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Quantitative Photoacoustic Tomography in Biological Tissues Assisted with Intrinsic Acoustic Measurements

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Abstract: Purely acoustic information can be extracted from photoacoustic measurements that can be used afterwards as prior knowledge in the reconstruction process. The fact is illustrated through experiments conducted on biological phantoms and on synthetic data.

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1. Introduction

The initial purpose of PhotoAcoustic Tomography (PAT) is to obtain, by the resolution of an inverse problem, the optical absorption coefficient spatial distribution of heterogeneously absorbing media, such as biological tissues. These tissues being heterogeneous both in terms (optical) absorption and scattering, quantification of this absorption coefficient can be achieved if the scattering process is taken into account. Multiple sources illumination [1–3] is a method to achieve Quantitative Photoacoustic Tomography (QPAT) by discriminating optical absorption and diffusion: the medium is actively probed, locally at different sources positions in order to sense the different responses of the medium.

Furthermore, local illumination generates near the interface a primary photoacoustic wave that propagates from the region close to the surface through the medium. Usually, one avoids to measure it, for example by placing the transducers in an orthogonal direction, because this signal can hide or at least pollute the photoacoustic signal coming from the abnormality of interest. However, this primary acoustic wave is going to be sensitive to any acoustic perturbations present in the medium. With local sources illumination, the interesting point is that the detection of this acoustically sensitive signal and the photoacoustic signal is detected within two separate times intervals.

In the present work, examples of measurements of this primary signal are performed on different phantoms. The filtering protocol is detailed and the motivation of using the acoustic echoes measurements as prior knowledge is illustrated through reconstructions performed on synthetic phantoms.

2. Fast localization of embedded objects introduced as a priori constraints in reconstructions of optical maps

The experimental setup (Fig. 1) is composed principally of a Nd:Yag pulsed laser at 1064 nm (Quantel, France, 330 mJ, 5 ns, 10 Hz). The energy, monitored by an attenuation system (rotating half wave plate and a polarizing cube) is thus reduced to 10 mJ and the frequency is doubled with a 8 mm³ KTP crystal in order to produce 532 nm optical pulses (energy left 3.6 mJ). The optical beam is driven through a periscope to the studied phantom, immersed into water for acoustic impedance matching. A cylindrical concave lens is used to produce a large illumination on the phantom in one dimension (35 mm), keeping the second one practically unchanged (5mm), leading to a maximal fluence estimated less than 3.5 mJ.cm⁻². The generated acoustic waves are collected with immersion acoustic transducers (diameter 10 mm, focal length 1", central frequency 3.5 MHz, Olympus, France). The collected signal is then preamplified (Olympus, France, gain 40 dB, cutting frequency 10 MHz) and driven to an oscilloscope (Tektronix, USA, 4 channels, bandpass with 200 MHz) for signal analysis. Sound and light speeds being so different, the optical fluence is going to be deposited instantaneously within the whole medium according to the spatial distribution of the absorption and the

diffusion coefficients. The deposited energy is dissipated into heat, microdilations are produced that give birth to a pressure wave that can propagate through the medium.

If the medium is optically and acoustically homogeneous, an acoustic wave pressure echo is going to be measured due to the deposition of energy, mostly near the illuminated surface because of the high level of scattering. Hence the time-of-flight of this echo corresponds mostly to the distance between the illuminated surface and the detector position. This measure can firstly serve as a rough estimate of the speed of sound in the medium. Note that in measurement shown in Fig. 1, the measurements were performed at 90° from the source illumination. Due to transducers directivity and wide field illumination, the surface signal is barely visible under this configuration. This pressure wave is propagating and probing the medium as a conventional acoustic wave. If acoustic heterogeneities are present in the medium, for example because of a different density or structure as in tumors (example mammary tumors visible with US echography), then this primary photoacoustic wave is going to be perturbed, and echoes (Fig. 1 (c)) related to the presence of acoustic heterogeneities will be sensed at longer times-of-flight. If optically absorbing heterogeneities are present within the medium, more energy will be absorbed locally giving birth to photoacoustic echoes (Fig. 1 (b)) arriving at a shorter times-of-flight compared to the above mentioned photoacoustic echo generated at the interface. Fig. 1 shows an example of such measurement performed on a homogeneous cylindrical phantom (Agar-agar 2%) with a black hair located in the center.

