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Generalized Sidelobe Canceler Beamforming Applied to Medical Ultrasound Imaging[1]

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Abstract—A generalized sidelobe canceler (GSC) approach is proposed for medical ultrasound imaging. The approach uses a set of adaptive weights instead of traditional non-adaptive weights, thus suppressing the interference and noise signal of echo data. In order to verify the validity of the proposed approach, Field II is applied to obtain the echo data of synthetic aperture (SA) for 13 scattering points and circular cysts. The performance of GSC is compared with SA using boxcar weights and Hamming weights, and is quantified by the full width at half maximum (FWHM) and peak signal-to-noise ratio (PSNR). Imaging of scattering point utilizing SA, SA (hamming), GSC provides FWHMs of 1.13411, 1.68910, 0.36195 mm and PSNRs of 60.65, 57.51, 66.72 dB, respectively. The simulation results of circular cyst also show that GSC can perform better lateral resolution than non-adaptive beamformers. Finally, an experiment is conducted on the basis of actual echo data of an ultrasound system, the imaging result after SA, SA (hamming), GSC provides PWHMs of 2.55778, 3.66776, 1.01346 mm at $z = 75.6$ mm, and 2.65430, 3.76428, 1.27889 mm at $z = 77.3$ mm, respectively.

Keywords: generalized sidelobe canceler, adaptive beamforming, minimum variance, unconstrained optimization beamformer, medical ultrasound imaging

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1. INTRODUCTION

Medical ultrasound imaging with characteristics of high transmission capacity and low harm to human body has become one of the major medical diagnostic technologies nowadays, and imaging algorithm is the key technology of medical ultrasonic imaging [1]. Delay-and-sum (DS) is the most widely used imaging method, but the DS is characterized by a low signal-to-noise ratio (SNR) and low resolution [2]. In order to improve the SNR and lateral resolution of echo imaging, Jørgen Arendt Jensen [3] proposed the synthetic aperture (SA) beamforming. Although SA can equivalently achieve focusing during transmitting and receiving for every point at the same time in the whole field, as well as improves the SNR and lateral resolution of system, it still has a high level of sidelobe energy. Applying of apodization calculations can reduce the sidelobe level at the loss of lateral resolution.

For decades, adaptive beamformers which use collected echo data for calculating a weight vector to process the collected echo data have been used in other fields of array signal processing, e.g. antenna and radar. The calculated weight vector is equivalent to a spatial filter which can retain the desired signal and suppress noise and interference signal of echo data, thus it improves the quality of echo imaging [4, 5]. For medical ultrasonic imaging, adaptive beamforming was first introduced by Capon [6]. Johan-Fredrik Synnevåg [7] et al. studied a minimum variance algorithm, and Iben Kraglund Holfort [8] proposed a minimum variance based on frequency domain. In order to improve the stability of adaptive beamforming, Johan-Fredrik Synnevåg [9] put forward a diagonal loading to ultrasonic imaging, and J. E. Evans [10] proposed a subaperture averaging algorithm to solve the coherence of ultrasound echo.

Literature studies demonstrate that the imaging quality of using adaptive beamforming outperforms that of using traditional non-adaptive beamforming [11–14]. In this paper we propose a novel adaptive beamforming algorithm named generalized sidelobe canceler (GSC) which is applicable to medical ultrasound imaging system. This new approach provides a set of new adaptive apodization weights for every point in image, thus effectively suppressing the interference and noise of echo signal while maintaining the desired signal. The newly proposed adaptive beamformer GSC compared with early proposed adaptive beamformers can separate the linear constraints with adaptive filter, therefore, the constrained optimization problem of adaptive beamforming is converted to the unconstrained optimization problem, which can be

[1] The article is published in the original.

further studied by researchers. In order to verify the validity of the proposed approach, simulation experiments of scattering points and circular cysts, as well as actual experimental data are shown. The simulation experiments using Field II and actual experimental data demonstrate superior lateral resolution in the echo imaging after GSC compared with traditional non-adaptive algorithms.

The outline of this paper is as follows: In Section 2 we introduce the sensor signal model, principle of synthetic aperture beamforming and its application to ultrasound imaging. Section 3 introduces the principle of GSC and realization of GSC, in detail. Section 4 presents the experimental results based on simulation data and actual echo data. Finally, the advantages of the proposed methodology and its comparison with non-adaptive beamformers are discussed in section 5. The whole study is concluded in Section 6.

