Real-time cylindrical curvilinear 3-D ultrasound imaging

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In patients who are obese or exhibit signs of pulmonary disease, standard transthoracic scanning may yield poor quality cardiac images. For these conditions, two-dimensional transesophageal echocardiography (TEE) is established as an essential diagnostic tool. Current techniques in transesophageal scanning, though, are limited by incomplete visualization of cardiac structures in close proximity to the transducer. Thus, we propose a 2D curvilinear array for 3D transesophageal echocardiography in order to widen the field of view and increase visualization close to the transducer face. In this project, a 440 channel 5 MHz two-dimensional array with a 12.6 mm aperture diameter on a flexible interconnect circuit has been molded to a 4 mm radius of curvature. A 75% element yield was achieved during fabrication and an average -6dB bandwidth of 30% was observed in pulse-echo tests. Using this transducer in conjunction with modifications to the beam former delay software and scan converter display software of the our 3D scanner, we obtained cylindrical real-time curvilinear volumetric scans of tissue phantoms, including a field of view of greater than 120° in the curved, azimuth direction and 65° phased array sector scans in the elevation direction. These images were achieved using a stepped subaperture across the cylindrical curvilinear direction of the transducer face and phased array sector scanning in the noncurved plane. In addition, real-time volume rendered images of a tissue mimicking phantom with holes ranging from 1 cm to less than 4 mm have been obtained. 3D color flow Doppler results have also been acquired. This configuration can theoretically achieve volumes displaying 180° by 120°. The transducer is also capable of obtaining images through a curvilinear stepped subaperture in azimuth in conjunction with a rectilinear stepped subaperture in elevation, further increasing the field of view close to the transducer face. Future work includes development of an array for adapting these modifications to a 6 mm diameter endoscope probe.

KEY WORDS: 2-D array; curvilinear; cylindrical; endoscopic; real-time; transesophageal; volumetric.

I. INTRODUCTION

A key element in adapting medical ultrasound imaging (US) to a specific application is the design of the transducer. For external 2D ultrasound, the three standard transducer types are sectored phased arrays, linear sequential arrays and curvilinear arrays. The first type employs all of its elements simultaneously to focus and steer its acoustic beam through the target location, producing a pie-shaped scan (Fig. 1A). Because of its scan shape and wide field of view away from the transducer face, these arrays are ideal for cardiac imaging, avoiding the interference posed by the ribs. For obstetric imaging or other applications where it is important to visualize structures close to the skin surface such as breast and vascular imaging, linear sequential arrays are preferred. By using small subsets of elements that are sequentially stepped across the aperture, these transducers are capable of scanning a rectangular area that includes proximal structures such as breast lesions (Fig. 1B). Similar to linear arrays, curvilinear arrays also employ sequential scanning to view regions close to the transducer face. However, due to their curvature, they are able to achieve a larger field of view (Fig. 1C).
In recent decades, endoscopic ultrasound has seen increased use in invasive applications for diagnostic imaging as well as surgical guidance. In addition to its cost effectiveness and lack of ionizing radiation, ultrasound has proven to be advantageous for invasive procedures because of its close proximity to the tissues of interest, enabling the use of higher frequency interrogation, thus increasing imaging resolution. For example, in echocardiography, imaging from within the esophagus provides clearer views of the heart without interference from the lungs and the ribs. Views acquired with transesophageal echocardiography (TEE) are particularly useful in patients who exhibit poor image quality in transthoracic scans due to obesity or pulmonary disease. TEE has already been established as an essential tool for diagnosis of valvular disease and wall motion abnormalities, as well as evaluation of cardiac function during open heart surgery. In addition to its cardiac applications, endoscopic ultrasound has been employed for locating cancerous lesions throughout the GI system, including the pancreas, liver and esophageal lining. Ultrasound has also seen use in transrectal imaging for diagnosing prostate cancer, rectal cancer and other abnormalities of the lower GI system. For vaginal imaging, applications include diagnosis of pelvic and cervical cancer as well as obstetric uses such as follicular monitoring in fertility disorders, differentiation of the normal and abnormal 1st trimester and diagnosis of ectopic pregnancy.

