easy-to-use, sensitive and inexpensive method for lesion detection, although further research is still necessary. In general, autofluorescence imaging may give good results for the distinction of lesions from normal mucosa. How-

ever, suspect lesions of the oral mucosa must be subjected to biopsy and other investigations, and certainly, it is inappropriate to place autofluorescence investigation in any role other than as an adjunctive scanning technique. If possible, autofluorescence spectroscopy could be used to find the optimal, most dysplastic location for biopsy, although the literature shows that autofluorescence is not specific enough for this purpose.

An allied area of laser-tissue interaction and spectroscopic analysis of re-emission is Raman scattering. This is a special, very weak form of light scattering in which energy is lost or gained to a molecule through a phenomenon known as inelastic scattering, where the frequency of photons in monochromatic light changes upon interaction with a tissue sample under investigation. The frequency (frequency and photon energy are inversely proportional) of the re-emitted photons is shifted up or down in comparison with original monochromatic frequency, and this is called the Raman effect. This shift provides information about vibrational, rotational and other low-frequency transitions in molecules.

Raman spectroscopy can be used to study solid, liquid and gaseous samples. Vibrational information is specific to the chemical bonds and symmetry of molecules. Therefore, it provides a fingerprint by which the molecule can be identified and has an important role within the area of tissue photo-analysis as it impacts disease and health.

3.14 Laser-Tissue Interaction and Photobiomodulation

Photobiomodulation (PBM) is the manipulation of cellular behavior using low-intensity light sources and the delivery of laser therapy (application of photonic energy at specific wavelengths) to induce a biological response through energy transfer. Sub-ablative photonic energy delivered into the tissue modulates biological processes within that tissue and within the biological system of which that tissue is a part. Key to the limits of benefit is the restriction of laser operating parameters to ensure that PBM has no appreciable thermal effect in irradiated tissue.

Phototherapy is characterized by its ability to induce photo-biological processes in cells [89]. This conforms to the first law of photobiology (light absorption by specific molecular chromophores). There is a so-called optical window in tissue (approx. 650–1100 nm), where the effective tissue penetration of light is maximized. The use of PBM therapy in patients almost exclusively involves red and near-infrared light (600–1100 nm) [90].

The absorption and scattering of light in tissue are both much higher in the blue region of the spectrum than the red. The principal tissue chromophores (hemoglobin and melanin) have high absorption bands at shorter wavelengths, tissue scattering of light is higher at shorter wavelengths and water strongly absorbs infrared light at wavelengths >1100 nm.

Wavelengths in the 600–700 nm range are chosen for treating superficial tissue, and those between 780 and 950 nm are chosen for deeper seated tissues, due to longer optical penetration distances through tissue with the latter group. Beam coherence is maintained as the laser beam penetrates the tissue and along with polarization may be an important factor in allowing the laser to effectively treat deeper tissues.

Incident photons of wavelengths as referenced above are absorbed into mitochondria and cell membranes of the target cells. Photonic energy is incorporated into a molecule to increase kinetic energy, activate or deactivate enzymes or alter physical or chemical properties of main macromolecules.

Growth factor response within cells and tissue may be seen as a result of increased ATP and protein synthesis [91], change in cell membrane permeability to Ca^{2+} uptake and cell proliferation, and overall, a cascade of metabolic "downstream" effects results in physiological changes resulting in improved tissue repair, faster resolution of the inflammatory response and a reduction in pain [92].

In summary, the results of laser-tissue interaction that promote PBM may be seen within three clinical areas of benefit—anti-inflammatory effects [93], analgesic and pain suppression effects [94] and effects that promote healing in the irradiated tissue [95].

Investigation into pain response during surgical laser use has revealed findings that are inconsistent with many anecdotal reports and may provide an opportunity for the essential subjective aspects of patient receptiveness to be accepted. Pain is a defence mechanism, and pain perception is innate and subjective. Equally, all stimuli applied to excess will result in pain.

Pain perception is multi-factorial and may be influenced through the following, either singularly or in combination:

- Emotion: fear, anxiety, stress syndrome, excitement
- Awareness: trust, previous experience, conditioning, e.g., hypnosis, activity subordination
- Threshold potential: age, infirmity, drugs, alcohol, social factors

The avoidance of pain during restorative and dental surgical procedures remains a strong factor in promoting patient acceptance of treatment, and many studies have been carried out to evaluate this in terms of laser-tissue interaction [96–98]. The use of the Nd:YAG (1064 nm) laser in developing pulpal analge-

sia, possibly through interference with the "gate theory" of neural stimulus propagation, was an early mainstay benefit of this laser following its launch into dental practice in 1990. However, investigation into the subjectivity or placebo effect has rendered its application inconsistent [99–101]. Chaiyavej et al. found that Er:YAG laser use, similar to rotary bur cutting of tooth tissue, caused neural response in both A and C intra-dental fibers [102].

Perhaps of greater significance in exploring this area may be the lack of tactile and thermal stimulation compared to rotary instrumentation during laser-assisted restorative dentistry. In seeking to understand the essentially anecdotal reports of soft tissue surgery using laser photonic energy in what is a thermally based interchange, there is the patient-centered factor of trust in the operator, together with a possible harmonization of micro-pulsed free-running emissions with regeneration potential of acetylcholine at synaptic junctions within the sensory neurone.