2. BACKGROUND

A. Synthetic Aperture (SA)

The SA belongs to non-adaptive beamformers, and it is the foundation of adaptive beamformer. For a phased array ultrasound system two steps are needed to realize the SA. Firstly, a single array is used to transmit ultrasound and all arrays to receive the echo, meanwhile, DS is used to deal with the echo data received by all arrays, and then received focused echo data will be got, which is called low resolution image (LRI). The delay time of each array is given by

$$k_{nm}(\mathbf{r}_p) = \frac{\left\|\mathbf{r}_n^{\mathrm{xmt}} - \mathbf{r}_p\right\| + \left\|\mathbf{r}_m^{\mathrm{rcv}} - \mathbf{r}_p\right\|}{c} fs, \qquad (1)$$

where n and m represent the transmitting and receiving arrays, respectively. N and M are the number of transmitting arrays and receiving arrays, which are equal to the number of total arrays, $n = 1, 2, ..., N$; $m = 1, 2, ..., M$; c is the velocity of ultrasound; fs is the system sampling rate; $\mathbf{r}_n^{\mathrm{xmt}}$ and $\mathbf{r}_m^{\mathrm{rcv}}$ are the spatial positions of the transmitting array and receiving array, and $\mathbf{r}_p$ is the spatial positions of a point. LRI is given by the expression

$$\mathrm{LRI}_n = \sum_{m=1}^{M} w_m x_m\left(k_{nm}(\mathbf{r}_p)\right), \qquad (2)$$

here $x_m(k)$ is the echo data of the m-th element.

Secondly, using all arrays to transmit ultrasound in turn, then the N's picture of LRI will be obtained. Finally, superposing all these LRIs, both transmitted and received focused ultrasound images will be obtained, and the ultrasound image is the beamforming response after SA, also known as the high resolution image (HRI):

$$\mathrm{HRI} = \sum_{n=1}^{N} w_n \mathrm{LRI}_n. \qquad (3)$$

For a traditional non-adaptive beamforming w_n is a series of fixed values, liken $\mathbf{w} = [1,1,\cdots,1]^{\mathrm{T}}$.

B. Sensor Signal Model

For traditional medical ultrasonic imaging, adaptive beamformers will process the echo data beamformed after SA, which means the echo data will achieve focusing during transmitting and receiving in advance. In order to describe more conveniently, we will redefine some variables later. To a linear array of M elements, the output of adaptive beamformers is given by [15]

$$y(k) = \mathbf{w}^{\mathrm{H}}(k)\mathbf{X}(k) = \sum_{i=1}^{M} w_i^*(k)\, x_i(k), \qquad (4)$$

where k is the time index. $\mathbf{X}(k)$ is the time-delayed version of array observations: $\mathbf{X}(k) = [x_1(k), x_2(k), \cdots, x_M(k)]^{\mathrm{T}}$, where $x_i(\mathrm{k}) = \mathrm{LRI}_i$. $\mathbf{w}(k) = [w_1(k), w_2(k), \cdots, w_M(k)]^{\mathrm{T}}$ is the complex vector of beamforming weights. $(\cdot)^{\mathrm{T}}$ and $(\cdot)^{\mathrm{H}}$ denote the transposition and conjugate transposition, and $(\cdot)^*$ denotes the conjugate complex, respectively. For adaptive beamformers $\mathbf{X}(k)$ can be divided into two parts:

$$\mathbf{X}(k) = \mathbf{S}(k) + \mathbf{P}(k), \qquad (5)$$

where $\mathbf{S}(k)$ represents the echo signal of the detected point, which is called desired signal, and $\mathbf{P}(k)$ is interference and noise signal, which include the sidelobe signal, thermal noise and reflection noise, etc.

For adaptive beamformers the weight vector $\mathbf{w}(k)$ is calculated from $\mathbf{X}(k)$ and varies with the echo data. An ideal weight vector $\mathbf{w}(k)$ corresponds to a spatial filter which can maintain the desired signal $\mathbf{S}(k)$ while suppressing the interference and noise signal $\mathbf{P}(k)$ of echo data, consequently, it improves the image quality of a medical ultrasound system.

Similarly, Eq. (4) can also show the non-adaptive beamformers. For $\forall x_i(k)$ $y(k)$ is the receiving focused echo signal of detected point, and the procedure of receiving focus is also known as DS. Setting $\mathbf{w}(k) = [1,1,\cdots,1]^{\mathrm{T}}$, the output of the beamformer $y(k)$ is a result of the SA, the beamformer response after SA $y(k) = \mathrm{HRI}$. Using the fixed window func-