

Quantitative, Free-Hand 3D Ultrasound Imaging based on a Modified 1D Transducer Array

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ABSTRACT

An approach for acquiring dimensionally accurate 3D ultrasound data, based on a modified 1D transducer array, is presented. The method avoids many of the drawbacks of conventional approaches to 3D ultrasound data acquisition. Scanning is simple and easy to perform in a clinical setting.

A modified 1D transducer array is employed comprising a central conventional 1D *'Imaging'* array and two perpendicular *'Tracking'* arrays – each integrally mounted at each end of the *'Imaging'* array. As the transducer is scanned in the elevation direction of the central array, the images acquired by the *'Tracking'* arrays remain coplanar and hence it is possible to accurately track image motion using any one of several image tracking techniques.

Methods for improving the performance and ergonomics of the transducer array are presented. In particular, a crossed electrode transducer structure is proposed for minimizing the total transducer 'footprint' (contact surface area). The versatility of the approach in terms of its suitability for scanning breast, carotid and prostate is discussed. We have acquired both phantom and *in-vivo* 3D ultrasound data with the prototype imaging approach. Initial studies suggest that the linear dimensional accuracy in the elevation direction (i.e., the reconstructed direction) is approximately 5%.

Keywords: Ultrasound imaging, Ultrasound transducers, Speckle tracking, 3D ultrasound

1. BACKGROUND

Although all anatomy is 3D dimensional in form, the vast majority of ultrasound imaging is 2D. Most of the time 2D provides sufficient information but there are clearly identifiable limitations:

1. Volume Measurements

While linear dimension and cross-sectional area can be determined from a 2D view, volume – arguably the most important clinical parameter – cannot be accurately determined. Current practice frequently calls for gross assumptions to be made regarding the relationship between the observed area and the estimated volume. This is frequently performed when estimating heart chamber volume using a limited number of 2D chamber cross-sectional views.

2. Some views are impossible to obtain

When using a 2D scanning probe it is not possible to obtain the 'C-Scan' view – i.e., a 2D slice parallel to the skin surface. Notice that an inability to obtain the 'perfect' 2D slice means that even an estimate of linear dimension or area may be erroneous. For example, if one is trying to measure the longest linear dimension of an organ of interest, then one must obtain a 2D view that includes the appropriate endpoints in 3D space.

3. Optimal 2D scan views may be missed

It is common practice, particularly in the United States, for a non-M.D. trained sonographer to acquire one, or more, 2D image slices which are later 'read' by a fully qualified (i.e., M.D.) radiologist who makes the diagnosis. The radiologist is dependent on the sonographer's ability to obtain the most clinically significant view. If the radiologist is not satisfied with the view, then it may be necessary to recall the patient and make a second study. This results in significant inconvenience, delay and added, potentially unnecessary, cost. A better clinical practice model would involve the sonographer acquiring a comprehensive 3D data set which would be later reviewed by the radiologist using a 'virtual examination' technique. In the 'virtual examination' the radiologist would navigate through the 3D volume to find the most diagnostically significant 2D slices. The reliability of the diagnosis would be further enhanced by the ability to capture multiple 2D views of the region of interest and thereby provide a means of corroboration.

4. 3D Images for clinical reports to referring, non-radiologist, physicians

3D Images are frequently easier to interpret – especially for users not experienced in viewing cross-sectional 2D slice B-Mode images. 3D Images may be more informative to referring physicians and may also assist in surgical planning where surgeons may feel more confident with their interpretation of a 3D view than with a set of 2D cross-sections.

The above issues strongly suggest the need for an easy-to-use, practical, low-cost quantitative 3D ultrasound acquisition method. In this context ‘quantitative’ is defined as meaning that the resulting images are dimensionally accurate to some known tolerance and that therefore clinically useful linear, area and volume measurements can be made. It is worth noting at this point that the majority of currently available 3D ultrasound acquisition techniques offered commercially are not sufficiently dimensionally accurate to permit quantitative measurements and make gross assumptions about the nature and speed of the volumetric data acquisition.

It is worthwhile briefly reviewing existing methods for acquiring 3D ultrasound data.

