Multi-line Transmission for 3d Ultrasound Imaging: An Experimental Study

Emilia Badescu, Denis Bujoreanu, Lorena Petrusca, Denis Friboulet, Hervé Liebgott

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Emilia Badescu, Denis Bujoreanu, Lorena Petrusca, Denis Friboulet, Hervé Liebgott

Univ Lyon, INSA-Lyon, Université Claude Bernard Lyon 1, UJM-Saint Etienne, CNRS, Inserm, CREATIS UMR 5220, U1206, F-69100, LYON, France

Email: emilia.badescu@creatis.insa-lyon.fr

Abstract — **Achieving a high frame rate in echocardiography is highly important for quantifying the short phases of the cardiac cycle that contain valuable information for medical diagnosis. Additionally, the 3D quantitative assessment of the heart would significantly improve the current measurements used in daily clinical routine. Nevertheless obtaining ultrafast images remains a challenge due to the trade-off between the image quality and a high frame rate, especially when volumetric data is acquired. Among the current ultrafast imaging methods, multi-line-transmit imaging (MLT) provides an increased frame rate but in the same time mostly preserves the image quality. In this paper we present the first real-time experimental implementation of the MLT in 3D ultrasound. The results indicate the potential of 3D MLT for achieving high contrast and resolution while increasing the frame rate. This study thus demonstrates the feasibility of 3D MLT in real-time and extends its possible applications to dynamic cardiac imaging.**

Keywords—ultrasound, high frame rate, multi-line trasmit, 3D imaging, experimental validation

I. INTRODUCTION

Ultrafast imaging is highly important in echocardiography for capturing rapid events that could be significant for diagnosis. Even if 3D ultrasound benefit from an increased clinical interest due to the possibility to assess accurate quantitative measurements, achieving a high frame rate remains a real challenge in 3D imaging [1].

Conventionally, with single focused transmissions, the frame rate can be increased by narrowing the field of view or diminishing the line density, which implies reducing the number of transmit events. In any case, the increase in frame rate comes with a compromise in terms of image quality (reducing the sector size or spatial resolution). As a solution to this inconvenient, multi-line acquisition (MLA) has been proposed and implemented on most of the clinical scanners [2]. This method is based on reconstructing multiple image lines in parallel from one emitted beam which results in a gain in frame rate related to the number of parallel acquisitions. However, MLA requires the broadening of the transmit beam which can be obtained by reducing the aperture size and therefore the transmitted energy. A lower transmit energy compromises the image quality by diminishing the signal to noise ratio (SNR).

Other approaches propose using electrocardiographic (ECG) retrospective gating for combining several subsectors over several cardiac cycles [3], [4]. But this technique may fail in patients presenting significant difference between cardiac cycles [1]. Despite the improvement in frame rate, these methods are still limited in current 3D implementation at frame rates on the order of a few tens of Hz.

Several approaches have been proposed in literature to increase further the frame rate such as plane/diverging wave imaging [5], [6]. However, the high temporal resolution is achieved to the detriment of the spatial resolution. Even though spatial compounding allows coping with this limitation [7], the compounding reduces the frame rate.

Alternatively, several studies showed that multi-line-transmit imaging (MLT) provides an increased frame rate but in the same time mostly preserves the image quality. However, most of existing MLT analysis focused on 2D ultrasound imaging [8]–[11] or they were limited to 3D simulations [12]. Moreover, the latest implementation of this method in 3D, even if tested experimentally, is based on generating the MLT data synthetically by summing up the raw data before beamforming [13].

