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

Harmonic Generation in Tissue with Matrix Arrays for 4D Cardiac THI

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***Abstract*— Combining 4D echocardiography and tissue harmonic imaging (THI) may improve image quality and consequently diagnosis of heart disease. Beamforming considerations such as methodology for sampling the volume with ultrasound beams, beam shape, and the magnitude of the 2nd harmonic component generated in tissue are important for 4D cardiac THI. No studies have yet investigated harmonic generation produced by diagnostic matrix arrays for 4D cardiac THI. The objective of the present work is to evaluate harmonic generation in tissue by matrix arrays by considering different 4D beamforming approaches with a numerical model based on the KZK equation. We evaluated the spatial extent, level of the produced 2nd harmonic component in tissue, and the volume rate of 4D cardiac THI performed by different beamforming approaches. We used different source pressures (applied voltage to the elements) that resulted in peak MI=1.9 in tissue. Beamforming approaches with shallow foci produced the highest peak 2nd harmonic in tissue (1.2 MPa) yet lowest volume rate (1.7 Hz). Beamforming approaches with deep foci produced the highest 2nd harmonic at imaging depth of 150 mm (0.3 MPa). The volume rate of 4D cardiac THI performed with focused ultrasound, PWI, and DWI is 1.7-20, 81, and over 1275 Hz, respectively.**

Keywords*— *echocardiography, tissue harmonic imaging, plane wave imaging, KZK, simulation, Ultrafast imaging.

I. INTRODUCTION

4D echocardiography has allowed physicians to examine the 3D structure of the heart in real time [1]. Current commercial 4D ultrasound systems rely on conventional beamforming techniques to sample the volume and are thus limited to a few volumes per second [2]. Recent developments in ultrafast imaging have improved the temporal resolution of 4D echocardiography to over 4000 volumes per second [2, 3], by sampling the volume with plane/diverging wave imaging (PWI/DWI), yet sacrificing the image quality.

Tissue harmonic imaging (THI) improves image quality and reduces image artifacts from phase aberrations and reverberations [4], which are often encountered in echocardiography. As ultrasound waves propagate in tissue, harmonic components are gradually generated due to nonlinear propagation. THI utilizes the second harmonic component in the backscattered echoes to form the image. Combining 4D echocardiography and THI will improve the image quality of 4D echocardiography and produce more accurate estimations of the physiology of the heart. However, to the best of our knowledge, no studies have yet investigated harmonic generation produced by diagnostic matrix arrays for 4D cardiac THI. Beamforming considerations such as methodology of sampling the volume with ultrasound beams, beam shape, and the magnitude of the 2nd harmonic component generated in tissue are important for evaluating the image quality and temporal resolution of 4D cardiac THI.

The Khokhlov-Zabolotskaya-Kuznetsov (KZK) equation has been used to study the nonlinear propagation of pulsed finite amplitude sound beams in thermoviscous fluids [5]. The equation describes the combined effects of diffraction, thermoviscous absorption, and nonlinearity. A recently developed KZK code that numerically solves the KZK equation in time domain and can model the field of arbitrary plane and diverging beams was used in present study [6].

The aim of the present study was to evaluate harmonic generation produced by a 2D (matrix) array with different beamforming approaches using the KZK numerical model we have previously developed. We have 2 specific objectives: first, to investigate the feasibility of combining 4D echocardiography and THI by quantifying the second harmonic component; and second, to investigate the temporal resolution of 4D cardiac THI performed by different beamforming approaches.

II. Methods

A. Matrix array with different beamforming approaches

A hypothetical matrix array which has the same overall physical dimensions as the Verasonics P4-2v array (19.2 × 14 mm) was modeled in present study. We evaluated the spatial extent and magnitude of the produced 2^{nd} harmonic component by this matrix array with different beamforming approaches. Six beamforming approaches in total using different azimuthal-elevation foci (AF-EF) were considered: (1) 75-75 mm, (2) 75-150 mm, (3) 150-75 mm, (4) 150-150 mm, (5) unfocused ("plane wave") in both planes, and (6) diverging wave in both planes. All approaches used different source pressure (applied voltage to the elements) that resulted in peak MI=1.9 in tissue.

