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Pixel based focusing for photoacoustic and ultrasound dual-modality imaging

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A B S T R A C T

It is desired that the same imaging functional modules such as beamformation, envelope detection, and digital scan conversion (DSC) are employed for the efficient development of a cross-sectional photoacoustic (PA) and ultrasound (US) dual-modality imaging system. The beamformation can be implemented using either delay-and-sum beamforming (DAS-BF) or adaptive beamforming methods, each with their own advantages and disadvantages for the dual-modality imaging. However, the DSC is always problematic because it causes blurring the fine details of an image, e.g., edges. This paper demonstrates that the pixel based focusing method is suitable for the dual-modality imaging; beamformation is directly conducted on each display pixel and thus DSC is not necessary. As a result, the artifacts by DSC are no longer a problem, so that the proposed method is capable of providing the maximum spatial resolution achievable by DAS-BF. The performance of the proposed method was evaluated through simulation and *ex vivo* experiments with a microcalcification-contained breast specimen, and the results were compared with those from DAS-BF and adaptive beamforming methods with DSC. The comparison demonstrated that the proposed method effectively overcomes the disadvantages of each beamforming method.

1. Introduction

In the last decade, photoacoustic (PA) imaging has received much attention due to its attractive features such as combining strong optical contrast with high ultrasound resolution [1–3]. The PA imaging uses non-ionizing laser energy to generate acoustic waves (i.e., PA signals) only inside a target that absorbs the incident laser energy. This is due to thermo-elastic expansion. An ultrasound (US) transducer detects the PA signals from which an image of initial pressure distribution is reconstructed. PA imaging is categorized into PA microscopy (PAM), PA tomography (PAT) and cross-sectional PA imaging (CS-PAI) based on imaging configuration [3,4]. Among them, CS-PAI is a viable method for clinical applications in real time and implemented by minimal modification of a current US scanner [5,6]. CS-PAI has potential to be used in clinical applications such as sentinel lymph node and microcalcification detection [7,8]. In addition, PA and US images are inherently easy to be co-registered together, and thus this dual imaging modality can simultaneously provide anatomical and functional information [7–11]. For this dual-modality imaging, it is efficient that the same imaging functional modules such as beamformation, envelope detection, and digital scan conversion (DSC) are used and thus it is necessary to seek suitable methods for the imaging functional modules.

To reconstruct an initial pressure distribution in PA imaging, several methods has been proposed: back projection, inverse Radon transform, and Fourier reconstruction [12–15]. In fact, the simple back projection is equivalent to delay-and-sum beamforming (DAS-BF) typically used in US imaging [14]. The inverse Radon transform and the Fourier reconstruction methods can enhance the spatial resolution if the frequency response of a PA signal receiver is sufficiently wide (e.g., all-pass frequency response); because of their data acquisition configuration, however, these methods are suitable for preclinical applications but limited to clinical applications, e.g., breast imaging. Also, the performances of the three beamforming methods are almost equivalent when the same transducer with a limited bandwidth (e.g., fractional bandwidth less than 80%) is used [16]. As a result, DAS-BF can be the simplest and most practical reconstruction method for CS-PAI as well as US imaging.

The CS-PAI based on DAS-BF generally suffers from deteriorating spatial resolution (i.e., high sidelobe level) because one-way

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beamforming is only viable. The problem can be partially solved using adaptive beamforming methods based on coherent factor (DAS-CF) or minimum variance (DAS-MV) at the expense of computational cost [17,18]. In addition, although these methods can enhance the lateral resolution compared to DAS-BF, these often do not work well when a signal-to-interference ratio (SIR) is less than 10 dB [19]; this drawback is generally problematic when a strong absorber (or reflector) is located near a weaker one in both PAI and US imaging. Especially, the adaptive beamforming methods for US imaging generate artifacts that are severe in the speckle region around strong reflectors, and thus lowering contrast resolution.

The CS-PAI also involves digital scan conversion (DSC) as in US imaging for efficient visualization; DSC is an essential functional block because either PA or US signal acquisition coordinates geometrically differs from display coordinates [20]. In DSC, interpolation is performed to acquire pixel values from adjacent received signals without Moiré artifacts. Among various interpolation methods (e.g., windowed sinc and bicubic), bilinear interpolation showed the best performance in terms of accuracy and complexity [21–23]. However, it introduces blurring fine details of an image. Furthermore, these blurring artifacts become severe as the number of transmitting scanlines decreases in US imaging, and these are exacerbated in zoom-in US images particularly. For US imaging, previously, we have proposed a pixel based focusing (PBF) method which can eliminate the blurring artifacts from DSC [24–26]. In the PBF, the beamformation is directly conducted on each display pixel and thus DSC is no longer necessary. Compared to DAS-BF, therefore, this method can enhance image quality (e.g., clear delineation of edges) because it effectively removes the artifacts from DSC. It should be noted that the PAT imaging systems have employed the pixel based beamformation [27], [28], but the way of beamformation and envelope detection differs from the PBF method for US imaging.

