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Analysis of the Functionality of New Techniques in Ultrasonic Systems and the Use of Array Transducers for Parallel Processing and Real-time 3D Imaging and Related Applications

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ABSTRACT

The optimal use of the functionalities of ultrasonic waves and the rational utilization of the capabilities of various multidimensional ultrasonic parameters have presented a unique ability in the fields and techniques related to the measurement of various physical quantities. Different types of probes, such as linear and phased arrays, are considered indispensable components in most ultrasonic imaging systems for performing many imaging processes and operations. The use of two-dimensional ultrasonic arrays is one of the solutions to prevent a decrease in contrast levels. A limiting factor in multidimensional arrays is the additional time required for data acquisition and signal processing. This paper, while refining and explaining the above-mentioned issues, analyzes the "functionality of new techniques in ultrasonic systems and the use of array transducers for parallel processing and real-time 3D imaging and related applications." In this context, other effective approaches, such as parallel processing, types and advantages of multidimensional (two-dimensional) arrays for achieving real-time 3D imaging, micro-machined ultrasonic transducers (cMUT), pyramid scanning, and straight-line 3D off-line imaging, are also analyzed in this paper.

1. INTRODUCTION

Recent advancements in ultrasonic imaging systems have significantly improved real-time imaging capabilities, particularly with the integration of array transducers. These transducers enhance parallel processing and enable high-resolution 3D imaging, making them invaluable in clinical applications. Huang and Zeng [1] discuss the role of real-time 3D ultrasound imaging, emphasizing the importance of data acquisition and reconstruction algorithms in optimizing image quality. Similarly, Kumar et al. [2] highlight how array transducers contribute to better imaging resolution, improving diagnostic precision. Wang et al. [3] further support these findings by comparing real-time 3D

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transesophageal echocardiography (RT3DTEE) with traditional 2D methods, demonstrating the superior accuracy of 3D imaging in assessing anatomical structures like the left atrial appendage. This improved diagnostic capability has been instrumental in medical decision-making, particularly in cardiology.

The integration of photoacoustic imaging (PAI) with conventional ultrasound techniques has also emerged as a promising approach to enhance imaging resolution and tissue penetration [4]. Recent developments in transducer technology, such as capacitive micromachined ultrasonic transducers (CMUTs) and piezoelectric micromachined ultrasonic transducers (PMUTs), have further improved imaging efficiency. Brenner et al. [5] discuss the impact of PMUTs with ScAlN thin films, showcasing their enhanced electromechanical properties, while Li et al. [6] emphasize the advantages of CMUTs in terms of thermal efficiency and broadband performance. Additionally, efficient data storage and processing solutions are crucial for managing the large datasets generated by advanced imaging techniques. Soroush et al. [7] highlight the significance of optimized storage management strategies in facilitating real-time data processing for ultrasonic systems. These technological advancements collectively contribute to improving clinical outcomes, as demonstrated by Thavendiranathan et al. [8], who explore the role of automated 3D imaging in evaluating mitral regurgitation severity, reinforcing the transformative potential of ultrasonic imaging in medical diagnostics.

In a paper authored by one of the writers of this article, it was stated that [9]: by intelligently employing ultrasonic tools and systems, as well as piezoelectric sparse arrays and ultrasound tomography capability to obtain multiple slices of an image, and by mounting the probe on a mechanical translator (position encoder), a wide and deep range of three-dimensional and real-time imaging can be achieved. A clear example of this is the reduced contrast outside the elevation focus range. The ultimate solution to this issue is the use of a two-dimensional array [10]. Unfortunately, for a 128×128 element two-dimensional array, if it is uncontrollable, the electronic channel and cable count become significantly large, making its implementation and usage very challenging. As an intermediate step to address part of the issues in slice thickness, multidimensional arrays such as 1.1/25 and 1.1/75 have been developed. The main issue all multidimensional arrays face is the increased time required for data acquisition and signal processing. Existing scanners have almost exclusively adopted a solution in which a pulse is only sent after all echoes within the field of view have been received. To gain extra time without compromising image quality, the presence of parallel processing capability is crucial.