Typical single plane arrays for TEE as well as other endoscopic applications consist of 64 or 128 channels at frequencies of 5 to 10 MHz, producing 2D sector scans with a field of view up to 90°. These arrays for transesophageal imaging are usually incorporated into a bidirectional flexible endoscope shaft of 60 to 110 cm in length, while for rectal and vaginal
In order to gain more flexibility while achieving different views within the patient independent of the probe’s mechanical maneuverability, multiplane and panoramic transducers have been developed. Commercially available multiplane transducers are capable of a wider field of view as well as a display of multiple planes in the patient by mechanically rotating a standard side-looking linear array. Although less probe manipulation is required in comparison with biplane transducers, the angles of rotation are in predetermined increments. Thus, multiplane probes are still incapable of true volumetric data, and the problem of decreased visualization close to the transducer face with phased array scanning is still an issue. Panoramic 2D TEE probes using motor driven annular arrays have been described as being capable of sector scans with a field of view as large as 270°. These have shown promise with regard to better visualization of the entire left atrium, the aortic arch, and better recognition of congenital abnormalities. However, like multiplane transducers, these are of limited effectiveness due to their reliance on mechanically driven arrays.

For ultrasound to be most effective in invasive procedures, the system must be capable of real-time time images with minimal probe manipulation while still providing as much information of the target volume as possible. Mechanical 3D scanners that translate or oscillate a linear or curvilinear array for external ultrasound imaging have been reviewed by Fenster and Downey and have been described as being less operator dependent while providing improved accuracy of volume data and orientations not available with 2D imaging. For 3D TEE imaging, ECG gating and respiration gating have been employed to acquire volumetric data that can then be reconstructed into a 3D image off-line. Another off-line reconstruction method for 3D TEE employs a panoramic transducer that is mechanically rotated about the probe’s axis in conjunction with ECG gating to provide data for a toroidal image. These 3D systems, however, are highly dependent on both mechanical translation and precise gating in order to obtain accurate data. In addition to normal respiration and heart motion, the presence of disease can cause distortion and measurement inaccuracies. Finally, off-line reconstruction of 3D images limits the clinical applicability of these transducers for biopsy and other surgical guidance procedures.

Real-time volumetric ultrasound was developed by von Ramm and Smith specifically for cardiac imaging. By using a two-dimensional array, the system was capable of employing sector phased array scanning in both azimuth and elevation in order to acquire a pyramidal volume of data (Fig. 1D) without relying on mechanical translation or reconstruction. The Model 1 system (Volumetrics Medical Imaging, Durham, NC) is capable of using up to 512 transmitters and 256 receivers with 16:1 receive mode parallel processing. This enables the scanner to insonify a pyramidal volume equivalent to 64 sector scans of 64 lines each stacked in the elevation dimension for almost 4,100 total scan lines of data at up to 60 volumes per second. Having acquired an entire 65° pyramidal volume of data, the scanner is then able to display in real-time up to five single planes at any desired angle depth and origin, as well as a display of real time volume-rendered data. As an illustration of the 3D scanner’s capabilities, figure 2 shows the azimuth B-scan and a C-scan of a tissue phantom (RMI Model 408; Middleton, WI) with 4 mm diameter spherical lesions of 35 dB contrast at a depth of 6 cm. The images were acquired with Model 1, using a 5-MHz, 2D array with 440 active piezo-
electric elements in a periodic geometry.\textsuperscript{14} The C-scan in figure 2B corresponds to the plane indicated by the arrows in the azimuth B-scan of figure 2A. The images in figure 2 will serve as a control for the current work.

