Non-contact photoacoustic tomography and ultrasonography for tissue imaging
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Abstract: The detection of ultrasound in photoacoustic tomography (PAT) and ultrasonography (US) usually relies on ultrasonic transducers in contact with the biological tissue. This is a major drawback for important potential applications such as surgery and small animal imaging. Here we report the use of remote optical detection, as used in industrial laser-ultrasonics, to detect ultrasound in biological tissues. This strategy enables non-contact implementation of PAT and US without exceeding laser exposure safety limits. The method uses suitably shaped laser pulses and a confocal Fabry-Perot interferometer in differential configuration to reach quantum-limited sensitivity. Endogenous and exogenous inclusions exhibiting optical and acoustic contrasts were detected ex vivo in chicken breast and calf brain specimens. Inclusions down to 0.5 mm in size were detected at depths well exceeding 1 cm. The method could significantly expand the scope of applications of PAT and US in biomedical imaging.

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1. Introduction

Biomedical imaging techniques are presently intensively developed [1]. Photoacoustic tomography (PAT), which provides images of the optical absorption contrast, holds promise for many biomedical applications [2,3]. Ultrasonography (US) is a well established imaging modality providing acoustic properties of tissues. Both PAT and US usually rely on ultrasonic transducers in contact with the tissue using a coupling fluid (water or gel). Unfortunately, a physical contact is not suitable for many applications [4]. For example, during brain surgery, the transducer array must be inserted in a sterile sleeve and the sterile saline solution used as coupling medium requires a horizontal working plane. Most extracorporeal applications can use coupling fluids but some, such as burn diagnostic, cannot. Applications related to ophthalmology may also benefit from a non-contact detection of acoustic signals [5]. For small animal imaging, immersion in water significantly complicates the procedure [6]. Consequently, non-contact optical detection of ultrasound in biological tissues is of great interest. Moreover, generation and detection of ultrasound by remote optical means would facilitate endoscopic implementations of PAT and US as well as compatibility with other imaging modalities such as optical coherence tomography (OCT) [2,5,7]. Air-coupled transducers [8] have been considered for PAT but their limited sensitivity could be difficult to overcome, especially when spatial resolution is needed. Attempts have been made to replace piezoelectric transducers by optical means but most of these attempts require a contact with the tissue [9–11].

Laser-ultrasonics (LU) is a well established optical technique allowing non-contact generation and detection of ultrasound [12]. LU is mostly applied to industrial materials such as metals, plastics and polymer-matrix composites. Ultrasound is laser generated by thermoelastic expansion or by slight ablation at the surface of the material and acoustic reflections coming back to the surface are detected using a laser beam reflected/backscattered by the material. The reflected/backscattered light, which is phase modulated by the surface displacement, is demodulated with a large throughput interferometer such as a confocal Fabry-Perot interferometer (CFPI) [13,14] or a photorefractive interferometer [15,16]. High sensitivity is obtained by using a high power detection laser and operating in a shot-noise-limited detection regime. The challenge in applying the optical detection scheme used in LU to biomedical imaging consists in obtaining an acceptable sensitivity without exceeding laser exposure safety limits [17]. Non-contact PAT has been recently demonstrated on materials [18] but, to our knowledge, not on biological tissues for which the optical properties can differ significantly and laser safety limits cannot be ignored.

In this work, the optical detection scheme used in industrial LU is adapted to the safe detection of ultrasound in biological tissues. The technique does not require any contact, coupling medium or surface preparation. Non-contact PAT and US (NCPAT and NCUS) are implemented using a single apparatus, the difference between both imaging modes being found in the reconstruction algorithms. This first demonstration is based on three strategies. First, to use a safe level of laser radiation by shortening high-energy detection laser pulses to the propagation time of ultrasound. Second, to reach a quantum-limited sensitivity using an
optical phase demodulator in differential configuration. Third, to approach a diffraction-
limited image reconstruction by sensing remotely the surface profile of the tissue. Those
strategies allow a safe detection of inclusions exhibiting optical and acoustic contrasts.
Inclusions down to 0.5 mm in size located at depths well exceeding 1 cm were detected \textit{ex vivo}
in chicken breast specimens. Promising preliminary results have also been obtained \textit{ex vivo}
with calf brain specimens. The technique is expected to be applicable \textit{in vivo} on
beating/moving internal organs or skin.

