Ablation Studies of Enamel Tissue using Pulsed HF Laser

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***Abstract*-We describe studies of the interaction of @400ns duration, multiline (2.6-3) mm HF laser with enamel tissue. The etch rate experiment showed an ablation threshold fluence of 47 Jcm^{-2} but using the dispersion theory for optical coefficients of n and k for enamel at dominant laser lines, this value reduced to @25 Jcm^{-2} making it more comparable with other investigations. The main mechanism of enamel ablation at 2.78mm where its absorption is very high thought to be due to microexplosive evaporation of water within enamel matrix ie. thermo-mechanical decomposition. Also, using photothermal deflection technique showed that at fluences below threshold a considerable amount of heat was emitted from the surface which was detected as a single compressive wave. The spectroscopic studies indicated that calcium was the main chemical element observed in plasma spanning from (420-620)nm.**

Index Terms- HF laser, enamel, photothermal deflection, Spectroscopy

I. Introduction

Laser ablation of dental tissue has extensively been studied over the past decades (1-2). The main reasons for applying lasers ranging from IR to UV in dentistry are thought to be due to having less pain compared with mechanical high speed drilling process, selective tissue removal, analgesic and anti-inflamatory effects and also the benefit of more cost-effective treatment. Increasing interest in the use of IR lasers for ablation of dental hard substances such as dentin as well as enamel has raised major concerns regarding a better understanding of laser-tissue interaction mechanism in a given case (3-4). A great deal of work has been done to demonstrate the potential use of pulsed HF laser for the ablation of organic materials (5-8). However, not much work is done about the analysis of a multimode hydrogen fluoride laser interaction with dental tissues. There area number of well established methods which can be used to analyze the laser-material interaction including photothermal deflection, PTD (9), photoacoustics, PA (10) and spectroscopy (11). In general, both PTD and PA are based on rapid heating of a sample after the absorption of optical energy. Consequently a thermal and acoustic change of refractive index ie. medium modulation is produced. The beam deflection can be due to thermal waves crossing the probe beam at below ablation threshold or plume and shock wave fronts at above threshold. This paper is to describe studies used to evaluate the interaction of pulsed HF laser with enamel tissue using etch rate, PTD and plasma spectroscopy experiments.

II. Materials and Methods

Sample preparation

In this study twenty freshly extracted non-carious human third molars (wisdom) were used. Some of about 2mm thick and 4x3 dimension and others were kept intact. Before the experiment the samples were stored in buffered saline containing 0.2% thymol solution to prevent bacterial growth. Immediately after the experiment the samples were placed in 0.9%sodium chloride. Prior to experiments the samples were air dried and mounted on a disk with paraffin wax. The cooling system was comprised of air and water spray in order to inhibit any heat expenditure around the treated area.

Experimental Technique

The home-built laser used in these experiments operated on a SF_6-C_3H_8 gas mixture at low pressure of60 torr and was excited by a fast, transverse, high voltage discharge. Multiline output energies up to 500mJ in a ≅400 ns FWHM was obtained at a pulse repetition frequency of ≅0.2 Hz. Spectral measurements revealed that 20 transitions spanning the wavelength range 2.67-2.96 µm appeared in the output with dominant emission at 2.76, 2.78, 2.82 and 2.92µm. The experimental set up is shown in figure 1 where the energy measurements were made using a Gen-Tec pyroelectric joulemeter. An InAs photodiode (JudsonJ12) allowed the relative laser output to be monitored on a shot-by shot basis.

Material removal measurements were made by exposing the samples to a predetermined number of laser pulses, n, and measuring the depth of material removed,Δ,using a high resolution optical microscope removed,(euromex ± 2µm depth resolution). The average etch rate (removed tissue depth per pulse)was then calculated from Δ/n(Fig.2). More information was achieved regarding the interaction mechanism using PTD technique. In these experiments, the probe beam was a He-Ne laser of 3mW power and was focused by a lens of 100mm focal length to a waist of few microns, passed parallel to the sample surface oriented normally to the HF laser. Deflection of the probe beam was measured as a function of pulse number (Fig.3) by a fast rising time of 1ns PIN-silicone photodiode. The output signal was then recorded by a 150MHz digital oscilloscope (Hitachi VC-7102).

The plasma produced during laser interaction was conducted to a high resolution monochromator by an optical fibre for analysis of ablation plume. The resolved signals were then detected by a PMT (RCA) detector whose output was fed to the computer for data storage and further processing.

