Laser Profilometry for the Characterization of Craters Produced in Hard Dental Tissues by Er:YAG and Er,Cr:YSGG Lasers

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ABSTRACT:

A new, highly accurate and repeatable methodology based on the principle of optical triangulation to measure ablation rates in hard dental tissues is introduced. Using this methodology, a comparison is made between the two leading laser wavelengths for hard tissue procedures in dentistry, Er:YAG (Fidelis Plus III, Fotona) and Er,Cr:YSGG (Waterlase MD, Biolase). In-vitro measurements of the maximum available drilling speeds (ablated volume per second) revealed ablation rates of the Er:YAG laser system to be 3.7 times higher in dentine, and 5.0 times higher in enamel compared to those achieved with the Er,Cr:YSGG laser system.

Key words: Er:YAG; Er,Cr:YSGG; optical triangulation principle, VSP technology, ablation speed; hard tissue procedures.

INTRODUCTION

Erbium lasers have been long recognized as the optimal dental lasers for effective, precise and minimally-invasive ablation of hard dental tissues.[1] Of all infrared lasers, they exhibit the highest absorption in water and hydroxyapatite, and are thus ideally suited for cold 'optical drilling' in enamel, dentin and composite fillings.

Early erbium and CO_2 lasers failed to gain wide acceptance by the dental community because their optical drilling speeds were lower in comparison to the mechanical bur. This has changed in the past years, with much higher ablation speeds now possible, and Variable Square Pulse (VSP) technology supported dental laser systems even exceeding the drilling speeds of conventional mechanical burs.[2]

In this paper we report on a new, accurate and fast method to measure ablated volumes in hard dental tissues, based on the optical triangulation principle,[4] that was used in an in-vitro study of the differences in ablation speeds between the two main erbium laser wavelengths currently used in dentistry, namely Er:YAG (2940 nm) and Er,Cr:YSGG (2780 nm).[6,21]

Many methods have been used to measure the volume of ablation craters in hard dental tissues. One of the earliest measurements of crater volume and ablation rate (AR) was performed in 1992.[8] Er:YAG laser ablated craters in dentine and enamel were sectioned along the craters' depths with a diamond blade to enable crater depth measurements with the ocular micrometer of a microscope.

An approximate measurement of the ablation rate of dental material has also been conducted by measuring the time needed for the complete perforation of a dental sample of a known thickness.[9]

Another method for volume measurement has been based on the scanning electron microscopy (SEM) technique.[10] Stereoscopic SEM images were taken by tilting the samples eccentrically around the ablated site in the surface plane. A three-dimensional (3D) digital elevation model was obtained by using stereoscopic imaging and post-processing analysis software.

Measurements of the ablated volume of a tooth have also been made by focusing a microscope objective onto the top of a tooth surface and subsequently onto the bottom of the ablated crater.[11] Camera images were taken while changing the focus of the objective in a vertical direction. The images at various focal points were then grouped together to construct a 3D crater model.

A non-destructive method to measure ablated volume is based on optical coherence tomography (OCT) imagery.[12] Measurements of the AR were made without requiring tooth repositioning, but with long time periods between subsequent laser pulses.

X-ray micro-tomography (XMT) is another rapidly developing non-destructive microscopic technique for the visualization and characterization of the internal 3-dimensional architecture of X-ray opaque materials. Using modern imaging technology and a fine point X-ray source it is now possible to image objects in 3D with micrometer resolution.[13] But the evaluation process of a single 3D model of an ablated crater on the tooth surface still takes a relatively long time (several hours).

Hard dental tissue AR has also been measured using a 3D laser scanning technique. Impressions of teeth were taken before and after the laser ablation, and measured by a triangulation scanner.[14] Impressions were used in order to avoid problems with the diffusive scattering of visible light on hard dental tissues. A similar scanning system has also been used for the detection of tooth wear.[15] A laser beam line was produced by a laser, and projected onto the sample surface using an optical set-up with a cylindrical lens. This image was then projected onto a CCD sensor under the triangulation angle. The method is still relatively complex and time-consuming since it requires several impressions to be made.

In the present study, we have developed a direct triangulation method for measuring cavities in hard dental tissues,[4] which does not require cavity impressions to be made. The technique was applied to, in-vitro, evaluate the difference in performance between Er:YAG and Er,Cr:YSGG hard tissue dental lasers.[6,21] Since this method does not require the laser handpiece to be in a fixed position in respect to the tooth, it allows measurements to be made under realistic conditions, identical to a manually performed laser treatment by the dental practitioner.

EXPERIMENTAL SETUP

Lasers

The Er:YAG laser used was that of a Fotona Fidelis Plus III laser system fitted with either an R02 non-contact handpiece or an R14 fiber-tipped contact handpiece. The Er,Cr:YSGG laser system used was a Biolase Waterlase MD fitted with a fiber-tipped 'Gold' handpiece. The comparisons were made between the two lasers using a range of pulse width, energy and fluence configurations, ranging from single pulses to longer bursts of pulses. The built-in water spray with manufacturer-recommended settings was used for all of the experiments.

Materials

Extracted premolar and molar teeth were selected and immediately following extraction, stored in a 10% formalin solution. Teeth were thoroughly cleaned of all residual debris using brushes and curettes. Prior to the procedure, all teeth were sterilized in an autoclave at 121°C and 2,1 atm for 30 minutes and stored in a physiological saline solution. The teeth were randomly chosen for the ablation experiments. Each data point represents an average of the effects of 6x80 laser pulses from 6 different tooth samples. Since the precision of ablation efficiency measurements is very sensitive to any aging of the laser beam delivery optics, special care was taken to make measurements only with undamaged fiber tips, protective windows and laser beam delivery systems.

