

Pulse Laser Parameters Effect on Tissue Thermal Damage Zone in Coagulation Process

Khalid Salem Shibib*, Ayad Zwayen Mohammed*
& Kholood Hasan Salih**

Received on: 19/10/2010

Accepted on: 3/3 /2011

Abstract

Owed to wide use of pulsed lasers in a medical field, a deep understanding of their effects on the temperature increase in tissue and the subsequent tissue thermal damage in a coagulation process may be a matter of importance. The influence of laser beam profile, repetition rate and pulse width on temperature distribution and the subsequent thermal damage in tissue are studied using finite element method, which solves the axis-symmetry bio-heat equation in tissue subjected to far IR pulse laser irradiation. Some conclusions are obtained: as energy/pulse remains constant, Gaussian laser beam profile rather than a top-hat beam will increase the in-depth tissue thermal damage at and near the center of the spot region, increasing in repetition rate will increase the temperature distribution and subsequent damage zone. As pulse width decrease, high temperature may be reached leading to cause a quantitatively and qualitatively damage which can be recognize as an increase in the size of the damage zone and a higher value of thermal dose. An increase in pulse width will reduce the rate at which energy deposited in the tissue which result in low extent of temperature increase which result in reduction of the damage zone quantitatively and qualitatively.

Keywords: pulse laser parameters, bio-heat equation, thermal damage, far IR laser, finite element analysis, coagulation process.

تأثير معالم الليزر النبضي على منطقة الضرر الحراري للنسيج في عملية التخثر

الخلاصة

نتيجة الاستخدام الواسع لليزر النبضي في المجال الطبي فقد أصبح الفهم العميق لتأثير معالم الليزر على زيادة درجة حرارة النسيج والضرر الحراري المرافق لها في عملية التخثر من الأمور الهامة. حيث تمت دراست تأثير هيئة حزمة الليزر ومعدل الاعاده وعرض النبضة على توزيع درجات الحرارة والضرر الحراري المرافق لها باستخدام طريقة العناصر المحددة . حيث تم استخدام معادلة الحرارة الاحيائية المتناظرة حول المحور على النسيج المعرض للإشعاع الليزري النبضي في المنطقة تحت الحمراء. بعض الاستنتاجات تم استنباطها حيث بثبات طاقة النبضة فان الهيئة الكاوسيه لحزمة الليزر تسبب ضررا اكبر من الحزمة ذات التوزيع المنتظم للطاقة في عمق وعند مركز البقعة المعرضة للإشعاع الليزري , وبزيادة معدل الاعادة ستزداد توزيع درجات الحرارة والضرر المرافق لها في النسيج. وبنقصان عرض النبضة ستزداد درجات الحرارة والتي تؤدي الى زيادة الضرر نوعيا وكما التي نستطيع تميزها من خلال زيادة عرض منطقة الضرر وزيادة قيمة الجرعة الحرارية. وزيادة

* Laser and Optoelectronics Engineering Department, University of Technology/ Baghdad

**Institute of Technology, Foundation of Technical Education /Baghdad

عرض النبضة سوف يقلل معدل الطاقة المترسبة في النسيج والتي تنتج تقليل مدى زيادة درجات الحرارة والتي ينتج عنها تقليل كمية الضرر نوعا وكما.

1. Introduction

Since its first invitation in the sixtieth of the last century, laser had been used extensively in medical applications. In the eightieth, the continuous wave laser dominates the medical application and due to advance in fiber optics manufacturing, laser material science and advantages of pulse laser mode, new useful pulse laser applications were observed in many medical fields.

Laser tissue interaction models occupied great interest due to the importance of safety and medical consideration where thermal damage may be a matter of importance. Many interaction mechanisms can be observed depending on specific tissue characteristics as well as laser parameters contribute to this diversity, which fill mainly in five categories of interaction types; photochemical interactions, thermal interactions, photo-ablation, plasma-induced ablation, and photo-disruption[1].

The theory of selective photothermolysis was firstly, proposed by Anderson and Parrish [2]. The first lasers to fully exploit this principle (i.e. selective thermolysis) was the Pulsed Dye Lasers introduced in the late 1980's for the treatment of port wine stains and strawberry marks in children, and shortly after, the first Q-switched lasers for the treatment of tattoos.

Now a day, the wide medical application of this laser mode invited a deep understanding of pulse laser

Parameter effects on tissue such as temperature increase and subsequent thermal damage, which is the major motivation of this work. The ability to produce highly localized heating at the desired location has made a pulse laser more attractive to tumor irradiation than continuous lasers. This is because of that for the same energy input; the instantaneous peak power attained during pulsed laser irradiation is greater than that obtained in continuous laser irradiation [3].

The focus of this text is to investigate the thermal effect of pulse far IR laser on tissue and the effect of pulse laser parameters on tissue coagulation, which have been studied theoretically and compare the result with an experimental data. A normal intra-beam laser hits a tissue where as the photons energy transmitted to tissue structure, then tissue temperature rises rapidly where coagulation, evaporation and ablation may occur.

