Ultrasound Array Modelling and Beamforming using Field II

[1]Jayaraj U Kidav, [2]B. A. Sujathakumari, [3]Laseena C.A.[*]
[1] Department of VLSI Design, NIELIT, Calicut, Calicut University, Calicut, Kerala, India
[2, 3] Department of E&CE, SJCE Mysoru, Visvesvaraya Technological University, Belgaum, Karnataka, India

Abstract—

U ltrasound medical imaging has become one of the most regularly and usual procedure for the diagnosis of diseases and as guidance in surgery. Its popularity arises as it provides high resolution images without ionizing radiation. The image acquisition method used here is digital beamforming and is done by manipulating the array sensor data to extract a signal of interest from noise when looking in a specific direction. In this paper, modelling of ultrasound array is done using Field II procedures. The study on delay sum beamformer is also done using the delay calculations performed in MATLAB. Finally a receive beamformer is designed and the image clarity, on varying the axial and lateral resolutions is observed. Initially the design of the transducer array and pulse-echo signal reception is done using Field II simulation program toolbox of MATLAB. Delay calculations for linear array transducers and the fractional delays are studied. A delay sum beamformer is designed according to the required specifications. By studying the coarse delay and fine delay calculations, an algorithm for Rx beamformer is implemented in MATLAB® and B-mode images for corresponding phantoms are obtained.

Keywords— Beamforming, Ultrasound, FIELD II, B-mode image, phantoms.

I. INTRODUCTION

Field II is a program for simulating ultrasound transducer fields and ultrasound imaging using linear acoustics. The programs use the Tupholme-Stepanishen method for calculating pulsed ultrasound fields. The Field program system uses the concept of spatial impulse responses. The program is run under MATLAB on a number of different operating system (Windows, Linux, Mac OS X).In medical ultrasound scanners, the ultrasound echoes received by the transducers determine the information about the properties of underlying tissues. The collected echoes are scaled and appropriately delayed to permit a coherent summation of the signals known as beamforming at the receiver.

A multi-element transducer is used for both transmitting and receiving the pulsed ultrasound field. The central frequency of the transducer is set as 3.5MHz, and speed of sound in tissue is taken as 1540m/s. Beam emission is controlled electronically. Here a single transmit focus at the depth of 60mm is taken. The reflected and scattered field is received by the transducer and is then amplified for compensating the loss in amplitude due to attenuation during propagation. After amplification, all the pulse echo signals are moved for receive beamformation.The beamformed signal is envelope detected, log compressed and scan converted to view as a gray-level display in real time.

The linear array transducers select the region of investigation by firing a set of elements situated over the region. The beam is moved over the imaging region by firing sets of contiguous elements. Transmit focusing is achieved by delaying the excitation of the individual signals. The receive focusing of the beam is done at the reception by delaying and adding responses from different elements. Focusing is done in order to align the pressure fields from all parts of the aperture to arrive at the field point at the same time. This is done by applying delays to individual signals, determined by the propagation time. The propagation time t_i from the center of the aperture element to the field point is

$$t_i = \frac{1}{c}\sqrt{(x_i - x_f)^2 + (y_i - y_f)^2 + (z_i - z_f)^2}$$

(1)

where (x_f, y_f, z_f) is the position of the focal point, (x_i, y_i, z_i) is the center for the physical element number i, and c is the speed of sound. A point is selected on the whole aperture as a reference for the imaging process. The propagation time for this is

$$t_c = \frac{1}{c}\sqrt{(x_c - x_f)^2 + (y_c - y_f)^2 + (z_c - z_f)^2}$$

(2)

where (x_c, y_c, z_c) is the reference center point on the aperture. The pressure field generated at the focus point depends on density of the medium, speed of sound and wave number. The emitted pressure is calculated by integrating the field over all elements of the aperture. A perception of the sound field for a fixed time instance can be obtained by employing Huygens' Principle in which every point on the radiating surface is the origin of an outgoing spherical wave. The spatial impulse response is found by observing the pressure waves at a fixed position in space over time by having all the spherical waves pass the point of observation and summing them. The scattered field and received signal by the transducer is measured using the spatial impulse response [1]. The received signal from the transducer is calculated as the spatial and temporal convolution of pulse-echo impulse, in-homogeneities in the tissue and pulse-echo spatial impulse response. Where, pulse- echo impulse is the convolution of transducer excitation, and the electro-mechanical impulse response during emission and reception of the pulse.