2. PARALLEL PROCESSING

Parallel processing can be achieved by sending multiple pulses simultaneously. However, this strategy might enlarge the scanners and complicate their electronics. Figure 1 illustrates a more logical solution proposed by Shattuck and colleagues in 1984, which is typically used in linear arrays [11]. In this approach, a wider beam is sent compared to the beams typically used in linear arrays, and the returning echoes are simultaneously detected in 4 to 16 detection paths. This is made possible by appropriately adjusting time delays.

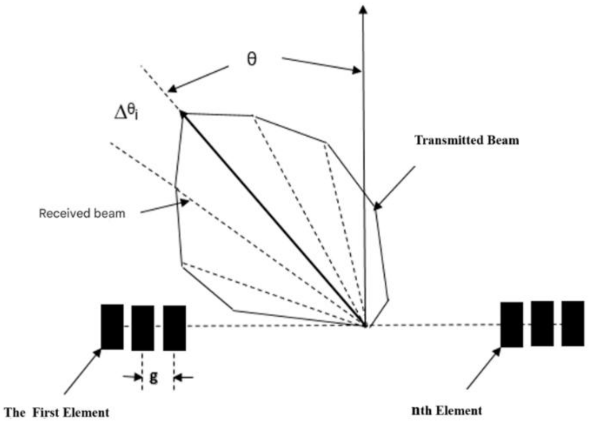


Fig. 1. General schematic of the solution proposed by Shattuck and colleagues.

In the initial approach [11], analog (continuous) delays were utilized. For reception at the angle $\theta + \Delta\theta_i$, where θ is the transmitted beam's steering angle and $\Delta\theta_i$ is the angle between **θ** and the **i**-th received beam ($i = 1...M$), depending on the specific design, M can range from 4 to 16. Additional delays, Δt_i, can be applied to the main delay at a steering angle θ, as expressed in Equation (1).

$$\Delta t_i = \Delta t_0(\theta, R_f) + \frac{ng}{c} sin(\Delta\theta_i) \tag{1}$$

where R_f is the focal distance of the beam, Δt_0 is the time delay required to focus the beam at R_f at an angle θ, n represents the number of elements, g is the pitch, and **c** is the speed of sound in the surrounding medium.

This expression is only valid when θ and $\Delta\theta_i$ are smaller than 26 degrees. For a transmitted pulse, when M = 4, four lines are received. This means that the time required to form a single frame is reduced to one-fourth, indicating a fourfold increase in frame rate. This achievement comes at the cost of a slight reduction in lateral resolution (due to the wider transmitted beam) and increased complexity of the electronic circuitry.

3. MULTIDIMENSIONAL ARRAYS

Multidimensional arrays are classified into four categories: 1.25D arrays, 1.5D arrays, 1.75D arrays, and 2D arrays [12]. The lateral views of these arrays are shown in Figure 2.

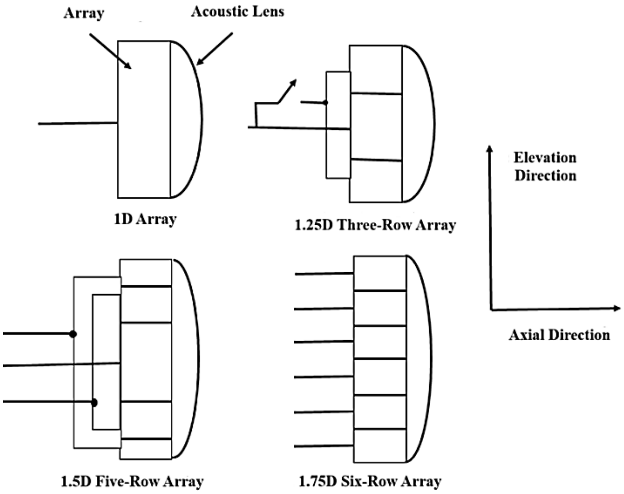


Fig. 2. Lateral View of Multidimensional Arrays

In a one-dimensional array, the vertical aperture and the focal distance are fixed. In a 1.25D array, the diaphragm aperture size is variable, but the focus is fixed. For near-field imaging, the switch is open, and only the central row is used. In a 1.5D array, dynamic focusing is achieved by adjusting the delays of the received echoes or transmitted pulses, similar to what can be done in the azimuth plane. A 1.75D array is similar to a 1.5D array, but with the difference that symmetry constraints are not present. Like circular arrays, to ensure uniform sensitivity and input impedance, the row environments are typically equalized. Only a full two-dimensional array allows dynamic focusing and beam steering in both the elevation and azimuth directions.

