agreed well with previously published results. At higher pressures, the evolution of Optison microbubbles appeared to involve a decrease in shell parameter, implying that the shell was thinning, cracking, or buckling. Changes in Sonazoid microbubbles did not show signs of changes in shell parameter, but rather, the bubble size appeared to change, implying that gas (in this case, air) dissolved into the bubble during insonification.

The authors thank L. Crum, A. Brayman, and W. Chen for many helpful contributions. This work is supported by NIH grant 8RO1 EB00350-2.

Session: 2D

PHASE ABERRATION
Chair: M. O’Donnell
University of Michigan

2D-1 8:30 a.m.

REAL-TIME ADAPTIVE IMAGING ON 1.75D, HIGH FREQUENCY ARRAYS
S. A. MCALEAVEY*, J. J. DAHL1, and G. E. TRAHEY1,2, 1Duke University, Durham, NC, 2Duke University Medical Center, Durham, NC.
Corresponding e-mail: mcaleave@duke.edu

We have created a system for real-time phase aberration correction functional with a 7.5 MHz, 1x192 element 1D linear array (Siemens) and a 9.5 MHz, 8x96 element, 1.75D linear array (Tetrad). The 1.75D array has a fixed elevation focus of 24 mm, a lateral element pitch of 0.20 mm, and an elevation pitch of 1 mm. RF echo signals for a span of axial ranges of up to 3 cm are acquired for each element. Echo data are transferred from a Siemens Antares scanner via 100 Mb/s Ethernet to a cluster of 6 dual-processor Linux OS computers. Parallel processing techniques are used to calculate and apply geometric focusing delays to the data, extract a range window corresponding to the brightest scatterer, and estimate an aberration profile. Aberration profiles are estimated with nearest neighbor correlation and weighted least squares algorithms. The calculated aberration profiles are used to modify the transmit and receive phasing of array elements.

The parallel implementation requires 0.9 seconds on 1 cluster processor, 0.7 on 2 processors, and 0.53 seconds on 6 processors, allowing a frame rate of 2 Hz. Nearest neighbor correction on the 8x96 probe, using a smaller range window, requires 2 seconds on a single processor and 0.8 seconds on 12 processors. We anticipate increased frame rates as we develop improved beam sequences for single-channel data acquisition. Computation and I/O rate limitations are discussed.

We present aberrated and corrected images of phantoms with point and speckle targets produced with one- and two-dimensional physical aberrators for both arrays. Images and processing times are presented for each algorithm and array.
We have imaged the thyroids, breasts, and carotid arteries of patients at Duke University Medical Center with this system. Pre- and post-correction in vivo images and estimated aberrators are presented. Issues of spatial and temporal stability of several adaptive imaging methods are discussed.

The authors wish to thank Siemens Medical Solutions USA, Ultrasound Division, for systems support. This work was funded in part by NIH Grant R01-CA-43334.

2D-2 8:45 a.m.

A CLINICAL STUDY OF ADAPTIVE BEAMFORMING USING TIME-DELAY ADJUSTMENTS ON A 1-D ARRAY
D-L. D. LIU*, J. A. BAKER2, and P. VON BEHREN1, 1Siemens Medical Solutions Ultrasound Group, Issaquah, WA, 2Duke Univ. Medical Center, Durham, NC.
Corresponding e-mail: donald.liu@siemens.com

Time-delay estimation is made using echoes from a user-selected target region, to which the transmit focus is applied to maximize signal coherence. The estimated time-delay profile is split into 2 components using a least-squares fitting: one component indicates an effective speed of sound, while the other is the residual time-delay error. On subsequent transmit and receive beamforming, the compensation can be made using the first component through modifying normal beamforming calculations, or, in addition, with the second component applied as well. These modifications were made in near real-time on the Elegra®, and were used to perform breast scanning on 7 patients. Images from the clinical scanning will be presented, along with the corresponding channel data and the estimated time-delay profiles. The results indicate that the compensation scheme resulted in significant improvement in image quality in some cases, particularly in patients with predominantly fatty tissue, while in some other cases, the effect is weak or unstable. One of the causes is that the selected target region may contain a cluster of a few strong scatterers and exhibit complex scattering signatures that are mistaken as propagation distortion. Other causes include the isoplanatic patch size, the elevation dimension of each array element, and the limitation of time-delay correction. These causes will be demonstrated using data obtained from clinical scanning and phantom measurements.

