COMPUTER PHANTOM SIMULATION

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Abstract: The aim of the current study was to analyze how number of elements of linear array and frequency influence the image quality in a homogenous medium. Linear arrays are most common for conventional ultrasound imaging, because of the advantages of electronic focusing and steering. First, was made comparison for simulated computer phantom with linear array with different number of elements (32, 64 and 128) and 3 MHz frequency. Second, the frequency was changed and the results were compared. Finally the results obtained were interpreted and discussed.

Keywords: linear array, computer phantom, image quality

1. INTRODUCTION

Phantoms are used for testing the performance of imaging equipment and to emulate acoustical properties of tissue. The acoustical properties of phantoms (such as speed of sound and attenuation) have to be close enough to soft-tissue properties. To meet this requirement, the phantom should enable the ultrasonic wave with some frequency to propagate to a defined depth. This path needs to include a sufficiently large number of small inhomogeneity, so that sufficient scatterers are presented. Therefore, the speckle shape of the phantom should be similar to that of tissue known in conventional ultrasonic scans. Another advantage of phantom modeling rather than using anatomical specimens is that phantoms can be shaped in any defined geometry.

A linear array contains N elements uniformly arranged with a distance between them. Each element performs piston-like vibration and the vibration profiles are uniform among all the elements. Ultrasonic imaging is based on a simple idea, which of obtaining the delay and sum beams. To get a scan line, array's elements are used for both emission and reception. Signal received from scattering in the medium is receive by transducer which by delaying proper inputs and their summation, maximizes the signal for a given direction not to be affected by noise or waves that propagate in other directions [1, 2].

Scanners are optimized to show the signals reflected by discontinuities, which are considerably lower than those reflected by the interfaces. These reflections are usually displayed on the screen in bright white. The image is typically grayscale and not a uniform gray or black as it should occur in a homogeneous medium. Figure 1 represent an example simulation of a 1 mm cyst (all-black area) obtained with Field II GUI.

Obtained signals are in fact the effects of constructive and destructive interference phenomena from structures whose dimensions are smaller than the wavelength. It is not possible to diagnose and view the microstructures but the signal strength may indicate pathology. Because the waves are running several contributors, it is recommended to make a characterization in terms of statistics. The distribution is Gaussian and, therefore, it is fully characterized by the average value and dispersion [3, 4]. Although the broadcast signals are characterized

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in statistical terms, they are not random, meaning that different values are obtained at each new measurement. For stationary structures there will be obtained each time the same values for the same position of the transducer. A slight change in the position sensor will however result in a correlated signal at the adjacent position. This correlation will be based on ultrasonic field width.

Since the echo depends on the interference waves from the microstructures, there are significant individual reflections of the analyzed structures. More important are variances within the tissue and its composition, key factors in determining signal strength.

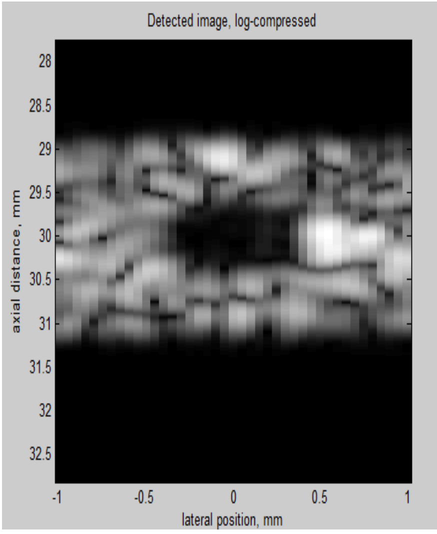

Fig. 1. Simulation of a 1mm cyst (all-black area) located at 30 mm.

