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Authors
Neev, Joseph
Huynh, Daniel S
Carrasco, William A
et al.

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THERMAL AND NOISE LEVEL CHARACTERISTICS OF HARD DENTAL TISSUE ABLATION WITH 350 FS PULSE LASER

Joseph Neev, Daniel S Huynh, William A. Carrasco, Petra Wilder-Smith
Beckman Laser Institute /
University of California, Irvine, CA

Luiz B. Da Silva, Michael D. Feit, Michael D. Perry, Alexander M. Rubenchik, Brent C. Stuart
Lawrence Livermore National Laboratories,
Livermore, CA

Abstract

In spite of intensive research, lasers have not replaced conventional tools in many hard tissue applications. Low removal rates, loud operation noise, and mechanical and thermal damage are among the main obstacles to successful application of lasers. Ultrashort pulse lasers offer several advantages in their high per-pulse-efficiency, negligible thermal and mechanical damage and low noise operation. Practical applications of these devices, however, depends critically on sufficiently high volume removal which should match or even exceed the high speed drill. In our study, acoustical output of the USPL is compared to the low and high speed dental drill, Er:YAG, and Ho:YSGG lasers. Noise levels of the USPL are shown to be negligible in comparison with all other tested system. In addition, thermal characteristics of hard dental tissue ablation by ultrashort pulse laser of low and high pulse repetition rates are presented. Encouraging results showing temperatures increases smaller than 10^0C even at the highest pulse repetition rates (1 KHz) are presented. A simple model for heat diffusion is discussed.

Introduction

Since their introduction to commercial use in the 1960's, lasers have been used for a variety of applications in the biomedical field. However, using lasers to ablate calcified tissue can result in irreversible damage of fractures and fissures because hard tissue is prone to both shear and compressive stresses.

In spite of these dangers, using lasers to ablate hard tissue is still attractive because these instruments offer the potential for precise and selective operation with minimal thermal or mechanical damage to surrounding tissue. Lasers have, therefore, been applied to a variety of sensitive hard tissue applications such as mid-ear bone ablation, implantation of joint prostheses in orthopedic surgery, removal of malignant tissue and dental procedures involving hard tissue. In the field of dentistry, laser applications include treatment of oral malignancies and periodontal diseases, dental caries control and removal, and removal of hard tissue for cavity and restoration preparation (1-8). Visible and infrared continuous wave and long pulse lasers are also used to optically drill holes through the stapes foot plate (stapedotomy operation) in the surgical treatment of otosclerosis, a disease in which the stapes bone fuses to the surrounding hard tissue in the middle ear (8-16). Although using lasers to remove bone has been considered for many years, it is still not widely practiced, in spite of considerable research in this area (17-24). In addition, orthopedic applications such as arthroscopic surgery (partial menisectomy, synovectomy, chondroplasty), cartilage and tendon removal, bone incisions, microperforation, and resurfacing and texturing of cartilage, tendon, and bone have all emerged as areas of laser applications (25). Finally, current research in areas of laser hard tissue ablation includes artherosclerotic plaque removal (26-28), preparation of vascular surfaces for stent insertion and treatment of numerous conditions found in the nail unit (29).

In spite of their promise, lasers today are still limited in their ability to remove sound tooth or

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hard tissue structure. Specifically, lasers can remove only small quantities of hard tissue and still avoid thermal and mechanical damage to remaining layers. Even with the application of air and/or water coolants and the use of higher repetition rates, lasers cannot cut tissue as efficiently as, for example, the conventional rotary drill (30).

Capitalizing on the new technology of ultrashort pulse lasers, we have identified a laser parameter regime which could provide hard tissue interaction characteristics that are superior to conventional mechanical drill technology or other commercial laser systems. The major advantages of this tissue ablation method are: 1) efficient ablation; 2) minimization of collateral damage; 3) ablation thresholds and ablation rates which are relatively insensitive to tissue type; 4) high control over ablation depth is achievable because only a small amount of tissue is ablated per pulse; 5) low operation noise level; and finally, 6) precise spatial control due to the multiphoton nature of the interaction.

In this study, we compared the acoustical output of the USPL to that of the high speed dental drill, the Er:YAG and Ho:YSGG lasers. In addition, thermal characteristics of hard dental tissue ablation by ultrashort pulse laser of low and high pulse repetition rates are presented.

