Computer Aided Tomography and Ultrasonics in Medicine, Raviv et al. (eds.)

RECONSTRUCTIVE TOMOGRAPHY BASED ON ULTRASONIC ATTENUATION

J. G. Miller, J. R. Klepper, G. H. Brandenburger, L. J. Busse,
M. O'Donnell, and J. W. Mimbs

Washington University
St. Louis, Missouri 63130, U.S.A.

ABSTRACT. This report addresses the potential of reconstructions based on quantitative indices of ultrasonic attenuation for characterizing the state of soft tissue. Emphasis is on those aspects of the method which are dominated by front-end considerations, i.e., by the propagation and detection of the ultrasonic signal. Investigations indicate that phase cancellation effects associated with the use of piezoelectric receiving transducers represent the dominant source of error in attenuation tomography. Results are presented which demonstrate the elimination of phase cancellation artifacts by the use of a receiving transducer based on the acoustoelectric effect. Reflection and refraction artifacts also serve to degrade the results of attenuation tomography. In the present studies, reflection effects are minimized and refraction effects are reduced by reconstructing the frequency derivative of the attenuation coefficient.

I. INTRODUCTION

The present work was motivated by previous studies from this and other laboratories suggesting that quantitative indices based on ultrasonic attenuation may be capable of differentiating normal and pathological tissue.[1,2] Thus the potential exists for characterizing the state of tissue in localized regions provided that quantitative values of attenuation can be obtained using computed tomography. An illustration of the potential use of a quantitative index of ultrasonic attenuation to characterize the state of tissue is presented in Figure 1, which summarizes the results of measurements from our laboratory.[3,4] These investigations were designed to determine whether quantitative alterations in ultrasonic attenuation are associated with myocardial changes occurring after ischemic insult. More than five hundred regions of myocardium from approximately 50 dogs were studied in vitro at one of several intervals after coronary occlusion: 15 minutes, 1 hour, 6 hours, 24 hours, 3 days, 2 weeks, 4 weeks, and 6 weeks. Regions of ischemic injury or infarction were defined independently either on the basis of demarcation of the ischemic zone following injection of colloidal carbon black dye after brief ischemia,[5,6] by an assay based on creatine kinase after more prolonged ischemia,[7] or by a biochemical assay for collagen content for remote myocardial infarction.[8] The ultrasonic attenuation coefficient characterizing each myocardial region was measured over the range 2 to 10 MHz. The ultrasonic attenuation in each region studied was characterized by the slope of a least squares line fit to the attenuation coefficient versus frequency data. In Figure 1, the average values of the slope of the attenuation in "ischemically injured" zones and in "normal" zones from the same hearts are displayed for groups of hearts studied at specified time intervals following coronary occlusion.

Results indicate that ischemically injured myocardial regions investigated 15 minutes, 1 hour, 6 hours, and 24 hours after coronary occlusion exhibit ultrasonic attenuation significantly decreased from that of nonischemic regions. In contrast, significantly increased ultrasonic attenuation is exhibited in zones of infarction investigated at 3 days, 2 weeks, 4 weeks, and 6 weeks following coronary occlusion. These findings suggest that quantitative indices based on ultrasonic attenuation may be suitable for characterizing the state of tissue.

This report addresses the potential of reconstructions based on quantitative indices of ultrasonic attenuation for characterizing the state of soft tissue. Emphasis is on those aspects of the method which are dominated by front-end considerations, i.e., by the propagation and detection of the ultrasonic signal. Investigations indicate that phase cancellation effects associated with the use of piezoelectric receiving transducers represent the dominant source of artifact in attenuation tomography. However, errors associated with phase cancellation are eliminated by the use of acoustoelectric receiving transducers which are under continuing development in our laboratory. Reflection and refraction effects also serve to degrade the results of attenuation tomography. In the present studies, we demonstrate that reflection effects are minimized and refraction artifacts reduced by reconstructing the frequency derivative of the ultrasonic attenuation coefficient.