2. Experimental setup

Using laser beams to generate and detect ultrasound in biological tissues immediately leads to
opposite requirements. First, the sensitivity of the measurement can only be increased by
using a higher power of laser radiation. Second, the laser power is limited by safety limits,
namely, the maximum permissible exposure (MPE), to avoid damaging the biological tissue
[17]. Those opposite requirements necessitate an appropriate laser exposure management.

Efficient laser generation of acoustic waves is obtained with sufficiently short laser pulses
compared to thermal and stress relaxation times [1]. Typically, few nanosecond pulses
produced by Q-switched lasers are appropriate. In practice, the tissue must be exposed
repetitively on a large surface area to obtain a uniform illumination throughout the tissue and
the generation laser pulse energy is limited by the MPE for a repetitive exposure [17]. The
laser exposure management is thus mostly related to the detection laser.

2.1. Detection laser

When probing soft tissues at depths limited to 1 or 2 cm, propagation delays of ultrasonic
waves are below 20 µs since the propagation velocity is about 1.5 mm/µs in soft tissues. In
order to minimize laser exposure, the detection laser illumination should be limited to the
ultrasound propagation delay. For most industrial applications, a master oscillator power
amplifier (MOPA) emitting powerful long pulses is appropriate. Such a laser source can be
adapted to biomedical applications by limiting the pulse duration without losing energy in the
laser system. This is done by adding an intensity modulator (IMOD) between the single-
frequency continuous wave (cw) master oscillator and the following pulsed power amplifier.
A schematic diagram of such a laser source is shown in Fig. 1a.

The IMOD allows inhibiting the amplifier output at the beginning of each pumping flash
by setting its transmission to zero. The energy then builds up in the amplifier laser rods (LRs).
Before the end of each pumping flash, the transmission of the IMOD is increased gradually to
release the stored energy and obtain the desired pulse duration and shape at the amplifier
output. Without pulse shaping (Fig. 1b), the output pulses at the exit of the pulsed power

Fig. 1. (a) Layout of the detection laser. MO, Nd:YAG cw master oscillator; ISO, optical
isolator; HW, half-wave plate; PBS, polarizing beam splitter; AM, amplitude modulator;
IMOD, intensity modulator; TFP, thin-film polarizer; LR, Nd:YAG laser rod; FL, flashlamp;
QW, quarter-wave plate. Other components are plane dielectric mirrors. (b) Pulse shape at the
output of the detection laser without pulse shaping. (c) Pulse shape tailored for the experiment
using the IMOD. The generation laser pulse occurs at \( t = 0 \) µs (green lines in b and c).
amplifier have an energy of 35 mJ, a peak power of 280 W and a duration of 110 μs (FWHM: full width at half maximum). With pulse shaping (Fig. 1c), the output pulses tailored for this experiment have an energy of 14 mJ, a peak power of 560 W and a duration of 24 μs (FWHM). Reducing the pulse duration thus increases the available peak power. More importantly, the peak power for a given energy is about 5 times higher with the tailored pulses. We have chosen using a tailored pulse shape with an increasing instantaneous power for two reasons. First, this eliminates strong signal oscillations at the output of high-pass filters preceding signal digitization. Second, the sensitivity increases with time to facilitate the detection of weaker signals arriving at larger time delays (coming from larger depths in the tissue).