Profilometer

The method is based on 3D measurement of the tooth surface using the optical triangulation principle [16] which is explained using Fig. 1.

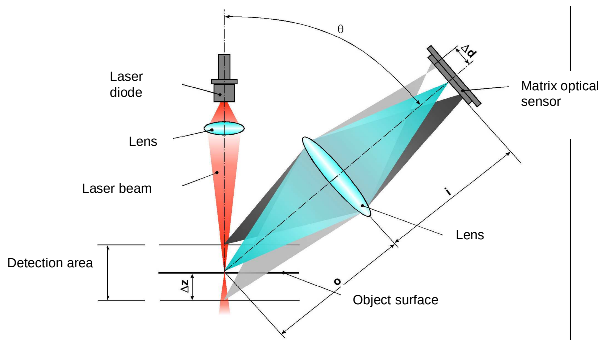


Fig.1: Schematic of the measurement system used for measurements based on the optical triangulation principle.

The surface is illuminated by a laser beam. Part of the diffusely reflected light from the surface is registered on the surface of an imaging sensor. If the surface changes its position relative to the laser by Δz, then the image of the irradiated area moves by Δd across the imaging sensor. By measuring Δd we can determine Δz once the parameters of the imaging system are known. A key parameter of the system is the (triangulation) angle between the optical axes of the laser and the imaging system.

In order to measure the shape of a surface, the surface is moved relative to the measurement system using a purpose-built positioning system, while Δz measurements are made during this process. To accurately characterize a laser produced crater in dental tissue it is necessary to make hundreds of thousands of Δz measurements. A new system we developed allows these measurements in just a few seconds and will be described in detail [3]. In this paper we have limited ourselves to presenting an overview of its main features for clarity purposes.

Several steps were undertaken in the development process with the aim to increase the measurement rate while not compromising accuracy. The selection of the shape of the measurement laser beam was one of these steps. The tooth surface under examination was illuminated using a laser beam with a highly elliptical cross-section (Fig. 2). A bright laser line is visible when the tooth surface is illuminated by such a beam. An image of the line is acquired by a camera. This arrangement enables the development of an accurate surface profile, consisting of about 500 Δz measurements, from a single acquired image. In order to measure the complete tooth surface, the tooth is translated in a direction perpendicular to the laser line using a precision translation platform driven by a stepping motor (Fig. 2).

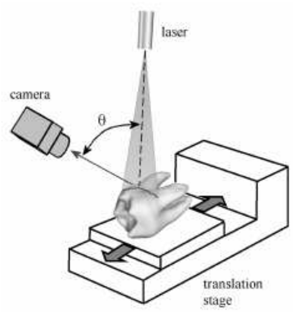


Fig.2: Operational design of the profilometer

Fig.3 shows a photograph of the measurement system, which includes the laser profilometer (left) and a personal computer (PC). The profilometer acquires profiles of a tooth surface under study and sends them to the PC, where they are developed into a 3D model of the surface. The model is analysed for characteristic geometrical parameters (e.g. ablated volume) which are then used to characterize the two laser processes under examination.

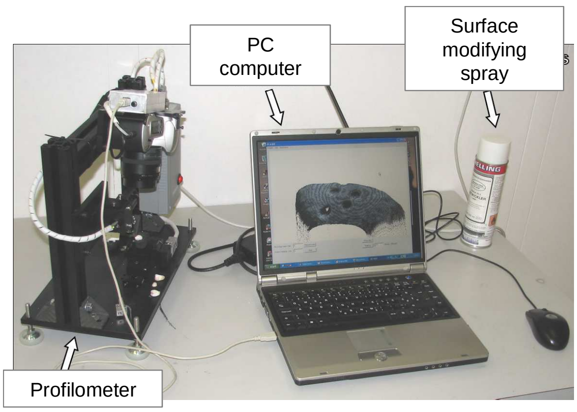


Fig.3: Photograph of the profilometer during the experiment

Another key element of the profilometer that boosts its measurement speed is a custom-built digital video camera with an integrated image processor. This allows the extraction of surface profiles at rates of up to 200 measurements per second.

We used a monochrome 656 x 494 pixel CMOS imaging sensor as the basis for the camera development. The sensor has an integrated on-chip ADC (Analog to Digital Converter) and a single 10-bit digital pixel output. It operates in "global shutter" mode in which all pixels stop collecting photo-generated charge carriers simultaneously. This eliminates geometric image distortion, known as the rolling shutter effect. 10] A dedicated image processor and interface to communicate with the host PC was developed using programmable logic (FPGA).

The imaging sensor was built into a modified Canon EOS 500N camera body with a Canon EF 50 2.5 Macro lens, which projects a 13.2 x 9.94 mm area on the tooth sample surface, onto the 6.61 x 4.97 mm active area of the sensor. The laser line is oriented along the shorter sensor side (perpendicular to the sensor row direction) as this enables fast laser line algorithm extraction. The 3D coordinates of the measured points were calculated with the triangulation model described in earlier studies [16]. The set-up was calibrated using a planar checker board as described in earlier studies [18, 19].

A linear translation platform driven by a stepper motor (travel range: 20 mm, resolution: 1.25 µm per full step) is used for sample translation. The translation platform is synchronized with the imaging sensor frame to attain accurate and repeatable measurements. Translation platform control logic is implemented inside the FPGA that controls the imaging sensor. The motor operates in a 1/8 micro-stepping mode which provides the smoothest motor operation.

