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Skin characteristics by Laser generated surface Waves

Zhihong Huang and Adèle L'Etang

Abstract— **This paper discusses a study into the suitability of using laser generated surface acoustic waves for the characterisation of skin properties without causing any damage to the skin thermally or by mechanical disruption. Using commercial Finite Element Code ANSYS, the effects of laser wavelength, laser beam radius and laser rise time on generation of laser generated ultrasonic waves in a 3-layered elastic isotropic model of human skin were studied. The FE model is an example of a sequential coupled field analysis where the thermal and mechanical analyses are treated separately. The heating of the skin model due to the short laser pulse is simulated by a dynamic thermal analysis with the laser pulse modeled as volumetric heat generation and the results from this analysis subsequently applied as a load in the mechanical analysis where the out-of-plane displacement histories are analyzed. The technique described in this paper also involves measuring the propagation velocity of SAWs, which are directly related to the material properties, and thickness of layers, this is done over a wide frequency range in order to obtain maximum information regarding the material under test.**

I. Introduction

THE use of laser generated ultrasonic has been widely used for the non-destructive evaluation of layered materials. Using velocity measurements of the generated ultrasonic bulk or surface waves, information can be extracted relating to layer thickness and mechanical properties of the layers in a material [1, 6]. The work on laser to generated acoustic waves by Wu and Liu [3] discussed the dispersion of the laser-generated surface waves in an epoxy bonded copper-aluminum layered specimen. The results show a clear influence of bonding layer thickness on the surface wave dispersion and the method could be applied to the non-destructive evaluation of the bonding properties.

In biomedical applications of laser ultrasonics, the temperature changes of the tissue due to short laser pulses should be limited to degrees or fractions of degrees in order to not destroy or damage the skin and keep the generation of ultrasonic waves within the thermoelastic regime. This temperature increase causes a rapid thermal expansion which in turn generates ultrasonic waves in the solid. In the thermoelastic regime various waves can be excited in the solid including longitudinal and transverse waves, surface acoustic waves and in thin plate Lamb waves.

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 and can be used to characterize 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.

In this paper we present research expanding this principle for the use of generated SAW for the characterisation of skin layer properties. A finite element (FE) modeling technique is presented to study the effect of the exciting laser pulse has on the properties of the generated Surface Acoustic Waves. Using the commercially available finite element code ANSYS, the effect of laser beam width, pulse rise time and laser wavelength has been studied in a 3-layered skin model. The SAW dispersion relations are calculated for Finite Element simulated SAW displacement waveforms over a range of source-detector separations in three-layered models of human skin. The simulations show that SAWs are extremely sensitive to changes in layer properties and will be able to be utilised to quantitively characterise the layer properties of human skin by the development of an inverse algorithm.

II. Finite Element Analysis

In this work the Finite Element code ANSYS is used to model the generation and propagation of ultrasonic waves where thermal diffusion and optical penetration depth of laser irradiation is being considered.

Sequential coupled-field analysis is carried out where the thermal and mechanical analyses are treated separately, as it

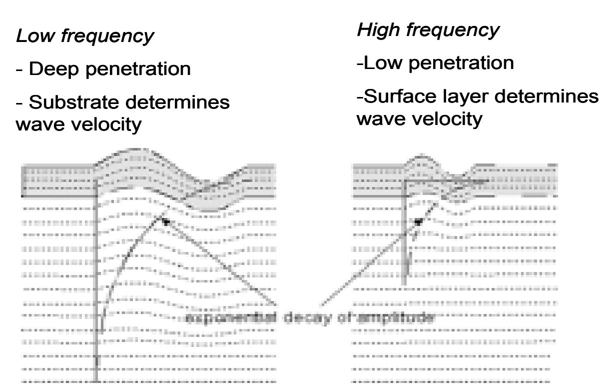

Fig 1.Schematic diagram of surface waves in layered media [2]

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Zhihong Huang and Adèle L'Etang, School of Engineering, Physics and Mathematics, University of Dundee, Dundee, DD1 4HN, UK
z.y.huang@dundee.ac.uk, and a.j.letang@dundee.ac.uk