The main advantage of using multidimensional arrays is better control over slice thickness and enhanced contrast. The limitations include:

- When a large number of element rows are present, lobes along the elevation direction become noticeable and unavoidable.
- Increased residuals and related artifacts.
- Increased complexity in electronic circuits.

- With the development of 1.5D arrays, where more than 1048 electronic channels are used for image acquisition, significant improvements in image quality have been achieved.

3.1.Two-Dimensional Arrays

To achieve real-time three-dimensional imaging, a 128×128 element two-dimensional array must be used. This allows for functionality similar to that of a 128-element one-dimensional array in the azimuth direction, but extended in the elevation direction. Three-dimensional imaging using a two-dimensional array, as shown in Figure 3, can be performed in two ways: pyramid scanning and straight-line scanning.

In pyramid scanning, a full two-dimensional aperture is used to steer and focus the beam in all directions (elevation and azimuth). In straight-line scanning, only a few apertures are utilized, and the beam is used in a linear and clustered format to acquire the full three-dimensional image.

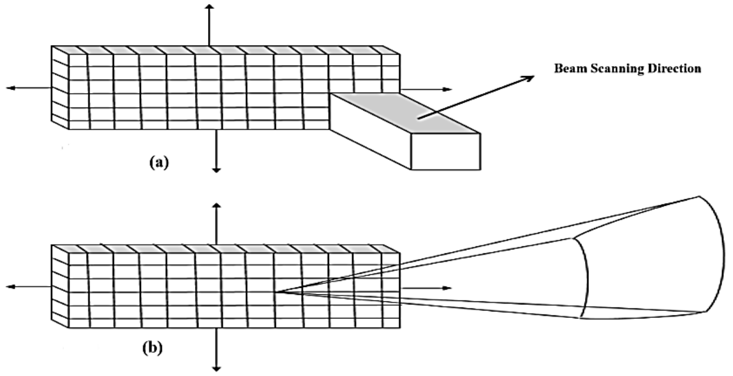


Fig. 3. (a) Straight-line scanning by a two-dimensional array, and (b) Pyramid scanning by a two-dimensional array.

Straight-line scanning offers advantages such as reduced complexity and, consequently, a lower cost, but it results in lower resolution due to the smaller diaphragm aperture. Constructing a 128×128 element two-dimensional array is not an unsolved issue and is feasible for implementation. However, challenges such as impedance mismatch due to the small diaphragm aperture, the large number of electronic channels when all elements are connected, and the internal connections present difficulties and problems associated with these arrays. These issues can be overcome by using piezoelectric materials with a high dielectric constant or multi-layer piezoelectric material that are acoustically in series and electrically in parallel [13].

Multi-layer piezoelectric materials reduce the impedance mismatch of the output element of the array by a factor of N, where N represents the number of layers. It is important to note that the electrical behavior of a piezoelectric transducer is similar to that of a capacitor near its resonant state. Using flexible multi-layer circuits can solve the internal connection problems, but the cable size remains challenging to manage. A flex configuration for a 5/1-dimensional array of 39×6 mm is shown in Figure 4, where Figure 4-a displays an image of the array set and Figure 4-b is an enlarged image of a section of the array. The number of channels can be feasibly reduced by multiplexing or adopting a sparse array strategy (an array where most of the inputs have zero values) [10].

