Photoacoustic tomography using integrating line detectors

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Abstract: Photoacoustic Imaging (also known as thermoacoustic or optoacoustic imaging) is a novel imaging method which combines the advantages of Diffuse Optical Imaging (high contrast) and Ultrasonic Imaging (high spatial resolution). In photoacoustic imaging, a short laser pulse excites the sample. The absorbed energy causes a thermoelastic expansion and thereby launches a broadband ultrasonic wave (photoacoustic signal). This way one can measure the optical contrast of a sample with ultrasonic resolution. For collecting photoacoustic signals our group introduced so called integrating detectors a few years ago. Such integrating detectors integrate the pressure in one or two dimensions (line or plane detectors). Thereby the three dimensional imaging problem is reduced to a two or a one dimensional problem for the pressure projections for line or plane detectors, respectively. Several reconstruction methods like Fourier or F-SAFT reconstruction or back projection are used for the two dimensional first step, but the model-based time reversal method shows a significant advantage: acoustical heterogeneity and attenuation, which both cause blurring of reconstructions, can be directly implemented in the reconstruction method. The integrating detectors are mainly optical detectors and thus can provide a high bandwidth up to several 100 MHz. Using these detectors the resolution is of ten limited by the acoustic attenuation in the sample itself, because attenuation increases with higher frequencies. For thin layers, small cylinders, and small spherical inclusions the effect of attenuation in human fat is simulated and the influence of dispersion on image reconstruction is shown.

1. Introduction

In 1880, Alexander Graham Bell discovered that pulsed light striking a solid substrate can produce a sound wave, a phenomenon called the photoacoustic effect [1]. It is only relatively recently that practical imaging methods based on this effect have been developed and reported [2]. Today, photoacoustic tomography (PAT), which is also referred to as optoacoustic tomography or as thermoacoustic tomography when using microwaves instead of light for excitation, is attracting intense interest for cross-sectional or three-dimensional imaging in biomedicine.

In photoacoustic imaging, short laser pulses are fired at a sample and the absorbed energy causes local heating. This heating causes thermoelastic expansion and the generation of broadband elastic pressure waves (ultrasound) which can be detected outside the sample, for example by a piezoelectric device.

A map or “image” of the photo-generated pressure distribution in the sample can be made by collecting the ultrasound at many different locations and processing it using a suitable algorithm e.g. by a filtered backprojection algorithm or by a time reversal algorithm. Only if the pulse is short enough, thermal expansion causes a pressure rise proportional to the locally absorbed energy density. Any photons, either unscattered or scattered, contribute to the absorbed energy as long as the photon
Using a Nd:YAG laser and an optical parametric oscillator (OPO) light pulses from the infrared to the visible regime can be selected with a repetition rate from 10 Hz up to 100 Hz. Some high speed PAT systems can even go up to 1000 Hz. The pulse duration in the nanosecond range enables a theoretical resolution of several microns in tissue (sound velocity similar to water at approx. 1500 m/s). For biomedical applications the light energy should not exceed 20 mJ/cm² in the visible spectral range.

Without consideration of acoustic absorption, the acoustic pressure \( p(r,t) \) can be expressed by the following wave equation:

\[
\Delta p(r,t) - \frac{1}{c^2} \frac{\partial^2 p(r,t)}{\partial t^2} = -\frac{\beta}{C_p} \frac{\partial}{\partial t} \Phi(r,t)
\]

where \( \Delta \) is the three-dimensional Laplace operator, \( c \) the sound velocity, \( \beta \) the thermal expansion coefficient, \( C_p \) the specific heat capacity and \( \Phi(r,t) \) the deposited energy per time and volume (“heating function”) caused by the absorption of the electromagnetic radiation in the sample.

For short electromagnetic pulses \( \Phi(r,t) = A(r) \cdot \delta(t) \), where \( A(r) \) is the energy density of the absorbed electromagnetic radiation. In such a situation the acoustic pressure \( p(r,t) \) solves the homogeneous wave equation with the initial conditions \( p(r,0) = p_0(r) = \beta c^2 / C_p \cdot A(r) = \Gamma \cdot A(r) \) and \( \partial / \partial t \ p(r,t = 0) = 0 \). The initial pressure \( p_0 \) at time \( t = 0 \) is therefore directly proportional to the absorbed energy density \( A \) with the dimensionless constant \( \Gamma \), the Grüneisen coefficient. The goal is to recover the spatial distribution of absorbed energy density inside the sample from acoustic pressure signals measured outside the sample (photoacoustic inverse problem).