Experimental Results

Figure 2 shows the etch depth per pulse as a function of fluence for the HF laser ablation of enamel tissue. In the general case of a sample in which i-the absorption is dominant ie. á>> â, ii-ablation commences instantaneously once a threshold fluence F_t is exceeded and iii-if the plume retains the same absorption coefficient as the condensed phase, then one can use the well known Beer's law where the etch rate varies as the logarithm of fluence.

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$$x = \alpha^{-1} \ln\left(\frac{F}{F_t}\right) \qquad (1)$$

Where X, a and F are etch depth per pulse, absorption coefficient and fluence respectively. A fit to the data using equation (1) yields $F_t \cong 47Jcm^{-2}$ and $\alpha \cong 1780cm^{-1}$ for the threshold and effective absorption coefficient, respectively. In order to characterize the interaction process, the PTD experiment was carried out and some quantitative information was obtained in a non-destructive manner. By assuming a Gaussian probe beam one can write its relation with respect to deflection as follows.

$$\Delta V = V_o \, \mathrm{erf}[\sqrt{2}\frac{\Phi}{\theta}] \qquad (2)$$

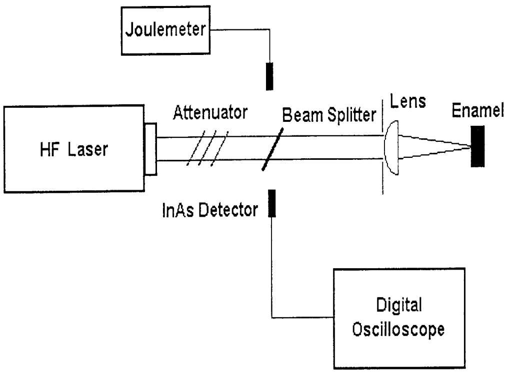
With ϕ and θ begin the beam angular deflection and angular divergence due to change in index of refraction respectively. ΔV is the photodiode output voltage at $\phi=0$, erf is the complimentary error function ie.

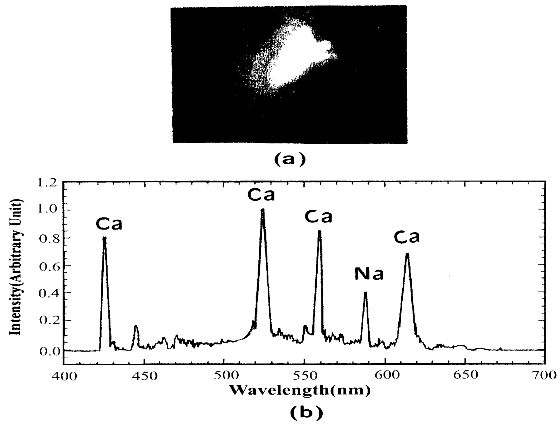
$$erf\ (x) = \frac{1}{2\sqrt{p}}\int_o^x e^{-t^2}\,dt$$

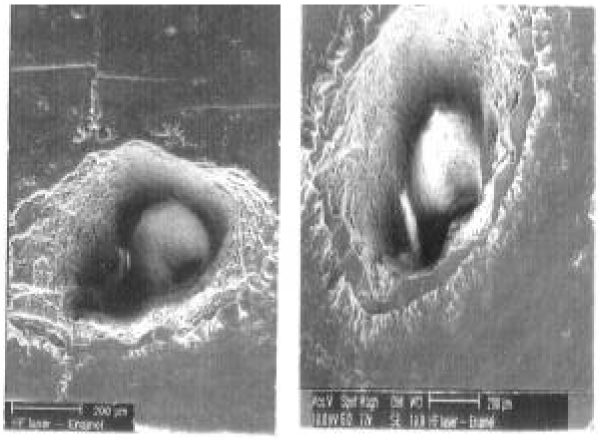
Figure 3 shows the output voltage (hence deflection) a function of laser pulse number at 23 Jcm^{-2}. It can be seen tha e voltage amplitude decreases with pulse number and soon it ren s constant after 4 pulses indicating that heat emission from the en l surface has saturated.

Figure 4a shows the ablated enamel surface at $80Jcm^{-2}$ and it appears a relatively clean ablation with smooth inner wal s obtained. However, in order to inhibit the enamel surface n dehydration and change of its optical properties, it was r sprayed after defined number of pulses. This is thought to b e main reason for observation of modified surface seen as rims e periphery of the ablation crater (Fig. 4b). An example of laser-induced plasma is shown in figure 5a where the combusion of fast ejected particles is clearly seen at the surface. The colour of plasma gradually changed from blue when the enamel was dry to orange-like when it was water sprayed.