Following, to achieve best results in simulated images, it is important to take care of dispersion phenomena. In literature, this issue is discussed in terms of spatial impulse response [5]. Jensen's model involves two convolutions of terms which take into account the transducer excitation, inhomogeneity in the tissue and the transducer geometry:

$$p_r(\vec{r},t) = v_{pe}(t) \underset{t}{*} f_m(\vec{r}) \underset{r}{*} h_{pe}(\vec{r},t) \tag{1}$$

$$v_{pe}(t) = \frac{\rho}{2c^2} E_m(t) \underset{t}{*} \frac{\partial^3 v(t)}{\partial t^3} \tag{2}$$

$$f_m(\vec{r}) = \frac{\Delta\rho(\vec{r})}{\rho} - \frac{2\Delta c(\vec{r})}{c} \tag{3}$$

$$h_{pe}(\vec{r},t) = h_t(\vec{r},t) * h_r(\vec{r},t) \tag{4}$$

So, to simulate a scan line the received signals from a collection of scatterers are summing, considering density and speed of sound perturbations in the tissue.

2. EXPERIMENTAL DATA AND DISCUSSIONS

For this work FIELD II [6 - 8] was used. A computer phantom which contains 5 point target and a cyst was created. The point target is separated at 5 mm. The diameter of the cyst is 10 mm. The phantom is modeled as a parallelepiped with dimensions $(x, y, z) = (40, 10, 50)$ mm.

The phantom was scanned with a 3 MHz linear array transducer. The element parameters are: height of element 5 mm, width of element 1 mm, distance between transducer elements 0.1 mm, and focal point at 70 mm. No apodization and no attenuation were applied. The generated image consists of 40 scan lines.

The main purpose of this study is to optimize a linear array transducer in order to attain high resolution and high contrast images. The optimization action is eased by simulation. Previously, we have made simulations for much type of ultrasound transducers [9 - 12]. This phantom model can be used for characterization the contrast-lesion detection capabilities of an imaging system.

In order to analyze linear array performances, there has been simulations for linear array with 32, 64, respectively 128 elements (see Figures 2, 3 and 4).

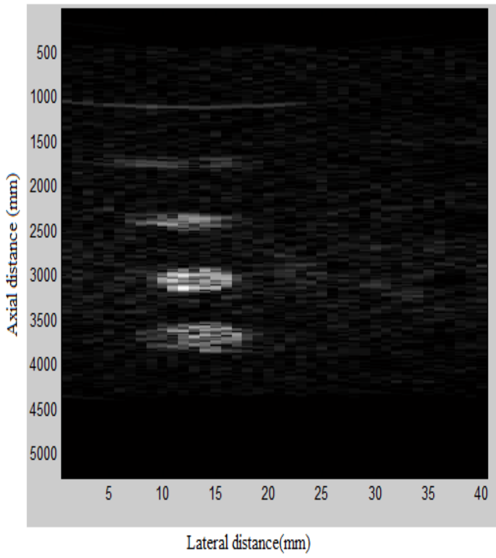

Fig. 2. Computer phantom scanned with a 32 elements linear array.

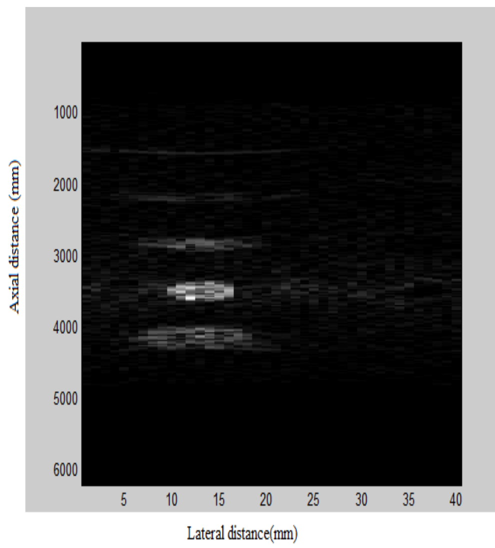

Fig. 3. Computer phantom scanned with a 64 elements linear array.

