The Use of Laser Energy for Therapeutic Ablation of Intraoral Hard Tissues

Position Paper: Science and Research Committee, Academy of Laser Dentistry

Adopted March 2007

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J Laser Dent 2007;15(2):78-86

EDITOR'S NOTE

This is the first of a series of position papers on various uses of lasers in dentistry, developed by the Science and Research Committee of the Academy of Laser Dentistry (ALD). This position paper was approved by the ALD Board of Directors in March 2007. The paper is not designed as a comprehensive literature review or as a detailed historical document. It covers aspects of the ablation of hard tissues relevant to dentistry, utilizing lasers currently available. The document will be revised and updated as technology changes and improves. It will require several other such papers to describe other applications. The reader is referred initially to the bulleted summary that highlights the key points of the paper, and then to the body of the text for supporting details.

SYNOPSIS

The present use of lasers in dentistry for the ablation of hard tissues is summarized in this publication together with a brief statement of scientific rationale.

SUMMARY

- When the wavelength of incident laser light is matched to the absorption band(s) of a target tissue component, light energy is converted primarily to heat which causes tissue change and/or ablation.
- To be therapeutically effective and efficient, it is necessary to deliver light of sufficient energy over time to effect tissue change, without causing unwanted collateral thermal damage by conduction of excess heat into the surrounding tissues.
- **Neodymium YAG (Nd:YAG, 1064 nm)**. While published studies have indicated some safe and effective usage, the clinical significance of the Nd:YAG laser wavelength was shown to be of only marginal benefit to the restorative dentist because of its very low absorption in hard tissues.
- **Continuous wave (CW) carbon dioxide laser (CW CO_2, 10,600 nm)**. The commonly available continuous wave CO_2 laser showed poor interactions with enamel, with reports of charring, cracking, and damaging heat buildup within tooth and bone structure.
- **Erbium YAG (Er:YAG, 2940 nm) and Erbium, Chromium YSGG (Er,Cr:YSGG, 2780 nm)**. The pulsed Er:YAG and Er,Cr:YSGG laser wavelengths are well absorbed by target hard tissue components, primarily water. These wavelengths offer safe use in cavity preparation.
- Both Er,Cr:YSGG and Er:YAG laser wavelengths are absorbed well in water, with the Er:YAG being somewhat more strongly absorbed in water than the Er,Cr:YSGG.
- The absorption in water is the primary absorption for these two wavelengths due to a relatively broad water band around 3,000 nm.
- In addition, there is a small absorption at around 2,800 nm by the hydroxyl group of the (carbonated) hydroxyapatite mineral of the tissues, but this is far outweighed by the water effects.
- Water is naturally present among the crystals in enamel, dentin, cementum, and bone deep into the tissue, filling every available pore. Enamel, dentin, bone, cementum, and carious tissue have, relatively, descending mineral density and ascending water composition.