The purpose of this study is to ascertain whether the PBF method is also suitable for CS-PAI and thus for the dual-modality imaging. Although it was verified that the PBF method is capable of improving image quality by removing the blurring artifacts due to DSC in US imaging, its performance evaluation has not been conducted in the case where the number of scanlines extremely increases so that scanline distance can be less than pixel resolution; in CS-PAI, any number of scanlines can be produced because there are no transmitting scanlines and the PA signals are simultaneously generated from the entire scanning volume. This property of CS-PAI may lead to alleviating the blurring artifacts even though DSC is used. In addition, no comparison of spatial resolution of the PBF method with those of DAS-BF, DAS-CF, and DAS-MV has been performed, in particular, as a function of the number of scanlines. To investigate the suitability of the PBF method for CS-PAI, therefore, its spatial resolution was compared with those of DAS-BF, DAS-CF, and DAS-MV with DSC based on bilinear interpolation as a function of the number of scanlines; for this, simulation and *ex vivo* experiments with a microcalcification-contained breast specimen were conducted. In addition, the computational complexity of the PBF method was also compared with that of DAS-BF which is the simplest method.

2. Methods

2.1. Conventional DSC-based image reconstruction method

In DAS-BF, DAS-CF, and DAS-MV methods, dynamic receiving beamforming is conducted at sampling points, n, along a transmitting scanline and the beamformed radio-frequency (RF) data, $b(n)$, can be expressed as

$$b(n) = c(n) \cdot \sum_{m=0}^{M-1} w_m(n) x_m(n) = c(n) \cdot \left[\mathbf{W}(n)^H \mathbf{X}(n) \right], \quad (1)$$

where $c(n)$ is the weighting value for the beamformed data, M is the number of channels, $w_m(n)$ and $x_m(n)$ are the aperture weighting function and the PA signal (or US signal) after applying receiving focusing delay at the m_{th} channel. Also, $\mathbf{W}(n)$ and $\mathbf{X}(n)$ are M $\times$ 1 matrices of the aperture weighting function and the PA signal (or US signal), respectively. H represents the conjugate transpose. If $c(n) = 1$ and $\mathbf{W} = [1\,1\,\cdots\,1]^H$, Eq. (1) becomes a simple DAS-BF. In DAS-MV, $c(n) = 1$ and $\mathbf{W}(n)$ is found to minimize the variance of $b(n)$, which is given by [17]

$$\mathbf{W}(n) = \frac{\mathbf{R}(n)^{-1}\mathbf{d}}{\mathbf{d}^H \mathbf{R}(n)^{-1}\mathbf{d}}, \quad (2)$$

where $\mathbf{R}(n)$ is the covariance matrix and expressed as $\mathbf{R}(n) = E[\mathbf{X}(n)\mathbf{X}(n)^H]$, and $\mathbf{d}$ is the steering vector. Since $\mathbf{X}(n)$ is obtained after the focusing delays are applied to the received signals, $\mathbf{d}$ becomes a vector of ones. For DAS-CF, on the other hand, $c(n)$ serves as a coherent factor (CF) [8], and it is defined as

$$c(n) = \frac{\left| \sum_{m=0}^{M-1} x_m(n) \right|^2}{\left| \sum_{m=0}^{M-1} x_m(n)^2 \right|}. \quad (3)$$

The CF value, i.e., $c(n)$ in Eq. (1) ranges between 0 and 1. In DAS-CF, $\mathbf{W}$ is set to a vector of ones and the CF values are adaptively computed at each beamforming point [16], [17].

After beamforming and subsequent envelope detection, the envelope signals are geometrically transformed from Cartesian coordinates to polar coordinates (or vice versa) in DSC for final display. In the DSC, the bilinear interpolation is generally carried out to find a pixel value P from the envelope signals, and the pixel value is obtained by [23]

$$P = \frac{\theta_\alpha \cdot (\alpha E_{l+1,n+1} + \beta E_{l+1,n}) + \theta_\beta \cdot (\alpha E_{l,n+1} + \beta E_{l,n})}{\theta_\alpha + \theta_\beta}, \quad (4)$$

where $E_{l,n}$ is the envelope signal from the l_{th} scanline at the n_{th} sample, θ_α and θ_β are the angular differences between the pixel P and the two adjacent scanlines, and α and β are the normalized radial differences between the pixel P and the two samples on the scanline, respectively.

2.2. Pixel based focusing (PBF)

Fig. 1 conceptually illustrates the proposed PBF method. To obtain one pixel value, the receiving beamforming is repeatedly performed at L beamforming points (x_l, y_l), where $l = 0, 1, \ldots, L-1$, along a virtual scanline. The beamformation can be expressed as

$$p_{PBF}(x_l, y_l) = \sum_{m=1}^{M-1} w_m r_m(n - \tau_m), \quad (5)$$

where $r_m(n)$ is the received PA signal or US signal at the m_{th} channel and τ_m is the focusing delay for the beamforming points that can be computed by

$$\tau_m(x_l, y_l) = \frac{\sqrt{(x_m^E - x_l)^2 + (y_m^E - y_l)^2}}{c}, \quad (6)$$

where c is the sound speed in a propagation medium and (x_m^E, y_m^E) is the location of the m_{th} channel, respectively. After L times beamformation along the virtual scanline, baseband inphase and quadrature components (i.e., I and Q, respectively) can be obtained by