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Precise dental ablation using ultra-short-pulsed 1552 nm laser

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ABSTRACT

A desktop diode pulsed laser having pulse width of 1.3 ps and wavelength of 1552 nm is utilized for precise targeted ablation of dentin, enamel, and composite material while minimizing thermal damage to the surrounding healthy tissue and nerve endings. A thermal imaging camera is used to measure the dental surface temperature rise during ablation. Following ablation, scanning electron microscopy (SEM) and optical microscopy are used to determine the quality of ablation and the volumetric ablation rate as a function of laser parameters. Surface temperature measurements are compared with the numerical modeling results obtained using the transient heat conduction equation. A good agreement between experimental and modeling results for the surface temperature is obtained which ensures accurate prediction of the temperature distribution throughout the tooth using numerical models. The SEM generates images of precise ablation of each dental material when the optimal laser parameters are used and the sample is scanned at a velocity to limit the number of overlapping pulses. During the ablation process there is minimal collateral damage to the surrounding healthy tissue and minimal heat spread throughout the tooth thus preserving the integrity of the pulp.

1. Introduction

Lasers have been of interest to the medical community for many years and have become a common instrument for many current medical procedures. For more than 40 years, lasers have been investigated for dental applications [1]. Currently, lasers in dentistry are primarily used for the ablation, or removal, of soft human tissue for periodontal disease and cosmetic surgery [2,3].

However, over this time, lasers have yet to become a common device used in dental offices for the ablation of hard tissue (enamel and dentin), caries, and filling material (composite and amalgam) for cavity preparation due to low ablation rates and thermal damage produced by many of the lasers studied over the last four decades. Caries removal and the preparation of cavities in dentistry are still primarily performed by the use of mechanical drills. These drills are often limited in precision resulting in a large amount of healthy enamel and dentin to be lost while removing the decayed tooth. This current technique is invasive and causes patient discomfort. The discomfort associated with mechanical drilling can be a psychological barrier for patients to overcome in seeking proper dental care. Due to the vibrations of the drills, it is necessary to use local anesthetic for the majority of dental procedures. A continuous water spray is used in conjunction with the drills to balance the temperature rise produced by friction between the drill and the tooth. Most tooth decay occurs in the enamel (outer surface) and the dentin, which is the region between the enamel and inner region of the tooth containing the nerve endings (pulp) [4].

In order to utilize lasers for caries therapy (removal of tooth decay), it is crucial to minimize the amount of heat diffusion to the surrounding tooth due to thermal energy produced by the laser irradiation. The human tooth is extremely sensitive to temperature variation such that a patient can sense a change in local temperature at the pulp of ±4 °C [5]. To protect the nerves contained in the pulp of the tooth from irreversible damage, such as permanent nerve loss, the temperature rise cannot exceed 10 °C [6–8]. Studies performed by Zach and Cohen [9] have shown that temperature rise of 20 °C in the pulp of the tooth almost always caused irreversible nerve damage. For temperature rises in the pulp of 10 °C there is a 15% chance that irreversible nerve damage will occur [9]. The pulp is extremely sensitive to temperature variations because of the nerve endings and blood flow contained within the pulp chamber. Alternatively, dental enamel and dentin can withstand higher temperature levels as long as thermal or mechanical damage (melting or cracking) is not induced.

Prior research on the ablation of human hard tissue has been primarily performed using a variety of long-pulsed lasers over a range of wavelengths. Recent investigation by some researchers has demonstrated precise selective ablation of composite

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Nomenclature

c	speed of light	r, z	spatial coordinates
C	specific heat of enamel	t	time
k_a	absorption coefficient of enamel	t_p	pulse width of laser beam
k_e	extinction coefficient of enamel	T	temperature
k_s	scattering coefficient of enamel	κ	thermal conductivity of enamel
q	heat flux	ρ	density of tissue

restorations and fissure sealants [10] using a frequency-tripled Nd:YAG laser. Though long-pulsed lasers have been used to ablate dental hard tissue efficiently, they produce a large amount of thermal diffusion throughout the tooth causing thermal/mechanical damage to the tooth structure as well as physical damage to the nerves [8]. It has been determined that direct ablation using long-pulsed lasers with pulse durations larger than 10 ps will increase the heat-affected zone, resulting in a thermal ablation in dielectrics such as teeth [11–15]. Thermal ablation occurs when a material absorbs enough radiant energy from the laser pulses to heat it to the boiling point temperature – at this temperature the material becomes vaporized [16]. According to Niemz [7], long-pulsed lasers and continuous-wave lasers generate a substantial amount of heat in hard tissue such as human teeth because the pulse duration is too long, allowing heat to diffuse significantly throughout the tooth and ultimately rising the pulp temperature far above the 10 °C temperature rise threshold for irreversible nerve damage [7].

On the contrary, ultra-short-pulsed (USP) lasers, with pulse durations less than 10 ps, are known for their ability to precisely machine materials, through the process of non-thermal ablation in which material is removed with very little heat deposited to the surrounding material. The precision that can be achieved with ultra-short-pulse lasers make them attractive for the ablation of hard biological tissue. Some applications for hard tissue ablation include mid-ear bone ablation [17], implantation of prosthetic joints [18], and dental ablation such as the removal of hard tissue and caries for cavity preparation [7,19–21].