- For both Er:YAG and Er,Cr:YSGG laser wavelengths, the laser energy is absorbed primarily by the water and is rapidly converted to heat, which causes superheating of the subsurface water, resulting in a disruptive expansion in the tissue.
    - The currently marketed mid-infrared (IR) laser wavelength lasers (around 3,000 nm) are free-running pulsed lasers. If the correct energy is used, application results in safe pulpal temperature rises of less than 5°C.
- Laser irradiation of enamel and dentin by Er:YAG or Er,Cr:YSGG lasers produces a "super-rough," micro-cavitated surface.
    - Studies have identified the fragility of laser-irradiated enamel, relative to the stability of the post-restoration margins. A combined approach of laser-irradiation and acid-etch techniques, to overcome such potential problems, is suggested. Regardless, there is the need to remove grossly overhanging and unsupported enamel to provide a stable post-restoration margin.
- The rate (speed) of ablation of dental hard tissue is a consequence of the amount of incident laser energy delivered to the tissue as well as the effects of wavelength, pulse duration, pulse shape, repetition rate, power density, the thermal relaxation time of the tissue, and delivery mode.
- Fluoridation of the tissue, incident angle of the delivery tip relative to the tooth, and presence of ablation products will affect the speed of ablation.
- The ablation threshold of human enamel has been reported to be in the range of 12-20 Joules/cm^2, and for dentin 8-14 Joules/cm^2 for the Er:YAG and Er,Cr:YSGG laser wavelengths, respectively.
- **Ablation of bone**. The development of Er:YAG and Er,Cr:YSGG lasers has enabled bone ablation to be carried out with minimal adjacent damage, and the use of erbium lasers in dento-alveolar surgery represents a less-traumatic experience for the patient.
- As with tooth tissue ablation, bone tissue cutting is a thermally induced explosive process and it is essential to maintain a coaxial water spray to prevent heat damage, which delays healing.
- Bone composition is very similar to dentin from the perspective of laser-tissue interactions. In maxillary alveolar bone, the speed of laser cutting is comparable with that of a bur, and slightly slower in the mandible.
- **Future developments and applications.** Several exciting new possibilities for the use of lasers on hard tissues are likely to become available to the practicing dentist in the near future. Two such examples are:
    - Selective ablation of calculus by wavelengths in the ultraviolet/blue region will offer less invasive dentistry.
    - Low microsecond-pulsed carbon dioxide lasers with a 9.3- or 9.6-µm wavelength have great potential for efficient and effective ablation of sound and pathological hard tissues, as well as modification of the mineral to increase resistance to caries attack.

INTRODUCTION

Laser light is unique in being emitted as a monochromatic, coherent, and collimated beam of non-ionizing electromagnetic (EM) energy which, for current dental purposes, spans the visible and infrared regions of the EM spectrum. Through a correct matching of incident laser wavelength with a target tissue element, light energy is converted primarily to heat, which causes tissue change or ablation.

For the present document the use of laser light for intra-oral hard tissue ablation will be the treatment that is primarily discussed. Other uses and mechanisms are known, but these are not the topic of the present paper.

In order for any laser-tissue interaction to be therapeutically effective and efficient, it is necessary to deliver light energy of sufficient value over time to effect tissue change, without causing unwanted collateral thermal damage by conduction of excess heat into the surrounding tissues.

LASER ENERGY AND DENTAL HARD TISSUES

Healthy oral hard tissues include enamel, primary and secondary dentin, cementum and alveolar bone. For the purposes of the application of laser energy in restorative dental procedures, demineralized and carious hard tissue must also be considered.

In addition to the prime interaction of laser energy with these tissues, there also exists a need to establish a rate of interaction that is commensurate with a time frame that allows such interaction to be clinically acceptable, by limiting the time for conduction of excess energy to occur, while at the same time being fast enough to be clinically acceptable.

Early investigations into the use of lasers for the ablation or modification of dental hard tissue were carried out using a ruby laser (red, visible, 694.3 nm), a carbon dioxide continuous-wave laser (infrared, 10.6 µm), and subsequently the neodymium:YAG laser (infrared, 1064 nm).

The Nd:YAG laser was the first to be marketed for soft tissue laser dentistry in the USA and numerous investigations were done to investigate its utility for hard tissue use.[1-9] Investigations included the ablation of (pigmented) diseased tissue, the antibacterial effect of this laser wavelength, and the possible effects on the vital dental pulp. While studies were published to establish some safe and effective usage, the clinical significance of this wavelength was shown to be of only marginal benefit to the restorative dentist, because of its very low absorption in sound enamel or dentin.[9] High fluences are needed for ablation unless the region to be ablated is pigmented or a pigment is applied to the surface. Furthermore, several studies drew conclusions that the Nd:YAG wavelength could cause unwanted heating side effects, such as cracking and melting of composite mineral structures.[1-9] Conversely, some workers published reports of the beneficial acid-resistance in enamel following exposure to low-power Nd:YAG energy and the resulting reconstitution of melted mineral in enamel.[10-12] However, what is not commonly realized is that an absorbing material (e.g., black ink) must be applied first for this wavelength to effectively increase the acid resistance of dental enamel, due to the very low absorption of Nd:YAG in enamel and dentin.

