Program Digest, 2019 IEEE International Ultrasonics Symposium (IUS)
Glasgow, Scotland, October 6-9, 2019

Multi-plane-transmit (MPT) Volumetric Imaging based on A Matrix Array: Experimental Validation

Yinran Chen[1, 2], Mathieu Pernot[3], Clement Papadacci[3], Jan D'hooge[4], and Jianwen Luo[2]
[1]Department of Computer Science, Xiamen University, Fujian, China
[2]Department of Biomedical Engineering, Tsinghua University, Beijing, China
[3]Physics for Medicine, Inserm, ESPCI Paris, PSL Research University, Paris, France
[4]Department of Cardiovascular Sciences, KU Leuven, Leuven, Belgium
Email: luo_jianwen@tsinghua.edu.cn

***Abstract*—Multi-plane-transmit (MPT) is a matrix array based 3D beamforming methodology, which combines the features of diverging wave compounding (DWC) and multi-line-transmit (MLT). The MPT beams are diverging in one lateral dimension, while focused in the other lateral dimension, leading to a plane-wise reconstruction process. As a result, compounding is performed within the beam planes, while parallel transmit scanning is performed across the planes. In our previous work, computer simulations have proved that the proposed MPT setup outperforms 3D DWC and 3D MLT at a similar frame rate particularly for moving targets such as the heart. In this work, the first experimental validation experiments are conducted on a standard phantom to further investigate the performance of MPT. The experimental results show that the proposed MPT setup has better contrast ratio (CR) and contrast-to-noise ratio (CNR) when compared with 5 × 5 DWC, while frame rate is similar. Furthermore, the performance of different MPT setups was also investigated. These preliminary phantom experiments demonstrate the advantages of MPT beamforming.**

Keywords—3D ultrasound imaging, beamforming, multi-plane transmit, phantom experiment

I. Introduction

High temporal resolution and three-dimensional (3D) imaging are two frontiers in the research community of ultrasound engineering [1, 2]. The former is necessary in order to capture fast-moving events, for example, the motion induced by myocardium contraction and relaxation, and the propagation of electromechanical and shear waves [3]. The latter can provide more comprehensive anatomical and functional information due to the additional dimension, for example, the 3D images of the heart and fetus [2].

Two main methodologies are proposed to increase the frame rate, i.e., diverging wave imaging (DWI) and multi-line-transmit (MLT). Compared with the conventional single-line-transmit (SLT) beamforming, DWI broadens the transmit beam by placing the transmit focal point behind the array (i.e., the virtual source) [4]. Parallel receive beamforming, or multi-line-acquisition (MLA) is used to reconstruct several lines within the beam at the same time to increase the frame rate. Furthermore, sequentially transmission of multiple virtual sources and spatial coherent compounding in receive beamforming, i.e., diverging wave compounding (DWC) can improve the image quality at the expense of frame rate. In 3D ultrasound imaging, i.e., 3D DWC, the virtual sources can also be placed behind the matrix array to transmit a 3D diverging wave, as illustrated in Fig. 1(a). The whole volume is then beamformed according to each virtual sources. The classical setups of 3D DWC includes 3 × 3, 5 × 5, and 9 × 9 virtual sources [5].

MLT is an alternative methodology for high frame rate imaging by transmitting multiple focused beams in parallel, with each beam scanning a sub-sector [6, 7]. Recently, MLT was also extended to matrix array based 3D ultrasound imaging, with the sub-volumes being reconstructed in parallel, as illustrated in Fig. 1(b). MLA can also be combined with MLT by beamforming several lines surrounding each transmit beam to further increase the frame rate. The classical setups of 3D MLT include 9MLT-4MLA and 16MLT-4MLA [8].

