Iraqi J. Laser, Part A, Vol.11, pp. 13-19 (2012)

IRAQI JOURNAL OF LASER

One dimensional Finite Element Solution of Moving Boundaries in Far IR Laser Tissue Ablation

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(Received 22 May 2012; accepted 18 November 2012)

Abstract: In this work, the finite element analysis of moving coordinates has been used to study the thermal behavior of the tissue subjected to both continuous wave and pulsed CO_2 laser. The results are compared with previously published data, and a good agreement has been found, which verifies the implemented theory. Some conclusions are obtained; As pulse width decreases, or repetition rate increases, or fluence increases then the char depth is decreased which can be explained by an increase in induced energy or its rate, which increases the ablation rate, leading to a decrease in char depth. Thus: An increase in the fluence or decreasing pulse width or increasing repetition rate will increase ablation rate, which will increase the depth of cutting. The selection of a proper laser parameters may be helpful for doctors in obtaining optimum advantages in such treatment.

Introduction

Since their first invitation in the 1960s, lasers have been used widely in medical applications. The science of laser tissue interactions has been studied extensively over the past decades [1,2]. Carbon dioxide CO_2 lasers are widely used as cutting tools in dermatological treatment such as laser resurfacing [3,4], and in many other medical fields [1,5]. One of the most useful characteristics of pulsed mode CO_2 laser in surgery is its ability to incise tissue accurately with minimal thermal damage of the surrounding healthy tissue and simultaneous bleeding of the small vessels [6].

Since the analytical solution has failed until now to adequately model the transient behavior of tissue subjected to the laser heat source, especially when considering pulsed mode operation, numerical methods have been used to calculate thermal and mechanical events in transient and qusi-steady state situation through the solution. Great amounts of work have been devoted to the photo-thermal modeling of laser tissue ablation during the past 20 years. Vasiliev and Serkov [7] considered a porous structure of biological tissue. They proposed a one dimension finite difference model of biological destruction using a multi-boundary Stefan problem. LeCarpentier et al. [8] investigated thermal and mechanical events during continuous wave (CW) argon laser ablation of biological media by 1-D Finite Element Analysis. Zhoui and Herwig [9] considered Moving Finite Difference to solve the bio-heat equation in tissue subjected to CW CO_2 laser based on the model suggested by reference [7]. They compared their result with experimental data and good agreement was found. Shibib et al [10], used Finite Element of Fixed Mesh technique to study the effect of tissue porosity on the thermal behavior of tissue subjected to CW CO_2 laser irradiation. Zhang et al [11], studied the dynamic photo-thermal model of CO_2 laser tissue ablation based on an improved version of McKenzie's three-zone model, where the finite difference method was used to solve numerically the bio-heat equation for the temperature, deposited laser energy fields and the boundary positions of the ablation and char zones. Compared with previously published data, a good agreement was found.

In this work, the Finite Element of Moving Coordinates technique is used to study the thermal behavior of tissue subjected to CW and pulsed CO_2 laser. A simple procedure in transform the coordinates to simplify the solution is followed. A coordinate transformation has been used so that the solution for temperature distribution and the location of ablation and evaporation fronts can be obtained.

Mechanism

In far IR lasers, such as CO_2, laser photons are predominantly absorbed in tissue, and their energy converted to heat. Thus scattering, reflection and transmittance of photons can be ignored at this wavelength (10.6 μm). If laser power is high enough, then the deposition of heat within the tissue will cause a rapid increase in its temperature, leading to denaturation of proteins, coagulation, evaporation (which leaves carbonized tissue), and ablation. In this work, CW and pulsed mode laser radiation are studied. The explosion effect, which can be produced at high intense laser power, is ignored. This limits the model to thermal behavior of tissue only where power intensity is not high enough to cause a water explosion and the pulse duration is much greater than the time period of stress confinement. In this case, only macro-scale thermal confinement or diffusion is applicable.

