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2-D image reconstruction of photoacoustic endoscopic imaging based on time-reversal

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ABSTRACT

Backgrounds: Photoacoustic endoscopic (PAE) imaging is a rapidly emerging interventional imaging modality for identification and characterization of intraluminal pathological tissues. Since the scanning aperture of PAE is enclosed in the lumen, image reconstruction techniques used in photoacoustic tomography (PAT) can not be directly applied.
Objective: The purpose of this work is to design an image reconstruction method based on time-reversal (TR) for a PAE imaging catheter equipped with a single-element NFU transducer with circular scanning.
Methods: Firstly, the back-propagation of photoacoustic waves emitted from the tissue absorbers was modeled and simulated. Then, two-dimensional (2-D) grayscale images of the acoustic pressure distribution were obtained displaying the morphological structure of luminal cross-sections. A computer-simulated vessel phantom embedded with atherosclerotic plaques was used to validate and quantitatively evaluate the method.
Result: The structural similarity (SSIM) of the images reconstructed with TR is comparable to algebraic reconstruction technique (ART), which is at least 65% higher than filtered back-projection (FBP). The time cost of TR is about 16 times that of FBP and 1/4 of ART under the same test condition.
Conclusion: The reconstructed image quality may degrade when the photoacoustic data are incomplete due to sparse measuring locations and limited-view scanning. The spline interpolation can be used to improve the image quality and eliminate artifacts.

1. Introduction

Photoacoustic endoscopic (PAE) imaging is a newly emerging and rapidly developed catheter-based imaging technique. Its emergence is based on the endoscopic ultrasound and photoacoustic tomography (PAT) imaging technique. It combines rich optical contrast with high ultrasonic resolution at depths up to several centimeters. A special catheter with a tiny probe mounted on its tip is directly inserted into the lumen and then pulled back slowly. A nanosecond pulsed laser or continuous laser after amplitude modulation is employed to illuminate the tissues. Upon the catheter revolves around its axis, the surrounding tissues absorb light and then acoustic waves, usually called photoacoustic (PA) waves, are generated and propagate through the tissues. The PA waves are then collected on the surface of the medium by conventional piezoelectric (PZE) ultrasonic transducer [1], polyvinyl difluoride (PVDF) hydrophone or optical sensor [2]. PA pressures are spatially resolved through inversion of the acoustic waves. A series of cross-sectional grayscale images are finally reconstructed from the PA pressures displaying the anatomic structures, functions, and molecular information of the luminal tissues [2–4].

There are essentially two inverse problems in PAE imaging [5]. The first one is acoustic inverse problem, which is usually considered as image reconstruction. It concerns about the reconstruction of the initial spatial distribution of acoustic pressure or deposited optical energy density from the pressure time series detected outside of the imaging object. The second one is optical inversion, so-called quantitative PAT. It consists of reconstructing the spatial distribution of the tissue optical absorption coefficient or the chromophores from the collected acoustic pressures or the deposited optical energy that is recovered from the acoustic inversion [6]. The focus of this paper is on the acoustic inverse problem.

The technique used to reconstruct PAT images depends on the complexity of the target structures. Furthermore, it relates to the way upon acquiring acoustic waves and the ultrasonic probe types [7–9]. There are three types of PZE ultrasonic probes used by current PA imaging devices: a focused ultrasonic sensor with linear scanning, a single-element none-focused ultrasonic (NFU) transducer with circular scanning, and a circular or linear array of

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ultrasonic transducers [10]. The linear scanning of a focused ultrasonic sensor is similar to B-scan. No mathematical reconstruction is needed and it is basically the adding of A-lines together with a coordinate transformation. However, the detection depth of the imaging system equipped with a focused ultrasonic sensor is limited in comparison with the latter two types of detectors. Besides, a wider range imaging can be implemented by using an NFU detector, which can be regarded as an ideal point-like detector under certain conditions. In the case of an ultrasonic transducer array, the acoustic waves generated by the surrounding tissues are simultaneously collected. Some PAE imaging systems adopted a single-element NFU transducer with 2-D circular scanning or a circular array of transducers with planar scanning by considering the special closed imaging geometry [1,3]. In [2], two all-optical forward-viewing PAE probes were developed, which employ a FP (Fabry-Perot) polymer film ultrasound sensor located at the distal end of the optical-fiber bundle. An array of ultrasonic transducers has the advantages of high data acquisition speed and fast image reconstruction in comparison with a single-element ultrasonic transducer. However, it has a higher requirement to the processing speed of the data acquisition circuit and the cache space. Besides, the spatial resolution of the reconstructed image highly depends on the density of the ultrasonic transducer array and the focusing capability. The lack of details, low lateral resolution and image artifacts may occur in the result images. A phase-controlled focusing reconstruction algorithm is one of possible solutions [11]. It is the most widely used image reconstruction method for the PA imaging system equipped an ultrasonic transducer array.