The deposition of heat in thermal laser-tissue interaction models is known to be a function of laser irradiation parameters and optical tissue properties, all these parameters play an important rule in heat deposition and this heat will diffuse to the adjusting tissue where an increase in its temperature is observed which may lead to subsequent thermal damage.

The finite element method has been used to solve an axis-symmetry bio-heat equation to predict temperature distribution numerically; these

temperature distributions are used to predict the thermal damage zone based on thermal dose equation. The effects of different pulse laser parameters have been studied to predict their effect on damage zone where the interaction of laser with tissue is proposed to stop as surface thermal damage initialized and the subsequent thermal damage is obtained till the tissue temperature dropped again below 39 °C where below this temperature, the skin thermal damage can be ignored.

2. Theory

2.1 Partial differential equation and boundary conditions

The partial differential equation that covers the heat transfer inside biological tissue is that proposed by Pennes, which was until now proved too well model the thermal behavior of living tissue [4], an axis-symmetry mathematical formulation of heat transfer equation will be used here due to the symmetrical nature of the problem:

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

Where ρ, c, k are the density, specific heat and thermal conductivity respectively, their values are taken from reference [5], where the water content (W) dependence of the three keys thermal parameters can be described as

$$\rho = 1000 / (0.06w + 0.938) \quad \dots(2a)$$

$$c = 1000(2.5w + 1.7) \quad \dots(2b)$$

$$k = \rho(0.45w + 0.174) / 1000 \quad \dots(2c)$$

The water content of tissue is assumed to be 0.7, which also affect the value of the absorption coefficient. The water is the dominate absorber having a value of 848 cm⁻¹ at 10.6 μ m wave length [5] and the tissue absorption coefficient can be written as [6]

$$\mu_a = w\mu_{a,w} \quad \dots(2d)$$

Terms that refer to living tissue (i.e. metabolic heat generation Q_m and perfusion heat due to blood flow Q_p) can be ignored comparing with heat generation. The studied section is assumed to be a cylinder having elevation and diameter of 1 cm. The initial condition throughout the region is assumed to be 37 °C.

Convection and radiation boundary is assumed at the top surface of the tissue which can be modeled mathematically as:

$$k \frac{\partial T}{\partial n} = h_{co} (T - T_\infty) + h_r (T - T_\infty) \quad \dots\dots(3)$$

Where h_{co} (convection heat transfer coefficient) = 10 W/m².K [7] $\dots(3a)$

h_r (Radiation heat transfer coefficient) = $4\sigma\epsilon \left(\frac{T_s + T_\infty}{2}\right)^3$ [8] $\dots(3b)$

The environmental temperature (T_∞) is 25 °C, while the emissivity (ϵ) of the tissue is taken to be that of skin = 0.96[9]. An adiabatic boundary

condition is applied through the axis of symmetry due to symmetrical nature of the problem. In the relative far in-depths boundaries, insulated boundaries can be assumed (see figure 1) because the temperature distribution there will not be affected by the instantaneous localized induced heat resulting from a laser strike, this assumption may be confirmed also by relatively low thermal conductivity, high specific heat of tissue.

2.2 Heat generation distribution

The heat generation due to laser-tissue interaction depends mainly on the absorption coefficient $\mu_a(z)$ (m^{-1}) and fluence rate $\phi(r, z)$ (W/m^2) then :

$$\dot{Q} = \mu_a(z) \phi(r, z) \dots\dots(4)$$

The terms r and z represent the axial and in-depth coordinates then heat generation can be written as [10,11]:

$$\dot{Q}(r, z, t) = \mu_a (1-rf) I_o \exp[-(\mu_a + (1-g)\mu_s)z] \dots\dots (5)$$

And if absorption dominates the scattering, which is the case for interaction of far IR Laser (such as CO₂ laser) with tissue, then fluence rate at any depth can be covered by Beer-Lambert law and the heat generation through the tissue due to laser photons absorption is:

$$\dot{Q}(r, z, t) = \mu_a \delta(1-rf) I_o \exp(-\mu_a z) \dots\dots(6)$$

Where δ a multiplication factor is used to simulate pulse laser, which has a value of one or zero corresponding to “on” or “off” duration, respectively. At

high wave length (i.e. 10.6 μ m) the reflectivity is almost zero, and waist diameter ($2 w_o$) is taken to be 1 mm to standardize the model. For Gaussian beam distribution the induced localized power intensity is:

$$I_o(r, z) = \frac{2P}{\pi w_o^2} \text{Exp}\left(-\frac{2r^2}{w_o^2}\right) \dots\dots (7)$$

Here P is laser power in watts and for top-hat beam profile the induced power intensity, wherever the radius is less or equal to the radius of the laser beam, is

$$I_o(r, z) = \frac{P}{\pi w_o^2} \dots\dots(8)$$

In the interaction of the far IR laser beam with tissue, μ_s is tended to zero and the general form is written here to simulate green laser-tissue interaction to compare the results of the created computer program with an experimental and theoretical model that used finite volume method, as we will see later.

