Comparison of wavefront distortion in the breast between optoacoustic imaging and ultrasonic imaging

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Abstract — Wavefront distortion in breast OA imaging is investigated and compared with that in conventional ultrasound. An adaptive OA imaging method is also proposed to compensate for the degradation caused by breast tissue inhomogeneities. Both OA and ultrasound wave propagation were simulated using a finite difference time-domain method. The proposed OA image correction method adaptively adjusted the sound speed at each image pixel in order to maximize an absorption-based image quality factor. As in reported *in vivo* and *in vitro* ultrasonic experiments in the literature, results showed that OA wavefronts suffer from significant time fluctuations and waveform distortion. For conventional ultrasound, the spatial coherence is limited by the van Cittert-Zernicke theorem and it decreases with the relative distance. The OA signals, on the other hand, exhibit relatively higher spatial coherence under the same conditions. Results also showed that the proposed correction algorithm can greatly enhance the OA image quality around a tumor-like area.

I. INTRODUCTION

In recent years considerable efforts have been devoted to the study of optoacoustic (OA) effects in human breast [1]. Current OA imaging algorithms, however, are based on the assumption of uniform acoustic properties throughout the medium to be imaged. Effects of acoustic heterogeneities on ultrasound breast imaging have been studied by many researchers. Zhu and Steinberg [2] observed severe amplitude distortion in narrowband ultrasound pulses in their one-dimensional *in vivo* transmission measurements. *In vitro* measurements by Hinkleman *et al.* [3] demonstrated serious waveform distortion and energy level fluctuation (ELF) in broadband ultrasonic wavefronts after propagating through breast tissue specimens. In breast OA tomography, Xu and Wang [4] proposed a ray approach similar to that in geometric optics to account for the OA wavefront distortion. Using acoustic ray tracing under simplified geometries, they inferred that amplitude distortion caused by refraction was not serious in breast OA tomography because OA signals are broadband, in a lower frequency range and experience only one-way transmission distortion. They also assumed that waveform distortion was negligible and concluded that only arrival time fluctuation (ATF) was dominant in breast OA tomography. However, *in vitro* one-way transmission experiments [3] [5] have shown that waveform

distortion and ELF was generally not negligible for broadband ultrasound. In addition, full-wave approaches such as the finite-difference time-domain (FDTD) method [5] showed that waveform distortion and ELF were accumulative results caused by acoustic impedance mismatch along all possible propagation paths, which cannot be exhausted with simple ray tracing. An OA wavefront could potentially undergo less waveform distortion as compared to an ultrasonic wavefront because it suffers less phase aberration due to its lower frequency range [6], but it still remains unclear that if ELF and waveform distortion are negligible in breast OA tomography because no counterparts of experiments or full-wave simulations as those in clinical breast ultrasound have yet been seen.

In this study the OA wavefront distortion is investigated through a two-dimensional (2-D) tissue map that was based on a measured two-dimensional breast model including acoustic scatterers with a distribution of optical absorption coefficient. The FDTD method is adopted to simulate the OA wavefront propagation inside the breast model. Moreover, pulse-echo wavefronts for a focused linear array are also obtained for both comparison and validation of the proposed model via the van Cittert-Zernicke theorem. A quality-factor (QF) based image correction method is proposed to improve the optical contrast of the OA image.

II. DESCRIPTIONS OF THE BREAST MODEL AND THE SIMULATION METHOD

The background acoustic models adopted in all simulations hereafter were constructed according to a breast phantom made by the Medical Sciences Center (Medical School, University of Wisconsin–Madison, USA). As shown in Fig. 1, the cross-sectional model consists of a zone of glandular parenchyma surrounded by subcutaneous fat. The sound speed in the parenchyma and the fat were set as 1523 m/s and 1459 m/s, respectively. The breast model has a radius of 5 cm. The averaged radius of the parenchyma is 4 cm and the thickness was varied in such a way that the ATF profile for an acoustic wavefront after propagating through the parenchyma wall will have a root-mean-square (RMS) value of 66.8 ns and a correlation length of 4.3 mm, which was based on the *in vitro* measurements obtained by Hinkleman *et al.* [3]. Acoustic scattering was also incorporated in our model by adding a normally-distributed, zero-mean random component with a RMS value of

0.1 g/cm^3 to the density map in which the density of the parenchyma and the fat were set as 0.99 g/cm^3 and 0.94 g/cm^3 , respectively. A typical density map is shown in Fig. 1. Two simulated breasts were considered, one had a disk with a radius of 1 mm placed at the center, another had three disks with the same radius of 4 mm placed around the model center. The absorption coefficients of these disks were set to be 2, 4, and 6 after normalization with respect to that of surrounding tissues. The sound speed inside the disk was 1550 m/s and the density was 1.12 g/cm^3 .

