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The Effect of Laser Wavelength in the Simulation of Laser Generated Surface Waves in Human Skin Model

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Abstract

A Finite Element (FE) Simulation of the thermoelastic laser generated surface waves in a 3-layered model of human skin is presented. Commercial finite element code ANSYS is used to study the effects of changing laser wavelength and hence the optical absorption has on the generated surface waves. The FE model consists of a thermal analysis with a volumetric heat generation boundary condition to simulate the thermal effect of the laser source penetrating into the skin. The results from the thermal analysis are then subsequently applied as a load in a mechanical analysis where the out-of plane displacement histories and temperature fields are analysed using two different laser sources to generate the ultrasonic waves.

Introduction

The use of laser ultrasonic waves has been widely utilised in the non-destructive evaluation of layered materials in industry [1,2]. In the thermoelastic regime various waves can be excited in the solid including longitudinal and transverse waves, surface acoustic waves (SAW) and in thin plate Lamb waves [1]. Information regarding material properties and layer thickness then can be extracted from ultrasonic bulk or/and surface waves [1,2].

In biomedical applications of laser ultrasonics, temperature changes of the tissue due to short laser pulses is limited by degrees or fractions of degrees. This temperature increase causes a rapid thermal expansion which in turn generates ultrasonic waves in the tissue. The characteristics of laser ultrasonic waves depend strongly not only on the optical penetration depth, thermal diffusion, elastic and geometrical features of the tissue as well as the parameters of the exciting laser pulse, including the shape, focus spot and pulse width [3] and can be used to characterise tissue properties. In non-metallic materials such as biological tissues, the phenomenon of laser generated ultrasound is dominated by the optical penetration depth and is determined by the properties of the material and the laser wavelength used [1]. To keep the generation of ultrasonic waves in the thermoelastic regime the energy incident on the tissue will be kept below the threshold at which irreversible changes occur, yet sufficient to produce thermoelastic waves that can be readily detected. Irreversible changes in tissues may occur at energy levels below the ablation threshold and consideration should also be given to the possibility of damage to the tissue by the mechanical disruption of tissue layers as a result of the generation and propagation of the thermoelastic wave. In this paper we present a FE modelling technique to study the effect that changing the laser wavelength and hence the optical penetration depth has on the generated SAWs.

Finite Element Model

The finite element method is a versatile modelling method due to its flexibility in modelling the complicated geometry and its capability in obtaining full-field approximate solutions. In addition the FE method is useful in calculating the excitation process, where thermal diffusion, optical penetration and other physical parameters are dependent on temperature. The modelling technique that is used is an

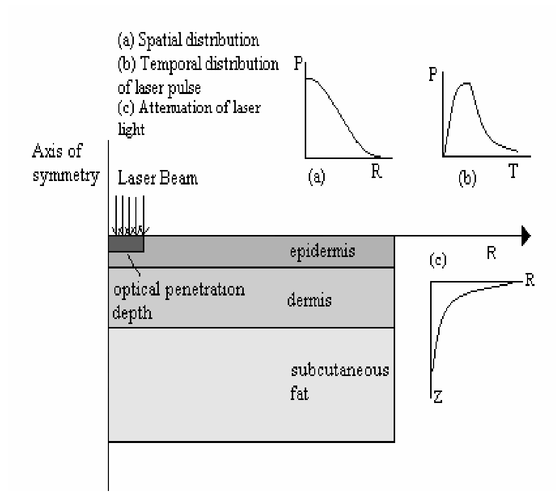


Figure 1. Schematic diagram of laser-irradiated sample showing model geometry, attenuation of laser light and surface spatial and temporal distribution.

example of an uncoupled thermal and mechanical finite element analysis, the two analyses can be solved separately as the effect that the stress field that is created has on the temperature is negligibly small. The heating of the skin model due to a laser pulse is simulated by a dynamic thermal analysis using a volumetric heat generation as a representation of the laser pulse. The nodal temperatures from the thermal analysis are input as loads in the mechanical analysis and the time-dependent out-of-plane displacement histories at various locations on the surface of the model are analysed.

