Acousto-optical imaging techniques in optically scattering media : towards tumors detection

M. Atlan[‡,*], F. Ramaz[‡], B. C. Forget[‡] and A. C. Boccara[‡]
[‡]Laboratoire d'Optique Physique, ESPCI, Université Paris VI, France
[*]Email: atlan@optique.espci.fr

Abstract

Acousto-optic imaging in (several centimeters) thick scattering media is a hybrid method combining both acoustics and optics in order to make tomographies. Our purpose is to reveal the local optical properties inside biological tissues. The use of light is motivated by the existence of optical contrasts between healthy and tumorous areas. But contrarily to ultrasound, light is highly scattered within biological tissues, and thus direct imaging cannot be performed through more than a few millimiters thick samples. We have chosen at the *Laboratoire d'Optique* to use an hybrid alternative to the purely optical technique: it combines light, which gives the contrast, and ultrasound longitudinal waves, which provide the resolution. We measure the quantity of transmitted light modulated at the ultrasound frequency with a CCD camera, used as an array of multidetectors, to increase the signal-to-noise ratio. This quantity is related to the local optical properties of the acoustic focal region.

We show that a linear sweep of the modulation frequencies (chirp) of the light and acoustic sources enhances the resolution by reducing the size of the Point Spread Function of our imager along the acoustic propagation axis. This concept is illustrated with images of optically absorbing regions inside phantoms and biological tissue samples. This achieves our first step towards a study in clinical environnement.

Introduction

The optical path of light crossing the volume in which the u.s.are focuseed is modulated (at a frequency of 2 MHz), and so is the intensity of the output speckle pattern : as a consequence, the amount of modulated light will depend on the local optical properties. A 3-D reconstruction of the object is obtained by a $x-y-z$ scan of the acoustic focal region. Our acoustic source induces an interaction volume close to a cylinder of 2 mm in diameter and 20 mm in length. The resolution of images perpendicular to the u.s. (x, y) plane have a millimeter size resolution (i.e transverse dimensions of the u.s.), while the situation is not so precise along the u.s. propagation axis (z) because the measured contrast depends on the ratio of the object size to the u.s. one (this is known as partial volume effect).

Acousto-optical tomography has already been carried out by several groups. Firstly, a recent improvement consisted in using multiarray detectors (CCD cameras) : each grain of speckle (random phase) is imaged onto a pixel in order to reduce the averaging effect of the camera and obtain more signal [1]. In this situation, one needs to modulate the light source close to the u.s. in order to translate the Fourier spectrum of the resulting beat within the camera bandwidth. Secondly, a good axial resolution has been obtained with a linear sweep of the u.s. frequency (chirp modulation), because in this case, the Fourier spectrum of the signal contains frequencies that are linearly related to the z-position along the u.s. [2] [3]. Finally, a recent study showed that the size of the virtual source could be reduced by a factor of 6 with a non linear (second harmonic) detection of the acousto-optical signal [4].

We have so far combined the first and the second approaches in recording a film of the chirp sequence, and taking its Fast Fourier Transform (FFT) in order to obtain at once the optical transparency along slices of the u.s beam [5].

Setup and Experiment

Setup

A several centimeter-thick sample under measurement is held between two tubes terminated with optical windows (transparent in the near IR) of 25 mm of diameter.

This sample is illuminated by a single mode axial source of 850 nm laser light provided by a 200 mW (maximum power) diode (SDL Corporation), over the surface of a first optical window (25 mm). It is immersed in water, in order to ensure a good coupling with an ultrasonic wave coming from a broadband PZT (Panametrics emitter, 4 cm output diameter), with a 2.25 MHz maximum-yield frequency, 70 mm of focal distance, 2 mm diameter focal spot, working with a peak pressure (measured) of 1 MPa at the focal point.

Two numeric synthesizers (NS1 & NS2) are used to provide modulation signals to the acoustic (PZT) and light (laser diode) sources, while a third synthe-

sizer (NS3) is used to generate a reference signal to start each chirp sequence.
Output light from the object plane is sampled by a $256 \times 256 \times 8\text{bit}$, 200 Hz bandwidth CCD camera (DALSA Corporation, CA-D1), working at rates from 100 to 160 Hz.