It has been demonstrated that ultra-short-pulse lasers can precisely machine human hard tissue (bone and teeth) and dental filling material through the process of plasma-mediated ablation. Plasma-mediated ablation of dental hard tissue was first introduced in 1993 by Niemz. It has been revealed that focused pulsed light in the picosecond range would produce quality ablation with very little heat conducted to the surroundings [7,22]. The process has minimal heat diffusion to the surroundings due to the fact that material is removed through a complex process involving ionization from the area of focused pulse light faster than heat would be transferred through conduction to the bulk material [11].

Plasma-mediated ablation occurs when the ablation threshold of a material is achieved. To reach this threshold, the light intensity must be greater than 10^{11} W/cm^2 thus causing an optical breakdown to occur since the electric field amplitude would be greater than 10^7 V/cm. Focusing the laser beam to a spot size in the dimension of tens of microns is necessary to attain such intensities. At the focus, matter in its normal state is transformed into plasma that will continue to absorb energy from the laser pulses causing immediate vaporization [23]. The material in the focus spot will be ablated due to ionization, resulting in only minor thermal or mechanical damage [7]. When using ultra-short-laser pulses, lower pulse energy is needed to achieve ablation thresholds because intensity is inversely proportional to pulse duration [24]. Attainment of a desired ablation rate with low-pulse energies will limit the amount of collateral thermal damage to the surrounding tissue [24].

Prior research performed by researchers such as Neimz [7], Lubatschowski et al. [20], and Strabl et al. [21] have demonstrated that ultra-short-pulse lasers can precisely ablate teeth and other dielectric materials, but these researches have not reported any thermal analysis to validate the advantages of using such ultra-short-pulse lasers. Rather, the thermal effects are assumed to be minimal due to theoretical hypothesis that the ultra-short-pulse laser interaction with teeth is free of linear absorption and the material will be removed faster than heat can be spread throughout the tooth [11]. The temperature distribution throughout the tooth from the area at which ablation occurs must be systemically characterized rather than assumed to be minimal. If the sensitivity of tooth structure and vulnerability of nerves to temperature gradients is considered, such an investigation represents crucial information missing from previous work in this field.

It has been demonstrated by Urig et al. [25], that a thermal imaging camera can be used to measure the surface temperature rise on dental materials during hard tissue ablation. These measurements can then be compared to calculated values of the surface temperature from the transient heat conduction equation. A good agreement between the experimental and modeling surface temperature will validate the model and the internal temperature such as in the pulp can be predicted [25,26].

Numerical analysis of heat diffusion in the tooth caused due to laser irradiation during dental ablation is extremely important to validate the safety of dental procedures for future clinical investigations. Knowledge of the temperature rise at the pulp chamber is crucial in order to guarantee that the nerves will not become permanently damaged from the heat. Experimental measurement of the temperature rise within the pulp of the tooth is very difficult to achieve. The structure of the human tooth is hard and brittle, which makes if very difficult to drill small holes to insert thermocouples, without causing a large amount of tooth loss. Even if a few holes are created to insert thermocouples a small number of points will be measured and the thermocouples can give false temperature measurements due to the absorption of light by the thermocouples [25]. The best way to determine the temperature rise at the pulp of the tooth for a variety of laser parameters and many points throughout the tooth is to use the heat conduction model to predict the temperature rise in the pulp chamber.

Scanning electron microscopy (SEM) is a common technique used to characterize the dental ablation by first determining if material has been removed, and secondly to determine the quality of ablation. The quality of the ablation is indicated by the appearance of the remaining material. If cracking and melting (thermal damage) occurs, the ablation quality is poor and it shows that heat conduction to the surrounding material has played a large role in initiating the ablation process [27,28]. If the surrounding material looks unaffected with a precise removal of material the ablation quality is good [8,29].

To determine the volumetric ablation rate of the dental materials, an optical microscope is used to measure the depth of ablation. Galbraith [30] demonstrated that the depth can be measured by using the optical microscope to focus on the surface of the test sample and later focus at the depth of ablation. Subtraction of the distance the objective moved to focus at the two locations will give the depth of ablation [30].

This paper demonstrates the use of a fiber-based desktop laser (Raydiance Inc., Orlando, FL) emitting light in the near-infrared region at a wavelength of 1552 nm with pulse duration of approximately 1.3 ps for ablation of various dental materials (extracted human teeth with and without composite filling). This wavelength of light is common for the communications industry, although unique to biomedical research. Ultra-short-pulse laser ablation of biological materials at this wavelength has never been reported before in the literature. Therefore, the aim of this paper is to demonstrate precise ablation of enamel, dentin and composite filling material in extracted human teeth for cavity preparation using an ultra-short-pulsed 1552 nm laser beam while limiting the maximum temperature rise within the pulp of the tooth to less than 10 °C. The measured surface temperatures of the extracted teeth during laser irradiation for various laser parameters are compared with the numerical results obtained by solving the transient heat conduction equation. SEM and optical microscopy are used to determine the quality and precision of ablation and the volumetric ablation rate of each material as a function of laser parameters.

2. Experimental methods and materials

A schematic of the experimental setup is shown in Fig. 1. The laser used in this study is a desktop ultra-short-pulse fiber laser (Raydiance Inc.) operating in the near-IR region having a wavelength of 1552 nm and pulse duration of approximately 1.3 ps. The laser beam is focused on extracted human teeth samples using a 20× microscope objective with a working distance of approximately 21 mm to produce a focal spot size diameter of roughly 10 μm. The array of time-averaged power values that can be produced by the laser varies from 0.040 W to 2.50 W through the adjustment of both the energy per pulse which varies from 1 μJ/pulse to 5 μJ/pulse and the repetition frequency which varies between 25 kHz and 500 kHz.

