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Second Harmonic Field Generation from a Phased Array Transducer and Its Beam Optimization

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Abstract— **In this paper, we have done the simulation of ultrasonic field propagation in human tissue medium from a phased array transducer. The ultrasound plane wave propagation has been modeled using the B/A model and has been solved using the time domain based pseudospectral method. The generation of second harmonic field has been studied along the on-axis and off-axis as well and has been compared with the fundamental. The principle of beam formation has been studied thoroughly and utilized for second harmonic field optimization. The second harmonic beam optimization has been performed by observing the variations of system parameters i.e. applied pressure, frequency, f-number and apodization techniques.**

Index Terms— **ultrasound imaging, second harmonic, pseudospectral, human soft tissues**

I. Introduction

Till recently the ultrasound propagation in biological tissues was assumed linear whereas the propagation practically found nonlinear [1]. It has been validated by many researchers that ultrasound waves undergo gradual distortion in almost every medical application. Thus tissue harmonic imaging (THI) is being used successfully as it provides maximum diagnostic information to the physicians and helps them to better diagnose the diseases [2, 3]. The THI is the formation of ultrasound image from the backscattered signal at twice the frequency of the transmitted signal. The second harmonic imaging is being preferred over fundamental imaging as it eliminates the wavefront aberration and attenuation on the forward path, narrows the beam and significantly suppresses the sidelobes [4]. This results in better lateral and axial resolution and improved signal-to-noise ratio [5]. For further development in ultrasound imaging, the researchers needs the better understanding of nonlinear acoustics and better computational tools capable of simulating the complex imaging system. This could help in designing the efficient instrumentation for second harmonic imaging. Modeling and simulation of ultrasound wave propagation in real tissue model described by full wave equation for heterogeneous, lossy and nonlinear medium serves the purpose. KZK (Khokhlov-Zablotskaya-Kuznetsov) [6] and related methods found restricted in forward and off-axis propagation due to the underlying paraxial basis. The pseudospectral method [7] provides high spatial accuracy and as it solves the nonlinear wave equation at every point in the 2D spatial domain, it can also be used to model propagation through heterogeneous model. At longer propagation distances, KZK solutions based on finite difference or finite element methods distorts signals unacceptably as they use lower order space and time derivative approximations. In this paper, we have used the same pseudospectral approach to model and simulate the ultrasound nonlinear wave propagation in an attenuating tissue medium.

In this paper, the pseudospectral method has been used to model the propagation of ultrasound in human tissues. The lateral and axial beam profiles have been observed at focal depth and also the frequency analysis has been performed. The principle of beam forming mainly composed of parameters related with excitation, probe and propagating medium. The control of these parameters gives the ability to improve the quality of the ultrasound images which ultimately will give the useful information to physicians for diagnosis. Few relevant parameters have been varied to observe their effects on lateral and axial resolution, penetration depth and their optimized selection has been suggested.

II. Pseudospectral method

The nonlinear acoustic wave equation's generalized form derived from the constitutive equation, popularly known as B/A model is used here which includes the effects of absorption and nonlinearity. The B/A model is given by the following equation[7, 8]:

$$\rho \frac{\delta^2 u}{\delta t^2} = \nabla p,\ p = -K\left(\nabla \cdot u + \frac{1}{2}\frac{B}{A}\left(\nabla \cdot u\right)^2\right) \qquad (1)$$

where p is the acoustic pressure, ρ is the density of the medium, $K = \rho c^2$ is the bulk modulus, c is the acoustic speed, B/A is the nonlinearity parameter of the medium and vector u represents the particle displacement velocity. In pseudospectral method, the time domain solution of the nonlinear acoustic wave equation achieved by using step-by-step integration in time and evaluation of special derivatives at each step on uniform mesh. For evaluating the spatial of function defined over uniform Cartesian grid, the discrete Fourier transforms has been used. After this the time dependent PDEs has been reduced to simple ODEs, which has been further integrated in time explicit linear multistep method. For integration in time along

Narendra D. Londhe, Research Scholar, EED, IIT Roorkee, Roorkee-247667, India. Phone No. +91-9411754514 propagation axis, staggered 4th order Adams Bashforth method has been used. Adams Bashforth is an explicit linear multistep method which gives improved accuracy and stability as compared to Runge-Kutta method and also gives reduced computation time. Secondly representing attenuation in the medium has been done using various relaxation models [9]. A perfectly matched boundary layer (PML) has been applied at the edges of the calculation domain to greatly reduce the effects of the domain periodicity i.e. wrap-around which inherently presents in the FFT base calculations [10].