Materials and Methods

For plasma mediated ablation studies, laser pulses generated by a 1053 nm Ti:Sapphire Chirped Pulse Amplifier (CPA) system were used (31). Seed pulses of 100 fs from a Kerr-lens mode locked, Ti:Sapphire oscillator were stretched to 1 ns in a four-pass, single-grating pulse stretcher. Amplification by nearly 10^9 to the 6 mJ range was achieved in the TEM_{00} stable cavity mode of a linear regenerative amplifier. Further amplification to the 60 mJ level was achieved in a Ti:Sapphire ring regenerative amplifier, which supported a larger (2.3 mm) beam diameter and reduced nonlinear effects. This system operated at 10 Hz; however, single pulses could be extracted for experiments. The experimental setup is shown in figure 1.

After amplification, we compressed the pulses in a four-pass, single-grating compressor of variable length. By varying the dispersive path length of the compressor, pulses of continuously adjustable duration from 0.3 ps to 1 ns were obtained. The root mean square energy stability was typically less than 3%. Due to saturated amplification in the regenerative amplifiers, the maximum energy never exceeded the average by more than 6%. Smooth, reproducible, high quality beam intensity profiles are important in order to ensure uniform interaction at the targeted surface.

All measurements were performed with laser spot size of 0.5 mm diameter (e^{-2} intensity). Laser pulses were focused onto the tissue sample by a 1 meter focal length lens, with a variable distance to the sample. The spot size was measured with a Charged Coupled Device (CCD) camera. The laser spatial mode at the sample had a 98% or better fit to a Gaussian, so the effective diameter as measured on the camera system was combined with the measured energy to give the pulse energy fluence. Estimated absolute uncertainty in fluence was 15%, but relative values are within 5%.

Dentin and enamel slices, 0.5 to 1.0 mm thick, were cut from the middle section of freshly extracted third molars, in a plane perpendicular to the occlusal-cervical direction. Teeth were treated with 0.5M EDTA for 2 minutes to remove the smear layer, then stored in 10% thymol solution until treatment. Tissue was sliced parallel to the crown and washed with demineralized water. Slices included an external ring of enamel tissue on the outside and dentin.

In order to evaluate sound level (in dB) associated with the ablation effect of longer pulse lasers and mechanical drills, we recorded ablation noise during dentin and enamel tissue treatment and compare them to those of the USPL. Signals were digitized and transferred to our data acquisition system on a computer (Apple Power Mac 8100/80 AV) for further processing and analysis. The ablation noise was analyzed using SoundEdit 16 (Macromedia Inc. San Francisco, CA), and the results display relative intensity and frequency content as a function of time.

A thermal camera (Inframetrics Model 600 Imaging Radiometer, Bedford, MA) with a scan speed of 32 frames per second was used for temperature measurements. The center temperature adjustment ranges from 20 to 1,500 °C. The camera field of view could be adjusted from 1:1 to 8:1, and was maintained at 2:1 or 4:1. The camera detector was a liquid nitrogen cooled HgCdTe with spectral sensitivity from 8 to 12 μm. The thermal camera was placed so that the surface temperature (ST) could be viewed directly. The camera was operated in the point mode so that both the entire temperature field could be viewed with false color presentation and a single point (specified by the operator) numerical value could be read directly with the camera specified accuracy of $\pm$ 0.5 degrees. Temperatures were measured on the back surface of 1 mm dentin slices while the laser was ablating the front surface.

For evaluation of temperatures at high pulse repetition rates, a 1 KHz laser ablation temperatures were monitored. Here a 100 μm beam from a 1 KHz, 600 fs system (Spectra Physics / Clark MXR) was utilized. Energy per pulse was 600 μJ.