Extracted human teeth samples were obtained from Bright Now Dental in Satellite Beach, FL. Teeth samples included 10 healthy teeth and 10 teeth containing fillings (composite material). The composite material (Prime and Bond NT, DENTSPLY Caulk, Milford, DE) is a light-cured dental adhesive system that combines composite materials to the dentin and enamel of the tooth for direct composite cavity restoration. The composition of Prime and Bond NT includes dimethacrylate and trimethacrylate resins, PENTA (dipentaerythritol penta acrylate monophosphate), nanofillers- amorphous silicon dioxide, photoinitiators, stabilizers, cetylamine hydrofluoride, and acetone [31].

The extracted teeth were machined first with a dental drill and then with a surface grinder so that the crown of the tooth was removed resulting in a flat surface exposing dentin, enamel, and composite filling on which the laser beam is easily focused. The laser beam is focused directly on the surface of the machined tooth where the ablation is desired. The sample was positioned on a three-axis stage driven by high-speed Newport SMC 100 motion controllers interfaced with National Instruments Lab view 8.0 to run the controllers. The maximum velocity of these controllers is 25 mm/s and can be varied in steps of 50 μm/s.

A detailed parametric study is performed to determine the ablation quality, surface radial temperature rise, and volumetric ablation rate as a function of laser parameters such as time-average power, repetition rate, and energy per pulse. During ablation, an infrared camera (IR Flexcam Pro, Infrared Solutions, Victoria, AU) is used to monitor the tooth's surface temperature. Thermal camera data is processed to generate radial surface temperature rise plots which will be compared to the calculated values for the surface temperature rise using the transient heat conduction equation. Following the dental ablation experiments, a scanning electron microscope (JEOL JSM 6380LV, Tokyo, Japan) is used to determine the quality, precision and width of the ablation trench created in each material (enamel, dentin and composite filling). An optical microscope on a Scanning Probe Microscope (PSIA, XE-100, Santa Clara, CA) is used to measure the ablation depth in the dental materials with an accuracy of 0.1 μm.

3. Mathematical modeling

To study the temperature distribution and analyze the heat diffusion zone during laser irradiation of dental materials, the energy equation is coupled with the Fourier constitutive equation, and the resulting parabolic equation is solved numerically. In this paper, the numerical models are used to predict the temperature rise at the surface (enamel) and at a depth of 2.5 mm which corresponds to the pulp location (see Fig. 2) during ablation. A pulsed laser beam of pulse width (t_p) is incident normally and focused on the sample. The laser beam has a Gaussian distribution in the radial direction and hence an axisymmetric cylindrical coordinate system is used to describe the geometry.

The Fourier heat conduction equation is given by:

$$q(r,z,t) = -\kappa \nabla T(r,z,t), \tag{1}$$

where T is the temperature, r and z are the spatial coordinates, t the time, q is the heat flux and κ is the thermal conductivity of enamel, and ∇ is the gradient.

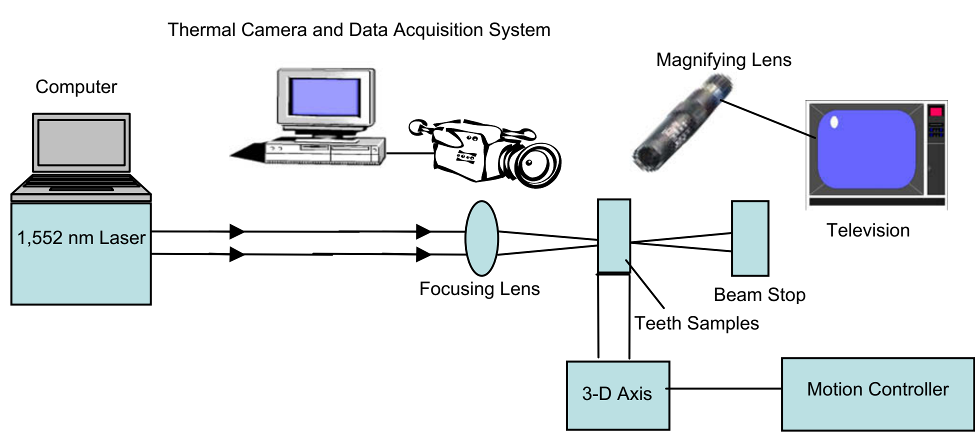

Fig. 1. Schematic of experimental setup.

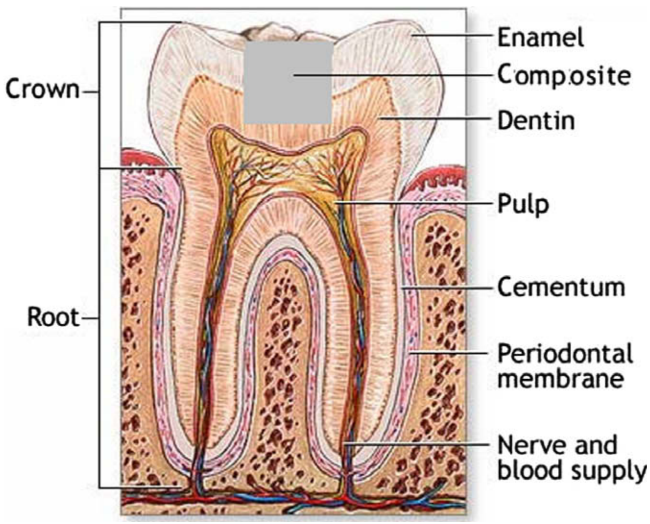

Fig. 2. Schematic of human tooth containing composite filling.