Other early studies on enamel ablation used the other commonly available laser wavelength, carbon dioxide (10,600 nm), but this laser gave poor interactions, with reports of charring, cracking and damaging heat buildup within tooth and bone structure.[13-14] The available carbon dioxide lasers at that time were continuous wave with no cooling water, which resulted in very high energy deposition in the hard tissue due to the combined absorption of this wavelength in both the mineral and in the water component. Excess heat was rapidly deposited with the continuous-wave mode rather than pulsed mode where there is time for the tissue to cool between pulses, and the energy density of the pulses can be better tailored to the needs. The continuous-wave CO_2 laser use resulted in reports of poor clinical benefit of this wavelength. The successful use of pulsed CO_2 lasers for ablation is expected to become a clinical reality in the near future.

The work of workers such as Keller and Hibst,[29] among others, illustrated the potential of the Er:YAG (2.94-µm wavelength) for effective ablation of dental hard tissues. This led to the development and marketing of free-running, mid-infrared wavelength (around 3,000 nm) lasers during the mid-1990s. This was a real achievement in addressing laser wavelengths that were complementary to target tissue elements, and offered clinically significant ablation rates that did not cause pulpal or collateral thermal injury if the right energy levels were used. The erbium:YAG (2940 nm) and erbium, chromium:YSGG (2780 nm) laser wavelengths are well absorbed by target hard tissue elements and appeared to offer safe use in cavity preparation.[15-17]

ABLATION OF HARD DENTAL TISSUES BY MID-INFRARED LASERS

It may seem incongruous, but to any clinician who may wish to use a laser in restorative dentistry, the high-speed rotary drill is seen as the "gold standard." Ease of use and speed are often accepted as plausible, even when several studies have shown that high-speed drilling gives rise to surface and pulpal temperature rise, tissue cracking, and unnecessary removal of healthy surrounding tissue during cavity preparation.[18-21] Such incongruity is compounded by the number of papers attesting to the precision, low thermal rise, and selectivity of mid-infrared laser wavelengths when used on dental tissue. Generally, the only drawback would appear to be the lower "speed" of cutting, when compared to the drill.[22-25]

Notwithstanding, the use of Er:YAG and Er,Cr:YSGG lasers in restorative dental procedures has progressed during the past 10 years, and within a given clinical setting the following factors will be significant in determining effective treatment outcomes:

- Target chromophores
- Mechanism of interaction
- Emission mode
- Pulse duration (pulsed or continuous wave, chopped)
- General thermal effects
- Relationship of laser action to cavity design and restoration retention
- Speed of "cutting"
- Power values.