For laser induced heat-transfer, heat loss by convection and radiation is neglected. The classical heat conduction equations for the finite element analysis with the heat capacity matrix *[C],* the conductivity matrix *[K]*, the heat flux vector $\{p_1\}$ and the heat source vector $\{p_2\}$ can be expressed as:

$$[K]\{T\}+[C]\{\dot{T}\}=\{p_1\}+\{p_2\} \qquad [1]$$

where $\{T\}$ is the temperature vector, $\{\dot{T}\}$ is the temperature rate vector. For the wave propagation, and ignoring damping, the governing finite element equations are:

$$[M]\{D\ddot{U}\}+[K]\{U\}=\{f_{ext}\} \qquad [2]$$

where $[M]$ is the mass matrix, $[K]$ is the stiffness matrix, $\{U\}$ is the displacement vector, $\{\ddot{U}\}$ is the acceleration vector and $\{f_{\mathrm{ext}}\}$ is the external force vector. For thermoelasticity the external force vector for an element is:

$$\int_{ve}[B]^T[E]\{\varepsilon_o\}dV \qquad [3]$$

where $\{\varepsilon_o\}$ is the thermal strain vector, $[B]^T$ is the transpose of the derivative of shape functions and $[E]$.

Thermal analysis

In order to develop the laser ultrasonic technique into an effective method of quantitative inspection it is first necessary to accurately simulate the laser induced temperature distribution. An exact representation of light propagation in skin requires a model that characterises the spatial distribution and the size distribution of tissue structures, their absorbing qualities and their refractive indexes. However for real tissues this is an almost impossible and a number of assumptions and simplifications have to be made.

The skin models consist of three layers, the epidermis, dermis and subcutaneous fat. We make the assumption that the thickness of the different layers in constant on a small scale. The meshes are placed in parallel according to the layer thicknesses are then connected to each other. A collimated laser beam normal to the surface has a small portion of the light reflected at the surface and the remaining light is attenuated in the tissue by absorption and scattering. The thermal effect of the laser pulse depends on the laser parameters in this paper, the surface spatial mode of the laser beam is assumed to be Gaussian in nature with the attenuation of laser light in the skin described using Beer-Lamberts Law as shown in equation 4.

$$\phi(r,z)=E_o\exp[-2r^2/r_o^{\,2}]\exp[-(\mu_a+\mu'_s)z] \qquad [4]$$

where $\phi(r,z)$ is the laser fluence, E_o is the radiant exposure at the tissue surface, r the radial coordinate, z is the coordinate that describes the depth below the surface, μ_a is absorption coefficient, μ_s the scattering coefficient and r_o is the beam radius. The highly forward scattering nature of soft tissue suggests that most of the scattered light is in the same direction as the collimated beam. It is therefore possible to improve Beer's law by replacing the scattering coefficient with the effective scattering coefficient [$\mu'_s = \mu_s(1-g)$]. Thus Beer's law can be improved for wavelengths where there is considerable scattering as:

$$\phi(r,z)=E_o\exp[-2r^2/r_o^2]\exp[-(\mu_a+\mu_s(1-g))z] \qquad [5]$$

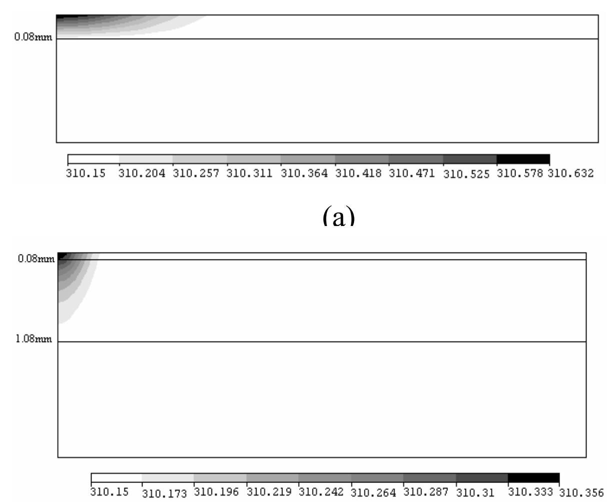
This is still an approximation and only considers collimated light and the forward scattered light in the z-direction. Light scattered in the other directions is neglected. The source term $Q(r,z,)$ describes the rate of heat deposition in tissue due to laser irradiation. The heat source term is a product of the absorption coefficient of the tissue and the laser irradiance. The rate of heat deposition per unit area is then given as

$$Q(r,z)=\mu_a\phi(r,z)) \qquad [6]$$

The temporal distribution of laser irradiation is assumed to be Gaussian in nature also given by:

$$g(t)=\frac{t}{to}\exp(-\frac{t}{to}) \qquad [7]$$

where t is time and t_o the rise time of the laser pulse.