1. Estimated position

A 1D transducer array acquiring 2D cross-sectional images is swept across the volume of interest in the elevation direction. The frame to frame location is estimated based on assumed motion of the transducer¹. This method requires a degree of skill in order to be useful and does not permit repeatable accuracy. It is worth noting in passing that errors in position estimation do not automatically result in evident breakdown of the resulting 3D images. In other words, the result of incorrect frame location estimation is distortion - not defocusing. Thus, the user does not get immediate guidance as to the extent of distortion that is due to imperfect scanning technique.

2. Position estimated using image frame decorrelation

This is similar to the approach described above in that the 1D transducer array is swept through the volume in an elevational direction resulting in a sequence of parallel 2D image planes. However, in this approach, frame to frame motion is estimated based on image data decorrelation². The extent of decorrelation is a function of focusing resolution in the elevation direction and hence, in a reverse process, knowing the calculated decorrelation value allows the associated motion to be estimated. Generally, a calibration acquisition is performed in advance so that an accurate decorrelation versus elevation motion function can be used. This technique is very elegant in that it uses currently available 1D transducers with no hardware modifications. However, one drawback is that the decorrelation function is symmetric and hence there is a substantial risk of a left / right ambiguity in the resulting 3D acquisition data. Although the method is in principle capable of providing quantitative data (i.e., dimensionally accurate), the authors are not aware of specific dimensional accuracy claims being made on behalf of any of the commercial products based on the approach.

3. Mechanical positioner

Mechanical, motorized guides for moving a 1D transducer array in a repeatable manner have been used for a long time³. They are well suited to research use but are less well accepted in general clinical use. The main limitation is that a different guide is required for every different type of scan – i.e. they lack versatility. Historically, they have tended to be large, heavy and clumsy. However, while the lack of versatility remains a problem, there has been recent progress in making mechanically scanned transducers less clumsy⁴. Additionally, motorized rotationally scanned trans esophageal echocardiography (TEE) acquisitions are reasonably well accepted. In this case, the electro-mechanical controls originally developed to enable rotational multi-plane TEE, are reused with minimal, or no, hardware modification and excellent 3D TEE cardiac images are obtained. It should be emphasized that in multi-plane TEE 3D reconstructions it is necessary to use some kind of cardiac cycle synchronization (e.g. ECG based).

4. Magnetic positioner

In this case a magnetic positioner is attached to the 1D transducer array⁵. Typically, the magnetic positioner on the transducer is a ‘receiver’ coil set approximately 2.5 cm in each dimension of a cube-based structure. This block causes a certain degree of ergonomic problems but is otherwise reasonably well accepted since the sonographer is free to acquire the volume data free of the constraints of a mechanical positioning system and without the cumbersome limitations of allowed motion sequences when using any of the estimated position based methods. The receiver coils detect the fields created by a transmitter coil set which is typically of the order of 10 cm on each side. However, smaller coil sets (for both transmitter and receiver) have been, or are being, developed.

5. Optical and Acoustic Positioners

As an alternative to the magnetic positioner, an optical positioning method can be used⁶. A set of infra red (IR) LEDs are mounted on the transducer. These are observed using dual IR cameras rigidly mounted above the field of interest. The relative positions of the LEDs, as they appear in each of the two offset camera views, allows the position and orientation of the transducer to be determined. This method is significantly more accurate than the magnetic positioner method but relies on maintaining a line of sight from the transducer mounted LEDs to the camera. It is also a relatively expensive technology – approximately \$10,000. Similarly, a method based on using acoustic sources (spark gaps) mounted on the transducer which are monitored by microphones rigidly mounted in the vicinity of the field of interest was proposed by King some years ago⁷. This latter approach has not achieved widespread clinical acceptance to-date.

6. 2D Arrays

2D Arrays for real-time acquisition of 3D data have been in development at Duke University for several years⁸. This has resulted in a commercial spin-off company (Volumetrics). 2D Arrays have the obvious technical advantage of being truly real-time. However, the cost and the complexity of both system and transducer have slowed development and acceptance of this technology. 2D arrays represent the preferred long-term path in cardiology where rapid tissue motion presents fundamental problems that cannot be addressed by 1D swept array based processes.

In summary, while 2D arrays are the preferred long-term solution for imaging highly dynamic regions (i.e., the heart), there remains a need for improved quantitative methods for scanning static, or slowly moving, regions using a sequence of 2D image frames captured using a sweep with a 1D transducer array. In this case, the problem reduces to one of accurately determining the location, and orientation, of each of the sequence of acquired 2D image frames.