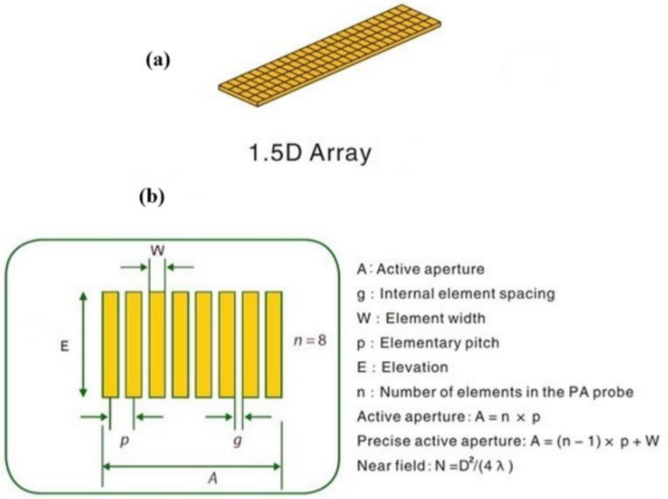


Fig. 4. (a) Image of a 5/1-dimensional array with multi-layer flex, and (b) Enlarged image of a section of the same array.

A reported complete two-dimensional array consisting of 60×60 elements, designed for real-time three-dimensional volumetric imaging, is currently commercially available. In this device, multiplexing and fast electronic components are likely used to reduce the number of channels while maintaining a rapid frame rate of 30 frames per second. Sparse arrays have been used for many years as an alternative to address this issue.

A recent significant advancement in transducer technology with considerable potential for constructing a two-dimensional array with high-cost efficiency and resolving internal connection problems is the capacitive Micro-Machined Ultrasonic Transducer (cMUT). This solution is entirely different from conventional transducer design strategies and offers the advantage of utilizing semiconductor technology, which enables the integration of transducers and electronic components, thus allowing their miniaturization. The general principle of this technology is illustrated in Figure 5.

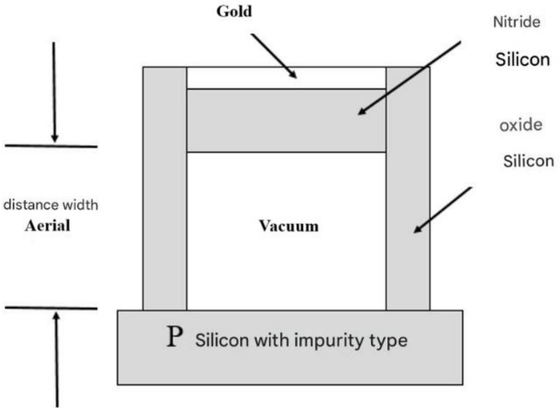
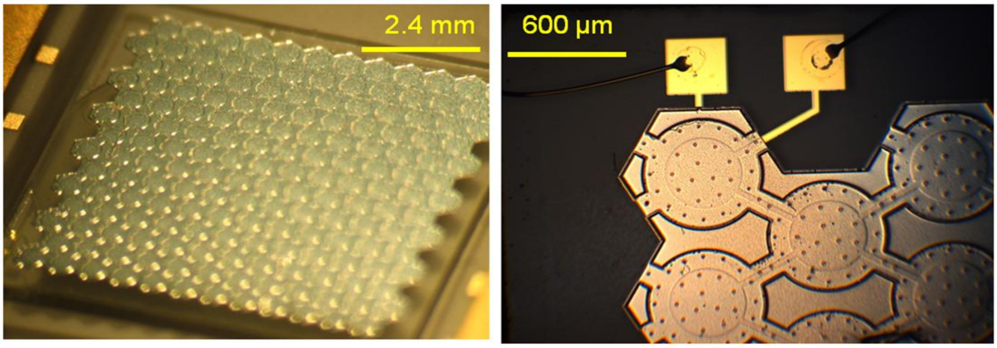


Fig. 5. Structure of a cMUT cell.

The silicon nitride membrane creates a capacitor that causes the membrane to vibrate when an AC voltage is applied. The electrode is made of a layer of gold. The sensitivity of the device is inversely proportional to the width of the air gap, so that as the gap width decreases, the sensitivity increases. The thickness of the membrane determines the resonance frequency of the device, and at a thickness of 12 micrometers, the resonance frequency is approximately f.