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is assumed that the effect that the stress field that is creates has on the temperature field is assumed to be negligibly small. The skin models consist of three simplified layers, namely the epidermis, dermis and subcutaneous fat. \in this work it is assumed that the thicknesses of the different layers are constant on a small scale. The meshes are placed parallel according to the layer thicknesses and are assumed to be bonded together. The heating of the skin model due to a laser pulse is simulated using a dynamic thermal analysis and the nodal temperatures from the thermal analysis are input as the loads in the subsequent mechanical analysis. The time-dependent out-of-plane displacement histories at various locations on the surface of the model are analyzed to show the differences in the surface waves due to the characteristics of the generating laser pulse. Due to the large computer resources required for these simulations an axisymmetric model is employed to reduce computer run times.

A. Thermal Analysis

In order to develop an accurate and close representation of the light propagation into skin would require a model which characterizes the spatial and size distribution of the tissue structures, their absorbing qualities and the refractive indexes, therefore a number of assumptions and simplifications have been made. Assumptions that are made in this work regarding the thermal response of the material model are that the thermal expansion that occurs due to the laser heating occurs over a time span close to that of the pulse duration and the thermal losses due to radiation and convection are also neglected in these simulations.

In the thermal analysis, the mesh is constructed of four-node axisymmetric quadrilateral elements. The model

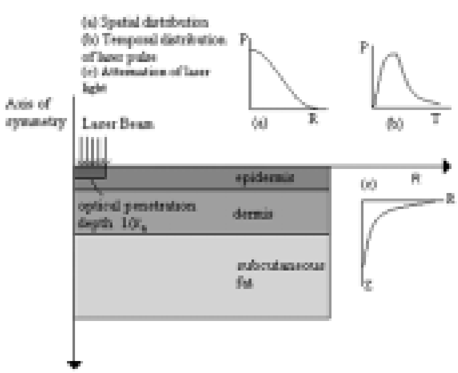

Fig. 2. Schematic diagram of laser-irradiated sample showing model geometry, attenuation of laser light, spatial and temporal distribution of the laser pulse.

dimensions are 20mm in length with an epidermis depth of 0.08mm, dermis depth of 1mm and subcutaneous fat depth of 10mm. The thermal properties of the skin layers used in the simulations are given in Table 1 [7,9] these skin property values have been used to present an average data set of the thermal and elastic properties of skin in the normal range. The heating of the multilayered skin model due to a short laser pulse is simulated by a dynamic thermal analysis with the laser pulse represented as a volumetric heat generation. The laser beam is modeled 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 equation 1. A cylindrical coordinate system is set up with the origin at the centre of the point of incidence and the z-axis directed into the material. The irradiated energy is assumed to be vertically incident on the material surface, the distribution of physical quantities such as fluence and heat generation are symmetrical to the axis of the incident light.

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

Where ϕ(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, μa is the 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 [μ_s = μ'_s(1-g)], where the anisotropy factor, g, indicates the angular deflection of a photon's trajectory caused by a scattering event.. Thus Beer's law can be improved for laser wavelengths where there is considerable scattering as:

$$\phi(r,z) = E_o \exp[-2r^2 / r_o^{\,2}]\exp[-(\mu_a + \mu'_s\,(1-g))z] \qquad [2]$$

This is still an approximation and only considers collimated light and the forward scattered light is in the z-direction. Light scattered in all 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 and the laser fluence. The rate of heat deposition per unit area is then described as:

$$Q(r,z) = \mu_a \phi(r,z) \qquad [3]$$

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

$$g(t) = \frac{t}{t_o}\exp\left(-\frac{t}{t_o}\right) \qquad [4]$$

Where t is time and t_o is the rise time of the laser pulse.
Therefore the volumetric heat generation boundary condition used to simulate the laser in the skin model is given as:

$$Q(r,z,t) = \mu_a \phi(r,z) g(t) \qquad [5]$$

In this paper a number of different laser sources used in these simulations, these are shown in table 2.

