Reviews

Effects of lasers on titanium dental implant surfaces: a narrative review

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Received: 26 February 2022 / Accepted: 12 August 2022 / Published online: 2 September 2022
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Abstract
Despite the many treatment modalities offered to prevent or manage peri-implantitis, there is currently a lack of high-quality evidence that supports any approach being regarded as a gold standard. Given that methods such as hand scaling with metal instruments and ultrasonic scaling may damage the implant surfaces, it is important to identify methods that are inherently safe for the surface being treated, and this is where interest in the use of lasers as alternative or adjunctive methods has arisen. This article provides a summary of the different types of lasers that can be used for the management and prevention of peri-implantitis. It also presents novel results from our research team related to the profile and surface characteristics of implants after treatment with different laser types and using different laser parameters. This review looks at the factors that should be considered when using lasers for the management or prevention of peri-implantitis. In conclusion, it is extremely difficult to formulate a reliable comparison between the available studies in the literature due to the high variability in laser types, settings and techniques used in each study. The review highlights the need for standardised studies in this field in order to provide recommendations to clinicians that would allow a more predictable treatment outcome.

Keywords Laser dentistry · Peri-implantitis · Titanium · Mid infrared laser · Near infra-red laser

Introduction

Despite the fact that very high success rates have been demonstrated for dental implants, they are susceptible to complications that may cause failure [1–3]. One such complication is plaque-induced peri-implantitis, which can be prevented by appropriate methods that decontaminate and debride the implant surface. Conventional methods of debridement of dental implant surfaces have been found to be insufficient and may cause damage or alteration to the implant surface [4, 5]. It is for this reason that lasers have been investigated for applications in the treatment of peri-implant disease.

Titanium implants have surfaces with macro, micro and nano-scale features that are designed to enhance osseointegration and thus long-term stability of the interface between the implant surface and the adjacent bone. In non-surgical treatment approaches for peri-implantitis with lasers, preservation of the micro and nano-scale features has been considered valuable, in the hope that these could contribute to bone formation in peri-implant bony defects. This differs from the situation during surgical treatment of advanced peri-implantitis where implantoplasty may be undertaken to remove surface features and simplify debridement. How much conservative non-surgical laser treatments may alter or damage implant surfaces has been an area of previous research, with studies using a variety of laser wavelengths and laser irradiation parameters on a range of implant surfaces. The large number of possible combinations of laser parameters and implant systems makes this a difficult area of literature to navigate. Lasers may absorb directly into the titanium or titanium alloy of the dental implant, as well as into the microbial deposits which are formed on its surface [6]. Absorption into titanium can lead to degradation of macro, micro and nano-scale surface features [7].

High survival rates of dental implants have been reported by a number of authors [1–3]. Systematic reviews of clinical outcomes with implant supported fixed dental prosthesis have shown 5-year survival rates of between 95.2% and 96.8% at 5 years [1–3] and between 86.7% and 92.8% at...
10 years [1, 3]. Despite this high survival rate, even successfully osseointegrated implants are susceptible to pathologic conditions that may eventually lead to failure [3, 4, 8]. Biological complications associated with dental implants may be defined as disturbances in the normal functioning of the implant as the result of biological factors that may affect the soft and hard tissues supporting the implant [9]. The incidence of biological complications such as peri-implantitis and soft tissue complications around implants has been reported between 8.6% and 9.7% at 5 years [1–3]. Where other risk factors are present, such as a history of periodontal disease, poor oral hygiene, smoking and diabetes, the risk of developing biological complications is even higher [10–12].

One of the main reported causes of post-integration failure is plaque-induced peri-implantitis [8, 13, 14]. Similar to periodontitis, the primary aetiologic factor for peri-implantitis is microbial colonisation of the implant surface, which drives an associated inflammatory reaction in the adjacent soft and hard tissues, resulting in bone loss [4, 8, 15–17]. To prevent this occurring, dental implants need monitoring and maintenance, including decontamination and debridement, at regular intervals throughout the lifespan.

In light of the similar aetiology of periodontitis and peri-implantitis, similar treatment modalities have been suggested [4, 18]. The treatment objective for both conditions is complete removal of the supra- and sub-gingival dental plaque biofilm, including areas which have mineralised to become calculus. Peri-implantitis treatment typically begins with a non-surgical approach to debridement and decontamination of the implant surface, which if unsuccessful will be followed by an open flap surgical approach and adjunctive treatments [16]. Both surgical and non-surgical therapy may include the adjunctive use of antiseptics or antibiotics [17, 19].