2.3 Finite element formulation

Equation (1) is discredited in axis-symmetry spatial dimensions and solved using the weak formulation and Galerkin procedure [12]. A total number of 148 triangular elements and 250 nodes are generated following special grid model. A finer element is used in the region of a highly expected temperature gradient and a courser mesh elsewhere. A computer program is created to follow the procedure so that to predict the temperature distribution through the domain and the subsequent thermal damage zone see figure 1.

2.4 Thermal dose equation

The calculation of the thermal dose for changing temperature exposure was done by using the technique suggested by Sapareto and Dewey [13]. This technique used a numerical integration to calculate the time that would give an equivalent thermal dose at a reference temperature (43 °C) under different temperature profiles. The concept of thermal dose [13, 14] had been developed for quantitatively describing the thermal effect on tissue of a temperature elevation. The thermal dose is:

$$D(r; z; t) = \sum_t R^{T_i - 43} dt \quad \dots \quad (9)$$

Where $T_i(r; z; t)$ is the average tissue temperature at a time interval dt in minutes. Thus, the thermal dose $D(r; z; t)$ is defined in terms of equivalent minutes at the reference temperature. The value of R is taken to be 2 for temperature greater than 43 °C and 4 elsewhere. The threshold for lesion formation (100% necrosis) is believed to be 50 to 250 equivalent minutes at 43 °C, depending on the tissue type [15] and the thermal dose of necrosis for beef is taken to be 240 minutes. It had been shown that the volume of a lesion predicted by this model correlates well with the volume of an experimentally produced lesion [15, 16]. Even this modulation is based on ultrasonic as a heat source; it can well model the damage zone where a laser is used as heat source since eq.(9) depends mainly on temperature and time at which the tissue is held on. [17]

3. Result and discussion

Half of the studied region and the used mesh are indicated in figure 1, the upper line of the boundary shows the tissue surface subjected to convection and radiation and the other three boundaries are insulated. A tissue of known thermal and optical properties is subjected to laser irradiation where the laser will strike the surface at waist beam and the laser is cut-off as the necrosis starts, (i.e. its thermal dose is greater than 240 minutes).

Pulse laser parameters are studied to show their effects on tissue damage zone. Typical laser parameters are used here, assume constant energy/pulse (0.2mJ), top hat beam profile, repetition rate of 250Hz, and pulse width of 100 μ s. An enlarged view for the predicted temperature distribution at the initialization of necrosis and at the end of cell necrosis after exposing to laser strike are shown in figures 2, 3.

The influence of different pulse laser parameters is as follows:

3.1 Laser beam profile

Two types of laser beam profile are examined, Gaussian and top-hat type where the repetition rate (500Hz) and energy/pulse (0.2mJ) are constant, the temperature history at the center of the spot region is shown in figure 4, which also shows an experimental temperature reading at the center of the spot region detected by thermal camera.

This is done on a beef tissue sample. The CO_2 laser beam is targeted to tissue with an angle of 45 ° and a paper is fixed between them until a hole is created where the thermal camera will

replace the laser so that it will read exactly the temperature at the center of spot region. Time of laser beam cut-off is 0.015 s, energy /pulse is 0.2mJ, 1ms pulse width and repetition rate is 500Hz. Gaussian laser beam is believed to increase the maximum temperature that could occur through the process more than top-hat beam because of high intense energy at and near the center of the beam, see figure 4. The calculation is terminated as the temperature drops below 39 °C after exposing to laser beam; this is due to the fact that the necroses below this temperature can be neglected. Even the center of the spot region will kept at a higher temperature when top-hat laser is permanently cut-off, its final thermal dose is less than that for Gaussian. This is because, the higher maximum temperature attended through Gaussian pulses is responsible for the increase in value and size of damage zone more than that could occur when top-hat laser pulse is permanently cut-off and forth after, see figures 4, 5, 6. Deeper and less width in the necrosis regions may expect since energy is concentrated near and at the center of Gaussian beam profile, see figure 6. (Line 2&4)

3.2 Repetition rate

Increasing the repetition rate while energy/pulse remains constant will certainly increase the total amount of energy induced to the tissue, this can fairly explain the increased in temperature at the surface of the affected tissue as the repetition rate increased, see figure 7. The value of the thermal dose attended in high repetition rate is greater than that of less repetition rate for the same reason;

see bold line in figures 5&8. Note that the two repetition rates are tested for top hat beam profile. Furthermore, an increased in necroses region may be expected due to the increasing in energy transferred to the tissue as repetition rate increased, see figure .6. (Line 1&2)

3.3 Pulse width

Figure 9 shows the temperature history at center of spot region where as the pulse is "on", the temperature of the tissue will increase, this is because that the generated heat is larger than the heat that transfer out of the targeted region and as pulse is "off" a reduction in temperature is observed, this is due to transfer of heat out of the spot region due to conduction, convection and radiation heat transfer. The intense heat deposition due to short pulse width will lead rapidly to increase in temperature within the spot region more than the expected from longer pulse assume repetition remains constant.

High temperature results from short pulse and due to the dependence of thermal damage directly to this temperature increase then an increase in thermal dose value (qualitatively) and damage zone size (quantitatively) may expect as pulse width decreases, see figures 6, 8. Also as pulse width decrease then, an extension in the damage zone will be expected since the high temperature induced will definitely cause a transfer of heat to the adjusting tissue assume conduction heat transfer will dominate the phenomena of heat transfer (conduction, convection and radiation), this will lead to an increase in the zone that has been affected by conduction heat transfer, which is

shown as an increase in the damage zone as pulse width decreases, see figure 6 (line 1&3).