III. NUMERICAL IMPLEMENTATION

The parameters used in performing the simulation of ultrasound plane wave propagation in soft human tissues are given below with their definitions and values. The excitation signal used is the 9 cycle Gaussian pulse with 60% bandwidth. The 16 element phases array transducer and the field generated is modeled using the methodology mentioned in the above section.

1) Parameters defining the excitation and medium

Central transducer frequency, f_0 = 1.7 MHz
Initial acoustic pressure, p = 1 MPa
Acoustic speed, c = 1540 m/s
Relative bandwidth = 0.6
Density of the medium, ρ = 1000 kg/m^3
Nonlinearity parameter, B/A = 7
Attenuation coefficient, α = 0.5 dB/cm-MHz

2) Parameters defining the phased array transducer

Number of elements = 16
Pitch of the array = c/f_0
Focal depth = 46.7 mm
Apodization technique = rectangular

The focused ultrasound field nonlinearly propagating through the biological tissue has been calculated by 16-element phased array transducer. The frequency spectrum of the field at focal depth is shown in Fig.1(a) and axial and lateral beam profiles are shown in Fig.1(b) and Fig.1(c) respectively. The frequency spectrum of the pressure field and beam pattern are observed at focal depth. The second harmonic component generated is very less in magnitude i.e. around 20 % of the fundamental. As shown in Fig.1(b) the sidelobe level in fundamental is 18 dB while in case of second harmonic, it is found 35 dB. The axial and lateral beam profiles in Fig. 1(b) & (c) for second harmonic are scaled down with respective fundamental beam profiles. In Fig.1(c), the amplitude of second harmonic rises because of the effect of nonlinearity and later starts reducing because of the attenuation effect.