Results and Discussion

Temperature measurement:

Temperature measurement of 350 fs (3 and 16.2 J/cm^2) and nanosecond (16 and 32 J/cm^2) during the ablation of dentin are shown in figure 1. As shown, ablation with 350 fs and 3 J/cm^2 is confined to within 2-3 ^{0}C of room temperatures even at very long exposure time. As the fluence is increased to 16 J/cm^2 considerably more energy is coupled to the plasma and tissue, and the amount of material removed per pulse (ablation efficiency) drops dramatically (34). Temperature of the tissue, therefore, increases considerably. Also shown in figure 1 is the back surface temperature of 1 mm dentin slices during 1 ns interaction with sub-threshold fluence of 16 J/cm^2. In spite of this subthreshold fluence, temperature clearly climbs linearly with time. At the higher 1 ns fluence of 32 J/cm^2, temperature increases linearly and rapidly to over 70 ^{0}C and charring occurs.

The best fit curve for the 350 fs at 3 J/cm^2 ablation data is compared to our diffusion model (equation 2, see discussion below) with excellent agreement (figure 2). As an illustration of temperatures during ablation with high PRR system, Figure 3 shows the front surface temperatures of a 1 KHz system (600 fs, 800 μJ/pulses). Again, temperatures are seen to be confined to within ten degrees of room temperature (the fluctuation in temperature are due to ablation stalling as the crater get deep. This phenomena was observed with the 1 KHz laser because of the operation fluence proximity to the threshold level).

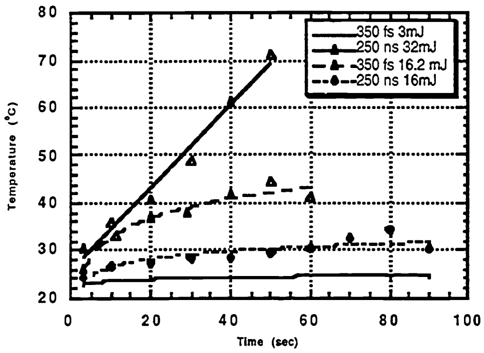

Fig. 1 Back surface temperatures of the 350 fs and 1 ns dentin ablation at various fluences

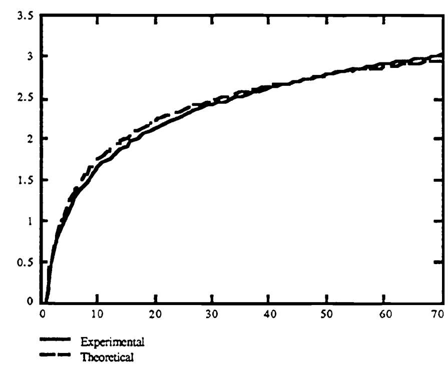

Fig. 2. Comparison of 350 fs 3 J/cm^2 measured temperatures to our thermal diffusion model

The accepted standard for tooth pulp safety was established by Zack and Cohen (35): pulp temperature increases must not exceed 5^0C for time duration longer than one minute. Our analysis demonstrate that use of the short pulse lasers results in only a moderate thermal loading in the ablation area, and therefore, ensure that this important safety limit is not approached.

To obtain initial insight into temperature distributions during USPL interaction with the tooth, we have conducted IR experiments on dentin slices (1 mm thick) where both front and rear surface temperatures were monitored. Back surface observations of a dentin slab with known thickness allow precise modeling of the time and space behavior of the temperature field. From our analysis (for a detailed discussion see 34), we arrive at the following expression (with a slab of thickness h, χ = the thermo-diffusivity coefficient, p = PRR, and a = the beam spot size).

$$T=\frac{2Q_0a^2}{\rho c\sqrt{\chi}}\int_0^t \frac{d\tau\, e^{-\frac{r^2}{a^2+4\chi\tau}}}{\sqrt{\tau}\left(a^2+4\chi\tau\right)}\sum_{n=0}^{\infty} e^{-\frac{(2n+1)^2h^2}{4\chi\tau}} \qquad (1)$$

If we normalize τ to $\tau_0=\dfrac{h^2}{4\chi}$ for the temperature at r=0 we have

$$T=\frac{Q_0a^2}{\rho c\chi h} f\left(\frac{t}{\tau_0},\frac{a}{h}\right) \qquad \text{where} \qquad (2)$$

$$f\left(\frac{t}{\tau_0},\frac{a}{h}\right)=\int_0^{\frac{t}{\tau_0}}\frac{dx}{\sqrt{x}\left(\frac{a^2}{h^2}+x\right)}\sum_{n=0}^{\infty} e^{-\frac{(2n+1)^2}{x}}$$

With a 0.5 mm beam (1/e^2), the value h=2a corresponds to our experimental data. The meaning of graphs with larger a values will be discussed later. One can see that, similar to experimental data, our calculations predict the slowing down of the initial fast temperature rise.