2.2. Scanning and detection systems

A schematic diagram of the scanning setup is shown in Fig. 2a. The generation laser (not shown) is a Q-switched laser emitting 5 ns pulses at 532 nm with a repetition frequency of 10 Hz. The generation laser beam (shown in green) is directed toward the biological tissue specimen after beam expansion and transmission through a line mask blocking the generation laser on the detection scan line. The purpose of the line mask was to eliminate any overlap between the detection and generation laser illumination during the entire scan to increase the allowed laser power (see also Subsection 3.2). This allows using the single shot safety limit for the detection laser and the repetitive exposure safety limit for the generation laser. The detection laser beam (shown in red) is transmitted to the setup with a multimode optical fiber. A sample of the detection laser beam is first deflected by a beam splitter (BS) and coupled into an optical fiber as a reference beam for phase noise elimination. The detection laser beam is then focused on a 400 μm diameter spot on the tissue specimen. A small prism mirror (P) in front of the collecting optics is used to direct the detection laser beam to the tissue without coupling stray light into the collecting optics. The resulting illumination/detection geometry is shown in the inset (Fig. 2a) where the elliptic green area is the generation laser spot (masked on the scan line) and the red area is the detection laser spot. The collecting optical system uses two lenses assemblies (LAs) with focal lengths of 33 mm for collimating and coupling the reflected/backscattered light (shown in yellow) into the optical fiber (400 μm core diameter, 0.39 numerical aperture) carrying the signal beam. The biological tissue specimen is located on a computer-controlled motorized stage allowing displacements along the scanning axis $x$ and the focusing axis $z$. The working distance between the biological tissue and the collecting optics was about 25 mm.

The phase demodulation of the collected light is done with an actively stabilized CFPI mounted in differential configuration to achieve both intensity and phase noise reduction (Fig. 2b) [19]. The horizontal polarization is used to analyze light from the tissue specimen (signal beam) and the vertical polarization is used to measure the phase noise of the detection laser (reference beam). The CFPI is a 1 m long cavity with mirrors’ reflectivity of 94.5%. The back mirror is mounted on a piezoelectrically actuated mount (PZM) fed by a 10 Hz servo loop ensuring stabilization at half-maximum of the carrier frequency transmission. The optical system inertia ensures a good stability within each measurement time window of about 20 μs. Such a CFPI is thus already compatible with in-vivo measurements. The calculated demodulation response of the CFPI is shown in Fig. 3 [14]. The peak demodulation response is around 2 MHz and remains significant between 400 kHz and 9 MHz. For both reference and signal beams, a differential detector (DD) was used to remove intensity noise (Fig. 2b). Each DD uses two InGaAs photodiodes in series DC-coupled to a transimpedance amplifier followed by a high-pass filter (300 kHz cutoff frequency) and a voltage amplifier. Phase noise of the detection laser was eliminated by the active filtering explained as follows. First, the amplitude of the signal channel before the generation laser pulse was multiplied to obtain an rms noise equal to that of the reference channel in the same time window. Then, iterations around this first value have allowed determining the optimum amplitude multiplication factor to minimize the residual noise on the difference between both channels. Typically, the residual noise was comparable to the shot noise level calculated from nominal values of the
transimpedance gain and InGaAs photodiodes’ quantum efficiency. The spectrum of the residual noise was practically white in the frequency range kept for the analysis (0.5 to 3 MHz). The active filter also included a fixed time delay between both channels to account for any electronic delay in the setup.

Fig. 2. (a) Schematic diagram of the biological tissue scanning setup: LA, lenses assembly; L, lens; M, plane mirror; BS, beam splitter; P, gold coated hypotenuse prism. (b) Schematic diagram of the CFPI in differential configuration: PBS, polarizing beam splitter; PZM, piezoelectrically actuated mirror mount; VA, variable attenuator; DD, differential detector.

Fig. 3. Calculated amplitude (blue) and phase (red) of the demodulation response of the CFPI.

2.3. Remote surface profilometer
The surface profile was measured prior to each scan using a slight modification of the scanning setup (Fig. 2a). A mechanical optical beam chopper was temporarily inserted between the signal beam optical fiber and the adjacent lenses assembly. A 5 mW green laser pointer was injected into the other end of the signal beam optical fiber. The laser beam backscattered by the tissue was collected back into the signal beam optical fiber and detected with an amplified Si photodiode followed by a lock-in amplifier. For each lateral position $x$, the vertical position $z$ was scanned (100 µm step size) over about 3 mm and the position of the maximum signal after a Gaussian fit with the $z$-scan intensity profile provided the position of the surface with an uncertainty of about 100 µm.

