Al-Nahrain Journal for Engineering Sciences (NJES) Vol.20 No.4, 2017 pp.894-903

Studying of Laser Tissue Interaction Using Biomedical Tissue

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Abstract:
In our work, three internal regions of rat are exposed to four different lasers with different power density, and then studying the histology of the tissues. Together the total absorption and transmission of the tissues at certain wavelength were determined.
Changing the wavelength across the absorption peak caused a significant difference in laser tissue interactions and changing the absorption coefficient, relaxation time, generated heat, and the intensity as a function of penetration depth. Furthermore, little mechanical damage could be seen in conventional histology**.**

Keyword: laser tissue interaction, many laser wavelengths, internal organs of rat.

Introduction:

Optical stimulation of the excitable tissues is a new therapeutic method in the medical field where the excitable tissue is stimulated by light-tissue interaction. Optical stimulation can be achieved by using several methods such as caged neurotransmitters, optogenetics, or lasers [1].
Interactions between light and biological tissue are essential and occur all the time. Life, Plants, algae and bacteria need sunlight to photosynthesise sugars such as glucose. Humans depend on photochemical reactions, e.g. Exposure to ultraviolet light is vital for the production of vitamin D in the skin. In addition light can also have undesirable effects such as causing cancers, e.g. UV damage to the DNA in melanocytes can lead to melanoma, a type of skin cancer. [2]

Lasers can be used significantly in medical diagnostics and treatment. At high power, these lasers are usually used as a scalpel [2] in surgery, especially those which have radiation wavelengths in the near IR and in far UV spectral regions because tissue has high content of water and collagen. The absorption peak of water in the near IR is around (3000nm) while for collagen in far UV is around (190nm) [3]. Biological tissue is turbid, with size of structure varies from tenths nanometers to hundreds micrometers [4].
Inhomogeneous optical medium have refractive index higher than the air about (1.4). Therefore part of light is rejected on the air/tissue interface. Turbid medium is described as medium with absorbing and scattering properties. Photons in such medium move in all directions and may be scattered or absorbed.
Tissue is also characterized by huge diversity and structural complexity.
When a laser beam incident on a tissue, then some of radiation is reflected at the tissue surface while the remaining part of laser radiation penetrates through the surface of the tissue and propagates inside it. Some of the propagated radiation is absorbed by successive tissue layers which it interacts with the tissue and do the required effect within it, while the other part of propagated radiation is scattered by the tissue so it will be a little biological impact [4].
The penetration depth of laser radiation in tissue is the most important property that determines its suitability for surgical procedure. The penetration depth is inversely proportional to the absorption coefficient of laser radiation in the tissue. With respect to exponential law of absorption, the intensity of an incident laser beam is attenuated in propagation through the tissue. This law describes the effect of the medium thickness with the absorption coefficient, and is commonly called as "Beer-Lambert's law" where expressed as [5]:

$$I(x) = I_0 \exp(-\alpha x) \ldots\ldots\ldots\ldots (1)$$

Where: $I(x)$: the intensity at distance (x), I_0: the incident intensity (assuming no reflection), α: absorption coefficient, x: the propagation distance along the optical axis, while the absorption length of the tissue is given as:

$$L = 1/\alpha \ldots\ldots\ldots\ldots (2)$$

The penetration depth is proportional to the wavelength of the laser radiation [6].
When laser beam incident on biological tissues several interaction mechanisms can be created related to the laser power density and its exposure time. There are five types of interactions may produce plasma-included ablation, photo-disruption, photo-ablation, thermal interaction, and photochemical interaction). There are another interactions may involve photomechanical phenomena in the same temperature range [2, 7]. The thermal interaction regime is the most importance for medical application.

The laser tissue interactions are characterized according to the tissue parameters and laser parameters as coefficients of absorption, reflection and scattering. The absorbed portion of the laser radiation can produce photochemical or photo-thermal effects depending on the wavelength of the laser radiation and the nature of tissue. Under certain conditions it can produce fluorescence from the tissue. The absorption of light in biomedical tissue is mainly happened by macromolecules such as proteins and pigments or water. The proteins and pigments can absorb the ultraviolet and the visible ranges of the spectrum while the water molecules can absorb the infrared region of the spectrum [5].
The ultraviolet lasers generally produce both the photo-thermal and photomechanical effects, while the infrared and visible laser can produce thermal effects only [2, 7].

When a laser beam interact with the tissue, a part of laser energy with high frequency electromagnetic waves can changes into thermal energy with low frequency electromagnetic waves and the heat generated in the tissue is directly related with the laser propagation in it.

As mentioned earlier, the part of the penetrated beam can attenuated in two different ways; as absorption and scattering which are characterized for absorption coefficient (α_a) and scattering coefficient (α_b) respectively. The absorption coefficient and the scattering coefficient depend on the tissue properties and on the laser wavelength, which represent the rate of energy loss per unit penetration length due to absorption and the photons scattering respectively.