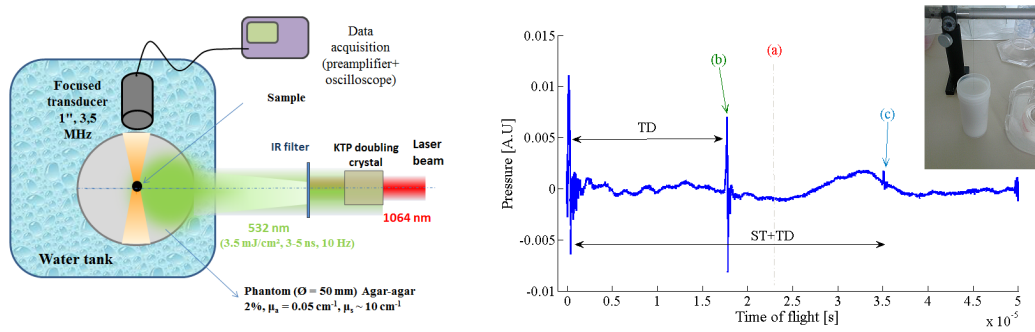


Fig. 1. Left, schema of the set up; Right: Example of acoustic measurement of a black hair inserted on a cylindrical Agar-agar 2% homogeneous phantom (diameter 5 cm, see insert), placed at the center of the phantom, angle between source and transducer, 90° .

In order to exploit this secondary acoustic echo, one has to clearly discriminate the two species of echoes, the photoacoustic ones from the acoustic ones. The two species of echoes belong to two different time spaces delimited by the position of the interface: (i) photoacoustic echo (b) localizes an object T belonging to a circle \mathcal{C} of radius $\overline{TD} = t_{(b)}/v_s$; (ii) acoustic echo (c) localizes an object T belonging to an ellipse \mathcal{E} of foci the position of the detector D and the projection of the laser beam at the surface of the phantom such that $\overline{ST} + \overline{TD} = t_{(c)}/v_s, /v_s$ being the sound velocity. Hence, the object T belongs to the intersection of \mathcal{C} and \mathcal{E} that occurs inside the phantom area. For multiple objects localization, one has to multiply the measurements and plot all possible occurrences. Here is illustrated the process with synthetic data computed by solving the multiphysics forward model for QPAT (optical diffusion and acoustic propagation equations) by considering the 2D disk phantom (see description on Fig. 2): the background is homogeneous ($\mu_{a0} = 10 \text{ m}^{-1}$, $D_0 = 3.33 \times 10^4$, $v_{s0} = 1485 \text{ m.s}^{-1}$, density $\rho_0 = 1000 \text{ kg.m}^{-3}$) and contains two abnormalities, one disk (1: $\mu_{a1} = 11 \text{ m}^{-1}$, $D_1 = D_0$, $v_{s1} = v_{s0}$, density $\rho_1 = 1500 \text{ kg.m}^{-3}$) and one square (2: $\mu_{a2} = 11 \text{ m}^{-1}$, $D_2 = 3.03 \times 10^4$, $v_{s2} = v_{s0}$, density $\rho_2 = 1500 \text{ kg.m}^{-3}$). From the sinograms, an algorithm of peak detections is applied [5], and the spatial histogram of probabilities of intersections between circle \mathcal{C} and \mathcal{E} is calculated. The gradient of the resulting image is taken to obtain enhanced contours and a final thresholding allows rejecting artifacts due to sources positions (Fig. 2, 1st row, last column).