The set-up measurement range is 20 mm, 10 mm and 5 mm along the x, y and z axes, respectively. The resolution along x, y and z axis is 156 nm, 20 µm and 5 µm, respectively. In a typical experiment we acquire 2000 profiles along a translation range of 10 mm which includes several craters and takes about 10 seconds. A typical measurement record takes approximately 16 Mbytes of memory. A typical result of the tooth measurement using the described set-up is shown in Fig. 4.

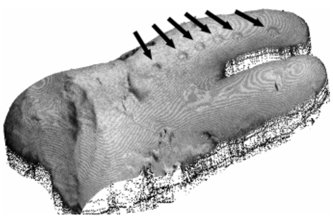


Fig. 4: Computer-rendered cloud of points of a tooth with ablated craters (marked by arrows).

Special software has been developed to control the measurement. The software allows key measurement parameters to be set up: projector laser power, translation platform position, travel range and the distance between consecutive profiles. The software also enables computer rendered images of 3D tooth surface models to be displayed. The 3D model images can be rotated, zoomed and panned interactively. Measured points within the area of interest can be arbitrarily selected and saved to file for further analysis.

Software ("Volume_analyser") was developed based on the GUIDE Matlab toolbox for further detailed analysis of the ablated crater. Fig. 5 shows screenshots of the graphical user interface during two data processing stages.

When measurement data is imported into the Volume_analyser, a rectangular region of interest (RROI) (rectangle in Fig. 5a) is selected and data outside the RROI is excluded from further analysis. Measurement outliers are also removed from the RROI at this stage. In the next processing stage the viccinity of the crater is examined and an area within the RROI is identified where the tooth surface has not been processed. These unprocessed surface points are used to reconstruct a reference surface $z_2(x,y)$, which approximates the tooth surface before laser processing. The reference surface is determined by surface fitting using bi-harmonic spline interpolation, which in 3D corresponds to multi-quadratic interpolation [20].

Denoting the measured crater surface by $z_1(x,y)$ we calculate the volume of the crater using the formula:

$$V \cong \sum_{i=1}^{m}\sum_{j=1}^{n}\left[z_2\left(x_i, y_j\right)-z_1\left(x_i, y_j\right)\right]\Delta x\Delta y ,$$

where $x_i = i.\Delta x$, $i = 1,2,\ldots,m$ and $y_j = j.\Delta y$, $j = 1,2,\ldots,$n are discrete Cartesian coordinates of the measured points. Δx and Δy are the distances between two neighbouring mesh points of the interpolated mesh in the x and y directions, respectively.

The Volume_analyser is also capable of calculating the crater depth and diameter. Fig. 5b shows a darker reference surface on the processed surface model of the tooth, and the calculated volume, depth and diameter.

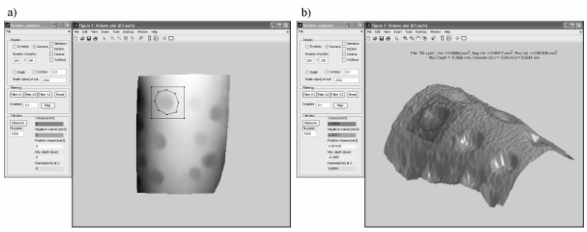


Fig. 5: Screenshots of the Volume analyser used for ablated volume determination. a) selection of region of interest; b) calculated reference surface superimposed on the measured data.

Tooth Surface Preparation

When a tooth is exposed to red laser light part of it penetrates deeper into the tooth volume. Consequently, a bright laser line on the tooth sample surface appears wider than it actually is, resulting in erroneous calculations of the profile position by the profilometer. This problem can be in principle suppressed systematically but sample material and structure that can distort the line width can vary greatly, which presents an added difficulty. Another problem is that the reflected light is brighter in the craters compared to the non-ablated tooth surface (see Fig. 6a). This is because the ablated craters are rougher, with more diffusely reflected light compared to the non-treated parts of the tooth. The digital camera detects these reflections as local bright and dark light sources with very high contrast. If the tooth is scanned without modifying its surface first, its 3D surface model will be unusable because it will present too much data noise (see Fig. 7a). We have solved the above problems by spraying a thin layer of a white powder on the tooth surface (see Fig. 6b). The powder is a few micrometers thick and is commonly used to detect fissures on material surfaces [22]. Fig. 7b illustrates the observation that the measurements made on modified surfaces exhibit much less artefacts and are therefore much better suited for quantitative evaluation than the measurements made on non-modified surfaces. We believe this to be the most appropriate method to attain equal levels of visible light reflectivity over the whole sprayed tooth surface region without any penetration of light into the tooth body.

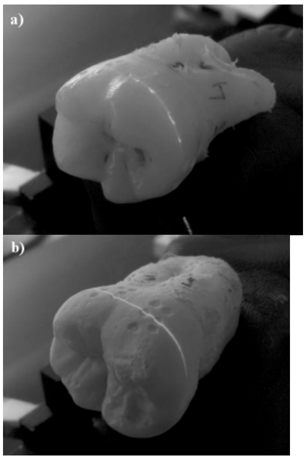


Fig. 6: Photographs of teeth under inspection. The bright curve is the reflected laser beam. a) Tooth with a non-modified surface; b) tooth with a modified surface to enhance diffuse reflection.

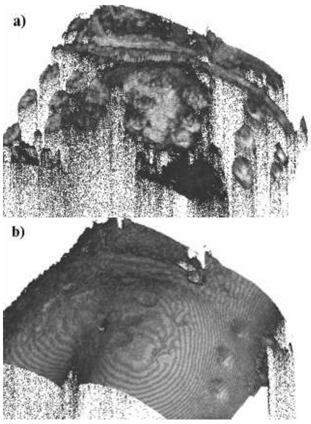


Fig. 7: Computer rendered images of a measured tooth surface: (a) non-modified surface, (b) modified surface.

