

ULTRASOUND MODULATION OF DIFFUSIVE OPTICAL WAVES FOR MEDICAL IMAGING

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ABSTRACT

Diffusive optical tomography [DOT] is a technique for imaging within the body. While DOT provides excellent results under good conditions, there are many situations (due to anatomy or other physical limitation) in which it suffers from a "limited view" problem. In this paper we discuss our work on a new technique for combining DOT with focused ultrasound to generate virtual sources of illumination. These virtual sources help overcome the limited view problem. We present our experimental results using laboratory tissue phantoms.

1 INTRODUCTION

The use of light in medical diagnosis is attractive for several reasons. First, optical imaging uses external light sources and is therefore a non-invasive procedure. Next, light is non-ionizing radiation. Optical imaging therefore can be used more frequently than current X-ray techniques, leading to earlier diagnosis of tumors and correspondingly lower mortality rates[Marks, 1993]. Finally, optical imaging can be used to simultaneously measure both total blood volume and percent oxygenation by exploiting differences in the hemoglobin spectrum (Fig. 1). This will provide information about the local metabolism and would be of great utility in, for example, detecting vascular tumor detection and stroke localization[Chance, 1988, Chance, 1998, Boas, 1994, Dwyer, 1997]. The major limitation on the use of light is that the human body is a highly scattering medium, make image reconstruction extremely difficult.

One technique for imaging in the body is diffusive optical tomography (DOT). In DOT, light is modulated at an RF frequency of up to a few hundred Megahertz and illuminates the tissue using an array of transmitters placed on the surface of the body[O'Leary, 1992, O'Leary, 1995]. The light propagates through the body as a diffusive wave[Knüttel, 1993, Haskell, 1994, Tromberg, 1993, van Rossum and Nieuwenhuizen, 1999]. The scattered light is then collected using an array of receivers configured to receive either reflected or transmitted light, depending on the physical and anatomical constraints. Imaging algorithms are then used to reconstruct the internal absorption and scattering inhomogeneities[Chang, 1995b, Chang, 1995a, Gaudette, 1999, Yao, 1995, Zhu, 1995, Zhu, 1997].

Current algorithms allow detection of anomalies at centimeter depths, but accuracy and resolution on the order of a few millimeters can only be achieved in the region immediately proximal to the receivers or transmitters[Graber, 1995a, Graber, 1995b, Gaudette, 1999]. This limitation arises because DOT is a form of near-field imaging (diffusive optical waves are evanescent waves with wavelengths around 10 cm[Gaudette, 1998]) and also because of the inherently weak signals due to dissipative nature of these waves. Thus, despite the great

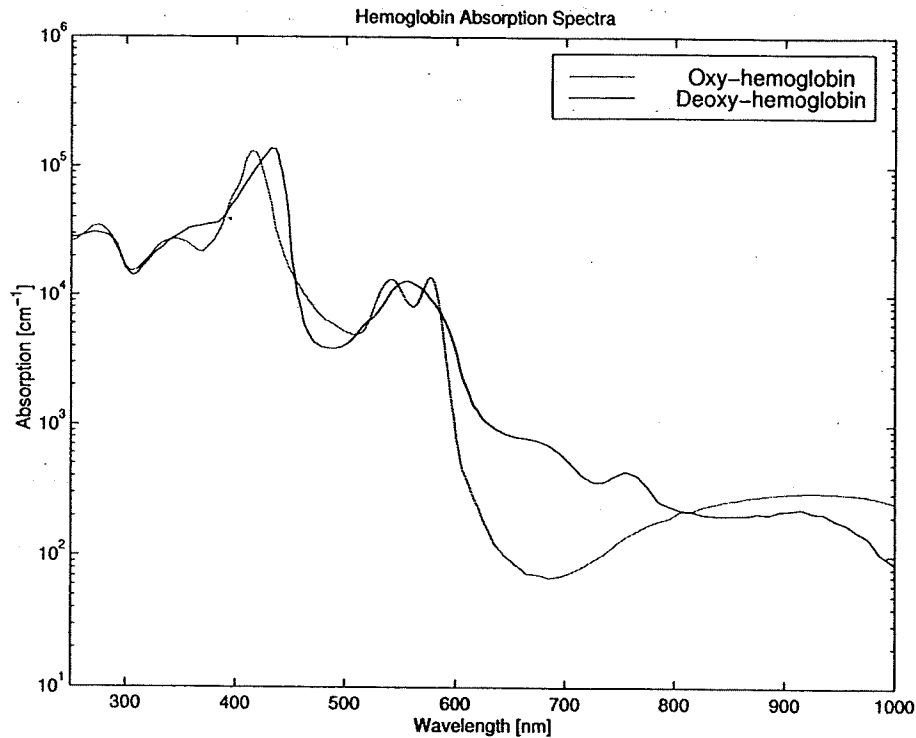


Figure 1: Hemoglobin absorption coefficient vs wavelength. Note the large difference between oxy- and deoxy-hemoglobin at approximately 700 nm.

promise of medical imaging with DOT, medically significant limitations persist. These include imaging inaccuracy in a reasonable depth of tissue and imprecise estimates of absorption or scattering [Gaudette, 1999, Chance, 1988, Graber, 1995a].