A beamformer is a spatial filter that operates on the output of an array of sensors in order to enhance the amplitude of a coherent wavefront relative to background noise and directional interference. A beamformer for a radio transmitter applies the complex weight to the transmit signal (shifts the phase and sets the amplitude) for each element of the antenna array. A beamformer for radio reception applies the complex weight to the signal from each antenna element, then sums all of the signals into one that has the desired directional pattern[5].This paper describes the design of transducer array using Field II and implementation of delay sum receive beamformer.The study and delay calculations are also described here. Finally, obtained algorithm is implemented in MATLAB and B-mode images for cyst phantom is obtained.

II. DESIGN

Implementation of delay sum beamformer is done by designing efficient algorithms for the processes as shown in Fig.1. The RF signal from the individual transducer array elements are digitised and taken as the input to this system. Array modelling and beamformer implementation is done by doing the following studies.
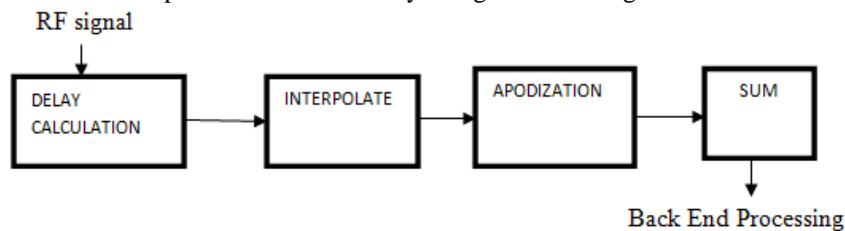
Fig.1 Block diagram of delay sum beamformer

A. Array design

An ultrasound array consists of number of transducer elements arranged in different methods to form linear, convex, annular, and phased arrays. Here a linear phased array transducer is designed with number of elements as 8, 64 and 128. The element width, kerf, height, focus and element subdivisions are designed as per the procedures of Field II and as shown in TABLE 1. Speed of sound in tissue is taken as 1540m/s and sampling frequency as 100MHz.

TABLE 1 ARRAY SPECIFICATIONS\

Array type	Linear focused array
Number of physical elements	64
Element height	13mm
Element width	0.265 mm
Element kerf	0.025 mm
Element pitch	0.290 mm
Elevation lens focus	60 mm
Emitter focus	[20 0 20] mm
Element sub division in x-direction	1
Element subdivision in y-direction	15
Transducer center frequency	3.5 MHz

B. Delay Calculation

An ultrasound linear phased array transducer contain over hundred transducer elements that may be multiplied and/or electronically steered and focused via phased array technique[2].Phase steering is accomplished by sequentially pulsing the array elements by calculation the inter – element delay using the equation and in detail is shown in Fig.2.

$$\Delta\tau = \frac{d \sin\theta s}{c} \qquad (3)$$

Where $\Delta\tau$ is the time delay between adjacent element, d the distance between element, the θs required steering angle, and c is the wave speed(1540m/s) in the medium. The ultrasound linear phased array transducer model stated in this paper is able to simulate the pressure field for steering and focusing within the transition zone(near field) , and beyond (far field). The focusing delay is calculated using the generalised focusing formula (4) for any number of elements N as stated in equation given below.

$$t_n = \frac{F}{c}\left\{\left[1 + \left(\frac{\overline{N}d}{F}\right)^2 + \frac{2\overline{N}d}{F}\sin\theta_s\right]^{1/2} - \left[1 + \left(\frac{(n-\overline{N})d}{F}\right)^2 - \frac{2(n-\overline{N})d}{F}\sin\theta_s\right]^{1/2}\right\} \qquad (4)$$

where t_n is the required time delay for element *n*=0,...N-1,N =(N-1)/2,*d* the center – to –center spacing between elements, *F* the focal length from the center of the array, θs the steering angle from center of array ,and *c* is the wavespeed.

