Computational Model of Photothermal Microscopy in Tissue

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Abstract: This research, through a rigorous optical, thermal and mechanical computational analysis, demonstrates the use of Photothermal Microscopy to tag light from the focus, and thereby improve contrast and depth of imaging limited by out-of-plane scatter. (180.1790) Confocal Microscopy; (190.4870) Photothermal Effects; (170.1870) Dermatology.

1 Introduction

The primary optical techniques for imaging skin are Confocal Microscopy and Optical Coherence Tomography (OCT). Each has its advantages and drawbacks regarding penetration and resolution. OCT achieves a penetration depth of 1mm-2mm with a resolution of <15µm. Confocal Microscopy produces significantly better resolution (<3µm), but has poor penetration resulting from clutter produced by light scattered from outside the focal plane.

An improvement upon these techniques is to tag the light at the focal plane by utilizing a pulsed focused heating laser to cause tissue expansion. A coherent confocal microscope can then be implemented, acting as a laser vibrometer, to measure this expansion. This method of optical detection of changes in elastic displacement due to the photothermal process, introduced by Eliyahu et al. [1] for a simple case, will be evaluated in this research.

2 Simulation

The numerical model employed was initially developed by Kowalski et al. [2] to simulate thermal effects in nonlinear optical media through the use of a finite difference scheme to solve the Fourier conduction equation:

\[ \nabla^2 T = \left( \frac{\rho c}{K_t} \right) \frac{\partial T}{\partial t} - \frac{q_{abs}}{K_t}. \quad (1) \]

where \( T \) = temperature, \( t \) = time, \( \rho \) = density, \( c \) = specific heat, \( K_t \) = thermal conductivity and \( q_{abs} \) = absorbed radiation per unit volume. The incident field, for this case a Gaussian beam symmetric about the x and y axes, is described by rays passing through a modeled sample composed of a grid of voxels in all three dimensions. Information is collected over several time steps for each voxel including.
optical path length (OPL) change, temperature, index of refraction, stress and displacement. At each step, the ray trajectories are corrected for changes in the index of refraction, $n$, caused by heating according to

$$ n(r) \frac{d^2 r}{d\ell^2} = \nabla n(r) - \frac{d r}{\ell} \frac{d n(r)}{d\ell} \quad (2) $$

The tissue displacement along each axis is found through Hooke’s Law after formulation of a thermo–elastic potential solved for the stress. In the $x$ direction, for example, this is:

$$ \Delta x = \frac{\sigma_x - \nu (\sigma_y + \sigma_z)}{E} dx + \alpha \Delta T dx \quad (3) $$

where $\sigma = \text{stress}$, $\alpha = \text{linear coefficient of expansion}$, $\nu = \text{Poisson’s ratio}$ and $E = \text{Young’s Modulus}$. These parameters are handled by a post processing code developed in MATLAB to calculate the phase of light waves scattered from particles throughout the sample with

$$ OPL = \int n d\ell \quad (4) $$

where $\ell$ is along the ray. Integrating the field amplitudes with these phase shifts over sets of particles at different depths (before, near and after the focus of the beam) shows how the changes in index of refraction and displacement contribute to a phase change in the total signal over time.

3 Conclusion

The temporal change in the coherent sum of scattered field amplitudes makes it possible to reject the clutter produced in measurements at deep penetration depths by the light backscattered from particles outside the focal plane. This will enhance the capabilities of the confocal microscope to image deeper while maintaining its high resolution. The model provides rigorous quantitative estimates of displacement, and also demonstrates the importance of accounting for heat–induced changes in the refractive index.

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References