At shorter pulse width, high temperature may be reached leading to localize high thermal damage (see figure 9) and as energy/pulse remains constant then an increase in pulse width will reduce the rate at which energy may be deposited in the tissue then a low extent of increasing in temperature may expect and consequently low damage zone, see figure 6 (line 1&3).

The motivation of using CO_2 laser in this thermal modulation process is its high absorptivity and very low reflectivity as it interacts with tissue so that any other laser of fewer wavelength will deposit less heat and consequently, reduce the thermal damage zone assume same laser parameters.

Finally, a comparison of the result using the program of this work with an experimental data using a type of laser and tissue properties as mentioned in references [1, 10, 18] which may verify the accuracy of the mathematical model and the finite element analysis of this work, see figure 10.

4. Conclusions

The finite element analysis has been used successfully to predict temperature distribution in tissue subjected to pulse laser irradiation from which thermal damage zone has been obtained using thermal dose equation. Different pulse laser parameters have been tested to predict their effects on coagulation processes. The program from which the result of this work is obtained had been verified by comparing its result with an

experimental and theoretical data and good agreement is obtained.

The effects of different pulse laser parameters in coagulation process assuming same energy/pulse (0.2mJ) are obtained:

a) Gaussian beam distribution has more in-depth and less width effect on thermal damage zone than that of a top hat beam due to intense energy at the center of Gaussian laser beam.

b) As repetition rate increased then more damage zone may expect due to increase in heat deposition in tissue, which led to increase in its temperatures, and the subsequent damage zone.

c) As pulse width decrease, high temperature may be reached leading to cause a quantitatively and qualitatively increase in damage which can be seen as an increasing the size of the damage zone and higher value of thermal dose. An increase in pulse width will reduce the rate at which energy deposited in the tissue then a low extent of increasing in temperature may expect and consequently reduce damage zone quantitatively and qualitatively.

Nomenclature

C - specific heat [$J \cdot kg^{-1} \cdot K^{-1}$]

D - thermal dose [minute]

g - anisotropy factor [-]

h - coefficient of heat transfer

[$W \cdot m^{-2} \cdot K^{-1}$]

I_o - power intensity at the surface

[$W \cdot m^{-2}$]

k - thermal conductivity [$W \cdot m^{-1} \cdot K^{-1}$]

n - vector normal to surface [-]

p - power [W]

\dot{Q} -heat generation [W. m⁻³]
 Q_p -perfusion heat due to blood flow [W.m⁻³]
 Q_m -metabolic heat generation [W.m⁻³]
 R -iso-dose constant [-]
 rf - reflectivity [-]
 T -temperature [° C]
 t -time[s]
 w - water content [-]
 w_o -waist radius [m]
 Greeks symbol:
 ρ - density [kg . m⁻³]
 μ_a -tissue absorption coefficient[m⁻¹]
 $\mu_{a,w}$ -water absorption coefficient[m⁻¹]
 μ_s - scattering coefficient[m⁻¹]
 σ - Stefan Boltzmann constant
 \mathcal{E} -emissivity

Subscript:

co-convection
 r-radiation
 s-surface
 ∞ -environment

References

[1] Niemz, M. H., Laser-Tissue Interactions. Berlin, Germany, Springer-Verlag Berlin Heidelberg, 2007
 [2] Anderson RR, Parrish JA, Selective photothermolysis :precise microsurgery by selective absorption of pulsed radiation. Science 220.(1983),4596 ,pp.524-527

[3] T.G. Van Leeuwen, E.D. Jansen, M. Motamedi, C. Borst, A.J. Welch, Physics of laser induced hyperthermia, in: A.J. Welch, M.J.C. van Germet (Eds.), Optical-thermal Response of Laser-irradiated Tissue, Plenum, New York, 1995
 [4] W. J. Minkowycz, E. M. Sparrow, and J. Y. Murthy., Handbook of Numerical Heat Transfer, 2nd Edition. John Willey & Sons, Inc., New York, 2006
 [5] M. J. P. Brugmans (et.al), Temperature response of biological materials to pulsed non-ablative CO2 laser irradiation, Laser Surg.Med 11(1991), 6, pp. 587–594
 [6] J.T.Walsh,T,F.Deutsch , Pulsed CO₂ Laser Ablation of Tissue: Effect of Mechanical Properties, IEEE Transactions on Biomedical Engineering , 36(1989), 12,pp.1195-1201.
 [7] Megan Jaunich (et.al) ,Bio-heat transfer analysis during short pulse laser irradiation of tissues .International Journal of Heat and Mass Transfer 51(2008),23-24 , pp. 5511–5521
 [8] Faye C. Mcquistion (et.al), Heating, Ventilating, and Air conditioning. 5th edition .John Wiley & Sons, 2000
 [9] G.L.LeCarpentier (et.al), Continuous wave laser ablation of tissue: analysis of thermal and mechanical. Events IEEE transactions on biomedical engineering, 40(1993),2 ,pp.188-200
 [10] Pradeep Gopalakrishnan, Influence of Laser Parameters on Selective Retinal Photocoagulation for Macular Diseases. Ph.D. thesis, University of CINCINNATI, 2005