Both 3D DWC and 3D MLT suffer from the trade-offs between frame rate and image quality. The volume-by-volume reconstruction of 3D DWC can reach ultra-high frame rate at the expense of image quality. The line-by-line reconstruction of 3D MLT can mostly preserve image quality, but only a limited gain in frame rate can be achieved. In our previous work, multi-plane-transmit (MPT) beamforming was proposed by reconstructing the volume plane-by-plane, as illustrated in Figs. 1(c) and 1(d) [9]. More specifically, one of the MPT beams, i.e., single-plane-transmit (SPT) beam, is diverging in one lateral dimension of the matrix array, while focused in the other lateral dimension. When running at a similar frame rate in computer simulations, the proposed MPT setup has higher lateral resolution and competitive contrast-to-noise ratio (CNR) when compared with 3D DWC and competitive image quality metrics with 3D MLT [9].

However, no attempt has been made to validate the feasibility and performance of MPT in a real ultrasound system. The objective of this study was thus to perform the first experimental validation of MPT beamforming by imaging a standard phantom. A preliminary comparison between MPT and 3D DWC was also conducted in the phantom experiments.

II. Methods

A. Principle of MPT

The transmit time delays of one of the SPT beams are illustrated in Fig. 2. Two time delay components, which generate a diverging wave in the azimuth dimension and a focused wave in the elevation dimension, respectively, are summed up to generate a unique (saddle-shaped) time delay profile. Based on the Huygens' principle, the resulting transmit beam is diverging in the azimuth dimension, while focused in

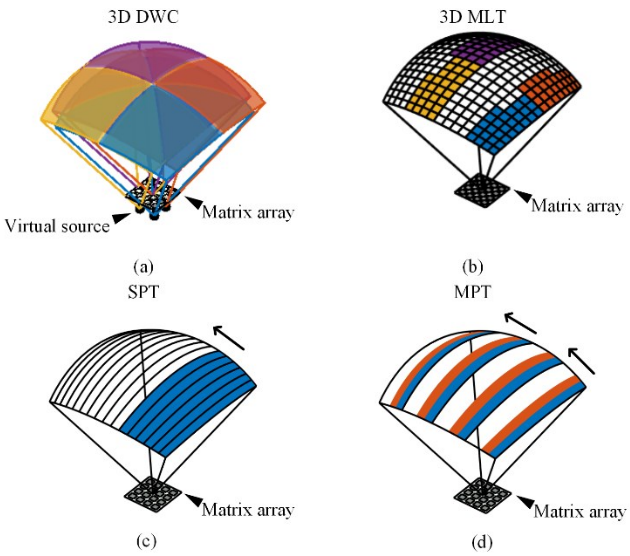

Fig. 1. Matrix array based volume reconstruction scheme of (a) 3D DWC, (b) 3D MLT, (c) SPT, and (d) MPT.

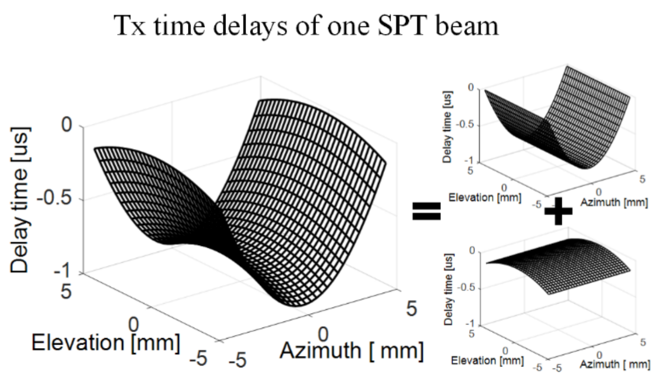

Fig. 2. Generation of the transmit time delay profiles of one of the MPT beams, i.e., one SPT beam. The unique profiles make the beam diverging in the azimuth dimension, while focused in the elevation dimension.

the elevation dimension, i.e., a *planar* diverging wave. In analogy to MLT, multiple SPT beams with a fixed inter-beam space can be transmitted into different planes, leading to parallel scanning in the elevation dimension. Compounding can also be performed within each plane in order to improve the image quality in the azimuth dimension. Furthermore, the concept of MLA can be transferred to MPT by reconstructing several planes corresponding to each transmit plane, in order to increase the frame rate. For example, a *m*MPT-*n*CMP-*p*MLA (CMP = compounding) setup means *n* virtual sources for compounding in the azimuth dimension, *m* parallel transmit beams in the elevation dimension, and *p* parallel received planes for each individual transmit plane. The coefficients, *m*, *n*, and *p* should be adjusted to maintain the balance between frame rate and image quality, as well as to keep image quality isotropic, i.e. similar in azimuth and elevation dimensions respectively.

