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Beamforming and Imaging Approaches for Array-Based Dual-Frequency Acoustic Angiography

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***Abstract*— Acoustic angiography (AA) using dual frequency (DF) ultrasound systems enables high-resolution and high-contrast imaging of the microvasculature within tissue. AA has been used to detect changes in vasculature due to cancerous tumours, and it has the potential to aid the evaluation of treatment outcomes. DF ultrasound imaging uses conventional frequency ultrasound transducers (~2-4 MHz, LF) on transmit (Tx), and high frequency (HF) transducers on receive (Rx) to detect higher order harmonics (~10-20 MHz) from microbubble contrast agents. In this paper, we show a DF array, with LF and HF arrays integrated within the probe head that overcomes limitations of the previously used single channel DF probes such as the limited focal depth. The integrated DF array has been connected to a configurable micro-US beamforming system to allow plane wave on Tx and parallel Rx in both HF and DF imaging modes. This work presents DF and HF imaging modes with plane wave imaging techniques to visualize a contrast-filled channel within a tissue-mimicking phantom.**

Keywords— dual-frequency, acoustic angiography, plane wave, compounding, beamforming

I. Introduction

Superharmonic imaging employs high order harmonics generated by the microbubble (MB) contrast agents for improved contrast-to-tissue ratio (CTR) compared to the conventional contrast imaging, which utilizes the second harmonics. Depending on their size, MBs are typically excited by low frequency (LF) pulses, around ~1-4 MHz, and could produce broadband signals at high frequency (HF, ~10-20 MHz) including superharmonics. Covering such a wide frequency band from 1 MHz up to 30 MHz is very challenging using one array for both transmit (Tx) and receive (Rx) and would be at the cost of element sensitivity, which in turn degrades the signal-to-noise ratio (SNR). Therefore, a dual-frequency (DF) transducer that uses separate LF and HF transducer arrays for transmission and reception is of interest. This allows the transducer to collect signals with a reasonable SNR over a broad bandwidth, and it allows better separation of the nonlinear MB signals from the tissues i.e. the Rx array does not detect the linear or 2nd harmonic signal from tissue.

There have been several DF transducer designs proposed. Bouakaz et al. developed phased array for diagnosic purposes with improved contrast where the interleaved odd and even elements are used to receive and transmit respectively with centre frequencies of 2.8 MHz and 0.9 MHz [1]. Later, Kruse and Ferrara investigated flow or targeted molecular imaging in clinical applications using two separate transducers for transmission (at 2.25 MHz) and reception (at 15 MHz) [2]. They observed a higher resolution, better rejection of tissue echoes and harmonics due to nonlinear propagation, and deeper penetration compared to the single-frequency imaging at 15 MHz. More recently, Hu et al. designed and fabricated a co-linear transducer that could transmit at 1.5 MHz on LF and receive at 5.4 MHz on HF, and they have tested the transducer for molecular imaging using targeted MBs [3].

Current broadband DF transducers used for acoustic angiography use single-element for each of the transmitting and receiving transducer elements. Lukacs et al. developed a broadband single-channel DF transducer with centre frequencies of 4 MHz and 25 MHz for LF and HF elements respectively [4]. The two elements are arranged confocally with the inner element being the HF transducer and the outer element being the LF transducer. This scanhead can be steered mechanically in the

azimuthal direction and has been demonstrated for imaging microvasculature in rat kidney and rat fibrosarcoma, including with molecular targeted contrast agents [5], [6]. It has been found that the CTR in rat kidney imaging has been improved and an image resolution of ~170 μm was achieved in rat fibrosarcoma imaging. Our team has developed a hybrid DF probe (Tx: 1.7 MHz, Rx: 20 MHz) with one LF transducer pair and a commercial HF transducer (MS 250, FUJIFILM-VisualSonics) and demonstrated its capability for superharmonic imaging on 200 μm cellulose tubes [7]. Though an array transducer was used on Rx, the useful FOV is limited by the fixed focus of single-element transducers, and 2D and 3D image frame rates are limited by the necessary mechanical translation. The limitations arising from single-element transducers could be resolved by developing an array-based DF imaging system with electronic beam steering and focusing.

In this study, we present the dual-mode imaging capability of an array-based DF (Tx: 2.1 MHz, Rx: 21.3 MHz) probe that has been designed and developed in collaboration with FUJIFILM-VisualSonics. Apart from being able to perform conventional HF B-mode scans, plane wave imaging with steering and compounding has been implemented for both the LF and HF transducers, aiming to show the potential of reconstructing both high-contrast and high-resolution images of both contrast agents and tissue-mimicking materials. Preliminary results are reported from in vitro HF plane wave imaging on wire targets and DF plane wave imaging on contrast flow in tissue-mimicking phantom.