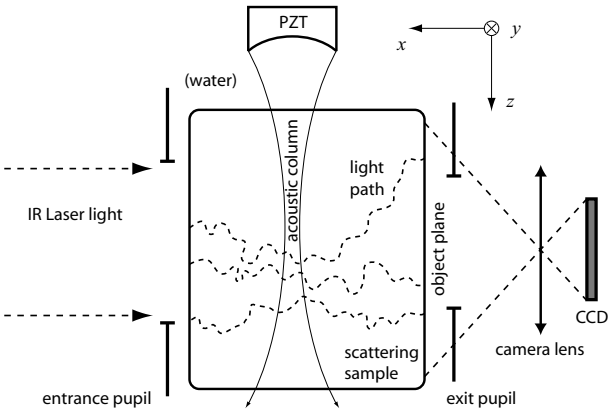


Figure 1: Setup

The chirp method

We used Wang's method [2] [3] to divide the local oscillator (virtual source of modulated light) into slices perpendicular to the sound propagation axis. We realize a heterodyne detection in which the u.s. frequency varies linearly with time. The laser illumination is modulated at the same chirped frequency, which can be shifted freely.
A chirp sequence consists of a linear sweep of the frequencies of NS1 ($f_{\text{u.s.}}$) & NS2 (f_{laser}) with time, typically from 1 to 3 MHz, during τ_{chirp} (approx 1s.) periods.

$$f_{\text{u.s.}}(t) = f^0_{\text{u.s.}} + b\,t \qquad f_{\text{laser}}(t) = f^0_{\text{laser}} + b\,t \quad (1)$$

Knowingly omitting unuseful terms for the understanding of the detection concept, the optical power sampled on each pixel of the CCD during a chirp sequence is proportionnal to [5]:

$$p_{1\text{px}}(t) \propto t(z)\,\cos\left(2\pi\nu t + \phi\right) \qquad (2)$$

Where ν is the apparent frequency of speckle intensity variations;

$$\nu = \Delta f + \frac{b\,z}{v_{\text{s}}} \qquad (3)$$

b denotes the velocity of the chirp (approx. $2\,\text{Mhz.s}^{-1}$); v_{s} is the velocity of the non-rotationnal acoustic wave inside the medium (approximately the speed in water, $1500\,\text{m.s}^{-1}$); $\Delta f = f^0_{\text{laser}} - f^0_{\text{u.s.}}$ is the constant difference between the two chirped frequencies.
Expression 3 shows that the Fourier spectrum of signal $p_{1\text{px}}(t)$ encodes the spatial position of the u.s. wave along its propagation (z-)axis, and is proportionnal to the local optical transmission of the medium
We can obtain the optical information along the acoustic column by computing the time-domain Fourier spectrum (FFT) of the sampling of speckle power (see raw results on figure 2).

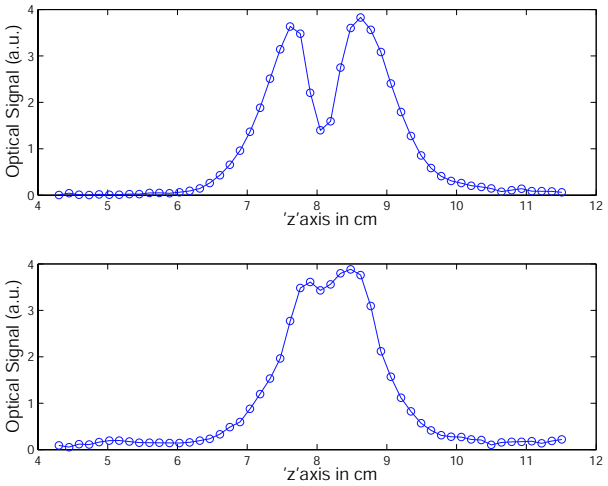


Figure 2: FFT of the time-domain samplings : u.s. positionned over an absorbing inclusion in a gelatin phantom (upper profile), and outside (lower profile).

Experiment

Each frame sequence recorded during a chirp sequence (approx. 1s) for a fixed position of the PZT gives a series of N images (presenting a speckle pattern) of the object plane, which are turned into optical transparency ($t(z)$ quantity, in equation 2) along the z-axis, by signal processing. Raw results are reported in figure 2.
Thus, a 2D tomography is performed with a 1D scan of the PZT (along x $\rightarrow$ figure 5 or y $\rightarrow$ figure 7).
The samples imaged are gelatin phantoms containing a suspension of 200 nm latex spheres to simulate scattering conditions of biological tissue (a reduced scattering coefficient of approximately $20\,\text{cm}^{-1}$, an anisotropy factor of 0.9 and weak absorption). Black coloured cylinders of the same medium (strong light absorption) are included in order to create inhomogenous optical properties in acoustically homogenous samples.
Other tomographies in (y, z) planes have been performed in turkey muscle samples into which we

added black ink with a syringe (see figure 7).