Conventional approaches to implant surface debridement have included mechanical treatment with hand scalers, sonic and ultrasonic scalers, using instrument tips made of plastic, carbon fibre or other materials in order to reduce damage to the surface. Other non-surgical treatment modalities include particle beams (air-powder abrasion), local delivery of antimicrobial agents and photodynamic therapy, in varying combinations with or without local or systemic antibiotics [20–28].

Whilst there are many possible treatment modalities, there is currently a lack of high-quality evidence that supports any particular approach being regarded as a gold standard [28].

It is important to consider whether novel approaches such as the use of laser treatments can give similar effects to existing conventional treatments but prevent damage to the implant surface. For example, in a study comparing the effect of Er:YAG laser treatment, chitosan brushes and titanium curettes, the curettes significantly altered the implant surface micro-texture, whilst the Er:YAG laser and the chitosan brush did not. Despite this, all three treatment modalities had a similar ability to remove Porphyromonas gingivalis [29]. Kotsovitis et al. in a 2008 systematic review of peri-implantitis therapy also pointed out that a significant limitation in the existing literature was that the duration of follow up may be too short to provide strong advice on which therapeutic methods have the highest efficacy over the longer term [17]. Given that methods such as hand scaling with metal instruments and ultrasonic scaling may damage the implant surface [3, 5], it is important to identify methods that are inherently safe for the surface being treated, and this is where interest in the use of lasers as alternative or adjunctive methods has arisen [30].

The overarching aim in treatment of peri-implantitis is to ensure that the treated implant fixture surface remains biocompatible, for adhesion of osteoblasts and fibroblasts. It is therefore necessary to balance the laser parameters being used for debridement against potentially deleterious effects on the surface and on the temperature of the metallic implant fixture. Ideal parameters for the therapeutic use of lasers in treating peri-implant disease should be determined for each laser wavelength in current use. Irradiation of the implant surface should maximise debridement and ensure thorough disinfection of the surface, as well as inactivation of toxins such as lipopolysaccharide, without causing deleterious changes to the implant surface. To optimise the laser parameters, it is necessary to consider not only the laser parameters but also the laser delivery system, since this will affect the power density and spatial distribution of energy provided onto the implant surface. Changing the wavelength has implications not only for absorption into the implant surface, into the biofilm and into adjacent targets such as hard and soft tissues, but also for the delivery of the laser energy through the delivery system. Adding to this are variables such as fibre or tip diameter and shape, and how these terminal parts of the delivery system interact with gingival crevicular fluid, blood or water. Laser debridement may require water irrigation to cool the implant surface as well as to flush away remnants of the biofilm which have been disrupted by the laser.

**Lasers for periodontitis and peri-implantitis**

A wide variety of lasers have been utilised in the treatment both of periodontitis and peri-implantitis [31–34]. The available wavelengths range from the visible spectrum (400–700 nm) through to the near infrared spectrum (700 nm–1500 nm), to the middle and far infrared spectrum (1500–11,000 nm) [35]. Within the infrared range, laser types of interest include the neodymium:yttrium aluminium garnet (Nd:YAG) laser at a wavelength of 1064 nm [36], the erbium:yttrium aluminium garnet (Er:YAG) laser at 2940 nm [37–42], the erbium, chromium:yttrium, scandium, gallium, garnet (Er,Cr:YSGG) laser at 2780 nm [43, 44],...
the carbon dioxide (CO₂) lasers at 9300 and 10,600 nm [36, 45] and various diode lasers at various wavelengths ranging from 810 to 980 nm [46, 47].

Of these lasers, the Er:YAG laser has been the most extensively researched for non-surgical periodontal therapy, providing clinical outcomes similar to those seen with conventional mechanical debridement [48].