3. Methodology
3.1. Image reconstruction
A typical image is obtained from 60 to 80 A-scans, each measured with a single detection laser pulse (no averaging). All A-scans are grouped in a raw B-scan image which is processed with the synthetic aperture focusing technique (SAFT) [20]. The SAFT algorithm was applied in the time domain and modified to take into account the surface profile of the biological tissue. When measuring remotely on curved surfaces with non-negligible height variations...
compared to the minimum ultrasonic wavelength (about 0.5 mm at 3 MHz), the surface profile is an essential input for a reliable image reconstruction. In our experiment, the surface profile was measured prior to the scan (Subsection 2.3). Each raw B-scan image, once linked with the surface profile, contains the necessary information for both NCPAT and NCUS reconstructions; the only difference being related to the expression of the propagation delay as schematically shown in Fig. 4.

In the case of NCPAT (Fig. 4a), the optically scattering tissue is considered to have a low background absorption at the generation laser wavelength, except for optically absorbing inclusions such as blood vessels. In this case, the thermoelastic generation occurs on each inclusion with a strength related to the product of the local laser fluence and the absorption coefficient of the inclusion. Ultrasonic waves then propagate up to the surface where they can be detected. In the case of NCUS (Fig. 4b), the combination of the weak endogenous optical absorption at the generation laser wavelength with the high generation laser light fluence at the surface of the biological tissue (or close to it) is considered to be sufficiently large to produce an acoustic wave by thermoelastic generation. The unprepared surface of the tissue is thus the source of a probing ultrasonic wave which has an initial waveform following the surface topography. Ultrasonic reflections produced by inclusions exhibiting an acoustic impedance mismatch with the surrounding tissue go back to the surface where they can be detected. Consequently, the technique is intrinsically bimodal; the same data being used for both processed images. The only difference is the temporal delay used for the image reconstruction: the one-way propagation delay is used for NCPAT and the two-way propagation delay is used for NCUS.

![Fig. 4. Reconstruction methods. (a) One-way path used in NCPAT imaging mode. (b) Two-way path used in NCUS imaging mode. Same color code as in Fig. 2.](image)

After applying the active filter (Subsection 2.2), each A-scan was numerically filtered with a third order Bessel-type bandpass filter using typical cutoff frequencies of 0.5 and 3 MHz. With a lateral step size of 400 µm, using frequencies higher than 3 MHz only increases the background noise without a significant gain in spatial resolution [21]. Each A-scan was delayed according to the measured surface profile and to the speed of sound in the specimen. The reconstruction grid included interpolation along the axis x to reduce the lateral pixel size to 100 µm instead of the experimental lateral step of 400 µm. The vertical size of the pixel was also set to 100 µm. Consequently, all images were obtained with a pixel size of 100 × 100 µm². For both imaging modes, the time derivative of each A-scan was used to account for the fact that pressure is proportional to the velocity of the surface, not to displacement [18].

### 3.2. Laser safety limits

For the generation laser, the average intensity must be considered since there is a significant overlap of the surface illuminated by the generation laser between successive points of the scan. The illuminated elliptic surface of 4.0 cm × 2.5 cm has an area of 7.9 cm². With 100 mJ pulses at 10 Hz, this corresponds to an average intensity of 130 mW/cm², well below the MPE at 532 nm (200 mW/cm²) [17]. The single-shot MPE (expressed in terms of fluence) can be used at the detection laser wavelength (1064 nm) since there is no overlap between successive points of the scan. The line mask also eliminates any overlap between the detection laser spots and the area illuminated repetitively by generation laser. Assuming a pulse duration t of about
25 µs, the MPE is given by \( C \cdot t^{0.25} = 0.39 \text{ J/cm}^2 \) since \( C = 5.0 \text{ J/cm}^2/s^{0.25} \) at 1064 nm [17]. The 400 µm diameter detection spot corresponds to an area \( A = 1.3 \times 10^{-3} \text{ cm}^2 \). The maximum energy per pulse is thus equal to \( E = A \cdot C \cdot t^{0.25} = 0.5 \text{ mJ} \), which is the pulse energy used in the experiment (after attenuation of the available 14 mJ).