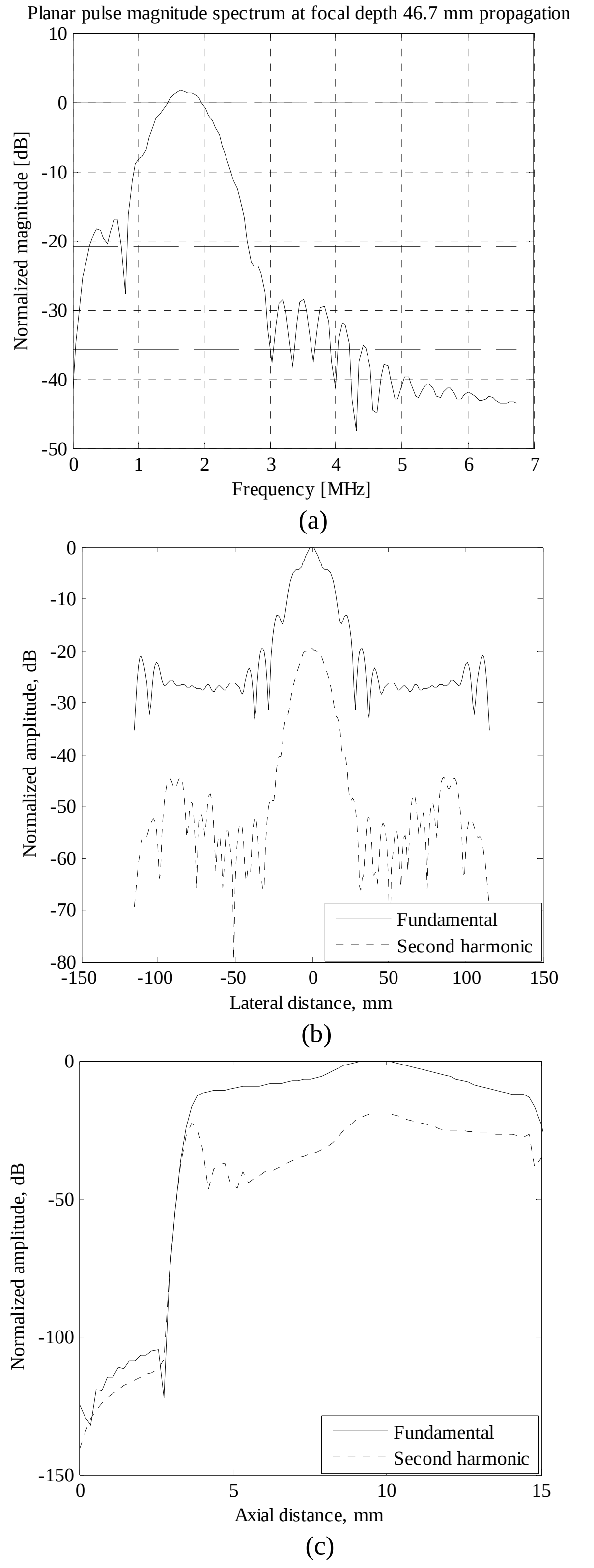

Figure 1. Simulated results of planar acoustic wave propagation in soft tissue medium with above mentioned excitation and medium properties (a) frequency spectrum of wave at focal point, (b) beam patterns of fundamental and second harmonic observed at focal point, (c) axial beam profile of both fundamental and second harmonic.

IV. Parameters variation

The transducer is the most expensive part of the ultrasonic imaging system. The design of transducer and rest parts of imaging system too depends on the optimized parameters selection so that the image quality can be improved. In this section, we have tried to observe the change of parameters on axial and lateral resolutions and penetration depth. The trade off between the parameters and their effects on resolution and penetration depth has also been analyzed. The parameters like center frequency of the transducer, applied pressure amplitude, apodization techniques, f-number are varied and their effects have been observed.

A. Change in transducer frequency and pressure amplitude

The center transducer frequency plays an important role in improving the lateral and axial resolution in ultrasound imaging as it directly proportional to the resolution while it is inversely proportional to depth of penetration. So the tradeoff between resolution and penetration depth with frequency and pressure amplitude has been observed here.

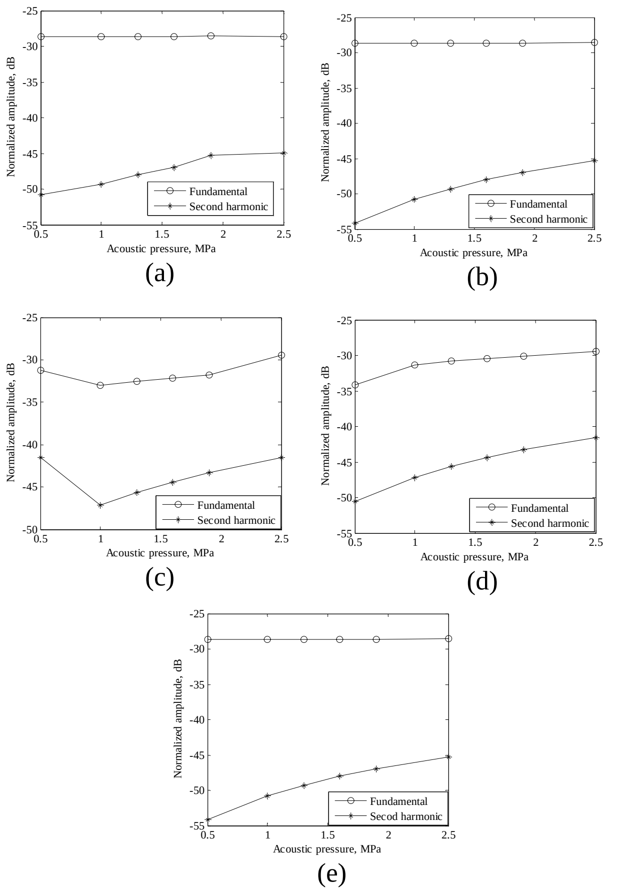

Figure 2. Normalized pressure value variation of fundamental and second harmonic component at various center transducer frequencies (a) f0 = 1MHz (b) f0 = 1.5 MHz (c) f0 = 2 MHz (d) f0 = 2.5MHz (e) f0 = 3 MHz

At fixed center transducer frequency, the strength of second harmonic component increases with increase in applied pressure while the fundamental frequency component remains almost constant as shown n Fig. 2 (a) – (e). As the frequency increases, the resolution will improve in second harmonic imaging and at the same time with improvement in applied pressure assured the strength of second harmonic component which ensures the higher penetration depth. So the trade-off between resolution and penetration depth can be controlled by using optimized combination of higher transducer operating frequency and pressure. This study can help in high frequency applications like intravascular ultrasound (IVUS).

