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Operational verification of a 40-MHz annular array transducer

Jeffrey A. Ketterling and Sarayu Ramachandran

J. A. Ketterling and Sarayu Ramachandran are with Riverside Research Institute, New York, NY (email:ketterling@rrinyc.org).

Orlando Aristizábal

O. Aristizábal is with Skirball Institute of Biomolecular Medicine and New York University School of Medicine, New York, NY.

Abstract

An experimental system to take advantage of the imaging capabilities of a 5-ring polyvinylidene fluoride (PVDF) based annular array is presented. The array has a 6 mm total aperture and a 12 mm geometric focus. The experimental system is designed to pulse a single element of the array and then digitize the received data of all array channels simultaneously. All transmit/receive pairs are digitized and then the data are post-processed with a synthetic focusing technique to achieve an enhanced depth of field (DOF). The performance of the array is experimentally tested with a wire phantom consisting of 25- μ m diameter wires diagonally spaced at 1 mm by 1 mm intervals. The phantom permitted the efficacy of the synthetic focusing algorithm to be tested and was also used for two-way beam characterization. Experimental results are compared to a spatial impulse response method beam simulation. After synthetic focusing, the two-way echo amplitude was enhanced over the range of 8 to 19 mm and the 6-dB DOF spanned from 9 to 15 mm. For a wire at a fixed axial depth, the relative time delays between transmit/receive ring pairs agreed with theoretical predictions to within ± 2 ns. To further test the system, B-mode images of an excised bovine eye are rendered.

I. Introduction

High-frequency ultrasound (HFU) imaging transducers (>20 MHz) provide a means to obtain high resolution images, albeit with moderate penetration depths in tissue. As these transducers become more widely available, they are being applied to numerous applications including ophthalmic imaging [1], [2], small animal imaging [3]-[5], dermatological imaging [6], and *in vivo* imaging via intravascular ultrasound (IVUS) probes [7].

The high-frequency transducers typically used in practice are spherically curved single element transducers [8]-[10]. These transducers have a lateral resolution on the order of a wavelength ($\lambda = 37 \mu\text{m}$ at 40 MHz), but they also have a fairly limited depth of field (e.g., 1 mm). Therefore, B-mode images generated with these transducers will provide the best resolution in the imaging plane near the geometric focus while features outside of this range will be out of focus (See Fig. 5 in Ref. [11]).

One means to improve the depth of field (DOF) is to use a multi-element array transducer. Arrays with the ability to focus in both the axial and lateral directions typically have many elements (>16) with element spacings on the order of $\lambda/2$ [12], [13]. These devices do not need to be mechanically translated to generate a B-mode image. However, they are still under development because of the complexity of their assembly and instrumentation. Arrays with fewer elements, such as annular arrays, permit an increased DOF but need to be translated to generate a B-mode image [14]-[17]. Because the size of individual elements of the annular

array are on the order of conventional HFU single-element transducers, standard fabrication and instrumentation techniques may be employed to construct them.

In an earlier article [18], we established the feasibility of fabricating a polyvinylidene fluoride (PVDF) based annular array. The array consisted of five elements with the array pattern etched onto a copper-clad polyimide backing film. The operation of each individual element was tested for complex impedance, pulse echo response, electrical cross talk, and insertion loss before and after impedance matching. The individual array elements were found to perform on par with typical single element PVDF transducers [8].

In this article, we expand upon the earlier work by developing a system to generate B-mode images using the annular array. An experimental system consisting of a crosspoint switch, digitizers, a pulser/receiver, and a XYZ axis linear scanning component is described. The system permits a single element to be pulsed and the received signals from all five elements to be digitized simultaneously. Data collected from all 25 transmit/receive pairs are then post-processed with a synthetic focusing technique to increase the DOF. To test the experimental system, we utilized a phantom with 25- μm diameter wires diagonally spaced at 1-mm by 1-mm increments. The data from the wire phantom are processed with the synthetic focusing algorithm to establish an improved DOF. A theoretical acoustic beam model based on the spatial impulse response method is employed to validate the operation of the array and the synthetic focusing algorithm. Finally, to demonstrate the improved DOF of the array and its utility to image inhomogeneous, attenuating tissue, a B-mode image of an excised bovine eye was rendered.