### 3.3. Preparation of tissue specimens and blood vessel phantoms

Chicken breasts purchased in a grocery store were cut parallel to the natural surface to obtain a uniform thickness of about 10 mm. The natural surface of the upper piece was used for laser measurements. In the lower piece of chicken breast, 1 to 3 mm deep incisions were made with a scalpel and gently filled with vegetable oil using a syringe before inserting exogenous inclusions. Vegetable oil was also poured between both pieces of chicken breast for a better ultrasonic contact. Calf brains obtained from a slaughter house were prepared similarly, keeping the natural upper surface of the brain for laser detection.

Blood vessels were simulated with polyester thin wall tubes filled with India ink diluted in water to obtain an absorption coefficient \( \mu_a \) comparable to that of whole blood at 532 nm, which is about 235 cm\(^{-1}\) for oxyhemoglobin and 217 cm\(^{-1}\) for deoxyhemoglobin. Those values were obtained using a whole blood concentration of hemoglobin of 150 g Hb/liter [22]. The polyester tubes were optically transparent at visible and near infrared wavelengths and the nominal wall thickness was 12.7 µm. The negligible wall thickness compared to the shortest ultrasound wavelength considered in the experiment (about 0.5 mm) also ensured a good acoustic transparency.

### 4. Results

Results obtained with a chicken breast specimen are shown in Fig. 5. The measured surface profile has been overlaid (black curve) on both reconstructed images in order to emphasize the fact that the specimen was not planar as expected in most non-contact biomedical imaging applications. White areas above the surface represent the surrounding air.

The embedded objects were chosen to produce different combinations of photoacoustic and ultrasonic responses. Blood vessel phantoms (20 mm long) were prepared to mimic the optical blood absorption coefficient at the generation laser wavelength while minimizing the acoustic impedance mismatch with the surrounding tissue (see Subsection 3.3). Metal wires (20 mm long) were used to provide a strong ultrasonic impedance mismatch. One of them was painted in white to minimize the photoacoustic response while the second one was unpainted to provide both a photoacoustic response and an ultrasonic impedance mismatch. All embedded objects were inserted parallel to each other at a depth of about 15 mm and the scan line was perpendicular to their common orientation.

The NCPAT image (Fig. 5a) and the NCUS image (Fig. 5b) of the same chicken breast specimen clearly exhibits the embedded objects highlighted by arrows and zoomed below their respective image. The 1.0 mm diameter and 0.5 mm diameter blood vessel phantoms (respectively, i and iii) are mainly seen in the NCPAT image. The 0.8 mm diameter white painted metal wire (ii) is only seen in the NCUS image and the 0.7 mm diameter grayish metal wire (iv) is seen in both NCPAT and NCUS images. This demonstrates the capability of the non-contact approach to discriminate photoacoustic and ultrasonic signatures of each object using a single data set. The NCPAT image also clearly exhibit many well defined endogenous absorption sites which should not be confused with noise spikes. In fact, successive measurements with the same specimen have shown a good repeatability of the resulting NCPAT and NCUS images. Faint hyperbolic artifacts seen in the upper part of the NCUS image (Fig. 5b) can be attributed to photoacoustic signals which are, as expected, not properly focused by the NCUS reconstruction algorithm. These artifacts are a consequence of the bimodal character of the non-contact method and the fact that photoacoustic signals necessarily precede ultrasound signals. For the NCUS reconstruction, a proper modeling of the source near the surface and the following non-planar acoustic wavefront propagation will be essential to reach the best resolution and signal-to-noise ratio (SNR) of the method. The lateral amplitude profiles (along \( x \)) extracted for each embedded objects are also shown in Fig.
5 using the same vertical scale (for each imaging mode) to exhibit the relative signal strength and the lateral resolution.