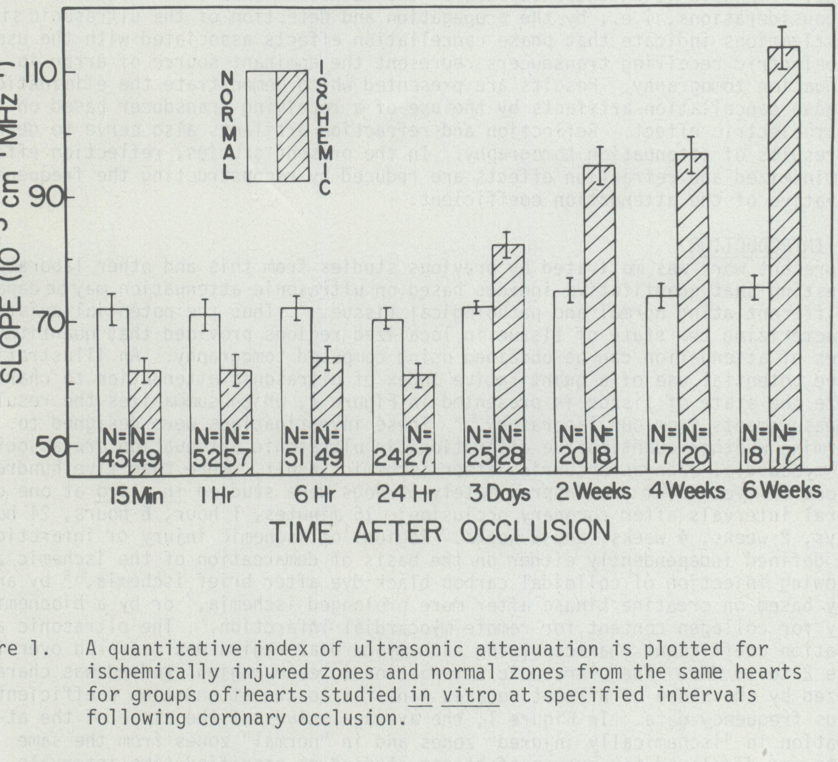


Figure 1. A quantitative index of ultrasonic attenuation is plotted for ischemically injured zones and normal zones from the same hearts for groups of hearts studied in vitro at specified intervals following coronary occlusion.

II. EXPERIMENTAL METHODS

All measurements were performed in transmission. Transmitting and receiving transducers were mounted on arms that extended down from a carriage into a water bath in which the specimen or test object was mounted. The mechanical scanning apparatus consisted of a carriage for linear motion supported from above in a mounting that permitted rotation about a central axis. Translation and rotation were achieved through the use of stepping motors under computer control. The transmitter and receiver were translated in a direction perpendicular to the direction of ultrasonic propagation producing a parallel ray projection of the specimen. Parallel ray projections up to 13 cm in length were obtained with points spaced uniformly at intervals of approximately .5 mm. For purposes of normalization each scan line began and ended with samples taken through paths with only water between transmitter and receiver. After completion of a linear scan the apparatus was rotated through a small angle and another linear scan was taken. This procedure was repeated until projections were obtained for a total angle of rotation of 180 degrees. The number of angles at which projections were carried out ranged from 40 to 120 depending upon the number of steps in the linear scan. All reconstructions were carried out by the filtered back projection technique using standard filters.

To generate the ultrasonic pulse, the output of a continuous wave rf oscillator was gated phase coherently to generate a narrow band pulse of 2 to 5 microsecond duration. The frequency of the voltage tunable cw oscillator was selected under computer control and was in the range 3 to 7.5 MHz. The gated rf pulse was fed to a voltage controlled variable gain preamplifier the output of which was applied to the input to an rf power amplifier. The output of this power amplifier drove the transmitting transducer. The control voltage to the preamplifier was generated by the computer in response to the magnitude of the signal at the receiving transducer. With this auto-ranging scheme, the amplitude of the rf pulse applied to the transmitting transducer was adjusted over a wide dynamic range to compensate for variations in the total attenuation. Thus the receiver was required to operate only over a modest dynamic range to achieve a substantial overall dynamic range.