In this study OA wave propagation and pulse-echo ultrasound were simulated by numerically solving the OA equations and the acoustic equations with the help of the FDTD method, respectively. The grid size Δx was set at $43 \mu\text{m}$ and a grid of 2690×2690 was employed. The time step Δt was set to 7 ns in order to meet the Courant stability condition [7]. The simulation procedures were similar to those described in [5] and [8], respectively. In OA simulations, uniform irradiation was assumed with the laser pulse duration set as 10 ns. Point ultrasound detectors were placed around the breast model at an interval of 0.25 degree (i.e., totally 1440 points), while the radius of detection is 6 cm. Received OA signals were obtained by summing the OA waveforms received at 6 consecutive points (i.e., totally 240 received OA signals). In pulse-echo ultrasound, a linear array with a focus at the breast center was placed 1 cm above the breast wall to be used both as a transmitter and as a receiver. A Gaussian pulse with a central frequency of 1.5 MHz and a 6-dB bandwidth of 1.5 MHz was employed as the transmitted waveform. This frequency range, which was lower than usually used in breast ultrasonography but was common in breast OA imaging [6], was chosen to form a fair basis for the comparison between the OA wavefront and the pulse-echo wavefront. The linear array consists of 1440 points spaced at an interval of 0.043 mm. Also acoustic signals coming from 6 consecutive points were summed together to form 240 received acoustic waveforms.

III. THE CORRECTION METHOD

The QF-based image correction method was first proposed by Muller and Buffington [9] to adaptively focus incoherent optical telescopes. In clinical ultrasound, Nock *et al.* [10] suggested the use of mean speckle brightness as a QF to adaptively adjust the transmit and receive time-delay to obtain an optimal image. In contrast to ultrasound imaging, the absorption profile, rather than the speckle brightness, is the quantity to be reconstructed in OA imaging. Therefore, the QF should be directly related to the absorption profile if a QF-based correction algorithm is to be used for OA imaging. For an

acoustically homogeneous medium, the absorption profile can be accurately reconstructed using the following time-domain algorithm proposed by Xu *et al.* [5]:

$$A(r) = -\frac{r_0^2 C_p}{2\pi c_0^4 \beta} \int_{\theta} d\theta \left. \frac{1}{t} \frac{\partial p(r_0, t)}{\partial t} \right|_{t(r, r_0) = |r_0 - r|/c_0}, \quad (1)$$

where $A(r)$ denotes the optical absorption, r_0 is the radius of detection, C_p is the specific heat, β is the coefficient of volume thermal expansion, and $p(r_0, t)$ is the OA signals received at each scanning angle θ . In (1) a constant sound speed was assumed and will become inaccurate when an inhomogeneous speed distribution $c(r)$ is encountered. Therefore, the time-of-flight required for sound propagation from r to r_0 should be modified as follows:

$$t(r, r_0) = \int_{L(r, r_0)} \frac{dl}{c(r)}, \quad (2)$$

where $L(r, r_0)$ is the straight line connecting r and r_0 . In the proposed correction method the imaged zone was first partitioned into pixels with equal areas, the sound speed over each pixel of the imaged region was then sequentially adjusted upward or downward to obtain a local maximum of the prescribed QF calculated according to (1) and (2). The entire procedure can be iterated until no significant improvements are available. In this study we adopt $A(r)^3$ as the QF in the proposed correction method. Even power of $A(r)$ were avoided because negative values of $A(r)$, which were unwanted errors but indistinguishable if $A(r)$ to even power were used, are usually encountered in (1) due to both acoustic heterogeneities and incomplete data acquisition.

IV. NUMERICAL RESULTS

The scattering property of the background model was validated by the spatial covariance for the pulse-echo wavefront. Fig. 2 shows the average result calculated using ten realizations with independent density distributions and choosing the first received waveform as a reference signal. Error bars indicate ± 1 standard deviation. It can be seen that the spatial covariance decreases approximately linearly with the distance, which is as predicted by the van Cittert-Zernicke theorem [11] and also justifies the use of the background scattering model.