In these simulations the laser heating is limited to degrees or fractions of a degree and it assumed that this will not cause any permanent damage to the skin tissue being examined. The skin's exposure to laser radiation must be limited for safety reasons to reduce the risk of permanent tissue damage [5,6]. The maximum permissible exposure (MPE) can be defined as the level of EM radiation that a person can be exposed to without hazardous biological changes in the tissue. The MPE are usually expressed in terms of radiant exposure in J/cm2 or as irradiance in W/cm2 and as a function of EM wavelength (or frequency), exposure time and pulse duration. Exposure to energy above the MPE can potentially result in tissue damage. In general the longer the wavelength the higher the MPE and the longer the exposure time the lower the MPE.

The temperature fields in the multiplayer skin models are calculated with time steps in the range of 0.1ns for the duration of the laser pulse and allowed to increase thereafter. The thermal analysis is run for 0.1s in order to provide a complete temperature history for the entire duration of the mechanical analysis.

B. Mechanical Analysis

The mesh used in the mechanical analysis is identical to the one used in the thermal analysis composed of four-node axisymmetric quadrilateral elements. The nodal temperatures from the thermal analysis are mapped onto the mesh as the load for the mechanical analysis. Temporal and spatial resolution is critical for the accurate convergence of the numerical results. The rule that applies is that 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 (λmin) is used.

TABLE 1
THERMAL AND MECHANICAL PROPERTIES OF SKIN LAYERS USED IN FE SIMULATIONS

	Epidermis	Dermis	Subcutaneous Fat
Density (gmm^3)	$1.2\text{x}10^{-3}$	$1.2\text{x}10^{-3}$	$1.0\text{x}10^{-3}$
Specific Heat (Jg^{-1}K^{-1})	3.590	3.300	1.900
Thermal conductivity (Wmm^{-1}K^{-1})	$2.4\text{x}10^{-4}$	$4.5\text{x}10^{-4}$	$1.9\text{x}10^{-4}$
Young's Modulus (Pa)	$1.36\text{x}10^{5}$	$8.0\text{x}10^{4}$	$3.4\text{x}10^{4}$
Poisson's Ratio	0.499	0.499	0.499
Thermal Expansion Coefficient (K^{-1})	$3.0\text{x}10^{-4}$	$3.0\text{x}10^{-4}$	$9.2\text{x}10^{-4}$

TABLE 2
CHARACTERISTICS OF LASER USED IN SIMULATIONS

Laser Source	1 mJ CO_2 10.6µm	1mJ CO_2 10.6µm	0.5mJ CO_2 10.6µm	50mJ Nd:YAG 532nm
Rise Time	10ns	10ns	20ns	10ns
Beam Radius	0.5mm	1.0mm	0.5mm	0.5mm
Absorption coefficient	86mm^{-1}	86mm^{-1}	86mm^{-1}	0.2mm^{-1}
Scattering Coefficient	Na	Na	Na	2.5mm^{-1}

C. Results

Figure 3 shows the contour plots of the model at the end of the laser pulse, in figure 3(a) that of the highly absorbing laser pulse the heat affected zone is localized and is absorbed within a 0.08mm depth with a maximum temperature increase obtained of 0.482K at the centre of the laser pulse. This is the characteristic temperature distribution for the three simulations using this type of laser source where a small volume of the tissue is affected by the laser pulse. This is in contrast to Figure 3(b) which shows the contour plot of the more highly scattering laser wavelength. When using this laser wavelength the laser energy is absorbed over a much larger volume of the tissue due to the scattering of the energy. The maximum temperature increase obtained here is 0.206K; this corresponds to a smaller increase in temperature than in the CO_2 laser simulation but there is a greater possibility of damage to the skin in this case due to the larger affected area. It is the distribution of optical energy in the skin model corresponds to the shape of an acoustic transmitter and has a strong influence on the characteristics of the generated ultrasonic waves.