B. Beamforming Setup

In this work, the beamformed volume was a rectangular pyramid with the apex located at the center of the matrix array. The 90 × 90 lines were evenly distributed in the volume, with an image depth of 100 mm and an opening angle of 60° in the cross-sections of azimuth-axial and elevation-axial planes. To cover this volume with such line density, a 3MPT-2CMP-2MLA setup is proposed, with a focal depth of 50 mm and an inter-beam spacing of 20° in the elevation dimension and opening angle of 60° in the azimuth dimension. As a result, 30 transmit events (i.e., 2 × 15 for azimuth × elevation) should be made, leading to a frame rate of ~166 Hz at a pulse repetition frequency (PRF) of 5 kHz. For comparison, a 5 × 5 DWC with opening angles of 60° × 60° requires 25 transmit events, which leads to a frame rate of ~200 Hz given the same PRF.

C. Phantom Experiments

Phantom experiments were performed on four synchronized Verasonics Vantage 256 systems (Verasonics, Kirkland, WA), controlling a matrix array with 32 × 32 effective elements, pitch size of 0.30 × 0.30 mm^2, and a center frequency of 3.0 MHz (Vermon, Tours, France). A standard tissue-mimicking phantom (model 054GSE, CIRS, Norfolk, VA) containing a series of wire targets and a hyperechoic cylindrical target was imaged.

Two 3D imaging sequences, i.e., 3MPT-2CMP-2MLA and 5 × 5 DWC were tested in the phantom experiments. The frame rates of the two sequences were similar for comparison. Please note that due to the restrictions of the system, only SPT beams without arbitrary waveform generator were generated and their corresponding channel data were restored. Consequently, the channel data of one MPT event were synthesized by summing up the channel data of respective SPT beams, according to the linear system theory. Additionally, different MPT setups were tested to evaluate their performance.

The contrast ratio (CR) and CNR of the hyperechoic targets, as well as the number of transmit events, which directly links to frame rate, were calculated for quantitative evaluation. The beam penetration was also compared by qualitative observation.

III. Results

A. Comparison between MPT and DWC

The three perpendicular cross-sections, i.e., the azimuth-axial, elevation-axial, and azimuth-elevation planes of the phantom are presented in Fig. 3. Figs. 3(a1) to 3(a3) show the results of 5 × 5 DWC, while Figs. 3(b1) to 3(b3) show those of 3MPT-2CMP-2MLA. The arrows indicate a much better penetration for MPT than for 3D DWC, showing that the wire targets at depth of ~80 mm remain visible. Otherwise, the average signal intensities of the background are slightly higher at a depth of ~50 mm in MPT, due to the focusing nature of the transmit beams, as shown in Figs. 3(b1) and 3(b2). However, it is highlighted that the image of the top wire target suffers from distortion due to the high overlap of the MPT beams in the near field of the matrix array. The quantitative evaluation shows that MPT has a CR and CNR of 13.29 dB and 10.31 dB respectively which is higher than those of 3D DWC, which were 11.40 dB and 9.73 dB respectively.