Figure 5b is the spectrum of enamel plasma spanning from 420-620 nm with its strongest emission lines of calcium at 526nm and 559 nm respectively.







(a) (b)

Figure 4. Scanning electron micrographs of enan ablated with HF laser in air. (a) 200 pulses at 67 Jcn (b) 190 pulses at 80 Jcm^{-2}. In both case the sample w

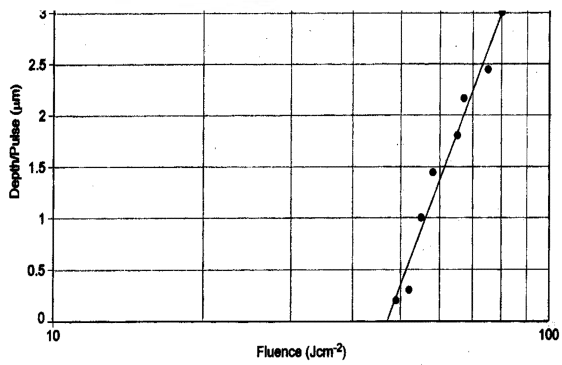


Figure 2. Plot of Etch depth per pulse Vs. logrithm of fluence.

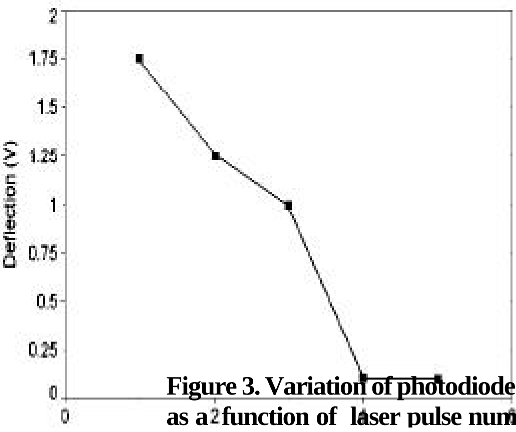


Figure 3. Variation of photodiode output voltage amplitude as a function of laser pulse number of 23 Jcm^{-2} and 2mm

III. Discussion

It is well known that the use of mid-IR lasers in the field of periodontology such as treatment of carious dentin and enamel has been successful. The reason is thought to be due to high absorption coefficients of hydroxyapatite [$Ca_{10}(Po_4)_6(OH_2)$] with a total composition weight of 90% at 9.6μm and 2.8μm respectively. Also, the remaining 10% water content has a high absorption peak at 2.9µm. Therefore, the effective ablation mechanism in (2.7-3)µm is microablation accompanied with tissue matrix inter-cellular water evaporation. Since hydroxyapatite has high energy absorption in this range, its melting and vaporization at high laser fluences also affect the interaction process (4,12).

It has been found that the values of fluence threshold and absorption coefficient are relatively high compared with similar investigations. To interpret and justify the results obtained we begin with the minimum energy per unit volume accumulated for enamel ablation which is about $\alpha F_t \cong 84$ Jcm^{-3} and that is much higher than the energy needed to vapourize the minerals in enamel (4). The key point we have to consider here is that not all the pulse energy is expended for tissue ablation. The propagation of electromagnetic wave in a linear medium can be determined by optical constants n and k which are the real and imaginary part of refractive index respectively. These constants in turn depend on dielectric constant, K and medium conductivity, δ. In order to correctly use the electromagnetic wave theory one ought to express the changes occurring in above parameters in terms of radiation frequency and dispersion. ie. Drude-Lorentz theory (13). Simulation of the theory for enamel showed that one would expect to have a great deal of surface reflection, R of 2.76µm and 2.78µm wavelengths near the resonance frequency of hydroxyapatite at 2.8µm. Assuming that the radiation is incident normally to enamel surface, then R can be defined in term of K as follow.

$$R = \frac{(n-1)^2 + K^2}{(n+1) + K^2} \tag{4}$$

Also, the real and imaginary part of K ie. K_r and K_i are related to n and k by the following expression.