The transmitting and receiving transducers used to collect the data displayed in Section III are identified explicitly for all results reported. In each case, data presented to contrast different receiving transducers or different processing combinations were obtained with the same transmitting transducer. One of two piezoelectric transmitting transducers of 5 MHz nominal center frequency was used in all investigations: i) a planar 12 mm diameter transducer, or ii) a focused 12 mm diameter transducer of 10 cm focal distance. Piezoelectric and acousto-electric receiver signals were obtained from single crystals of cadmium sulfide plated with indium electrodes. Under appropriate conditions, cadmium sulfide exhibits both a piezoelectric and an acoustoelectric response to an incident ultrasonic pulse. The piezoelectric response is sensitive to the amplitude and phase of the pulse, but the acoustoelectric response is sensitive only to the intensity of the pulse.[9] Selection of either the piezoelectric or the acousto-electric response is accomplished by filtering the output of the cadmium sulfide crystal. The piezoelectric response is an rf pulse centered at the carrier frequency, in contrast with the acoustoelectric response which is a pulse centered at dc with frequency components determined by the width of the transmitted ultrasonic pulse. Thus high pass filtering yields the piezoelectric signal and low pass filtering yields the acoustoelectric signal. Two cadmium sulfide receiving transducers were used in these investigations: i) a 12 mm diameter circular cross section receiver, and ii) a 1.2 cm^2 rectangular cross section receiver of approximate dimensions 14 mm by 8 mm.

III. RESULTS

The present work focuses on those features of the propagation and detection of the ultrasonic pulse which degrade reconstructions based on attenuation. The dominant source of error appears to arise from phase cancellation. Phase cancellation effects occur when an ultrasonic wave exhibiting distorted phasefronts is received by a spatially extended piezoelectric receiver. The electrical output of a piezo-electric receiver is proportional to the instantaneous integral of the pressure across the face of the transducer. The output of the receiving transducer is thus the instantaneous sum of local voltages which are of one sign for local compressions and of the opposite sign for local rarefactions. Hence variations across the face of the receiving transducer in the phase of the ultrasonic waves result in a degraded electrical output. Variations in the phase of the ultrasonic pulse across the face of the receiver result from propagation through media exhibiting variations in the index of refraction. Nevertheless, phase cancellation effects are purely instrumental in character since they are a direct consequence of the phase sensitive nature of piezoelectric receivers.

Previous reports from our laboratory documented the role of phase cancellation effects as a source of error in ultrasonic transmission studies of soft tissue carried out with piezoelectric receivers.[9] Elimination of artifacts due to phase cancellation effects was demonstrated with the use of an intensity sensitive ultrasonic receiving transducer based on the acoustoelectric effect. The magnitude of the output voltage from an acoustoelectric receiver is proportional to the intensity of the received ultrasonic wave. The sign of the output voltage is determined only by the direction of propagation of the wave, i.e., by the direction of momentum transfer, and remains unchanged for local compressions or local rarefactions. Thus the voltage is not affected by the phase of the received ultrasonic wave.

A comparison of the performances of piezoelectric and acoustoelectric receivers of identical aperture (1.2 cm^2) is presented in Figure 2. The apparent attenuation as a function of frequency is displayed for 6 adjacent and morphologically similar regions of dog myocardium. The measurements were carried out using a 12 mm diameter