Accuracy and Repeatability

In order to evaluate the accuracy and repeatability of the measurement system, we manufactured a CNC precision milled and drilled aluminium alloy block. Nine holes with different shapes and sizes were drilled, approximately covering the range of the expected crater sizes (volumes). In table 1 we show the obtained repeatability of the corresponding volume measurements as obtained with the profilometer. Three situations were evaluated: without sample repositioning, with sample repositioning, and with repeated surface modification with the white powder. During sample repositioning the tooth was removed from the translation platform before each surface measurement, and then placed back on the translation platform in different coordinates and tilt angles. This procedure simulated the situation when the tooth is removed from the translation platform in order to have its surface modified with the powder. Repeated surface modification was performed by removing the powder from the surface with water, drying the surface, and then re-modifying the surface with the powder again. The obtained repeatability was better than 2% for all three situations, with the largest deviations obtained in repeated surface modifications.

Measurement condition	Repeatability (St. dev. / Mean value) [%]
Without sample repositioning	0.8
With sample repositioning	1.5
With repeated surface modification	1.8

Table 1. Repeatability of volume measurements using a reference sample with conical borehole (2.60 mm surface diameter, 60° aperture angle, and 1.50 mm depth). Number of measurements per borehole is 5.

The accuracy of the volumetric measurements with the profilometer was also evaluated by measuring the depths and diameters of reference holes with an optical microscope. The results obtained with both methods differed by less than 5% on average.

RESULTS

Erbium laser pulse durations

It has been shown that the pulse duration of an ablating laser can have a significant effect on its ablation efficiency.[2,7] Namely, if the energy required is delivered into the target within a very short time, then the energy has little time to escape from the ablated volume, and so less heat is diffused into the surrounding tissue, resulting in a higher ablation efficiency. For this reason, we decided to measure the pulse durations for both erbium laser systems used in our experiments. The Waterlase MD laser system allowed two pulse durations settings, H and S, while the Fidelis Plus III system allowed five pulse duration settings (SSP, VSP, SP, LP and VLP). Fig. 8 shows the measured pulse durations at 300 mJ of laser energy as obtained with the Fidelis Plus III laser system. Depending on the laser pulse energy, the Fidelis Plus III system's Er:YAG laser pulse durations were adjustable from 50 µs to 1000 µs.

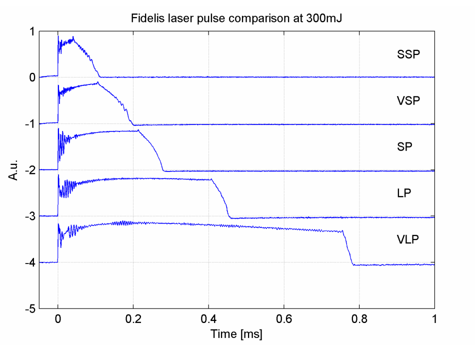


Fig.8: Measured variable pulse duration range of an Er:YAG laser (Fidelis Plus III, Fotona)

The measured range of pulse durations with the Waterlase MD system's Er,Cr:YSGG laser was found to be smaller, and towards longer pulse durations (See Fig. 9).

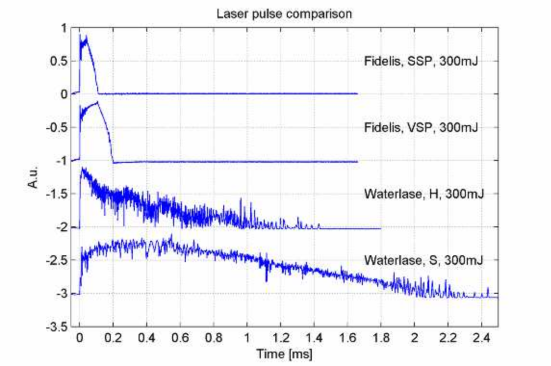


Fig.9: Comparison of shortest pulsewidths for the tested laser sources.

Note that the Waterlase MD laser system employs relatively short pump pulses of only 140 µs in the H mode, and 700 µs in the S mode. In spite of this, due to the presence of the Cr^{3+} ion in the Er,Cr:YSGG laser crystal, the generated laser pulses are much longer, and are in the shortest H pulse mode in the order of 400-1000 µs.

Maximum ablation rate and speed

The initial measurements concentrated on the ablation rate (AR), i.e. the ablated volume per pulse energy (in mm^3/J), for a 300mJ pulse of both lasers (300mJ was the maximum pulse energy available from the Waterlase MD Er,Cr:YSGG system). All AR data represent average values for a single pulse. It was found (see Fig. 10) that the volume of dentine per pulse energy ablated by the Er:YAG system (73 mm^3/mJ) was greater, by a factor of 1.4, than that ablated by the Er,Cr:YSGG system (53 mm^3/mJ). For comparison, results of an earlier published study are included that show a lower rate of volume removal of 16 mm^3/mJ for the Er,Cr:YSGG.[6] We attribute this difference to the high sensitivity of the Er,Cr:YSGG laser ablation process to any reduction in intensity of the beam (which can be caused for example by an aging fiber tip) that can result in the laser moving from cold ablation to a less efficient thermal regime.

In enamel, the AR of the Er:YAG system (32 mm^3/mJ) was greater, by a factor of 1.5, compared to that achieved with the Er,Cr:YSGG system (21 mm^3/mJ).

It is important to note that the AR of the Er:YAG laser system increased with higher laser pulse energies. Thus when the Er:YAG laser pulse energy was increased from 300 to 450 mJ the AR in enamel and dentine increased by approximately 10%.

