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Simulation of one Dimensional Photoacoustic Imaging

Mohammed A. Hussain and **Luma W. Jameel**

Department of Laser and Optoelectronics Engineering, Al-Nahrain University, Baghdad , Iraq

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Abstract: The present work provides theoretical investigation of laser photoacoustic one dimensional imaging to detect a blood vessel or tumor embedded within normal tissue. The key task in photoacoustic imaging is to have acoustic signal that help to determine the size and location of the target object inside normal tissue. The analytical simulation used a spherical wave model representing target object (blood vessel or tumor) inside normal tissue. A computer program in MATLAB environment has been written to realize this simulation. This model generates time resolved acoustic wave signal that include both expansion and contraction parts of the wave. The photoacoustic signal from the target object is simulated for a range of laser pulse duration 10ns-10µs emitted from Nd:YAG laser, depth of target object 0.3-3 cm, distance from the object to the detector 0.7-3 cm and the diameter of target object 0.1-0.6 cm. The diameter of the object computed by the simulation is always being 75% of its value. The amplitude of the signal is directly proportional with the laser pulse energy and inversely proportional with the depth of target object and the distance from the object to the detector. The PA signal is fully generated in Nano second laser pulse duration range as it is short enough to fulfill the stress confinement condition.

Introduction

One of the techniques for non-invasive as well as non-destructive imaging is a hybrid biomedical imaging which is called photoacoustic (PA) imaging. It is developed based on the PA effect, a phenomena in which the absorbed energy from the light is transformed into kinetic energy of the sample by energy exchange processes, then results in local heating and thus a pressure wave or sound[1]. Its advantages comparing with conventional imaging techniques based on x-rays, ultrasound, or magnetic resonance are the applications of nonionizing radiation, an enhanced contrast for many important tissue structures, and the relatively low cost (mainly compared to magnetic resonance imaging). It combines the advantages of pure optical imaging with those of ultrasound imaging [2].

Photoacoustic also known as optoacoustic and thermoacoustic, is obtained by the absorption of light energy, and the subsequent emission of an acoustic wave [3].

Laser technology has provided a useful short-pulsed and wavelength-tunable light source for photoacoustic imaging. At some wavelengths of light, the absorption coefficient of blood can be ten times higher than that of its surrounding tissue, which results in excellent intrinsic contrast for blood-vessel imaging. Photoacoustic imaging has been successfully applied to imaging vascular structures and tumor angiogenesis a few mm under the tissue[4].

The simulation is used to find the location and the size of target object under a normal tissue.

Theoretical Background

Photoacoustic imaging relies upon irradiating the tissue surface with low energy nanosecond pulses of visible, or more deeply penetrating near-infrared (NIR) laser light. Absorption of the light by subsurface anatomical features such as blood vessels leads to impulsive temperature rise, typically in milliKelvin range, accompanied by rapid thermoelastic expansion and the subsequent generation of broadband (tens of megahertz) ultrasonic wave[5]. An acoustic transducer then detects these waves, and time resolution of the resulting signal determines the one dimensional of the detected object[6].

The excited PA signal is locally determined by the EM absorption and scattering properties, the thermal properties, including the thermal diffusivity and thermal expansion coefficient, and the elastic properties of the sample.

The formation of photoacoustic waves can be described by the thermally generating wave equation for acoustic pressure [7], and as presented in Figure1;

$$\frac{\partial^2 P(r,t)}{\partial t^2} - {C_s}^2 \nabla^2 P(r,t) = \frac{\partial H}{\partial t} \quad (1)$$

where, H is a function defined as the heat deposited in the medium per unit volume per unit time.
C_s is speed of sound in tissue.
r is the target diameter.

The pressure acoustic wave amplitude can be represented by [7]

$$P(r,t) = \frac{\beta}{4\pi C_p \Delta r}\frac{\partial}{\partial t}\left[\frac{E_a}{\sqrt{\pi}\tau_e}\exp-\left[\frac{\left(t-\frac{\Delta r}{Cs}\right)}{\tau_e}\right]^2\right] ,$$
$$Ea = \alpha E_0 \quad (2)$$

E_a is absorbed laser energy, E_0 is incident laser energy, α is medium absorption coefficient.
τ_e is a single effective time scale, has been suggested, in which

$$\tau_e = \sqrt{(\tau_p^2 + \tau_a^2)} \quad (3)$$

where, τ_a acoustic transit time across the radius of acoustic source.
τ_p laser pulse duration. β is thermal expansion coefficient of the medium. C_p the specific heat capacity (J/kg.K) and Δr detection distance.