B. Variation of apodization techniques

Apodization is method aperture weighting of normal velocity for reducing side lobes on either side of the mainlobe in array transducers. It gradually decreases the vibration of the transducer surface with distance from its center. In arrays, the apodization can be achieved by exciting elements with different voltage amplitudes and selection of that amplitudes generally specified by a function like triangular, rectangular etc. The proper selection of apodization function leads to better sidelobe suppression.

In this section, four apodiztion window techniques named triangular, rectangular, Gaussian and Kaiser are observed. Fig. 3 shows the frequency spectrum of pressure field generated at focal depth from the transducer with four apodization techniques. The amplitudes of fundamental and second harmonic remains almost same.

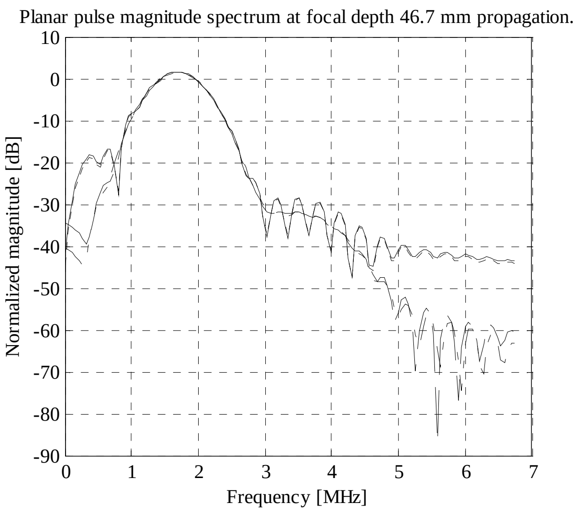

Figure 3. Frequency spectra of pressure field at focal depth using various apodization techniques, ___(solid) rectangular, ---(dashed) triangular, -.-(dot-dashed) Gaussian,….(dotted) Kaiser.

The beam pattern is the real measure to understand the effects of apodization techniques; Fig.4 shows the beam patterns calculated with four different apodization techniques. There is a trade-off between sidelobe suppression and beamwidth. As shown in Fig. 4, with rectangular apodization, the sidelobe level is improved but the beamwidth is increased compared to fundamental. The same happened with the use of Kaiser apodization. While with triangular apodization, the beamwidth of second harmonic is narrow compared to fundamental, sidelobe level is improved but the number of sidelobes is more. So the apodization techniques suppresses the sidelobes but at the same time the beamwidth increases which may further degrades the lateral resolution. So the apodization should be selected in such a manner that the sidelobe suppression as well as narrow beamwidth can be achieved. From Fig.4, it has been observed that the Gaussian apodization serves both the purposes in the best way.

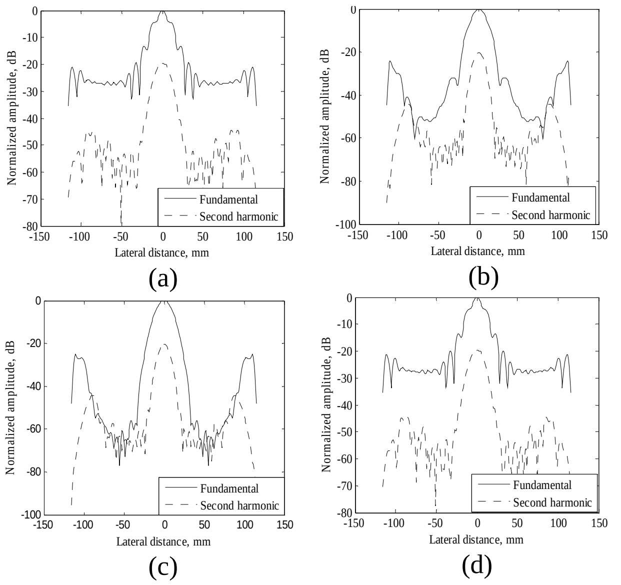

Figure 4. Beam patterns calculated at focal depth with different apodization techniques (a) rectangular, (b) triangular, (c) Gaussian, (d) Kaiser