In this paper we present the first implementation of a real-time 3D MLT acquisition scheme on a research ultrasound system. Our aim is to prove its feasibility and to compare it with conventional focused imaging

II. METHODS

A. Multi-line transmission

The increase in frame rate is achieved by transmitting simultaneously several steered focused beams as described in 2D by Tong et al. [14]. In order to study the compromise between the high frame rate and image quality the number of MLT was varied from 4 to 8 and 16. The transmission scheme of the three transmission set-ups is represented in Figure 1. As a general principle, the angular aperture is divided into 32 angles ($\alpha_Y, Y = 1..32$) between [-30º, 30º] in elevational direction and 32 angles in azimuthal directional

($\theta_X, X = 1..32$). Considering n the number of simultaneous beams (n=[4,8,16]), for each set of $nTX = 32/n$ transmissions θ_X is kept fixed while α_Y is varied according to $\alpha_{iTX}, \alpha_{iTX+nTX}, ..., \alpha_{iTX+(n-1)\cdot nTX}$, where $iTX = 1..nTX$. Therefore, for each transmission the following pairs of angles are used to focalize simultaneously in n different directions: $(\theta_j, \alpha_{iTX}), (\theta_j, \alpha_{iTX+nTX}), ..., (\theta_j, \alpha_{iTX+(n-1)\cdot nTX})$, where θ_j is changed for each set of nTX transmissions in the same order as the angles in the elevational direction.

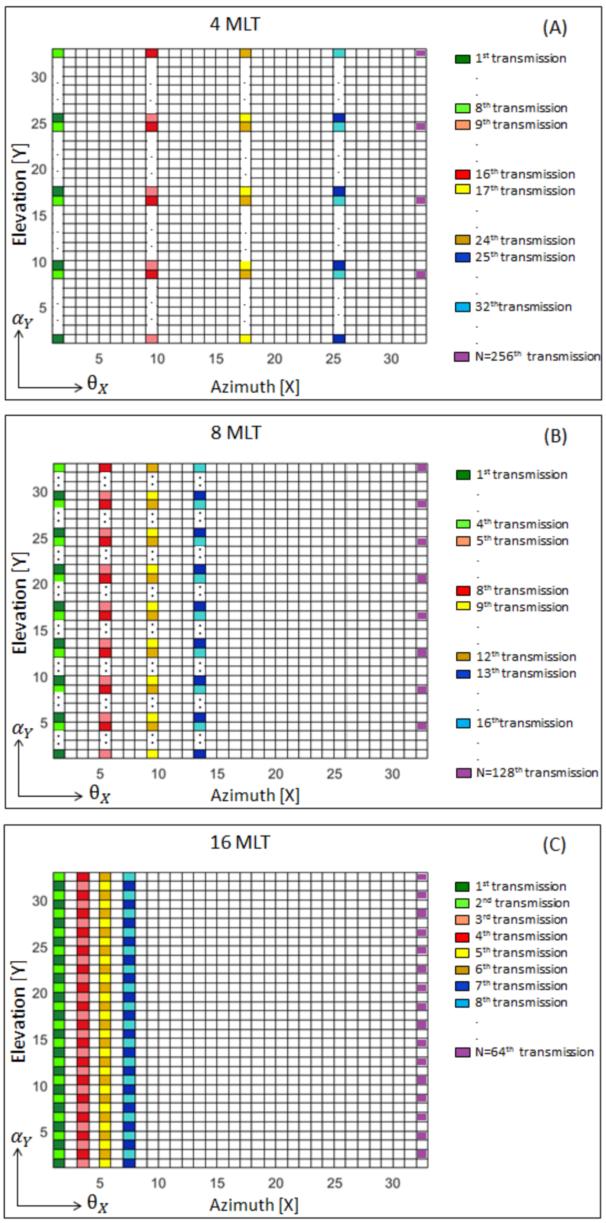

Figure 1: Transmission scheme for 4 MLT (A), 8 MLT (B) and 16 MLT (C)

In order to transmit multiple focused beams simultaneously, a superposition of waveforms needs to be applied to each element of the transducer. The Verasonics system allows the user to define arbitrary waveforms by using pulse codes [15]. Yet, as the number of multiple transmission increases, the number of superposed waveforms increases. Consequently, the pulse which codes the resulting waveform becomes more complex, requiring long transmits sequences that compromise the (thermal) safety of the imaged media and of the transducer. A solution for dealing with safety issues is to use short waveforms in transmission.