Scanning assistance provided by Anna Fernandez (then graduate student at Duke Univ.) is acknowledged with pleasure. This project is partially supported by NIH grant R29 CA61688.

2D-3 9:00 a.m.

ULTRASONIC SHEAR (SH) MODE IN THE HUMAN SKULL
G. T. CLEMENT*, P. J. WHITE, and K. HYNYNEN, Harvard Medical School, Brigham and Women's Hospital.
Corresponding e-mail: gclement@hms.harvard.edu
Noninvasive treatment of brain disorders using focused ultrasound requires a reliable model for predicting the physical distortion of the field due to the skull using physical parameters obtained in vivo. A previous study indicated that control of ultrasound phase alone is sufficient for correcting phase distortion by the skull using a phased ultrasound array [Clement, Phys. Med. Bio. 2002]. However, this study assumed only longitudinal propagation through the bone; an assumption that breaks down as a focus approaches the skull bone. Incident waves experience a partial or full mode conversion from an incident longitudinal wave into a shear wave within the bone. As a result, a measurable amount of energy may reach the brain through this conversion. In order to accurately model transskull propagation, it may therefore be necessary to obtain precise data on shear wave speeds and attenuation in the human skull. Five ex vivo human calvaria were examined. Each sample was also imaged in water using computerized tomography (CT). The information was used to determine the inner and outer skull surfaces, thickness as a function of position, and internal structure. Sound speed measurement over a series of points was obtained by placing a skull fragment between a transducer and a receiver with the beam high incident angles relative to the skull. Shear wave speed fluctuation and absorption changes as a function of density and layer structure are discussed. Finally, we demonstrate instances of significantly improved field modeling, when the SM is considered.

2D-4 9:15 a.m.

OFF-AXIS SCATTERER FILTERS FOR IMPROVED ABERRATION MEASUREMENTS
J. J. DAHL* and G. E. TRAHEY, Duke University, Durham, NC.
Corresponding e-mail: jjd@duke.edu

Successful adaptive imaging requires accurate measurements of the aberration profile across the array surface. This task is straightforward when a point target is present. However, the profile measurement is degraded for clinically realistic diffuse and extended scatterer targets due to the incoherent nature of the source and the effect of off-axis scatterers. Off-axis scatterers generate echo wavefronts from an extended range of arrival angles which, upon summation at the transducer surface, generate artifacts in the measured aberration profile. We have developed off-axis scatterer filters and applied them to simulation and phantom data to remove the effects of off-axis scatters on aberration measurements. These filters were applied to individual channel rf data with lateral spatial cutoff frequencies ranging from 0.2 mm$^{-1}$ to 1.75 mm$^{-1}$. Individual channel data was acquired from both a conventional 1x192 and a custom built 9.0 MHz, 1.75D transducer (Tetrad Corp.) with an 8x96 geometry and interfaced with a Siemens Antares Scanner. Electronic aberrations were applied to the face of transducer to simulate a near field phase screen while data was acquired on a spherical lesion phantom with 2 mm cysts. Off-axis scatterer filtering reduced the amount of residual error after phase correction by up to 36% for the 1x192 array and as much as 21% for the 8x96 array. Furthermore, aberrations measured in the
near field and at outer edges of the array were more accurately depicted for filtered data. Contrast measurements show that lesion contrast can increase up to 18% with filtered data. We applied these filters to clinical thyroid data. For 15 images in 7 volunteers the mean increase in measured aberration strength was 20% (range 8-40%) with application of the filter. This agrees with simulation data which shows that for aberration strengths typically found in thyroid data, adaptive imaging underestimates phase errors as much as 30%. Profiles generated by filtered data show improved spatial and temporal stability and exhibit smooth surfaces. Spike artifacts normally present in profiles created from non-filtered data are also eliminated.

NIH grant RO1-CA43334

2D-5 9:30 a.m.

