study of thermal effects for dentin tissue ablation by 350fsec laser pulses

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Abstract:

The mechanism of photo ablation for life tissues had been shown. The thermal effects for dentin tissue ablation by 350fsec laser pulses of a (1.05 μm) Ti:Sapphire as maximum temperature for tissue ($T_{max}$), thermal gradient ($\frac{\partial T}{\partial z}$), threshold ablative intensity (I), released thermal power per area (Q), vaporization energy per volume ($E_c$), thermal diffusivity ($\chi$), effective thermal diffusivity length ($\ell_\chi$), thermal relaxation time ($\tau$), threshold ablative fluence ($F_1$) and final ablative fluence ($F_2$), as well as coefficients which specify ablation achievement success as ablation depth (d), ablation velocity (v), absorption coefficient (α), start ablation time ($t_1$) and final ablation time ($t_2$) had been calculated as tissue temperature variation in one time and at incident laser fluence variation in other. These results had been interpreted and the relation between thermal effects and ablation success coefficients were specified. It can be concluded that the higher laser fluence causes ablation achievement in higher velocity. The main conclusion is that the maximum temperature ($T_{max}$), released thermal power per area (Q) and vaporization energy per volume ($E_c$) causes high ablation depth (d) within shortest time.
key words: laser ablation, dentin, tissue, thermal effects.

1- Theoretical part:

Many studies were directed to laser irradiation interaction with biological tissues [1-29], J.Neev, et al., presented a study of using ultra short pulse lasers in hard tissue ablation [30]. In this research, we present a study of the parameters which control the dentin tissue ablation by 350-fsec laser pulse.

Many interaction mechanisms may occur when applying laser light on biological tissue. These mechanisms are depends on the optical tissue properties and the laser radiation characteristics itself such as: exposure time, applied energy, focal spot size, energy density and power density. These mechanisms are photo chemical interactions, thermal interactions, photo ablation, plasma-induced ablation and photo disruption [31].
photo chemical interactions take place at very low power densities (1 Watt/cm²) and long exposure times ranging from seconds to continuous wave. They occur when the laser light induce chemical effects and reactions within macromolecules or tissues, where the energy release due to photo synthesis [31].

The thermal interaction stands for a large groupe of interaction types, where the increase in local temperature is the significant parameter change. However, depending on the duration and peak value of the tissue temperature achieved, different effects like coagulation, vaporization, carbonization and melting may be distinguished [31].

The temperature of the tissue below the surface (T) may be calculated as following [30]:

\[ T = \frac{2QT}{\rho c \chi} \] ........................ (1)

where (Q) is the thermal power released per unit area, (t) is the exposure time, (\(\rho\)) , (c) are the density and heat capacity of the tissue, respectively and (\(\chi\)) is the thermal diffusivity of tissue which given as [30]:

\[ \chi = \frac{1}{(\frac{\partial T}{\partial t}) A} \] ............ (2)

where (\(\frac{\partial T}{\partial t}\)) is the time variation of temperature, A is affected area by laser and \(\Delta T\) is the difference between initial tissue surface temperature and that in time (t).

The vaporization energy per unit volume (\(E_v\)) for tissue is given by [30]:

\[ E_v = \frac{Q}{\Delta a} \] ............ (3)

where (a) is the dentin tissue thickness.

The temperature gradient (\(\frac{\partial T}{\partial z}\)) within tissue surface was given as below [30]:

\[ \frac{\partial T}{\partial z} = - \frac{Q}{\rho c \chi} \] ............ (4)
The effective thermal diffusion length \( \ell_{\chi} \) is given by [30]:

\[
\ell_{\chi} = \frac{1}{(\chi \tau_p)^{1/2}} \quad \ldots \ldots (5)
\]

where \( \tau_p \) is the laser pulse duration. Thermal relaxation time \( \tau \) for laser pulse within biological tissue and maximum of tissue temperature \( T_{\text{max}} \) may be shown as eqs. (6) and (7), respectively [2,30]:

\[
\tau = \frac{1}{\chi \alpha^2} \quad \ldots \ldots (6)
\]

\[
T_{\text{max}} = \frac{\alpha \tau I}{\rho c} \quad \ldots \ldots (7)
\]

where \( \alpha \) is the tissue absorption coefficient which given as [30]:

\[
\alpha = \left( \frac{\delta x}{\delta t} \right) \rho c \chi \ell_{\chi} \quad \ldots \ldots (8)
\]

where \( I \) is the laser intensity which causes the ablation. Photo ablation is other interaction where occurs at \( 10^7 - 10^8 \) Watt/cm\(^2\) laser intensity and laser pulse durations in the nanosecond range. The threshold laser intensity to start ablation \( I \) is given by [31]:

\[
I = I_o \exp \left( -\alpha d \right) \quad \ldots \ldots (9)
\]

where \( I_o \) is the incident laser light on tissue and \( d \) is the ablation depth within tissue which may be approximated to [31]:

\[
d = \frac{2.3}{\alpha} \log_{10} \frac{I_o}{I} \quad \ldots \ldots (10)
\]

laser intensity \( I \) can be given in other form as [2]:

\[
I = v c \rho T \quad \ldots \ldots (11)
\]

where \( v \) is the velocity of ablation which illustrated as [30]:

\[
v = \left( \frac{Q \tau}{\rho c} \right)^{1/2} \quad \ldots \ldots (12)
\]

The threshold laser fluence to start ablation \( F_{1} \) can be calculated as [2]:

\[
F_1 = \frac{\chi \rho}{\alpha \tau} \quad \ldots \ldots (13)
\]

and the needed time to start ablation \( t_1 \) was given by [2]:

\[
t_1 = \frac{\rho c T}{\alpha I_o} \quad \ldots \ldots (14)
\]
To achieve whole laser ablation, it may be needed to laser fluence ($F_2$) as [2]:

$$F_2 = F_1 \sqrt{\frac{t_2}{\tau}} \quad \cdots \quad (15)$$

Ablation process covers time ($t_2$) to achieve it completely which given by [2]:

$$t_2 = \frac{\chi c^2 p^2 T^2}{I^2} \quad \cdots \quad (16)$$

When obtaining power densities exceeding $10^{11}$ Watt/cm$^2$, plasma-induced ablation will occur and very clean and well-defined removal of tissue without evidence of thermal or mechanical damage can be achieved.

Photo disruption is the physical effects associated with optical breakdown as plasma formation and shock wave generation. If breakdown occurs inside soft tissues, cavitation and jet formation may additionally take place. During photo disruption, the tissue is split by mechanical forces. Whereas plasma-induced ablation is spatially confined to the breakdown region, shock wave and cavitation effects propagate into adjacent tissue, thus limiting the localizability of the interaction zone.

In this research, we are focus on thermal changing effects in laser ablation for dentin tissue.

Figures (1-4) show the micrographs of dentin craters which were resulted by ablation at different laser fluence. These micrographs were taken on a Philips 515 (Mohawk, N.J) scanning electron microscope.
**Fig. 1** ablated dentin crater at 1J/cm² [30]

**Fig. 2** : side view of dentin ablation crater [30].
**Fig. 3:** profile of a 3-J/cm² dentin crater floor and underlying tissue [30].

**Fig. 4:** dentin crater following ablation by 100 pulses of 34J/cm², 1-nsec pulses [30].
2- Experimental part:

Laser pulses generated by a (1.05μm) Ti:Sapphire chirped pulse amplifier (CPA) system were used. The basic laser system is illustrated in Fig.(5). Seed pulses of (100fsec) from a Kerr-lens mode–locked, Ti:Sapphire oscillator were stretched to (1nsec) in a four–pass, single–grating pulse stretcher. Amplification by nearly \((10^9)\) to the (6mJ) range was achieved in the TEM\(_\infty\) 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 (2.3mm) beam diameter and reduced non linear effects. The system operated at (10Hz); however, single pulses could be extracted as well [30].
After amplification, the pulses are compressed 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 psec) to (1 nsec) were obtained. Smooth, reproducible, high-quality beam intensity profiles are important in order to ensure uniform interaction at the targeted surface [30].