The motivation of this research is to extend the advances of real-time volumetric scanning with 2D arrays to endoscopic ultrasound, particularly transesophageal echocardiography, by using novel array geometries and scanning methods. 2D phased arrays have been implemented in catheters with a diameter as small as 2 mm and as many as 112 active channels for 3D intracardiac imaging.\textsuperscript{15} In addition, Yen and Smith have developed a 2D array capable of rectilinear scanning to acquire a rectilinear volume (Fig. 1E).\textsuperscript{16, 17} Much like the linear sequential array, it employed small subsets of elements to direct transmit beams directly perpendicular to the transducer face. While this format does enable visualization of structures close to the array, its field of view is still a limiting factor when applied to TEE and other invasive applications. A spherical curvilinear adaptation of this rectilinear array was also developed by Yen and Smith\textsuperscript{17, 18} by extending 1D curvilinear arrays (Fig. 1C) to two dimensions (Fig. 1F). However, this prototype had a large radius of curvature of 4 cm, providing only slight gains in FOV.

In this paper, we describe a prototype cylindrical curvilinear scanner for real-time volumetric ultrasound. A cylindrical array that utilizes curvilinear sequential scanning across its curved surface and either phased array scanning or rectilinear scanning in elevation (Figs. 1G and 1H, respectively) can take advantage of the natural curvature of an endoscope probe while simultaneously providing an overall increased field of view, especially close to the transducer. Using this prototype array, the first real-time cylindrical curvilinear volumetric scans of tissue-mimicking phantoms are shown along with volume rendered views. 3D color flow Doppler images of a string phantom are also presented. This form of transducer array may increase the effectiveness of TEE as well as transrectal, transvaginal and other endoscope-based ultrasound procedures.
II. MATERIALS AND METHODS

Transducer fabrication

A 5 MHz prototype cylindrical array was developed from the geometry of the 12.6 mm diameter transthoracic array of figure 2 that has been previously described.\textsuperscript{14,19} The polyimide multilayer flexible circuit (MLF), fabricated as detailed by Fiering et al.,\textsuperscript{20} accommodates 440 active elements at a pitch of 0.350 mm in a periodic sparse array design shown in figure 3A. Of the active channels, 256 are shared transmit/receive channels, while the remaining 184 are solely for transmit. Curved to a 4 mm radius of curvature, this array will look as shown in figure 3B. A 12.6 mm x 12.6 mm x 0.340 mm wafer of fine grain PZT-5H (TRS Ceramics, State College, PA) was attached to the flex circuit with a layer of conductive silver epoxy. Once bonded, a 25 μm dicing blade was used to dice the PZT at a pitch of 0.350 mm in both directions. Once the major cuts were complete, additional cuts were made through the center of each element in both azimuth and elevation in order to subdivide each element into four equal parts. With each subelement 0.175 mm on a side, a width-to-thickness ratio of approximately 0.5 was achieved to suppress lateral mode coupling. The entire MLF is shown in figure 4A with the array in the center already covered by the diced piezoelectric elements. The solder pads located at either end of the flex are used for connection to the transducer handle.

Once element dicing was complete, the array was molded by, and applied to, a lossy epoxy backing. The backing was designed to have a 28 mm x 24 mm base that would accommodate connections to the transducer handle. From the base, the backing narrows to a half-cylinder with a 4.4 mm radius of curvature, as shown in figures 4B and 4C. This backing geometry would ensure that all elements in the array would be on the curved surface of the transducer. A curved well with a radius of 4.9mm was used to hold the array in place during bonding, ensuring good adhesion across the curved surface of the backing. The completed array, covered with a grounded foil layer of 7 μm thickness, can be seen in figure 4B. In order to ease fabrication, this prototype array does not include a matching layer. Impulse response and power spectrum results were acquired using pitch-catch experiments in a water tank. These results were then compared to KLM simulation results performed with PiezoCAD.
Simulations