B. MPT with Different Setups

The zoom-in images of the phantom acquired with different MPT setups are summarized in Fig. 4. These setups include SPT, 3MPT, 5MPT, and 9MPT in the elevation dimension, in combination with 1CMP, 3CMP, and 5CMP in the azimuth

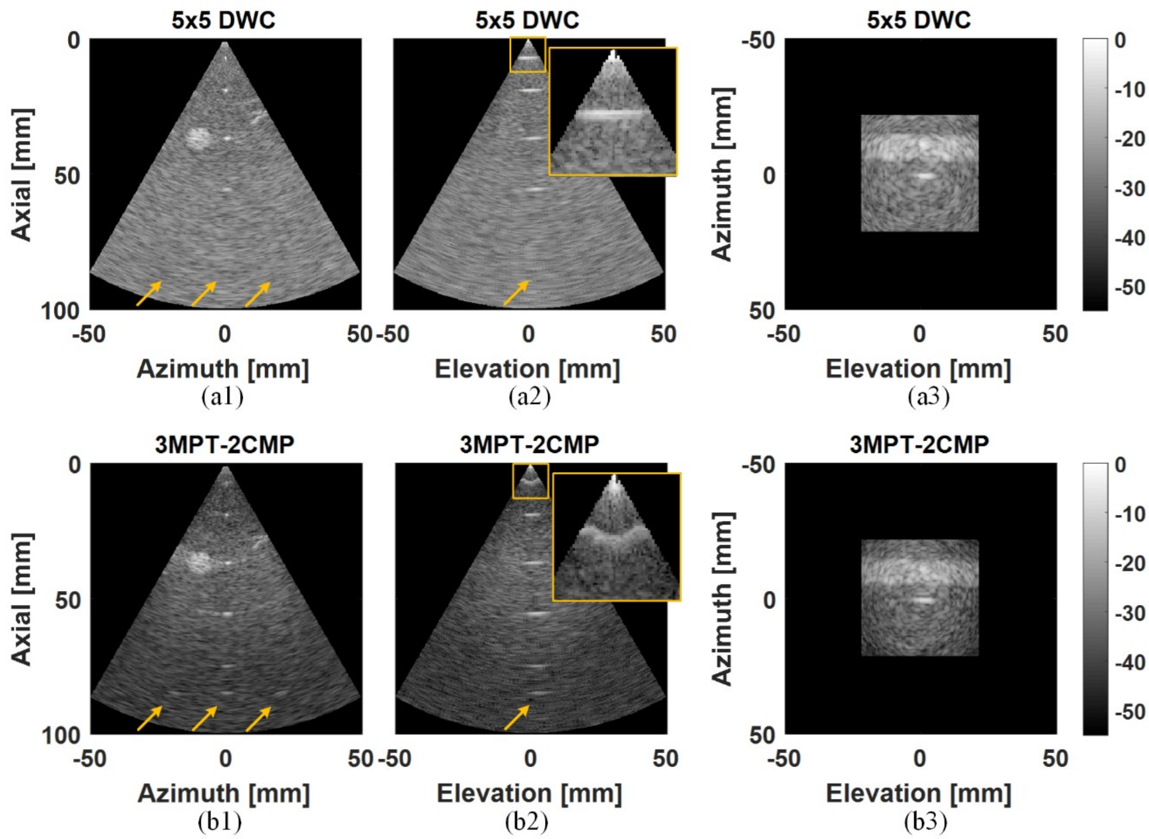

Fig. 3. Different cross-sections of the phantom acquired with 5 × 5 DWC and 3MPT-2CMP-2MLA. (a1-a3) the azimuth-axial, elevation-axial, and azimuth-elevation planes of 5 × 5 DWC; (b1-b3) the same cross-sections of 3MPT-2CMP-2MLA. The yellow arrows indicate the wire targets at depth of ~80 mm, while the yellow boxes highlight the top wire targets.

dimension. It shows that more MPT beams lead to higher level of crosstalk artifacts, and thus lower image quality. On the contrary, a larger number of compounding angles leads to higher image quality. Consequently, SPT-5CMP-2MLA has overall the best image quality, 9MPT-1CMP-2MLA has the worst image quality, as illustrated in Figs. 4(a3) and 4(d1), respectively.

IV. Discussion

In this paper, the first experimental validation of MPT beamforming based on a standard phantom was performed with 3D DWC taken for comparison. The results show that under similar frame rate, the proposed MPT setup outperforms 3D DWC in terms of CR and CNR. Moreover, MPT with different setups were also compared.