C. Variation of f-number

F-number is defined as the ratio of focal length to transmit aperture length. The variation of f-number means the change in focal depth. The lateral resolution is the product of wavelength and f-number, so larger the aperture and shorter wavelength (high frequency) gives better lateral resolution. Even the change in focal depth with constant transmit aperture length can also give the

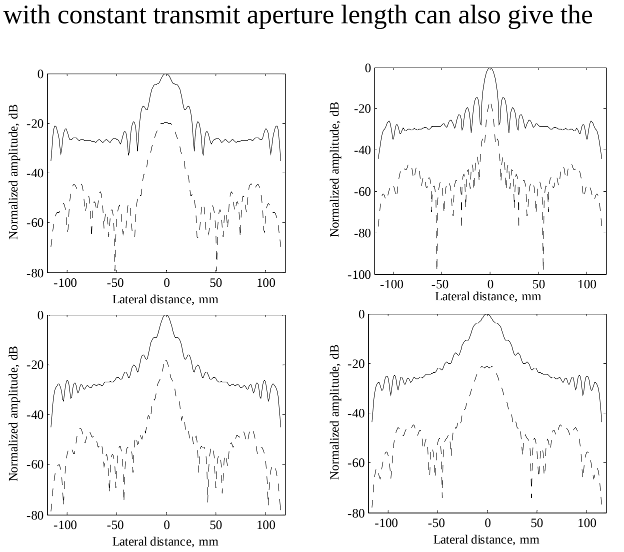

Figure 5. Beam patterns calculated at focal depth with f-number (a) f-number =1 (b) f-number =1.3 (c) f-number =1.6 (d) f-number =1.9

improved lateral resolution at high frequency. Scan depth can be divided into smaller regions and taking images at multiple transmit foci is done in dynamic focusing which shows the change in f-number only. The images taken at various focal depths combined together gives improved image resolution. But the focus has limited strength, so at higher value of focus distance as shown in Fig. 5(c) beam started to get broaden and that will ultimately reduce the resolution. In Fig. 5(b), the value f-number is 1.3, has shown the possible optimized value of f-number as beam is narrower amongst all beam patterns at other values of f-number.

V. CONCLUSIONS

Image quality is the most concerned issue for researchers, as the improved image quality can give more information to the physicians which may help them understanding of the behavior of the human body parts. The most influencing factor to image quality is the nonlinear nature of the propagation of ultrasound in human tissues. The harmonics generated plays an important role in improving the resolution of the images, at the same time there is a difficulty in understanding their behavior. Modeling is the best way to analyze the behavior of wave propagation and it helps in designing the instrumentation. In this paper, we have tried the same by analyzing the variation of parameter variations on the image enhancement and the trade-offs between them.

The parameters which influence image quality are instrumentation parameters and medium parameters. We have focused on instrumentation parameters like transducer frequency, applied pressure, apodization techniques and f-number (which depend on focal depth, number of elements and pitch of the array). Proper selection of these parameters and better understanding of the trade-offs of their effects on beam parameters leads to optimization of beam profiles. This leads to improved lateral and axial resolution and better penetration depth. The use of proper combination of transducer frequency and applied pressure improves both resolution and penetration depth. And the selection of apodization technique affects the sidelobe suppression and beamwidth which decides the lateral resolution. The variation of f-number using of multiple foci may lead to reduced frame rate and change of number of active elements and pitch is physically not possible. The synthetic aperture imaging can be used in which selected elements is used to be excited which can provide variation in aperture size and pitch also. The dependence of these factors upon each other is also need to be studied and mathematical optimization techniques can be used to optimize every parameter performance and complete system performance as well.

VI. References

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