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Simulating Ultrasound Fields for 2D Phased-Array Probes Design Optimization

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***Abstract*—Nowadays, ultrasound diagnostic imaging is one of the non-invasive techniques mostly used in the clinical practice. Recent advances in this field have brought to the development of small and portable systems. New bidimensional probes consisting of 2D phased arrays, allow to obtain real-time 3D representations of moving organs and blood vessels anatomy. Being the complexity of such 4D ultrasound imaging systems significantly increased, new challenges concerning electronics integration arise for designers. In this paper a software simulator is described, which has been developed in order to model ultrasound wave generation, pressure field distribution and echoes reception, with the aim to become a useful tool for optimizing the probe design. The paper mainly focuses on linear ultrasound field modeling; preliminary results on non-linear interactions with contrast agents are also here introduced.**

I. Introduction

MODERN realt-time volumetric (4D) imaging based on ultrasound (US) technology allows to achieve a moving 3D view of organs and structures inside the patients' body [1]. 4D ultrasound diagnostic devices take advantage of progress in integrated circuits technology, computation power and processing methods, which make it possible to design new portable systems employing 2D probes based on phased arrays of piezoelectric transducers and electronic scanning. As a matter of fact, it is the system electronics which is now in charge of building the delays profile to achieve correct focusing and of steering the beam to scan the whole volume of interest. Moreover, being real-time, 4D ultrasound imaging presents many benefits in terms of quality and interactivity if compared to other standard diagnostic imaging techniques such as MRI and CT, especially when temporal concerns are critical (i.e., in cardiac applications).

Among the imaging techniques employed by diagnostic ultrasound systems, harmonic imaging (HI) allows to achieve significantly better quality images in terms of spatial resolution, contrast-to-noise ratio, increased penetration and improved diagnostic reliability [2]. For these reasons, almost all state-of-the-art commercial US scanners implement HI. In this case, not only fundamental frequency signals are acquired and displayed by the system but also second harmonic echo ones, exploiting harmonics generation due to non-linear propagation of ultrasound in tissue or to the interaction with contrast agents [3] [4].

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In order to improve the designers understanding on how system performance is affected by choices made in integrated-circuits design [5], US field simulators can be employed. Many examples have been already described in the literature, such as the Field II simulator [6], Ultrasim [7], the DREAM Toolbox [8] and also commercial softwares, i.e. PZFlex (Weidlinger & Associates Inc., New York, NY, USA) or Wave3000 (CyberLogic Inc., New York, NY, USA). Simulating how the ultrasonic wave, generated by a phased-array of transducers, propagates can be undoubtedly a useful means to verify how front-end choices reflect on the beam properties. A simulator, in fact, may allow to test if the set delays give in turn the desired focusing or, for example, if employing a certain apodization function helps lowering possible energy dispersions (side lobes) and how the mechanical wave interacts with human body tissues. Moreover, foreseeing (at least, approximately) the echo signals the system would receive can help to properly tune the reception chain components (i.e. low noise amplifiers, time gain controlled amplifiers and ADCs) and to develop beamforming algorithms for image reconstruction. This is particularly true in the case of 4D imaging devices, in which the system complexity drastically increases [1]. A 2D probe in fact usually involves thousands of transduction elements (e.g., 4096 elements in a 64×64 matrix) and connections (coaxial cables) to the transmission/reception channels inside the system. This is why the most appropriate choice will be to integrate part of this electronics directly inside the probe, but of course this requires an accurate study of the problem and design of the probe.

The objective of the work here presented is to develop a simulator to be used by electronic designers in order to optimize the transmit and receive circuitry of a bidimensional ultrasound probe for medical applications. This tool will allow the user to set all the transducer-related parameters and to verify how they affect beam propagation and echoes reception. Aiming at this, an *ad-hoc* ultrasound field model will be implemented and made step by step more complex and accurate, as well as the one accounting for interactions between mechanical waves and tissue. Unlike other already existing simulators, interfacing the acoustic side with the electrical one is made possible by implementing an electric equivalent model of the piezoelectric transducers.

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Thus, a complete electro-mechanical description of the US beam formation, focusing, propagation, and scattered wave reception will be provided. This way, simulations will give the user all the required information for front-end electric signals validation, aiming in the end to a real ASIC design.

In this paper we integrate the results already introduced in [9] - which concerned only the transmission of the ultrasonic beam and pressure field generation in the 3D space - with scatterers inclusion inside the medium and echoes reception modeling. We also describe some preliminary results regarding non-linear US modeling and harmonic distortion in the case of a wave reflected by contrast agent microbubbles.