Govering Partial Differential Equation And Boundary Conditions

Thermal laser tissue interactions can be described by the bio-heat equation. Ignoring metabolic heat generation and perfusion of heat due to blood flow, the heat transfer equation through tissue subjected to laser as a heat source can be reduced to [11]:

$$\rho c \frac{\partial T}{\partial t} = k\nabla^2 T + \dot{Q} \qquad (1)$$

Where

In char layer $\rho c = \rho_c c_c \, , k = k_c$

In virgin layer $\rho c = \rho_v c_v \, , k = k_v$

Furthermore, the model includes an element where char and virgin layers exist. Here equivalent properties can be easily determined due to the position of the evaporation front, see Figure 1. Assuming that the thermally affected depth is much smaller than the spot diameter ,then eq.(1) can be reduced to a one dimensional model, i.e.:

$$\rho c \frac{\partial T}{\partial t} = k \frac{\partial^2 T}{\partial x^2} + \dot{Q} \qquad (2)$$

Knowing that the tissue pore diameter (6 μm) [7],is less than CO_2 laser wavelength (10.6 μm) and the emissivity of char layer is close to unity, the heat generation due to incident laser photons can be approximated as heat adding at the surface [7,9]. In addition, considering convection, radiation ,heat transfer from the surface and ablation of the char; the boundary condition at the char surface can be written as:

$$q_s = q_o - \dot{m}_c H_c - h(T_s - T_\infty) \qquad (3)$$

Where $\dot{m}_c = \rho_c V_c$ (3a)

And $V_c = a_c \exp(\frac{-H_c M_c}{R T_c})$,[7]

$h = h_{co} + h_r$ Where $h_{co} = 25 W/m^2.K$,[12]

and $h_r = 4\varepsilon\sigma\left(\frac{T_s + T_\infty}{2}\right)^3$,[13] .

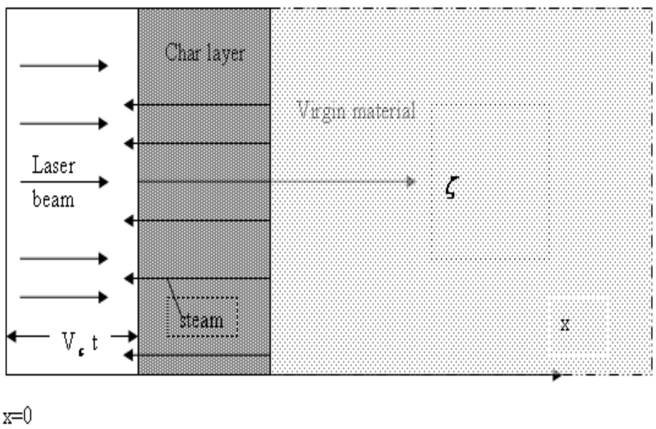

Fig. (1): Studied domain and coordinates.

Also the boundary condition at evaporation front is [7]:

$$-k\frac{\partial T}{\partial x}_{x+} = -k\frac{\partial T}{\partial x}_{x-} - \dot{m}_w H_{ev} \qquad (4)$$

Where $\dot{m}_w = \varphi\ \rho_w V_w$

and

$$V_w = a_w \frac{\exp(\frac{-H_{ev} M_w}{RT_w})}{1 + \frac{d}{\delta}}$$

In the solution the porosity is equal to 0.7 and the environmental temperature is 25° C.

In CW mode laser irradiation, the above equation is applied directly as it is. In pulsed mode, if the temperature of the upper tissue is high enough to supply the necessary latent heat of water vaporization, evaporation occurs at the evaporation front. Otherwise evaporation stops due to insufficient heat conduction and eq.(4) returns to its ordinary form.

This model is valid for pulse widths much greater than the time required for stress confinement (i.e. pulse width $>> 1/\ \mu_a a$) , By following the work of Payne [14] in his definition to specify the macro-scale and micro-scale heat confinement, a condition for heat confinement is achieved if:

$$Fo \equiv \alpha l^{-2} t_p \leq 1 \qquad (5)$$

Where $l = 1/\mu_a$ for macro scale heat confinement and $l = L$ for micro scale heat confinement ,here L is heat penetration depth= 0.7 μm [1], then using eq(5), the pulse width $\leq$ 700 μs is sufficient to achieve macro-scale heat confinement and pulse width ≤ 1.2 μs can achieve micro-scale heat confinement .The mechanism and dynamics of pulse laser ablation for pulse width where there is no micro-scale thermal confinement(i.e. pulse width ≥ 1.2 μs) are well understood as a photo-thermal event[14].