(a)

(b)

Figure 2 contour plot of temperature distribution at 20ns after (a) 1mJ laser in which absorption predominates (b) 50mJ laser in which scattering predominates.

Therefore the volumetric heat generation equation used to simulate the laser in the skin is given as:

$$Q(r,z,t) = \mu_a \phi(r,z) g(t) \quad [8]$$

In the thermal analysis, the finite element mesh is constructed using four-node axisymmetric quadrilateral elements. The skin model dimensions are 20 mm in length with epidermis depth of 0.08 mm, dermis depth of 1mm and subcutaneous fat depth of 10 mm. The thermal properties of the skin layers are shown in table 1. The laser beam is modelled as having Gaussian spatial and temporal distribution with the intensity of the laser beam decreasing with depth according to Beer Lambert's Law as shown in Figure 2 using equation 5 with the fluence rate calculated using equation 1 or 2 depending on the laser wavelength used. The laser beam is perpendicular to the surface of the laser so the distributions of physical quantities such as fluence and heat generation rate are symmetrical to the axis of the incident light; therefore a cylindrical coordinate system is assumed.

Two laser sources are used in the simulations. The first is assumed to be a 1mJ CO_2 10.6μm laser in which absorption predominates and the laser energy will be absorbed at the surface of the skin. The optical properties of all the skin layers are assumed constant with the absorption coefficient of the skin layers 86 mm^{-1}. The second laser is assumed to be 50 mJ frequency doubled Nd:YAG 532nm laser where scattering predominates and the laser will penetrate further into the skin, the absorption and reduced scattering coefficient of the skin layers in this simulation is 0.2 mm^{-1} and 2.5 mm^{-1} [4] respectively. Both lasers have a rise time of 10 ns and a spot radius of 0.5 mm. The temperature fields in the multilayer skin models are calculated with the time steps in the analysis are kept in the range of 0.1ns during the duration of the pulse and are allowed to increase after the laser pulse. The thermal analysis is run for 0.1 seconds in order to provide a complete temperature history for the duration of the mechanical analysis.

<u>Mechanical Analysis</u>

The mesh in the mechanical analysis is composed of four node axisymmetric quadrilateral elements. The mechanical properties used in the simulations are given in table 1. The nodal temperatures from the thermal analysis are mapped onto the mesh for the mechanical analysis. The element size was kept the same as the mesh in the thermal analysis, with the analysis run for the same time as the thermal analysis and the out-of-plane displacements at various locations along the surface of the model are recorded.

Temporal and spatial resolution of the finite element model is critical for the accurate convergence of the numerical results. The rule that is followed is the time steps should be small enough to measure 20 points per cycle of the highest frequency component (f_{max}). The minimum element size is chosen in the same manner so that the propagating waves are spatially resolved. As a rule more than 20 nodes per minimum wavelength (λ_{min}) is used.

Results

The contour plot of the temperature distribution in Figure 2(a) shows that the heat affected zone during laser irradiation is localised and is absorbed within 0.08 mm depth, a maximum temperature increase of 0.482 K is achieved at the centre of the laser pulse. This is in contrast to Figure 2(b) which shows that the laser energy is absorbed over a much larger volume of the tissue sample due to scattering. In the laser irradiated sample in which scattering predominated the maximum temperature is 0.206 K which is smaller than the temperature increase of the 1mJ laser irradiated sample but the affected area is greater therefore the possibility of skin damage is increased. Due to the assumptions made in the boundary condition of the simulated laser pulse the shape of the heat affected are is cylindrical in shape as the effects of radial scattering are neglected in these simulations. The pulse duration of the laser pulse in these simulations is in the nanosecond range.