II. MATERIALS AND METHODS

A. The Linear Model for US Wave Propagation

The modeling of propagation, echoes generation and reception of the ultrasonic wave has been carried out in successive steps. First of all, a linear model, well known in the literature, has been used to simulate US transmission and reception in an homogeneous medium. This model is based on the *Spatial Impulse Response* (SIR), firstly introduced by [10] [11], then further described in the literature and also implemented in several simulation software tools, e.g. [6] and [8]. The SIR allows to describe the spatio-temporal response of a transducer to a delta function excitation. For a flat transducer mounted onto an infinite rigid baffle [12], the pressure field distribution in the 3D space can be derived by the Rayleigh integral:

$$P(\vec{r}_o,t)=\int_S \frac{\rho}{2\pi|\vec{r}|}\frac{\partial v_n(t-|\vec{r}|/c)}{\partial t}dS=\rho\frac{\partial v_n(t)}{\partial t}\underset{t}{*}h(\vec{r}_o,t), \quad (1)$$

where * denotes temporal convolution. Here, $|\vec{r}|=|\vec{r}_T-\vec{r}_o|$ represents the distance between the field observation point in r_o and the transducer in r_T (with respect to the same reference system); c is the speed of sound in the propagation medium (whose density is ρ); v_n is the vibration velocity normal to the transducer surface S and assumed to be uniform over it. $h(r_o,t)$ represents the SIR, which is related to the transducer geometry and to its distance from the point of observation:

$$h(\vec{r}_o,t)=\int_S \frac{\delta(t-|\vec{r}|/c)}{2\pi|\vec{r}|}dS. \quad (2)$$

Equations (1) and (2) state nothing else than the Huygens' principle, saying that every infinitesimal surface area dS of the transducer behaves like a point source from which a spherical wavefront originates. Thus, the pressure field value in a specific point can be obtained by summing up all the contributions from each source element. By simply extending the summation (integral) to all the elements making up the probe, we can model the behavior of an array of transducers.

Moreover, focusing delays can be included in the SIR model, so as possible apodization (i.e., weighting) coefficients can [9]. Also propagation inside an attenuating medium can be modeled, as described in [9] [13].

B. Pulse-Echo Field Representation

Linear systems theory can be used also when modeling the reception of the US wave echoes by the transducers. In this case, the received pressure field is computed as:

$$P_R(\vec{r}_T,t)=\frac{1}{c^2}\frac{\partial^2 v_n(t)}{\partial t^2}\underset{t}{*}h(\vec{r}_o,\vec{r}_T,t)\underset{t}{*}h(\vec{r}_T,\vec{r}_o,t), \quad (3)$$

where the convolution between the two SIRs (h_T in transmission and h_R in reception) gives the so called "pulse-echo SIR". The received voltage signal, can be then easily obtained by convolving (3) with the transducer electro-mechanical impulse response.

C. Non-Linear Response Due to Contrast Microbubbles

In order to study the generation of non-linearities due to the presence of inhomogeneities inside the propagation medium, contrast microbubbles have been considered. Recently in fact, new contrast agents have been developed [14] [15] which are suspensions of gas-filled microbubbles, generally behaving as non-linear volume oscillators when hit by an incident US pulse. In this case, the scattered pressure is due mainly to bubble volume (and radius, $R(t)$) variations and can be expressed as:

$$P_s(r,t)=\rho\frac{R(t)}{r}\left(2\dot{R}(t)^2+R(t)\ddot{R}(t)\right), \quad (4)$$

where r is the radial distance from the bubble centre and ρ is the surrounding medium density [16]. To describe a single spherical bubble radial oscillations, the Rayleigh-Plesset equation [16] [17] (or one of its many variants) is usually employed:

$$\ddot{R}R+\frac{3}{2}\dot{R}^2=\frac{1}{\rho}\left[p_L-P_o-P_i+\frac{R}{c}\dot{p}_L-4\nu\frac{\dot{R}}{R}-\frac{2\gamma}{\rho R}\right], \quad (5)$$

where P_o is the ambient pressure (e.g. 1 atm), P_i is the driving pressure, p_L is the pressure of the gas inside the bubble [15], ρ and υ are the medium density and viscosity, γ is surface tension and c is the speed of sound.

D. Simulator Design

The ultrasound field simulator has been developed using Matlab (The MathWorks, Natick, MA, USA). At the moment, it allows to simulate the behavior of a 2D probe (both during beam transmission and echoes reception) assuming the US wave to propagate in an homogeneous medium according to the SIR model equations, implemented following the discrete representation concept [18].