However, the consequence of using Dirac coded waveforms in transmission instead of longer waveforms such as a 3 cycle sinusoidal signal is illustrated in Figure 2, for a focused transmission.

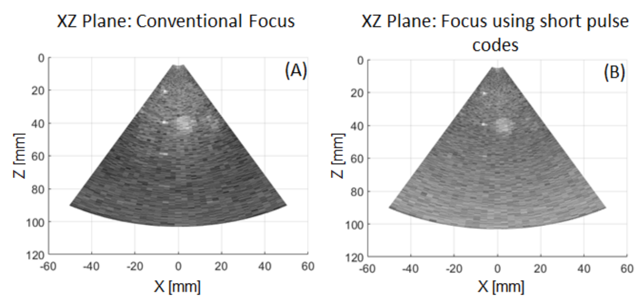

Figure 2: Focus sequence: XZ plane obtained by using 3 cycle signal (A), short coded pulse (B)

As it can be observed, using short pulse codes results in an improvement of axial resolution but with the tradeoff of lower contrast. The contrast to noise ratio (CNR) was calculated between the hyperechoic cyst placed at the focal point and the background, according to the expression:

$$CNR = 20\log_{10}\left(\frac{\left|\mu_{bck} - \mu_{cyst}\right|}{\sqrt{\sigma_{bck}^2 + \sigma_{cyst}^2}}\right) \quad (1)$$

where μ_{bck}, μ_{cyst} are mean and σ_{bck}^2, σ_{cyst}^2 are the corresponding variances of the background and the cyst regions calculated for a B-Mode image, having a dynamic range of 60 dB.

The CNR decreased from 7.48 dB to 5.03 dB when short pulse codes were used. This is due to the low energy carried by the short excitation waveforms that results in a low Signal to Noise Ratio (SNR) for the received signals. The contrast degradation is more noticeable as the distance to the focus and the penetration depth are increased. For example, the media points placed at 80 and 90 mm are no longer visible for the image obtained using pulse codes.

Since our acquisition system allows the implementation of MLT just by using short pulse codes, for a fair analysis the 3D MLT acquisitions were compared in this study with a focused transmission obtained also using short excitation signals, as the one shown in Figure 2 (B).

B. *Experimental set-up*

Data acquisition was performed by using 4 Verasonics systems synchronized to drive a 32x32 elements Vermon probe having a central frequency of 2.97MHz [16]. The pitch was 0.3 mm and the images were acquired using a sampling frequency of 11.9 MHz. A volume of 32ºx32º was insonified having a depth of 10.35 cm. The focal point was set to 4 cm.

III. Results

The feasibility of 3D MLT was evaluated based on the data acquired on a CIRS ultrasound phantom (Model 054GS). Figure 3 illustrates three sections of its 3D volume, obtained by steering 4 focused beams simultaneously.

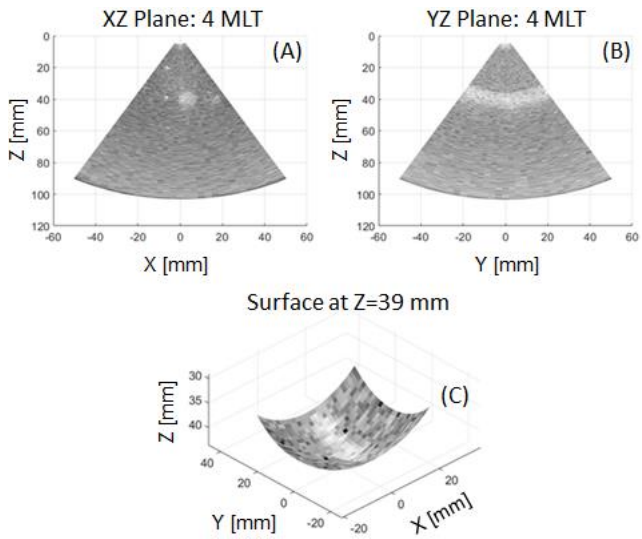

Figure 3: 3D results for 4 MLT.
(a) XZ and (b) YZ reconstructed planes and (c) a surface at a constant depth (Z=39 mm)

Rectangular (i.e. no) apodization was used on both transmitting and receiving elements. For comparison, the same apodization was used for the conventional focused sequence.