The imaging resolution of large-aperture detectors is limited by the acoustic bandwidth and therefore also by the acoustic attenuation, which can be substantial for high frequencies. This effect is usually ignored in reconstruction algorithms but can have a strong impact on the resolution of small objects or structures within objects. Stokes could already show in 1845 that for liquids with low viscosity, such as water, the acoustical absorption increases by the square of the frequency [3]. Not taking into account the acoustic attenuation for photoacoustic image reconstruction, especially small structures (corresponding to shorter wavelengths and therefore higher frequencies) appear blurred [4].

To what extent this blurring can be compensated by regularization methods (as performed in [4]) and how much information is lost due the irreversibility of attenuation is still an open problem (see section 5). In this paper for thin layers(1D), small cylinders(2D), and small spherical inclusions(3D) the effect of attenuation in human fat is simulated and the influence of dispersion on image reconstruction is shown.

2. Integrating planar and line detectors

If the acoustic pressure outside the illuminated sample is measured with an integrating large-area detector, the signal at a certain time is given by an integral of the generated acoustic pressure distribution over an area that is determined by the shape of the detector. For example a planar detector measures the projections of the initial pressure distribution over planes parallel to the detector plane, which constitute the Radon transform of the initial pressure distribution [5]. Stable and exact three-dimensional imaging with a planar integrating detector requires measurements in all directions of space and so the receiver plane has to be rotated to cover the entire detection surface.

Recently we presented a simpler setup which requires only a single rotation axis and therefore the fragmentation of the area detector into line detectors perpendicular to the rotation axis [6]. Using a two-dimensional reconstruction method and applying the inverse two-dimensional Radon transform afterwards gives an exact reconstruction of a three-dimensional sample with this setup. Such integrating line detectors can be made of a stripe of a piezoelectric film or can employ optical sensing techniques where the line is formed by an optical waveguide or by a free propagating laser beam. The signal acquisition and image reconstruction procedures can be broken down into two steps. First at a given orientation of the line the detector is translated around the object to be imaged (Fig.1). Due to the integration of the pressure field over the length of the line the recorded signals are described by a solution of the two-dimensional (2D) wave equation with a source given by a projection of the initial pressure distribution along the line direction into a plane perpendicular to the line. Applying a 2D reconstruction algorithm to the set of signals yields an image of this projection. The object is then
rotated about an axis perpendicular to the line detector and another projection is obtained. This procedure is continued until projections from sufficient directions are sampled. In the second step the three dimensional (3D) initial pressure distribution is reconstructed by applying the inverse linear Radon transform to projection data in planes perpendicular to the rotation axis.

**Figure 1.** Line detector scanning around an object

3. **Fourier and F-SAFT reconstruction**

For the 2D reconstructions in the frequency domain an exact and very efficient method for image reconstruction is the two-dimensional form of a frequency-domain reconstruction algorithm developed for a planar detector array \([7,8]\). If this reconstruction algorithm is applied to signals measured with integrating line detectors e.g. parallel to the \(z\) direction along a line perpendicular to the detector (e.g., along the \(x\)-axis) it gives the two-dimensional source distribution in the \(x\)-\(y\)-plane, which is equivalent to the line integral along direction \(z\) of the pressure generated at time \(t = 0\) by the incident laser pulse.

The signals measured with the line detector along the \(x\)-axis are given by \(p(x, y=0, t)\), where \(p\) is the integrated acoustic pressure along the detector line parallel to the \(z\)-axis in the \(y=0\) plane where all detector lines are located.

