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Acquisition and Processing of V-wave Ultrasound Beams for Fast Frame Rate

Background, Motivation and Objective

Commonly used ultrasound data acquisition for medical applications includes focused beams, divergent beams, and plane wave beams. Most commercial ultrasound scanners employ focused beams for higher signal to noise ratio. Plane wave beams are used for ultrafast imaging applications such as Doppler and microvascular imaging. There is still a need in medical diagnostic ultrasound for an even faster frame rate. We propose a V-wave beamforming method that is twice faster in frame rate than plane wave beamforming method and, at the same time, has higher signal to noise ratio in center areas of an image domain thanks to its ability to direct acoustic energies towards the center. The ability to rapidly illuminate a large volume of tissues with ultrasound in-sonification with improved focusing capability, and properly image all echoes reflected from acoustic contrasts in the tissues makes the V-wave beamforming a very useful tool for diagnosing, to name a few, cardiovascular diseases, heart diseases, blood blockages, malignant cancers where blood flows are fast and plenty.

Statement of Contribution/Methods

A V-wave ultrasound beam is collected with a novel design of transmission pattern of a transducer whose elements are arranged in a linear or curved array. In the design of transmission pattern half of the elements on the transducer are used to transmit a local coherent wave in one inward direction and the other half of elements are used to transmit another local coherent wave in a conjugate direction. The two propagation directions are opposing each other and are symmetric with respect to the normal direction to the transducer face at the center. The transmit time advance (negative of time delay) is minimal (or zero) at the center and increases towards both edges with a time slope that is a function of the sine of a flex-angle (measured with respect to the array). If one holds the transducer horizontally, the left half of transmitters will illuminate the center and the right portion of the image domain, and the right half of transmitters will illuminate the center and left portion of the image domain. Compared to data acquisition using plane wave beams, only half the number of beams is required to achieve comparable image quality and resolution at the center part of the image domain, effectively doubling the frame rate of an ultrasound scanner. The beamforming of V-wave RF data includes the following steps:

1. Take one input time series (or data trace) of a V-wave beam at a given receiver

2. Perform necessary frequency filtering to protect the data from aliasing or wavelet distortion during beamforming, if desired

3. Spray the data along impulse response curves of the V-wave beam. Also compute necessary attributes such as a transmitter-receiver offset on the transducer, reflection angle at an image point, wavelet stretch, anti-aliasing frequencies etc.

4. Accumulate image contributions, with options to form partial images for generation of common image point gathers

5. Perform amplitude weighting for true reflection amplitude preservation, if required

6. Repeat steps (1) – (5) for all input data traces and all V-wave beams

7. Perform gather processing and coherent compounding to obtain the final image

Results/Discussion

The first example is a numerical phantom study using a modified version of Fresnel Simulator from Ultrasound Toolbox (USTB, https://www.ustb.co). The use of this simulator is subject to the citation rule on its website. The phantom model contains: two rectangular boxes with a depth range between 7 – 9 mm; 4 flat continuous reflectors at 20mm, 40mm, 60mm and 80mm depth; a hyperechoic target of 8mm radius at 70mm depth and a second hyperechoic target of 6 mm radius at 50mm depth; a row of scatter points at 30mm depth and a column of scatter points at the center of the model. We have simulated 37 V-wave beams with flex angles ranging from 0 to 32 degree as well as 74 planewave beams with tilt angles from -32 to 32 degree. Fig. 1(a) is an image of the 74 planewave beams. Fig. 1(b) is another image of the 37 V-wave beams. We see similar resolution and image quality between the two images. In the central part of the image domain the image using V-wave beams looks stronger because of the enhanced focusing effect.

The second example is an in-vivo study of upper trachea and thyroid lobes performed in two separate scans. Verasonics Vantage ultrasound system (Verasonics Inc., Kirkland, WA, USA) equipped with a linear array transducer (6.5 MHz GE L3-12D, GE Healthcare, Wauwatosa, WI, USA) is used in both scans. Fig. 1(c) is produced using 256 focused beams on an 18-year-old healthy volunteer. Fig. 1(d) is produced using 64 V-wave beams on a 45-year-old healthy volunteer. Trachea and thyroid are well imaged in both scans. Carotid arteries on the edges are well imaged using focused beams and a hint of them can be seen on the image using V-wave beams. This is as expected: 256 focused beams walking from left to right uniformly in Fig. 1(c), and in Fig. 1(d) 64 V-wave beams all directing acoustic energies towards the center area. As a matter of fact, a lot of small blood vessels in thyroid can be seen on the image using V-wave beams even if the number of beams is much less (<u>more resolution with much less beams</u>). This demonstrates a utility of V-wave beams for clinical applications where a high frame rate and higher image resolution are both warranted. Such applications include color Doppler, power Doppler, microvascular imaging (MVI), elastography, and ultrasound localization microscopy (ULM). The downside is, of course, loss of signal strength on the edges of an image.

