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Stress Wave Focusing Transducers

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Introduction

Conversion of laser radiation to mechanical energy is the fundamental process behind many medical laser procedures, particularly those involving tissue destruction and removal. Stress waves can be generated with laser radiation in several ways: creation of a plasma and subsequent launch of a shock wave, thermoelastic expansion of the target tissue, vapor bubble collapse, and ablation recoil. Thermoelastic generation of stress waves generally requires short laser pulse durations and high energy density. Thermoelastic stress waves can be formed when the laser pulse duration is shorter than the acoustic transit time of the material:

$$\tau_c = d/c_s \qquad (1)$$

where d = absorption depth or spot diameter, whichever is smaller, and c_s = sound speed in the material. The stress wave due to thermoelastic expansion travels at the sound speed (approximately 1500 m/s in tissue) and leaves the site of irradiation well before subsequent thermal events can be initiated. These stress waves, often evolving into shock waves, can be used to disrupt tissue. Shock waves are used in ophthalmology to perform intraocular microsurgery and photodisruptive procedures as well as in lithotripsy to fragment stones. We have explored a variety of transducers that can efficiently convert optical to mechanical energy. One such class of transducers allows a shock wave to be focused within a material such that the stress magnitude can be greatly increased compared to conventional geometries. Some transducer tips could be made to operate regardless of the absorption properties of the ambient media. The size and nature of the devices enable easy delivery, potentially minimally-invasive procedures, and precise tissue-targeting while limiting thermal loading. The transducer tips may have applications in lithotripsy, ophthalmology, drug delivery, and cardiology.

Materials and Methods

The geometry of acoustic lenses was studied with respect to focusing laser-produced stress waves. Laser radiation was supplied by a frequency doubled Nd:YAG laser (model 250-10 GCR, Spectra Physics, Mountain View, CA) emitting 5 ns-duration pulses of 532 nm light. The laser radiation was coupled into quartz optical fibers having diameters ranging from 0.2-1.0 mm. Several transducer focusing tips of differing materials and dimensions were produced and tested for their coupling efficiency and focusing ability. A variety of transducer tips could be produced, ranging from shaping the fiber tip itself, to complicated mechanical devices. The basic design of one transducer is shown in Figure 1. This tip has a cavity for holding an absorbing media such as a dyed liquid. The laser-produced stress wave is formed in the absorbing media and transmitted to the transducer tip which is shaped to focus the stress wave.

Visualization of pressure waves was accomplished with the use of a schlieren shadowgram technique. A Nitrogen pumped-Dye laser tuned to a wavelength of 630 nm was used as the probing illumination beam. The pulse duration of the Ni-Dye laser was 4 ns allowing excellent temporal resolution. The repetition rate of the Ni-Dye laser and the capture rate of the CCD allow only one image per event. A variable delay generator was used to allow photos at any time after the probe beam was delivered. The consistency of the process is excellent so that repeated images with various time delays creates an accurate depiction of stress wave and bubble evolution.

Computer simulations of the generation and propagation of stress waves using the two-dimensional LATIS simulation code have been used to design and analyse experiments with various acoustic lenses and focusing transducers. This code produces a finite-difference time dependent solution of the hydrodynamic equations with realistic equations-of-state and material strength models for the various materials, such as the tissue, water in the transducer, and the transducer material. The space-time distribution of the stress field is calculated. The simulation results are used to determine the optimal curvature of focusing surfaces and the optimal materials for the generation of laser thermo-elastic stresses and for the coupling of those stresses to the targeted tissue.

Results

Experimental evaluation of initial transducer designs demonstrated the feasibility of focusing laser-induced stress waves. A typical time history of a stress wave created by a bare optical fiber in a dye solution (600 cm^{-1}) is shown in Figure 2a. Encapsulating the same dye in the transducer shown in Figure 1, yielded the stress wave excursion in Figure 2b. Better acoustic coupling was achieved with alternate transducer materials. Stress waves were able to be focused by a factor of two, theoretically improving the intensity by a factor of four.

Discussion

We have demonstrated the potential of focusing laser-produced stress waves. These stress waves may be used for applications of targeted tissue disruption such as in lithotripsy, interventional cardiology, or ophthalmology. In addition, less destructive stress waves may be beneficial in drug delivery. The use of transducer tips incorporating an absorber allows for predictable and reproducible events independent of the absorption of the ambient media. The incorporation of tips and a focused geometry also protects the optical fiber from stress induced damage. Further, the increased intensity at the focus enables one to use less laser energy to achieve the same intensity as unfocused stress waves, reducing the non-beneficial thermal load.

Initial transducers were made of aluminum for ease of machining. However these materials have significantly different acoustic impedances compared to tissue, water, or blood. The acoustic impedences of the interfacing materials determine the intensity of the reflected and transmitted stress wave. The reflection coefficient is given by:

$$R = (1-z_1/z_2)/(1+z_1/z_2) \qquad (2)$$

where z = acoustic impedance of the media. Acoustic impedences of water and blood are 1.48 and 1.61, respectively, while aluminum is 17 g/cm^2s, resulting in a reflection coefficient greater than 83%. Subsequent transducers were produced from polystyrene ($z = 2.94\ g/cm^2s$) to better match impedances (R = 29%). The theoretical diffraction limited focal spot radius (84% of energy) is given by:

$$r_0 = 0.61\lambda f/R \qquad (3)$$

where λ = wavelength of sound, f = focal length of lens, and R = initial radius. The focal length of the acoustic lens is given by:

$$f = R/(1-c_L/c_s) \qquad (4)$$

where c_L = liquid sound speed and c_s = solid sound speed. Using practical values ($\lambda = 12\ \mu m$, f= 1 mm, R = 200 μm) a spot radius of less than 40 μm is predicted. Taking into consideration the effective focal depth, this could increase the shock intensity at the focus by approximately 10 times that at the transducer tip. Designs can be further improved by better impedance matching materials and potentially absorbing into the transducer material directly.