II. Material and Methods

A. Transducer

The spherically-curved annular array utilized in these studies consisted of five elements, had a total aperture of 6 mm, a geometric focus of 12 mm, and a nominal center frequency of 40 MHz. The annuli were designed to have equal areas and there were 100- μm kerf spacings between elements. The active acoustic component of the array was a 9- μm PVDF membrane with a single conductive side. The actual array pattern and trace lines to each array element were formed on a copper clad polyimide film (flex circuit) using standard printed circuit board (PCB) techniques (Fig. 1). To assemble the transducer, the PVDF was bonded to the polyimide film with epoxy, pressed into a Teflon tube with a ball bearing, and then the tube was back-filled with epoxy. For a more detailed description of the array fabrication along with electrical and acoustic properties of the individual array elements the reader is referred to Ref. [18].

The present version of the array is packaged in an UHF connector (Fig. 2). The UHF connector acts as a rugged protective shield for the array and mechanically stabilizes the flex circuit linking the array to the PCB. To mount the array in the UHF connector, the inner conductive pin was first ground down with a milling machine and a small notch was cut into its side. The epoxy backing plug of the array was then machined on a lathe to a depth appropriate for the UHF connector. Next, the array was placed into the connector and sealed into place with epoxy. Finally, silver epoxy (EE129-4 Epoxy Technology, Billerica, MA) was used to bridge the outer conductive layer of the PVDF to the UHF housing. Because the epoxy plug had a larger diameter than the array aperture (13 vs. 6 mm), spreading the silver epoxy partly onto the PVDF did not affect the acoustic performance of the device.

B. Theory and data processing

1) *Spatial impulse response model*: In order to better characterize the sound field of the annular array as well as to test the synthetic focusing technique, a transmit/receive spatial impulse response (SIR) model was developed [19].

The SIRs of individual annuli were convolved to obtain two-way echoes from a point source permitting both two-way beam profiles and simulated B-mode images to be calculated.

The SIR at a point $\mathbf{r}$ in the sound field of a transducer of surface area $\mathbf{S}$ is

$$\mathbf{h}(\mathbf{r}, t) = \frac{1}{2\pi} \int_{\mathbf{S}} \frac{\delta(t - r'/c)}{r'} d\mathbf{S} \quad (1)$$

where $\delta(t)$ is the Dirac delta function, r' the distance between $\mathbf{r}$ and a surface element $d\mathbf{S}$, and c is the speed of sound [19]. The pressure $p(\mathbf{r}, t)$ at $\mathbf{r}$ is expressed as

$$p(\mathbf{r}, t) = -\rho \frac{\partial v(t)}{\partial t} * \mathbf{h}(\mathbf{r}, t), \quad (2)$$

where $v(t)$ is the transducer surface velocity and ρ is the fluid density. Figure 3 shows the time-domain pulse and corresponding spectrum used for $v(t)$ in the simulations.

The transmit/receive voltage is proportional to the double convolution [20]

$$E_{T,R}(\mathbf{r}, t) \propto -\frac{\partial^2 v(t)}{\partial t^2} * \mathbf{h}_T(\mathbf{r}, t) * \mathbf{h}_R(\mathbf{r}, t) \quad (3)$$

where $\mathbf{h}_T(\mathbf{r}, t)$ is the transmit SIR and $\mathbf{h}_R(\mathbf{r}, t)$ is the receive SIR. This equation may then be used to compute individual radio-frequency (RF) scan lines for every transmit/receive combination and simulate the two-way beam profile of the transducer.

A wire at a single depth was simulated by assuming the wire was a point source and moving it laterally relative to the transducer. RF data for wires over a depth range of 5 to 20 mm were calculated using 100 scan lines spaced 10 μm apart. At each wire location, the RF lines for each of the 25 transmit/receive ring pairs were computed. The RF data were then passed through the synthetic focusing algorithm and the lateral beamwidth was determined. The algorithm shifted the focus of the array to the wire location and reduced the 25 RF lines at each lateral position to a single composite RF line.