For laser interaction with dental materials incorporating scattering and absorption of laser energy in the tissue medium, the energy equation is given by [32]:

$$-\nabla q(r,z,t) + k_a L_0 \exp\{-4\ln 2 \times [(t - z/c)/t_p - 1.5]^2\} \exp\left(\frac{-2r^2}{\sigma^2}\right) \times \exp(-zk_e) = \rho C \frac{\partial T}{\partial t}, \tag{2}$$

where ρ is the density of enamel, C is the specific heat of enamel, c is the velocity of light, L_o is the maximum intensity of the laser beam at the sample surface, k_a is the absorption coefficient of the enamel, k_e is the extinction (scattering and absorption combined) coefficient of the enamel, t_p is laser pulse width, and σ is the spatial variance of the Gaussian laser beam.

The boundary conditions that are used are the following: (i) all the boundaries except the laser incident surface are insulated, (ii) at the laser-incident surface convective heat exchange (convective heat transfer coefficient = 10 W/m^2 K) with surrounding ambient air (25 °C) is considered, (iii) the temperature profile is symmetric about the z-axis, and (iv) initially (t = 0) the temperature is equal to the ambient temperature.

Eqs. (1) and (2) are discretized and combined to obtain a finite-difference equation using alternating direction implicit (ADI) scheme with Backward in Time and Central in Space (BTCS) formulation. Details of this discretization scheme and Van Neumann stability analysis can be found in literature [33]. The solution is obtained by using values of Δr and $\Delta z = 1.5625 \times 10^{-4}$ m and 2.5×10^{-5} m, respectively and $\Delta t = 0.05 \times 10^{-12}$ s. The resultant finite difference equation is solved by tri-diagonal matrix algorithm (TDMA).

The optical and thermophysical properties of enamel and dentin used in the above equations are given in Table 1.

4. Results and discussions

Extracted healthy human teeth and teeth containing composite filling material were irradiated with the ultra-short-pulse 1552 nm fiber laser to quantify and understand laser interaction with dental materials. A parametric study was performed to determine the quality of ablation, the surface radial temperature rise, and the volumetric ablation rate as a function of laser parameters such as repetition rate and time-averaged power.

Table 1
Optical and thermal physical properties of human enamel and dentin at 1552 nm

Dental material	Extinction coefficient (m^{-1})	Absorption coefficient (m^{-1})	Density (kg/m^3)	Thermal conductivity (W/m °C)	Specific heat (J/kg °C)
Enamel	380	128	2970	0.931	753.77
Dentin	900	400	2140	0.569	1171.52

Computer-controlled motion controllers were used to translate the teeth samples perpendicular to the laser beam focus at a constant velocity of 1 mm/s. Scanning experiments were performed by varying the repetition frequency of the laser from 50 kHz to 500 kHz while keeping the pulse energy constant at 5 μJ, resulting in time-averaged laser powers of 0.25–2.5 W. If the power loss due to the converging optics is accounted for, the time-averaged power incident on the dental samples is reduced to nearly half the output laser power and is varied from 0.15 W to 1.31 W. Fig. 3 is a schematic drawing demonstrating an extracted human tooth as it is translated perpendicular to the laser beam focus which is incident on the sample surface.

During irradiation of the teeth samples, the surface-average temperature was monitored using an infrared camera. After the completion of experiments, scanning electron microscope (SEM) images were taken to obtain a detailed image of the ablated region to determine the quality of ablation and to detect any resulting thermal or mechanical damage to the surrounding material. Analysis software (Scandium, Olympus Soft Imaging Solutions, Münster, Germany) was used in conjunction with the SEM images to measure the width of the ablation trenches. In addition, a high-power optical microscope attached to the Scanning Probe Microscope was used to measure the depth of ablation, and used to calculate the volumetric ablation rate of the dentin, enamel, and composite material, as a function of the laser parameters.

Fig. 4 shows the radial temperature distribution during ablation of an extracted normal tooth sample for various laser parameters when the sample is scanned at a constant velocity of 1 mm/s. The surface-average temperature rise increases with the increase of time-averaged power and repetition rate. For time-averaged power equal to 1.31 W, 0.73 W, 0.29 W, 0.18 W, and 0.15 W the maximum surface temperature rise is approximately 43 °C, 30 °C, 17 °C, 11 °C, 8 °C, respectively. The error bars plotted in Fig. 4 represent uncertainty in experimental measurements. If a 99% confidence interval is used, the precision index for a total of three runs can be calculated. The standard deviation between the three runs at each individual nodal point is evaluated. Thus, the total uncertainty values at each nodal point is the product of the precision index times the standard deviation. It is observed that a maximum total uncertainty 2.86 °C is obtained for the case of time-averaged power of 1.31 W and repetition rate of 500 kHz. Error bars for other cases are calculated and is in the same range but is not shown in the figure.

The nature of human teeth make if very difficult to experimentally monitor the local temperature rise at the pulp chamber during laser irradiation Mathematical modeling of the transient heat

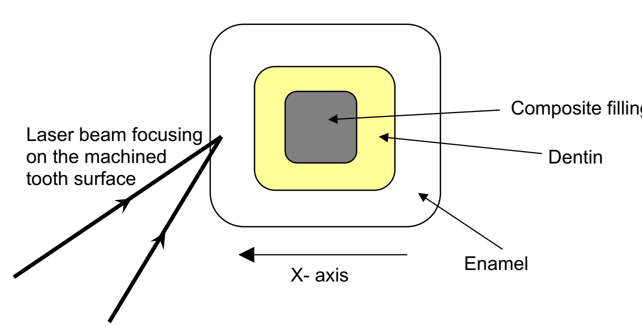

Fig. 3. Schematic of extracted tooth as it is scanned through the laser beam focus.