Representing the virtual source of modulated light

Recording the optical signal

Expression 3 shows the linear relation between ν and z, whose coefficients Δf and $\frac{b}{v_s}$ are freely tunable. Thus, information $t(z)$ can be stretched along the restituted frequencies of the FFT profile. In other words, this gives total control over the virtual probe's position and (z) size.
One can consider that this virtual probe is a local source of coherent light blinking (its intensity) at an apparent frequency ν. The "optical signal" is the amount of "blinking" light reaching the CCD.

Visualization of the virtual source of modulated light

In order to visualize the virtual source, we record on the CCD a geometrical image of the object plane. One movie of measurements gives 64k (256×256 pixels) time samplings of independant speckle power variations $p_{1\mathrm{px}}(t)$. These samplings are grouped into packets of typically 2×2 or 2×4 pixels to make "phase-averaged" frequency spectra (FFT of $p_{1\mathrm{px}}(t)$).
Hence we get a collection of frequency profiles. Then, we make cartographies of the amount of modulated light into the object plane (exit window), giving an image for each restituted frequency.
This image is computed for the whole collection of frequencies of the spectrum, resulting in an new frame sequence in which each frame represents the spatial density of modulated light at a selected frequency.
Figure 3 represents several frames of such a sequence, computed in the case where the acoustic column is set next to the object plane, in order to reveal a virtual probe as confined as possible, moving along the acoustic propagation (z) axis, as ν increases. This experiment shows evidence of relation 3 between the apparent frequency of modulation of the speckle (ν) and the location (along z) of the virtual source of modulated light inside the medium.
The theoretical z-size of this source is determined by the sampling conditions. Figure 4a is computed in the following case : $N = 128$ images, $\nu_{\mathrm{cam}} = 100\,\mathrm{Hz}$ (camera BW), $b = 1.43\,\mathrm{MHz.s}^{-1}$. Thus, we expect a size of $\delta z = 0.8\,\mathrm{mm}$ (equation 3) along the acoustic propagation axis, which is confirmed by experiment (see figure 4b). This millimetric-sized probe is used to perform high resolution and contrast images presented in the next section.

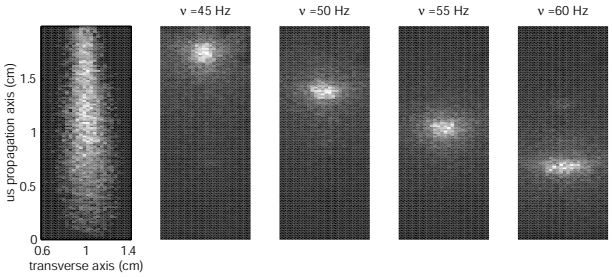
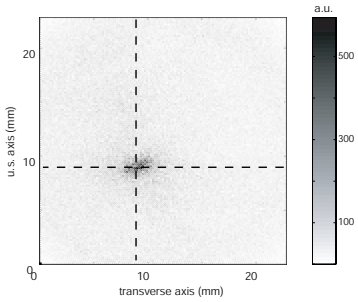
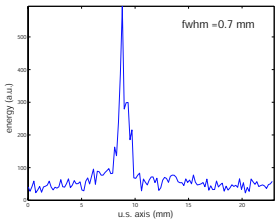


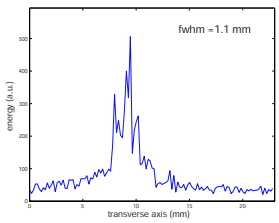
Figure 3: Virtual sources of modulated light. Left : full spectrum AO source . Next figures : images at several apparent frequencies ν of time-sampled optical power of speckle (computed images)



(a) Virtual source of modulated light at 40 Hz



(b) Slice along the acoustic column (z)