The values of thermoconductivity for dentin and enamel are not very different, 0.45 W/mK and 0.65 Wt/mK, respectively. The products of density and heat capacity are close: 2.3 J/cm^{3}K in dentin and 2.175/cm^{3}K for enamel. If we consider the thermal diffusivity median between dentin and enamel at 0.0025 cm^2/sec., the diffusion time, h^2/4χ, is about one second. Using these values, the graph of Figure 2 shows our theoretical calculation for the temperature evolution of up to 70 seconds, as well as the best fit curve for our experimental data.

From our thermal camera data (Figures 1 and 2), the value of absorption efficiency α can now be estimated. From equation (3) and with a pulse energy 3 mJ used in the experiment, the temperature increase was about 3 ^{0}C. Using the value f=2, we have a reasonable value for α=0.07. This value looks reasonable because most of the pulse energy is reflected and a big part of it already absorbed is ejected with vapors. This value is also consistent with the results of our hydrosimulation, and is in good agreement with the experimental data.

In our experiments, the temperature was measured on the rear side of the slab. In reality, the thermal transport inward makes the temperature lower than in the slab example. The data indicates that the temperature is lowered by a factor of $2f/\sqrt{\pi}$. In addition, from Figure 2 we see that f is approximately 2 after 60 seconds, and as was pointed out above, in a realistic (whole) tooth model, the temperature must be 2-3 times smaller than in our experiments.

For comparison, long pulse temperatures, Figure 2, were also measured with an IR camera 1 mm behind the ablated surface at 10 Hz and 34 mJ/pulse (sufficiently above the threshold to be identified as efficient AR energy with Dentin AR of about 4 μm/pulse). Within our model, the long pulses must have a similar temperature dependence different only with larger coupling coefficients. Nevertheless, we observe more steep temperature increases and postponement of saturation. This can be explained if we take into account that for long pulses there is sufficient time for the plasma spot size to expand and exceed the laser spot size. Effective spot size will then increase. A larger spot size makes the geometry closer to the one-dimensional case and results in postponement of saturation time for the temperature increase. This effect also increase the risk for collateral tissue damage due to lateral transport by the plasma (and its high thermal conduction).

One of our goals is to demonstrate USPL high tissue removal rates which are associated with high PRR. From our preliminary ablation rates studies (see below), we concluded that such removal rates correspond to a pulse repetition rate of five to seven hundred Hz. From the discussion above, we saw that the temperature in semi-infinite media must be 3 times smaller than in our slice experiments. From the above simplified analysis, we already estimate that PRR as high pulse repetition rates of several hundred pulses per second can be tolerated without any external cooling. A further increase in PRR which will also be safe, can be achieved by the use of yet shorter pulses (members of our team have already developed the technology for pulses as short as 30 fs, two orders of magnitude shorter than our current 350 fs). Also, as the above analysis show, a small reduction in laser spot size will dramatically reduced temperatures and can, thus, easily allow a significant increase in operating PRR. In view of the above discussion, we feel confident that safe removal of dentin and enamel with high repetition rate ultrashort lasers is a realistic possibility. However, in order to get an idea of actual thermal effect at the very high end of our PRR regime, we have conducted a preliminary IR camera study of temperatures generated by a 1 KHz, 600 fs pulse commercial system. Front surface temperatures are shown in Figure 3. As can be seen from the data, temperature increases even at this high PRR are limited to below 10 ^{0}C. Further modeling and experimental studies at the more effective ablation fluence range of 2-4 J/cm^2 are necessary, of course, and are awaiting the development of adequate, higher energy, high PRR systems.