Coordinate Transformation

A solution for temperature can be simplified by eliminating the moving boundary through the transformation to a moving coordinate ζ whose origin coincides with the instantaneous ablation surface as

$$\zeta = x - V_c t \qquad (6)$$

Then the transformed boundary-value problem becomes

$$\rho c \frac{\partial T}{\partial t} = k \frac{\partial^2 T}{\partial \zeta^2} + \rho\ c V_c \frac{\partial T}{\partial \zeta} \qquad (7)$$

And all the above boundary conditions (i.e. eq (3), (4)) can be easily transformed to the moving coordinate depending on their location with respect to the moving coordinate, see Figure1.

Finite Element Analysis

The discretization of eq.(7) having the aforementioned boundary conditions can be accomplished using Galerkin method [15] .The studied region is divided into 299 elements with 300 nodes and the application of Galerkin method results in a system of ordinary differential equations [15]which are solved using an iterative pattern. A concentration of nodes near severe gradient boundaries is used to ensure non-oscillating result. A computer program has been created using Visual Basic 6 coding to follow the procedure of predicting the temperature distribution through the domain and the two front positions in the region of interest with an iterative procedure.

Results And Discussion

A moving coordinate method is used to deal numerically with tissue subjected to both pulse and CW laser radiation. A computer program is created based on a Finite Element Analysis to predict the temperature distributions, positions of fronts and char depths for both CW and pulse laser modes. No deficiencies exist in using the transformation of coordinate and this simple coordinate transformation can simplify the solution to deal easily with this complex problem. In CW mode, the program has been compared with previously published works [9,10] and a good agreement has been found. The results from this work are closed to the theoretically and the experimentally determined value of the char depth. For power levels in W/cm^2 of 175, 298, 627, the char depths in μm are found from the result of this work to be in order of 60.1, 40.05, 20.1 respectively while in the experimental work of reference [9], these values in μm is found to be 70,39,20 respectively. This may verify the theoretical basis and coding of this program in CW laser mode. To verify the accuracy of the numerical method in pulsed mode, the result of this program is tested against experimental data extracted from reference [16] and an acceptable

agreement has been found. In this work, the theoretical ablation front position is found to be 940 μm, see Figure 2, and in the experimental work of reference [16], it was found that the ablation front was equal approximately to 784 μm after 30 pulses, each of which has18 mJ, repetition rate of 600Hz, pulse width of 160 μs with a spot radius of 230 μm. The damage zone (char depth plus necroses depth) is predicted numerically which is found to be approximately 72 μm while the experimental value for the same situation in reference [16], is found to be equal to 100 $\pm$ 10 μm .The fronts depths for the above pulsed laser parameters are shown in Figure 5, assuming that the thermal damage due to tissue temperature elevation for a period of time can be described by the Arrhenius equation having constants extracted from reference [11].
The difference in results between this study and the experimental data published elsewhere is due to:
1) The approximation of using the one dimensional model. The more precise simulation is the axis-symmetry model.
2) The approximation of heat adding at the surface. A more accurate approximation may be obtained from an absorption model which converges to the real condition where the heat generation is declined through the tissue depth. Also the effect of neglecting scattering in the carbonized region and the mass transfer of water within the tissue, both may reduce the used model accuracy.
3) The scarce in the char thermal properties and their variation with temperature, which may affect the results.
4) Ignoring the effect of water explosion on tissue, which may significantly affect the result especially at high laser intensity where significant ablation mass is lost due to spalliation resulting from this phenomenon [16].
The program is tested for pulsed mode laser irradiation with the parameters that may be used to test its applicability, see table 1;at constant energy per pulse (E/p=2J/cm2), repetition rate of 500 Hz and a pulse width of 1 ms, the temperatures at interfaces are shown in Figure 3. As the pulse is “on”, a dramatic increase in surface temperature is observed reaching a value of 740°C as the domain reaches a quasi-steady state. As pulse is “off” a dramatic reduction in surface temperature is also observed where it will reach a value of 280°C. Evaporation will continue as long as the temperature of the overlying tissue is higher than the evaporation temperature where eq.(4b) is applicable. Stopping of the evaporation condition seems to occur as the pulse width is reduced to 0.4 ms, where at constant repetition rate, the decrease in pulse width will increase the time through which heat may transfer out of the targeted region. In this case, a greater reduction in temperature will be expected. If the temperature required to achieve evaporation at the current location is equal to or higher than the overlying tissue, then evaporation stops.