A. *Analysis of MPT*

In the phantom experiments, the proposed MPT setup, 3MPT-2CMP-2MLA, has much better beam penetration than that of 5 × 5 DWC, as well as a better performance in terms of CR and CNR. This is mainly due to the one-dimensional focusing nature of the MPT beams. As a result, the signal-to-noise ratio of the backscattered echoes remain higher than that of 3D DWC, which spreads out the transmit energy to the whole volume. However, the number of MPT beams and the number of compounding angles should be carefully designed according to the specific application scenario.

B. *Limitation of the Experiments*

The present phantom experiments preliminarily validate the feasibility and performance of the proposed MPT beamforming. However, more aspects should be taken into consideration in the future. Firstly, the “real” transmit of MPT beams, which requires an arbitrary waveform generator in the system, is to be conducted better. Next, the total transmit energy of the compared imaging sequences should be equalized for a more fair comparison. Then, a more complex phantom, as well as a phantom in a dynamic situation, should be involved in future experiments. As a result, post-processing methods, like motion compensation and elastography could be validated [10]. Furthermore, acoustic safety should be investigated thoroughly before moving towards *in-vivo* application.

C. *Outlook of MPT*

MPT beamforming provides an alternative way to 3D DWC and 3D MLT in matrix array based 3D ultrasound imaging, showing potential in maintaining a better balance between frame rate and image quality. The possible application scenarios include but are not limited to 3D B-mode imaging of the heart, 3D myocardial elastography, and 3D blood flow imaging.

V. Conclusion

In this work, the first experiments of MPT beamforming were conducted in order to validate its feasibility and performance. The proposed MPT setup shows higher image