To simulate the axial beam profile for a series of focal zones, axial beam profiles were first calculated for each of the focal depths. Then, the axial curve at each focal depth was windowed to extract the axial range centered around the focal depth without any overlap into the adjacent focal zone [Eq. (3)]. Finally, a composite axial beam profile was assembled by stitching together all of the windowed axial segments.

2) *Synthetic focusing technique*: The synthetic focusing technique extends the natural DOF of the transducer by introducing appropriate delays to each transmit/receive RF line to simulate focusing the transducer over a range of focal depths. To focus the array at a depth f , the one way delay t_n of each element is [20]

$$t_n = \frac{a_n^2(1/R - 1/f)}{2c}, \quad (4)$$

where a_n is the average radius of the n^{th} array element and R is the geometric focus. This only accounts for the delay in a single direction, thus the round trip delay is found from the sum of the transmit and receive delays $t^{\text{tot}} = t^{\text{T}} + t^{\text{R}}$.

An increase in the axial DOF of the array is achieved by shifting the focus over q focal depths, or focal zones of width Δf ranging from f_{min} to f_{max} , and then windowing the data at each focal depth to extract just the focal range of interest. This process may be described by the summation

$$s(t) = \sum_{o=1}^q \left(\sum_{l=1}^n \sum_{m=1}^n E_{l,m}(\mathbf{r}, t - t_l^T - t_m^R) \right) w^o(t) \quad (5)$$

where $E_{l,m}$ is the RF line for the transmit/receive ring pair l to m time shifted by $t^{tot} = t_l^T + t_m^R$ and $w^o(t)$ is the rectangular window encompassing only focal zone o . The summation scheme in (5) essentially time shifts the 25 RF data lines at each lateral spatial location, sums them to give a single RF line, and then windows them at the relevant focal depth. If the data is processed with $t^{tot} = 0$ and a single focal zone, the result simulates the case of a single element transducer having the same total aperture and geometric focus as the annular array.

By implementing transmit and receive synthetic focusing after the RF data have been acquired, we have a great deal of flexibility when post-processing the acquired RF data. While a HFU analog beam former would be faster, its implementation is technically challenging and currently would require customized electronics [21].

C. Instrumentation

The experimental system was designed to permit quasi-real time scanning by obtaining digitized transmit/receive data and then post processing them with the digital synthetic focusing algorithm. The system consisted of a XYZ motion system, a crosspoint switch, a pulser/receiver, a digital storage oscilloscope, and a PCI digitizer card (Fig. 4). The basic premise is to obtain data from all 25 transmit/receive ring pairs of the array and then process the data with the synthetic focusing algorithm to obtain an image. The crosspoint switch permits a data set to be collected in five passes rather than 25 if each ring pair were acquired separately. While this is not optimal for real-time imaging applications, it does provide an efficient method of establishing the functionality of the array.

The link between the system components is the bi-directional crosspoint switch (CXL/8X8 Cytec, Penfield, NY). The switch is configured with the five array elements on input lines (In 1-5), the pulser on one output line (Out 5), and a digital oscilloscope (6050A LeCroy, Chestnut Ridge, NY) on four other output lines (Out 1-4). The crosspoint switch permits any input to be connected to any output which means that a single array element may be pulsed and then the received signal from all five elements digitized simultaneously. One ring, the pulsed ring, is digitized (RF IN) via a PCI digitizer (DP110 Acqiris, Monroe, NY) while the other four are digitized with the oscilloscope. The signals sampled by the oscilloscope are first amplified with 45.5 dB of gain (Miteq AU-1313, Hauppauge, NY). While not indicated in the figure, the cable lengths into and out of the pre-amps are of equal length. Because two separate digitizers are used with different electrical components and path lengths, a small relative delay needs to be added to one of the digitizers to properly align the signals in time.

Two pulsing units were used for these experiments. For the wire experiments a pulser/receiver (Panametrics 5900, Waltham, MA) was utilized with the receive signal from the unit going to RF IN after 26 dB of internal gain. For the *in vitro* bovine eye scans, a monocyte pulser was employed (Avtech AVB2-TA-C-CVA, Ottawa, Ontario, Canada). In this configuration, a T-junction was incorporated to connect the pulser, transducer and receive digitizer. A diode expander was placed between the pulser and T-junction to isolate the pulser from the reflected acoustic signal. A diode limiter was placed between the T-junction and a 32.5 dB pre-amp (Miteq AU-1114) and the output signal of the pre-amp was then captured at RF IN.