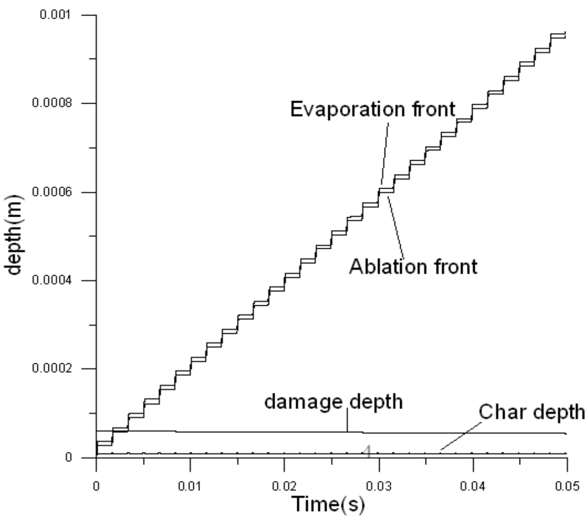

Fig. (2): Influence of pulse CO_2 laser on interfaces depth.

However, this situation will not be maintained long, since the subsequent pulse will cause again a rapid increase in the upper tissue temperature causing evaporation to continue, see Figure 3. The simulation also predict that the fluctuation in interface temperature is increased, this due to
1) The increasing in the time that heat (in its three types) can be transfer out of the region without heat addition which lowers the surface temperature.
2) the increase in maximum recorded temperature when the pulse is “on”.
Before a further study, some discussion is required. As the induced rate of energy or the total induced energy increases, the increase in ablation temperature is much greater than the increase in evaporation temperature. Assuming the decrease or increase in char depth depends mainly on the variation in ablation rate, which itself depends on ablation temperature,

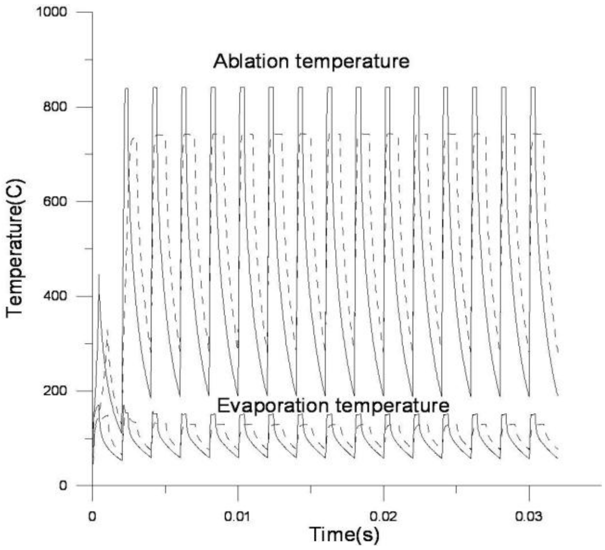

Fig. (3): Temperature at ablation front and evaporation front (solid line) with energy/pulse of 2J/cm^{2} , pulse width of 1 ms ,repetition rate of 500Hz (dashed line) and pulse width of 0.4ms (solid line).