2. NEW ACQUISITION APPROACH

A new method based on a modified 1D transducer array has been developed^{9,10}. The transducer array has a regular 1D *Imaging* array in the center. On each end of the *Imaging* array a dedicated *Tracking* array is integrally mounted. The *Tracking* arrays are oriented perpendicular to the *Imaging* array. Therefore, as the transducer is scanned in an elevation direction (with respect to the central *Imaging* array), the image sequence acquired by the *Tracking* arrays is coplanar and moves from side to side. It is possible to track image motion between successive coplanar images in a sequence and therefore it is possible to estimate the frame to frame motion of the *Imaging* array. Ultimately, as the names suggest, only data from the *Imaging* array is used in 3D reconstruction. Image data acquired by the *Tracking* arrays is used merely to determine the position and orientation of the sequence of *Imaging* data frames. A modified transducer is illustrated diagrammatically in Fig. 1. The transducer is scanned in the manner illustrated in Fig. 2.

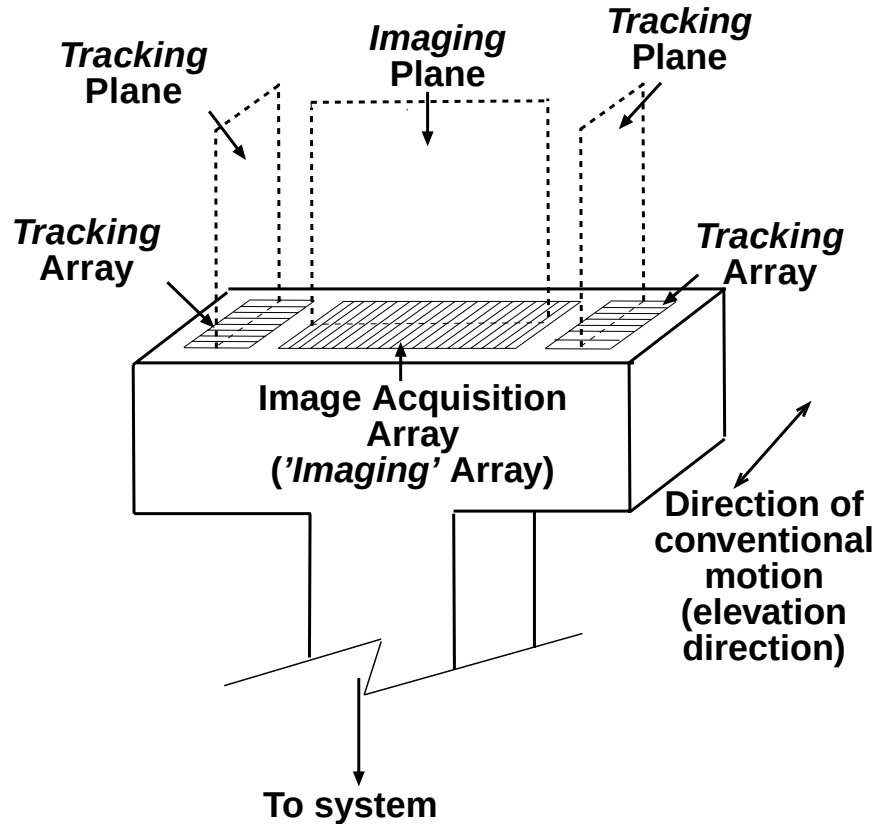


Fig. 1 Modified transducer array comprising a central *Imaging* array and two perpendicular, integrally mounted *Tracking* arrays.