Images are formed by scanning the transducer across a sample using an automated motion system. To obtain all 25 sets of transmit/receive pair data, five translational passes are necessary; one pass for each pulsed ring. Because the crosspoint switch had a relatively long switching time (100 ms), it was not practical to obtain all the data in a single pass. A motion controller (PCI-7534, National Instruments, Austin, TX) with a high spatial trigger rate was utilized to minimize the translation motion time during a scan.

All components of the scanning system were integrated into a PC based system under LabVIEW software control (National Instruments). The software permitted configuration of all relevant parameters and allowed data collection with a single push of a button. The software also permitted data to be viewed or saved as individual transmit/receive pairs or as a single, processed image using the synthetic focus method described in Sec. II-B. A typical full data set run took ≈ 40 s with the majority of the time devoted to data capture and transfer from the digital oscilloscope.

D. Wire Phantom

The performance of the array was tested with a wire phantom. The phantom consisted of 25- μm diameter wires diagonally spaced at 1 mm by 1 mm intervals. The phantom accomplished several things. First, it allowed the acoustic beam to be characterized for individual transmit/receive pairs. By scanning the transducer across an individual wire, two-way lateral beam profiles could be measured [22]. Because there were wires at multiple depths, the lateral beam profile could be determined at several depths with a single translational pass.

Second, the wire simulated a point source at different depths and permitted the relative pulse/echo delays to be compared between the transmit/receive pairs. Comparing these delays to the theoretical predictions allows for verification that the array elements are operating as expected. In particular, when the wire is placed at the geometric focus where the relative delays between transmit/receive pairs is zero, the signals sampled with the digital oscilloscope and the PCI digitizer can be properly aligned.

Finally, the phantom permitted the synthetic focusing algorithm to be tested with targets at distinct depths. With zero time delay data, the reflected signal amplitude is greatest when the wire is at the geometric focus and then severely drops off when moving in either axial direction. When the data are processed with focal zones at the different wire depths, the reflected signal increases over an expanded axial range.

III. Results and Discussion

A. Theoretical

Two-way beam simulations for individual transmit/receive ring pairs were computed using (3), and then (5) was used to examine how the axial beam profile changed as the number of focal zones were increased (Fig. 5). Figure 5(a) presents the behavior of the beam when all annuli were excited with no delays added to the RF data representing the case of a single element transducer. The axial profile shows a strong peak at the geometric focus of 12 mm and the 6-dB axial DOF is 0.89 mm.

Figure 5(b) shows the effect of shifting the focus from 5 mm to 20 mm in steps of 1 mm. Multiple peaks are now visible with each peak corresponding to one of the focal zones. This leads to an effective increase in the DOF but, in this case, with strong scalloping between focal zones. Also apparent in the figure is a slight shift of the maximum axial peak toward the transducer. While not visible in the figure because the windowing scheme of (5) suppresses them, secondary axial peaks begin to form as the focus shift becomes more extreme, effectively lowering the peak reflected amplitude at the desired focal location.

In Figure 5(c) the number of focal zones was further increased to 31 with a step size of 0.5 mm. The additional focal zones have helped smooth out the axial profile and have reduced the scalloping seen when the focal zones were 1 mm in width [Fig. 5(b)]. Further increases in the number of focal zones will lead to additional smoothing of the axial profile. While a continuous dynamic focus with a focal zone for every RF point would be the ideal, this approach is not practical from a computational standpoint.

From these axial beam simulations and by also examining B-mode images taken from continuous scattering media, we determined that the focal zone width needs to be < 0.25 mm for a fairly smooth axial beam profile. For focal zone widths > 0.25 mm, the scalloping between focal zones becomes significant and may lead to brightness banding in the final B-mode images. The results in Fig. 5(c) show that signal reflection improves from ≈ 8 to 18 mm. However, a more reasonable DOF range for *in vivo* applications would be the 6-dB points from roughly 9 to 15 mm. Outside of this range the two-way reflected signal starts to substantially decrease and the lateral beamwidth begins to increase causing loss of focus.