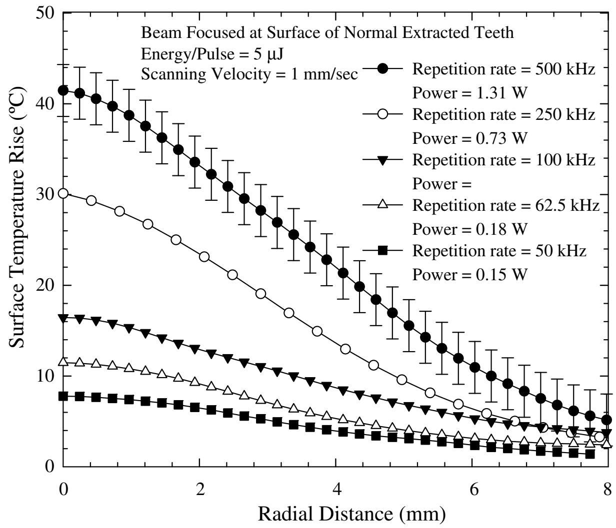

Fig. 4. Surface radial temperature distribution during ablation of normal teeth sample for different laser parameters.

conduction equation is used to predict the local internal temperature rise at the pulp chamber. The predicted values obtained using numerical models for the surface temperature rises match reasonably well with the experimentally measured values as observed in Fig. 5. The predicted values for the temperature rise at the interface between the dentin and the pulp (see Figs. 2 and 3) within the pulp chamber obtained using the numerical model is found to be very low, which is desired. For all the laser parameters considered, the predicted temperature rise at the pulp is less than 10 °C, which is the threshold for irreversible nerve damage.

The quality of ablation of the dentin and enamel at different laser parameters are next examined using SEM images. Figs. 6a and 6b show SEM images for the case of repetition rate = 50 kHz and time-averaged power = 0.15 W for the case of dentin and enamel, respectively. The dentin and enamel are both ablated precisely at these laser parameters and there are no visual signs of thermal or mechanical damage to the surrounding hard tissue. Figs. 7a and 7b display SEM images of the ablated dentin and enamel, respectively for repetition rate = 500 kHz and time-averaged power = 1.31 W. It is observed that a significant amount of thermal damage occurs in the dentin while enamel is precisely ablated with no signs of thermal damage at these laser parameters.

Figs. 8 and 9 show the volumetric ablation rate of extracted human dentin and enamel, respectively as a function of time-averaged power. The ablation rate for dentin increases in a linear fashion with the increase of the time-averaged power. On the other hand the ablated volume of the enamel remains the same as the time-averaged power is increased. This demonstrates a true non-thermal ablation in enamel in which the material is removed independently of the linear absorption of the 1552 nm light. Dentin and composite material have a higher concentration of water than the enamel. In the regions where more water is present, the light (1552 nm) will be highly absorbed with respect to the water content. Tables 2 and 3 summarize the ablation width and depth measurements and also the calculated ablation volume of dentin and enamel, respectively for different laser parameters. Fig. 10 shows the experimental results during ablation of teeth samples containing composite material for various laser parameters when the sample is scanned at a constant velocity = 1 mm/s. The surface temperature rise increases with the increase time-averaged power and repetition rate as in the case of normal extracted teeth. The

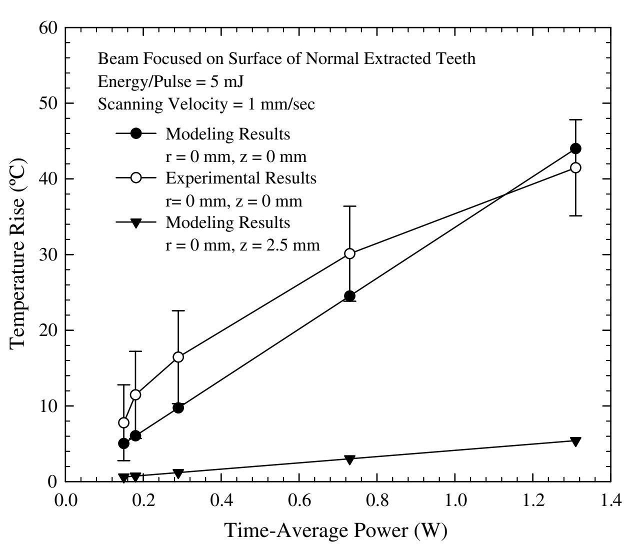

Fig. 5. Modeling validation of experimental results for the surface temperature rise and the predicted pulp temperature rise as a function of time-averaged power.

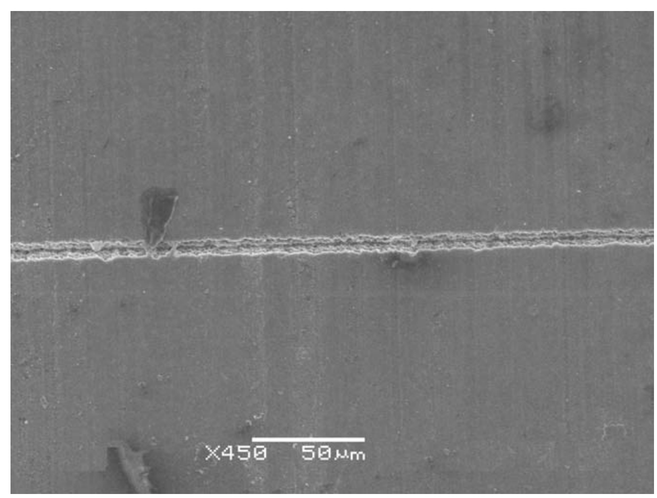

Fig. 6a. SEM image of ablation of dentin for the case of time-averaged power = 0.15 W and repetition frequency = 50 kHz; magnification = 100×.