Ablation measurements were performed with laser spot size of (0.5-mm) diameter. Laser pulses were focused onto the tissue sample by a 1m 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 [30].

Ablation studies were performed using thin dentin slices (0.5-1)mm thick were cut from the middle section of freshly extracted third molars, in a plane perpendicular to the occluso–cervical direction. Teeth were treated with 0.5 MEDTA for (2min) to remove the smear layer, then stored in 10% thymol solution until treatment. Slices were prepared parallel to the crown and washed with demineralized water [30].

Ablation rates were determined by viewing the crater edge and bottom with an optical microscope coupled to a calibrated micro positioning depth gauge (digital gauge DG-2100s, sony, Japan). Craters generated by 100 pulses were examined [30].

Following irradiation, the dentin was washed briefly with 100% ETOH and mounted on stubs using colloidal silver liquid (Ted pella, Inc. Redding, CA.) with the laser treated areas pointing upward and then gold coated on a PAC-1 pelco advanced coater 9500 (Ted pella, Inc. Redding, CA).

3- calculations & results:

The thermal effects such as thermal diffusivity ($\chi$), effective thermal diffusivity length ($\ell_\chi$), thermal power per area ($Q$), vaporization energy per volume ($E_v$), maximum temperature for tissue ($T_{\text{max}}$), thermal gradient ($\frac{\partial T}{\partial z}$), threshold ablative intensity ($I$), threshold ablative fluence ($F_1$) and final ablative fluence ($F_2$), as well as coefficients which specify ablation achievement success as ablation depth ($d$), ablation velocity ($v$), absorption coefficient ($\alpha$), start ablation time ($t_1$) and final ablation time ($t_2$) had been calculated at different values of time at (3J/cm$^2$) fluence and incident laser intensity on tissue ($I_o$) of ($3x10^{14}$Watt/m$^2$) dependent on (t-T) data in Ref.(30) as shown in table (1). In other hand, the data of (fluence, ablation depth) variations was dependent in
calculation each of ((I) , (\(\frac{\partial T}{\partial z}\)) , (Q) , (E_e) , (T_{max}) , (v) , (F_1) , (t_1) , (F_2) , (t_2) , (t)) at \((\chi =2.2 \times 10^{-7} \text{ m}^2/\text{sec} )\), \((\ell_{\chi} = 3.603 \times 10^{16} \text{ m}^{-1})\), \((T_o= 303 \text{ K})\), \((\alpha = 10.25 \times 10^{-16} \text{ m}^{-1})\) and \((\tau = 4.326 \times 10^{16} \text{ sec} )\) had be shown in table (2).

The temperature of dentin tissue surface (T), thermal diffusivity (\(\chi\)), evaporisation energy per volume (E_e), temperature gradient \((\frac{\partial T}{\partial z})\), effective thermal diffusivity length (\(\ell_{\chi}\)), thermal relaxation time (\(\tau\)) and maximum temperature for dentin tissue surface (T_{max}) had been calculated using eqs.(1-7), respectively where (a) effected tissue area by laser of \((0.1962 \mu\text{m}^2)\).

The absorption coefficient (\(\alpha\)), threshold ablative intensity (I), ablative depth (d), ablative velocity (v) had been calculated according to eqs.(8-11), while eqs.(13-16) were used to calculate each of ablative threshold fluence (F_1), time needed to start ablation (t_1), final ablative fluence (F_2) and final ablative time (t_2).