Figures 6-a and 6-b show images of several cMUT cells, where the lighter-colored structure corresponds to the membrane cells and the conducting paths, and an atomic force microscopy (AFM) image of a single cell is displayed.



(a) (b)

Fig. 6. (a) Image of several cMUT cells, (b) Atomic force microscopy image of a single cMUT cell.

Figure **7** shows a 192-element array constructed with this technology, and Figure 8 displays the corresponding image obtained from a conventional PZT array. An improved axial resolution, resulting from the wide bandwidth of the cMUT, is clearly observable.

Other advantages of cMUT include the elimination of the need for acoustic impedance matching between the transducer and the loading medium. However, this technology also has some drawbacks, such as a slight decrease in sensitivity and the need for a bias voltage on the order of 100 volts.

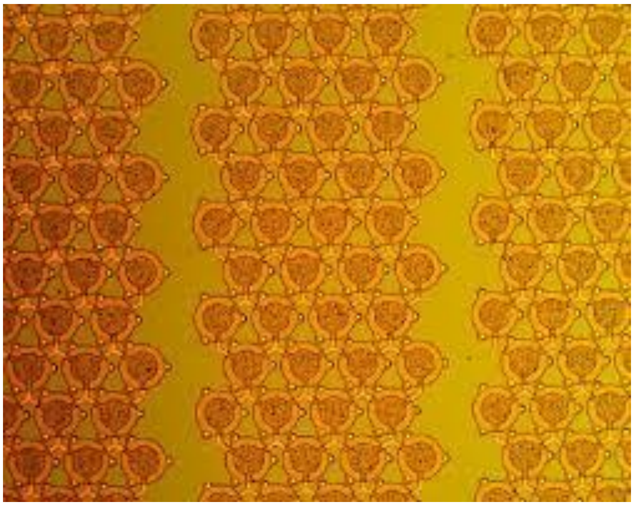


Fig. 7. Image of a 64-Element Linear cMUT Array

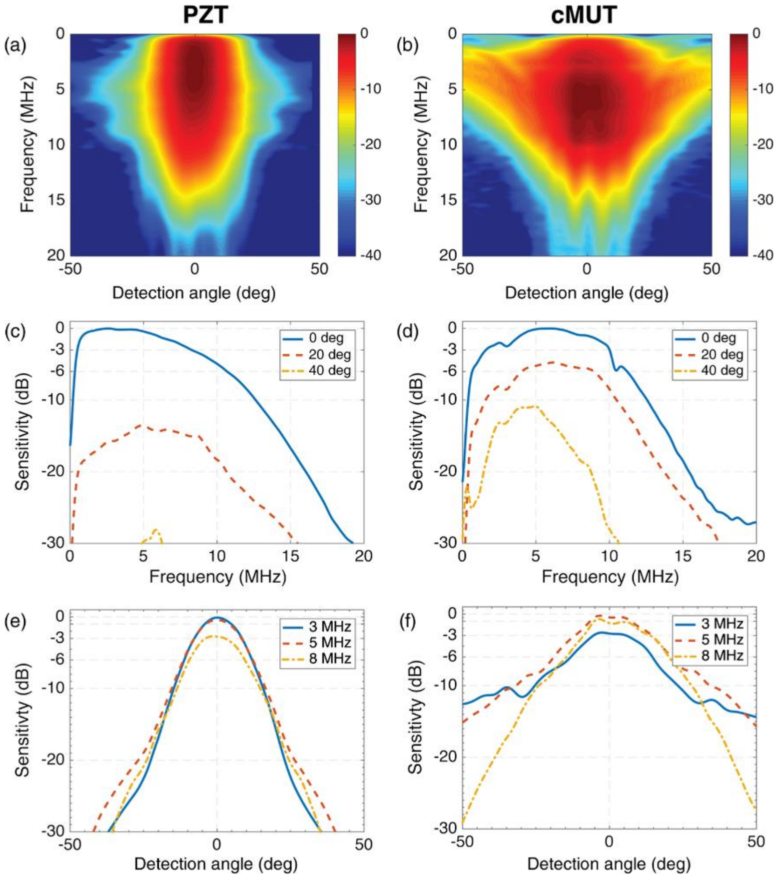


Fig. 8. Comparison of Images Obtained from a Breast Tumor at a Frequency of 9 MHz
(a) Image Obtained from a cMUT Array, (b) Image Obtained from a Conventional PZT