II. Methods

A. Transducer Characterization

The HF array has 256 elements in total, and half of the aperture(128 elements) was used for each plane wave transmit. HF and LF plane wave acoustic pressure fields were measured at steering angles of 0°, ±5°, ±7°, ±10°, and ±15° by applying element-specific delays. A programmable imaging system (Vantage 128, Verasonics, WA, USA) was used to drive the LF array on Tx to measure acoustic fields. Both one-cycle and two-cycles waveforms at 2.1 MHz was used in LF measurement. Data were acquired using a 40 μm aperture calibrated needle hydrophone (NH0040, Precision Acoustics Ltd, Dorchester, UK). A configurable micro-ultrasound imaging system (FUJIFILM-VisualSonics, Toronto, Canada)) was used to drive the HF array stack, and an 85 μm aperture calibrated needle hydrophone (HGL0085, ONDA, Sunnyvale, CA) was used for data acquisition. A one-cycle waveform with a centre frequency of 21.3 MHz was used in HF measurement. Averaging has been applied in both LF and HF pressure measurements to increase SNR.

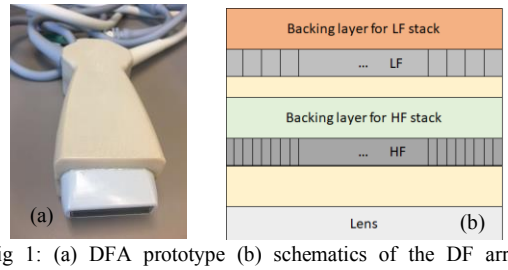

Fig 1: (a) DFA prototype (b) schematics of the DF array geometry.

B. DF Contrast imaging in a tissue mimicking phantom

A tissue-mimicking phantom containing a 1.1 x 1.1 mm^2 square wall-less channel was used for DF and HF imaging of contrast. The tissue-mimicking material was 2% (w/w) agar (BD Difco™ Agar Technical, Ref. 281210, Fisher Scientific Canada) with 1% (w/w) silica (S-5505, Sigma-Aldrich, St Louis, MO, USA). MicroMarker® (FUJIFILM-VisualSonics, Toronto, Canada) microbubble contrast agents were diluted using phosphate-buffered saline (PBS) to a concentration of 2×10^6 MBs /mL and were fed into the channel by gravity. Continuous stirring was applied to assure uniform MB concentration during data acquisition.

Plane wave imaging with the HF array was performed using the configurable micro-ultrasound imaging system (FUJIFILM-VisualSonics, Toronto, Canada). Coherent compounding was implemented with 9 steering angles and 64 parallel receive channels were used for reception upon each transmit. A total of 4 Tx/Rx events were acquired per image frame at each single steering angle. A one-cycle waveform at a centre frequency of 21.3 MHz was used at all steering angles, and sampling on receive more than $2\times$ Nyquist frequency. DF imaging was performed with two configurable micro-ultrasound imaging systems connected together, one to drive the LF array, and the other to operate the HF array. A two-cycle waveform at 2.1 MHz was used on LF transmit. All 32 LF elements were used on Tx for flat transmit (0° steer) for the presented results. Meanwhile, the HF system acquired the backscattered acoustic signals.

Unbeamformed radio-frequency (RF) channel data were acquired from the HF imaging system for image reconstruction. Frequency spectra of the RF data was analyzed to design the proper filter. A low-pass finite-impulse-response (FIR) filter (< 30 MHz) was applied to raw RF data, and data were beamformed at f-number of 1.5 in Rx focusing. No apodization or gain compensation were applied to the channel data. Beamformed HF data were coherently summed at all steering angles.

III. Result and Discussion

A. Transducer Characterization

The HF beams have less than 2 dB pressure variation across the beam width for each steering angle and to 28 mm depth. The focal depth was at 10 mm (-6 dB beam width: 0.7 mm) as shown in Fig. 2 when driven at the standard voltage for micro-

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ultrasound imaging. The pressure at the elevation focus was 1.89 MPa at 0° steering and 1.14 MPa at 15° steering.

The LF pressure profiles (Fig. 3) had less than 5 dB pressure variation across the beam width for all steering angles and depths to 25 mm, but they have stronger edge effects than the HF arrays. With 0° plane wave transmission, the LF pressure maps show a focal depth of 5 mm (-6 dB beam width: 2 mm). Driven at 51 Vpp, the pressure across the central, most uniform portion of the 0° beam was 190 kPa at 5 mm depth, decreasing to 150 kPa for ±15° steering. These pressures have previously been shown to produce high harmonic response from MBs for DF imaging. These results are promising for DF compound plane wave imaging.