Fig. 5. Images of a chicken breast specimen. (a) NCPAT image obtained with the following embedded objects (respective diameters in parenthesis): i, blood vessel phantom (1 mm); ii, white painted metal wire (0.8 mm); iii, blood vessel phantom (0.5 mm); iv, unpainted greyish metal wire (0.7 mm). (b) Corresponding NCUS image. All scales are in mm except for amplitude profiles (in arbitrary units).

Results obtained with calf brain specimens are shown in Fig. 6. Again, the measured surface profile has been overlaid (black curve) on each reconstructed image. Similar embedded objects were used to test the imaging capability. It is easily seen that brain is more challenging considering the numerous endogenous structures. The NCPAT image shown in Fig. 6a displays all embedded objects highlighted by their respective arrow and zoomed below the main image. The photoacoustic response of the white painted needle (iv) may be due to blood residue accidentally deposited during the preparation or to misidentification with a

Fig. 6. Images of calf brain specimens. (a) NCPAT image obtained with the following embedded objects (respective diameters in parenthesis): i, unpainted grayish metal wire (0.7 mm); ii, blood vessel phantom (1.1 mm); iii, blood vessel phantom (0.7 mm); iv, white painted metal wire (0.8 mm). (b) NCUS image obtained with a second calf brain specimen in which two metal wires were embedded (0.7 mm diameters). All scales are in mm except for amplitude profiles in a (in arbitrary units).
naturally occurring photoacoustic source. The surface of a brain being much more structured than that of a chicken breast, satisfying results were not obtained systematically by applying our two-dimensional (2D) NCUS algorithm based on a one-dimensional (1D) surface profile measurement. Figure 6b shows the NCUS image of two embedded metal wire in a second calf brain specimen. Although the lateral resolution is poor, the signal strength is good and would be much better using a three-dimensional (3D) reconstruction algorithm, a 2D measurement of the surface topography and the numerical propagation of the non-planar acoustic wave probing the tissue.

5. Discussion

5.1. Spatial resolution

Experimental images shown in Section 4 exhibit the embedded objects as expected. However, the extracted lateral profiles do not provide a diffraction-limited sizing since objects with different diameters can give lateral profiles with similar FWHMs: see, for example, profiles (i) and (iii) in Fig. 5a. This is attributed in part to the inhomogeneity of the acoustic speed in real tissues. The surface profile measurement is also perfectible to reduce the uncertainty (of about 100 µm) which was not negligible compared to the minimum ultrasonic wavelength considered in the reconstruction (0.5 mm at 3 MHz). The uncertainty could be significantly reduced by using an OCT system. Combining our system with an OCT system would allow a simultaneous measurement of the surface profile and the raw B-scan used for NCPAT and NCUS imaging. This would also allow real time remote tracking of low frequency surface displacements encountered in applications involving beating or moving living tissues.

The present system was optimized for a 2D reconstruction algorithm. Clearly, a 3D reconstruction would be desirable for applications involving highly structured tissues such as brains. A 3D algorithm would intrinsically increase the SNR by increasing the number of A-scans involved in the reconstruction of each point of the volume. This would allow to further reduce the detection laser energy and to superimpose generation and detection laser beams without reaching the MPE, thus eliminating the necessity of a line mask.

More fundamentally, the thermoelastic generation of the ultrasonic wave near the surface of the tissue and its propagation need to be modeled more carefully in order to implement a NCUS reconstruction algorithm achieving diffraction-limited resolution and sensitivity. An efficient NCUS mode must be based on a 3D reconstruction algorithm and a 2D surface topography measurement. Coupling this non-contact system with an OCT system would provide both a real time measurement of the surface topography and the OCT image near the surface of the tissue. The PAT-OCT combination has already been shown to be useful [2,5,7].