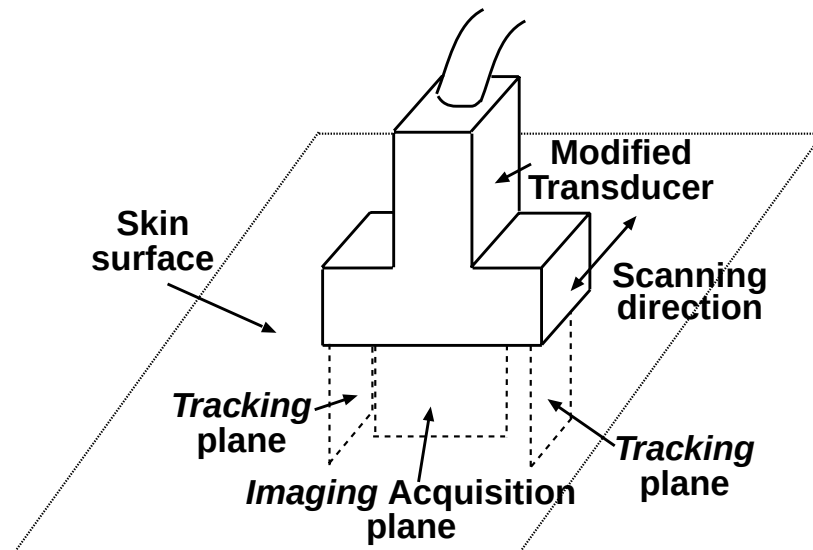


Fig. 2 Illustration of the how the modified transducer array is swept through the volume of interest. Notice how the *Imaging* plane sweeps through the volume (in a direction perpendicular to the orientation of the *Imaging* plane) while the *Tracking* images planes remain coplanar.

Frame to frame image matching has been exploited with considerable clinical success to form Extended Field of View images¹¹. The precise algorithms used in these products are generally proprietary. However, Weng, in several papers and patents^{11,12}, has described the basic principles of efficient image matching with the purpose of determining frame to frame

translation and rotation. In the current example – for determining 3D location – it is necessary to determine the translation of at least three points associated approximately with at least three of the four corners of each 2D image frame. In practice, multiple additional redundant estimates are made to improve accuracy and reduce the impact of isolated bad estimates. The Minimum Sum of Absolute Differences (MSAD) algorithm is preferred. This algorithm produces similar results to those attainable using simple multiplication intensive cross-correlation but with substantially less computational expense¹³. The MSAD algorithm, as its name suggests, uses only comparison, summing, modulus and subtraction operations. Thus computationally expensive multiplies are avoided. Additionally, since it is possible to avoid integer overflow in the Absolute Difference kernel portion of the algorithm it is well suited to efficient implementation in integer oriented (fixed point) IC processors. As an example, Weng's work was implemented using a Texas Instruments TMS320C80 – Multimedia Video Processor (MVP)¹¹. Similarly, Intel Single Instruction Multiple Data (SIMD) commands are capable of using the Pentium 64 bit register to operate on eight 8 bit wide words simultaneously. This type of efficiency would be impossible if the 8 bit operations overflowed. These MSAD operations can be used to determine the best matching motion estimate at multiple locations on the *Tracking* array images. Significant further improvement can be obtained by estimating the motion to subpixel resolution using a quadratic curve, or hyperbolic curve, fitted to the SAD values at the neighboring locations to the pixel offset associated with the actual Minimum SAD.

3. METHODS

The modified array was based on a modified Acuson 8L5 transducer with 128 elements on a 0.3 mm pitch. 96 Elements were dedicated for the *Imaging* array and 16 elements were dedicated to each *Tracking* array. This array was used in conjunction with an Acuson Sequoia® 512 with very minor software modification. For the sake of simplicity, it was preferable to limit the total element count to 128. However, it is recognized that it would have been preferable to make both the *Imaging* and *Tracking* arrays wider. Fig. 3 illustrates the distribution of elements between the various arrays. Notice that the elements, from the system's viewpoint, are still numbered sequentially from 0 to 127. The significance of this is that the system can address the acoustic lines sequentially along the array in a conventional manner. The system may be blind to the fact that the array has effectively been chopped into three segments and reattached with portions of the array perpendicular to each other. Actually, the *Tracking* arrays were specially fabricated but the foregoing sentence illustrates the concept most simply. The only modification that is strictly required is that the aperture functions associated with the sequence of acoustic lines are artificially cropped so that the apertures *never* cross the boundaries between the *Tracking* and *Imaging* arrays. This concept is illustrated graphically in Fig. 3. It is worth emphasizing that by using this elegant control of transducer apertures, and by maintaining a common element pitch and center frequency for both *Tracking* and *Imaging* arrays, a single, conventional beamformer, is all that is required for operating what are, in reality, three discrete transducer arrays. When the modified transducer is scanned a modified 2D image is obtained. An example image is illustrated in Fig. 4. Modest discontinuities in the image are evident between the image portions associated with each *Tracking* array image and the central *Imaging* array image. Having acquired a beamformed set of acoustic line data, it is then necessary to process the composite data sets to determine image motion. Image motion estimation is primarily applied to the *Tracking* images but can also be applied to the *Imaging* array data to determine 'in plane' motion. In this way, it is possible to determine full six degree of freedom (6 DOF) motion of the 2D *Imaging* array data sets. It is worth noting that the technique used here only measures relative motion. In other words the initial image has an assumed location within 3D global space. While initially this appears to be a potential problem it does have important advantages over systems using a fixed frame of reference – such as a magnetic positioner based system. Since the new method uses relative motion, any gross tissue motion – for example caused by patient breathing, or some other involuntary motion - is compensated out of the final 3D reconstruction.

