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Adaptive field-of-view imaging for efficient receive beamforming in medical ultrasound imaging systems

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Abstract

Quadrature demodulation-based phase rotation beamforming (QD-PRBF) is commonly used to support dynamic receive focusing in medical ultrasound systems. However, it is computationally demanding since it requires two demodulation filters for each receive channel. To reduce the computational requirements of QD-PRBF, we have previously developed two-stage demodulation (TSD), which reduces the number of lowpass filters by performing demodulation filtering on summation signals. However, it suffers from image quality degradation due to aliasing at lower beamforming frequencies. To improve the performance of TSD-PRBF with reduced number of beamforming points, we propose a new adaptive field-of-view (AFOV) imaging method. In AFOV imaging, the beamforming frequency is adjusted depending on displayed FOV size and the center frequency of received signals. To study its impact on image quality, simulation was conducted using Field II, phantom data were acquired from a commercial ultrasound machine, and the image quality was quantified using spatial (i.e., axial and lateral) and contrast resolution. The developed beamformer (i.e., TSD-AFOV-PRBF) with 1024 beamforming points provided comparable image resolution to QD-PRBF for typical FOV sizes (e.g., 4.6% and 1.3% degradation in contrast resolution for 160 mm and 112 mm, respectively for a 3.5 MHz transducer). Furthermore, it reduced the number of operations by 86.8% compared to QD-PRBF. These results indicate that the developed TSD-AFOV-PRBF can lower the computational requirement for receive beamforming without significant image quality degradation.

Keywords: Front-end; Receive beamformer; Efficient phase rotation beamformer; Adaptive field-of-view imaging

1. Introduction

Digital receive beamforming (DRBF) has significantly improved the image quality of medical ultrasound systems [1]. However, DRBF requires very fast analog-to-digital converters (ADCs) to support the fine time delay resolution necessary for dynamic focusing. To reduce the high ADC sampling frequency, interpolation beamforming (IBF) and quadrature demodulation-based phase rotation beamforming (QD-PRBF) are commonly used [2–4]. However, both IBF and QD-PRBF need a computationally-demanding finite impulse response (FIR) filter for each receive channel (e.g., 32 complex FIR filters for a 32-channel system with QD-PRBF). The hardware requirement to support multiple FIR filters becomes especially challenging during the development of ultrasound systems with large channel counts (e.g., 3D ultrasound systems with 2D transducer [5]). Furthermore, it would be also difficult for ultraportable systems (e.g., handheld ultrasound systems [6]) to afford the needed computing resources for multiple FIR filters.

To reduce the computational requirement in DRBF, various beamforming approaches have been proposed

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[7–12]. For example, 2nd-order sampling-based demodulation lowers the computational requirement in PRBF by eliminating the need for demodulation filtering [7]. However, this method suffers from artifacts since it assumes narrowband signals and no frequency-dependent attenuation, which is typically not the case in medical ultrasound imaging. Similarly, sigma–delta oversampled beamformers were proposed to alleviate the need for complicated delay circuitries by utilizing one-bit sigma–delta modulators running at a high frequency. However, it yields a large reduction in image contrast (e.g., ~20 dB) due to dynamic focusing artifacts [8–10].

Previously, we developed an efficient beamforming method, i.e., two-stage demodulation-based PRBF (TSD-PRBF), which reduces the number of demodulation filters by performing dynamic focusing on the mixed signals, instead of demodulated signals [13]. From simulation and phantom studies, we found that TSD-PRBF provides comparable image quality to QD-PRBF when the beamforming frequency (f_{bf}) is greater than or equal to the ADC sampling frequency (f_{adc}). However, when the number of beamforming points is reduced (i.e., $f_{bf} < f_{adc}$), its image quality could be degraded because of the presence of signal harmonics at $\pm 2f_0$, where f_0 is the center frequency of the received signals [13].