It is without question that when used correctly and with recommended operating parameters to maximize laser-tissue interaction, laser-assisted surgical procedures on soft and hard tissue are less physically injurious when compared to both scalpel and rotary bur. Patient acceptance, peer pressure and a general acceptance of "hi-tech" approach to treatment may all propose an enhancement of tolerance of sensory stimulation.

3.15 Conclusion

An overview has been presented to explore the physical and biological aspects of interaction of laser photonic energy with target oral hard and soft tissue. Any inconsistency may be viewed in terms of the precise mechanisms governing molecular energy dynamics but also the great diversity in tissue types and their close approximation within the oral cavity. The clinician is faced with technical challenges in manipulating any given laser wavelength to employ it as broadly as possible during clinical procedures; the additional facility of power density phenomena in helping to initiate an essentially photothermal event may help to deliver predictable and precise surgical outcomes. In Fig. 3.21, it is possible to appreciate the complex yet interconnected relationship that exists between incident laser choice and operating parameters, the consequent gradient of effects that result from such choice and how the laser-tissue effects so produced can represent a breadth of clinical application— in terms of both therapy and diagnosis. The photon distribution and concentration (fluence) delivered will influence the degree of (essentially) photothermal effect and consequent reaction of the target tissue. At low fluences, the benefits can be seen as diagnostic value and PBM (photochemical) effects; at higher fluences (with the same laser wavelength or another), the photothermal interaction predominates with ablative and per-

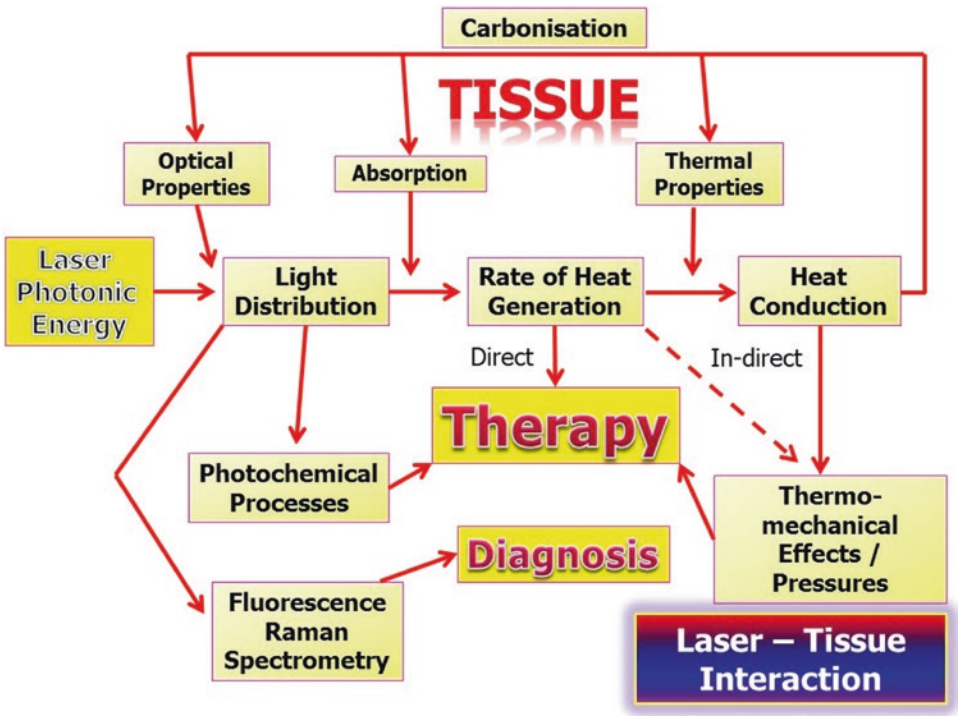


Fig. 3.21 Summary of laser-tissue interaction to demonstrate the breadth of application, relative to photon delivery

manent structural tissue change. The overload of fluence values may give rise to destructive collateral effects such as carbonization and consequent changes in tissue optical properties and absorption potential.

In summary, laser photonic energy offers a degree of "purity" through empirical properties—wave coherence and single (mono) wavelength. Unique wavelength value confers predictable photonic EM energy value through an inverse proportional relationship.

All matter has constituent atomic and molecular energy, consistent with intra-, inter- and extra-atomic and molecular binding forces. For any system, the "resting" gross energy value determines a ground-state physical form (solid, gas, liquid) relative to temperature.

Predictable ("pure") interaction with tissue can only occur if incident energy is absorbed by the tissue although levels of laser photonic energy may be viewed as being largely insufficient to overcome interatomic covalent or intra-lattice ionic binding forces within target tissue.

A better explanation might emerge, based on chromophore absorption of photonic energy leading to temperature rise within the system, and based on such assumption, laser photothermolytic interaction with tissue is mainly due to the indirect consequences of the conversion of EM photonic energy into thermal energy.

Photoacoustic and photochemical effects may be viewed as further consequential effects of primary photothermolysis.

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