3.2. Sparse Arrays

To reduce the number of elements and channels, sparse arrays can be employed. Piezoelectric elements, as illustrated in Figure 9, are randomly positioned on a pre-determined diaphragm aperture. All elements may be used for both transmission and reception, or a subset may be designated for transmission and reception. A drawback of this sparse array is that its sensitivity decreases due to the reduced size of the diaphragm aperture, resulting in an increase in the sidelobe level or background noise outside the main lobe. It has been demonstrated that this effect is inversely proportional to the number of elements.

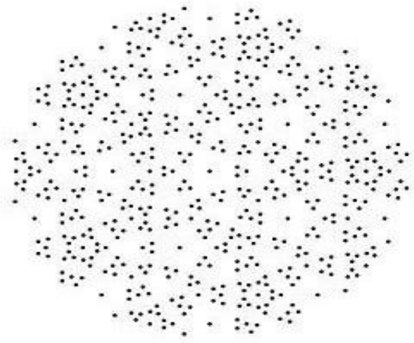


Fig. 9. A randomly scattered array, where the squares represent the piezoelectric elements.

The sidelobes resulting from a large step in a sparse periodic array can be mitigated by selecting different steps for the transmit and receive arrays. If L represents the array width, the transmitted radiation pattern in a direction φ_x and at a distance $r >> L$, as shown in Figure 10, is obtained using Equation (2).

$$H_T(u) = \int a_T\left(\frac{x}{\lambda}\right) e^{jk\left(\frac{x}{\lambda}\right)u} d\frac{x}{\lambda} \tag{2}$$

Where $u = sin\,\varphi_x$, λ is the wavelength, **k** is the wave number, and a_T represents the aperture window function of the transmitter.

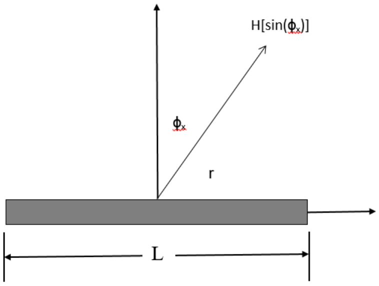


Fig. 10. Geometric Representation of Radiation for a Linear Array

This expression fundamentally states that the radiation pattern of a diaphragm aperture is the Fourier transform of the diaphragm aperture function. For a linear array, the diaphragm aperture function can be represented by Figure 11, where b denotes the element width. If the selected independent acoustic variable is pressure, $a_t(x/\lambda)$ consists of a series of pulses with pressure amplitude P_t. Here, the distance xxx is normalized with respect to λ. Additionally, the radiation pattern is derived from Equation (3).

$$H_R(u) = \int a_R\left(\frac{x}{\lambda}\right) e^{jk\left(\frac{x}{\lambda}\right)u} d\frac{x}{\lambda} \tag{3}$$

Where a_R is the diaphragm aperture function for reception. The two-path pulse-echo radiation pattern is given by Equation (4):

$$H_{TR}(u) = H_T(u)\,H_R(u) = FT[a_T]\,FT[a_R] = FT[a_T * a_R] \tag{4}$$

Where $***$ denotes convolution, and $E(x/\lambda) = a_T(x/\lambda) * a_R(x/\lambda)$ is commonly termed the "effective diaphragm aperture function" or "array complement function."