The pulse-echo acoustic wavefront and the OA wavefront formed by aligning the waveform calculated using $\partial p(r_0, t)/\partial t$ for each scanning

angle are displayed in Fig. 3. $\partial p(r_0, t)/\partial t$ was used instead of $p(r_0, t)$ because $\partial p(r_0, t)/\partial t$ is the waveform used to coherently reconstruct the absorption profile, as is shown in (1). The relative-time delays between individual received waveforms in these wavefronts have been removed. Obvious ELF's can be seen from both wavefronts. It can be seen that the OA wavefront has noticeably higher coherence than that of the pulse-echo wavefront. More specifically, the waveform similarity factors [3] of the OA wavefront and the pulse-echo wavefront are 0.939 and 0.822, respectively.

The reconstructed absorption profile for the simulated phantom is shown in Fig. 4. The display area is 3.3 cm x 3.3 cm. Panel (a) shows the result corrected using 1x1 pixel. The contrast between the tumors and the background are 3.29 dB (I), 2.27 dB (II) and 0.95 dB (III). The OA image corrected using the proposed method with 12x12 pixels is shown in Fig. 4 (b). The contrast between the tumors and the background are 4.26 dB (I), 2.90 dB (II) and 1.09 dB (III). Undistorted values of contrast are also calculated as 5.27 dB (I), 3.62 dB (II) and 1.47 dB (III).

Fig. 5 (a) shows the simulated B-scan image for the same phantom used in Fig. 4, while Fig. 5 (b) shows the image corrected using the method proposed by Nock *et al.* [10] (i.e., adjust the time-delay on each channel to maximize the mean speckle brightness). The standard deviation of density variation in the background is 20 dB higher than that inside the tumors. The two images are visually close, with calculations show that the contrast between the tumors and the background are 14.88 dB (I), 13.02 dB (II), and 18.08 dB (III) in Fig. 5 (a) and 14.74 dB (I), 12.77 dB (II), and 18.71 dB (III) in Fig. 5 (b).

V. CONCLUDING REMARKS

From numerical results it can be seen that both the OA image and the ultrasound image are significantly degraded by acoustic heterogeneities in breasts. ATF can be compensated only if the relative time-delay between channel signals can be accurately estimated. For pulse-echo ultrasound, speckles are inevitable in ultrasound imaging because the acoustic structure for a soft tissue is inherently granular (i.e., randomly distributed scatterers with a mean size much smaller than the involved signal wavelength). On the contrary, the OA wavefront usually is composed of thermoelastic pulses with individual lengths of approximately the corresponding tissue sizes, which were considerably large as compared to acoustic scatterers and lead to no speckles after being used for optical absorption reconstruction. The non-scattering property also results in the high coherence of an OA wavefront, which explains the good efficacy of the proposed correction method in

improving the image contrast. On the contrary, the low spatial coherence of the pulse-echo wavefront makes the time-shift-based compensation less effective for the B-scan ultrasonic images. However, the OA wavefront was still distorted in waveform similarity after propagating through the breast tissue (see Fig. 3 (b)), which was ignored by the ray-tracing approach [4] and cannot be compensated using time-shift-based methods such as the proposed one and the method proposed in [4].

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Fig. 1. Background density map for the breast model. Glandular parenchyma is enclosed by the irregular wall, subcutaneous fat is bounded by the irregular wall and the circular wall.

Fig. 2. Average spatial covariance of the pulse-echo wavefronts calculated using ten realizations. Error bars denote ± 1 standard deviation.

Fig. 3. (a) Pulse-echo wavefront and (b) OA wavefront for the simulated breast phantom with a tumor placed at the center. In (a), the vertical coordinate denotes the array direction. In (b), the vertical coordinate denotes the receive angle. The horizontal coordinates in (a) and (b) are time with a length of $22.7\mu s$ and an interval of $0.03\mu s$. Signal amplitude is displayed on a 128 linear gray scale, with the maximum represented by white and the minimum represented by black.

(a)

(b)

Fig. 4. Reconstructed OA images for the simulated phantom with three tumors around the center: (a) Without correction and (b) With correction using 12×12 pixel. The images are displayed on a 128 linear gray scale.

Fig. 5. B-scan images for the same phantom as in Fig. 4: (a) Without acoustic heterogeneities. (b) With heterogeneities and corrected by maximizing the mean speckle brightness inside the marked area. The images are displayed on a 40 dB logarithmic scale.