- [11] Bo chen, Experimental and modeling study of thermal response of skin and cornea wavelengths laser irradiation. Ph.D thesis , University of Texas at Austin, 2007
- [12] R. W. Lewis, K. Morgan, H. R. Thomas, K. N. Seetharamu, The Finite Element Method In Heat transfer Analysis. Chi Chester, U.K ,1996
- [13] Sapareto SA, Dewey WC., Thermal dose determination in cancer therapy. Int. J. Radiat. Oncol. Biol. Phys. 10(1984),6 , pp.787-800
- [14] W. C. Dewey, L. E. Hopwood, S. A. Sapareto, L. E. Gerweck, Cellular responses to combinations of hyperthermia and radiation, Radiology, 123(1977), 3, pp. 463-474
- [15] Damianou CA, Hynynen K, Fan X. Evaluation of accuracy of a theoretical model for predicting the necroses tissue volume during focused ultrasound surgery. IEEE Trans. Ultrason. Ferroelect. Freq. Contr., 42(1995),2 ,pp.182-187
- [16] C. Damianou and K. Hynynen, The effect of various physical parameters on the size and shape of necrosed tissue volume during ultrasound surgery. J. Acoust Soc. Amer, 95(1994), 3, pp.1641-1649
- [17] K. S. Shibib, Thermal damage due to incidental continuous Co₂ laser Irradiation on human skin. thermal science 14(2010)2, pp. 451-458.
- [18] C. Framme, G. Schuele, J. Roeder, R. Birngruber and Brinkmann, Influence of pulse duration and pulse number in selective RPE treatment , Laser in Surgery and medicine ,34(2004),3, pp.206-215

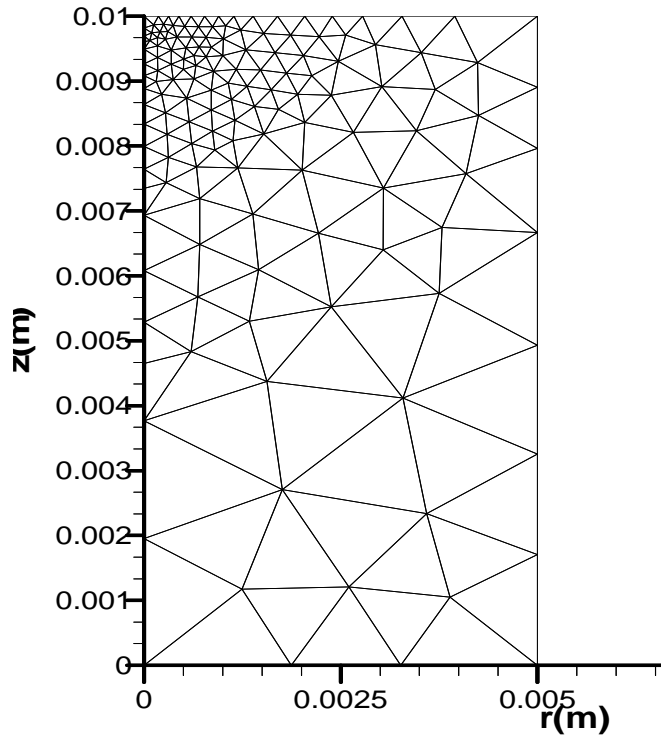


Figure (1) Selected domain and the used mesh

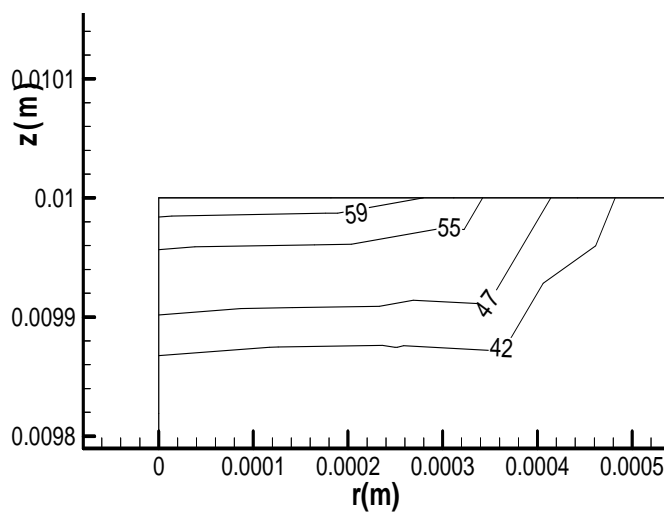


Figure (2) Enlarge view of temperature distribution at the cut-off time, $100 \mu s$ pulse width and $Hz=250$, top hat beam profile, energy/pulse is $0.2mJ$.