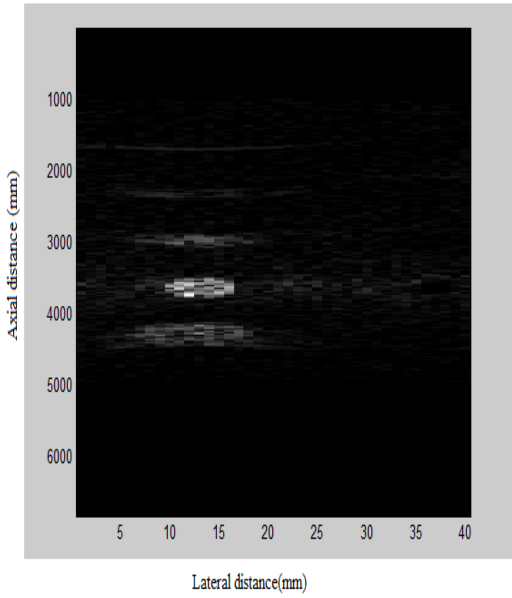

Fig. 4. Computer phantom scanned with a 128 elements linear array.

In order to identify the influence of frequency for 64 element array, we generated an image at 5 MHz, in which case a best resolution was obtained (see Figures 5 and 6).

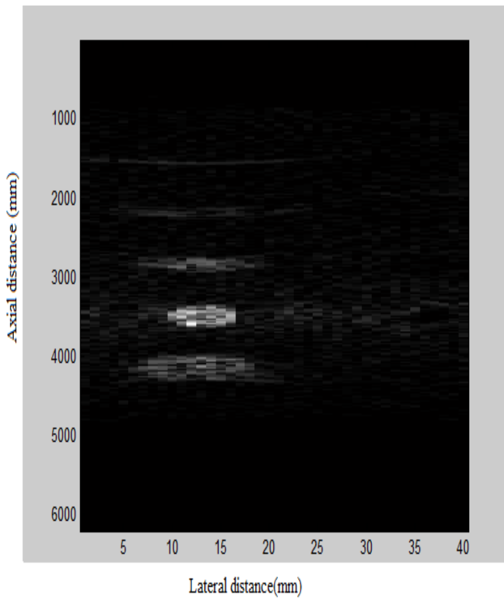

Fig. 5. Computer phantom scanned with a 64 elements linear array at 3 MHz.

Our method used three different transducers type and the same scatters. The effect of different experimental choices becomes easier to evaluate. To find the best solution we generate a computer phantom. A simple cyst phantom with point scatters has been defined and has been used in imaging. A homogeneous speckle pattern is seen along with all the features of the phantom.

The simulation was performed with an Asus x59sl series, 1660 MHz and lasted 3 hours for 32 array, 4 hours for 64 array and 6 hours for 128 array. As an observation, the simulation times increase nearly linearly with the number of elements. An improvement for a linear array image with 128 elements can be observed; the size and location of the cysts are better identifiable. In the case of the 32-transducer array there is smearing of the image, with a homogeneous speckle pattern seen along with all the features of the phantom. For 128-transducer array with the larger aperture size, there are results in higher resolution and detail in the image, in addition to a reduction in the speckle pattern.

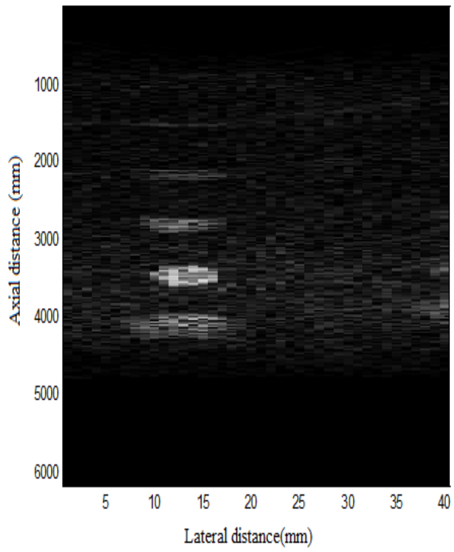

Fig. 6. Computer phantom scanned with a 64 elements linear array at 5 MHz.

3. CONCLUSION

The experiments were conducted for linear arrays of 32, 64 and 128 transducers, and the latter delivering better resolution. Increasing the number of elements will improve the results. If the frequency was increased the results were worse.

Following the simulations performed, it was found that increasing the number of elements and of the actual diameter of the transducer too, the results are superior but the generation of scan lines consume more computing resources and time.

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