**Laser use on dental implants and the accompanying surface effects**

**Erbium lasers**

In the treatment of peri-implantitis, the erbium family of lasers (Er:YAG and Er,Cr:YSGG) has received the most attention. Operating at a wavelength of 2940 nm, the emissions of Er:YAG lasers are highly absorbed by water, with very short laser pulses creating rapid fluid movement and cavitation events which can be useful for disrupting biofilms [49]. Due to the strong absorption in water, the Er:YAG laser has high bactericidal activity against periodontal pathogens. Its photothermal actions also inactivate bacterial toxins such as lipopolysaccharides [33, 50]. Several studies have shown that the Er:YAG laser can effectively remove deposits of dental plaque and calculus from implant surfaces without causing damage to the implant surface or producing excessive heat within the implant fixture [50–52]. Several studies have reported no significant changes in surface roughness when titanium implants are irradiated with Er:YAG lasers at 60 to 180 mJ/pulse for up to 2 min, when longer pulse durations are used such as 250–250 ms [53–55]. Using high peak powers and extended exposure times will however cause alterations to titanium implant surfaces [56]. With excessively high exposure parameters, Er:YAG laser irradiation may cause deleterious effects on titanium implant surfaces, which will reduce adhesion of osteoblastic cells [57]. Adverse surface changes such as melting and flattening of surface projections can be prevented by keeping the laser peak power low and also by using concurrent water irrigation [55].

Er:YAG laser treatment can have potent antimicrobial actions on biofilms on implant surfaces and can decontaminate the implant surface [58, 59]. Using a tip-shaped like a chisel can give an appropriate geometry for delivering the beam onto the implant surface, whilst reducing the risk to adjacent structures [60]. The geometry of beam delivery is an important consideration since a high peak power could roughen the surface, thereby provide an increased surface area on which biofilm could accumulate [61]. The laser parameters must be adjusted to prevent any adverse changes to the oxide layer on the titanium, as this could impair the subsequent attachment and proliferation of cells such as osteoblasts and fibroblasts [62]. The use of high pulse energies and short pulse durations (and hence high peak powers) is particularly problematic. Fenelon et al. in a 2020 report noted that using high pulse energies (e.g. 600 mJ/pulse) could influence machined titanium surfaces, causing roughening of the surface, with a complex pattern of peaks and troughs. Rough implant surfaces with micro and nano-scale features may be more prone to the absorption of laser energy, putting them more at risk of unintended modification of the surface at lower power settings [63]. On many occasions, it is impossible to completely remove the biofilm without causing surface alterations [64]. Removal of contamination on the outermost layer of titanium oxide using Er:YAG laser irradiation/micro-explosions under water irrigation should enhance success when treating peri-implantitis [65].

Dental implants with moderately rough surface features (Sa in the range of 1–2 mm) have an increased surface area compared to a machined surface, and this greater surface area should enhance bone ingrowth [66]. The presence of many microscopic projections on rough surfaces and the high surface area from these projections means that they tend to absorb more Er:YAG laser energy. This results in greater surface changes than with machined surfaces. The overall effects and their dose response are shown in Figs. 1 and 2. Note that these show the influence of pulse energies for Er:YAG lasers that are greater than those that would normally be used in clinical practice for implant treatment. High laser pulse energies cause degradation of the rough surface architecture (Fig. 1g, h, k, l). In contrast, smooth machined implant surfaces, on the other hand, undergo less surface alterations when the same laser parameters are used (Fig. 1a, b, e, f, i, j). When surfaces are lazed without accompanying water irrigation, photothermal actions are greater, and hence surface alterations are more noticeable than when the surface is treated in the presence of water (Fig. 1a vs. 1b; 1c vs. 1d; 1e vs. 1f).

It is important to take account of the laser spot size when the terminal aspect of the delivery system is changed. This will occur when changing handpieces, when changing tips and when changing the diameter of the fibre that is being used. Most fibres used with Er:YAG lasers have significant attenuation properties; hence, only a small proportion of the energy entering the fibre will be delivered onto the implant surface. The effect on Er:YAG laser energy on the surface of the titanium is also influenced greatly by the tip design. George and Walsh in 2009 assessed various flexible optical fibres for dental applications and reported details of beam profile and energy output [67]. A plain or bare tip will disperse the energy in a conical configuration over an angle of 16–20°. In this configuration, the inverse square law predicts that a close distance between the laser tip and the titanium surface will result in a high fluence, with resulting carbonisation. On the other hand, if the fibre has a conical shaped
end, this will disperse the laser energy over a large area, so the same laser panel setting will have little or no effect on the titanium surface even at the same distance from the tip to the target. As a result, there may be little or no surface alteration. Figure 2 below clearly shows the effects of irradiation using a 200 μm fibre optic tip, with no appreciable modification of the surface in machined samples, regardless of the irradiation conditions, with only minor alterations observed in rough surfaces in under dry irradiation conditions at 600 mJ per pulse.