The generated heat in the tissue is equal to absorbed energy in it and can be expressed as:

$$h(z) = \frac{I(z) - I(z+\Delta z)}{\Delta z} = -\frac{\partial I(z)}{\partial z} = \alpha_a . I(z) \quad \ldots\ldots\ldots\ldots (3)$$

where *h(z):* is the generated heat per unit area and per unit time, in a very small thickness Δz [5, 8].
In the most cases, the light is absorbed and scattered into the tissue simultaneously, which the attenuation coefficient of the Beer's law can be described as the sum of the absorption and scattering coefficients and is called the total attenuation coefficient, which can be given as [9,10,11]:

$$\alpha_t = \alpha_a + \alpha_s \ldots\ldots\ldots (4)$$

The following analyses are basically intended for vaporization phenomenon of soft tissues and the related thermal effects using highly absorbed laser radiation. The heat propagation in the tissue after an exposure time can be calculated by the heat conduction equation in a material medium [12, 13]:

$$\frac{\partial T}{\partial t} = \frac{k}{\rho . c}\nabla^2 T + \frac{h}{\rho . c} \ldots\ldots\ldots (5)$$

While the diffuse gain can be given by:

$$g = -kAdt\frac{d^2T}{dz^2}dz \ldots\ldots\ldots (6)$$

where T is the temperature, t is the time, k is the thermal conductivity, ρ is the tissue density, c is the specific heat, A is the cross sectional area of the irradiated tissue.
The variation of temperature can be given as:

$$\Delta T = constant \times exp\left(-\frac{uz}{\mu}\right) \ldots\ldots\ldots\ldots (7)$$

and then for biological tissue, the temperature is calculated by:

$$T = 37 + 63\, exp\left(-\frac{uz}{\mu}\right) \ldots\ldots\ldots (8)$$

In order to calculate how the heat propagation in a soft tissue after an exposure time, when the laser beam is off, it is only necessary to be the overall heat balance equal to zero.
The thermal penetration length which describes the propagation extension per time and it is related to heat propagation in the tissue can be given by:

$$z_{th}(t) = \sqrt{4\mu t} \ldots\ldots\ldots (9)$$

where μ is the tissue thermal diffusivity and it is given by:

$$\mu = k/\rho c \ldots\ldots\ldots\ldots (10)$$

Taking into account the temperature threshold *T* for Hyperthermia and coagulation, the depth of effect (damage) for each of them, ahead of the vaporization front can be estimated as "equilibrium damage depth",

$$z_{eq} = -\frac{\mu}{u}\ln\left(\frac{T-37}{63}\right) \ldots\ldots\ldots\ldots (11)$$

where *u* is the propagation speed of the vaporization front, which can be estimated from the overall heat balance as:

$$u = \frac{H}{\rho(c\Delta T + L_v)} \ldots\ldots\ldots\ldots (12)$$

where, ΔT = 63, and L_V is the latent heat of vaporization of water[13,14].
The time that necessary to heat propagates from the tissue that irradiated until the optical penetration length is the relaxation time. This time is obtained when the optical penetration length (absorption length) equal thermal penetration length, then thermal relaxation time is given by [15]:

$$t_{th} = \frac{1}{4\mu\alpha_a^2} \ldots\ldots\ldots\ldots (13)$$

For pulsed laser processing, where the pulse duration (t_p) is less than the equilibrium time (t_{eg}) which corresponds to equilibrium depth of damage,

the temperature and the equilibrium time follows that [16]:

$$T = 37 + 63erfc\left(\frac{z}{2\sqrt{\mu t}}\right) \text{.....(14)}$$

$$t_{eq} = \frac{\mu\rho^2(c\Delta T + L_v)^2}{1.64H^2} \quad \text{.....(15)}$$

Increasing the body temperature leads to several effects such as hyperthermia, coagulation and other irreversible tissue effects. By increasing the temperature some molecular bonds are destroyed and the membrane is altered. The reduction in enzyme activity is observed. For temperatures around 60°C, denaturation of proteins and collagen occurs which leads to the coagulation of tissue and it can necrotize cells. At higher temperature the equilibrium of chemical concentration is destroyed as the permeability of membrane of cells increases. The vaporization of water occurs at 100°C. Within the vaporization phase, the temperature of tissue does not alter and gas bubbles are formed. The propagation of these bubbles accompanied with the alteration of their volume causes thermal decomposition of tissue fragments. If all water molecules are vaporized, carbon atoms are released and the adjacent tissues are blackened and smoke rises from the skin. This stage is called carbonization. Finally beyond 300°C melting might occur. [17]

Method:

At our work, three entrails of rat have been exposure to different wavelength of different lasers and the absorbance of their entrails will examined before and after laser exposure. The histology for these entrails has been examined.

Specific Heat Capacity (C = 4350 J/kg.K), Latent Heat of Vaporization (Lv = 2,260,000 J/kg), Thermal Conductivity (k = 0.556 W/m.K), Density (ρ = 1000 kg/m³) and Thermal Diffusivity (α = 1.324*10^{-7} m²/s).

Histology:

Liver:

Figure (1) shows the absorbance spectrum and the histology for the liver before explosion with laser. Figure (1-A) shows the absorbance of the used wavelength which indicate to a small value of absorbance with high transmission reached to (99.84%) at 405nm while the transmission at 532nm is of limit (97%). In wavelength of (785, 1060nm) the absorbance can be larger than the previous wavelength with transmission limit of (89% and 93%) for 785 and 1060nm respectively.

Figure (1-B) show the normal structure of liver tissue.