Multiple sub-pixel motion estimates are made on each of the *Tracking* images. In a typical example, five estimates spaced along the range direction, are made on each of the two *Tracking* images. Obviously this results in a grossly over constrained estimate of motion – if one assumes that a rigid body motion criterion is appropriate for estimating image frame motion. While tissue is inherently non-rigid, we believe that assuming rigid body motion provides the preferred basis of accurate initial estimation of image frame motion. The best fitting six degree of freedom rigid body motion was determined using a quaternion-based approach¹⁴. This approach used a least squares fitting process. While the current implementation uses a simple, unweighted, least squares approach, further incremental improvement is possible if one uses a weighted least squares approach. For example, weights for each estimate are based on some weight associated with each individual motion estimate. The ratio of Mean SAD to Minimum SAD gives one estimate of the reliability of a particular motion estimate.

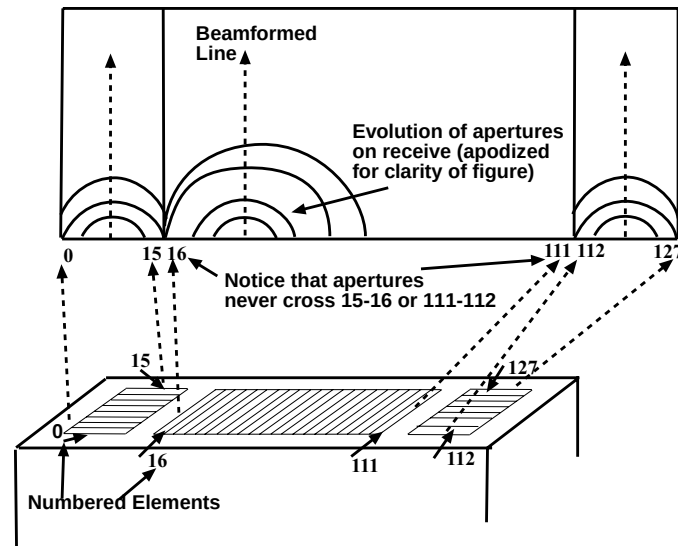


Fig. 3 Illustration of how physical elements of the modified transducer are processed by the system as a single composite 1D array with artificial aperture boundaries between channels 15 & 16 and between 111 & 112. This ‘hard’ programmed boundary between apertures associated with successive lines is the only significant software modification necessary for prototype implementation.

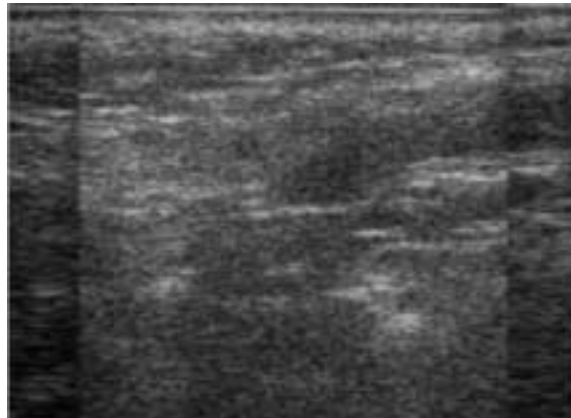


Fig. 4. A composite image from the Acuson Sequoia® system illustrating the central *Imaging* frame and adjacent *Tracking* images. Slight discontinuities are evident towards each end. The end portions are darker because the apertures used for capturing the *Tracking* data acoustic lines are inherently reduced in comparison with those available to the central *Imaging* array.