If a focusing handpiece is used, the much smaller spot size gives a dramatic increase in the power density at the point of focus. This is why using a focusing handpiece (such as the model 2060 on the KEY-3 + laser, KaVo, Biberach, Germany) causes much greater surface changes than using a handpiece with germanium-doped fibres with plain ends (Fig. 2). Using the same panel setting, changing the handpiece or tip will alter the attenuation within the delivery system as well as the dispersion of energy from the tip. In practical terms, this means that thresholds need to be considered for different laser delivery systems as well as for different laser wavelengths [63].

A further consideration is what happens to the terminus of the laser fibre during use. A high fluence in the generation of plasma can cause degradation of the tip, changing its dispersion pattern for light as part of the cladding is burnt off or degraded at high temperatures. These types of effects can be seen with plain fibres when used at high pulse energies. Examples of this effect are shown in Fig. 2 for 200-μm plain fibre tips used in both dry and wet conditions on machined and roughened titanium surfaces (Fig. 2b, d, f, h). The treating dentist must
consider not just the profile of laser beam but also the type of implant surface being ablated when selecting the laser pulse energy to be used for treatment. If the incorrect settings are chosen, the effects on the titanium surface could range from carbonisation, as shown on a roughened surface (Fig. 2c, d, g, h) through to melting and fusion of the roughened surface, especially when treated under dry conditions at higher pulse energies (Fig. 2c, g).

A second laser in the erbium family is the Er,Cr:YSGG laser, with a wavelength of 2780 nm. This laser has also been used to treat peri-implantitis and gives efficient removal of dental plaque biofilm and contaminants from the implant surface [68, 69]. There is also some potential for surface alterations to titanium surfaces when pulse durations are very short and peak powers are very high [56]. Using this laser type with conical tips which disperse the energy will reduce the likelihood of adverse surface changes. This laser type when used with conical tips at a pulse energy of 50 mJ does not cause any damage to titanium implant surfaces. Increasing the pulse energy will begin to create small changes on the implant surface, e.g. at 84 mJ/pulse, including small areas of surface melting [70].

**Nd:YAG laser**

The Nd:YAG laser has been used in periodontal applications and is a potential candidate for the treatment of peri-implant pathology [71]. This laser operates at a wavelength of 1064 nm. Unlike the erbium lasers, this wavelength is poorly absorbed in water, limiting the ability to create strong cavitation events [71]; however, the wavelength is well absorbed in many metals including titanium and its alloys, which raises concerns for severe surface effects when used at anything above low peak powers [6]. Romanos et al. in 2000 documented damage to titanium plasma-sprayed (TPS) and hydroxyapatite (HA)-coated titanium discs when irradiated with a Nd:YAG laser [5], whilst Kreisler et al. in 2002 concluded that Nd:YAG lasers are not suitable for decontamination of implant surfaces due to their inherent risk of causing damage to the implant surface [72]. More recent studies have explored the possible use of Nd:YAG lasers at very low pulse energies (20–30 mJ), with air cooling, to reduce such concerns. Such low pulse energies have been shown to have bactericidal effects without dramatically altering titanium implant surfaces or excessively increasing the temperature of the titanium dental implant fixture [73–75].

**Fig. 2** Light microscopy images, at 10× magnification, of Er:YAG treated surfaces, using 200-μm plain fibre tips to irradiate a machined surface. 500 mJ dry (a) and wet (b); 600 mJ dry (e) and wet (f). Irradiation of a rough surface is shown as follows: 500 mJ dry (c) and wet (d); and 600 mJ dry (g) and wet (h). The KaVo KEY-3 model 2062 Er:YAG handpiece with a 200-micron plain tip was positioned at 1 mm from the target, with a spot size of 200 μm, giving power density values of 13.25 and 15.45 W/cm [2] for pulse energy settings of 500 and 600 mJ respectively.
As well as using a low pulse energy, a much longer pulse duration can be used, going from the microsecond range to the millisecond range. Goncalvez et al. in 2010 demonstrated the benefit of this approach using an Nd:YAG laser with a 1-ms pulse duration [76].