US IMAGE QUALITY OBTAINABLE THROUGH MAMMOGRAPHY COMPATIBLE COMPRESSION PADDLE

J. F. KRÜCKER*, R. HOCTOR, A. KAPUR, M. M. GOODSITT, P. L. CARSON, and K. THOMENIUS,

1University of Michigan, Department of Radiology, Ann Arbor, MI, 2General Electric Global Research Center, Schenectady, NY.

Corresponding e-mail: jokr@umich.edu

Automated 3D ultrasound (US) scanners were developed and evaluated for whole breast imaging. The scanning systems include acoustically transparent compression paddles for compression of the breast akin to that used for mammography. Two prototype units were built, one for use in combination with a digital mammography system (GE Senographe 2000D), the other a stand-alone system for research and future clinical US studies. The systems use automated 2D probe mover assemblies for precise scanning of a high-frequency, matrix array probe. A key design challenge is the compression paddle material, which must satisfy mechanical as well as acoustic and, for the combined system, X-ray constraints. Two design options were investigated, use of a rigid plastic plate similar to those found in mammography systems, or use of a thin, stretched film. The materials tested were a 3 mm thick, low acoustic impedance TPX plastic plate and sub-mm thin Mylar film membranes. Evaluation of image quality was performed using commercial and custom-built phantoms as well as in-vivo scans. For imaging through the plate, the beamforming of the scanner was modified to account for the higher sound speed in the plate compared to the assumed sound speed in tissue. B-mode images and RF data from line targets were analyzed to show the efficacy of the correction. For both plate and films, attenuation, reverberations, contrast, and the full width at half max (FWHM) of the point spread function were measured. Without beamforming corrections, the image through the plate was severely degraded (86% increase in lateral FWHM, 3.5dB contrast reduction in a -30dB hypoechoic structure, 8dB signal loss averaged over 4cm depth). With corrections, image quality with the TPX plate was comparable to that with the thickest Mylar membrane tested (7% increase in lateral
FWHM, 0.5dB contrast reduction, 3dB signal loss). Elevational FWHM was affected only modestly in both cases (increase up to 6%). Modest reverberations from the plate in the first cm of the image may be reduced by adding matching layers. With both types of compression paddles, the advantages of generating whole breast image volumes with high fidelity, resolution, and comparability to mammography should outweigh the minor losses in image quality.

This work was supported in part by NIH grant RO1 CA 091713.

2D-6 9:45 a.m.

TWO-DIMENSIONAL PHASE ABERRATION CORRECTION USING AN ULTRASONIC 1.75D ARRAY: CASE STUDY ON BREAST MICROCALCIFICATIONS
A. T. FERNANDEZ*1 and G. E. TRAHEY2, 1Philips Research USA, Briarcliff Manor, NY, 2Duke University, Durham, NC.
Corresponding e-mail: anna.fernandez@philips.com

Ultrasound phase aberration resulting from tissue velocity inhomogeneities reduce the focusing ability of ultrasound waves and degrade image quality. Two-dimensional phase aberration measurements and correction with higher-order 1.75D arrays are expected to reduce this problem. We implement a least-mean-squares measurement algorithm and discuss implementation decisions for phase aberration correction techniques. We present clinical results from using individual channel RF signals from an 8x128 1.75D array (Tetrad Co.) interfaced with a Siemens Elegra scanner. The average patient aberration measurement in the thyroid (9 patients) was 22.2 ± 4.6 ns r.m.s amplitude with 4.8 mm ± 1.3mm FWHM autocorrelation length. Analysis of breast scans (8 patients) resulted in a patient average aberration measurement of 33.1 ± 7.0 ns r.m.s amplitude and 6.4 ± 1.5 mm FWHM. The results show a statistically significant difference (p-value = 0.01) in aberration measurements in clinical breast imaging patients in age groups 20-40 years old and 50-70 years old. Improvements from aberration correction are challenging to evaluate in clinical images of soft tissues. Microcalcifications act as in vivo point targets and allow quantification of image quality improvement with adaptive imaging. We present results of receive-only aberration correction in clinical images and describe quantitative improvements and the results from using specific correction implementation techniques in a case study of breast microcalcifications after phase aberration correction. On average, microcalcification brightness and resolution were improved by 53 % and 16 % respectively.

This work was supported by NIH grant RO1-CA43334, and with in-kind and technical support from Siemens Medical Solutions USA, Inc., Ultrasound Division.