B. Numerical model

The nonaxisymmetric version of the KZK equation suitable for focused diagnostic arrays can be numerically solved in dimensionless form by introducing five normalized variables [6]:

$$\frac{\partial P}{\partial \sigma} = \int_{-\infty}^{\tau} \left(\frac{1}{4G_x}\frac{\partial^2 P}{\partial \chi^2} + \frac{d_{xy}}{4G_y}\frac{\partial^2 P}{\partial \psi^2}\right) d\tau' + A\frac{\partial^2 P}{\partial \tau^2} + NP\frac{\partial P}{\partial \tau}. \quad (1)$$

where, $P = p/p_0$, $\sigma = z/d_x$, $\chi = x/a_x$, $\psi = y/a_y$, $\tau = \omega_0 t'$, p and p_0 is the acoustic and source pressure, respectively, z is the propagation axis of the sound beam, d_x and d_y are the azimuthal and elevation focus of the array, respectively, as it is shown in Fig.1, and a_x and a_y are the half-apertures of the array in azimuthal and elevation direction, respectively. ω_0 is the characteristic angular frequency, and t' is the retarded time such that $t' = t - {}^{z}/_{c_0}$, where c_0 is the speed of sound. d_{xy} is the ratio between focal length in azimuthal and elevation dimension ($\text{i.e.}, d_{xy} = d_x/d_y$). x and y are the azimuthal and elevation coordinates, respectively. The operating frequency is 2 MHz.

The first, second, and third term on the right-hand side of equation (1) account for diffraction, thermoviscous dissipation, and quadratic nonlinearity of the medium, respectively. Each term is described by a dimensionless parameter as follows:

$$G_x = z_0/d_x; \quad G_y = z_0/d_y; \quad (2)$$

$$A = \alpha_0 d_x. \quad (3)$$

$$N = d_x/\bar{z}. \quad (4)$$

G_x and G_y in equation (2) are the focusing gain in the azimuthal and elevation direction, respectively. z_0 is the Rayleigh distance, which $z_0 = \omega_0 a^2/2c_0$. A in equation (3) is the attenuation parameter, where α_0 is the thermoviscous attenuation coefficient, and $\alpha_0 = \delta{\omega_0}^2/2{c_0}^3$. N in equation (4) is the nonlinearity parameter, where $\bar{z}$ is the plane wave shock formation distance, and $\bar{z} = \rho_0 c_0^3/\beta\omega_0 p_0$, where β is the nonlinearity coefficient.

An appropriate boundary condition for the source at $\sigma = 0$ is:

$$P = f\left(\tau + G_x\chi^2 + G_y\psi^2\right)H(1-\chi)H(1-\psi). \quad (5)$$

The source function $f(\tau)$ used in the present study was:

$$f(\tau) = \exp[-(\tau/N_c\pi)^{2m}]sin[\tau + \phi], \quad (6)$$

where m and N_c indicate the shape of the envelope and number of cycle of the pulse, respectively. In the present study $N_c = 3.5$ cycles and $m = 2$ (hyper-Gaussian envelope). The various step sizes for the numerical solutions of different beamforming approaches are shown in Table 1.

Table 1: Step sizes used in the numerical code

Parameters for Numerical Solutions	Focused ultrasound	PWI	DWI
Focus	75 or 150 mm	10^4 m	-12.5 mm
Azimuthal steps (x) per aperture	100	100	150
Elevation steps (y) per aperture	100	100	100
IBFD step size ($\Delta\sigma$)	5×10^{-4}	1×10^{-9}	2×10^{-3}
CNFD step size ($\Delta\sigma$)	2×10^{-3}	4×10^{-9}	8×10^{-3}

C. Matrix array for PWI and DWI

For producing an unfocused "plane wave" beam with our existing numerical model we place the focus at infinity (10^4 m) as it is shown in Fig. 1. For producing a diverging beam suitable for DWI, we place a virtual focus behind the transducer surface, at -12.5 mm (Fig. 1).

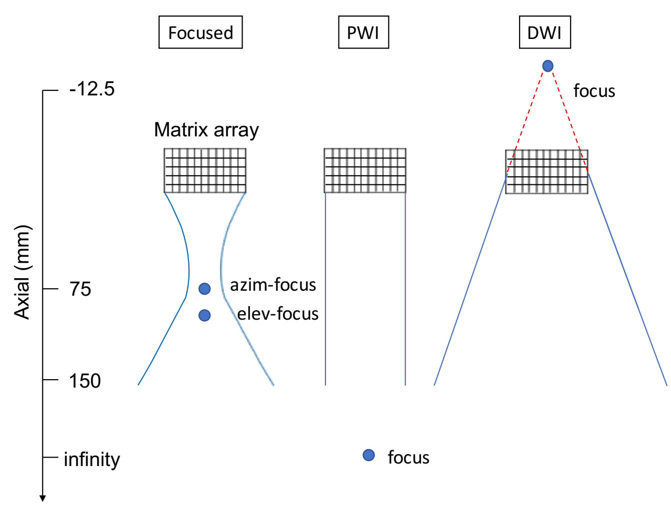

Fig. 1. Schematic of how conventional focusing, PWI, and DWI were modeled in the numerical solution; Conventional focusing has an azimuthal and an elevation focus, PWI has one focus at infinity, and DWI has a virtual (negative) focus behind the array.