When multiple Nd:YAG laser pulses are delivered onto the same site, extensive changes to the titanium surface will result. Fenelon et al. in 2020 showed this effect for both machined and rough surface titanium samples irradiated with plain fibre tips and documented severe surface modifications, which increased in severity with increasing pulse energy [63]. The nature of such effects can be documented using measurements of surface roughness, as well as 3D imagery (Fig. 4). Melting and fusion of a roughened surface are particularly evident at 300 mJ/1 Hz (3 W) and above. The adverse effect of Nd:YAG laser energy on titanium has also been described in earlier work [50, 51]. A possible explanation for the more pronounced surface effect seen on a roughened (etched grit-blasted) surface than a machined surface discs lies in the greater surface area of the former. Fenelon et al. in 2020 used 3D profilometry to show the dense pattern of fine peaks and troughs on a roughened surface results in a greatly increased surface area [63]. This rough surface reflects less but absorbs more laser energy than a machined surface (Fig. 3) [48].

Regarding the influence of irrigation on laser effects on surface characteristics of machined surfaces, variability will occur depending on how well the laser is absorbed in water. With strong absorption in water (as seen with Erbium lasers), water that is located between the laser tip and the surface will be converted to steam, and the resulting expansion will displace further water. The formation of plasma between the tip and the surface can then result in changes to the surface. As water flows back after the steam bubble has

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**Fig. 3** An example of 3D reconstruction imagery. Sample images produced using TelyMap software: (a) 3D reconstruction of profile top view, (b) 45° angle zoomed image, and (c) a lateral view displaying the profile of a treated area. The samples were titanium surfaces irradiated with an Nd:YAG laser.
burst, there will be a cooling effect on the surface. This cycle will repeat with each laser pulse. If the pulse energy is very low and the pulse duration is long, all of the emitted laser energy from the tip may be absorbed in the water, and thus none will reach the surface. The temperature at which water boils is far below the melting point of titanium and its alloys, which means that effects such as melting are most likely due to direct absorption of laser energy into the surface, or a result of the plasma generated when the pulse duration is short. The subtle effects on the surface shown in Fig. 2 are best explained by the influence of plasma.

When water absorbing laser wavelength is being used, large volumes of water on the surface when combined with low laser peak powers will result in a dampening or protective effect for the titanium surface. This is relevant to water as a coolant as well as to crevicular fluid, saliva or blood. The presence of such fluid films will reduce the tendency for surface modification due to thermal effects, and this has been noted for both Er:YAG and Nd:YAG lasers [77, 78]. The latter emit energy which is poorly absorbed in water but strongly absorbed in titanium. Hence, a thin film of water in this situation will not protect the surface from thermal damage [79]. Dry or wet irradiation of machined titanium surfaces can therefore result in ablation of the surface. Irradiation under dry conditions creates a spindle-like pattern of peaks and troughs radiating from the centre, along with peripheral effects due to plasma (Fig. 4a, b, e, f, i, j). As the pulse energy increases, the extent of these surface

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**Fig. 4** 3D reconstructed images with corresponding microscopic photographs showing profile and surface characteristics of flat machined titanium surfaces irradiated with an Nd:YAG laser, using a plain tip on machined surface in dry and wet conditions. 50 ml/pulse at 20 Hz dry (a and b) and wet (c and d); 100 ml/pulse at 20 Hz dry (e and f) and wet (g and h); 150 ml/pulse at 20 Hz dry (i and j) and wet (k and l). A scale in mm is included at image borders of profiled images. A 200-micron plain tip was positioned at 1 mm from the target and used with a spot size of 200 μm. The power density values were 38.0, 75.0 and 109 W/cm² respectively for 50, 100 and 150 ml per pulse, corresponding to average powers of 1, 2 and 3 W respectively.
changes becomes more severe with extensive ablation and melting (Fig. 4i, j). As a result of some surface cooling and attenuation of plasma due to the presence of water, machined surfaces that are irradiated by the Nd:YAG laser under wet conditions do not show the spindled pattern, but rather severe melting and fusion of the surface (Fig. 4c, d, g, h, k, i) [63].

When titanium with a roughened surface is irradiated with the Nd:YAG laser using plain fibre tips, the surface will absorb the laser energy strongly, and this will also result in melting and fusion. As a consequence, there is loss of the micro-roughened surface provided by the manufacturer. For the same reasons as discussed earlier, this effect is greater in samples irradiated under dry conditions (Fig. 5) and is more severe as laser pulse energy and peak power increase. At very high pulse energies, the difference between irradiation under dry and wet conditions is less pronounced (Fig. 5). If the Nd:YAG laser is used with a conical fibre tip, the dispersion of the laser energy means that the irradiation conditions at the surface of the titanium are below the ablation threshold for surface modification.