At this point the advantages of the new transducer based 3D acquisition approach can be summarized.

1. Ergonomic design with unrestricted scanning motions permitted

The transducer is modified in a relatively innocuous manner and can be manipulated freely by the user. This puts the new method at an advantage when compared to mechanical scanning systems – that do not allow unrestricted motion – and magnetic positioner systems – that require an additional coil attachment to be placed on the transducer thereby somewhat impacting ergonomic acceptance of the transducer by the sonographer. Simultaneously, these characteristics are achieved while retaining a high degree of dimensional accuracy. It is believed that the dimensional accuracy achieved with the prototype device ($2\sigma = 5\%$) is competitive with other quantitative positioning methods (e.g., magnetic positioners)

2. No sources of interference

It is known that some of the alternative methods of tracking transducer position are subject to interference from the environment. For example, magnetic positioners are known to be affected by ferrous, and sometimes non-ferrous, metals in their vicinity. They can also be affected by electromagnetic fields. While these interfering effects can in principle be controlled, it is very difficult to categorically ensure that such interference will not occur and it is also nearly impossible to detect when such interference is occurring. (An independent measure of location is necessary to determine if there is interference and associated positioner error.) The new method uses only ultrasound images and hence is largely immune to environmental based interference.

3. Uses relative motion

As discussed above, the new method is based on measures of relative motion and hence is unaffected by gross tissue motion. Examples of gross tissue motion result from breathing or the consequence of an illness that prevents a patient from remaining still.

4. Competitive cost

It is believed that the *Tracking* arrays may be added to a conventional 1D array at a modest cost. It has not been possible to accurately quantify the incremental cost of the *Tracking* arrays but it is worth noting that these arrays may potentially be made to a lower specification than the central *Imaging* array that determines ultimate imaging quality. Additionally, it is worth noticing that this method places all the incremental cost in the transducer. This makes the approach suitable for low cost imaging systems where the only cost penalty for upgrading to 3D capability is in the transducer purchase. This assertion relies on the reasonable assumption that the CPU processing power within a typical state-of-the-art low cost system is sufficient to perform the motion tracking, 3D reconstruction and 3D visualization operations.

4. ACCURACY AND IMAGING RESULTS

The accuracy of this new 3D acquisition method was tested by translating the prototype transducer across a Gammex RMI small parts phantom and forming a 3D reconstruction of the wire targets. Using 16 measurements, with an imaging depth of 20 mm, a 2σ % accuracy of 5% was obtained. Based on other measurements made with a similar conventional 1D array, with a selectable aperture size, it is believed that reasonably good accuracy is attainable out to a range corresponding to $F\# = 4$. This result suggests that any future version of the array should have as large an aperture as possible and operate at the highest possible center frequency. It is also worth observing that the accuracy observed here is consistent with the results obtained by Weng¹¹ when Extended Field of View images were being generated using image matching algorithms broadly similar to those used here.

Fig. 5 illustrates a surface rendered 3D reconstruction of the carotid bifurcation of a healthy volunteer. This scan was made at Stanford Hospital, CA. Once the 3D volumetric grid data set was acquired, the Color Doppler data was isolated and processed using VitalImages VoxelView®. A clinically useful and dimensionally quantitative result has been obtained. Recently, our collaborators at Stanford have been investigating the application of this new technology for the clinically important application of low-cost, rapid screening for carotid artery stenosis¹⁵. Clinically significant stenosis occurs in approximately 5% of the over 60 years old population¹⁶.



Fig. 5 Surface rendered 3D view of a normal human carotid artery bifurcation. The view is made somewhat complex by the presence of the jugular vein near the carotid artery. The jugular vein lies near the carotid artery below the bifurcation. This image illustrates a section of carotid artery approximately 50 mm long. This image was provided by Dr. Sandy Napel, Radiology Department, Stanford University, CA.