	Epidermis	Dermis	Subcutaneous Fat
Density (gmm^3)	$1.2x10^{-3}$	$1.2x10^{-3}$	$1.0x10^{-3}$
Specific Heat (C) ($Jg^{-1}K^{-1}$)	3.590	3.300	1.900
Thermal Conductivity (K) ($Wmm^{-1}K^{-1}$)	$2.4x10^{-4}$	$4.5x10^{-4}$	$1.9x10^{-4}$
Young's ModulusE (Pa)	$1.36x10^5$	$8.0x10^4$	$3.4x10^4$
Poisson's Ratio (ν)	0.499	0.499	0.499
Thermal Expans' coeff' (α) (K^{-1})	$3.0x10^{-4}$	$3.0x10^{-4}$	$9.2x10^{-4}$

Table 1. Thermal and Mechanical properties of skin layers used in FE simulations [6].

In the regime of stress confinement the main mechanism of energy transformation is non-radiative relaxation where the absorbed laser energy is converted into heat and produces an acoustic source in the absorbed region which produces elastic waves. The distribution of this optical energy corresponds to the shape of an acoustic transmitter and the characteristics of the generated ultrasonic waves are strongly dependent on the optical penetration of the laser beam.

Figure 3 and 4 show the laser generated surface acoustic waves at different locations on the skin model surface for both lasers. SAWs penetrate into a solid by a few wavelengths deep with amplitudes that decay exponentially with depth and a penetration depth that varies with wavelength. In order to develop the laser ultrasonic technique into an effective method of characterising skin properties it is necessary to obtain an amplitude of wave that is easily measured by interferometric techniques. It can be seen that the amplitudes of the ultrasonic waves and the shape of the waveform varies considerably depending on the laser source used. The amplitudes obtained from the 1mJ laser source are smaller than the amplitudes measured using the 50 mJ laser. It can therefore be assumed that more useful surface wave amplitudes can be obtained by having a larger acoustic transmitter in the tissue, however this will greatly increase the possibility of thermal damage to the skin but also damage to the tissue by mechanical disruption as a result of the generation and propagation of the thermoelastic wave.

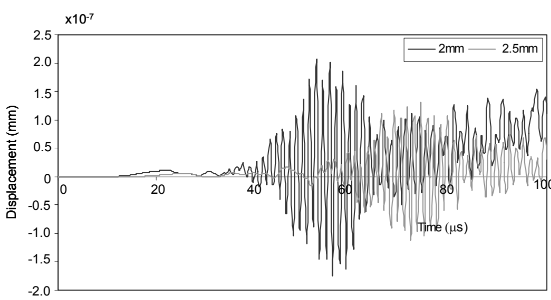


Figure 3. Out-of plane displacement history at varying source receiver distances using 1mJ laser source

Conclusion

Numerical simulations using FE code ANSYS in a three-layered model of skin are presented to study the effect that changing the wavelength of the generating laser has on the generated SAWS. A sequential coupled-field thermal and mechanical analysis with a BC applied to simulate the laser pulse taking into account the spatial, temporal distribution of the pulse along with the material absorption and scattering characteristics. The results obtained using a laser which is strongly absorbed in skin and a laser that is strongly scattered in skin are presented showing that much greater amplitudes of waves are generated by using a laser source that penetrates deeper into the skin will keeping the maximum temperature rise at less than 1 degree. The downside of using a laser that penetrated deeper into the skin however is that a greater volume of tissue is affected and the possibility of damaging the skin increases. Therefore further research will be done using a number of different laser sources and by varying the pulse characteristics to obtain the best combination of laser characteristics to obtain amplitudes that are easily measured while minimising any risk of skin damage.

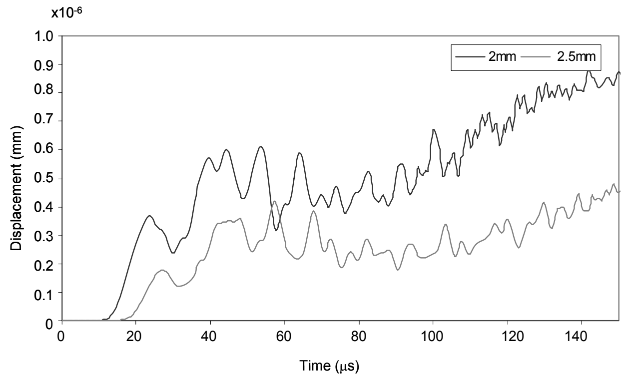


Figure 4. Out-of plane displacement history at varying source receiver distances using 50mJ laser source

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