In addition to using the KLM model to predict the electromechanical performance of the transducer, simulations using Field II\textsuperscript{21} were used to predict the beam characteristics of the prototype array. The original flat transthoracic array design (Fig. 4A) was first simulated at 5 MHz with a 0.350 mm element pitch. Views for the on-axis beamplot for \((x, y, z) = (0, 0, 50 \text{ mm})\) are shown in figures 5A and 5B. The \(-6\text{dB}\) beam width when focused at 5 cm is 2 degrees, while grating lobes reach a peak relative amplitude of \(-48\text{dB}\) at approximately 25° off-axis in both azimuth and elevation. These beamplots correspond to the images acquired in figure 2.

The Volumetrics Model 1 scanner was originally intended for phased array scanning in azimuth and elevation to acquire a pyramidal volume of data. Thus, the beam former delay software had to be modified to allow for curvilinear scanning across the curved surface of the prototype transducer. The delay software was modified to activate a 5.95 mm diameter subaperture for each transmit line with unfocused elements to accommodate receive mode parallel processing. In receive, the active elements are determined by an expanding F/1 aperture. All elements not within the subaperture parameters are inactive during a line acquisition. Thus, as seen in figure 3, as transmit lines move further away from the center of the array, the subapertures become more sparse with a lower element density. This will adversely affect the off-axis sensitivity and resolution of our transducer.

These modifications were simulated using Field II to predict beam characteristics as well as compare with the results from the flat transthoracic simulation. For an on-axis simulation, the center 5.95 mm diameter subaperture was used to acquire beam characteristics of the central image line. This subaperture, containing 233 transmit elements and 140 total receivers, is shown in black in figure 3B. Focusing this array at 5 cm away from its front face, the beam plots shown in figures 5C and 5D were acquired. In the curved azimuth direction (Fig. 5C), the \(-6\text{dB}\) beamwidth is 4° with significant \(-28\text{dB}\) grating lobes appearing at 42° off-axis. In the flat elevation direction (Fig. 5D), the \(-6\text{dB}\) beamwidth is slightly narrower at 3 degrees with no major grating lobes present off-axis from the main beam.

FIG. 4 (A) Multilayer flexible interconnect circuit with bonded and diced PZT, (B) curved prototype cylindrical array with grounded foil cover and epoxy backing and (C) prototype cylindrical array attached to Model 1 handle.
In addition to the central image line, an off-axis simulation was performed for a subaperture centered at 45° from the curvilinear axis. This simulation used the same size transmit and receive subapertures as previously employed for the center line beamplots; however, due to the sparse sampling of elements away from the center of the array, the results (Figs. 5E and 5F) suffer from larger grating lobes. In the azimuth direction, there is no discernible widening of the main beam, but the grating lobes are larger due to the reduction down to 140 transmit elements and 122 receive elements.

Display

In addition to the modifications made to the beam former delay software, the display software was also altered in order to display a standard 65° phased array scan in elevation and a 120° curvilinear B-scan in azimuth. Displays for C-scan slices were altered in order to accommodate the increased information in the azimuth direction.

III. RESULTS

Transducer testing

Pulse excitation KLM simulations shown in figures 6A and 6B predict an element response with a center frequency of 4MHz and a 6dB fractional bandwidth of 30.1%. Using a water tank, individual element measurements were made to determine experimental impulse response and power spectrum. These pitch-catch measurements were recorded by pulsing each element in turn with a pulser/receiver (Panametrics Model 5073PR) and monitoring the returned echo with an oscilloscope (Tektronic TDS 744A, Wilsonville, OR) and spectrum analyzer (Hewlett-Packard HP358A, Everett, WA). Figure 6C and 6D show a typical pulse
and spectrum recorded using this method. The displayed pulse had a center frequency of 3.49 MHz and a -6dB fractional bandwidth of 28.1%. Using this pitch-catch method to determine element operation, it was determined that 75% of the 440 possible channels were working. Most of the nonfunctioning elements were located at the extremes of the array. A 10% sampling of the operational elements yielded an average bandwidth of 28.9%.