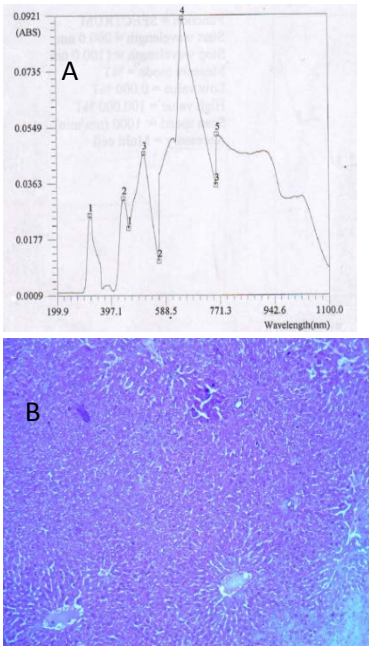


Figure (1): A) the absorption spectrum of liver B)normal structure of liver tissue.

The following figure show the tissue of liver when exposed to different laser wavelength with different power density. Figure (2-A) show the structure of liver after exposed to diode laser of 405nm with power density of (1.949W/cm^2) .The section of liver showing normal structure apperance of central vein and sheet of hepatocyte cells. Figure (2-B) show the structure of liver after exposed to SHG Nd:YAG of (532nm) of power density of (198.94 x10^{2}W/cm^2) which the section showing local discrete necresis of parenchyal tissue with inflantary cell infiltral. Figure (2-C) shown the tissue after exposed to laser diode of (785nm) with power density of (0.00497W/cm^2) . The section shows look like normal structure with local necresis and inflantary cells. When the tissue was exposed to CO_2 laser of (10.6μm) with power density of (19.894W/cm^2) the section shown look like normal structure also but with sinusoidal dilatation. (table (1))

Kidney:

Figure (3) shows the absorbance spectrum and the histology of scot-free kidney. Figure (3-A) shows the absorption spectrum of kidney: the maximum absorbance at (405, 532, 785 and 1060) nm correspond to (0.0756, 0.041, 0.068 and 0.048,) respectively.

As a result, the minimum transmission at 405 nm reach to 92% whiles the maximum transmission at 532nm approximately of (96%).

Figure (3-B) shows the normal structure of kidney.

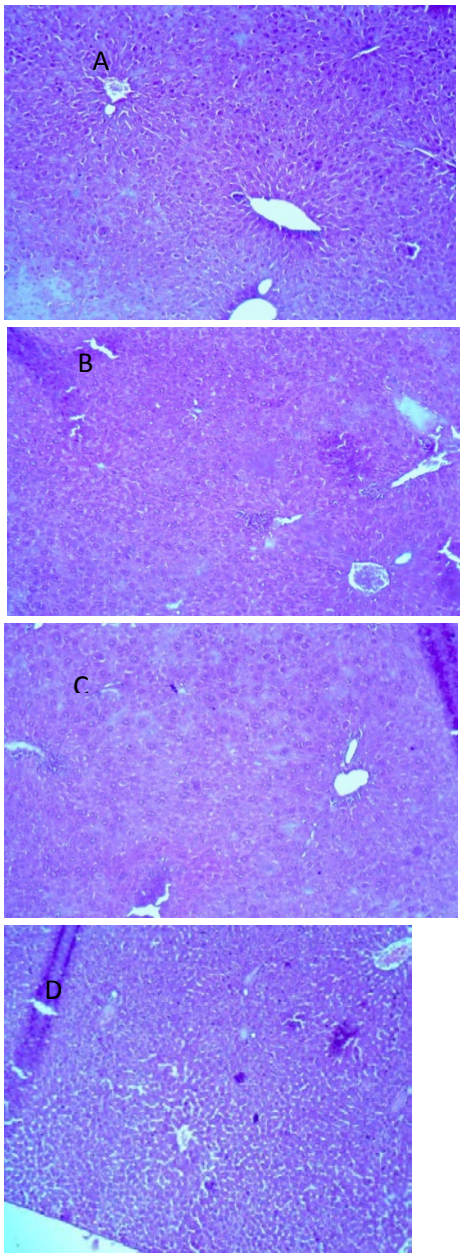


Figure (2): shows liver structure after exposed to different laser wavelengths

A: laser diode of (405nm), B: SHG:Nd:YAG of (532nm), C: laser diode of (785nm) and, D: CO_2 laser of (10.6µm).

Table (1): The changes in the liver tissue aftar exposed to laser with different wave length.

Wavelen-gth in nm	Power density	Transmit-ttance before	After (liver)
405	1.949 W/cm^2	99.84%	normal structure apperance of central vein and sheet of hepatocyte cells.
532	198.94 x10^2 W/cm^2	97%	local discrete necresis of parenchyal tissue with inflantary cell infiltral
785	0.00497 W/cm^2	89%	look like normal structure with local necresis and inflantary cells.
1060	19.894 W/cm^2	93%	look like normal structure also but with sinusoidal dilatation.