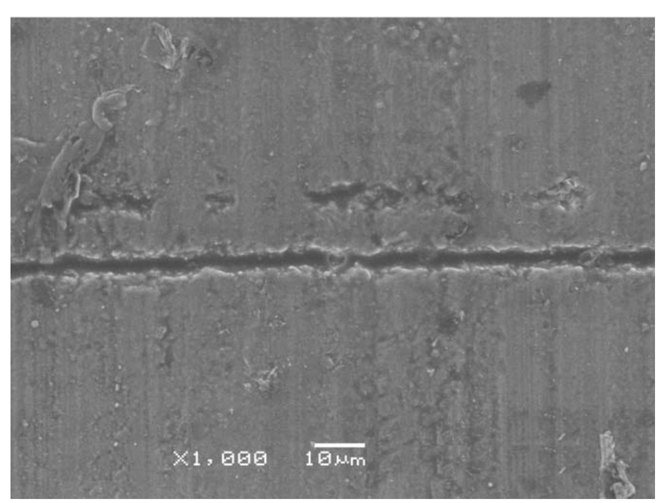

Fig. 6b. SEM image of ablation of enamel for the case of time-averaged power = 0.15 W and repetition frequency = 50 kHz; magnification = 1000×.

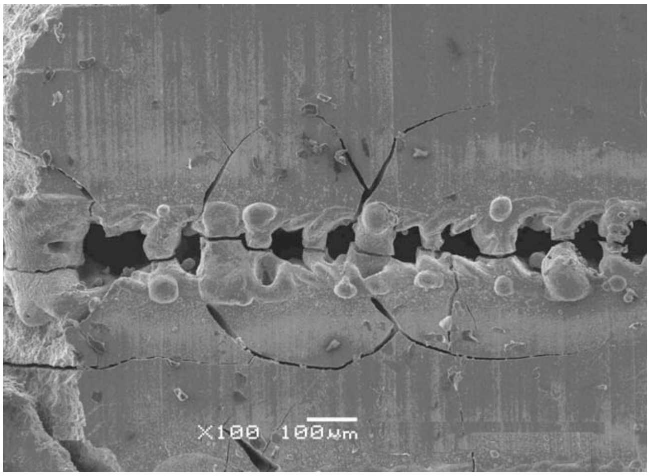

Fig. 7a. SEM image of ablation of dentin for the case of time-averaged power = 1.31 W and repetition frequency = 500 kHz; magnification = 100×.

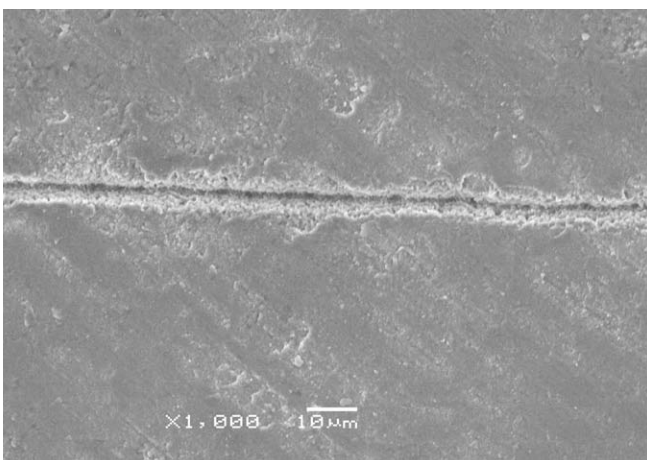

Fig. 7b. SEM image of ablation of enamel for the case of time-averaged power = 1.31 W and repetition frequency = 500 kHz; magnification 1000×.

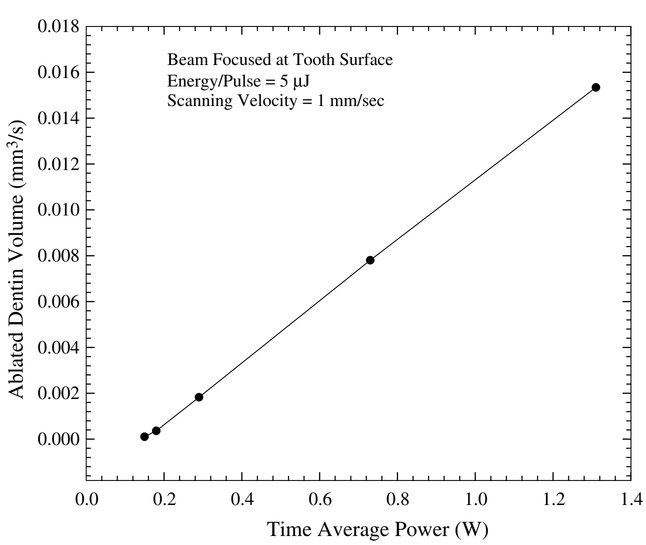

Fig. 8. Ablated volume of dentin as a function of time-averaged power.