Diode lasers

Because of their photothermal disinfecting actions, diode lasers have been used in the treatment of peri-implantitis. Common examples are the gallium aluminium arsenide lasers (from 670 to 830 nm), gallium arsenide lasers (810–840 nm) and indium aluminium arsenide lasers (904–980 nm) [80]. The 980-nm diode lasers appear to have little if any effect on titanium implant surfaces (titanium plasma-sprayed (PTS) or coated in hydroxyapatite (HA)).
even when used in continuous wave mode at powers up to 15 W [5].

Bacterial reduction has been demonstrated with diode lasers operating at 980 nm [76] and 809 nm [81], when used to treat bacteria grown on titanium with machined, TPS, HA coated or sand-blasted large grit acid-etched (SLA) surfaces, for average laser powers up to 3 W, with no damage to the implant surface. Complete elimination was not achieved, and this suggests a need for further optimisation of laser parameters based on the nature of the surface. The most challenging surface was the SLA surface, which has significant micro and nano-scale features.

When diode lasers are being used, consideration must be given to the safety of adjacent structures such as the alveolar bone [81, 82]. As with erbium and Nd:YAG lasers, optimisation of the laser parameters to suit the implant surface is necessary when using diode lasers. The wavelength of the diode laser will influence the extent of reflection versus absorption. Since diode lasers are available over a broad range of wavelengths, from the visible region extending into the near infrared, careful attention must be paid to how the diode laser energy interacts with blood, gingival tissues and the dental implant components. Some diode laser wavelengths will absorb very strongly into blood, such as those in the visible blue and visible green regions, and this will change the temperature responses at the site of irradiation [83]. For this reason, manufacturers of diode laser devices should provide detailed advice on the appropriate parameters for an implant decontamination. Avoiding continuous wave emission would limit extent of local heating, as would using low average laser powers [83]. In view of the large number of diode laser wavelengths available, comparison between studies is difficult, and each laser will need bespoke settings to maximise effectiveness and safety.

**Carbon dioxide (CO₂) laser**

The strong absorption of far infrared energy (10,600 nm) from CO₂ lasers in water makes them useful for the removal of contaminated soft tissue and residues without changing the surface characteristics of implants [84–86]. It has also been reported that CO₂ laser energy can kill microorganisms from the implant surfaces without changing the surface topography, since most of the incident laser energy is reflected from the surface [87].

Literature shows that average power settings for management of peri-implantitis have been in the range of 2 to 4 W [88, 89], with both chopped continuous wave and pulsed continuous wave emission modes being used [88–90]. The high rate of soft tissue ablation with this laser makes it suitable for exposing implants, debulking inflammatory tissue and ablating granulation tissue, to gain access to supragingival parts of implants. This laser type is not considered useful in non-surgical (closed) debridement due to challenges in delivering the energy to subgingival sites in a controlled manner so that the energy is directed laterally onto the implant surface rather than apically. There are a range of delivery systems and tips, but the challenges of directing the energy laterally within a tip that can be inserted into the periodontal pocket alongside an implant are yet to be resolved. Such access issues do not exist if an open flap surgical debridement approach is being used. Furthermore, similar issues with regard to the parameters and techniques were highlighted when CO₂ lasers were used for prevention of early caries. The available studies provide insufficient information, incorrect settings and parameters, or suggested solutions that were not applicable to clinical practice [91].

**Influence of the laser delivery system**

The specific action of each laser wavelength is dependent not only on the absorption of that laser energy into the target, but also on the characteristics of the delivery system. Depending on the system used, laser energy may be delivered to the target surface using optical fibres made of glass for visible and near infrared lasers or made of rare earth element oxide materials for near and middle infrared lasers. Other delivery systems used with metal and far infrared lasers include flexible hollow waveguides and articulated arms. Each delivery system can be fitted with handpieces or with tips or optical fibres, and the terminal part of the delivery system can either focus the energy or disperse it [92]. When tips or optical fibres are used, the design of the tip has a major influence on the spatial distribution of laser energy, and this influences how the laser interacts with the target [92]. Various modifications to fibre tips have been described, which alter the emission direction, beam divergence and spatial distribution of laser energy [92–96].