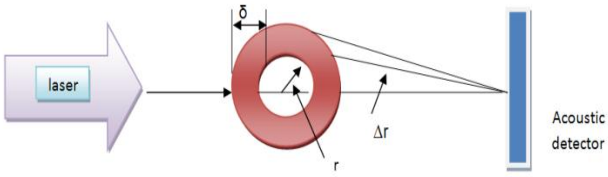

Fig. (1): Geometry for detecting Photoacoustic pressure wave from a uniformly absorbing sphere located a distance from an ultrasonic detector.

Hence, the acoustic pressure detected in the non-boundary case is equal to [7]

$$P(r,t) = \frac{\beta E_a}{4\pi^{3/2} C_p \tau_e \Delta r}\frac{\partial}{\partial t}\left[\exp-\left[\frac{\left(t-\frac{\Delta r}{Cs}\right)}{\tau_e}\right]^2\right] \quad (4)$$

The photoacoustic pressure spatial profile is dependent on the absorption coefficient of the target object [8].

The product of temporal width of laser induced stress transient multiplied by the speed of sound depicts the dimension of object under tissue in the direction of stress detection [9], the dimension of sample along the detector line is found by [4]

$$z = c_s t \quad (5)$$

where, z is target dimension; t is delay time between the laser pulse irradiation and the transient stress wave arrival to the detector.

The product of the speed of sound in tissue and the delay time between the moment of the laser pulsed irradiation and the moment of transient stress wave arrival to the detector, displays the depth of object location from the measurement surface [9].

Results and Discussion

This simulation is performed in order to obtain the sensitivity required by photoacoustic imaging system that will effectively detect the target object(a blood vessel or a tumor) of a certain size located at a certain depth using MATLAB program. This work simulates an object that has a spherical shape as shown in Figure 2, the tissue around the object is considered as a uniform turbid medium. The incident laser beam shape is considered as Gaussian, with different object radii and at different depths.

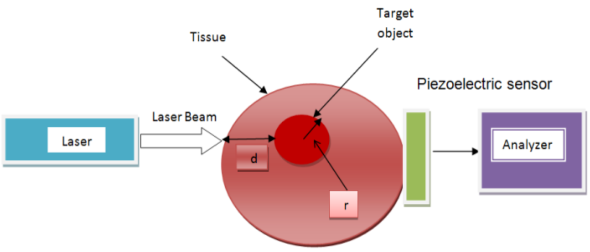

Fig. (2): Arrangement for detecting Photoacoustic pressure wave from a uniformly absorbing sphere located at a distance d from the laser source.

This simulation is based up on the spherical mathematical model equation 4. It is assumed that the distance from the surface of the tissue to the target object is d (depth of object), r is the radius of tumor, the sound velocity in tissue is 1500 m/s, and t is the time-of-flight from the center of target object to the transducer.

The laser pulse is considered between 5ns and 1µs. The incident laser beam of the infrared radiation of 1.064µm wavelength, laser energy 100mJ, the absorption coefficient of tissue for this wavelength is 18 m^{-1} while the absorption coefficient of target object which is the same of the blood because the tumor is having a large blood contents. Figure 3 shows the simulation results of the temporal profile of acoustic signal, assuming laser pulse duration 5 ns, laser pulse energy is 100 mJ, the depth of target object is taken at 1 cm, 0.3cm diameter of target tumor, and the distance from the target object to the detector is 0.7, 1.0, 2.0 and 3.0 cm respectively.