B. Experimental

The experimental results will be broken into two components. First, the behavior of individual transmit/receive ring pairs are compared to theoretical predictions in terms of relative delays. If the ring pairs don't behave as predicted, the synthetic focusing method cannot be properly utilized. Second, data from all 25 transmit/receive pairs are employed to verify that a significant improvement in DOF may be achieved with the annular array. Unless otherwise noted, all time domain data are digitized at 1 GS/s.

Figure 6 shows collections of time domain signals for all transmit/receive pairs at multiple wire depths. Each column represents a single wire depth and the traces are shown relative to the channel pair with the earliest time of arrival. The key feature to note is the distinct increase in relative delays between ring pairs when moving away from the geometric focus at 12 mm. When moving closer to the transducer (< 12 mm), the center element (Xmit-Rcv 1-1) is the closest element and the outer element (5-5) is the furthest as observed in Fig. 6. When moving in the opposite direction (> 12 mm) the reverse behavior holds true and the ring pair (5-5) arrives earliest.

At the geometric focus of 12 mm, the round trip distance is identical and we expect to see zero relative delay between channels. This property of the array allows us to calibrate the two digitizers in terms of time delays as discussed in Sec. II-C. Using this method, we found a 43-ns relative delay between the two digitizers. In terms of actual 1-ns steps, the traces at 12 mm align to within ± 2 ns.

Of more interest are the relative delays between transmit/receive pairs at depths other than the geometric focus. Table I shows a comparison between experimental delays and theoretical predictions for a wire located at 9 mm. The tables are normalized to the time of ring pair (1-1) and the units of measurement are 1-ns time steps. The tables are characterized by a symmetry along the diagonal and increasing delays when moving towards the lower right corners of the tables. The agreement between experiment and theory provides confirmation that the array elements are operating as expected. While they are not shown here, similar levels of agreement were observed for wires in 1 mm increments between 8 and 19 mm. Outside of this range, the data became too noisy to analyze.

With the operation of the individual ring pairs verified, the next phase of testing focused on the implementation of the synthetic focusing. Put simply, at each wire depth the digitized time domain signals are time shifted by intervals such as those in Tab. I, effectively aligning the

signals in each column of Fig. 6. A final image was constructed as a composite of data from each wire depth.

Figure 7(a) shows the baseline case of zero time shifts and represents the DOF attainable with a single element transducer having a 6-mm total aperture and a 12-mm geometric focus. Note the peak at 12 mm and the rapid drop off of signal when moving away from 12 mm. The data processed with the synthetic focusing algorithm are shown in Fig. 7(b). The experimental data show a clear improvement in terms of DOF relative to the non-time shifted case. The figure also demonstrates that the two-way echo amplitude is enhanced over the range of 8 to 19 mm.

One benefit of an annular array is its ability to maintain a fairly uniform two-way lateral resolution over the axial range of the synthetic focusing. Figure 8(a) shows theoretical two-way lateral beam profiles after synthetic focusing at a range of wire depths. The trends in peak echo amplitude are the same as what was observed in Fig. 7(b). The sidelobe levels are observed to steadily decrease as the axial position increases and no significant sidelobe peaks are observed.

The corresponding experimental lateral beam data are shown in Fig. 8(b). The overall trends are similar to those observed in Fig. 8(a). However, the sidelobe levels of the experimental data are somewhat higher than the theoretical calculations and the wires at 9, 10, 12, and 14 mm display small sidelobe peaks. The difference between the experimental and theoretical data most likely arises because of a small deviation from perfect spherical curvature for the array, small errors in the time shift delays because of system vibration and numerical rounding, and less than ideal SNR for some of the transmit/receive pairs.

The 6-dB beamwidth values for experimental and theoretical data at a range of wire depths are shown in Fig. 8(c). The values agree to within $\approx 15\%$ and show a gradual increase in beamwidth starting at 9 mm. The lateral beamwidth varies from 60-100 μm over the practical imaging axial depth range of 9 to 15 mm. This is consistent with the theoretical estimate of the DOF for the annular array utilized in these studies [20].