D. *Tissue parameters for nonlinear simulations*

We used the following acoustic parameters for simulations in tissue: speed of sound $c_0 = 1540\ \mathrm{m/s}$, density $\rho_0 = 1000\ \mathrm{kg/m^3}$, nonlinearity coefficient $\beta = 5$, absorption coefficient $\alpha_0 =\ 0.3$ dB/cm at 1 MHz (a quadratic relationship with frequency is assumed in the calculations).

III. Results

A. *2nd harmonic by different beamforming approaches*

Beam plots of the 2nd harmonic component produced by the matrix array with 5 beamforming approaches in the azimuthal and elevation plane are shown in Fig. 2. The characteristics of the 2nd harmonic component produced by each beamforming approach is shown in Table 2. Beamforming approaches with deeper foci have lower focusing gain and therefore require higher source pressure than the conventional beamforming approach with both azimuthal and elevation foci placed at the center of the image depth (e.g., 2.3 MPa compared to 1 MPa) in order to reach equivalent peak 1.9 MI in tissue. Limited by the peak MI, the highest source pressure that PWI and DWI could use was 2.7 MPa at f=2 MHz.

As expected, AF=75/EF=75 produced the highest peak 2nd harmonic component compared to other beamforming approaches (Table 2). However, PWI produces a field that has wider 6 dB width that may improve the volume rate of 4D cardiac THI. For example, the 6 dB width at 75 mm of the 2nd harmonic produced by PWI was roughly 5 times wider than that produced by AF=75/EF=75. Using a wider beam for 4D cardiac THI would cover more area in a single transmit and thus improve the volume rate of 4D cardiac THI.

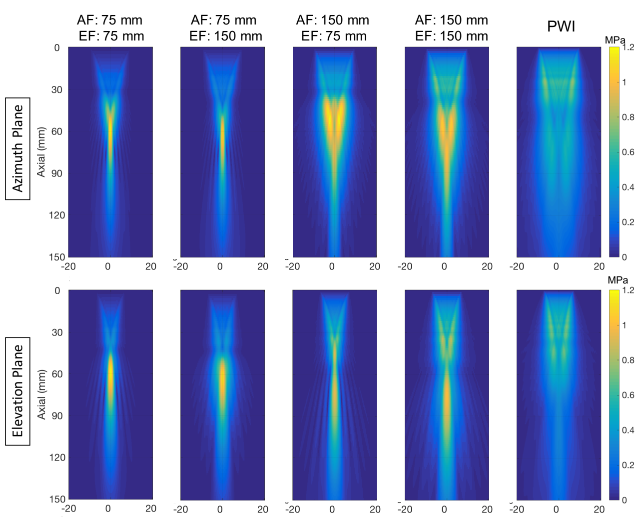

Fig. 2. Beam plots of the 2nd harmonic in azimuthal and elevation plane produced by the matrix array with different beamforming approaches. Pressure is in MPa.

Table 2: Source pressure, 6 dB width, peak 2nd harmonic of the field, and volume rate of different beamforming approaches.

Beamforming approaches (mm)	AF:75 EF:75	AF:75 EF:150	AF:150 EF:75	AF:150 EF:150	PWI	DWI
Source pressure (MPa)	1	1.2	1.8	2.3	2.7	2.7
Peak 2nd harmonic (MPa)	1.2	1.12	1.18	1.05	0.76	0.13
6dB range of axial field (mm)	44	48	86	107	87	33
6 dB width at 75mm-azim (mm)	1.9	2.1	5.6	7.1	13.1	
6dB width at 75mm-elev (mm)	2.8	3.6	3.6	4.2	6.72	
Transmit pattern (azim×elev)	61×51	51×41	21×25	12×21	7×9	2×2
Volume rate (Hz)	1.7	2.5	10	20	81	1275

B. *4D cardiac THI by different beamforming approaches*

The volume in front of the image probe is filled with beams such that the whole volume is sampled by placing the 6 dB down points of each individual beam touching each other and generating a uniformly yellow appearance as shown in Fig. 3. The contour plots are in dB normalized with respect to the maximum pressure in each beam. The transmit pattern (see table 2) indicates the number of individual beams required for each beamforming approach to sample the image volume. For example, for AF=75/EF=75, the image volume is sampled with 61x51 beams (table 2) and this would result in a volume rate of 1.7 Hz. For PWI, the image volume is sampled with 7x9 beams and the volume rate would be 81 Hz. Finally, for DWI the image volume is sampled with 2x2 beams and this results in a volume rate of 1275. However, as it is shown in Fig. 3, the penetration depth would be significantly reduced since diverging beams spread out the energy over a large area.