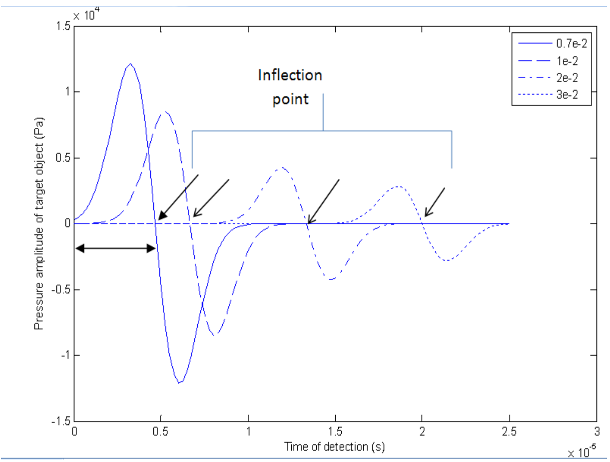

Fig. (3): Temporal prolife of acoustic signal at a distance from the target object to the detector of 0.7, 1.0, 2.0 and 3.0 cm.

It can be observed that each signal has a delay time for the PA wave to reach the acoustic detector. The location of target object can be measured by calculating the time from zero time to the inflection point of the curve and multiplied by the velocity of sound in tissue. Therefore, the delay time of each of the four signals presented in Figure 3 increases with the distance of the target from the detector.

The amplitude of acoustic signal is inversely proportional with distance between the target object and the detector. Figure 4 presents the simulation results of the temporal profile of acoustic signal, assuming pulse duration of 5ns, laser pulse energy is 100 mJ, the distance from the target object to the detector is 1 cm, diameter of target tumor is 0.4cm, and the depth of target object under normal tissue is 0.3, 0.8, 1.5 and 3.0 cm respectively.

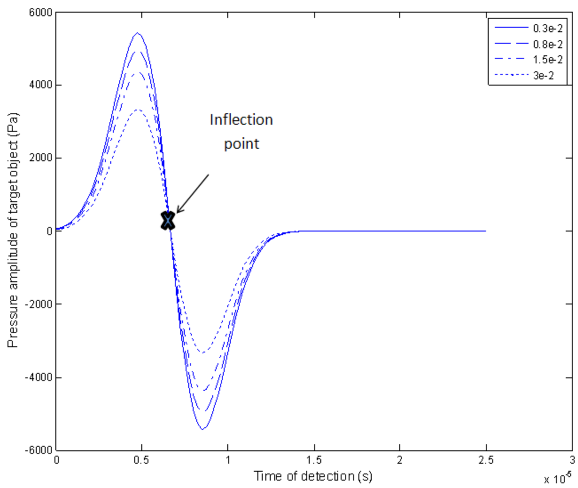

Fig. (4): Temporal prolife of acoustic signal at different depth of target object (0.3, 0.8, 1.5 and 3 cm) under normal tissue.

These results reflect identical findings with those observed in Figure 3where the amplitude of acoustic signal reduces exponentially according to Beer-Lambert Law because of the absorption of the surrounding normal tissue of laser pulse energy before the laser radiation reaches the target object.

The location of target object under the normal tissue can't be calculated from this results as only the amplitude of the signal is changed with the depth of target object, as this parameter has not been included within the spherical model analysis, i.e, Eq. (4). Figure 5 shows the simulation results of the temporal profile of acoustic signal at following assuming pulse duration of 5 ns, laser pulse energy is

100mJ, the depth of target tumor is 1 cm, the distance from the target tumor to detector is 1cm, and the diameter of the target tumor is 0.1, 0.2, 0.4 and 0.6 cm respectively.

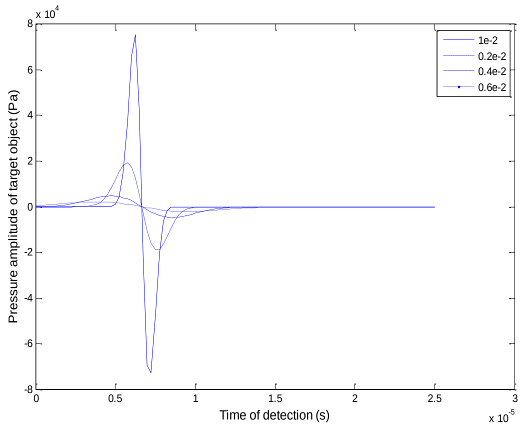

Fig. (5): temporal prolife of acoustic signal at different target object diameter (0.1, 0.2, 0.4, 0.6 cm).