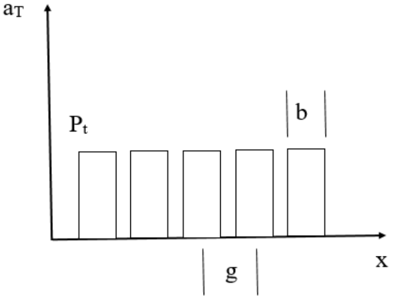


Fig. 11 Transmission Diaphragm Aperture Function for a Linear Array with Pitch g and Element Width b

In the far field of the array, i.e., r≫l, square pulses can be represented as impulses, as shown in Figure 12. Assuming that the elements NTN_TNT and NRN_RNR are present in the transmission and reception diaphragm apertures, respectively, based on convolution, the number of elements in the effective diaphragm aperture should be $N_{TR} = N_T + N_R - 1$, with a width of $2L$ and a pitch of half the wavelength λ.

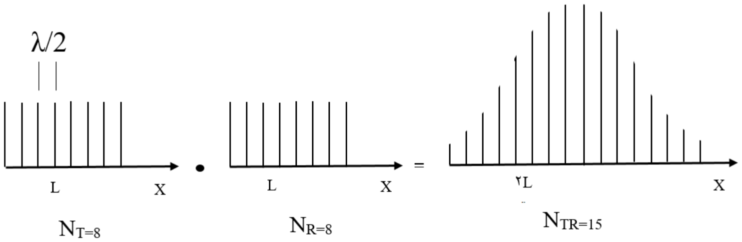


Fig. 12. Desired Effective Diaphragm Aperture Function Obtained from the Convolution of the Transmission Diaphragm Aperture Function with the Reception Diaphragm Aperture Function

The desired effective diaphragm aperture can be reconstructed from the transmission and reception diaphragm aperture functions using various methods. In Figure 13, two such methods are illustrated. This approach is valid only for the far field of an unfocused array and in the focusing of a phased array.

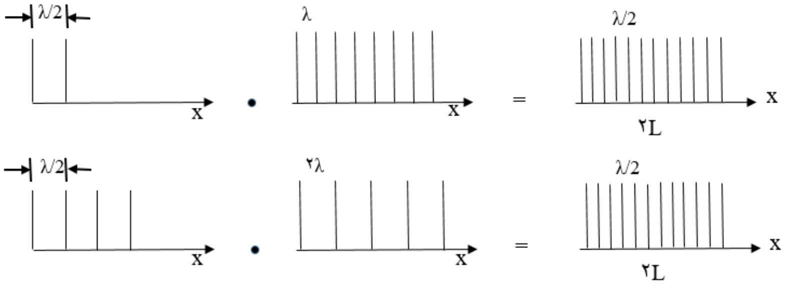


Fig. 13. Two approaches used to achieve the desired effective diaphragm aperture function.

4. THREE-DIMENSIONAL IMAGING

Three-dimensional ultrasound (3DUS) is a newly developed and significant field that utilizes the tomographic capability of ultrasound by acquiring multiple slices of an image. Three-dimensional reconstruction can be performed either offline or in real-time (real-time), using two-dimensional arrays and parallel processing [7]. If a scanner is capable of displaying volumetric images in real-time, it is typically referred to as a four-dimensional scanner.

Offline three-dimensional imaging is performed using a free-hand scan with an electromagnetic position sensor or by mounting the probe on a mechanical translator, whose position is encoded. The acquired data, after image processing performed by optimized algorithms, are displayed on a high-resolution screen.

Currently, three-dimensional ultrasound images are displayed in two ways: as a series of images perpendicular to each other, and as images that depict the three-dimensional structure. Image rotation is provided as an option. A three-dimensional image of a fetus obtained offline, and a three-dimensional image of a heart obtained in real-time or by a four-dimensional scanner, are shown in Figures 14 and 15, respectively [15].