5. IMPROVED DESIGN VARIANTS

One potential limitation of the method proposed here is that it depends on tissue contact at the ends of the transducer array. This *potential* problem is presented graphically in Fig. 6. *However, it should be stressed that in recent clinical evaluations this limitation has been overcome easily by using careful scanning technique – even when scanning convex anatomy such as the neck.* It is possible to place the *Tracking* transducer arrays ‘in board’ (illustrated in Fig. 7 – middle view) but this results in a far wider array with a significant amount of wasted transducer ‘footprint’. Ideally, common transducer surface area is used for both *Imaging* and *Tracking*. This may be achieved using a crossed-electrode transducer array design – illustrated diagrammatically in Fig. 8. This crossed-electrode type of design is moderately well proven and was used by Kobayashi et al ¹⁷ in their bi-plane TEE transducer several years ago. In the crossed-electrode design, parallel strip electrodes run in one direction on one side of the piezoelectric substrate and in the perpendicular direction on the opposite of the substrate. Typically, it is necessary to use a 1:3 Piezo-ceramic:Epoxy composite substrate in order to minimize strong mechanical coupling (cross-coupling) and associated corrupted acoustic field patterns. In any case, using a 1:3 composite substrate, with the elements defined by electrode alone (i.e., not by mechanical separation) results in some degree of residual cross-coupling. However, it is fortuitous that the ‘linear array’ designs used here (focused but not steered) are far less susceptible to imaging degradation due to cross-coupling effects than ‘phased array’ designs commonly used in Cardiology which are required to both steer and focus.

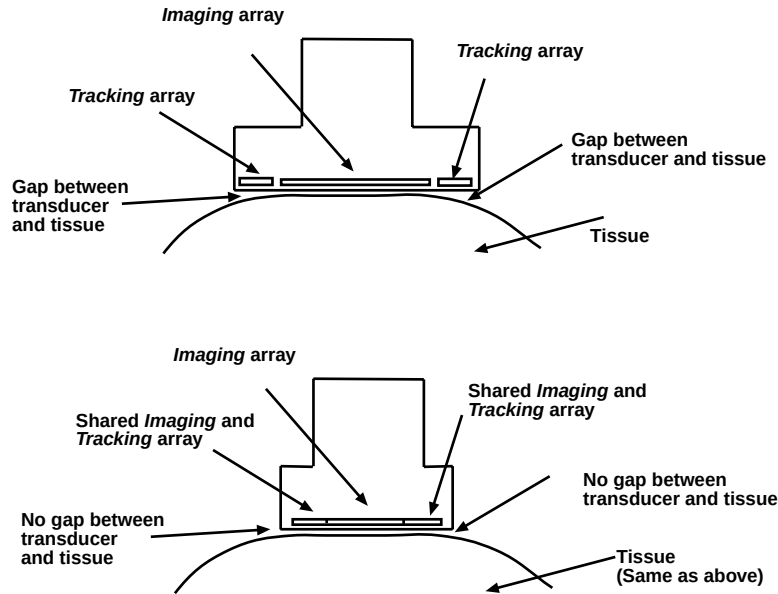


Fig. 6 Illustration indicating why it is preferable to mount the *Tracking* arrays superimposed with the ends of the *Imaging* arrays. In the top view, where the *Tracking* arrays are ‘out board’ of the *Imaging* array, there is a *potential risk* of poor tissue contact when scanning a convex tissue surface. In the bottom view, the *Tracking* arrays are superimposed on the *Imaging* arrays and total transducer ‘footprint’ is reduced thereby improving tissue contact. It should be observed that the *Imaging* array azimuthal dimension is identical in each of the above examples.