Measurements of the maximum drilling speeds available from the two laser types (see Fig. 11) were then made. Each laser was configured to the settings recommended for maximum drilling efficiency. A test drilling procedure with fixed duration was completed in samples, and the ablated volume measured, resulting in an ablation speed measurement in mm^3/s. Two settings were used for the Er:YAG system, one for precise ablation (300 mJ at 30 Hz), and the other a specialized mode, designed specifically for very high speed removal of hard tissue (MAX mode - 1000 mJ at 20 Hz). For the Er,Cr:YSGG laser system maximum recommended settings of 300 mJ and 25 Hz were used.

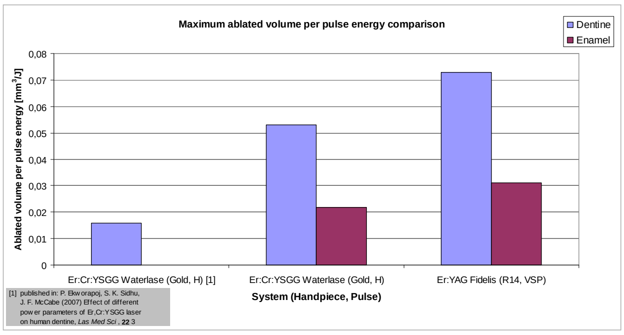


Fig.10: Plot of measured results of ablated volume per pulse energy of dentine and enamel for both laser sources.

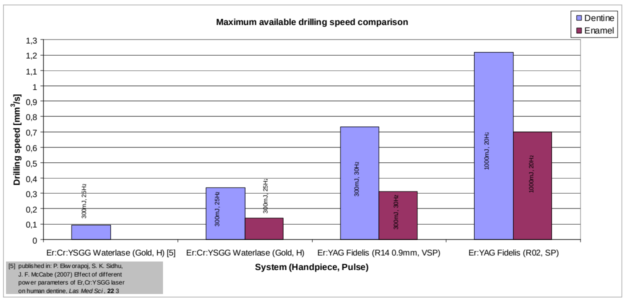


Fig.11: Plot of measured results of drilling speeds in dentine and enamel for both laser sources.

Measurements show that the Er,Cr:YSGG laser removed enamel at a speed of 0.14 mm^3/s, and dentine at the speed of 0.33 mm^3/s (faster than the 0.1 mm^3/s measured in the previous study[6]). The Er:YAG laser precise ablation settings resulted in removal speeds of 0.72 mm^3/s for dentine and 0.31 mm^3/s for enamel; approximately a factor of 2.2 faster than the Er,Cr:YSGG laser. When we consider MAX mode in the Er:YAG laser, the results show ablation speeds of 1.21 mm^3/s (3.7 times faster compared to the ablation rate of the Er,Cr:YSGG) in dentine and 0.70 mm^3/s for enamel (5.0 times faster compared to the AR of the Er,Cr:YSGG).

It is important to note that even at the very high Er:YAG system MAX mode ablation speeds, the ablation regime remained cold and therefore no thermal damage to the teeth could be observed (see Fig. 12).

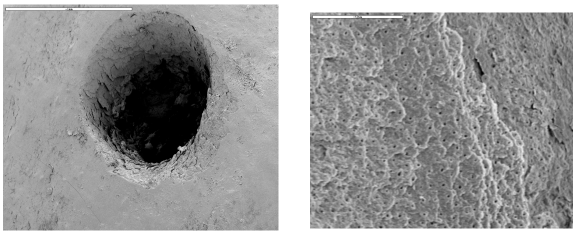


Fig.12: Surface electron microscope (SEM) pictures as obtained following the ablation with the Er:YAG MAX mode in enamel (left) and dentine (right). There are no fissures or charring at the cavity border, and the dentinal tubules are wide open.

On the other hand, with the Er,Cr:YSGG laser thermal damage in the form of brownish discolored spots could be consistently observed in the dentine, almost independently of the laser energy or the repetition rate (See Fig.13). This was in spite of the water spray that was used in all of our experiments. While further research is needed to determine the cause of this effect we tentatively attribute this thermal effect to a higher absorption of the Er,Cr:YSGG laser wavelength (2.78 µm compared to the 2.94 µm wavelength) of the Er:YAG laser [1]) and the high, 20% content of the organic material (consisting mostly of collagen type 1) in dentine.

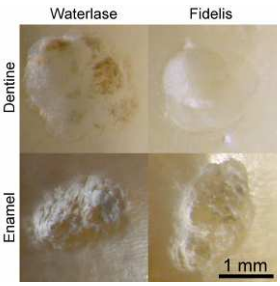


Fig.13: Photographs of ablated craters in dentine and enamel following the irradiation with an Er,Cr:YSGG laser (Waterlase MD, Biolase) and an Er:YAG laser (Fidelis Plus III, Fotona). No thermal damage could be seen in enamel for either of the systems. In dentine, however, brownish spots were consistently observed following the Er,Cr:YSGG laser treatment.

Our AR measurements show consistency with previously published study results.[7,23] One previously published study using the Er:YAG laser [7] reported approximate crater depths of 0.03-0.05 mm per 50 J/cm^2 pulse in dentine, and approximate crater depths of 0.01-0.02 mm per 50 J/cm^2 per pulse in enamel. Our triangulation measurements with the Er:YAG, and under similar conditions reveal approximate depths (calculated by dividing the measured ablated volume with the laser spot area) at 50J/cm^2 of 0.03-0.04 mm in dentine and 0.015 mm in enamel. Similarly, a recent research study with the Er,Cr:YSGG laser [23] revealed a single 50 J/cm^2 pulse ablation rate in enamel of 0.005 mm^3, consistent with our measured rate of 0.004 mm^3.