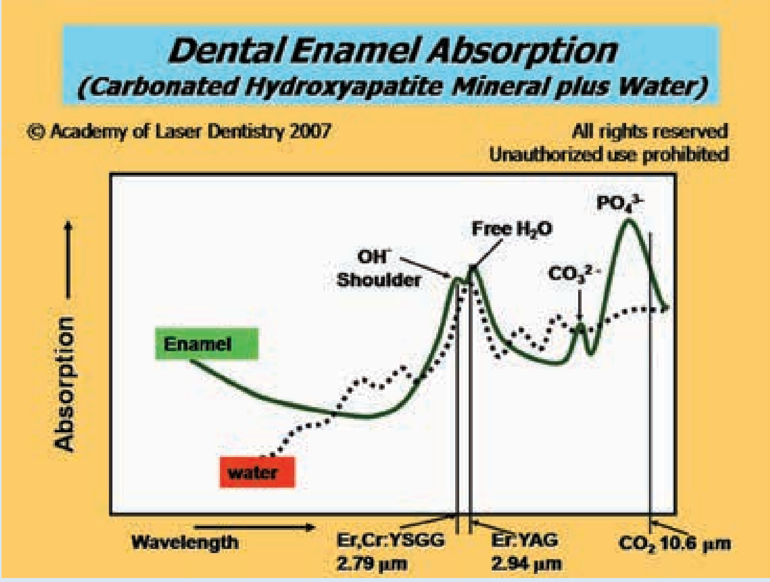
a) Target chromophores

Both Er,Cr:YSGG and Er:YAG laser wavelengths are absorbed well in water, with the Er:YAG being somewhat more strongly absorbed in water than the Er,Cr:YSGG. This absorption is several orders of magnitude greater than that seen with the Nd:YAG wavelength. The absorption in water is the primary absorption for these two wavelengths due to a relatively broad water band around 3,000 nm.[9, 51]

This is water that is naturally present among the crystals in enamel, dentin, cementum, and bone deep into the tissue, filling every available pore. In carious tissue there is an even higher quantity of water that replaces the lost mineral. The key to understanding hard tissue ablation by these wavelengths is that it is primarily due to this absorption in water and superheating of the water below the surface (see below, and Figure 1). Enamel, dentin, bone, cementum and carious tissue have, relatively, descending mineral density and ascending water composition.[26-27]

In addition, there is a small absorption at around 2,800 nm by the hydroxyl group of the (carbonated) hydroxyapatite mineral of the tissues,[28-32] but this is far outweighed by the water effects. Unfortunately many publications about laser effects on hard tissues have perpetuated the erroneous statements that dental mineral strongly absorbs these wavelengths. Not only is this incorrect, but it misleads us in understanding the mechanism of how ablation occurs due to laser application at these wavelengths and misdirects the use of these laser wavelengths.

Figure 1. Schematic absorption curve of dental enamel (carbonated hydroxyapatite (HA) plus water) and emission wavelengths of the Er,Cr:YSGG, Er:YAG, and CO_2 lasers. Carbonated HA exhibits a small peak at approximately 7,000 nm, coincident with carbonate $(CO_3)^{2-}$ ion absorption. The solid line presents the absorption bands for dental enamel with the tissue components labeled. The dashed line represents the absorption bands for water. The major broad enamel absorption band that spans the 3.0-µm (3,000-nm) region is due to the water content of the tissue, not the hydroxyapatite mineral. (Reprinted with permission from Parker SPA. The use of lasers in bone surgery. *J Laser Dent* 2007;15(1):9-13).



b) Mechanism of interaction

When incident laser energy directed onto hard dental tissue is absorbed by the prime chromophores, either water or carbonated hydroxyapatite, one of two effects occur. For both Er:YAG and Er,Cr:YSGG wavelengths this energy is absorbed primarily by the water and is rapidly converted to heat, which causes super-heating and a phase transfer in the subsurface water, resulting in a disruptive expansion in the tissue. Through this mechanism, whole tissue fragments are ejected and a hole is cut in the tooth, with little or no alteration to the mineral itself.

If laser light is effectively absorbed by the mineral, the crystals themselves may be heated above their melting point and some disruption of the crystal structure occurs with subsequent re-solidification in a different form, or direct ablation of the mineral, but there is also conductive heat transfer to interstitial free water.[33-37] Relatively high fluences are needed at these wavelengths for this to occur.

c) Emission mode and pulse duration

The emission mode of currently marketed mid-infrared wavelength lasers is defined as free-running pulsed. Currently commercially available lasers emit a pulse train of 50-250 microsecond pulses on average which, when delivered in rates of 3-50 Hz (pulses per second) values, represent duty cycle values of approximately 80%. While pulse durations are close to the thermal relaxation times of enamel and dentin, it is evident that there exists a need to examine further ultrashort pulse durations – and associated high peak power values – in an attempt to create sufficient ablative force without inducing collateral thermal damage.[38-39]