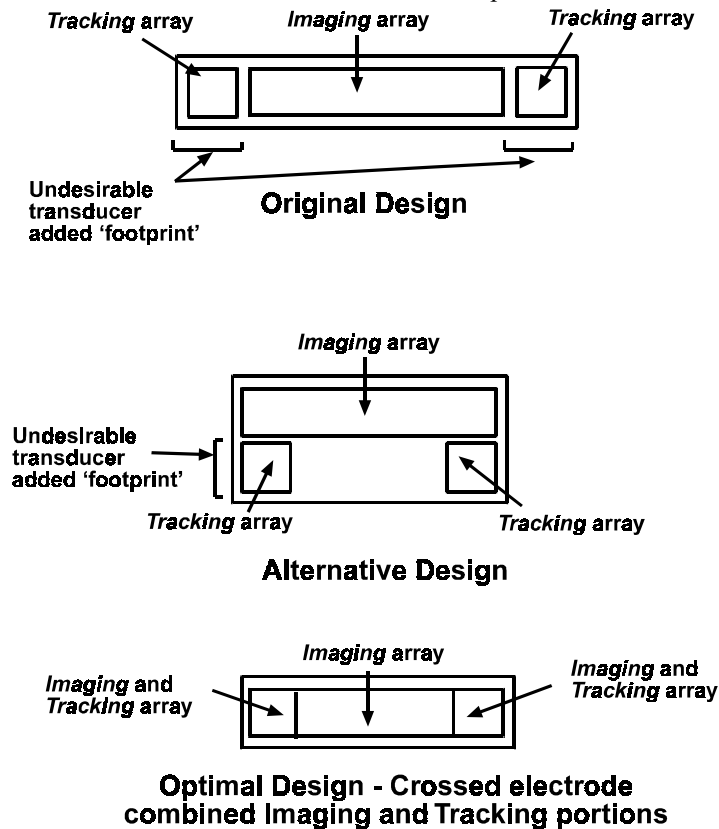


Fig. 7 Illustration of original *Tracking* array configuration (plan view – looking down in array surface), one non-preferred alternative design and an optimal design that saves transducer ‘footprint’ but requires more complex crossed-electrode transducer design

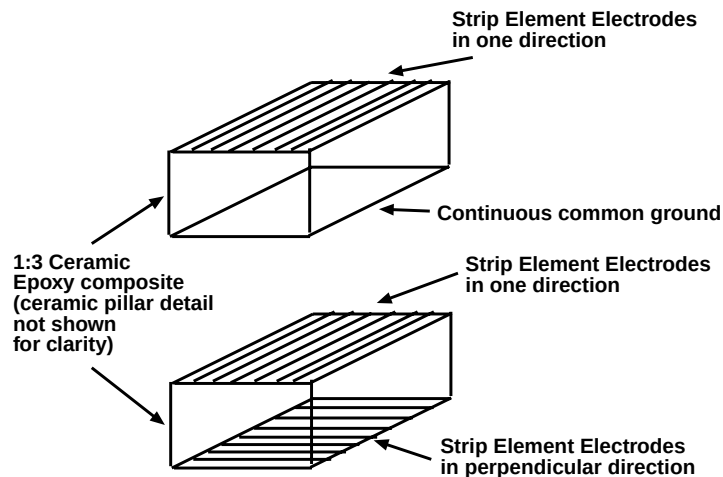


Fig. 8 (Top) Illustration of conventional 1D transducer array with defined electrodes on one surface and a common ground electrode on the opposite side. (Bottom) In the crossed-electrode design, sets of electrodes – running in perpendicular directions – are attached to each of the top and bottom surfaces. The roles of the electrodes are swapped between live signal and ground in order to activate the transducer in each of the two perpendicular directions. 1:3 Piezoelectric ceramic: Epoxy composite (comprising ceramic posts embedded in an Epoxy polymer matrix) are typically used in this configuration.

6. CONCLUSIONS

An improved approach to 3D ultrasound acquisition based on a modified transducer array has been presented. This approach has a number of advantages over conventional approaches by incorporating all of the following characteristics – 1) Quantitative, 2) Ergonomically satisfactory, 3) Uses relative motion, 4) Immune to environmental interference and 5) Low Cost. Preliminary evaluations suggest that a 2σ accuracy of 5% is achievable. Analysis indicates that optimal performance is achievable when the *Tracking* array is as wide as possible and uses the highest practical operating frequency. One issue that has been identified relates to the desirability of using a crossed-electrode array so that *Tracking* and *Imaging* functions can be performed using a coincident substrate and so that added transducer ‘footprint’ is avoided. Since the method uses measures of relative motion in the determination of transducer displacement, it is particularly well suited to mobile tissue regions such as the breast. Additionally, the technique has recently been applied successfully in a prototype investigation of the appropriateness of the technology as the basis for a screening technique for carotid artery stenosis¹⁵. It is also envisaged that the technique can be extended to enable 3D prostate imaging. In this case, the *Tracking* arrays would be integrally mounted within the casing of a modified Endo-Rectal ultrasound probe.

7. ACKNOWLEDGMENTS

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