A comparative measurement of the ablation (drilling) speed of a high speed drill has been recently made.[24] The high speed drill used in the study was an S 68535KR 090, Komet/Gebr. Barasseler, Lemgo, Germany, with diamond particles size of 125µm and diameter of 0.9 mm. The applied pressure of the drill on the tooth was 15N. Measurements with the drill and with the Fidelis Plus III laser were made under the same conditions, and using the optical triangulation method. The obtained drilling (ablation) speed with the high speed drill of 0.3 mm^3/s was lower compared to the ablation speeds of the Er:YAG laser (see Fig. 11).

ANALYSIS

If we look at the theory behind these two laser types we can see that laser physics is only one aspect of the explanation of the performance disparities measured. An investigation of the specification and operating differences between the Er:YAG and the Er,Cr:YSGG laser systems used can clarify further reasons for the differences.

Wavelength Considerations

Wavelength is a key factor in the suitability of any laser for hard tissue procedures in dentistry. The absorption of laser energy in water and hydroxyapatite is related to the laser wavelength in accordance with the curve shown in Fig. 14.

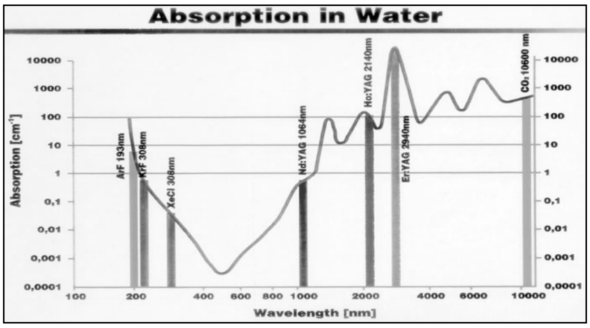


Fig.14: Absorption curve for water showing the absorption peak inhabited by the Erbium lasers, just below 3000nm, and the absorption of various other laser sources.

Erbium laser wavelengths all operate in the region of the major absorption peak for water, and are thus the most suited to hard tissue ablation treatments. Both CO_2 and Ho:YAG lasers show significantly lower absorption in water and are thus less suited for treatments in this field.

Closer study of the absorption peak associated with Erbium lasers shows a 300% difference between the absorption coefficients µ of Er,Cr:YSGG (400 mm^{-1}) and Er:YAG (1200 mm^{-1}) (see Fig. 15). Because of the different water and hydroxyapatite content levels in human dentine, the absorption coefficients for the Er:YAG lasers are approximately 150 mm^{-1} in enamel, and 200 mm^{-1} in dentine. The corresponding absorption coefficients for the Er,Cr:YSGG are approximately three times lower.

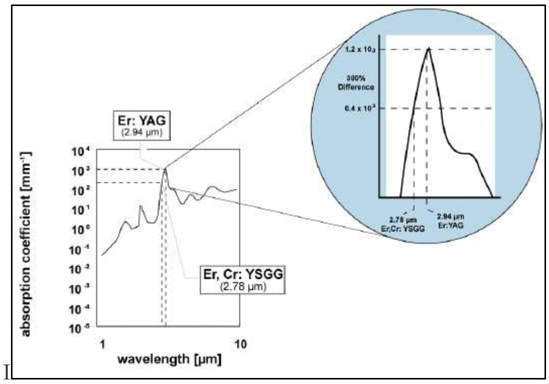


Fig.15: Detail of the absorption curve in the region of both Er:YAG and Er,Cr:YSGG showing the magnitude of the difference in absorption between the two.

The Er:YAG laser wavelength thus penetrates approximately $1/\mu = 7\mu m$ in the enamel, and 5 µm in the dentine. The Er,Cr:YSGG laser wavelength penetrates deeper, 21 µm in enamel, and 15 µm in dentine.

This difference influences the volume of the directly illuminated tissue that needs to be rapidly heated to ablative temperatures by the laser light before the absorbed energy is spread out into the surrounding tissue by the process of thermal diffusion (see Fig. 16).

The higher the penetration depth, the larger the volume of directly heated tissue that needs to be rapidly heated up, and the higher the laser pulse power that is required for efficient and cold ablation.

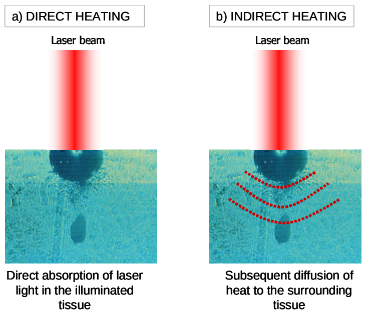


Fig.16: Two steps in tissue heating upon laser irradiation. Indirect heating must be avoided when efficient cold ablation of hard tissues is needed as the indirect heating leads to undesirable thermal effects.

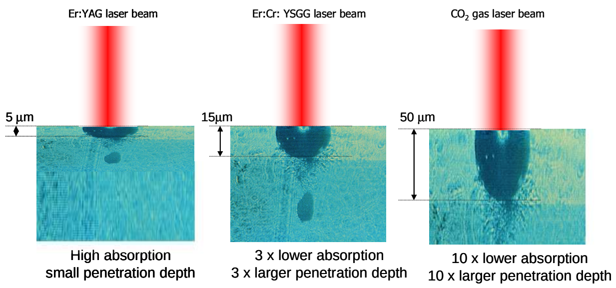


Fig.17: Depending on the laser type, different volumes of the illuminated tissue need to be directly heated. The penetration depths are for human dentine. At lower absorption coefficients, and therefore higher penetration depths more laser pulse power is required in order to avoid secondary heating of the tissue.

