7.3 L1 NORM DECONVOLUTION FOR MEDICAL ULTRASOUND USING QUADRATURE SIGNALS, X. M. Lu and J. M. Reid, Biomedical Engineering and Science Institute, Drexel University, Philadelphia, PA 19104.

In pulse-echo ultrasound system, the axial resolution is limited by the transmitted pulse duration. Assuming a one-dimensional ultrasound system, the echo can be described as,

$$r(t) = w(t) * s(t) + n(t) \qquad (1)$$

where r(t) is the received echo signal at the transducer, w(t) is the transmitted pulse or wavelet, s(t) is the medium impulse response and n(t) is the measurement noise. The asterisk denotes convolution. The wavelet w(t) is a "blurring" function that smears the shape of s(t). Various deconvolution techniques have been investigated for the resolution enhancement. The objective of deconvolution is to estimate s(t) based on the measurements of r(t) and w(t) with minimum distortion.

Most-well known deconvolution techniques, such as the Wiener pulse sharping filter, minimum variance deconvolution and frequency domain deconvolution, are based on the least-square error criterion (L2 norm). All of these techniques share the same problems: they are not robust and "ring" effects exist in the sharpened pulse [1]. In contrast, the L1 norm deconvolution is superior for those problems. By minimizing the absolute error, L1 deconvolution is much less sensitive to the large errors than is L2. Large errors occur in deconvolution, since it is a highly ill-conditioned problem [2].

The L1 norm deconvolution can be formulated as a linear programming problem. The implementation requires large amount of memory and computing time. It may be one of its limitations for real applications. The quadrature signal, which defines the complex envelope of a bandpass signal demodulated at carrier frequency, has a slowly changing waveform. Therefore, the sampling rate for the demodulated signal can be reduced several fold. Two L1 norm deconvolution methods for data in quadrature form have been derived. These not only greatly reduce the size of the problem but also lessen the effects of undersampling on s(t). One algorithm uses the complex envelope of the wavelet and the other only needs the magnitude, if the phase of quadrature wavelet is constant. The limitations for each algorithm are discussed.

Preliminary tests on simulated and experimental data showed that the L1 norm method performs very well, even in the presence of moderate noise level, for both algorithms. The quadrature algorithms are over 10 times faster and use about 4 to 8 times less memory than rf methods. The quadrature algorithms were also applied to signals from biological tissues. The deconvolved ultrasound images show more details of the microstructure of tissues.

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[2] Ekstrom, M.P., *IEEE Trans. Audio. Electroacoustics AU-21*, 344 (1973)

7.4 SIDELOBE REDUCTION OF NONDIFFRACTING PULSE-ECHO IMAGES BY DECONVOLUTION, Jian-yu Lu and James F. Greenleaf, Biodynamics Research Unit, Department of Physiology and Biophysics, Mayo Clinic and Foundation, Rochester, MN 55905.

B-scan images produced by a J_0 Bessel nondiffracting transducer have large depths of field and high resolution but present higher sidelobes than those obtained by conventional focused transducers. In this paper, two-dimensional deconvolution was employed to improve both the lateral and axial resolution as well as to suppress the sidelobes of nondiffracting Bessel beams. For pulse-echo images of a bead phantom, deconvolution was done on both simulated and experimented images. Results show that about 10 dB suppression of sidelobes was achieved in addition to resolution improvement. Sidelobe suppression was also observed for the B-scan images of an RMI413A tissue-equivalent phantom and for excised human tissue samples.

Because of the large depth of field of the nondiffracting beam, only a few deconvolution kernels are required for image deconvolution over the entire region of interest, allowing for the possibility that deconvolution can be done in real time.

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7.5 NONDESTRUCTIVE EVALUATION OF MATERIALS WITH A J_0 BESSEL TRANSDUCER, J-y. Lu and J. F. Greenleaf, Biodynamics Research Unit, Department of Physiology and Biophysics, Mayo Clinic and Foundation, Rochester, MN 55905.