5.2. Sensitivity

According to the obtained experimental images, the sensitivity of the non-contact approach appears sufficient for many applications. However, a more quantitative analysis facilitates the comparison with other methods. The sensitivity can be expressed in terms of the minimum detectable pressure, sometimes called the noise equivalent pressure. For a shot-noise-limited measurement, the root-mean-square (rms) amplitude of the photocurrent fluctuations is

\[ \sigma_i = \sqrt{2eB}, \]

where \( e = 1.6 \times 10^{-19} \) C is the electron charge (magnitude) and \( B \) is the detection bandwidth. The average photocurrent \( I \) is given by

\[ I = \eta \frac{e\lambda}{hc} P, \]

where \( \eta \) is the quantum efficiency of the photodiode, \( \lambda \) is the detection laser wavelength, \( h \) is the Planck constant \( (6.6 \times 10^{-34} \) J·s), \( c \) is the speed of light \( (3.0 \times 10^8 \) m/s) and \( P \) is the incident optical power. According to the demodulation response of a CFPI [14], the rms value of the smallest measurable displacement of the surface is given by
where \( S \) is the amplitude of the demodulation response (see Fig. 3 for the CFPI used in the experiment). The pressure inside the tissue is related to the displacement \( \varepsilon \) of the surface by

\[
p = \frac{Z \varepsilon}{2} \frac{\partial}{\partial t},
\]

where \( Z \) is the acoustic impedance of the medium (about \( 1.5 \times 10^6 \) Pa·s/m for soft tissues). The factor of 2 in Eq. (4) is due to the fact that the surface displacement is twice that of the particle displacement inside the medium (free surface boundary condition). According to Eqs. (1), (3) and (4), the rms noise equivalent pressure is given by

\[
\sigma_p = \frac{Z \lambda}{4S} \sqrt{\frac{2eB}{I}},
\]

where \( f \) is the ultrasonic frequency. At \( f = 2 \) MHz, where \( S = 1 \), and for a detection bandwidth \( B = 3 \) MHz, we find that, in typical experimental conditions (\( I = 10 \) mA), \( \sigma_p = 8 \) Pa or, equivalently, \( \sigma_\varepsilon = 0.8 \) pm. This first implementation of non-contact optical detection appears significantly less sensitive (by about one order of magnitude) than piezoelectric transducer detection when the sensitivity of piezoelectric transducers is calculated by considering only thermo-electrical fluctuations [23].

### 5.3. Extension to in vivo applications

The photon lifetime in the cavity of the CFPI can be estimated as follows. The resonance linewidth of the cavity is given by \( \Delta = \frac{FSR}{F} \) where \( FSR \) is the free spectral range (75 MHz) and \( F \) is the finesse of the cavity (about 28). Consequently, in our case, \( \Delta = 2.7 \) MHz. According to the time-bandwidth product of an exponential decay, we obtain the photon lifetime \( \tau = (2\pi \Delta)^{-1} = 60 \) ns. This value is indicative of the response of the CFPI to speckle decorrelation. The value of \( \tau \) being much smaller than speckle decorrelation time in tissues (1 ms), there is no limitation in using a CFPI for in vivo applications.

### 6. Conclusion

The results presented herein give a first demonstration of NCPAT and NCUS performed ex vivo on soft biological tissues. Sub-millimeter endogenous and exogenous inclusions have been detected at depths well exceeding 1 cm. It is clearly shown that the highly scattering surface of a soft tissue provides a sufficiently strong diffused reflectivity to measure the ultrasonic displacement of the surface while remaining below the MPE. This however requires limiting the pulse duration of the detection laser to the propagation time of ultrasound. Using a differential configuration of the CFPI has also been essential to reach a quantum-limited sensitivity by removing both intensity and phase noise of the detection laser. For NCPAT, the surface profile measurement was used to obtain nearly diffraction-limited images using time-domain SAFT. Preliminary results were also obtained for the NCUS imaging mode but were limited by the 2D reconstruction algorithm and the 1D surface profile measurement.

Although the achieved sensitivity is below that of piezoelectric transducers, the technical benefits of a non-contact method may outweigh the drawbacks of transducer-based detection using a coupling medium. Among possible applications, a multiwavelength version of the present system could provide a fast and spatially resolved evaluation of blood oxygenation which is critical in brain surgery [6]. The need for a coupling medium being removed, a wealth of applications could emerge from the present method.

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