The logistics of delivering laser energy to the threads or subgingival regions of a dental implant are considerable. A range of side firing fibres and tips have been developed specifically for this purpose. Some of these are rigid, whilst others are semi flexible or highly flexible. The rigidity of the delivery system tip influences whether any tactile feedback is provided to the operator. Tactile feedback may help the operator correctly position the tip into the region between the threads of a dental implant. A plain fibre tip will disperse the energy in a forward direction with a conical-shaped emission pattern. A typical glass fibre or rare earth element oxide fibre will disperse the energy with the divergence of less than 20° [97]. This low divergence does not suit the situation of a periodontal pocket around a dental implant. It is hence important to use fibre tip designs with greater lateral emissions to direct laser energy on the implant bone interface. Various modifications to optical fibre tips have
been explored to increase the lateral emission of laser energy for the wavelengths of interest [97]. Techniques include grinding and polishing to give side firing or tapered tips [98], heating and pulling [99, 100] and chemical etching [101–103]. A technique of combined etching and abrasion steps has been shown to produce conical fibre tips with multiple facets in a ‘honeycomb’ surface pattern, giving greatly increased lateral and decreased forward emissions compared to plain fibres [97].

Factors influencing ablation

The currently accepted theory for the mechanism of erbium laser ablation of dental hard tissues (enamel, dentine, cementum and bone) is a photo-mechanical action based on micro-explosions of water within the target. Conversion of the water to steam generates an increase in pressure, leading to micro-explosions [104, 105]. These same mechanisms also apply to the ablation of dental plaque biofilms and dental calculus. The explosions are accompanied by photothermal disinfection and by rapid fluid movement caused by cavitation events. The combination of these effects results in decontamination of the implant surface [106]. In contrast, photothermal events dominate for the Nd:YAG laser and for diode lasers. Bacteria are inactivated by direct heating actions [107]. Some diode lasers (940 and 980 nm) have sufficient water absorption to generate fluid agitation, and this can assist in debridement of implant surfaces [108, 109].

Other factors that may influence the ablative effect of lasers on mineralised and non-mineralised dental plaque biofilms on implant surfaces include the presence of water [110], the laser pulse energy and the spatial distribution of laser energy [111], the beam diameter [112] and the pulse duration [113]. Use of a water coolant during laser irradiation attenuates temperature changes on the target and thus minimise adverse effects on adjacent tissues [77]. As discussed earlier, the presence of water can also influence the interaction of water absorbing laser wavelengths on implant surfaces [63]. Optimisation of irrigation protocols is important for both safety and effectiveness. When a water absorbing laser is being used, excessive irrigation will dampen the ablation effects [77]. Under clinical conditions, Er:YAG lasers typically with pulse energies in the order of 50–200 mJ and with a continuous water mist spray are used [63]. Several studies indicate that water spray or a moist surface is essential for effective ablation of dentine, as it not only reduces thermal stress but also reduces charring of the ablated surface [114–117]. Provided a moderately long pulse duration is used (in the order of 250–350 μs), no adverse effects on the smooth titanium abutment surfaces should occur [63]. Further work is needed to optimise water flow rates for effective ablation of implant surfaces. Considerations around irrigation should also consider whether the implant service being treated is supragingival or subgingival, and hence whether there is saliva or gingival crevicular fluid present.

Laser parameters for treating dental implant need to consider the pulse shape, pulse duration and pulse energy. The pulse shape (power versus time) is important because it determines how much of the pulse is above the ablation threshold of the target. In this situation, there are at least two targets with vastly different ablation thresholds—the dental plaque biofilm (whether mineralised or not) and the titanium or titanium alloy of the implant. There needs to be sufficient energy to ablate microorganisms in the biofilm, but not to ablate the titanium surface. The pulse duration influences the peak power. The greater the peak power, the more likely that cratering of the surface will occur and that plasma will be generated. Using longer pulses with lower peak power means that the likelihood of damaging the surface will be less. The influence of laser pulse energy on ablation has been well documented for dental hard tissues [118, 119], and the same principles in relation to surface alterations with erbium family lasers [50, 51, 53, 56, 68] and Nd:YAG lasers [5, 72, 76].