d) General thermal effects

The vital dental pulp is acutely sensitive to thermal change. Studies have established that rotary instrumentation can cause conductive thermal rise in excess of 20°C above 37.4°C.[40-41] With regard to laser irradiation of dental tissue, the explosive defragmentation resulting from water-assisted mid-infrared laser wavelengths allows much of the heat to escape from the cavity carried in the ablated particles, resulting in pulpal thermal rises of less than 5°C.[42-44] The affinity of mid-infrared laser wavelengths with water allows the main absorption to take place in demineralized tissue richer in organic material and with a higher percentage of water, thus protecting the sound underlying tissue with a reduced penetration of the beam. Contrary to this, the accumulation of ablation debris within a deep cavity can lead to "super-heating" which can lead to conductive heat damage.[45-46]

e) Relationship of laser action to cavity design and restoration retention

The emergence of conservative restorative cavity design, at variance with the classical "G.V. Black" model, represents a move toward minimal intervention with the development of acid-etch retention of composite resin restoratives. Laser irradiation of enamel and dentin by Er:YAG or Er,Cr:YSGG lasers results in a "super-rough," micro-cavitated surface that may predispose to ideal retention of composite resin. A succession of studies has identified the fragility of laser-irradiated enamel, relative to the stability of the post-restoration margins. Studies have proposed a combined approach of laser-irradiation and acid-etch techniques to overcome such potential problems.[47-52] Irrespective, there may well remain the need to remove grossly overhanging and unsupported enamel with a rotary bur, in order to either expedite cavity preparation or provide a stable post-restoration margin. Such consideration places patient care above the ideology of "pure" laser dentistry.

Furthermore, mention should be made as to the suitability of current lasers in the provision of full-veneer crowns. With regard to the preparation of single-surface veneers, using either direct or indirect materials, there is acceptance of the benefits that laser techniques may bring. Conversely, the use of lasers alone in the preparation of full-coverage indirect restorations is to be deprecated, due to the time required, the very large total energy input that would be required, and the possible irreversible damage to the pulp.

f) Speed of "cutting" as related to laser parameters

The rate (speed) of ablation of dental hard tissue is a consequence of the amount of incident laser energy delivered to the tissue as well as the effects of wavelength, pulse duration, pulse shape, repetition rate, power density, the thermal relaxation time of the tissue, and emission mode.[53-54] In addition, it is necessary to avoid the possibility of heat buildup in the tissue (and undesirable heat conduction to the pulp) and also to prevent the accumulation of products of ablation, or char.

Mid-infrared ablation of dental hard tissue has given rise to the concept of the existence of two wave fronts of interaction – an ablation front and a thermal front. It is important that the ablation front should always precede the thermal front, if the possibility of damaging heat rise is to be avoided. Studies, therefore, have looked at the effects of too much incident power and the buildup of ablation products, or their removal by means of a coaxial water spray.[55] It is also evident that the desire to match cutting speeds with those of rotary instruments has led to power delivery far in excess of that postulated by Keller and Hibst, relative to the ablation threshold of enamel. Coexistent with such power levels and heat conversion, studies have been carried out to determine the effect of reducing the pulse duration of the laser energy. It has been shown that by reducing the pulse duration, peak power values rise, ablation is more efficient, and heat transfer is minimized.[56-59]

In addition to the above, other factors such as fluoridation of the tissue, incident angle of the delivery tip relative to the tooth, and presence of ablation products will all affect the speed of ablation. Several reports have shown the effectiveness of addressing the delivery tip parallel to the axis of the enamel prisms in order to access the inter-prismatic, higher-water content structure.

Generally, the rate of tissue ablation with a laser, when compared to a high-speed rotary instrument, has given rise to claims of 80% slower in enamel, and comparable speed in dentin, when matched against a slow-speed drill. In addition, the use of sharp curettes in removing gross caries can allow lasers to be used within an acceptable time frame.