In this paper we will demonstrate a new technique for combining DOT and focused ultrasound, in which the DOT signals are modulated by interactions with the acoustic wave, thereby “tagging” the light to indicate that it has passed through the focus of the ultrasound. The resulting light has a different frequency of modulation and can be detected easily in the presence of other stronger, but unmodulated, DOT signals. Thus, the ultrasound focus acts as a “virtual source” of DOT waves. Because ultrasound travels easily through most tissue, sources can be placed in locations inaccessible to other non-invasive probes.

2 ACOUSTO-PHOTONIC IMAGING

In order to understand conventional DOT images, it is common to think in terms of the probability-weighted paths over which photons travel from the source to the receiver (to avoid confusion with the virtual source, we will refer to this actual source of photons as the photon density wave [PDW] source). Figure 2 (left) illustrates this concept for a single active PDW source. Photons from the active PDW source spread out in all directions (drawn as a fan for

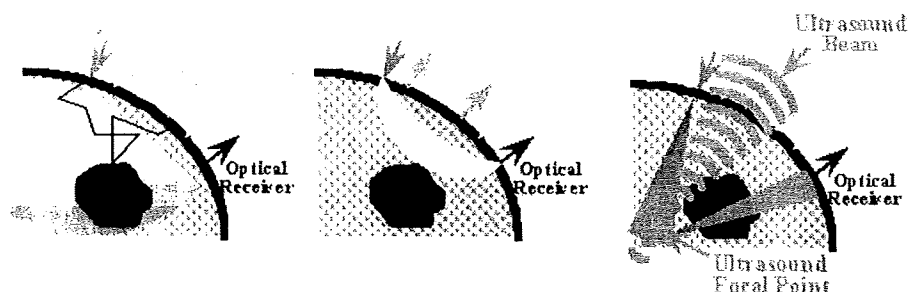


Figure 2: =bf Light Propagation in Tissue (Left), Diffusive Waves in DOT for one Source-Receiver Pair (Center) and API (Right)

clarity), changing directions randomly as they encounter scatterers, and being absorbed as they encounter absorbers. Most of those photons never reach the receiver. A sample photon path is shown, and the shaded arc shows the decaying density of photons at increasing distances from the PDW source.

The photons that do reach a particular receiver follow a smaller number of paths, concentrated along a path which is usually banana-shaped as shown in Fig. 2 (middle). The boundaries of the banana path are ill defined, with a few photons passing well beyond the nominal boundaries shown. Among these few photons are some which pass through the anomaly (e.g. a tumor) shown heavily shaded, and can carry information about it. The width of the banana path is determined by the PDW source-receiver separation and its depth depends on the optical properties of the medium and the overall geometry (typical depth is 1-3 cm although longer paths are possible).

Each inactive PDW source, shown in light gray, becomes active in turn until all n_t have been activated. Each of the n_r receivers is sampled simultaneously. Thus while there are a total of $n_t n_r$ banana paths [DiMarzio, 1999], many of them carry little or now useful information because of the limited PDW source positions available (widely separated pairs have very weak signals, very closely separated pairs sample almost entirely along the surface). This leads to an ill-posed inversion problem.

In API, Fig. 2 (right), the very small amount of light which passes through the focus of the ultrasound is further modulated by the ultrasound. The scattered light energy is thus formed at the acousto-optic sum and difference frequencies, and is therefore easily discriminated from the stronger signal along the conventional DOT banana path using narrow-band electronic filtering. By creating virtual sources located inside the body, API allows us to sample a variety of different paths not accessible in traditional DOT and the problem becomes less ill-posed. The size of these virtual sources is proportional to the size of focus of the ultrasound transducer.