A further consideration is the beam profile, which is the spatial distribution of laser energy. All the energy can be concentrated into a single spot with a Gaussian beam profile, or the energy can be divided into a series of geometrical shapes, such as rings or dots, when the laser is in multimode operation. The configuration of the laser cavity influences the beam profile, and for this reason that can vary between lasers of the same wavelength. The energy distribution in the spot affects the laser to target interaction and the ablative effect of the laser [120]. Larger spot size means a lower power density and therefore a reduced ablative action. As the spot size becomes larger, any defects such as craters that form on the surface will be shallower, due to defocusing of the beam [112]. This defocusing action applies regardless of the spatial beam profile of the laser. Laser manufacturers should provide information about how the power density changes with distance for particular handpieces and fibres, stating actual measured values rather than calculated values, since actual measured values will take into account losses within the delivery system itself. Users can then make informed judgements regarding what the focused laser spot is likely to do on the target [111, 120]. Typically, manufacturers will provide information about the effective beam diameter that can be achieved with the delivery system components provided. Clinicians must bear in mind that changing the diameter of the fibre optic tip and the shape of the tip will alter the spot size [92].

Laser treatments of implant surfaces should avoid short pulse durations since these will have the highest peak power and the greatest ability to form plasma. The influence of pulse duration on ablation is well known in the literature.
for dental hard tissues, and the same principles apply with titanium and its alloys [113]. A short pulse duration can cause ablation at a lower pulse energy [113], whilst there will be little or no ablation at longer pulse durations for the same pulse energy [113]. This line of thinking explains how a titanium absorbing laser wavelength such as that from the Nd:YAG laser can be used safely to decontaminate implants, provided that an extremely long pulse duration is employed, to avoid causing surface alterations [76].

In addition to laser parameters, clinicians must bear in mind the nature of the target surface, both in a chemical sense and in terms of morphology. Different implant components may have compositional differences, such as grade 4 titanium being used for a fixture, but grade 5 titanium alloy (which contains aluminium and vanadium) being used for an abutment. With regard to the surface characteristics, a surface that appears smooth and highly polished to the naked eye will interact differently with laser light than the surface which appears matte. On reaching the surface, longer wavelengths of laser light may reflect even from surfaces that do not appear to be extremely smooth [35, 80]. Laser energy may reflect from highly polished titanium surfaces onto adjacent soft tissues, causing accidental ablation [80]. Clinicians need to carefully follow the manufacturers’ recommended maximum laser settings for treating implant surfaces and be aware of this risk of specular reflection. The issue has not been documented well in the literature for diode laser wavelengths, despite being well described for the far infrared carbon dioxide laser [121]. This laser type is not used widely in dental implant therapy, with diode lasers being much more commonplace. There is a significant gap in the literature regarding the interaction of diode laser wavelengths with dental implant surfaces of various types, and further work on this topic is necessary [122].

A significant deficiency in the literature is that, whilst clinical studies have been undertaken, the results for those are specific for the laser systems and implant materials used, for all the reasons mentioned above. It is not possible to generalise all lasers of the same wavelength, nor to all types of implant materials or implant surfaces. On the basis of laser to target interactions, results may vary even within the same laser system when there are changes to the delivery system, such as changing to a disposable fibre of a different diameter. This emphasises why it is important that further studies be done to establish what parameters can be used safely with implant components of different types.

**Conclusion**

The current literature supports the use of lasers in the treatment of peri-implant pathology, as an adjunct to traditional treatment methods, or in some cases as a replacement for them. The major considerations are around whether the use of lasers for surface debridement enhances the end result from a biological perspective (e.g. removal of biofilm, decontamination of the surface, and maintaining or enhancing its biocompatibility), whilst limiting the extent of change to the surface, at both the chemical and morphological levels. At the present time, there is no one ideal method for implant debridement and decontamination using conventional devices. Lasers are appealing; however, careful consideration must be given to the effect of laser irradiation on the implant surface. Whilst many studies have documented the extent of surface changes caused by laser irradiation of titanium implant surfaces, the results are bespoke to the particular laser system and target material combination that was used. Variables between lasers of the same wavelength or type (such as pulse duration, pulse shape, beam profile and delivery system) as well as between laser parameters (such as pulse energy, pulse frequency and irradiation characteristics) make comparisons between outcomes difficult. Adding to these factors is the influence of the terminal component of the delivery system on the spatial distribution of laser energy. Using tips that have been specially modified to improve the lateral emission of laser energy onto the side of the implant fixture may be appropriate given the morphology of the peri-implant pocket. The current study highlights the gaps in the current literature and the need for standardised studies to formulate recommended clinical protocols/settings for the use of lasers in different dental procedures.

**Funding**  Open Access funding enabled and organized by CAUL and its Member Institutions

**Declarations**

**Ethics approval**  This article does not contain any studies with human participants or animals performed by any of the authors.

**Conflict of interest**  The authors declare no competing interests.

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