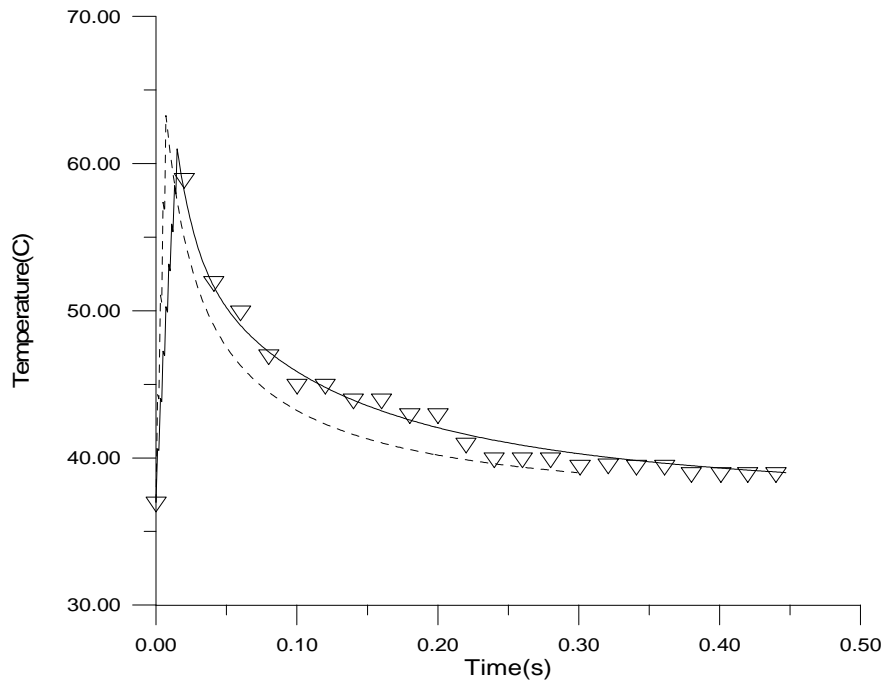


Figure (3) Temperature history at the center of spot size, the dashed line for Gaussian and solid for Top-hat beam. Both have 0.2mJ, 1ms pulse width and 500Hz, the triangle indicates experimental data for top-hat beam with camera resolution of 50 Hz (i.e. time of record is 0.02 s)

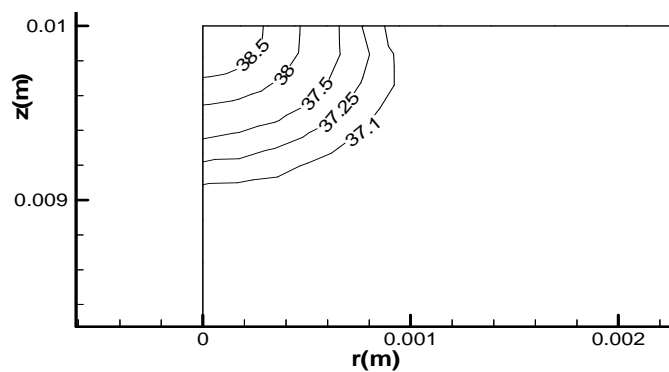


Figure (4) Enlarge view of temperature distribution at the end of cell necroses ,the distribution is at temperature just below 39° C for 1 ms pulse width and Hz=250,top hat beam profile, energy/pulse is 0.2mJ

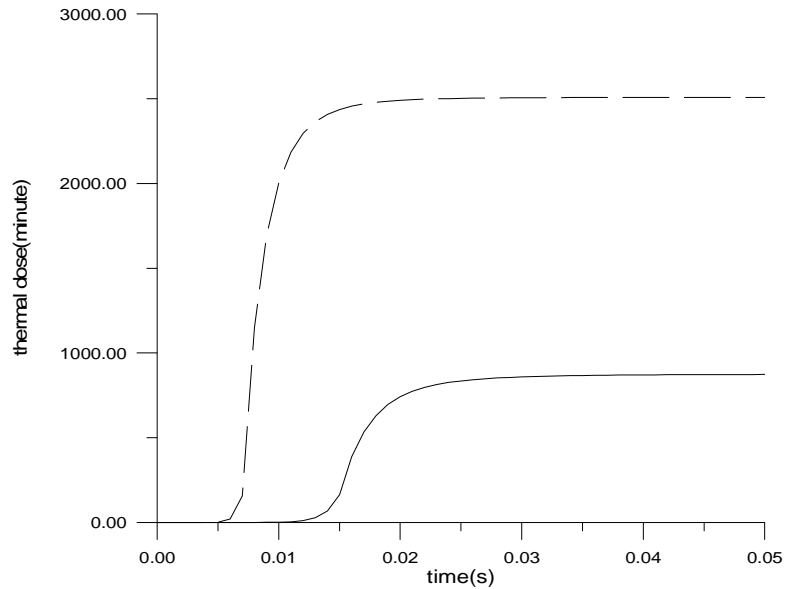


Figure (5) Thermal dose at center of spot size, dashed line indicates Gaussian laser beam, bold solid line indicates top hat laser beam. Both have same repetition rate (500Hz) energy /pulse (0.2mJ), pulse width 1ms.