The frequency-domain algorithm is given by \([7,8]\):

\[
P_0(k_x, k_y) = \frac{2c_k}{\text{sign}(k_x) \sqrt{k_x^2 + k_y^2}} A\left(k_x, \text{sign}(k_y) c \sqrt{k_x^2 + k_y^2}\right)\tag{2}
\]

where \(P_0(k_x, k_y)\) and \(A(k_x, \omega)\) are the 2D Fourier transforms of \(p(x,y)\) and \(p(x,y=0,\omega)\), respectively and \(\text{sign}\) is the signum function. The arguments of the Fourier transforms are the spatial wave vector components \(k_x\) and \(k_y\) and the temporal angular frequency \(\omega\). This algorithm gives only an exact reconstruction if the scan length along the \(x\)-axis is infinite. To test the reconstruction with a finite scan length we performed the following simulation. The source in this 2D simulation is a Gaussian shaped peak shown in Fig.2a. The calculated temporal signals for a scan are shown in Fig.2b. The reconstruction using the finite scan data along the \(x\)-axis is shown in Fig.2c. It can be clearly seen that the resolution in the \(x\) direction is worse than in the \(y\) direction. This is a result of the missing information in frequency space, where a large wedge-shaped area near the \(k_x\) axis contains no data.

Another frequency domain algorithm, often used in laser ultrasonics, is F-SAFT (Fourier domain synthetic aperture focusing technique) \([9]\). Here the auxiliary function \(A\), which is the 2D Fourier transform of the measured data \(p(x,y=0,\omega)\), can be “propagated” to a certain depth \(y\) using a propagator:

\[
A(k_x, y, \omega) = \exp(iy \sqrt{\frac{\omega^2}{c^2} - k_x^2}) A(k_x, y=0, \omega)\tag{3}
\]
Using the inverse 2D Fourier transform, $p(x,y,t)$ for this depth $y$ can then be calculated. We could show that if the discretization goes to zero, the Fourier transform and the F-SAFT give the same results for reconstruction. For finite discretization F-SAFT shows fewer artifacts (see Fig. 2d for simulation example) than frequency domain reconstruction (Fig. 2c). As the wave vectors $k$ for a constant frequency $\omega$ are located on a circle an interpolation step is necessary before the inverse FFT can be used in the frequency domain reconstruction. More accurate interpolation shows fewer artifacts (in Fig. 2c nearest neighbor interpolation is used). During the F-SAFT reconstruction algorithm no interpolation is necessary, but the calculations have to be performed for every depth. Therefore the numerical effort is higher than using the frequency domain algorithm.

4. Frequency dependent acoustic attenuation
The propagation of the attenuated pressure wave obeys the following equation [4]:

$$\Delta p(r,t) - \frac{1}{c_0^2} \frac{\partial^2}{\partial t^2} p(r,t) + L(t)*p(r,t) = 0 \quad \text{with} \quad L(t) = \frac{1}{2\pi} \int_{-\infty}^{+\infty} \left[ K(\omega)^2 - \frac{\omega^2}{c_0^2} \right] e^{-i\omega t} d\omega$$  (4)
Here \( K(\omega) = \frac{\omega}{c(\omega)} + i \cdot \alpha(\omega) \) is a complex wave number.

For tissue the attenuation coefficient \( \alpha(\omega) \) grows approximately linearly with frequency, which gives according to the Kramers-Kronig relations a monotonic increasing function for \( c(\omega) \) [10].

The pressure of the attenuated wave \( p_{\text{att}} \) at detection distance \( r = r_5 \) is obtained from the signal without attenuation \( p_{\text{ideal}} \) by:

\[
p_{\text{att}}(r_5, t) = \frac{1}{2\pi} \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} p_{\text{ideal}}(r_5, t') \exp(i c_0 K(\omega) \cdot t') d\omega \cdot \exp(-i \omega \cdot t) d\omega
\]

\((5)\)

Figure 3: Pressure at detection point 10 mm away from a thin layer (1D) with a thickness of 0.2 mm, a cylinder (2D), and a sphere (3D) with a diameter of 0.2 mm (right) and reconstructed initial pressure (left); without attenuation (solid) and with attenuation (dashed); the attenuation corresponds to human fat tissue.

Figure 4: Pressure at detection point (right) and reconstructed initial pressure (left) in 3D; no attenuation (solid), with attenuation but no dispersion (dashed), with attenuation and dispersion (dotted).