A fractional delay filter is used for the band-limited interpolation between samples. A discrete time signal *x(n)* is delayed to yield an output *y(n)* as*y(n)=L{x(n)}=x(n-D)*,where L is a linear operator and D is a positive real number that can be split into the integer and fractional part as *D= Int(D) +d* [3]. In ideal case, when the desired delay D assumes an

integer value, the impulse response of the signal reduces to a single impulse at n=D, but for non integer values of D the impulse response is an infinitely long, shifted and sampled version of the original signal. Thus an approximating (real-coefficient) filter must be implemented to solve this problem. There are several methods proposed in array signal processing for the implementation of fractional delay filter. Here, for the design of Rx beamformer, a Farrow structure fractional delay FIR filter with MMSE interpolator is used.

C. Dynamic focusing

Each frame is divided into number of scan lines. The lateral resolution determines the number of scan lines. As per the axial resolution, the maximum depth of each scan is divided into number of focal depths as shown in Fig.3.Time to travel these focal depths is taken as the focal times. The apodization is valid only at these focal times. The focus of receive aperture is thus varied in each focal time. The delayed signals are summed to obtain a beamsum data for single scan line. Similarly the row data is obtained for the total number of scan lines.
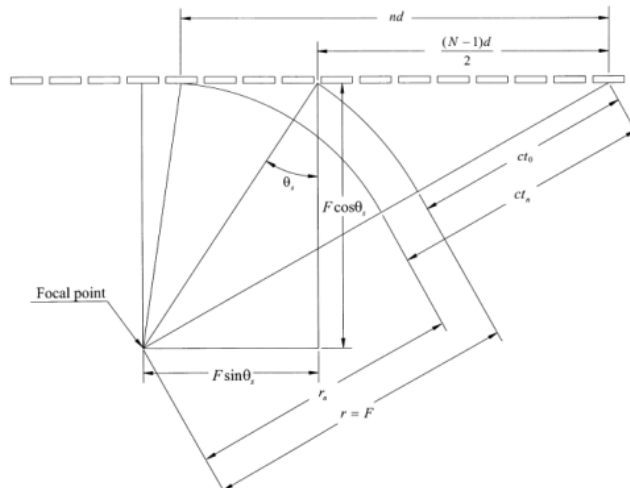

Fig.2. Geometry of liner phased array used in deriving the focusing formula for even-numbered arrays
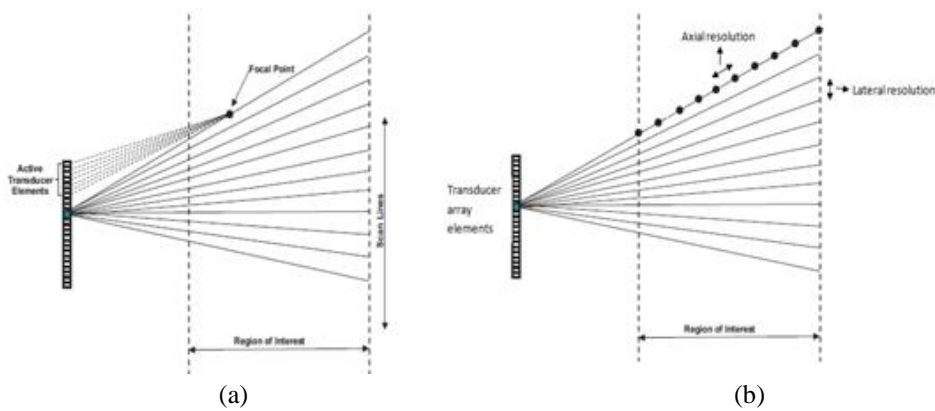

| (a) | (b) |

Fig.3 (a) Dividing scan area into number of scan lines. (b) Each scan line is divided into number of focal points.

D. Interpolation

Fractional delay interpolation is used to generate fine delay. Wiener –Hopf Filter is used as the FIR filter for implementing fractional delay. [4] $w_{opt}=R_x^{-1}r_{xd}$.where R_x is the input autocorrelation matrix and r_{xd} is the cross correlation vector. In this paper FIR filter with length 2 is considered. Thus

$$R_x = \begin{bmatrix} r(0) & r(1) \\ r(1) & r(0) \end{bmatrix}$$

where r(t) represents the autocorrelation function of X(n). If f_2 and f_1 represents upper and lower cut-off frequencies of the bandpass signal X(n), the R_x is calculated as:

$$R_x = \begin{bmatrix} 1 & \cos(w1)\sin(w)/w \\ \cos(w1)\sin(w)/w & 1 \end{bmatrix}$$