Emitting several beams in the same time leads to non-negligible cross-talks artefacts, which become visible as the number of transmissions increases. Such an artefact is presented in Figure 4 (D) for the media point placed at 20 mm when 16 MLT was used.

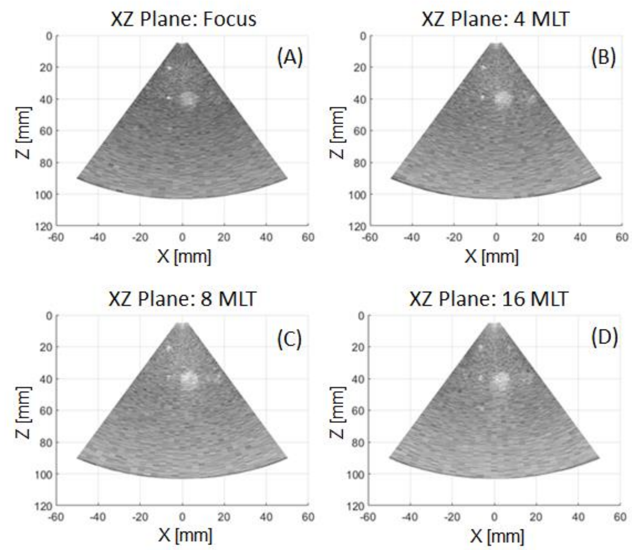

Figure 4: Comparison XZ planes for focus (A), 4 MLT (B), 8 MLT (C) and 16 MLT (D)

As shown in Figure 4, the image quality degrades with the number of simultaneous transmissions. For evaluating the degradation of the images, we calculated the CNR and the resolution which was computed using full width at half maximum (FWHM) at –6 dB. The contrast was investigated for a hyperechoic cylinder, whilst the resolution was measured for a media point, both placed on the focal point (4 cm).

While for 4 MLT the contrast has a value (4.83 dB) close to the one obtained using a conventional focused transmission (5.03 dB), it is reduced by 40% for 16 MLT. In the same time, the axial resolution degrades from 0.65 mm to 3.55 mm with a 16-fold increase in frame rate. All the contrast and axial resolution values are reported in Table I.

Table I: Axial Resolution and contrast values

Sequence/ Image quality	Focus	4MLT	8MLT	16MLT
Axial Resolution	0.65mm	1.25mm	2.1mm	3.55mm
Contrast	5.03 dB	4.83 dB	4.05dB	3.61dB

IV. Discussion and Conclusion

The contrast and resolution values obtained when transmitting 4 focused beams simultaneously were comparable with those of a 3D focused sequence obtained by using the same emission parameters. The frame rate can be increased further but with the cost of the image quality. However, different methods that could improve the image quality and reduce the cross-talks even for a high frame-rate have been proposed in literature. Such techniques focused on aligning the transmit directions along the transverse diagonal of the transducer [12], using a proper apodization [13], using different frequency bands for the transmitted beams [17], [18] or the minimum variance beamformer [10], [19]. Most of these studies were conducted in 2D or in simulations in 3D. The exploration of these directions in experimental 3D will be the topic of future research.

The results presented in this paper indicate the experimental feasibility of MLT in 3D and its potential for achieving high contrast and resolution while increasing the frame acquisition rate, even for a simple transmission scheme and when no apodization is used. The real-time 3D MLT implementation presented in this study extends its possible applications to dynamic cardiac imaging.

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