Table (1): Pulsed laser parameters

line no.	Energy/ pulse (J/cm^{2})	Repetition rate (Hz)	Pulse width (ms)
1	1	500	1
2	1	500	0.4
3	2	500	1
4	2	500	0.4
5	2	100	1
6	2	100	0.4
7	1	100	1

Then one can conclude that an increase in surface temperature (or energy or energy rate) will decrease char depth and vice-versa. The effects of pulse laser parameters on char depth are shown in Figure 7,which indicates the char depth history. The reduction in the fluence will reduce the induced energy which will reduce ablation rate. The reduction in ablation rate will then increase the char depth as discuss above, see lines 3, 1and lines 4, 2,in Figure 4. It also shows that the reduction in pulse width will increase the rate of induced energy which will increase ablation rate. The increase in ablation rate will decrease char depth see , lines1,2and lines 3,4 in Figure 4. It also shows that the increase in repetition rate will increase the induced energy to the tissue , then the increase in induced energy will again increase ablation rate, which will decrease char depth, lines 3,5 in the same Figure. Figure 5 indicates ablation front positions with respect to time for different pulse laser parameters. The increase in repetition rate will increase the rate of induced energy (increase maximum recorded surface temperature)which will increase the depth of cutting, see lines 4, 6 and 1, 7 in Figure 5. The decrease in pulse width will increase the rate of induced energy (increase the maximum recorded surface temperature), which will increase depth of cutting, see lines 1,2 and 3,4 in Figure 5. The increase in the fluence will increase the induced energy (or increase the maximum recorded surface temperature), which will increase the depth of cutting, see lines 1,3 and 5, 7 in Figure 8. A Small increase in the thermal damage zone is observed as repetition rate increased in experimental work of reference [16], the authors define the thermal damage zone as the distance measured from the ablation crater to the region where temperature $\geq 65^{\circ}$C, which means that the thermal damage zone as defined in that reference is the sum of the char and the necroses region then his work is not conflicted with this work where a decrease in char depth will be combined with an increase in necroses zone as repetition rate increases [16]. The condensation of vapor which will happen at low repetition rate will increase the thermal damage zone since the transformation region and its surrounding will be kept at condensation temperature, this will increase the size of the damage zone, also a reduction in char depth is expected since ablation will be continued while the evaporation is stopped.

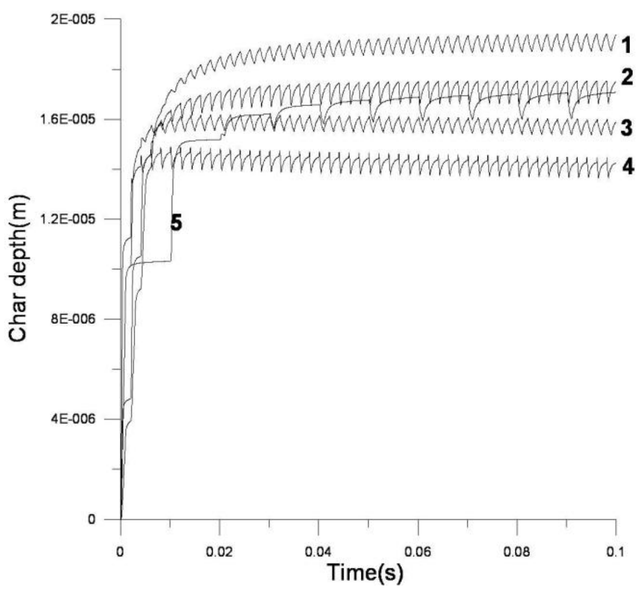

Fig. (4): Char depth history at different pulse laser parameters.