A minimization based approach has been adopted for the resolution of the inverse problem, in which it is sought to find (μ_a, D) which minimizes the error functional $F = \frac{1}{2} \sum_s \int_{\Omega} (H_s^m(\mathbf{r}) - H_s(\mathbf{r}, \mu_a, D))^2 d\mathbf{r}$, where H_s^m is the measured initial pressure distribution map due to source s , that can be obtained from the measured pressures, after resolution of the acoustic propagation inverse problem. Here, to test the feasibility, we considered an ideal initial pressure distribution map $H_s^m = \Gamma \mu_a \phi_s$, obtained from the multiplication of the Grueneisen coefficient Γ (kept constant, here), the absorption coefficient distribution map and the fluence deposited by source s calculated by solving the diffusion

equation. The reconstructions of the optical parameters have been performed with an adjoint-assisted gradient-based method [4]. The reconstructions obtained with and without use of the localization map are shown on Fig. 2 (two last rows).

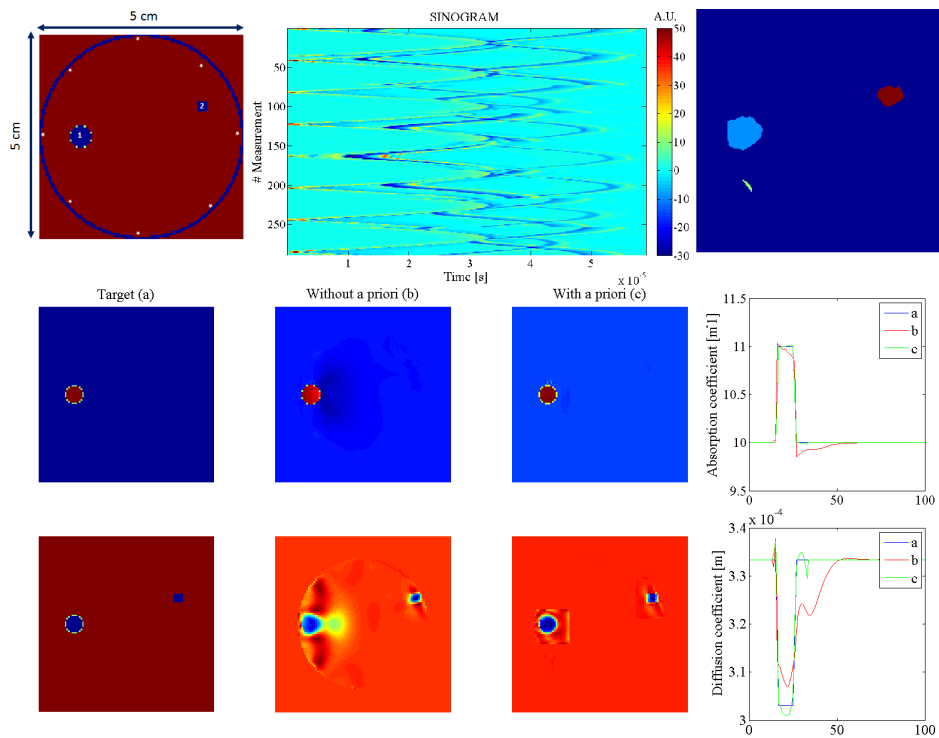


Fig. 2. 1st row: schematic of the numerical phantom, homogeneous disk ($\varnothing=5\text{cm}$), containing two small inhomogeneities (1: disk $\varnothing=0.5\text{cm}$; 2: square $0.25\times 0.25\text{cm}^2$), sinogram of the measurements (8 sources and 36 detectors positions), map of occurrences of intersections between circles \mathcal{C} and ellipses \mathcal{E} after post-treatment; 2nd row: absorption coefficient maps μ_a (target (a), without (b) and with (c) a priori constraints, and horizontal cross-plot passing through the inhomogeneity; 3rd row: same as 2nd row for diffusion coefficients D .

The results clearly exhibit reconstructions of much higher quality, especially for the reconstruction of the diffusion coefficient. The robustness of the approach on experimental data will be presented.

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