where $w1= \pi(f_1-f_2)/f_s$ and $w= \pi(f_2+f_1)/f_s$, with f_s as the sampling frequency. The cross correlation vector $r_{xd} = [p(0)\ p(-1)]^T$, where p(0) and p(-1) represents the cross correlation function between d(n) and X(n) for lags of 0 and -1 respectively; and calculated as

p(0)= cos(w4).*sin(w3+eps)/(w3+eps)

p(-1)=cos(w1-w4).*sin(w-w3+eps)/(w-w3+eps)

where w3= w x *del* ,w4=w1 x *del* ,eps is the floating point accuracy and *del* is the fractional part of the delay calculated by the equation (4).

III. IMPLEMENTATION

The linear phased array is designed and implemented using Field II. The pressure field calculations and its variations on changing focus are obtained. On increasing the number of transducer array elements a better B- mode image with more information was obtained. Thus for the implementation of beamformer a linear phased array with 64 elements were taken. The coarse delay and fine delay are calculated. The data flow diagram is as shown in Fig.4. A two tapped interpolation filter is designed. According to the lateral and axial resolution, the number of scan lines and focal points are respectively varied. A cyst phantom obtained from Field II is taken as the test material for all the simulations.
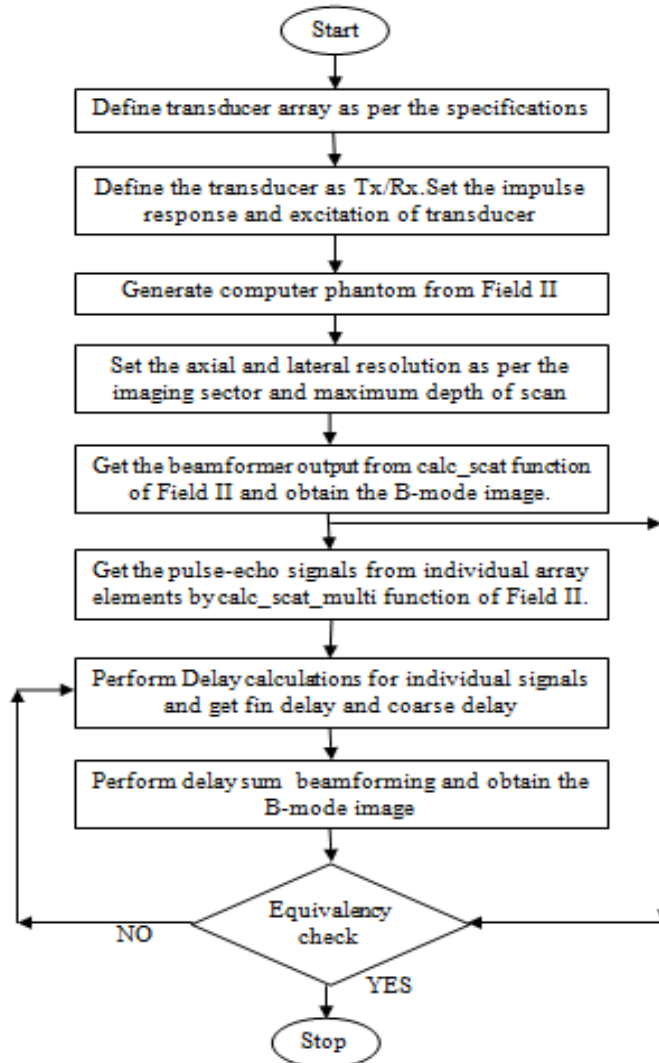

Fig.4 Data flow diagram

IV. RESULTS

Ultrasound linear focused array transducers are modelled using field II procedures and the specifications are shown in TABLE 1. The pressure field is measured at the focus as shown in Fig 5 (a).Apodization is done to reduce the side lobes to improve the directivity and done using boxcar, hanning and cosine-taped window functions. The variation in transmit pressure field by different apodization is shown in Fig, 5.As apodization with hanning window gave more directive pressure field, the further apodization processes was done using it. Transmit, receive and pulse-echo fields are measured.