In Fig. 3 for thin layers (1D) with a thickness of 0.2 mm, cylinders (2D), and spheres (3D) with a diameter of 0.2 mm the signal with \( p_{\text{att}} \) (dashed line) and without \( p_{\text{ideal}} \) (solid line) attenuation are shown (right) at a distance of 10 mm in human fat (0.6 dB MHz\(^{-1}\) cm\(^{-1}\)). The reconstructed images...
from the attenuated signals are shown on the left (dashed line) and look very similar for the different dimensions. In Fig. 4 the influence of dispersion is shown for a sphere with a diameter of 0.1mm (half of the diameter of Fig. 3). Dispersion mainly shifts the attenuated signal in time (right).

5. Conclusions and outlook on future work
For the 2D reconstruction of the initial pressure integrated on lines parallel to the detector line F-SAFT shows less artifacts than the frequency domain reconstruction algorithm, because no interpolation step is necessary. The influence of attenuation on the reconstruction is the same for integrating planar detectors, line detectors, and point detectors. Dispersion for small structures causes an additional time shift of the “measured” signal and can show a significant influence on the reconstructed initial pressure distribution. As $c(\omega)$ is a monotonic increasing function, signals from small structures arrive earlier. For other wave equations describing attenuation, e.g. the Stokes equation, the dispersion and therefore this time shift is often much smaller at the same attenuation. The linear increase of the absorption as a function of frequency in tissue is empirically proven only in a certain frequency range [10]. The behavior of the absorption outside this frequency range is still an open question. As the sound velocity as a function of the frequency $c(\omega)$ is determined by the absorption in the whole frequency range according the Kramers-Kronig relations, the actual dispersion can be different. Therefore additional measurements of absorption and dispersion are foreseen.

Compensation of the frequency-dependent attenuation is an ill-posed problem and is limited by the thermodynamic fluctuations of the measured pressure around its mean value. In future work these fluctuations and their influence on the reconstructed image will be studied to get a theoretical limit of the achievable resolution taking acoustic attenuation into account.

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References
[1] A. G. Bell, “On the production and reproduction of sound by light: the photophone,” American Journal of Science 20, 305–324 (1880).
[2] M. Xu, and L. V. Wang, “Photoacoustic imaging in biomedicine,” Review of Scientific Instruments. 77(4), 041101-041122(2006).
[3] G. G. Stokes, “On the theories of the internal friction of fluids in motion, and of the equilibrium and motion of elastic solids,” Trans. Cambridge Philos. Soc. 8, 287-319(1845).
[4] P. J. La Rivière, J. Zang, and M. A. Anastasio, “Image reconstruction in optoacoustic tomography for dispersive acoustic media,” Optics Letters. 31(6), 781-783(2006).
[5] M. Haltmeier, O. Scherzer, P. Burgholzer, and G. Paltauf, “Thermoacoustic tomography with integrating area and line detectors,” Inverse Problems 20(5), 1663-1673(2004).
[6] P. Burgholzer, C. Hofer, G. Paltauf, M. Haltmeier, and O. Scherzer, “Thermoacoustic tomography with integrating area and line detectors,” IEEE Trans. Ultrason., Ferroelect., Freq. Contr. 52(9), 1577-1583(2005).
[7] K. P. Kostli, M. Frenz, H. Bebie, and H. P. Weber, “Temporal Backward Projection of Optoacoustic Pressure Transients Using Fourier Transform Methods”, Physics in Medicine and Biology 46, 1863-1872 (2001)
[8] Y. Xu, D. Z. Feng, and L. V. Wang, “Exact Frequency-Domain Reconstruction for Thermoacoustic Tomography – I: Planar Geometry,” IEEE Transaction on Medical Imaging 21, 823-828 (2002).
[9] D. Lévesque, A. Blouin, C. Néron and J.-P. Monchalin, „Performance of laser-ultrasonic F-SAFT imaging”, Ultrasonics 40 (10), 1057-1063 (2002).
[10] N. V. Shushilov and R. S. Cobbold, “Frequency-domain wave equation and its time-domain solutions in attenuating media”, J. Acoust. Soc. Am. 115, 1431-1436 (2004).