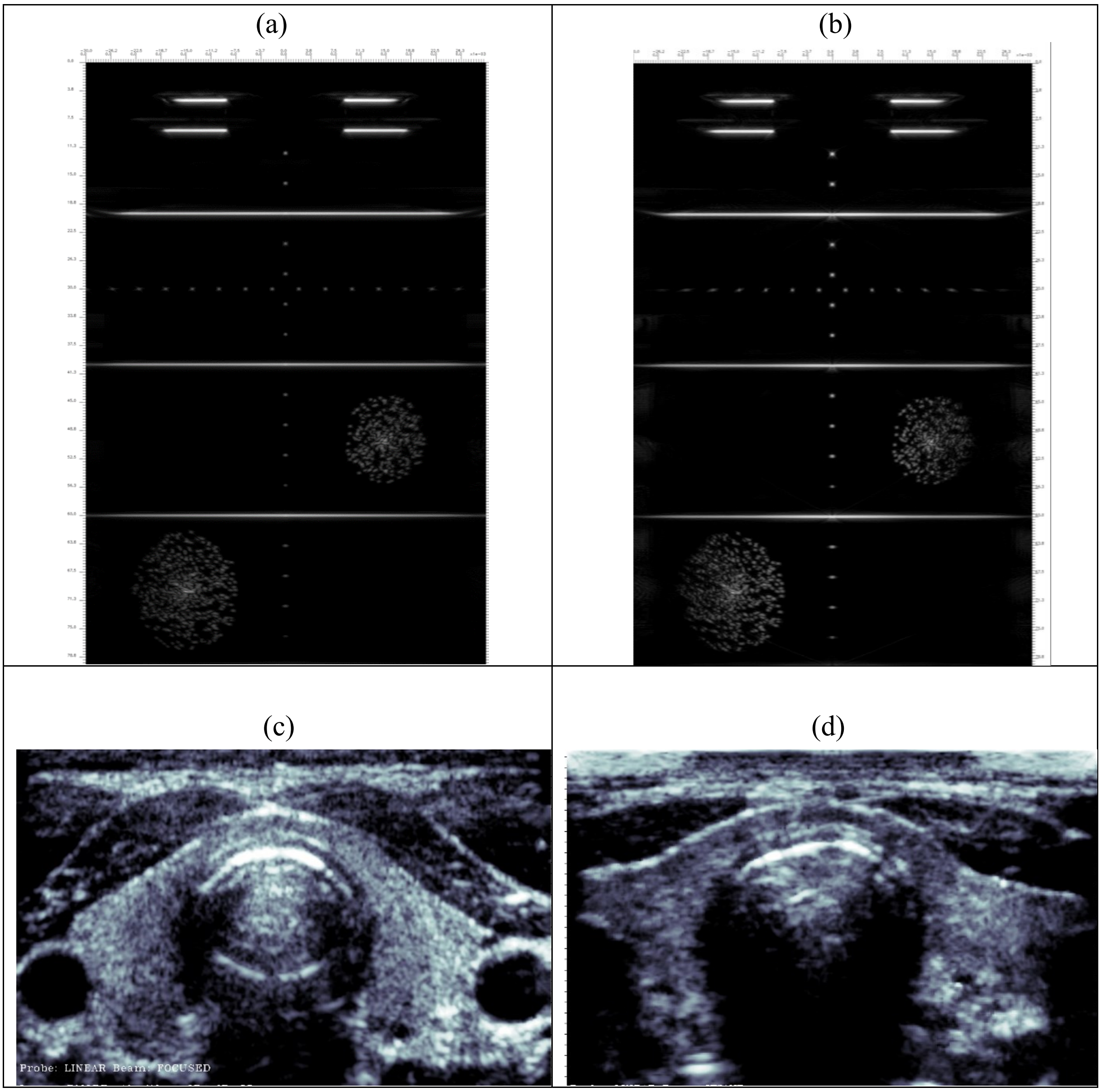

Fig. 1