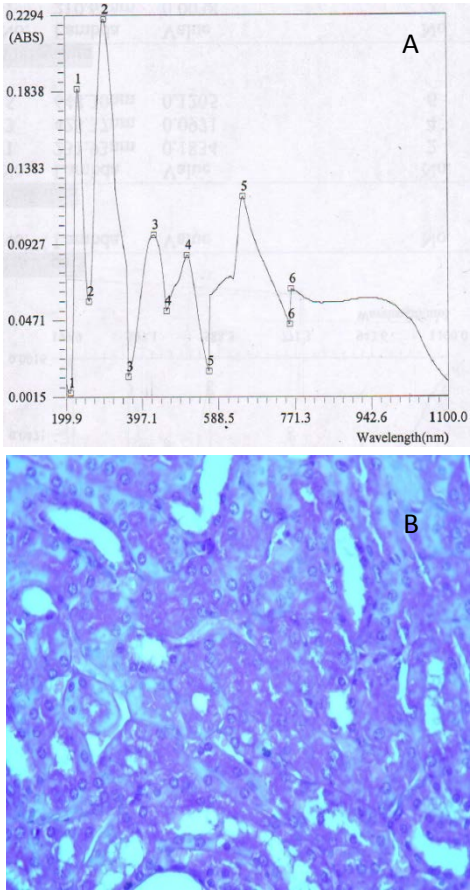


Figure (3): A) the absorption spectrum of kidney, B) normal structure of kidney tissue.

The following figures show the structure of kidney after being exposed to four different wavelength of laser with different power denstiy. figure (4-A) shows the structure of kidney after being exposed to laser diode of (405nm) . The

section shows normal structure apperance of renal tubules. Figure (4-B) and figure (4-C) show section of kidney after being exposed to SHG Nd:YAG laser at (532nm) with power density of (198.94 $x10^2 W/cm^2$) and laser diode of (785nm) of power density of ($0.00497 W/cm^2$) respectively. These sections show degenerative changes and necresis of renal tubules eqillibrium. Figure (4-D) shows section of kidney after being exposed to CO_2 laser of (10.6μm) . This section shows degererative and necresic of renal tubules.

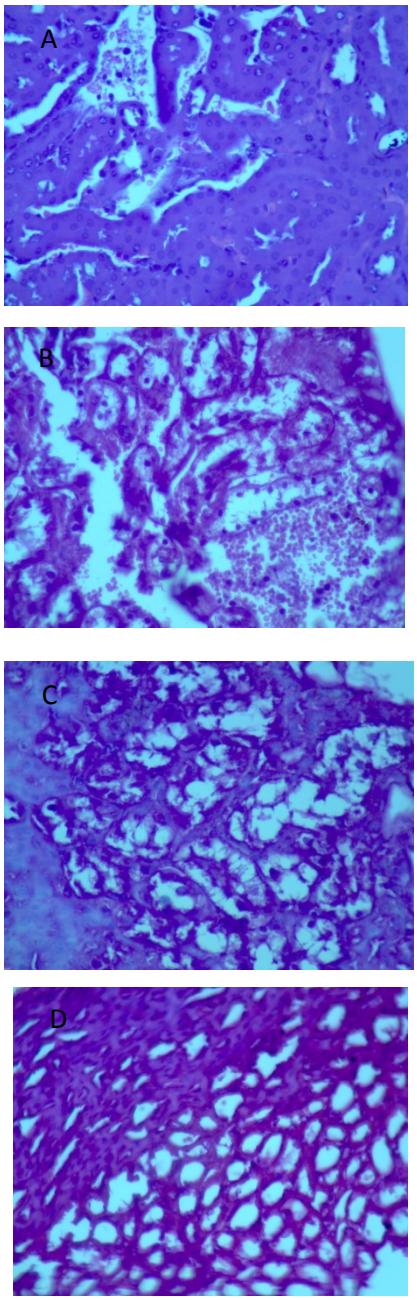


Figure (4): showing kidney structure after exposed to different wavelength of laser:

A: laser diode of (405nm), B: SHG:Nd:YAG of (532nm), C: laser diode of (785nm) and, D: CO_2 laser of (10.6μm).

Table (2): Kidney tissue when exposed to different laser wavelength with different power density.

λ in nm	Power density in watt/cm^2	transmittance before	After(kidney)
405	1.949	84%	normal structure apperance of renal tubules
532	198.94	91%	These sections show degenerative changes and necresis of renal tubules eqillibrium.
785	0.00497	86%	These sections show degenerative changes and necresis of renal tubules eqillibrium.
1060	19.894	90%	degererative and necresic of renal tubules.

Lung:

Figure (5) shows the absorbance spectrum (A) and histology of normal lung (B). The lung have a transmission approximately of (99.9%, 97%, 89% and 93%) at (405, 532, 785 and 1060) nm respectively.