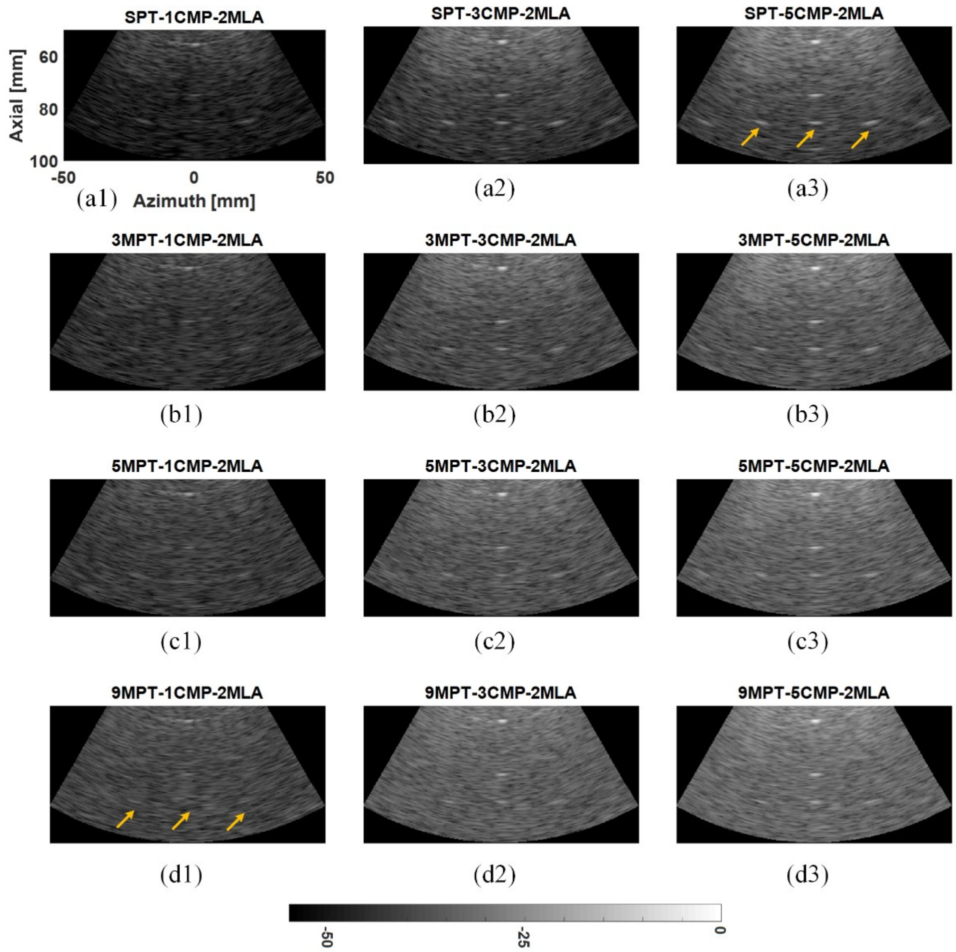

Fig. 4. Zoom-in images of the phantom acquired with different MPT setups, including SPT, 3MPT, 5MPT, and 9MPT in the azimuth dimension, and 1CMP, 3CMP, and 5CMP in the elevation dimension. The yellow arrows indicate the visible (a3) and invisible wire targets (d1) cases due to different level of crosstalk artifacts and number of compounding angles.

quality than 3D DWC when operating at a similar frame rate. The preliminary phantom experiments demonstrate that MPT is promising for high frame rate and high image quality 3D ultrasound imaging.

ACKNOWLEDGMENT

This study was supported in part by the National Natural Science Foundation of China (81471665, 81561168023 and 61871251).

REFERENCES

[1] M. Tanter and M. Fink, "Ultrafast imaging in biomedical ultrasound," IEEE Trans. Ultrasonics, Ferroelectrics, and Frequency Control, vol. 61, pp. 102-119, 2014.

[2] R. M. Lang, V. Mor-Avi, L. Sugeng, P. S. Nieman, and D. J. Sahn, "Three-dimensional echocardiography: the benefits of the additional dimension," Journal of the American College of Cardiology, vol. 48, pp. 2053-2069, 2006.

[3] L. P. Badano, R. M. Lang, and J. L. Zamorano, Textbook of Real-time Three Dimensional Echocardiography: Springer, 2011.

[4] G. Montaldo, M. Tanter, J. Bercoff, N. Benech, and M. Fink, "Coherent plane-wave compounding for very high frame rate ultrasonography and transient elastography," IEEE Trans. Ultrasonics, Ferroelectrics, and Frequency Control, vol. 56, pp. 489-506, 2009.

[5] J. Provost, C. Papadacci, J. E. Arango, M. Imbault, M. Fink, J.-L. Gennisson, et al., "3D ultrafast ultrasound imaging in vivo," Physics in Medicine and Biology, vol. 59, p. L1, 2014.

[6] L. Tong, H. Gao, and J. D'hooge, "Multi-transmit beam forming for fast cardiac imaging-a simulation study," IEEE Trans. Ultrasonics, Ferroelectrics, and Frequency Control, vol. 60, pp. 1719-1731, 2013.

[7] L. Tong, A. Ramalli, R. Jasaityte, P. Tortoli, and J. D'hooge, "Multi-transmit beam forming for fast cardiac imaging—Experimental validation and in vivo application," IEEE Trans. Medical Imaging, vol. 33, pp. 1205-1219, 2014.

[8] A. Ortega, J. Provost, L. Tong, P. Santos, B. Heyde, M. Pernot, et al., "A comparison of the performance of different multiline transmit setups for fast volumetric cardiac ultrasound," IEEE Trans. Ultrasonics, Ferroelectrics, and Frequency Control, vol. 63, pp. 2082-2091, 2016.

[9] Y. Chen, L. Tong, A. Ortega, J. Luo, and J. D'hooge, "Feasibility of multiplane-transmit beamforming for real-time volumetric cardiac imaging: A simulation study," IEEE Trans. Ultrasonics, Ferroelectrics, and Frequency Control, vol. 64, pp. 648-659, 2017.

[10] Y. Chen, J. D'hooge, J. Luo. Doppler-based motion compensation strategies for 3-D diverging wave compounding and multiplane-transmit beamforming: A simulation study [J]. IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control, 2018, 65(9): 1631-1642.