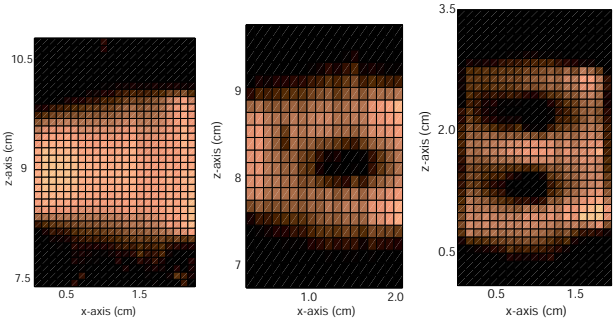
(c) Slice along the transverse (y) axis

Figure 4: Characteristics of the virtual source of modulated light in the object plane (computed image)

Tomographies

Tomographies shown in figure 5 have been generated by stacking 1D-profiles of raw results (represented in figure 2). Experiments are performed on 2 cm-thick gelatin phantoms described in section . These (x, z) tomographies are obtained in blank phantom (figure 5a), phantom with one (figure 5b), and two (figure 5c) added, black ink coloured, cylinders. These cylinders present no acoustical contrast, as shown on figure 6, representing an *echo B* in a (x, z) plane inside the second sample (containing

two inclusions).
Figure 7 represents two (y, z) tomographies performed at the same location, in the same medium (2 cm-thick turkey muscle), at several moments in time. Reference for normalization is an image captured in a fresh, blank sample. Figure 7a is obtained after a first injection of black ink with a syringe, and figure 7b after a second injection in the same area. These figures show images obtained after normalization by the blank image.



(a) blank phantom (b) one $\phi = 4.0\,\text{mm}$ inclusion (c) two $\phi = 4.0\,\text{mm}$ inclusions

Figure 5: 2D-tomographies into (x, z) planes (raw results)

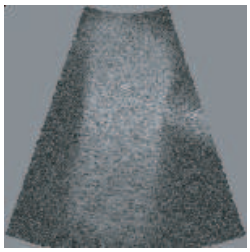
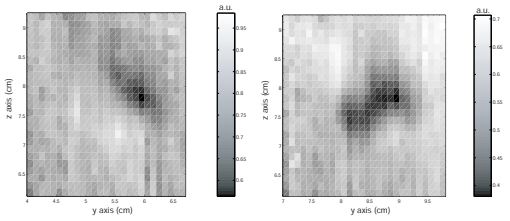


Figure 6: Echo B in a (x, z) plane inside a phantom containing two optical inclusions

Conclusion

The described imaging technique is a fast, non harmful way of performing IR tomographies in biological tissue. It is based on acousto-optical interaction into scattering media, under coherent illumination, and produces images of optical transmission which varies both with absorption and diffusion.
The signal processing needed to extract the information involves only simple 1D FFTs.
In the direction of the u.s. propagation axis, this resolution is improved by the detection technique.



(a) after first ink injection (b) after second ink injection

Figure 7: 2D-tomographies into (y, z) planes (turkey meat). Normalized raw results

In the transverse direction, it is given by the acoustic part of the imaging system : the size of the focal spot. Resolution is then millimetric in all directions , as we have demonstrated. This induces a better contrast power for the tomographies, as the apparent volume of acousto-optical interaction is reduced. We demonstrated the possibility of performing imaging of tracer propagation into a static scattering medium with this technique.

References

[1] S. Lévêque, A. Boccara, M. Lebec, and H. St-Jalmes, "Ultrasonic tagging of photon paths in scattering media : parallel modulation processing," Optics Letters, vol. 24, pp. 181–183, 1999.

[2] L. V. Wang and G. Ku, "Frequency-swept ultrasond modulated optical tomography of scattering media," Optics Letters, vol. 23, pp. 975–977, 1998.

[3] G. Yao, S. Wang, and L. V. Wang, "Frequency-swept ultrasond modulated optical tomography in biological tissue by use of parallel detection," Optics Letters, vol. 25, pp. 734–736, 2000.

[4] J. Selb, L. Pottier, and A. C. Boccara, "Nonlinear effects in acousto-optic imaging," Optics Letters, vol. 27, pp. 918–920, 2002.

[5] B. Forget, F. Ramaz, M. Atlan, J. Selb, and A. Boccara, "High contrast fast fourier transform acousto optical tomography of phantom tissues with a frequency-chirp modulation of the ultrasound," Applied Optics, vol. 42, no. 7, pp. 1379–1383, 2003.