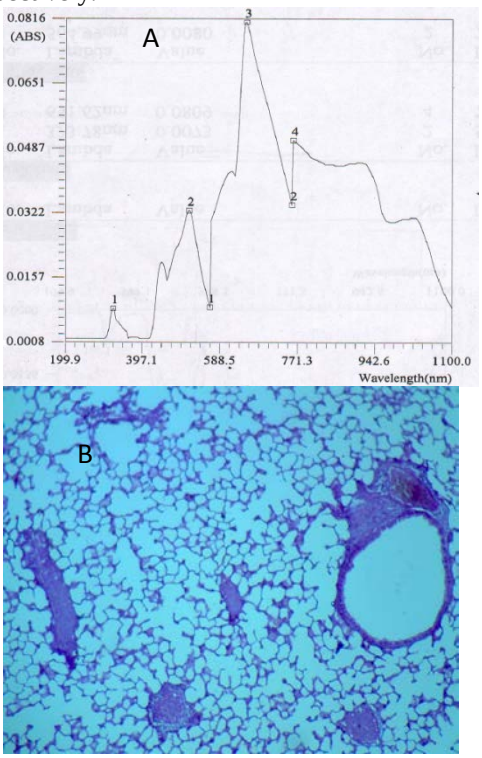
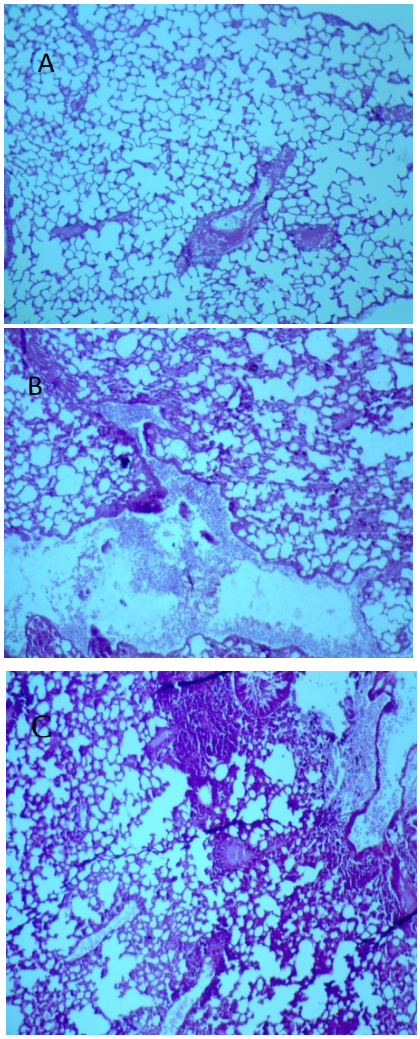


Figure (5): A) the absorption spectrum of lung, B) normal structure of lung tissue.

The following figures show the tissue strucure of lung after being exposed to different laser wavelength with different power density. Figure (6-A) shows the tissue after being exposed to (405nm). This section shows normal structure consisting alreali and alreals spaces with bronkinol. The section in figure (6-B) shows the tissue after being exposed to (532nm). This section of lung shows thickening of alrealar septae with presence bronchiolas secet. Figure (6-C) showing the struture of lung after exposed to (785nm). These sections show degenerative changes and necresis of renal tubules eqillibrium. Figure (6-D) showing the struture of lung after exposed to (1060nm) which this section showing thickening of alrealar septae with necresis and inflammatory cells infiltrt with fornat of emphyscma: Table (3).



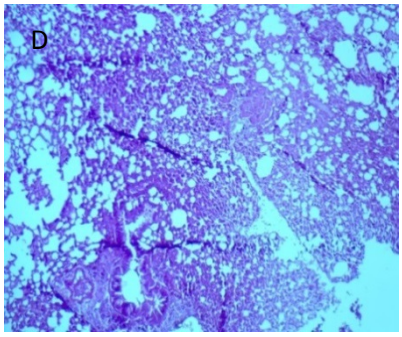


Figure (6): showing lung structure after exposed to different wavelength of laser.

A: laser diode of (405nm), B: SHG:Nd:YAG of (532nm), C: laser diode of (785nm) and, D: CO_2 laser of (10.6μm).

Table (3): The viration in the lung tissue when exposed to different laser wavelength with different power density

λ in nm	Power density	Transmi-ttance before	After(lung)
405	1.949 W/cm^2	99.9%	normal structure consisting alreali and alreals spaces with bronkinol.
532	198.94 x10^2 W/cm^2	97%	shows thickening of alrealar septae with presence bronchiolas secet
785	0.0049 7 W/cm^2	89%	showing thickening of alrealar septae with necresis and inflanatory cells infiltrt with fornat of emphyscma.
1060	19.894 W/cm^2	93%	

Absorption coefficient:

From the present work, it was observed for the three organs tissue: as the wavelength is increased the absorption coefficient was increased The organs which have the greatest affected with the increasing of the wavelength is the liver then kidney then the lung which has the smallest absorption coefficient, Figure (7) and table (4).

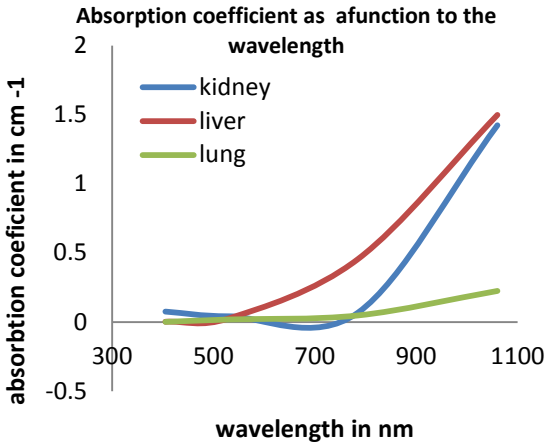


Figure (7): absorption length as a function of wavelength

Table (4): absorption coefficient for kidney, liver and lung at different wavelength of exposure

λ in nm	Absorption coefficient		
	Kidney	Lever	Lung
405	0.075597	0.0021	0.00096
532	0.041399	0.0282	0.0186
785	0.068101	0.455996	0.04771
1060	1.424501	1.49925	0.225023
t-test	0.00857	0.00856	0.00860

Absorption length:

It was observed for the three organs tissue: as the wavelength is increased the absorption length was decreased, Figure (8) and table (5). Statically by using t-test the organs which have the greatest affected with the increasing of the wavelength is the lung.