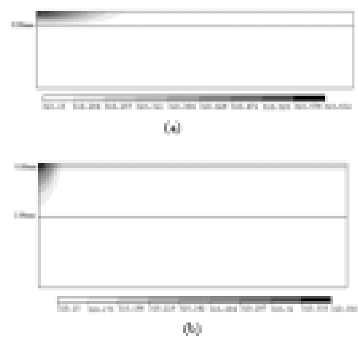

Fig. 3. Contour plot of temperature distribution at 20ns after (a) 1mJ laser in which absorption predominates and (b) 50mJ laser in which scattering dominates.

Figure 4 shows the example of simulated out-of-plane displacement histories at different points on the surface of the skin models for the simulations. SAWs penetrate into a solid by a few wavelengths with amplitudes that decay exponentially with depth and a penetration depth that varies with the wavelength of the elastic wave. As the SAW travels along the distance of the model the characteristics of the waveform changes which is due to the dispersion effect.

Rayleigh waves are SAW in which the longitudinal and shear displacements are coupled together and travel at the same velocity. To determine the dispersion relation the SAWs (out-of-plane displacement histories) [10-12] are measured at two distances from the source on x_1 and x_2 on the model surface. The phase spectra $\phi_1(f)$ and $\phi_2(f)$ is calculated by applying a Fourier transform to the measured waveforms. The phase velocity c depending on frequency f (dispersion curve) is then determined by inserting $\phi_1(f)$ and $\phi_2(f)$ into the following equation:

$$c(f) = \frac{(x_2 - x_1)\omega}{\phi_2(f) - \phi_1(f)} \quad [6]$$

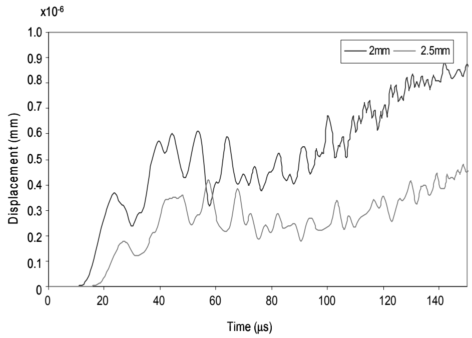

Fig. 4 Out-of-plane displacement histories recorded on the surface of the skin models using 50mJ Nd:YAG laser source.

where ω denotes the angular frequency.

Figure 5 shows the calculated dispersion curves in three models with varying Young's Modulus of the dermis layer. It is clear that the amplitudes, frequency and shape of the generated waveforms vary considerably depending on the laser pulse characteristics. Higher frequency waveforms which are confined closer to the surface are produced by using laser beam with a smaller radius and rise times and by using laser sources that are more highly absorbing instead of one that penetrates further into the skin layers. It can also be seen from the above results that larger amplitudes of waves can be generated when there is a large acoustic transmitter in the tissue, however when the temperature increase is larger or affects a larger section of the sample the possibility of thermal damage has to be considered as well as the possibility of mechanical damage due to disruption caused by the generation and propagation of the elastic wave.

III. Conclusion

The simulated waveforms in multilayered skin models using four different laser sources are presented in this paper. For the use of laser generated surface waves to characterize skin properties it is necessary to generate waves that can be easily measured using interferometric techniques for these simulations. In order to efficiently generate SAWs in the skin model the laser needs to be strongly absorbed in the sample rather than using a source which is strongly scattered this results in more elastic wave energy being generated closer to the surface of the material being tested and of higher frequency waves being produced. The shorter the laser pulse time also results in the generation of higher frequency SAWS which will be closely confined to the surface of the sample this is also true of using smaller laser beam radii.

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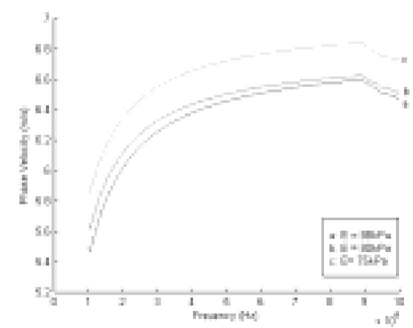

Fig. 5. Phase Velocity dispersion curve measured on three layered skin models with variation in Young's Modulus of dermis.