For designing beamformer, initially, a linear array transducer is designed and used to get the echo signals from phantom generated by Field II. Delay Applying integer delay, fin delay and apodization, all the eight pulse echo signals are summed to form the beamformed data. To view as a B-mode image, the RF data was undergone decimation, envelope detection, log compression and scan mode conversion. The scan mode converted RF data formed a 2D B-mode image as shown in Fig. 7.

The B- mode images obtained for a single point scatter using number of array elements as 8, 64 and 128 , are shown in Fig.8.For the design of receive beamformer, initially an array of 8 elements are taken with transducer center frequency as 3.5GHz and sampling frequency as 100GHz. Fig. 9 shows the output of the beamformer focused at 45°.

The algorithm implemented in Matlab is used to obtain a B- mode image. Both images obtained by Field II beamformer and delay sum beamformer is shown in Fig.10.As the two images are identical, a receive beamformer with 64 channels is implemented and the corresponding B- mode image for cyst phantom is obtained as in Fig 11.
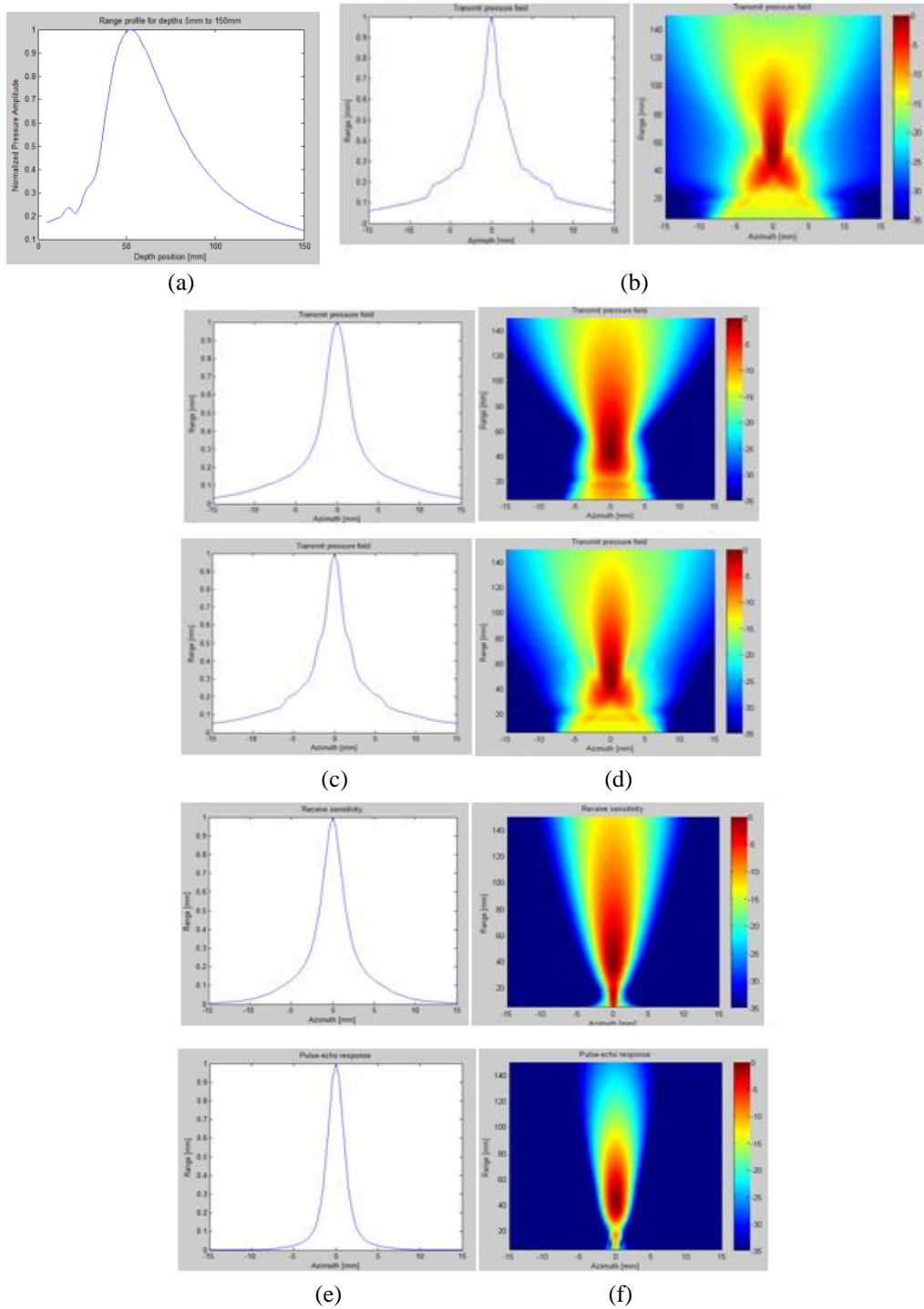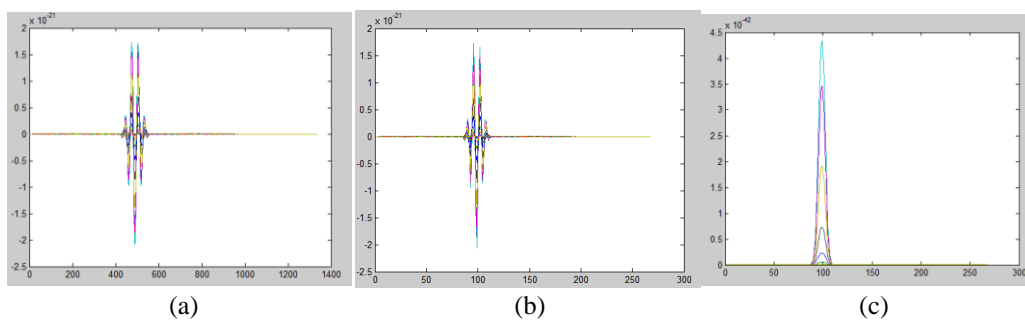