These results indicate that the size of target tumor is measured by calculating the time between the maximum positive peak and the minimum negative peak then multiplied by the velocity of sound in tissue using Eq. (5).

The calculated target object diameter represents 75% of the object diameter which has been assumed for the simulation for all of the simulated results. This is due to the mathematics of the spherical source model which has been adopted in this work. Thus a correction factor (division by 0.75) can be simply introduced to these results in order to estimate more accurate target diameter. It can be observed also that the amplitude of the PA signal has decreased with the diameter of the target object. This is because of increased attenuation of the PA signal with increased volume of the target. Figures 6, 7, 8 and 9 show the simulation results of the temporal profile of acoustic signal at following assuming the depth of target tumor is 1 cm, the distance from the target tumor to detector is 1cm, and the diameter of the target object is 0.6 cm and laser pulse duration of 10 ns 0.1 µs, 1 µs, and 10 µs respectively as a function of detection time. These results showed that PA wave amplitude is not affected as long as the laser pulse duration is short enough for stress confinement condition, as in Figures 6 and 7. Figure 8 reveals slightly reduced amplitude for 1µs pulse duration. But for 10 µs (Figure 9) the acoustic signal is not fully generated as the stress confinement condition has not been met where the pulse duration is longer than stress confinement time.

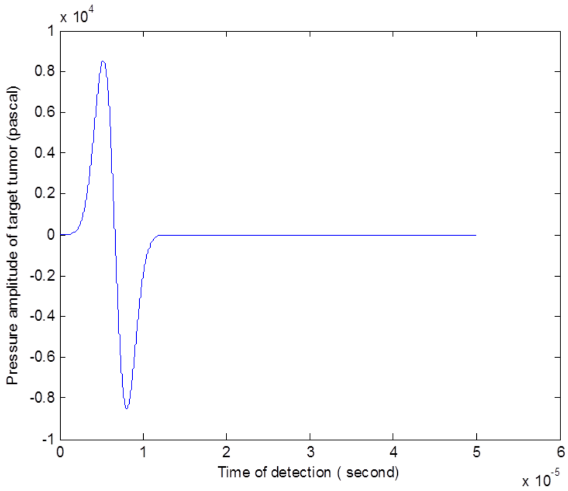

Fig. (6): Temporal profile of acoustic signal at 10ns laser pulse duration

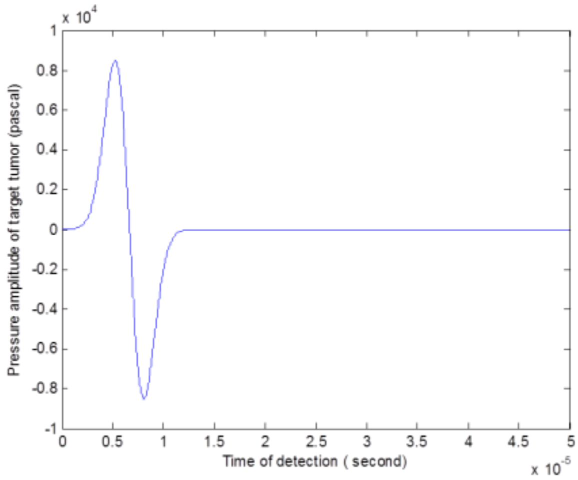

Fig. (7): Temporal profile of acoustic signal at 0.1µs laser pulse duration

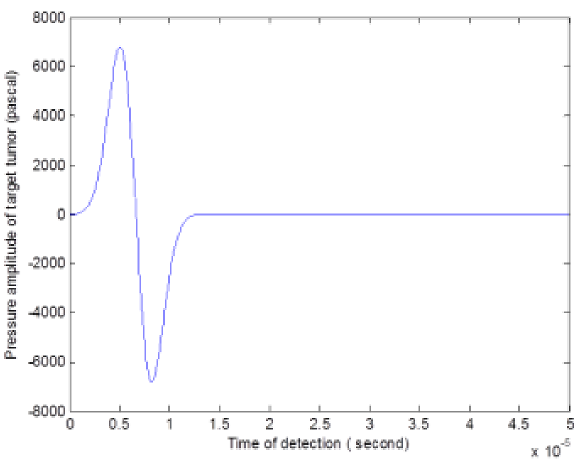

Fig. (8): Temporal profile of acoustic signal at 1µs laser pulse duration

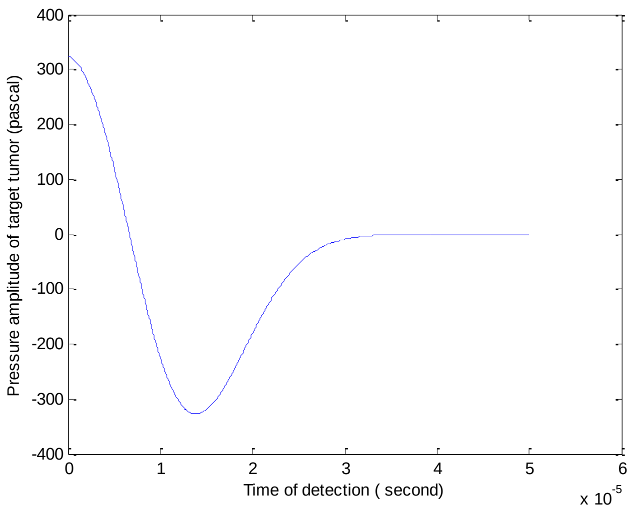

Fig. (9): Temporal profile of acoustic signal at 10μs laser pulse duration.

Conclusions

The basic principle of laser photoacoustic technique is the generation of transient stress in tissue under a short laser pulse irradiation. Two conditions must be met to generate a photoacoustic signal, stress and thermal confinement as the laser pulse duration must be shorter than a thermal diffuses length of tissue and shorter than the time of the stress to transit the heat region. The main conclusions noted from the present work are:

i- The distance from the zero time to the inflection point of the signal represents the delay time, which can be used to calculate the location of object inside the normal tissue.

ii- The amplitude of the signal is linearly proportional with laser pulse energy and inversely with the target depth, diameter of the target object and the distance to the detector.

iii-The duration of acoustic signal wave generated is not effected by depth of object and laser pulse energy.

iv- The signal from the simulation has a sine wave shape with the same amplitude in the positive and the negative regions.

v- The time between thepositive peaks and the negative peaks of the PA signal which shall represent the size of the target object, is 75% of the assumed real dimension for the whole range of the simulation results for the spherical model simulation.

vi- The acoustic signal wave is not filling generated at 10 μs laser pulse duration in order to fill generated the laser pulse duration must be the shorter than the stress conferment time.

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محاكاة التصوير الضوئي-الصوتي احادي البعد

محمد عبد الامير حسين **لمى وليد جميل**

قسم هندسة الليزر والالكترونيات البصرية، جامعة النهرين ، بغداد ، العراق

الخلاصة تم في هذا العمل استعمال نظرية التصوير الضوئية- الصوتية لكشف وقياس جسيم (اوعية دموية واورام) داخل النسيج السليم. ان الهدف الرئيسي هو الحصول على اشارة صوتية تساعد في قياس حجم وموقع الجسيم داخل النسيج. تم استعمال النموذج الرياضي الكروي للتعبير عن الجسيم في داخل النسيج. برنامج الحاسب الالي في بيئة MATLABللتحقيق هذه المحاكاة. ان هذا النموذج يولد اشارة موجة صوتية لجزئيها الانضغاطي والتخلخلي. تمت محاكاة الاشارة الضوئية- الصوتية لجسيم الهدف لمدى10نانو ثانية – 10 مايكرو ثانية للنبضة اشعة الليزر منبعث من Nd:YAG الليزر ،عمق الجسيم 0.3-3 سم، بعد الجسيم عن الكاشف 0.7-3 سم ونصف قطر الجسيم 0.1- 0.6سم. ان نصف قطر الجسيم من خلال المحاكاة هو دائما 75% من نصف القطر الاصلي المفروض ،سعة الاشارة تقل عند انخفاض طاقة نبضة الليزر ،زيادة عمق الجسيم وبعده عن الكاشف. الاشارة الضوئية-الصوتية تتولد كاملة عند نطاق نبضة الليزر في نانو ثانية فهي قصيرة بما فيه الكفاية للحدوث شرط حبس الاجهاد.