3 EXPERIMENTAL APPARATUS

Our experimental setup is shown in Fig. 3. A 70 MHz oscillator serves as the local oscillator signal for a pair of demodulators. An HP 8647A tunable RF source set to 72.3 MHz is split using a tee; one arm of the tee is used to modulate a small solid-state laser and the other arm of the tee is connected to the first demodulator. The modulated laser light is sent down a plastic

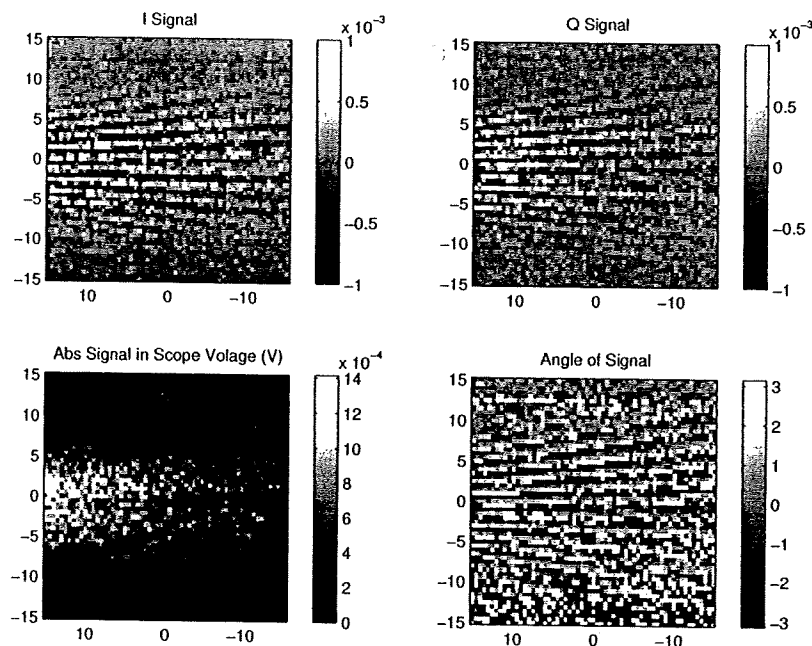


Figure 4: Sample data run, homogeneous conditions. The top figures are the inphase (left) and quadrature (right) components of the signal. The lower-left figure is the amplitude of the signal and the lower-right figure its phase. The X and Y axes give the detector displacement (in mm) from the zero position (arbitrarily determined). Signal strength (in volts) is indicated by color.

4 EXPERIMENTAL RESULTS

The detector fiber is attached to a mechanical arm, moving in a plane normal to the line containing the laser source and the ultrasound focus. The arm is controlled by a PC and can be moved with sub-millimeter precision. In this way we are able to map out the acousto-optical field within the entire plane.

Using our experimental setup, we are able to detect modulated signals in both homogeneous (Fig. 4) and inhomogeneous (Fig. 5) media. These figures were generated in transmission. Work on signals acquired in a reflective configuration is on-going.

5 CONCLUSIONS

We have demonstrated a new technique for combining DOT with focused ultrasound known as acousto-phonic imaging. In acousto-phonic imaging, an ultrasound field modulates the applied optical diffusive wave allowing light to be “tagged” as having passed through a well-defined region space. This acousto-optical signal appears at the sum and difference frequencies which allows for easy discrimination between modulated and unmodulated signals. In effect, the focus of the ultrasound creates a “virtual” source of diffusive waves located *inside* the body.

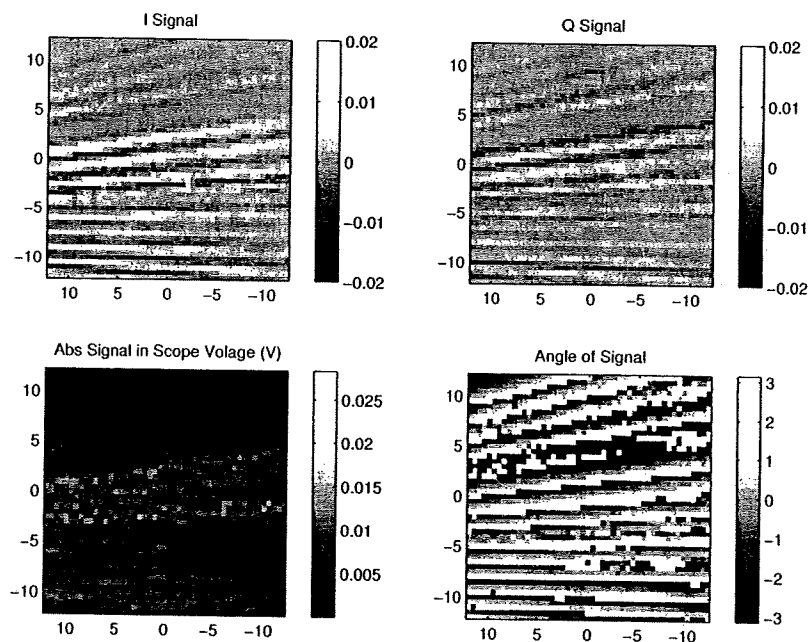


Figure 5: Sample data run with optical inhomogeneity. The geometry was the same as that of Fig. 4.

We have demonstrated experimentally that this is a viable imaging technique using a Titanium Oxide tissue phantoms. The experimental images (with and without synthetic inhomogeneities) show periodic variations in the phase of the signal with a spatial period comprable to the wavelength of the ultrasound source.

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