The PAE scanning aperture is enclosed in a closed lumen. The acquisition of acoustic waves generated by the tissues are highly limited, thus exact reconstruction formulas do not exist [12]. Those well-developed methods for PAT can not be directly applied to reconstruct PAE images. Fortunately, image reconstruction is not constrained by the scanning way or the scanning trajectory of the ultrasonic detector. It is possible to apply conventional image reconstruction algorithms to recover cross-sectional PAE images, including filtered back-projection (FBP) [12–14], time-reversal (TR) [15,16] and algebraic reconstruction techniques (ART) [17,18], etc. FBP is one of closed-form analytic inversion formulas in tomography. The FBP-based algorithms have the advantages of simple formulation and fast reconstruction. However, the back-projection is performed along an arc surrounding the current measuring location. In consequence, artifacts may exist in the reconstructed image. Besides, FBP requires that the collected acoustic pressures are complete. The detector is needed to rotate 360° around the imaging object and collect all of the acoustic waves generated by the surrounding tissue absorbers. Unfortunately, all-view scanning is difficult to be implemented for PAE imaging due to the enclosed imaging geometry. ART is one of well used iterative methods in approaching inverse problems. The continuous problem is discretized and then solved in an iterative manner. Such techniques have been well developed in CT imaging. Compared with FBP, ART is more suitable for the case of the limited-view scanning due to its ability to incorporate accurate imaging physical models and the instrument response [17]. The main concern of the iterative image reconstruction methods is that they may lead to long reconstruction time even when accelerated by use of modern computing hardware [19]. Most experimental studies of PAT to date have employed analytic image reconstruction algorithms [20]. TR reconstruction method is based on the finite difference solution of the wave equation [21]. It is easy to implement for any shape of the observation surface and acoustically inhomogeneous media [22].

In this study, an image reconstruction method based on TR was designed for a PAE imaging catheter equipped with a single-element NFU transducer with circular scanning. The method has been demonstrated with a computer-generated vessel model. Its performance was compared with FBP and ART based on quantitative evaluation. The advantages as well as limits were discussed. Possible factors that may influence the reconstruction quality were also investigated.

2. Methods

2.1. PA imaging principle

The acoustic pressure, $p(\boldsymbol{r}, t)$, collected at a scanning location $\boldsymbol{r}\in\Omega$ and at time t, can be derived according to the PA wave equation [7],

$$p(r, t) = \frac{\beta I_0 c}{4\pi C_p}\tau\frac{\partial}{\partial t}\oint_{|r-r'=ct|}\frac{A(r')}{t}dr', \tag{1}$$

where Ω is the imaging region (i.e., the propagation region of the acoustic wave), $A(\boldsymbol{r})$ represents the spatial distribution of the optical absorption coefficient, I_0 is the laser pulse energy, c is the acoustic velocity in the biological tissue, β and C_p are, respectively, the isobaric thermal expansion coefficient and specific heat capacity of the biological tissue, and τ is the laser pulse width. According to Eq. (1), the amplitude of the PA wave generated by an absorber is actually proportional to β and I_0. At time $t=|\boldsymbol{r}-\boldsymbol{r}'|/c$ and a scanning position $\boldsymbol{r}$, a NFU transducer detects acoustic pressure signals, which are the integral of the acoustic waves generated by all absorbers along a circle of radius $|\boldsymbol{r}-\boldsymbol{r}'|$ around the current scanning position.

The image reconstruction aims to determine $A(\boldsymbol{r})$ from $p(\boldsymbol{r}, t)$. An image reflecting the spatial distribution of the optical absorption is obtained through solving the initial distribution of the acoustic pressures within the imaging region at $t=0$. The physical model used to describe the lossless propagation of ultrasonic waves in the acoustically homogeneous medium is given by three coupling equations [23],

$$\begin{cases} \frac{\partial}{\partial t}u(\boldsymbol{r}, t) = -\frac{1}{\rho_0}\nabla p(\boldsymbol{r}, t) \\ \frac{\partial}{\partial t}\rho(\boldsymbol{r}, t) = -\rho_0\nabla\cdot u(\boldsymbol{r}, t)' \\ p(\boldsymbol{r}, t) = c^2\rho(\boldsymbol{r}, t) \end{cases} \tag{2}$$

where $u(\boldsymbol{r}, t)$ denotes the vibration velocity of the medium, and $\rho(\boldsymbol{r}, t)$ and ρ_0, are acoustic density and medium density, respectively.

2.2. Constructing initial image

For simplicity, the NFU transducer mounted on the catheter tip is supposed to be a point-like detector. As shown in Fig. 1, the imaging plane centers on the detector and is perpendicular to the catheter axis. The radius of the circular scanning trajectory is equal to the catheter radius, d_0. The reconstruction region is located outside of the scanning circle. Both the width and length of the initial image are l mm. The initial image is discretized into $M\times M$ grids in a Cartesian coordinate system centered on the catheter center. The node at the intersection of row m and column n is denoted as (m, n), where $m, n = 1,2,\ldots,M$. The distance between adjacent nodes is $\Delta x = l/M$.

The reconstructed grayscale image is represented by a matrix $\boldsymbol{B}$ of $M\times M$ dimensions. $p(\boldsymbol{r}_s, t)$ represents the acoustic pressure measured by the detector at a point $\boldsymbol{r}_s$ and at time $t\in[0,T]$, where T is the temporal limit.