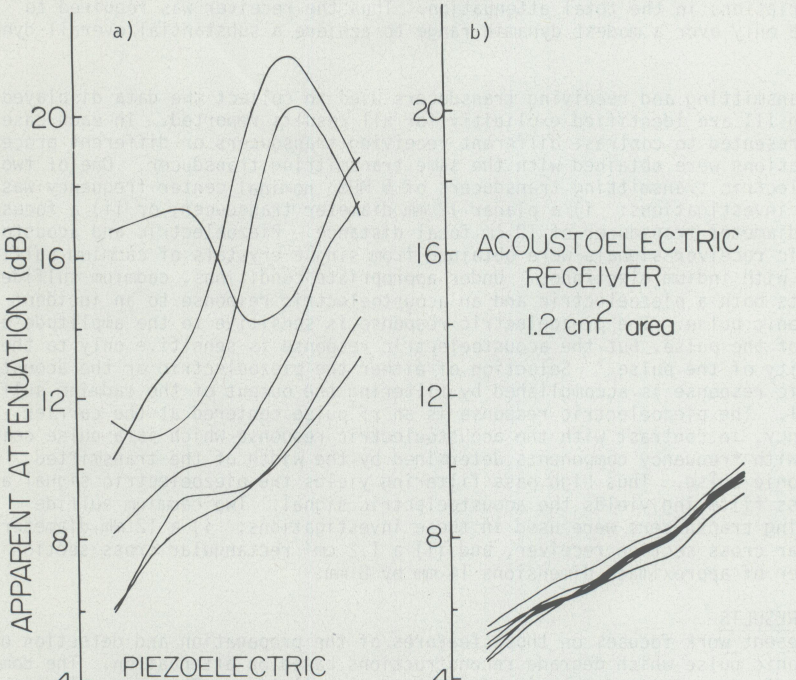


Figure 2. The apparent attenuation at 6 adjacent sites of dog myocardium is plotted as a function of frequency for measurements carried out with a piezoelectric receiver (panel a) and an acoustoelectric receiver (panel b).

planar transmitting transducer. Data obtained with the piezoelectric receiver (Figure 2a) exhibit marked variability, presumably due to phase cancellation effects. In contrast, data obtained at the same sites with the acoustoelectric receiver (Figure 2b) are mutually consistent and exhibit the monotonic and approximately linear dependence on frequency characteristic of soft tissue. The uniformity and consistency of the attenuation data obtained with the acoustoelectric receiver presumably reflect the elimination of phase cancellation artifacts.

The data of Figure 2 suggest that errors associated with phase cancellation may result in significant overestimation of the ultrasonic attenuation if piezoelectric receivers are used. This hypothesis is supported by the statistical results presented in Figure 3. The apparent attenuation at 5 MHz was determined in 40 sites of dog myocardium using the same transducers as in Figure 2. The upper panel of Figure 3 consists of a histogram showing the number of sites exhibiting a specified value of apparent attenuation determined with the piezoelectric receiver. The lower panel of Figure 3 consists of a similar histogram for data obtained on the same sites with the acoustoelectric receiver. The statistical results summarized in Figure 3 indicate that data obtained with a piezoelectric receiver estimate substantially larger and less consistent ultrasonic attenuation than data obtained under identical conditions with an acoustoelectric receiver.