Table (5): absorption length for kidney, liver and lung at different wavelength of exposure

Wavelength in nm	absorption length in cm		
	Kidney	liver	Lung
405	13.228	476.19	1041.67
532	24.155	35.461	53.763
785	14.684	2.193	20.96
1060	0.702	0.667	4.444
t-test	0.05	0.172	0.009

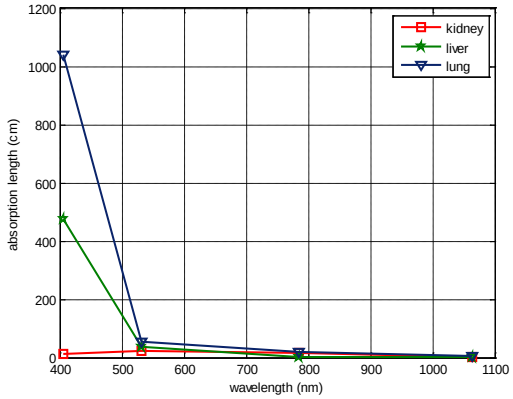


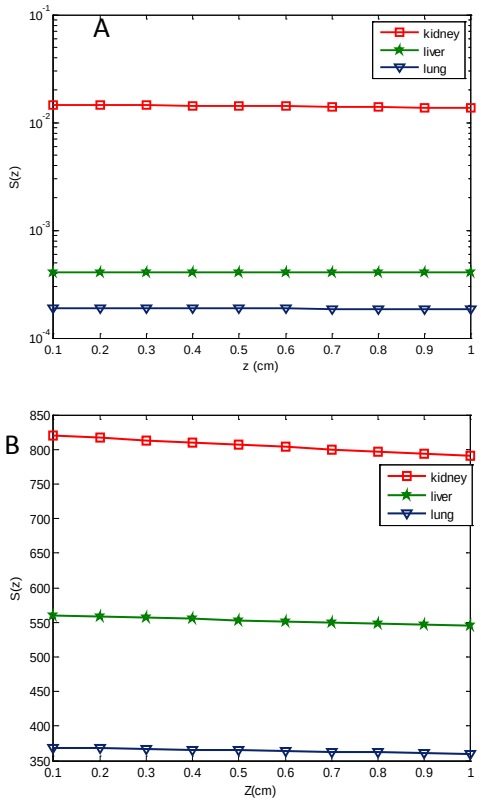
Figure (8): absorption length as a function of wavelength.

Our results have a good agreement with Almaty [1] for rat liver.

Generated heat:

For our work the thermal penetration =1.673mm (for 5s).

The generated heat as a function of penetration depth shown in figure (9). The smallest level of generated heat as a function of penetration depth is the lung for all explosion wavelengths.



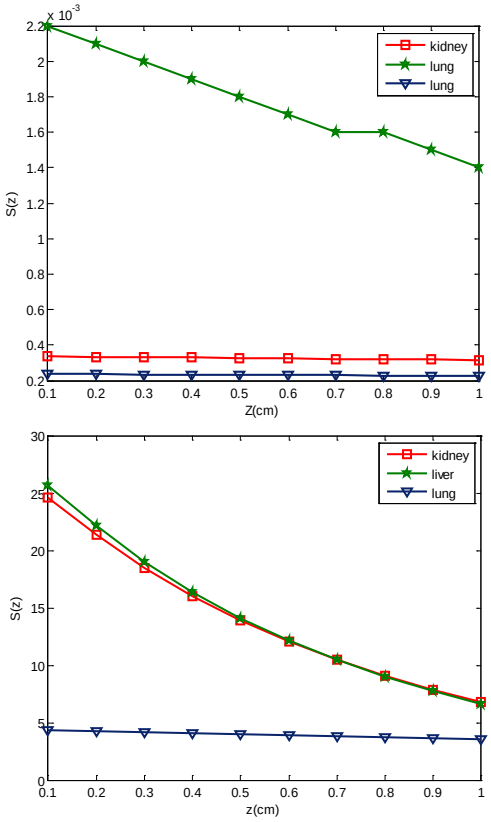


Figure (9): the relation of generated heat as a function of penetration depth; A: at 405, B: at 532, C: at 785, and D: at 1060.

Table (6): The relaxation time for the kidney liver and lung after exposure to laser of Different wavelength

λ in nm	Kidney	Liver	Lung
405	23.62 x10^6	11.76x10^6	1.86x10^6
532	43.133x10^6	63.323x10^6	96.006x10^6
785	26.22 x10^6	3.92x10^6	37.44x10^6
1060	1253x10^6	1.190xE10^6	7.936x10^6

Thermal relaxation time:

In our work, it can be observed that the maximum thermal relaxation time for the three organs appear at the range between (500-600) nm.