Nondiffracting beams such as the J_0 Bessel beam and X-waves are newly discovered propagation invariant solutions of the isotropic/homogeneous scalar wave equation [1,2]. These beams can be almost realized exactly with a finite aperture physical device producing a much larger depth of field than that of the conventional diffracting beams such as focused and Gaussian beams. We applied a finite aperture approximated J_0 Bessel nondiffracting beam to nondestructive evaluation (NDE) of materials. A 10-element, 50-mm diameter, 2.5 MHz central frequency broadband J_0 Bessel annular array transducer was used to image a steel block phantom containing 11 parallell-drilled holes of different diameters that act as "flaws" of the material. The phantom was placed at several depths in water within the depth of field of the transducer. Resulting B-scan images of the phantom show uniform lateral and axial resolution of about 2 mm over a large depth of interest (230 mm) using the lensless J_0 Bessel transducer with a flat surface. The difficulties of applying a focused beam to nondestructive evaluation of materials due to sound speed diversities of the materials could be eliminated with the diffraction-controlled beams. The axial resolution of B-scan images can be increased using a diffraction-controlled transducer of higher central frequency and broader bandwidth.

This work was supported in part by grants CA 54212 and CA 43920 from the National Institutes of Health.

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7.6 TWO-DIMENSIONAL ARRAY TRANSDUCERS USING MULTILAYER CERAMIC TECHNOLOGY, S.W. Smith and E.D. Light, Duke University, 4816 Montuale Drive, Durham, NC 27705.

We have previously described 2-D array transducers consisting of 16 x 16 = 256 PZT elements operating at 2.5 MHz that use 128 transmit elements and 32 receive elements. Element size is 0.4 mm x 0.4 mm. There are severe fabrication difficulties in electrical connection to such elements, which are less than one ultrasound wavelength on a side. To solve this problem, we have used a multilayer ceramic connector (hybrid microelectronic technology) consisting of 20 thick films of alumnia and screen printed metallization with customized interconnections between the layers called vias. Nineteen ground layers are included between the signal layers to reduce electrical crosstalk. A $\lambda/4$ mismatching layer of conductive epoxy is bonded between each PZT element and the silver metal pad of the MLC connector to provide a low impedance backing. In the current configuration, a 16 x 16 transducer array, 0.6 mm element spacing, is expanded to a 16 x 16 grid of connector pins at a standard spacing of 2.5 mm. Hybrid microelectronic circuit technology shows promise for solving the fabrication problems of 2-D array transducers over a thousand elements at frequencies exceeding 5 MHz.

7.7 3-D VELOCITY ESTIMATION USING A DUAL APERTURE TRANSDUCER, Keith S. Dickerson[1], V.L. Newhouse[1], D. Cathignol[2] and J-Y Chapelon[2], [1]Drexel University Biomedical Engineering and Science Institute, Philadelphia, PA 19104 and [2]INSERM Unite 281, Lyon, France.

Conventional Doppler techniques estimate only the axial velocity component of 3-D blood flow. We propose a 3-D estimation of blood flow based on the analysis of the Doppler spectrum produced by a transducer that has two apertures. It is known that for line flow, the Doppler spectral width is proportional to the velocity transverse to the sound beam. It was shown recently that line flow insonified by a transducer with two apertures, both of which simultaneously transmit and receive, can produce a tripled peaked Doppler spectrum when suitably oriented. The spacing of the peaks, combined with the spectral widths and knowledge of the geometry of the dual element transducer, yields an estimate of he three velocity components. At low velocities, the peaks tend to merge so that individual, successive firing of the apertures has to be employed. This method may be readily adaptable to existing Doppler units since it makes use of FFT spectral information. Experiments were performed using a pair of circular aperture transducers, focused to 20 mm. An accuracy analysis of the 3-D velocity estimation will be presented.

Work supported in part by the National Science Foundation.

7.8 REAL-TIME ANGLE-INDEPENDENT ULTRASONIC IMAGING OF BLOOD FLOW: INITIAL RESULTS, L.N. Bohs, B.H. Friemel and G.E. Trahey, Department of Biomedical Engineering, Duke University, Durham, NC 27706.

We have previously described a system which uses the Sum-Absolute-Difference (SAD) algorithm to track the motion of small regions from one ultrasonic frame to the next in order to produce a two dimensional color velocity map [1]. In this paper, we report on enhancements to the system that

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