In general, lasers with longer penetration depth cause more thermal damage. This is due to the fact that even at high ablation rates a larger volume (one penetration length deep) of non-ablated heated material at the bottom of the ablated craters always remains (see Fig. 17). In addition, the ablation thresholds are higher for higher penetration depths resulting in more heat being transferred to the tissue before the surface ablation is initiated.

Pulse duration considerations

In laser ablation we generally talk about four ablation regimes.[7] At high energies and low pulse durations (i.e. at high laser pulse powers), the ablation speed is higher than the rate at which heat diffuses into the tissue. All laser energy is thus used up in COLD ABLATION (see Fig. 18). With decreasing energies and/or longer pulse durations (i.e. with lower laser pulse powers), the layer of tissue that has indirectly been heated becomes thicker. Thermal effects become more pronounced and, with these, ablation efficiency is considerably reduced (WARM ABLATION and, at even lower energies, HOT ABLATION). At energies below the ablation threshold there is NO ABLATION and all the energy is released in the form of heat, irrespective of the laser pulse duration.

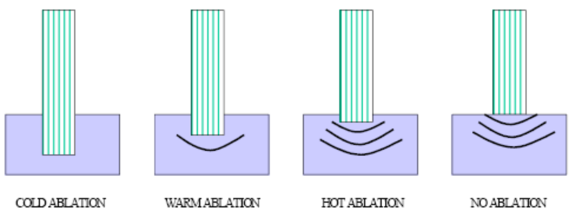
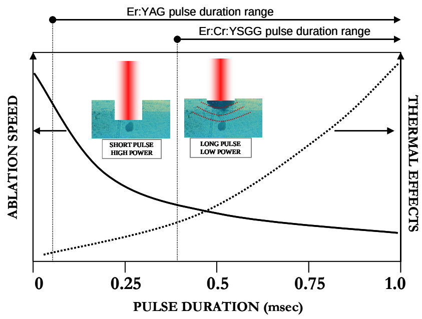


Fig.18: The effect of the laser beam on tissue in the four ablation regimes.

One of the key factors that determines the regime and efficiency of laser ablation is the laser pulse duration. If the energy required is delivered into the target within a very short time, then the energy has little time to escape from the ablated volume, and so less heat is diffused into the surrounding tissue (see Fig. 19).



Fyig.19: Keeping the pulse energy constant, the ablation efficiency increases, and the thermal effects decrease towards shorter pulse durations. Due to the long cross-relaxation time of the Cr^{3+} ion, the Er,Cr:YSGG cannot be operated bellow approximately 400 µs.

In this respect, the Er:YAG laser is at an advantage, since it offers variable pulsewidths down to 50µs. The Er,Cr:YSGG laser is limited to a minimum pulse width of approximately 400µs due to the long cross-relaxation time of the Cr^{3+} ion (see Figs. 8 and 9).

Pulse shape considerations

Pulse shape should also be considered, as this has a strong influence on the 'true' pulse width and power. Figs 8 and 9 indicate that the pulse profile of the Er:YAG laser with VSP technology is controlled and ensures that the power within the pulses is approximately constant. This ensures that the pulse modality does not uncontrollably shift during a pulse from "cold ablation" at the beginning of a pulse (where short Er,Cr:YSGG laser pulses have a peak), to "warm ablation" at the middle of a pulse, and to "hot ablation" towards the end of a pulse.

Discussion

For precise and safe hard tissue procedures in both enamel and dentin, it is recommended to operate at laser energies and pulse durations that are significantly above the ablation threshold. Based on the wavelength and pulse duration considerations, the Er,Cr:YSGG laser is found to be suitable for soft tissue applications where some degree of thermal coagulation effects is desirable. The Er,Cr:YSGG laser has limitations when used on hard tissues. On the other hand, the Er:YAG laser, especially when pumped with Variable Square Pulse (VSP) technology can be operated at adjustable pulse durations, from super short pulses (SSP) that are ideal for precise ablation of hard tissues, to very long pulses (VLP) for soft tissue procedures (see Fig. 13).

With the unavailability of Er,Cr:YSGG systems capable of generating pulse energies greater than 300 mJ, and pulse durations bellow approximately 500 µs it can be understood that the ability to operate in a purely cold ablative regime with these systems is limited. This means that any reduction in intensity of the beam (which effectively reduces pulse power, and can be caused by a strong water spray, an aging fiber tip, or scattering from the ablation plume) can result in the laser moving from cold ablation to a warm, or even a hot regime. Note that an important cause for the reduction in laser pulse power can also be the varying of tip/beam angle by the practitioner.

CONCLUSIONS

A novel, highly accurate and repeatable methodology for the measurement of ablated volumes in teeth has been developed. Since this method does not require the laser handpiece to be in a fixed position with regard to the tooth measurements can be made under realistic conditions and identical to manually administered laser treatments by a dental practitioner. Using this methodology, a detailed comparison has been made between the two leading laser wavelengths for hard tissue procedures in dentistry, Er:YAG (Fidelis Plus III, Fotona) and Er,Cr:YSGG (Waterlase MD, Biolase).

At 300 mJ output laser energy, the ablation rate (ablated volume per pulse energy) in both enamel and dentine was greater, by a factor of 1.4 -1.5, with the Er:YAG laser. It was not possible to make comparative measurements of the two technologies at pulse energies greater than 300 mJ as these pulse energies are not available from any commercially-available Er,Cr:YSGG system. However, the measured ablation rates at higher Er:YAG laser pulse energies indicate an approximately two times higher ablation rate compared with the Er,Cr:YSGG laser. Measurements of the maximum available drilling speeds (ablated volume per second) for both laser systems revealed 3.7 times higher ablation rates in dentine, and 5.0 times higher ablation rates in enamel with the Er:YAG laser.