Each transducer is modeled as a flat rectangular element, being the centre of the whole phased array (lying on the xy plane) placed in the origin of the reference system. The SIR computation is then performed by subdividing each element of the matrix in small areas considered as single-point sources, whose contributions are summed up to build the whole array response. A more detailed description of all the simulation parameters can be found in [9].

A graphical user interface has been also designed, which allows to visualize the acoustic pressure field in the 3D space

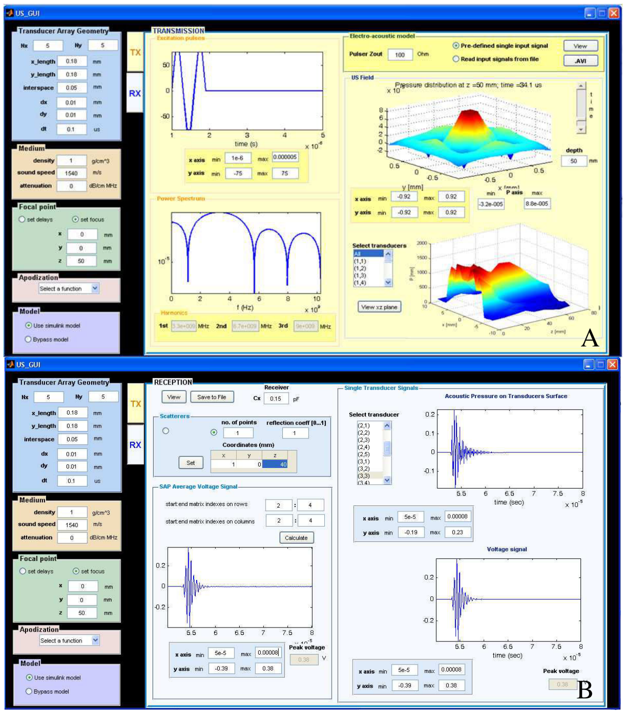

Fig. 1. Simulator GUI panels: A) transmission of the US wave; B) signals involved in echoes reception. The panels on the left allow the user to set those parameters related to the transducers geometry and to the propagation medium, to choose an apodization function and to set the focusing point.

during the transmission phase (Fig. 1A) and also the reflected wave (Fig. 1B). Besides, an equivalent electric model of each piezoelectric transducer element inside the probe has been designed using Simulink. This way, a complete electro-mechanical simulation of the transducers behavior can be performed and also the involved electric signals can be viewed and analyzed. Anyway, the user can choose whether to set directly the acoustic input signals or the electrical ones (running the Simulink model) before starting the simulations. Similarly, it is possible to view the acoustic and/or electric received signals.

The user can tune not only the electric model parameters and those related to the transducers geometry and to the medium but, more importantly, he/she can set the voltage excitation pulses applied to the piezoelectric elements, with proper focusing delays. The GUI then shows the spectral content of these signals and a 3D view of the pressure field distribution on the *xy* (azimuth & elevation) plane at the desired depth. Also the pressure peaks evolution along the *z* axis can be visualized (Fig. 1A). For what concerns reception instead, the GUI allows to set the number and position of desired scattering points, to see the electric voltage signals generated when the echo reaches the transducers and also to calculate the average of the signals received by a selected aperture (Fig. 1B).

III. Results

A software tool has been developed, which allows to simulate the behavior of 2D phased arrays of piezoelectric transducers and implements a SIR-based model for US wave propagation. First results concerning the transmission of the ultrasonic beam were presented in a previous paper [9], showing how our first prototype was able to simulate the generation of a focused, possibly apodized beam and its propagation also in the case of lossy medium. This work extends the US system modeling including also the echoes reception phase.

Results here presented have been obtained simulating propagation in an homogeneous medium (water: c=1540 m/s, ρ=1000 kg/m^3) in which two scattering points have been placed at (x,y,z) = (1,0,40) mm and (-1,0,40) mm. The phased array has been modeled as a 5×5 matrix of square transducers (l=0.18 mm, *kerf*=0.05 mm). The excitation velocity signal $v_n(t)$, uniform over the whole 2D array surface, consisted of an 8 periods sinusoid, oscillating at a frequency of 3 MHz and weighted using the Hanning window. The beam has been focused at (x,y,z)=(0,0,40) mm; user-defined focusing delays have been set, so that the maximum delay is applied to the central array element and then they decrease while moving towards the matrix corners.

Fig. 2 shows how the acoustic wave propagates till reaching the scatterers which reflect it back to the transducers surface plane. The represented z axis (depth) has been discretized with a step of 5 mm.