(a)

(b)

(c)

(d)

(e)

(f)

Fig. 5 (a) Pressure field variation along the depth of scan region. Transmit pressure field variation for apodization done by (b) boxcar, (c) hanning and (d) cosine-taped window functions (e) receive sensitivity (f) pulse –echo response.
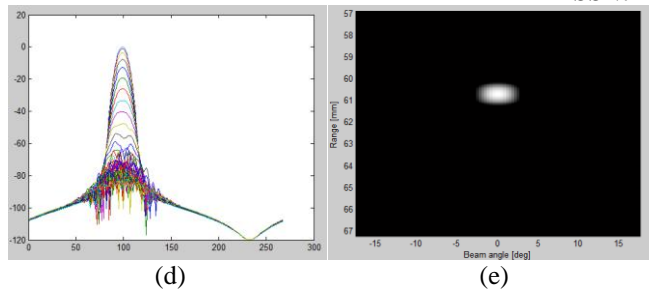

(a)

(b)

(c)

(d)   (e)

Fig. 7 (a) Beamformed RF data,(b) RF data after decimation,(c) envelope detected image,(d) log compressed image and B-mode image formed after scan conversion.
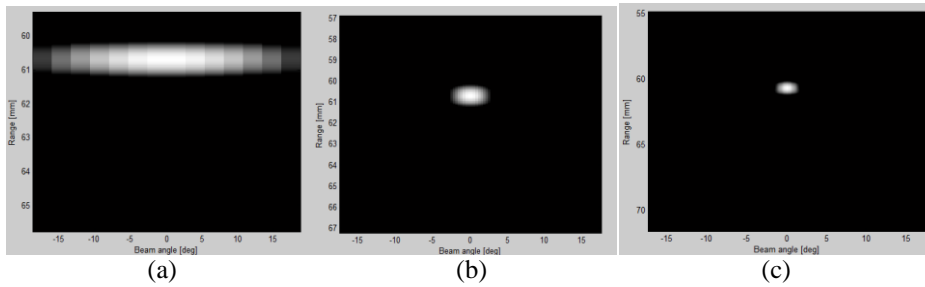

(a)   (b)   (c)