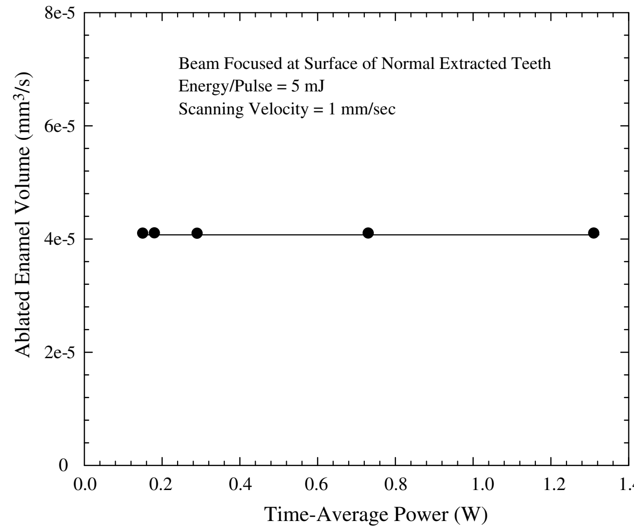

Fig. 9. Ablated volume of enamel as a function of time-averaged power.

Table 2
Ablated volume of dentin for various laser parameters

Repetition rate (kHz)	Time-averaged power (W)	Ablation depth (optical microscopy) (mm)	Ablation width (Scandium software from SEM images) (mm)	Ablation volume of dentin (mm^3)
50	0.15	0.016	0.0065	0.0001
62.5	0.18	0.028	0.0098	0.0004
100	0.29	0.033	0.0173	0.0018
250	0.73	0.065	0.1230	0.0078
500	1.31	0.108	0.1420	0.0153

Table 3
Ablated volume of enamel for various laser parameters

Repetition rate (kHz)	Time-averaged power (W)	Ablation depth (optical microscopy) (mm)	Ablation width (Scandium software from SEM images) (mm)	Ablation volume of enamel (mm^3)
50	0.15	0.0086	0.0039	0.00003423
62.5	0.18	0.0053	0.0065	0.00003604
100	0.29	0.0061	0.0059	0.00003516
250	0.73	0.0050	0.0068	0.00003405
500	1.31	0.0057	0.0061	0.00003477

maximum temperature rise for time-averaged powers of 1.31 W, 0.73 W, 0.29 W, 0.18 W and 0.15 W are approximately 38 °C, 26 °C, 12 °C, 10 °C, and 8 °C, respectively. The quality of the composite filling material ablation for two laser parameters at a frequency of 50 kHz, time-averaged power of 0.15 W and at a frequency of 500 kHz and time-averaged power of 1.31 W as obtained using SEM is shown in Figs. 11 and 12, respectively. These images show a very precise and uniform removal of composite material. There are no signs of visual thermal or mechanical damage to the surrounding dental composite material. The SEM images demonstrate that this laser is capable of precisely removing composite material with minimal thermal damage to the surrounding material.

Fig. 13 displays the volumetric ablation rate of the teeth samples containing composite filling material as a function of time-averaged power. It is determined that the ablation rate increases linearly with the increase of time-averaged power as for the case of dentin. At the maximum time-averaged power of 1.31 W, the

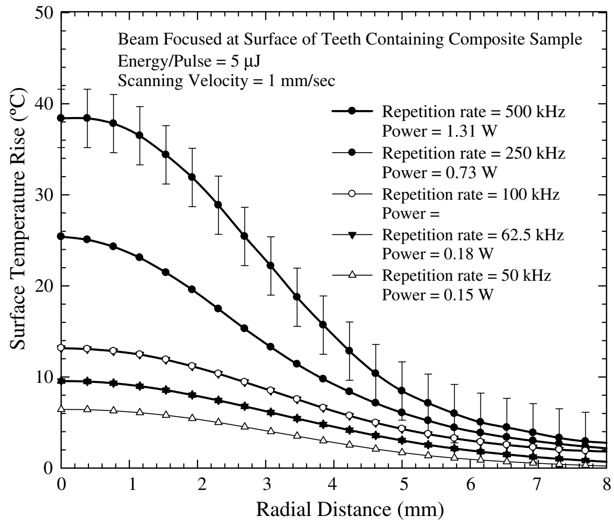

Fig. 10. Surface radial temperature distribution during ablation of teeth sample containing composite filling material for different laser parameters.

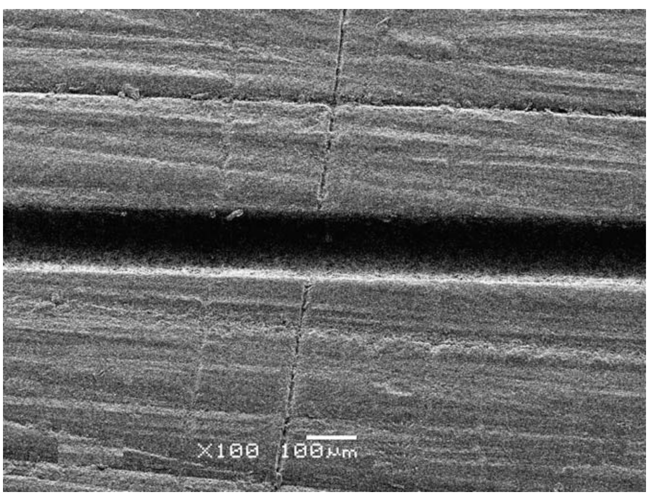

Fig. 11. SEM image of ablation of teeth sample containing composite filling material for the case of time-averaged power = 0.15 W and repetition frequency = 50 kHz; magnification = 100×.