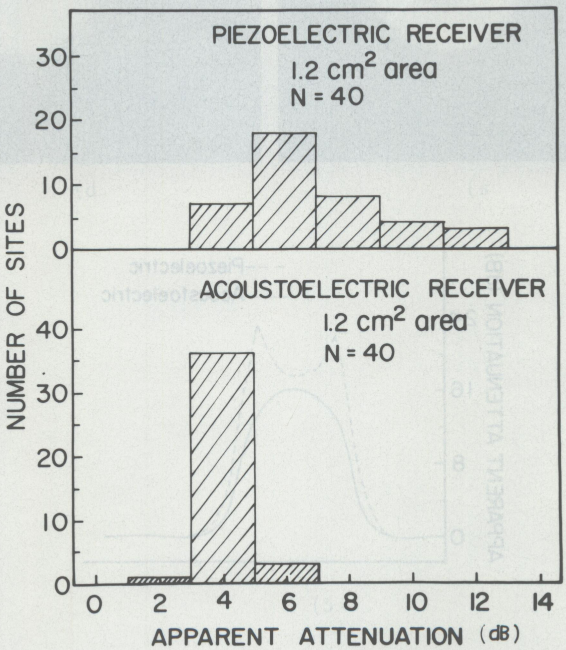


Figure 3. The number of myocardial sites exhibiting a specified value of the apparent attenuation is plotted as a function of the apparent attenuation. The determinations in the upper panel were made using a piezoelectric receiver. The determinations in the lower panel were made at the same sites using an acoustoelectric receiver of the same aperture.

A comparison of results obtained with piezoelectric and acoustoelectric receivers in ultrasonic reconstructive tomography is presented in Figure 4. A cylindrical specimen of castor oil in a 2 cm diameter finger cot was selected as a test object in order to illustrate phase cancellation effects. The characteristic impedance and index of refraction of castor oil are nearly equal to those of water, so reflection and refraction effects are modest. Furthermore, the ultrasonic attenuation coefficient of castor oil is substantial, reducing the relative importance of reflection and refraction losses. A planar, 12 mm diameter transmitting transducer was used. In Figure 4a the results of scanning the castor oil specimen at a frequency of 4 MHz using a 12 mm diameter piezoelectric receiver are illustrated.

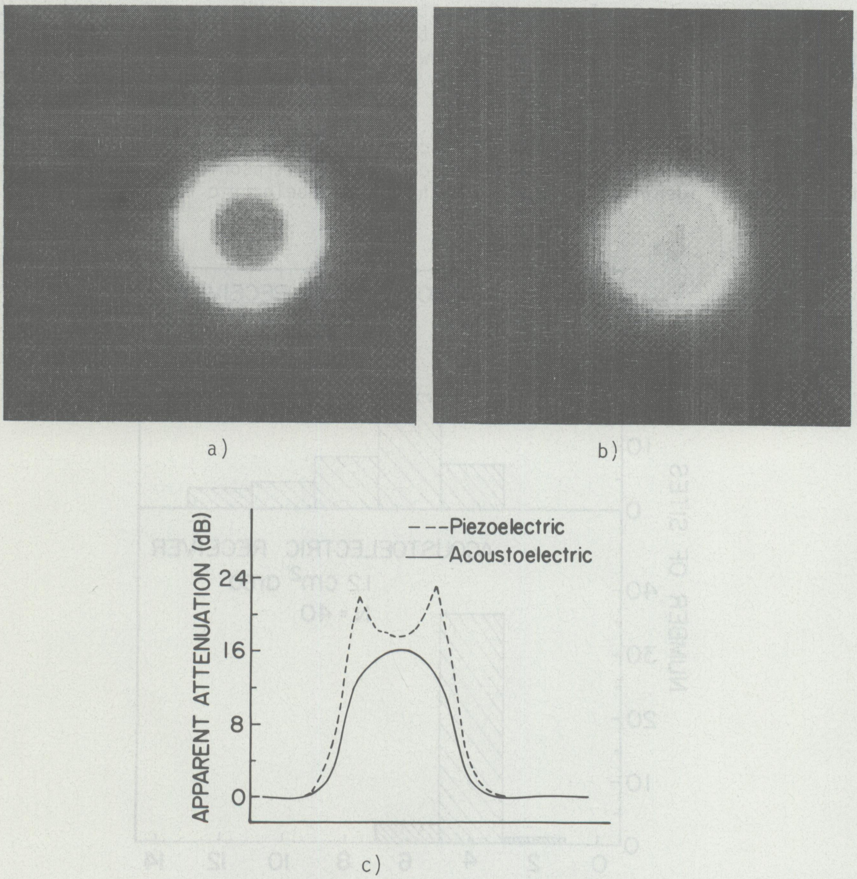


Figure 4. Results of measurements carried out on a cylindrical specimen of castor oil using a 12 mm diameter piezoelectric receiver and a 12 mm diameter acoustoelectric receiver. Panel a) consists of a reconstructed image based on the apparent attenuation at 4 MHz estimated using the piezoelectric receiver. Panel b) consists of a reconstruction of the apparent attenuation at 4 MHz using the acoustoelectric receiver. In panel c) typical parallel ray projections based on the apparent attenuation estimated using the piezoelectric and the acoustoelectric receivers are presented.