The debate over what constitutes a "recommended" power value for laser-assisted ablation of dental hard tissue is compromised by many conflicting factors, not least the danger of the anecdote. The ablation threshold of human enamel has been reported to be in the range of 12-20 J/cm^2, and for dentin, 8-14 J/cm^2 for the Er:YAG and Er,Cr:YSGG laser wavelengths, respectively. For an average laser delivery spot size, with the use of a free-running pulsed emission mode, this may equate to approximately 150-250 mJ/pulse. What is of paramount concern is the delivery of sufficient laser energy, within a minimal time, to achieve clinically acceptable ablation rates without causing adjacent tissue damage. Apart from those studies that have determined minimal levels of power necessary, there does seem to be a plethora of anecdotal reports. It would seem prudent for the clinician to follow the manufacturer's guidelines in establishing laser treatment protocols for a given laser, bearing in mind the differing operating parameters of air, water, spot size, and any power losses that may occur within differing delivery systems.

g) Bone ablation

Clinical procedures that may involve the cutting or ablation of bone include surgical extraction, periodontal surgery and infrabony pockets, clinical crown lengthening, and apicoectomy. The development of Er:YAG and Er,Cr:YSGG wavelengths has enabled bone ablation to be carried out with minimal adjacent damage, and the use of erbium lasers in dentoalveolar surgery represents a less traumatic experience for the patient when compared to the intense vibration of the slow-speed surgical bur.

As with tooth tissue ablation, tissue cutting is a thermally induced explosive process [60-61] and it is essential to maintain a coaxial water spray to prevent heat damage which would delay healing. Bone composition is very similar to dentin from the perspective of laser-tissue interactions. The mineral is similar, the protein content is similar, as is the water content.

In maxillary alveolar bone, the speed of laser cutting is comparable with that of a bur and slightly slower in the mandible, reflecting the greater mineral density of cortical bone. It is considered important that excessive power parameters be avoided to reduce the "stall-out" effect of debris and minimize blood spatter. Laser parameters of 350-500 mJ, 10-20 Hz (average power range 3.5-7.0 Watts) with maximal water spray appear to effect good ablation rates.

Studies into the healing of laser-ablated bone support the contention that the reduction in effects such as physical trauma, tissue heating, and bacterial contamination may lead to uncomplicated healing processes, when compared to conventional use of a surgical bur.[62-64] The microanalysis of the surface of bone that has been ablated using lasers shows little evidence of thermal damage, and any char layer appears to be restricted to a minimal zone of 20-30 µm in depth.[65-66]

FUTURE DEVELOPMENTS

There are several exciting new possibilities for the use of lasers on hard tissues that are likely to become available to the practicing dentist in the near future. A couple of examples are presented here. In each area years of research have set the stage for the development of commercially viable lasers.

Selective ablation of calculus by wavelengths in the ultraviolet/blue region is one example.[67-68] Ablation of carious enamel, dentin, and cementum, as well as bone may be more efficiently done with wavelengths not currently available commercially. The strongest absorption bands for the carbonated hydroxyapatite mineral of teeth and bone are in the 9.3 - 10.6-µm wavelength region, with 9.6 µm being the strongest. Pulsed carbon dioxide lasers have great potential for ablation and modification of mineral to increase caries resistance. Recent studies have shown that by matching the pulse duration to the thermal relaxation time of the tissue and optimizing the fluence per pulse, very efficient ablation of enamel, dentin, and carious tissue can be achieved with little peripheral damage.[69-70] Low microsecond pulse-duration lasers (e.g., 5-10 µs) of 9.3- or 9.6-µm wavelength have great potential. Furthermore laboratory studies have shown that similar irradiation conditions, but at lower fluences, can beneficially alter the mineral of enamel to make it more resistant to acid and consequently to inhibit caries progression.[31, 71-72] Clinical studies will be needed to confirm the viability of this methodology in the mouth. It will be possible to produce lasers that can ablate dental hard tissues and bone, while at the same time inhibiting subsequent caries progression, and keeping peripheral damage to a minimum.

SAFETY CONSIDERATIONS

All laser-tissue interaction using surgical lasers carries general and specific safety concerns. With regard to the statutory instruments ANSI 136.1 (2007) and IEC 60825-1/A2:2001, suitable precautions to protect intraoral nontarget tissue and patients' and operator's eyes and skin should be employed.[73-74]

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