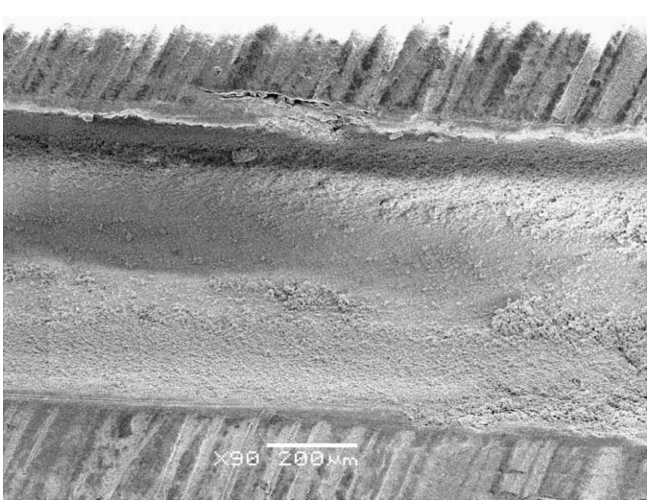

Fig. 12. SEM image of ablation of teeth sample containing composite filling material for the case of time-averaged power = 1.31 W and repetition frequency = 500 kHz; magnification = 90×.

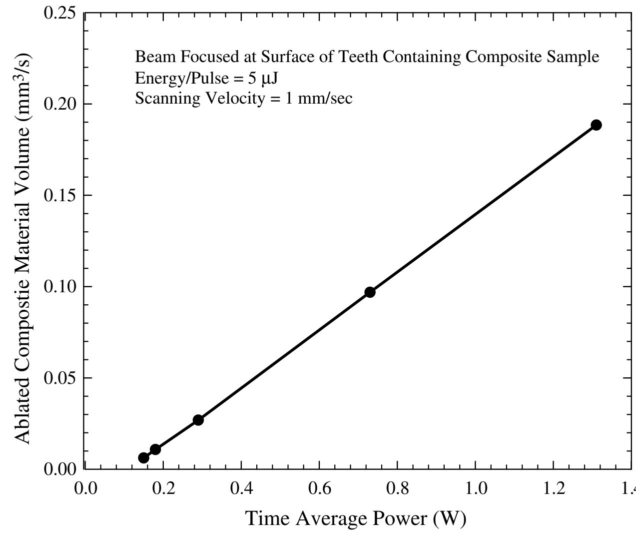

Fig. 13. Ablated volume of composite filling material as a function of time-averaged power.

Table 4
Ablated volume of composite filling material for various laser parameters

Repetition rate (kHz)	Time-averaged power (W)	Ablation depth (optical microscope) (mm)	Ablation width (Scandium software from SEM image) (mm)	Ablation volume of composite material (mm^3)
50	0.15	0.0548	0.1125	0.006159
62.5	0.18	0.0720	0.1510	0.010801
100	0.29	0.1361	0.1979	0.026903
250	0.73	0.2516	0.3569	0.096905
500	1.31	0.3208	0.5913	0.188398

ablation rate is approximately 0.20 mm^3/s. At the lowest time-averaged power of 0.15 W, the ablation rate is approximately 0.02 mm^3/s. Table 4 summarizes the values for the ablation depth, width and ablated volume as a function of repetition rate and time-averaged power.

5. Conclusion

This paper demonstrates the use of an ultra-short-pulse laser operating at 1552 nm for precise ablation of dental materials such as enamel, dentin, and composite materials. Composite filling material grossly absorbs the 1552 nm laser light causing the energy to be distributed as heat to the surrounding composite material. This process is traditionally how long-pulsed-lasers thermally ablate material. The composite material absorbs the radiant energy at this wavelength due to the water content in the composition and possibly other materials that strongly absorb at 1552 nm. It is observed that as the time-averaged power and repetition frequency increase, so does the average surface temperature and the amount of material removed. As the repetition rate decreases from 500 kHz to 50 kHz the time between consecutive pulses increases from 2 μs to 20 μs. This time increase allows the radiant energy to dissipate between laser pulses resulting in decreased surface temperature rise and collateral thermal damage at the lower repetition rates. To create precise non-thermal ablation of the composite material, the repetition rate needs to be reduced or the scanning velocity needs to be increased to minimize the spatial overlap of the pulses. For the experiments performed in this paper the energy per pulse remained constant at a value of 5 μJ.

Extracted human enamel is precisely ablated with consistent material removal and no collateral thermal damage as the time-averaged laser power and repetition rate is varied. This demonstrates that the enamel is ablated independent of the traditional absorption of radiant energy. The enamel experiments represent successful precise ablation with no thermal damage to the surrounding healthy hard tissue. On the other hand, extracted human dentin is a more porous material with more water content than the enamel region of the human tooth. The dentin interacts similarly to the composite material such that it grossly absorbs the radiant energy causing a large amount of thermal and mechanical damage induced on the surrounding tooth structure as the time-averaged power and repetition rate increases. It is observed that the removal rate of dentin increases as the time-averaged power and repetition rate increase. However, the thermal damage and heat-affected zone also increases significantly. The ablation rate for enamel remains the same ($\approx$0.000035 mm^3/s) for different time-averaged laser power and repetition rate.

The experimental results for the surface temperature rise during ablation are in good agreement with the values obtained using the transient heat conduction model. Therefore, the predicted temperature rise at the pulp chamber obtained using the mathematical model is accurate. The predicted values for the pulp temperature rise are very low and never rise above 10 °C during laser ablation of the extracted human teeth.

With the data presented in this paper, future research on a potential smart ablation device for cavity preparation is possible. Initial tests have been successful proof of concept work demonstrating that precise ablation with minimal thermal damage can be produced with this ultra-short-pulse laser.

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