The temporal variation for each of \((E_e\text{ with } I)\), \((Q\text{ with } d)\), \((\frac{\partial T}{\partial z}\text{ with } v)\), \((F_1\text{ with } t_1)\) and \((F_2\text{ with } t_2)\) had been drawn in figs.[(6-a)-(6-e)] , respectively , while figs.[(7-a)-(7-e)] illustrate the variation of these parameters at different values of fluence.

**Table (1) :** The parameters which depended in calculation at (t,T) variation

<table>
<thead>
<tr>
<th>data</th>
<th>Symbol</th>
<th>Value</th>
<th>Notes</th>
</tr>
</thead>
<tbody>
<tr>
<td>Temporal variation of</td>
<td>(\frac{\partial T}{\partial t})</td>
<td>1.0083 K/sec</td>
<td>calculated using T-t data in Ref.[30]</td>
</tr>
<tr>
<td>temperature</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Laser pulse duration</td>
<td>t_p</td>
<td>350fsec</td>
<td>Ref.[30]</td>
</tr>
<tr>
<td>Heat capacity of dentin tissue</td>
<td>C</td>
<td>1x10^{10} J.m^{-1}</td>
<td>Ref.[30]</td>
</tr>
<tr>
<td>Density of dentin tissue</td>
<td>(\rho)</td>
<td>1.07 \times 10^3 Kg/m^3</td>
<td>Ref.[30]</td>
</tr>
<tr>
<td>Thickness of dentin tissue</td>
<td>(a)</td>
<td>1x10^{-3} m</td>
<td>Ref.[30]</td>
</tr>
<tr>
<td>Incident laser intensity</td>
<td>I_o</td>
<td>3x10^{14} Watt/m^2</td>
<td>calculated using energy-duration data in Ref.[30]</td>
</tr>
</tbody>
</table>

**Table(3) :** The parameters which depended in calculation at (Fluence,ablation depth) variation
<table>
<thead>
<tr>
<th>data</th>
<th>Symbol</th>
<th>Value</th>
<th>notes</th>
</tr>
</thead>
<tbody>
<tr>
<td>Initial temperature of dentin surface</td>
<td>T</td>
<td>303 K</td>
<td>Ref.[30]</td>
</tr>
<tr>
<td>Laser pulse duration</td>
<td>tp</td>
<td>350fsec</td>
<td>Ref.[30]</td>
</tr>
<tr>
<td>Heat capacity of dentin tissue</td>
<td>C</td>
<td>1x10^10 J/m^3</td>
<td>Ref.[30]</td>
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</tr>
<tr>
<td>Thickness of dentin tissue</td>
<td>a</td>
<td>1x10^-3 m</td>
<td>Ref.[30]</td>
</tr>
<tr>
<td>Absorption coefficient of dentin tissue</td>
<td>( \alpha )</td>
<td>10.25x10^-16 m^-1</td>
<td>Calculated using Eq.8</td>
</tr>
<tr>
<td>Thermal diffusivity of dentin tissue</td>
<td>( \chi )</td>
<td>2.2x10^-7 m^2/sec</td>
<td>Calculated using Eq.4</td>
</tr>
<tr>
<td>Effective thermal diffusivity of dentin tissue</td>
<td>( \ell \chi )</td>
<td>3.603x10^16 m^-1</td>
<td>Calculated using Eq.5</td>
</tr>
<tr>
<td>Thermal relaxation of dentin tissue</td>
<td>( \tau )</td>
<td>4.326 x10^36 sec</td>
<td>Calculated using Eq.6</td>
</tr>
</tbody>
</table>

![Graph](image-url)
Fig.(6): The temporal variation for each of:

(a) evaporation energy $E_e$ and ablative intensity $I$
(b) thermal power $Q$ and ablation depth $d$
(c) temperature gradient $\delta T \delta Z$ and ablation velocity $(v)$
(d) start ablative fluence ($F_1$) and start ablative time ($t_1$)
(e) Final ablative fluence ($F_2$) and final ablative time ($t_2$)
Fig. 7: The effect of laser fluence variation in each of of:

(a) evaporation energy $E_e$ and ablative intensity $I$
(b) thermal power $Q$ and ablation depth $d$
(c) temperature gradient $\frac{\delta T}{\delta Z}$ and ablation velocity ($v$)
(d) start ablative fluence ($F_1$) and start ablative time ($t_1$)
(e) Final ablative fluence ($F_2$) and final ablative time ($t_2$)

4- Discussion:

The continuously applying for constant laser energy on dentin tissue with spent time, leads to available high incident laser power density on tissue which causes increasing in each of tissue temperature (high $T$) and abundant ablative intensity (high $I$) as shown in table (2-a,b) and fig.(6-a), respectively. Large ablation depth has been achieved according that (high $d$) and laser ablation has been occurred with high velocity (high $v$) as shown in table (2-b) and in fig.(6-b,c) because increasing in maximum temperature for tissue (high $T_{max}$) as illustrated in table (2-a).

It can be noticed that the increasing in time which required to ablation start ($t_1$) as given in table (2-c) inspite of high temperature for dentin tissue. This can be attributed to increase in loss energy due to high thermal diffusivity (high $\chi$) as shown in (table 2-a) on tissue surface and decreasing of effective thermal diffusivity length within deeper layers of tissue (low $l_\chi$) as shown in table (2-a) which can be interpreted by decreasing in absorption coefficient inside tissue (low $\alpha$) as given in table (2-b). Each of temperature gradient $\frac{\delta T}{\delta Z}$ and temperature difference ($\Delta T$) as shown in table (2-a) had been decreased with tissue surface temperature increasing because of tissue temperature becomes equal approximately.

Laser intensity has been exploited in ablation achieving which leads to decrease in each of released thermal power per area ($Q$) (low $Q$) as shown in table (2-b) and vaporization energy per volume (low $E_v$) as illustrated in table (2-b).

We can notice from table (2-c) that thermal relaxation time ($\tau$) has been increased with temperature increasing due to high temperature of dentin tissue surface which may be arrived. That is the same reason to achieve complete ablation process in less threshold ablation fluence (low $F_1$),
less end ablation fluence (less F₂) within less time (low t₂) as shown in table (2-c) and fig.(6-d,e).

The higher laser fluence means increasing in incident laser intensity on dentin tissue which causes abundant in intensity for ablation as shown in in table (4-a) and fig.(6-a). Ablation process will be achieved in high (v) and with less time (less t₁) and less (t₂) because high quality of available laser fluence to start and end ablation (high F₁ and F₂) as given in fig.(7-c,d,e) and table (4-b).

Laser fluence has been utilized in tissue thickness ablation to depth (d) which increases with fluence as table (4-b) and fig.(7-b). This can be attributed to high temperature gradient \( \frac{\partial T}{\partial z} \) as shown in table (4-a) and fig.(7-c) which means high temperature difference between affected area in dentin tissue and other adjacent areas.

This difference increases maximum temperature in affected tissue area (high T∧max) as illustrated in table (4-a). In result, high thermal power per area will be released (high Q) and large energy per volume (large Eₑ) has been available to vaporization, as shown in table (4-a) and fig.(7-a,b).

5- Conclusions:

It can be concluded that the high laser fluence is the best way to achieve laser ablation in high velocity and in less start ablation time (t₁) and final ablation time (t₂) because large available laser intensity to achieve ablation. Anormouse reduce for each of thermal diffusivity, effective thermal length and absorption coefficient has the main role for ablation accelerate. The main conclusion is that the large temperature difference between affected tissue area and other areas, increasing in maximum temperature and higher thermal power per area, are help to achieve higher ablation depth and with higher ablation velocity.
6-References :