Besides developing this first simulator prototype, we were also interested in modeling the simplified case in which wave non-linearities generate, (at the moment) only due to the presence of a single-point inhomogeneity inside the propagation medium. For this reason, a real model used in the case of contrast harmonic imaging, but with some simplifications, has been chosen. We have considered only a single microbubble, (till now) without any encapsulating shell. The Rayleigh-Plesset model (eq. 5), modified by including radiation damping [14] [17], has been implemented to describe the interaction of the incident US wave with the bubble and the scattered pressure (eq. 4). Preliminary obtained results are here presented, which

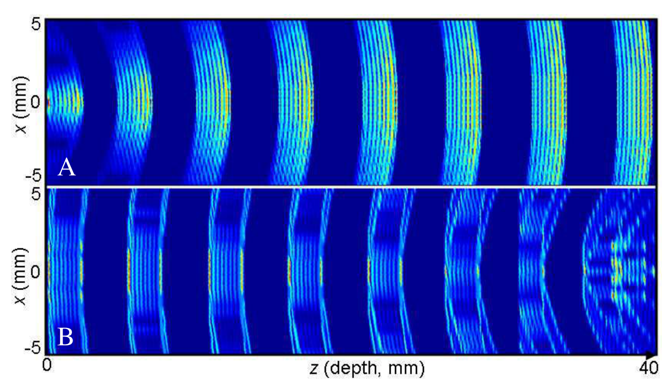

Fig. 2. Simulated US wave propagation in water: top wiew (xz plane). Figure 2B is temporally subsequent to figure 2A. (A) The acoustic pressure wave generates from the surface of the transducer (aligned along the x axis) and then propagates (from left to right) until reaching two scattering points, placed at a depth of 40 mm. (B) The wave is then partially reflected back (from right to left).

demonstrate that actually the volume oscillations of a bubble, hit by an incident US wave, cause the generation of higher harmonics in the reflected wave spectrum. The medium considered is again water; undriven bubble radius has been set e.g. to R_o=1.5 µm; the incident US wave has been modeled as a 3 cycles Hanning-window-modulated sinusoid (f_o=2 MHz, peak amplitude=0.3 MPa). The Runge-Kutta method implemented by the *ode45* function in Matlab has been employed for differential equation (5) solving. In Fig. 3, the spectra of both the incident and scattered wave have been plotted: as expected, harmonics at multiples of the fundamental frequency f_o, generated by the interaction with the bubble, are clearly visible in the latter.

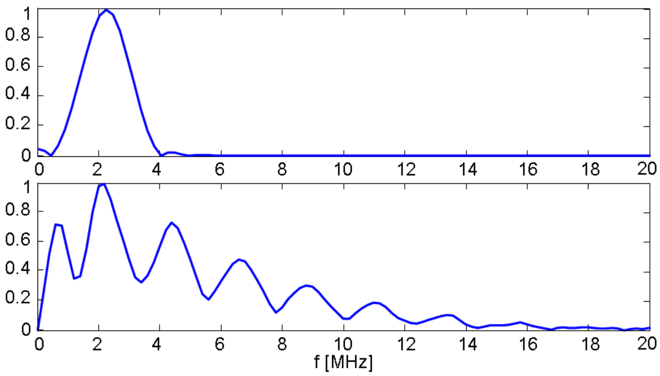

Fig. 3. Normalized spectra of the incident (top) and scattered (bottom) acoustic pressure waves. The latter clearly shows the presence of harmonics at frequencies which are multiples of the fundamental one (2 MHz).

IV. Discussion and Conclusions

The first objective of this work was to investigate the behavior of 2D phased arrays of piezoelectric transducers, by simulating US field generation and wave propagation in an homogeneous medium. For this reason, a software has been developed which implements a linear model of the US field and allows the user to set all the desired parameters related to the transducers geometry, to the medium and to the excitation signals used to generate the beam (i.e. focusing delays). We have demonstrated that this tool can be used to view the distribution of the generated pressure field in the 3D space and time and also to model echoes reception, after having included scatterers inside the propagation medium. Particularly, the spatio-temporal acoustic model can be integrated with the transducers electric one. By this way, it is possible to simulate the beam generation directly starting from the voltage excitation signals to be applied to the piezoelectric elements in the matrix.

Moreover, we have also begun to address those concerns regarding second harmonic imaging implementation. Modeling of non-linearities arising inside the US wave has been investigated, first of all starting form harmonics generation due to contrast agent microbubbles oscillations.

Further research is foreseen in this direction, in order to achieve a complete non-linear description of the ultrasonic wave propagation, which could allow to perform more realistic simulations. Finally, all the implemented models will be used to run complete tests of the designed electronics, in which the role of the transducers will be held by the simulator. In particular, we will first use it to verify if the ASIC implementation of the focusing delays generator actually allows to focus the beam as desired.

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