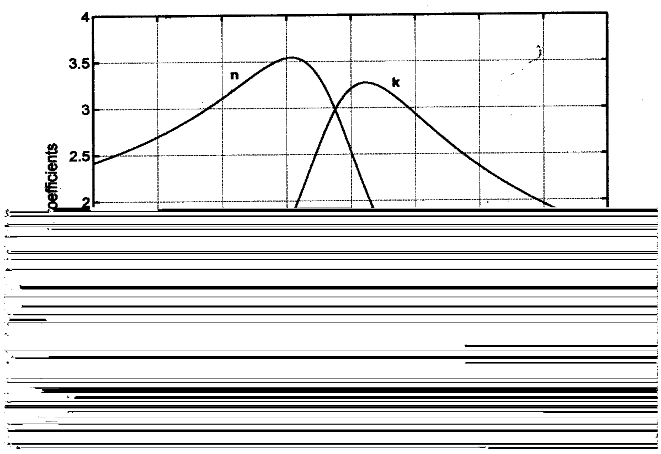


Figure 6. Plot of optical coefficients (n,k,R) of enamel as a function of inverse wavelength.

$$\begin{aligned} n &= [\frac{1}{2}(k_r + \sqrt{k_r^2 + k_i^2})]^{\frac{1}{2}} \\ k &= [\frac{1}{2}(-k_r + \sqrt{k_r^2 + k_i^2})]^{\frac{1}{2}} \end{aligned} \tag{5}$$

Where k_r and k_i are each defined in terms of resonance frequency, ω_o and the subsidery frequency, ω near ω_o. Figure 6 shows the variation of n, k and R with inverse wavelength and hence ω_o The values of incident energy E_i, reflectivity of the surface R and transmitted energy in tissue for interaction E_T at each dominant laser modes are given in table below.

It can be seen that only ≅ 54% of fluence threshold plays an effective role in ablation process. Thus, the initial value of 47Jcm^{-2} becomes ≅ 25 Jcm^{-2} which is closer to 18 Jcm^{-2} obtained by seka etal (12) using Er:YSGG laser. It must be emphasized that at 2.91 µm only 12.2% of incident energy is spent for the enamel ablation and despite the fact that this wavelength is very close to the absorption peak of water at 2.94µm no enamel removal takes place. The reason is simply because the 2.91µm wavelength carries only 5.5 $Jcm^{-2}$of fluence threshold (7Jcm^{-2}) for enamel ablation via direct tissue evaporation using Er:YAG laser (14). It is well known that the water presence plays an important role in the material ablation by energy absorption and hence making the cavitation effect more renounce. Initially, the voltage amplitude was very high (Fig.3) but as the interaction proceeded it decreased due to dehydration of enamel. Thus by keeping the wetness of tissue surface at a relatively constant level one can also hope to keep the plasma intensity constant. This is important since water spraying at high fluences can easily cause mechanical damage seen as "rims" at around the ablation zone (Fig.4b). This effect can be explained in terms of shockwaves and ejected particles escaping with high velocity. The result of spectroscopy in figure 5b shows that almost all the peaks corresponds to calcium with one sodium line at 589nm. It is interesting to notice that no emission line representing the phosphor element was obtained. This may be due to fact that the energy of HF laser in this experiment was not high enough to cause the dissociation of Ca_{10} $(Po_4)_6$ $(OH)_2$ to release the $(PO_4)_6$ and hence remained intact in bulk form.

Table 1. Calculated values of incident enegry, reflectivity and transmitted energy of different HF laser lins for enamel tissue.

Laser (µm) mode	E_i%	R%	E_T%
2.76	15.5	58	6.5
2.78	44.5	55	20
2.82	27.7	46	15
2.91	12.2	-	12.2
	100		≅ 54

IV. Conclusion

We believe that in the range of 2.76-2.78µm where the secondary absorption peak of hydroxyapatite lies, a less violent and cleaner ablation of enamel compared with its main absorption peak at 9.6 µm is produced. Also, by constantly keeping the surface of enamel damp, most of the pulse energy will be spent to vaporize the water during the ablation hence screening the tissue from possible plasma damage. Change of enamel state due to variation in its absorption coefficient was monitored below fluence threshold using PTD experiment. The main mineral element detected in spectroscopy was calcium. It seems that a pulsed HF laser with advantages such as i-wavelength tunability between (2.6-3) µm, ii-high absorption coefficient of hydroxyapatite at 2.8µm, iii-ability of optical fibre delivery and iv-lack of carcinogenetic effects make this laser potentially a suitable candidate for some medical applications.

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