If we look at the theory behind these two laser types we can see that laser physics only partly explains the measured performance disparities (explaining the difference in the ablation rate by the factor of 1.5-2). Another part (explaining the remaining difference in the ablation speed by the factor of 5.0) is the difference in the operating capabilities between the particular Er:YAG laser system (20W in the Fidelis Plus III laser system) and the Er,Cr:YSGG laser system (8W in the Biolase Waterlase MD laser system) used in the experiments.

The study shows that both wavelengths are suitable for hard dental tissue treatments, but that the absorption and pulse duration characteristics of the Er:YAG laser wavelength mean that it is the more efficient and safer of the two. The study also proves that the latest technology Er:YAG laser systems are capable of matching or improving on the ablation speed of high speed drills.

REFERENCES

1. R. Hibst, Lasers for Caries Removal and Cavity Preparation: State of the Art and Future Directions. J. Oral Laser Appl. 2:203-211 (2002).
2. M. Lukac, M. Marincek, L. Grad, Super VSP Er:YAG Pulses for Fast and Precise Cavity Preparation, J. Oral Laser Appl. 4:171-173 (2004).
3. A. Gorkic, T. Perhavec, D. Bracun, J. Diaci, Using Laser Profilometry for Characterisation of Craters Produced in Hard Dental Tissue by Laser Ablation, to be published.
4. D. Bracun, M. Jezersek, J. Diaci, Triangualation Model Taking into Account Light Sheet Curvature, Meas.Sci.Techn., 17: 2191-2196 (2006).
5. P. Ekworapoj, S. K. Sidhu, J. F. McCabe (2007) Effect of different power parameters of Er,Cr:YSGG laser on human dentine, Las Med Sci, 22 3.
6. A. Gorkic, T. Perhavec, D. Bracun, M. Marincek, J. Diaci, Using Laser Profilometry for Characterisation of Craters Produced in Hard Dental Tissue by Laser Ablation, to be published.
7. B. Majaron, D. Sustercic, M. Lukac, U. Skaleric, N. Funduk. Heat Diffusion and Debris Screening in Er:YAG Laser Ablation of Hard Biological Tissues. Appl. Phys. B 66,1-9 (1998).
8. Li Z Z, Code J E and Van De Merwe W P 1992 Er:YAG laser ablation of enamel and dentin of human teeth: determination of ablation rates at various fluences and pulse repetition rates *Lasers Surg Med* **12** 625-30
9. Sarafetinides A A, Khabbaz M G, Makropoulou M I and Kar A K 1999 Picosecond laser ablation of dentine in endodontics *Lasers Med Sci* **14** 168-174
10. Forrester P, Bol K, Lilge L and Marjoribanks R 2006 Effects of heat transfer and energy absorption in the ablation of biological tissues by pulsetrain-burst (>100 MHz) ultrafast laser processing *Proc. of SPIE* **6343** 63430J-1-63430J-7
11. Rode A V, Gamaly E G, Luther-Davies B, Taylor B T, Graessel M, Dawes J M, Chan A, Lowe R M and Hannaford P 2003 Precision ablation of dental enamel using a subpicosecond pulsed laser *Aust Dent J* **48** 233-9
12. Ohmi M, Tanizawa M, Fukunaga A and Haruna M 2005 In-situ observation of tissue laser ablation using optical coherence tomography *Optical and Quantum Electronics* **37** 1175-1183
13. Mercer C E, Anderson P and Davis G R 2003 Sequential 3D X-ray microtomographic measurement of enamel and dentine ablation by an Er:YAG laser BDJ **194** 99-104
14. Albert M, Leo K, Katrin S and Reinhard H 1997 3D volume-ablation rate and thermal side effects with the Er:YAG and Nd:YAG laser Dent Mater **13** 246-251
15. Mehl A, Gloger W, Kunzelmann K H and Hickel R 1997 A new optical 3-D device for the detection of wear *J Dent Res* **76** 1799-807
16. Bracun D, Jezersek M and Diaci J 2006 Triangulation model taking into account light sheet curvature *Meas. Sci. Technol.* **17** 2191-2196
17. Liang C-K, Peng Y-C and Chen H H 2005 Rolling Shutter Distortion Correction *VCIP* **5960** 1315-1322
18. Heikkilä J and Silvén O 1997 A four-step camera calibration procedure with implicit image correction *CVPR* 1106-1112
19. Bouguet J Y Camera Calibration Toolbox for Matlab, http://www.vision.caltech.edu/bouguetj/calib_doc/index.html
20. Sandwell D T 1987 Biharmonic Spline Interpolation of GEOS-3 and SEASAT Altimeter Data *Geophysical Research Letters* **2** 139-142
21. M. Lukac, T. Perhavec, J. Diaci, Compariosn of Er:YAG and Er,Cr:YSGG dental lasers, presented the International Conference of the SOLA Society in Bangalore, India, Feb 8-9 (2008).
22. Standard-Check Entwickler Nr. 3 (121.200.501) http://www.bsksiegen.de/index.php?option=com_content&task=view&id=70&Itemid=115 http://www.bsksiegen.de/images/stories/datenblaetter/Pruefmittel/sdc_medium_nr3_entwickler_spray.pdf

23. H.W. Kang, I. Rizou, A.J.Welch, Hard tissue ablation with a spray assisted mid-IR laser, Pys.Med.Biol. 52, 7243-7259 (2007).

24. I. Miletić, A. Baraba, S. Krmek, Ž. Božić, T. Perhavec, I. Anić, Comparison between Er:YAG laser and high speed handpiece: prepared cavity volumes and ablation speed rate in enamel and dentin in vitro. Temperature changes and SEM observations associated with Er:YAG preparation, to be published.

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