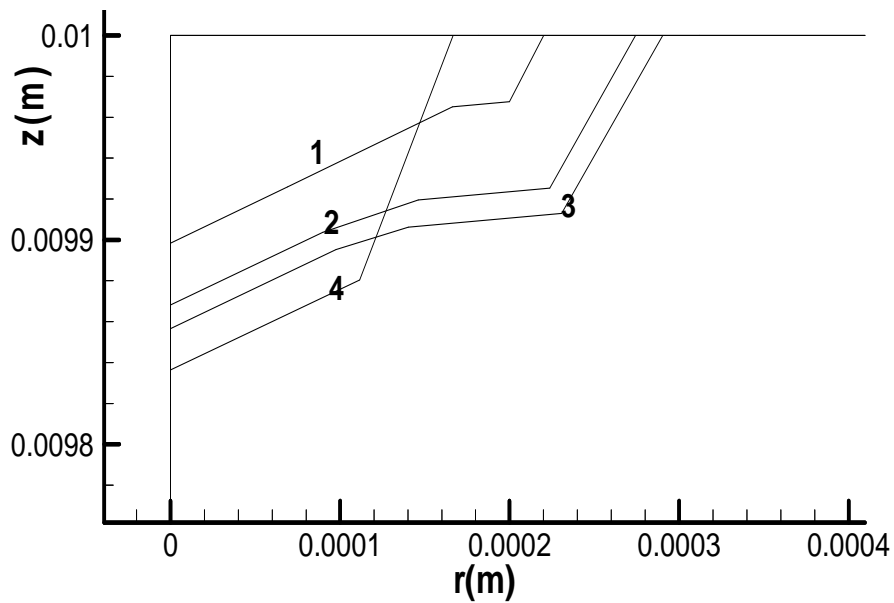


Figure (6) Enlarge view of thermal damage zone, Line1- pulse width of 1 ms, Hz=250, Top hat beam, Line 2- pulse width 1 ms, Hz=500, Top hat beam, Line 3- pulse width of 100 μ s , Hz=250, Top hat beam, Line 4-pulse width of 1ms, Hz=500, Gaussian beam

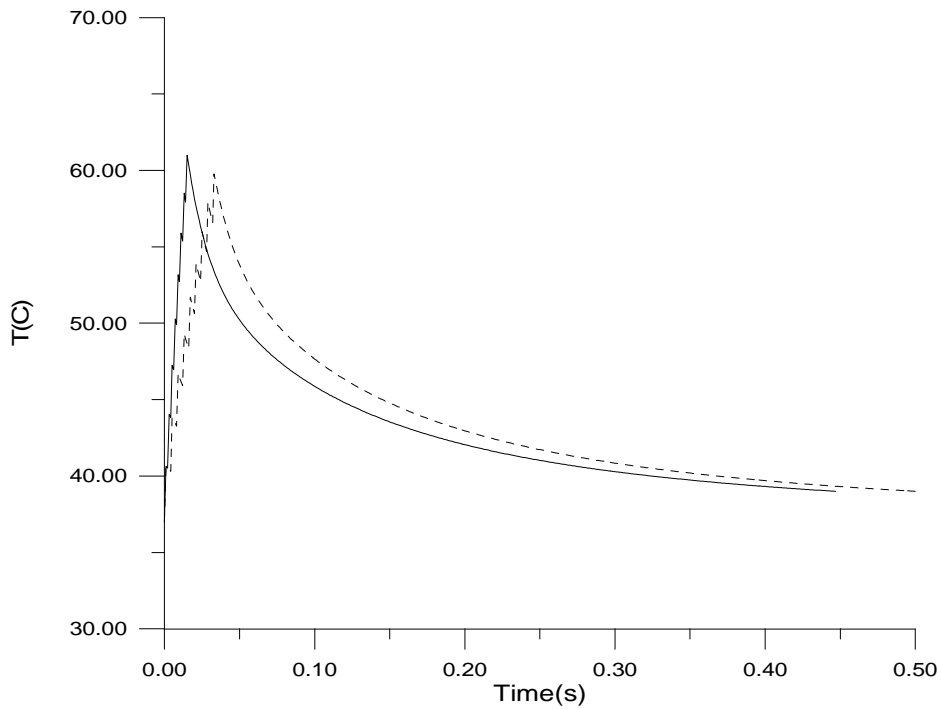


Figure (7) Temperature history at the center spot size of a tissue .The solid line for repetition rate of 500Hz and the dot line for 250 Hz, both have 0.2mJ, 1ms pulse width, top hat beam profile.