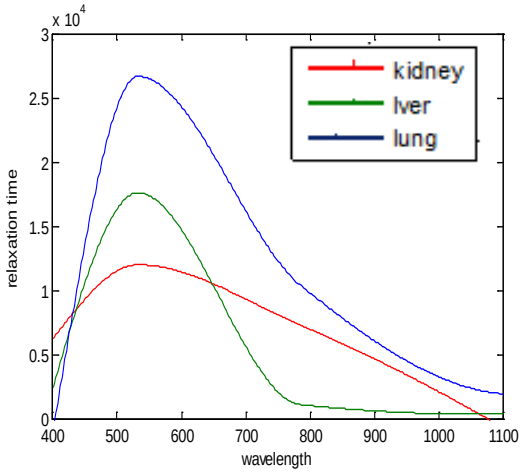


Figure (10): the thermal relaxation time as a function of wavelength.

Figure (11) shows that the ability of kidney and liver for absorption the three kind of laser versus the thermal relaxation time is the same approximately, while the lung tissue take a long time for relaxation due to the smallest level of generated heat as a function of penetration depth.

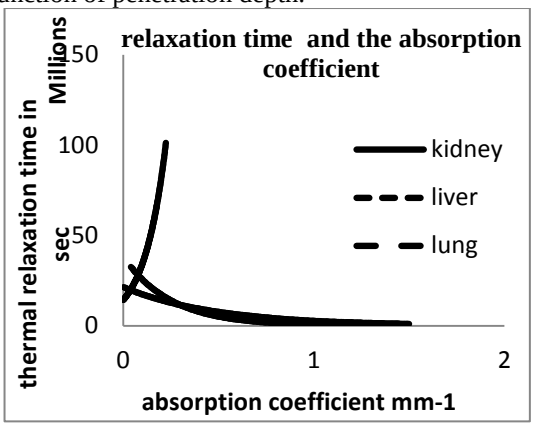


Figure (11): the relation between thermal relaxation time with the absorption coefficient for the three organs.

Power Intensity as a function of penetration depth I (z):

From figure (12) for the intensity as a function of penetration depth, it can notice that the most organs which effected with the wavelength change is the kidney then the liver while the lung has the highest intensity (the absorbance was the lowest).

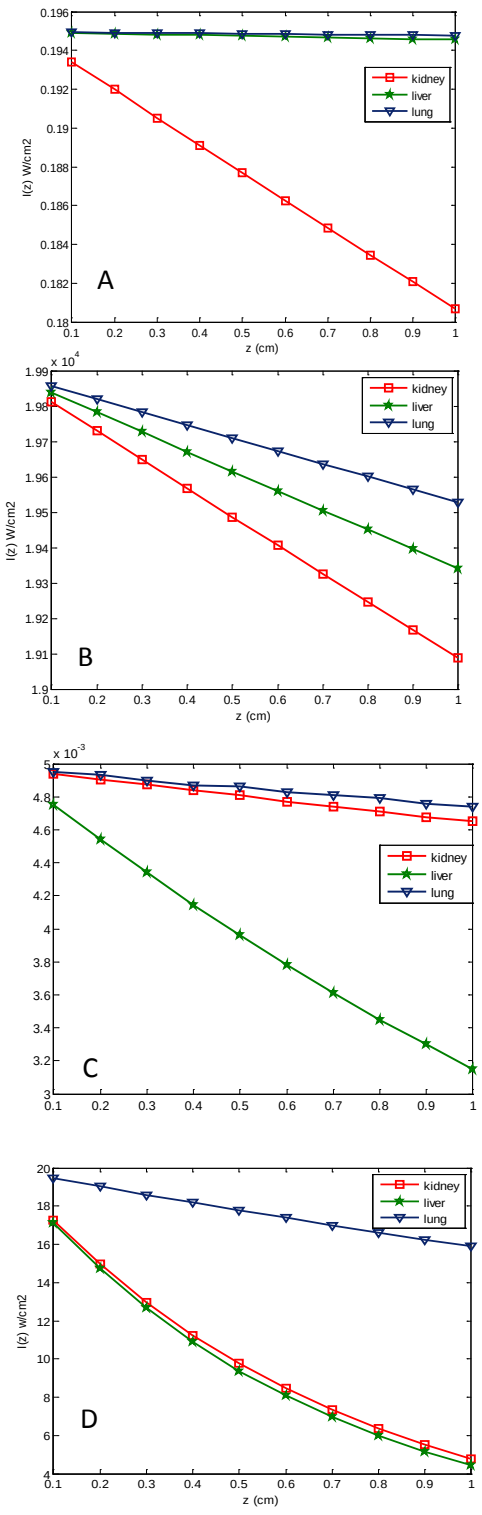


Figure (12): the relation of the intensity as a function of penetration depth; A: at 405, B: at 532, C: at 785, and D: at 1060

Conclusions:

Subjacted three internal region of rat to four different lasers with different power density, caused several effects such as hyperthermia, coagulation degererative and necresic and others. the structure of the liver tisuue showing local discrete necresis of parenchyal tissue with inflantary cell infiltral,. The kidney show degenerative changes and necresis of renal tubules eqillibrium, degererative and necresic of renal tubules.The lung thickening of alrealar septae with presence bronchiolas secet and thickening of alrealar septae with, necresis and inflammatory cells infiltrt with fornat of emphyscma. The main reason for these changes is high temperature caused denaturation of proteins and collagen occurs which leads to the coagulation of tissue and it can necrotize cells.