Figure 9 shows a B-mode scan for a collection of 3 wires at 1 mm depth increments. The figure shows both the case of an image formed by the summing the RF data with no time delays [Fig. 9(a)] and an image processed with the synthetic focusing algorithm using focal zones corresponding to the three wire depths [Fig. 9(b)]. Both B-mode plots show the log-compressed A-line envelopes and are normalized to unity to better show the effect of the synthetic focusing. The signal-to-noise ratio (SNR) was evaluated from data before log compression by taking the ratio of the peak signal value and the root mean square of the background noise. The zero time delay image represents what would be observed with an equivalent single element transducer. For this case, the beam appears relatively wide and sidelobes are visible. After the data has been processed with the synthetic focusing algorithm [Fig. 9(b)], the wires become more distinct, no sidelobes are visible, and the relative brightness at each wire location becomes more uniform.

As the first practical test of the annular array data acquisition system we generated images from an excised bovine eye (Fig. 10). The raw RF data were acquired with 200 scan lines, 100- μm spacing between lines, and at a 500-MS/s sampling rate. For this scan, the Avtech monocycle pulser was utilized. The figure shows the angle, the iris, the transition from the cornea to the sclera, and the ciliary body. Fig. 10(a) represents the sum of all 25 ring pairs with no processing and Fig. 10(b) the composite image after synthetic focusing with 41 focal zones (0.12 mm/zone). An improvement in resolution, edge sharpness, and depth penetration is visible in Fig. 10(b). These are all features that are consistent with the enhanced DOF and resolution expected for the annular array.

IV. Conclusion

In a prior publication [18], we established the feasibility of our PVDF annular array devices. Here, we have extended the previous work by developing a data acquisition system to simultaneously acquire RF data from multiple transmit/receive ring pairs and to post-process the acquired data with a synthetic focusing technique to improve the effective DOF of the array.

To test the array functionality, we utilized a wire phantom with 25- μm diameter wires diagonally spaced at 1 mm by 1 mm increments. When a wire was placed at a specific depth, the relative time domain delays between transmit/receive ring pairs agreed with theoretical predictions to within ± 2 ns. This established that the array elements were operating as predicted. For each wire depth, RF data were acquired for all 25 transmit/receive ring pairs and then were post-processed with a synthetic focusing technique. Improvement in the echo amplitude was observed over an axial range of 8 to 19 mm. However, for *in vivo* imaging applications this range will be reduced because of acoustic attenuation and a finite SNR. A more reasonable practical imaging range will be the 6-dB DOF from 9 to 15 mm.

Our current efforts are to perform further system tests by acquiring data from *in vitro* cadaver eyes, *in vivo* rabbit eyes and *in vivo* mouse embryos. In addition, we will continue to optimize the array and the data acquisition system to improve both the SNR and the speed of data acquisition. The current speed of data acquisition (≈ 40 s) is relatively long and will make it difficult to acquire high quality *in vivo* data unless the object being scanned remains perfectly still. One method to significantly improve acquisition time is to utilize a digitizer crate linked to the control computer over a high speed bus. An additional method is to use just a single mechanical scan rather than the five we currently used. This would require custom electronics that permit a rapid sequence of 5 pulses, one for each array element, at each spatial location that RF data are acquired.

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Biographies

Jeffrey A. Ketterling (Member '02) was born in Seattle, WA, in 1970. He received the B.S. degree in electrical engineering from the University of Washington, Seattle, WA in 1994. He received the Ph.D. in mechanical engineering from Yale University, New Haven, CT, in 1999. His thesis focused on experimental studies of phase space stability in single bubble sonoluminescence. Since 1999 Dr. Ketterling has been a Member of the Research Staff at Riverside Research Institute, New York, NY. His research interests include high frequency arrays, transducer sound field modeling, high frequency response of acoustic contrast agents, and ultrasonic tissue characterization.

Sarayu Ramachandran was born in 1976 in Bombay, India. She received the B.S. degree in biomedical engineering from Bombay University, Bombay, India and the M.S. degree in biomedical engineering from Columbia University, New York, NY. She has been a Member of the Research Staff at Riverside Research Institute since 2001. Her research interests include signal processing, ultrasound beam simulations, image segmentation of high frequency ultrasound images and ultrasonic tissue characterization.