In this paper, we propose adaptive field-of-view (AFOV) imaging to enhance the image quality in TSD-PRBF when the number of beamforming points is small.

2. Methods and materials

2.1. PRBF with two-stage demodulation

In two-stage demodulation-based phase rotation beamforming, QD is performed in two steps [13]. While mixing is performed before dynamic receive focusing, the computationally-demanding demodulation filtering is done after coherent summation, thereby substantially reducing the number of filters compared to QD-PRBF [13]. On the other hand, artifacts could be introduced due to (a) nonlinearity in dynamic receive focusing and (b) signal aliasing at lower beamforming frequencies (f_{bf}). Nonlinearity in dynamic focusing was found to have a negligible impact on image quality [13]. On the other hand, signal aliasing at reduced beamforming frequencies can lead to image quality degradation. Utilizing bandpass sampling principles, it was found that f_{bf} of $1.33f_0$ provides the minimum signal aliasing for beamforming frequencies between $1.33f_0$ and $2.66f_0$ [13,14]. Thus, the beamforming frequency needs to be selected based on the center frequency of received signals. However, f_0 varies with the imaging depth due to frequency-dependent attenuation [15].

2.2. Adaptive FOV (AFOV) imaging

We have developed a new method, where the beamforming frequency is adjusted based on the displayed FOV properties (i.e., axial size and mean center frequency) and the number of available beamforming points per scanline. Since only the displayed area is reconstructed in AFOV imaging, high spatial resolution can be achieved for small field-of-view sizes even with a reduced number of beamforming points as shown in Fig. 1. The maximum beamforming frequency in adaptive FOV imaging is given by

$$f_{bf} = \frac{c \times L}{2 \times D} \tag{1}$$

where c is the speed of sound in the medium, D is the axial FOV size, and L is the number of beamforming points. While Eq. (1) determines the maximum beamforming frequency, the center frequency of received signals needs to be estimated prior to selecting f_{bf} due to the fact that the ratio of f_{bf} to f_0 has a large impact on signal aliasing in TSD-PRBF.

To better understand signal aliasing at different beamforming frequencies, Fig. 2a shows an example frequency spectrum of the received signals with the center frequency f_0 and when the bandwidth (BW) equals $0.67f_0$. The spectrum of the received signals is a complex-valued function centered at $+f_0$ and $-f_0$, which may be represented by $A + jB$ and $A - jB$, respectively. Fig. 2b depicts the real and imaginary components of the frequency spectrum. These RF data are multiplied with cosine and sine at f_0 to generate the *inphase* (I) and *quadrature* (Q) data, respectively, during mixing. From Euler's formula, cosine is an even function and can be represented in the frequency

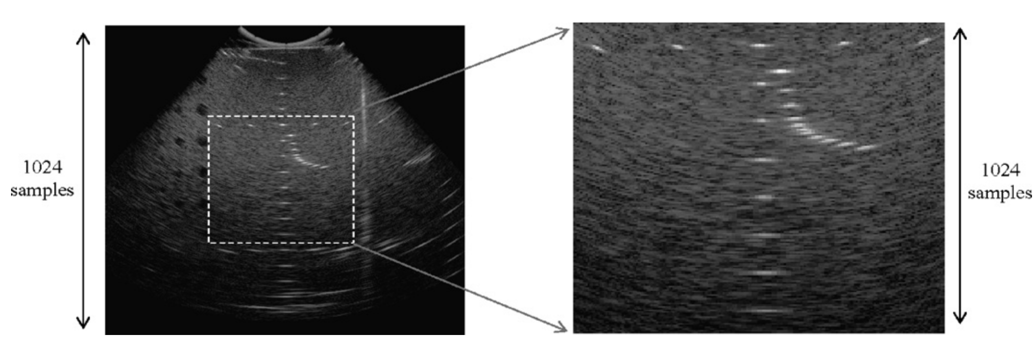

Fig. 1. Adaptive field-of-view (AFOV) imaging.