Fig. 8 B- mode images obtained for a single point scatter using number of array elements as (a) 8, (b) 64 and (c)128 .
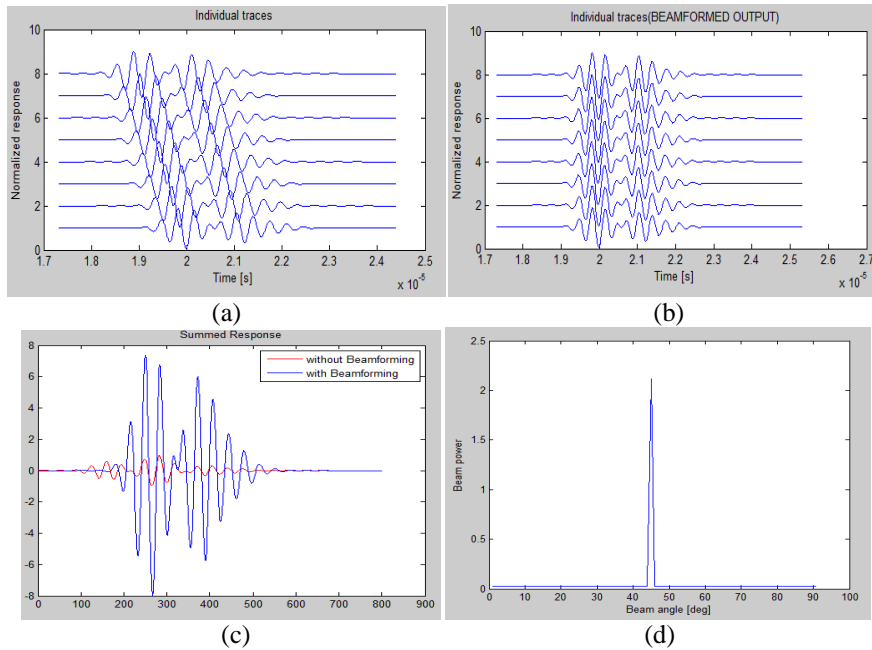

(a)   (b)

(c)   (d)

Fig.9: (a) Signals of individual elements.(b) Delayed signals of individual elements, (c)Summed response of individual signals with and without beamforming, (d) beam power plot showing maximum power only at 45°.
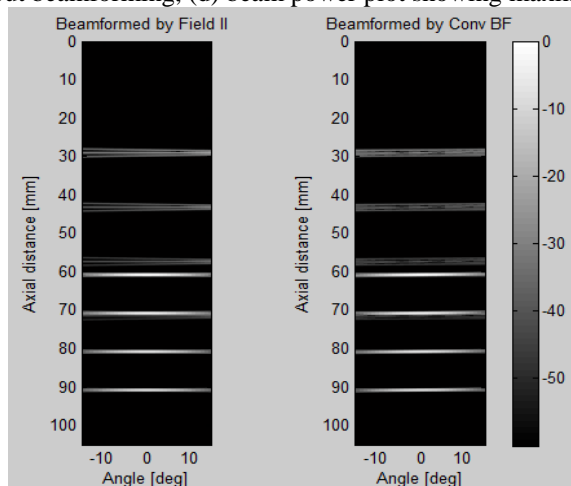

Fig.10:Comparison of Beamformer data obtained by Field II and Delay-Sum Beamformer.
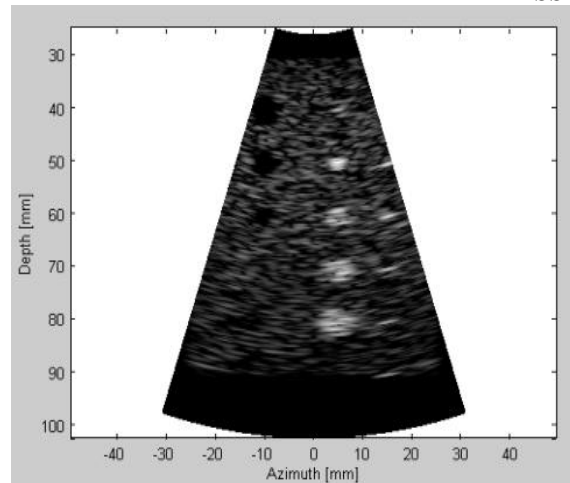
Fig. 11: B- mode image of cyst phantom from delay sum receive beamformer implemented in MATLAB.

V.   CONCLUSIONS

Initially using Field II program procedures ultrasound transducer is designed. The variation of ultrasound field in different focusing points and apodization were studied. There are many studies undergoing on beamforming in ultrasound and a better algorithm for delay calculation by increasing axial and lateral resolution may leads to new developments in the designing of Ultrasound backend systems. Here the beamformer is designed by taking phased array transducer and its calculated delay equation. The algorithm is implemented in computer phantom and obtained B-mode image is identical to that of Field II image.

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