Orlando Aristizábal was born in Popayán, Colombia, on April 11, 1964. He received the B.S. degree in Physics in 1986 from City College of New York and the M.Phil. degree in Physics in 1993 from the Graduate Center of the City University of New York. During this time he studied the acoustomagnetic properties of Type I and Type II superconducting films. From

1993 to 1995 he was a visiting scholar at the University of Illinois in Urbana-Champaign where he studied the electrical properties of YBCO and other high temperature superconducting thin films. Since 1996 he has been a Research Scientist in the in vivo micro-imaging laboratory headed by Dr. Daniel Turnbull at the Skirball Institute of Biomolecular Medicine, New York University School of Medicine. His research interests are focused on the development and use of high frequency ultrasound imaging methods and their applications.

References

1. Lockwood GR, Turnbull DH, Christopher DA, Foster FS. Beyond 30 MHz - Applications of high-frequency ultrasound imaging. *IEEE Eng. Med. Biol. Mag* 1996;15(6):60–71.
2. Silverman RH, Rondeau MJ, Lizzi FL, Coleman DJ. Three-dimensional high-frequency ultrasonic parameter imaging of anterior segment pathology. *Ophthalmology* 1995;102(5):837–843. [PubMed: 7777285]
3. Foster FS, Pavlin CJ, Harasiewicz KA, Christopher DA, Turnbull DH. Advances in ultrasound biomicroscopy. *Ultrasound Med. Biol* 2000;26(1):1–27. [PubMed: 10687788]
4. Turnbull DH, Foster FS. Ultrasound biomicroscopy in developmental biology. *Trends Biotechnol* 2002;20:S29–S33.
5. Phoon CKL, Turnbull DH. Ultrasound biomicroscopy-doppler in mouse cardiovascular development. *Physiol. Genomics* 2003;14(1):3–15. [PubMed: 12824473]
6. Turnbull DH, Starkoski BG, Harasiewicz KA, Semple JL, From L, Gupta AK, Sauder DN, Foster FS. 40-100 MHz B-scan ultrasound backscatter microscope for skin imaging. *Ultrasound Med. Biol* 1995;21(1):79–88. [PubMed: 7754581]
7. de Korte CL, van der Steen AFW, Céspedes EI, Pasterkamp G, Carlier SG, Mastik F, Schoneveld AH, Serruys PW, Bom N. Characterization of plaque components and vulnerability with intravascular ultrasound elastography. *Phys. Med. Biol* 2000;45(6):1465–1475. [PubMed: 10870704]
8. Sherar MD, Foster FS. The design and fabrication of high-frequency poly(vinylidene fluoride) transducers. *Ultrason. Imaging* 1989;11(2):75–94. [PubMed: 2734975]
9. Cannata JM, Ritter TA, Chen WH, Silverman RH, Shung KK. Design of efficient, broadband single-element (20-80 MHz) ultrasonic transducers for medical imaging applications. *IEEE Trans. Ultrason. Ferroelect. Freq. Contr* 2003;50(11):1548–1557.
10. Lockwood GR, Turnbull DH, Foster FS. Fabrication of high frequency spherically shaped ceramic transducers. *IEEE Trans. Ultrason. Ferroelect. Freq. Contr* 1994;41(2):231–235.
11. Aristizábal O, Christopher DA, Foster FS, Turnbull DH. 40-MHz echocardiography scanner for cardiovascular assessment of mouse embryos. *Ultrasound Med. Biol* 1998;24(9):1407–1417. [PubMed: 10385963]
12. Ritter TA, Shrout TR, Tutwiler R, Shung KK. A 30-MHz piezo-composite ultrasound array for medical imaging applications. *IEEE Trans. Ultrason. Ferroelect. Freq. Contr* 2002;49(2):217–230.
13. Cannata, JM.; Shrout, TR.; Shung, KK. *Proc. IEEE 13th Int. Symp. Appl. Ferroelect. Nara; Japan: 2002. A 35 MHz linear ultrasonic array for medical imaging; p. 343-346.*
14. Alves CHF, Snook K, Ritter T, Shung KK. High frequency single element and annular array transducers incorporating PVDF. *Proc. SPIE Med. Imag* 2000;3982:116–121.
15. Snook KA, Zhao JZ, Alves CHF, Cannata JM, Chen WH, Meyer JJ, Ritter TA, Shung KK. Design, fabrication, and evaluation of high frequency, single-element transducers incorporating different materials. *IEEE Trans. Ultrason. Ferroelect. Freq. Contr* 2002;49(2):169–176.
16. Morton CE, Lockwood GR. Design of a 40 MHz annular array. *Proc. IEEE Ultrason. Symp.* 2001 2001;2(2001):1135–1138.
17. Brown JA, Démoré CEM, Lockwood GR. Design and fabrication of annular arrays for high-frequency ultrasound. *IEEE Trans. Ultrason. Ferroelect. Freq. Contr* 2004;51:1010–1017.
18. Ketterling JA, Aristizábal O, Turnbull DH, Lizzi FL. Design and fabrication of a 40-MHz annular array transducer. *IEEE Trans. Ultrason. Ferroelect. Freq. Contr* 2005;52(4):672–681.
19. Arditi M, Foster FS, Hunt JW. Transient fields of concave annular arrays. *Ultrason. Imaging* 1981;3(1):37–61. [PubMed: 7195094]