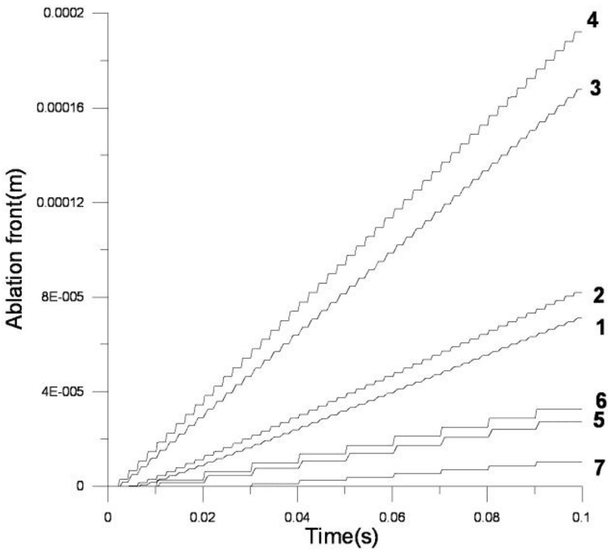

Fig. (5): Front ablation at different pulse laser parameters.

The volume of vapor that remains in tissue is very small, so heat induced to tissue when the vapor is condensed can be ignored and the model can be applicable even when time between pulses is greater than the characteristic time for condensation, which is known to be 20-120 ms [14] where evaporation velocity is zero.

Conclusion

The moving boundary finite element solution has been shown to successfully model thermal ablation phenomena in tissue subjected to far IR laser radiation. The thermal model is limited to a case where mechanical damage may not occur, or it has less effect on the result. Two laser modes are tested and the results give good agreement compared with experimental and theoretical data published elsewhere. In pulse laser with the above conditions, some predictions are made the following predictions can be made from the simulation: An increase in the fluence or decreasing pulse width or increasing repetition rate will increase ablation rate, which will increase the depth of cutting. The selection of proper laser parameters may be helpful for doctors in obtaining optimum advantages in such treatment.

Nomenclatures:

a sound velocity [m/sec]
c specific heat[J/kg .K]
d pore diameter [m]
Fo Fourier number[-]
h convection heat transfer coefficient [w/m^2 .K]
H enthalpy of evaporation for water and sublimation for char[J/kg]
k thermal conductivity[w/m .K]
L heat penetration depth [m]
l depth[m]
$\dot{m}$ mass flow rate[kg/s.m^2]
M molecular weight [kg/kg.mole]
q heat flux[w/m^2]
$\dot{Q}$ heat generation[w/m^3]
R universal gas constant [J/kgmole.K]
t time[s]
T temperature[$^\circ$ C]
V velocity[m/sec]
x dimension variable[m]

Greeks:

μ_a absorption coefficient[m^{-1}]
δ char layer depth [m]
ρ density [kg/m^3]
ε emissivity
σ stefanboltizmann constant
φ porosity
ζ moving coordinate[m]
α thermal diffusivity [m^2 /sec]

Subscripts

c char tissue
co convection
ev evaporation
o induced
p pulse
r radiation
s surface
v virgin layer
w water
∞ environmental

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حل العناصر المحددة احادي البعد للاحداثيات المتحركة في القطع الحراري للانسجة بواسطة ليزر الاشعة تحت الحمراء البعيدة

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الخلاصـة:
في هذا البحث، تم استخدام تحليل العناصر المحددة للاحداثيات المتحركة لدراسة السلوك الحراي للنسيج المعرض لليزر ثنائي اوكسيد الكربون (CO_2) بالنمطين النبضي والمستمر. تمت مقارنة النتائج بالنشريات السابقة ووجد تقارب جيد بينهما مما يحقق النظرية المستخدمة والبرنامج الخاص بها. تم استنباط بعض الاستنتاجات حيث وجد ان عمق الطبقة الفحمية يقل عند انقاص عرض النبضة لليزر النبضي، كما وجد ان زيادة معدل تكرار النبضة او شدة الطاقة (fluence) سيقلل ايضاً من عمق الطبقة الفحمية. زيادة شدة الطاقة او انقاص عرض النبضة يزيد من سرعة القطع الحراري والذي يؤدي لاحقاً الى زيادة عمق القطع. ايضاً تم ملاحظة زيادة في عمق القطع وذلك بزيادة معدل تكرار النبضة. ان اختيار معلمات الليزر المناسبة يكون مجدياً للاطباء في الحصول على الفوائد المثلى في مثل هذه المعالجة.