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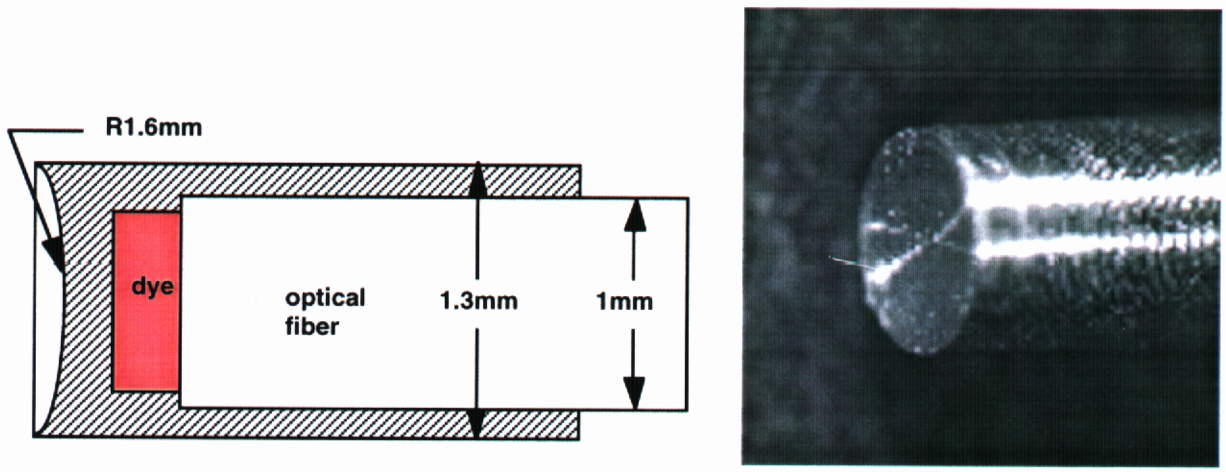

Figure 1. Optical-to-mechanical focusing tip transducer.

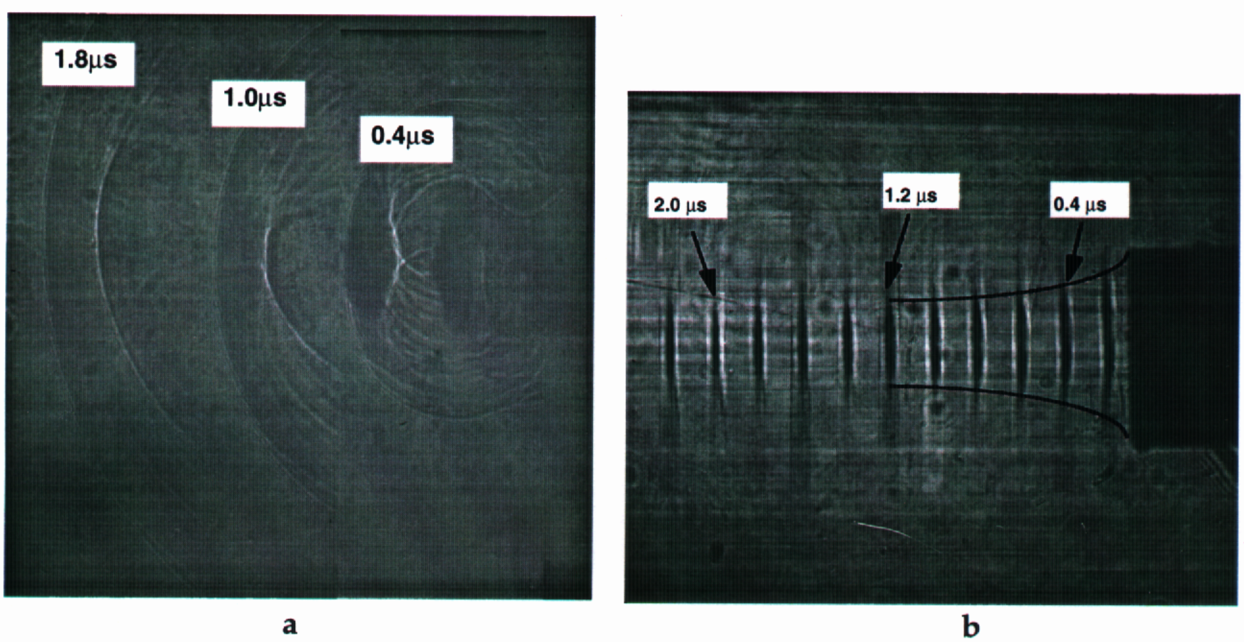

Figure 2. a) Unfocused (bare fiber), **b)** focused (transducer) stress wave evolution: 10mJ/pulse, 8 ns pulse, α=600cm^{-1}

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