20. Arditi M, Taylor WB, Foster FS, Hunt JW. An annular array system for high-resolution breast echography. *Ultrasonic Imaging* 1982;4(1):1–31. [PubMed: 7199769]
21. Talman JR, Garverick SL, Morton CE, Lockwood GR. Unit-delay focusing architecture and integrated-circuit implementation for high-frequency ultrasound. *IEEE Trans. Ultrason. Ferroelect. Freq. Contr* 2003;50(11):1455–1463.
22. Raum K, O'Brien WD. Pulse-echo field distribution measurement technique for high-frequency ultrasound sources. *IEEE Trans. Ultrason. Ferroelect. Freq. Contr* 1997;44(4):810–815.

Fig. 1.

Array pattern etched onto the copper clad polyimide film. The center element is Ring 1 and the outermost annulus is Ring 5. The horizontal lines that break the symmetry of the annuli lead to electrical connection pads.

Fig. 2. Annular array mounted in a UHF connector. Silver epoxy forms the ground plane from the PVDF to the UHF case. A PCB is utilized to go from the flex connector of the array to standard coaxial cables. Electrical impedance matching is accomplished on the PCB.

Fig. 3.
The transducer surface velocity signal (solid) and corresponding spectrum (dotted) used in the simulations.

Fig. 4.
Experimental system diagram.

Fig. 5.

Theoretical axial beam profiles for the full 25 transmit/receive combinations as the number of focal zones is increased. The initial case, (a), represents a single focal zone or the case of no delay corrections. The number of focal zones are then increased to 16 and 31, representing Δz intervals of (b) 1 mm and (c) 0.5 mm, respectively.

Fig. 6. Relative time delays for all 25 transmit/receive pairs at multiple wire depths. All traces have been normalized to one.

Fig. 7.
Two-way theoretical axial response (solid) compared to experimental data (diamonds) with a) no focusing and b) with 0.5 mm focal zones.

Fig. 8. Two-way lateral beamwidths after synthetic focusing at a range of axial depths for a) theoretical and b) experimental data. The corresponding 6-dB beamwidths are compared in c).

Fig. 9. B-mode images for a collection of three wire depths for both a) a zero time delay image (34-dB SNR) and b) an image processed with the synthetic focusing algorithm (54-dB SNR). The data in each image has been log compressed and normalized to unity.

Fig. 10.

Images of excised bovine eye for a) case with no time delays added (46-dB SNR) and b) with synthetic focusing applied (51-dB SNR). The transducer is positioned at the top of the image.

TABLE I
RELATIVE TWO-WAY XMIT-RCV DELAYS* FOR A WIRE AT AN AXIAL DEPTH OF 9 MM

	1	2	Theory 3	4	5	1	2	Experiment 3	4	5
1	0	-18	-37	-55	-75	0	-20	-38	-56	-75
2	-18	-36	-55	-74	-93	-21	-36	-55	-72	-92
3	-37	-55	-73	-92	-111	-37	-55	-73	-90	-110
4	-55	-74	-92	-111	-130	-56	-73	-92	-108	-129
5	-75	-93	-111	-130	-149	-74	-92	-110	-128	-148

* Values are in ns.