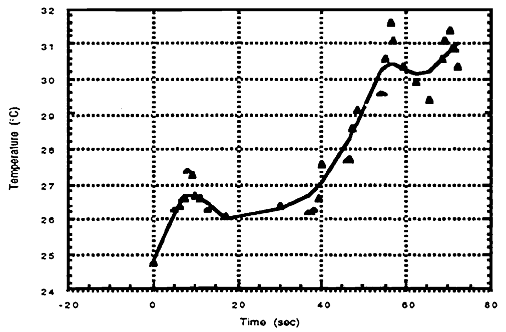

Figure 3. Front surface temperatures of a 1 KHz, 600 fs, 800 μJ system. Note that temperatures are confined to within 10 ^{0}C of room temperature

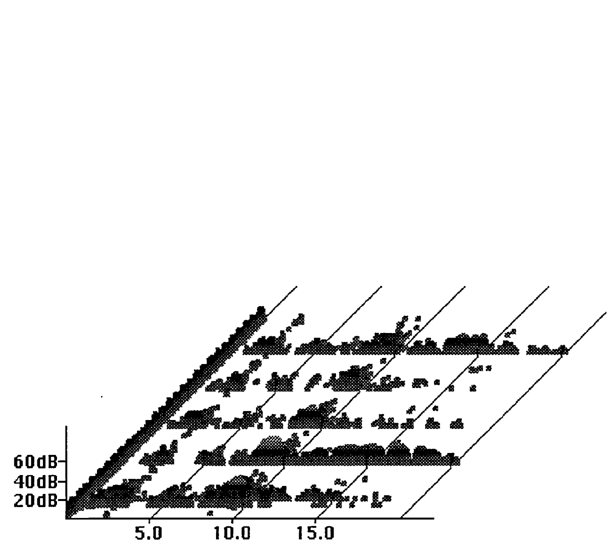

Figure 4a: Er:YAG laser noise level vs time and frequency.

Acoustical noise level comparison:

The potential for hearing damage (both to patient and dental care provider) is worth noting. As the laser pulse energy is increased and larger portion of it is coupled to the plasma, a significant and steady increase in audible spark is recorded. If acoustical output is compared with ablation rates as a function of pulse energy, it is clear that some of the additional energy beyond 3 mJ/pulse which is not being used effectively in material removal is converted to noise. On the other hand, the sound level at or below 3 mJ/pulse is very low.

The results (Figure 4), showing Er:YAG, high speed drill, Ho:YSGG, and USPL (dB noise level as a function of frequency and time) clearly demonstrate the negligible sound level of the USPL in comparison to either the mechanical drill or conventional long pulse lasers. Note that the very low frequency noise with the USPL is merely a background room noise and not the USPL noise. Similar observations for Er:YAG (not shown) demonstrated considerably higher noise in comparison to the USPL operation. However, if the pulse energy level is increased (e.g. to 16 J/cm^2 not shown) considerably larger portion of the light energy is coupled into the plasma which releases its access energy in a louder audible spark. Since increasing energy above 3-4 J/cm^2 does not result in increased ablation rates (see Neev et al).

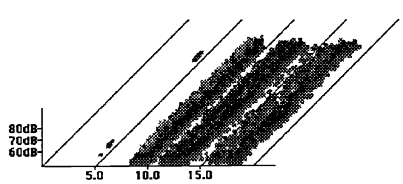

Figure 4b: High speed dental drill noise level vs time and frequency.

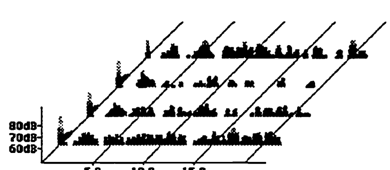

Figure 4c: Ho:YSGG noise (250 μs, 60 mJ/p) vs time and frequency

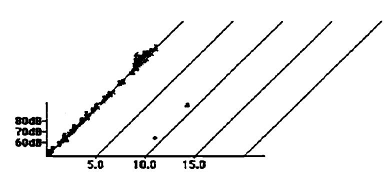

Figure 4d: Noise From an USPL (350 fs laser 3.0 mJ/p). Note: low frequency noise is background.

Conclusions

We have investigated the thermal and acoustical characteristics of the interaction of ultrashort pulse lasers with hard dental tissue. Minimal temperature increase and negligible noise output was identified at the most efficient interaction parameters and ten Hz PRR. Preliminary temperature measurements of a KHz system also illustrated relatively small temperature increases (< 10 ^{0}C) in agreement with our model predictions. Considered together with our ablation rates and ablated surface characteristics discussed in other papers in this conference, we feel that the potential for an efficient, selective, accurate, and damage-free operation of USPL systems was demonstrated.

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