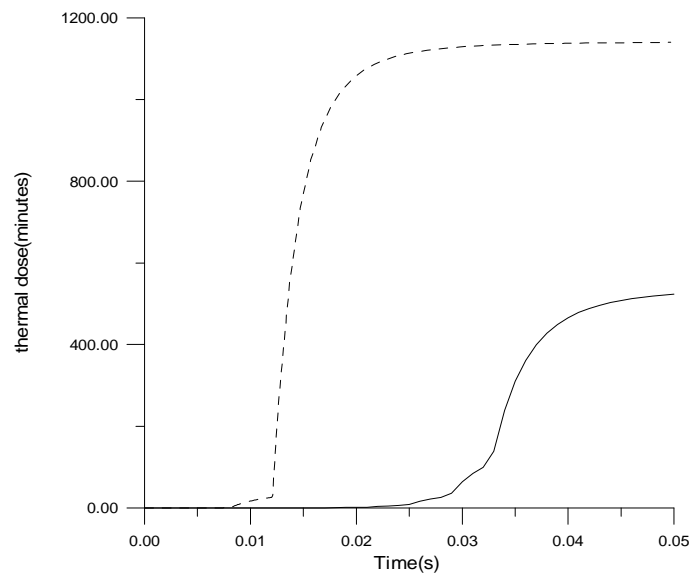


Figure (8) Thermal dose at center of spot size, dashed line indicates pulse width of 100 μs , bold solid line indicates pulse width of 1 ms. both have same repetition rate (250Hz) ,energy/pulse (0.2mJ) ,Top-hat beam.

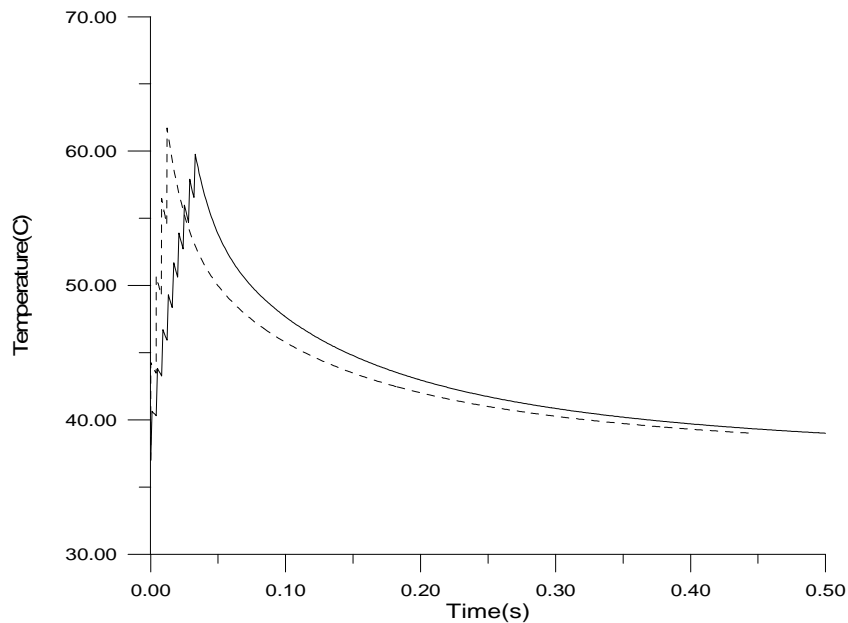


Figure (9) Temperature history for two pulse width at the origin of laser beam hitting a tissue dashed line indicates pulse width of $100 \mu s$, bold solid line indicates pulse width of 1 ms. assume same frequency(250Hz) and energy/pulse(0.2mJ),top-hat beam profile.

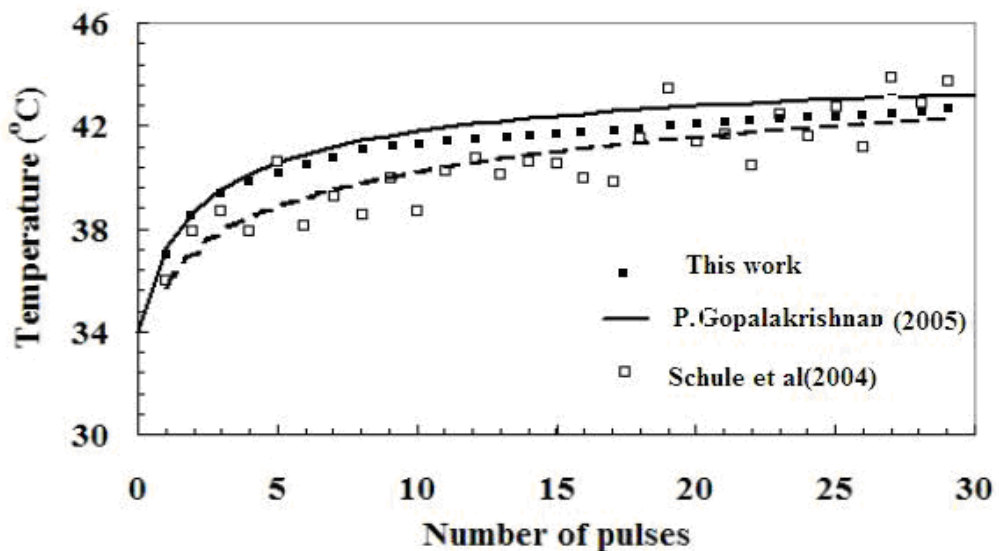


Figure (10) Comparison of baseline temperature with experimental result by references [10,17] the laser parameter are :Wave length –green, profile top-hat, pulse width- $1.7 \mu s$,energy/pulse- $100 \mu J$, repetition rate-100Hz, number of pulses-30,spot diameter- $160 \mu m$