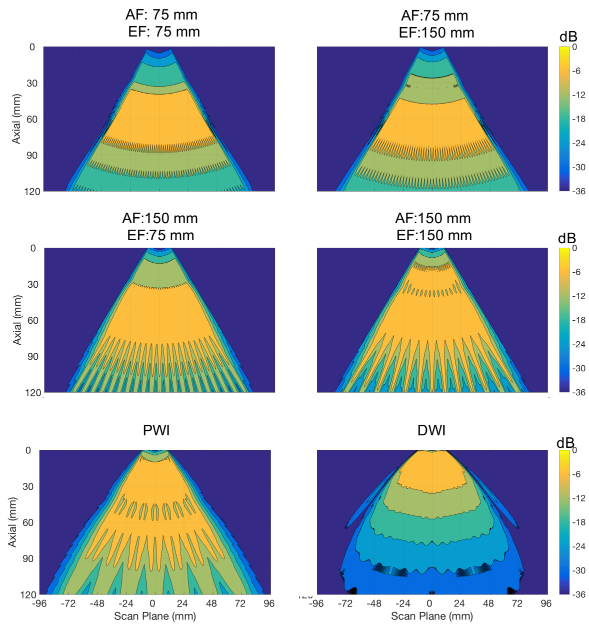

Fig. 3. Sampling of the azimuthal plane with individual beams for the various beamforming approaches considered. A single beam is steered in order to fill the volume in such a way that the 6 dB down points touch each other.

C. *Magnitude of the 2nd harmonic*

To better visualize the amount of produced second harmonic with each beamforming approach we plot it along the axis of each beam in Fig. 4. The y-axis is in a logarithmic scale and the maximum values are those shown in table 2. From Fig. 4 we see that the beamforming approach AF=75/EF=75 produced the highest peak 2^{nd} harmonic component in tissue which is ~1 dB higher than that produced by AF=150/EF=150. However, AF=150/EF=150 produced the highest 2^{nd} harmonic component at a depth of 150 mm. In addition, as it is shown in Table 2, the axial field produced by AF=150/ EF=150 has the largest 6 dB range compared to other beamforming approaches, indicating that the field is more uniform with deep foci. The peak 2^{nd} harmonic produced by PWI is 4 dB lower than that produced by AF=75/EF=75 but still 7.5 dB higher than that at a depth of 150 mm.

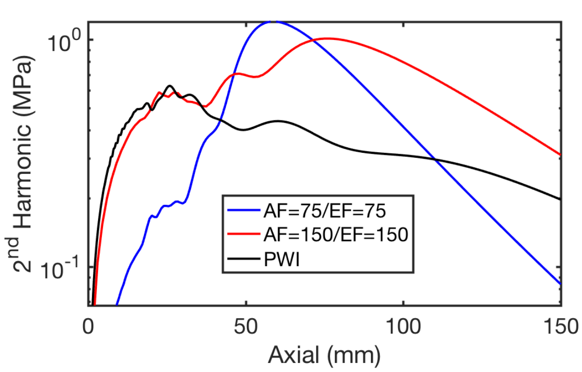

Fig. 4. Axial fields of the 2^{nd} harmonic in logarithmic scale at the center of the matrix array produced by 3 beamforming approaches.

IV. Conclusion

We have investigated the harmonic generation of a virtual matrix array for 4D cardiac THI with a numerical model based on the KZK equation. A beamforming approach with both azimuthal and elevational foci at 75 mm produced the highest peak 2^{nd} harmonic component in tissue and yet the narrowest beam that would limit the volume rate of 4D cardiac THI. A beamforming approach with azimuthal and elevational foci of 150 mm produced slightly lower peak 2^{nd} harmonic component (1 dB) but higher 2^{nd} harmonic component (11 dB) at a depth of 150 mm than the approach with shallow foci. The volume rates for the beamforming approaches we have considered for 4D cardiac THI are in the range 1.7-20 Hz with focused ultrasound (depending on the focal depths), 81 Hz with PWI, and 1275 Hz with DWI.

Acknowledgment

The research reported in this study was supported by the National Institute of Biomedical Imaging and Bioengineering of the National Institutes of Health under Award Number T32EB001650

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