In addition to pitch-catch experiments, a point target system response was obtained using the Model 1 Volumetrics scanner in order to approximate the beam characteristics of the prototype array. The tip of a 1 mm diameter rod was used as the point target. Placing the rod in a stable position so that its tip was centered at a depth of 4 cm away from the transducer face, a C-scan image of the tip was recorded with Model 1. A plot of the echo amplitude in the image for the curved azimuth direction can be seen as a dashed line in figure 7A, while the experimental beam in elevation is shown similarly in figure 7B. These results are compared to new simulated beamplots produced with Field II. This new set of simulated data was produced by approximating the experimental conditions used for the point target acquisition. The Model I clock frequency of 40 MHz was applied, as well as a sound speed of 1,470 m/s since the experimental results were acquired in a water tank. Also, nonfunctioning elements were removed from the simulation array, and a random apodization was applied in order to approximate element nonuniformity. The results from this Field II simulation are plotted against those acquired from the Model 1 scanner as solid lines in figures 7A,B. It can be seen from comparison that, although there is reasonable correlation between profile characteristics, the experimental beams are noticeably wider in both azimuth and elevation and the clutter floor is significantly higher. The wide experimental beams can be accounted for by the use of parallel processing with the Volumetrics scanner, as well as undersampling in the azimuth direction. In addition, the use of a relatively large rod for a point target response experiment further widens the result. The high clutter floor indicates that the electronic noise from the volumetric scanner outweighs the clutter produced by the transducer’s ultrasound beam.
Images

Real-time images of various phantoms using the prototype cylindrical curvilinear array are shown in figures 8 to 11. These images were acquired with the Model 1 Volumetrics scanner using the modified display program to view an expanded 120° curvilinear B-scan in azimuth and a standard 65° sectored phased array B-scan in elevation. Horizontal arrows in the azimuth scan indicate the plane at which the displayed C-scan is acquired, while vertical arrows show the location of the elevation B-scan.

Figures 8 and 9 show images of a tissue phantom (Univ. of Wisconsin) using a scan depth of 6 cm. The echogenic ‘tumor’ in figure 8 has a diameter of 2 cm and is located at a depth of 4 cm. Its inherent contrast is +20dB. In figures 8A-8C, the hyperechoic inclusion is clearly visible at the center of the scan in both azimuth and elevation. In the sectored elevation B-scan, speckle is notably smaller than in the curvilinear azimuth B-scan. In figures 8D-8F, the inclusion is still detectable when located 45° off-axis. The signal strength in the azimuth B-scan is lower since the subapertures off-axis have a lower element density than those located closer to the center of the array. The loss in image quality at this angle is as predicted from the Field II beamplots acquired through simulations (Figs. 5E-5F). Figure 9 shows similar scans of a 2 cm diameter hypoechoic inclusion at an equal depth and with a contrast of −20dB. Although not as clear as the target in figure 8, the ‘cyst’ is still distinguishable both on-axis (Figs. 9A-9C) and 30° off-axis (Figs. 9D-9F). The grating lobes that were noted in Field II simulations contribute significant background clutter to the images in the curved azimuth scan, reducing contrast resolution. Due to the contributions of these grating lobes in combination with the sparse off-axis subapertures, the hypoechoic inclusion was not viewable as far off-axis as was possible with its hyperechoic counterpart in figure 8.