For the absorption it was observed for the three organs tissue: as the wavelength is increased the absorption length was decreased (the absorption coefficient was increased), and the organs which have the greatest affected with the wavelength variation is the kidney it has the smallest absorption length (long absorption coefficient), while the lung, has the longest absorption length (absorption coefficient is the less), and this result has a good agreement with [1] especially with rat liver.

The smallest level of generated heat as a function of penetration depth is the lung for all explosion wavelengths.

The ability of kidney and liver for absorption the three kind of laser versus the thermal relaxation time is the same approximately, while the lung tissue takes a long time for relaxation due to the smallest level of generated heat as a function of penetration depth.

For the intensity as a function of penetration depth it can notice that the most organs which affected with the wavelength change is the kidney then the liver while the lung has the highest intensity (the absorbance was the lowest).

References:

[1] Mohamed A. A., "Optical Properties Measurements of Rat Muscleand Myocardium at 980 and 1860 nm Using Single Integrating Sphere Technique", University of Miami Scholarly Repository, 2014). *Open Access Theses.*

[2] Martin B. H. ,"Measurement of tissue optical properties", MSc. thesis, 9th May 2013.

[3] Fodor L. et al., "Light laser interaction", *Aesthetic Applications of Intense Pulsed Light*, Springer-Verlag London Limited, 2011.

[4] Raulin C. and Karsai S. (eds.), "Laser-tissue Interactions, *Laser and IPL Technology in Dermatology and Aesthetic Medicine"*, Rudolf Steiner, Springer-Verlag Berlin Heidelberg, 2011.

[5] Daniel J. E., "Optical sensing for tissue differentiation", 18 November 1978.

[6] Sharon T., "Pathologic Analysis Of Photothermal And Photomechanical Effects Of Laser–Tissue Interactions, Photochemistry and Photobiology", Volume 53, Issue 6, Pages 825–835, June 1991.

[7] Mohit G., "Analyzing Thermal and Mechanical Effects of Pulsed Laser Irradiation on Tissues", MSc. in Biomedical Engineering, Florida Institute of Technology, 2014.

[8] Khalid Salem Shibib,. Ayad Zwayen Mohammed, Kholood Hasan Salih , "Pulse Laser Parameters Effect on Tissue Thermal Damage Zone in Coagulation Process"Eng.& Tech.journal,Vol 29,No,5, 2011.

[9] Steven L. J., "Laser-Tissue Interactions", Depts. of Biomedical Engineering and Dermatology Oregon Health & Science University, Student Workshop, Univ. of Oulu, Oulu Finland, June 5, 2010.

[10] Carlos D., Jan S., Laura R. and Susana M., "Suitability of Filofocon A and PMMA for experimental models in excimer laser ablation refractive surgery", OPTICS EXPRESS 20955, Vol. 16, No. 25, 8 December, 2008.

[11] Kyunghan K. and Zhixiong G., "Thermal Relaxation Times in Biological Tissues Subjected to Pulsed Laser Irradiation", 9th AIAA/ASME Joint Thermophysics and Heat Transfer Conference , San Francisco, California ,5 - 8 June 2006.

[12] Denise M. Z., Patricia A. A., Thiago M. P., Paulo R. C. and Walter V. Jr. "Heat Generation and Transfer on Biological Tissues Due to High-Intensity Laser Irradiation", Developments in Heat Transfer, Dr. Marco Aurelio Dos Santos Bernardes (Ed.), ISBN: 978-953-307-569-3, In Tech., 2011.

[13] Frank P. I., DAVID P. D., THEODORE L. B., ADRIENNE S. L., "Fundamental of heat and mass transfer", Copyright 2007 John Wiley & Sons,

[14] Bernard Ch. MS, PhD* and Ashley J. W. PhD, "Analysis of thermal relaxation during laser irradiation of tissue, Lasers in Surgery and Medicine" Volume 29, Issue 4, pages 351–359, November 2001.

[15] Markolf H. N., "Laser-Tissue Interactions: Fundamentals and Applications", second edition, Springer.

[16] Menzel R., "Photonics", Springer, Berlin, 2001.

[17] Mohammad A. A., Mohsen E., and Ezeddin M., "Mechanisms of Laser-Tissue Interaction: II. Tissue Thermal Properties", J Lasers Med Sci., Summer; 4(3): 99–106 2013..

دراسة تفاعل الليزر مع الأنسجة باستخدام النسيج الطبي الحيوي

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الخلاصة:

في هذا البحث ثلاث مناطق داخلية لفأرة تم تعريضها الى اربعة انواع من الليزر بكثافة طاقة مختلفة ودرست التغيرات الحاصلة بانسجتها. وتم تعين معاملات الامتصاص الكلي والنفاذية الكلية للانسجة لكل طول موجي مستخدم.

تغيير الطول الموجي خلال قمة الامتصاص سبب فروقات ذات اهمية في تفاعلات الليزر مع النسيج وتغير معامل الامتصاص، زمن الاسترخاء، الحرارة المتولدة، والشدة كدالة الى عمق النفوذ. وعلاوة على ذلك لوحظ تلف ميكانيكي صغير في الانسجة.