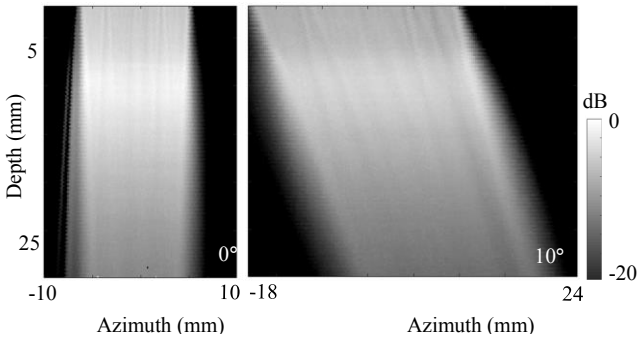

Fig. 2: Flat and steered HF plane wave (0° and 10° steer) over 28 mm depth. Pressure amplitude normalized to the peak at flat plane wave.

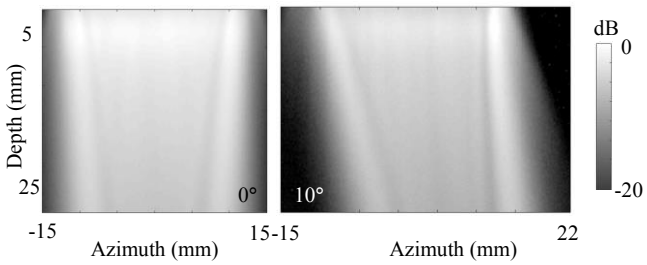

Fig. 3: Flat and steered LF plane wave (0° and 10° steer) over 28 mm depth. Pressure amplitude normalized to the peak at flat plane wave.

B. DF Contrast and HF Image of Phantom

The coherently compounded HF plane wave image of the phantom has been reconstructed using 9 angles, and the DF contrast image, reconstructed from a single, flat transmit (0°), is overlaid in false colour, as shown in Fig. 4. The image is displayed with a dynamic range of 60 dB for the HF image and 20 dB for the DF contrast image. The CTR was calculated from the HF image, defined as $CTR^{HF} = 20log_{10}(\frac{\overline{S_{MB}}}{\overline{S_m}})$, where $\overline{S_{MB}}$ and $\overline{S_m}$ are signals from in the contrast-filled channel and backscatter from the tissue mimicking matrix outside the channel but at the same depth. CTR^{HF} is estimated to be 4.18 dB. The CTR for the DF overlay is estimated from the SNR since away from the contrast-filled channel the signal minimum is limited by noise, not backscatter from the tissue mimicking material. Similar to the HF calculation, the SNR^{DF} is computed to be 21.79 dB. This is a factor of 5.2 improvement in CTR.

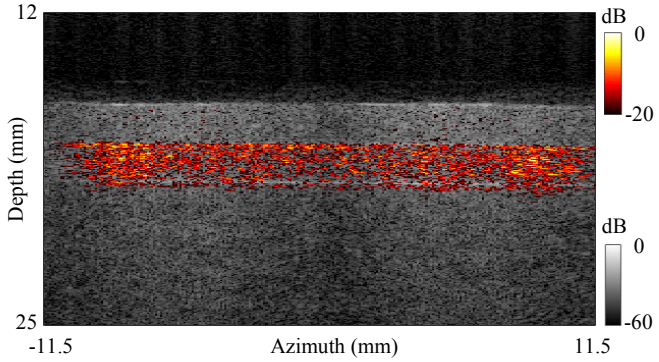

Fig. 4: Fused image of contrast flow within tissue-mimicking matrix phantom. Color-rendered image from DF contrast imaging mode and gray-scale image from HF plane wave imaging with beam steering and compounding.

IV. Conclusion

We have implemented both HF and LF plane wave imaging with steering on Tx and compounding on Rx with a configurable micro-ultrasound system. The acoustic pressure profiles have been measured to confirm beam uniformity and sufficient pressure for MB contrast excitation. We have presented in vitro imaging of MB contrast flowing in a tissue-mimicking phantom using both HF and DF plane wave imaging modes. The quantified SNR^{DF} and CTR^{HF} shows that array-based DF imaging using a single plane wave LF transmit can produce contrast images with SNR suitable for acoustic angiography. We conclude that array-based DF plane wave imaging has potential for visualizing both microvasculature and the surrounding tissue.

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