Fig. 14. Three-Dimensional Image of a Fetus in the Womb Captured Offline [15]

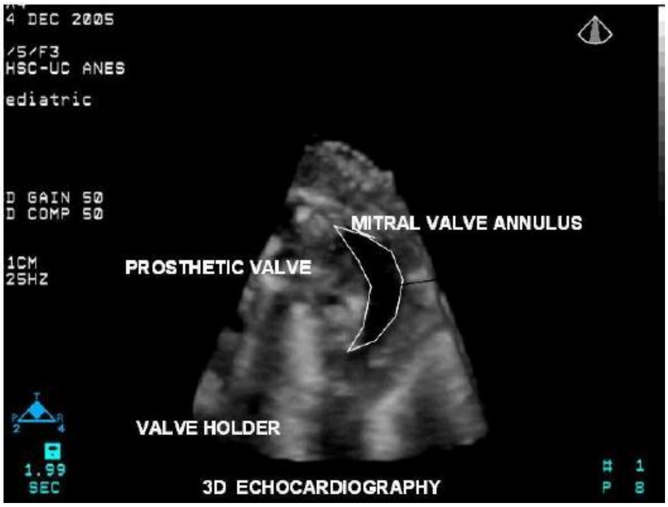


Fig. 15. Three-dimensional image of a heart valve obtained in real-time or by a four-dimensional scanner [15].

Some clinical applications of three-dimensional ultrasound include:

- Coronal plane imaging of a newborn's heart, which cannot be displayed using two-dimensional ultrasound;
- Measurement of the volumes of organs such as the heart, etc.;
- Guiding invasive methods such as needle biopsies of the prostate.

Like other imaging techniques, the clinical impact of three-dimensional imaging has not yet been definitively established and requires further research. However, it is evident that for 3DUS to have significant clinical effects, its performance must align with that of two-dimensional ultrasound in terms of image quality and cooperative functionality.

5. CONCLUSION

The equipment in most ultrasonic imaging systems, including various linear probes, Doppler transducers, etc., is essential and unavoidable for performing many imaging operations and preventing issues like contrast degradation. Multidimensional arrays have longer data acquisition and signal processing times. The reduction and elimination of this limitation, through the optimal use of relevant scanners and parallel processing capabilities, has provided exclusive solutions without compromising image quality.

The strategy of parallel processing and using linear arrays, among other approaches, has reduced the time required to form a frame by a quarter, in other words, it has increased the frame rate by four times (considering M=4). It is evident that this optimization results in a slight reduction in lateral resolution and increased complexity in the associated electronic circuits. The use of multidimensional arrays (in their four types) has been considered an invaluable approach to reducing the limitations and flaws mentioned in this paper. To ensure consistent input sensitivity and electrical impedance, the row environments are typically made equal. Only a complete two-dimensional array allows dynamic focusing and beam steering in both elevation and azimuth directions.

A two-dimensional array can be used to achieve real-time three-dimensional imaging.

Three-dimensional imaging can be performed in two ways: (I) Pyramid scanning (for beam steering and focusing in the elevation and azimuth directions using two two-dimensional apertures), and (II) linear scanning (only one aperture is used where the beam is linear for acquiring a complete image). While linear scanning has advantages such as reduced complexity of electronic circuits and cost, it has poor resolution due to the small aperture size. Furthermore, using piezoelectric materials with high dielectric constants or piezoelectric materials arranged in parallel acoustically and in series electrically can overcome the issue of "electrical impedance mismatch."

The design and construction strategy of capacitive micro-machined ultrasonic transducers (cMUTs) offers advantages over traditional transducers by utilizing semiconductor technology, which enables the integration of transducers and electronic components and, consequently, their miniaturization. Despite the need for a bias voltage of approximately 100V and a slightly lower sensitivity, cMUTs have the significant advantage of eliminating the need for acoustic impedance matching between the transducer and the loading environment.

Three-dimensional ultrasound (3DUS), which performs three-dimensional reconstruction both offline and online using two-dimensional arrays and parallel processing, represents a significant and developed field with tomography capability.

The data obtained after image processing, performed by optimized algorithms, are displayed on a high-resolution monitor. Like other imaging techniques, the clinical impact of three-dimensional imaging has not yet been definitively confirmed and requires further research. It is evident that for 3DUS to have significant clinical effects, it must align with the performance of two-dimensional ultrasound in terms of image quality and other factors.

Transparency Statement

The data supporting this study are available upon reasonable request to the corresponding author, subject to ethical and confidentiality considerations.

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Declaration of Interest

The authors declare that they have no competing interests.

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