Figure 10 shows the color flow Doppler capabilities of the prototype cylindrical curvilinear array. A string rotated by a motor set at a rate of 30 revolutions/min was imaged in a water tank using a 12 cm depth scan. The transducer was held stationary in the tank so that the string moved in the azimuth (curved) direction of the array. The sectored phased array B-scan (Fig. 10A) was set at 30° off-axis while the C-scan (Fig. 10C) was oriented to display the length of the passing string. Both the curvilinear azimuth B-scan (Fig. 10B) and the aforementioned C-scan show that the transducer, when used with color flow Doppler, can successfully differentiate between the portions of the string perceived as approaching the transducer (red) and the portions moving away from it (blue). Examination of each section

FIG. 7 Simulated and experimental beamplots of the central image line. Simulated results (solid line) were acquired using Field II. Experimental results (dashed line) were acquired by imaging a 1mm diameter hydrophone needle in a water tank.
of the string shows a gradual increase in the magnitude of string velocity when moving away from the center of either the azimuth B-scan or the C-scan in figure 10C.

Images from a sponge phantom are shown in figure 11. Rather than displaying a standard C-scan slice as in the previous images, figure 11C shows a real time rendered display taken from the data between the planes indicated by the arrows in the curvilinear azimuth B-scan (Fig. 11B). The sponge contained 20 mm and 5 mm diameter holes, the larger of which is visible in cross-section in both B-scan views. The 5 mm hole is not distinguishable in the long-axis view of the azimuth B-scan, but is clearly seen in the volume rendered scan with its advantage of spatial integration.

IV. SUMMARY AND DISCUSSION

To our knowledge, we have developed the first real-time cylindrical curvilinear volumetric scanning system. Adapting this type of array to a suitable size could prove useful for
transesophageal echocardiography and other endoscope-based ultrasound procedures by providing a wide field of view in conjunction with good visualization close to the transducer face. A 5 MHz 440 channel prototype array has been fabricated from a previous trans-thoracic design. This array has been used with delay and display software modifications to obtain cylindrical curvilinear scans that utilize curvilinear scanning in azimuth and phased array scanning in elevation. If an element yield higher than 75% were achieved, this prototype array could conceivably achieve scans greater than 120° in azimuth. In addition, phased array sector scanning in elevation could be increased for a total possible field of view of 180° in azimuth by 120° in elevation. Visualization close to the transducer face could be further increased by implementing rectilinear scanning in elevation.22

Images of cysts in tissue-mimicking phantoms have confirmed the detrimental effects of grating lobes observed in Field II simulations and point target experiments. These contribute significant background clutter and, in conjunction with poor element yield at the extremes of the array, degrade off-axis resolution. Improving element yield as well as

![Diagram](image-url)
Improving fabrication techniques could partially remedy some of these problems. However, the transthoracic array design was not intended for use with cylindrical curvilinear scanning; therefore, a new sparse array design tailored for this new ultrasound technique must be developed.

**FIG. 10** Color flow Doppler results using the prototype cylindrical array. (A) The elevation B-scan shows the cross section of the moving string at the plane indicated by the arrows in the curvilinear azimuth B-scan (B). The azimuth scan shows that the cylindrical scan successfully differentiates between the portions of the string that are traveling towards (red) and away from (blue) the transducer. This is also shown in the C-scan (C).

**FIG. 11** B-scans and a real-time volume rendered view of a sponge phantom with 20 mm and 5 mm diameter holes. The arrows in the sectored elevation scan (A) and the curvilinear azimuth scan (B) indicate the data used for the volume rendering (C).
Due to improvements in cable technology, we believe we can successfully develop a 6 mm diameter TEE probe with as many as 504 active channels in a cylindrical curvilinear array that takes advantage of the endoscope’s natural curvature. Current work involves design of a suitable array geometry with a higher subaperture element density. In addition, unlike the prototype array, the subapertures used for each scan line would be uniform in element density and geometry. We believe that such improvements will improve pulse-echo sensitivity, reduce grating lobes and increase off-axis resolution. Preliminary Field II simulations of currently in-progress TEE array designs have shown that the grating lobes can be lowered to at least $-32\text{dB}$. Current design and simulation studies are aimed at further reducing these. It is our hope that the completed TEE cylindrical array will be capable of a 180° field of view in azimuth as well as rectilinear scanning in elevation to maximize visualization for endoscopic procedures.

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