The reconstruction exhibits a very pronounced artifact, reflecting a substantial overestimation of the attenuation near the edges of the object. The origin of this artifact is easily identified in the representative parallel ray projection lines shown in Figure 4c. Results obtained under otherwise identical conditions using a 12 mm diameter acoustoelectric receiver are shown in Figure 4b. The reconstruction is notably more uniform than that in Figure 4a, manifesting the elimination of phase cancellation artifacts. A comparison of the projection lines shown in Figure 4c provides a quantitative estimate of the improvements achieved by replacing a piezoelectric receiver with an acoustoelectric receiver. Use of the piezoelectric receiver results in an incorrect estimate of the attenuation as a result of phase cancellation effects. For the homogeneous, cylindrical specimen with index of refraction only slightly different from that of the surrounding medium illustrated in Figure 4, phase cancellation effects are most pronounced for ultrasonic paths near the edges of the object. The origin of the phase cancellation effects in Figure 4a lies in the small time of flight differences between adjacent rays of the spatially extended ultrasonic beam that reach the piezoelectric receiver. Under conditions such as those exhibited by soft tissue, in which differences in indices of refraction are more significant and specimens are not homogeneous, phase cancellation effects are expected for many ultrasonic paths through the specimen.

Phase cancellation effects such as those illustrated in Figures 2, 3, and 4 appear to be the dominant source of error in attenuation tomography. An additional source of error results from refraction of the ultrasonic beam. The most general definition of refraction includes lateral and axial displacements of any portion of the ultrasonic beam as a result of variations in the index of the medium. Qualitatively, an axial displacement may be manifested as a shift in the phase of the ultrasonic beam impinging on the receiving transducer. If the axial displacements of adjacent rays result in variations in phase across the face of the receiving transducer, substantial phase cancellation artifacts may result when the receiver is piezoelectric. These variations in phase are, however, of no consequence when an acoustoelectric receiver is used. In the present context, therefore, attention is focused on lateral beam displacements. Specifically, refraction artifacts may be expected when variations in the index of refraction result in lateral deflections of sufficient magnitude that some or all of the spatially extended ultrasonic beam misses the receiving transducer. Under these conditions, the magnitude of the integrated attenuation will be over estimated.

Artifacts which result from small lateral shifts in the position of the ultrasonic beam due to refraction can be reduced by the use of a relatively large aperture receiving transducer. This approach for reducing refraction effects is illustrated in Figure 5 which shows segments of projections obtained with two orientations of the rectangular receiving transducer, one with lateral dimension 8 mm and the other with lateral dimension 14 mm. The focused transmitting transducer was used in this investigation. The specimen consisted of a cylinder of olive oil contained in a 3.8 cm diameter condom. Olive oil is markedly less attenuating than castor oil, making the effects of refraction losses relatively more significant. To eliminate artifacts arising from phase cancellation, acoustoelectric receivers were used to produce both scan lines. A comparison of the projection obtained using the 8 mm receiver with that obtained using the 14 mm receiver indicates a substantial overestimate of attenuation near the edge of the specimen when the smaller lateral dimension receiver is employed. This overestimate of the attenuation presumably arises from the fact that a substantial fraction of the beam missed the smaller receiver as a result of lateral refraction. Refraction effects can be reduced by adjusting the speed of sound in the coupling medium (e.g., water) to equal that of the object imaged. Modest adjustments in the velocity of sound in water can be achieved by varying temperature or salt concentration.

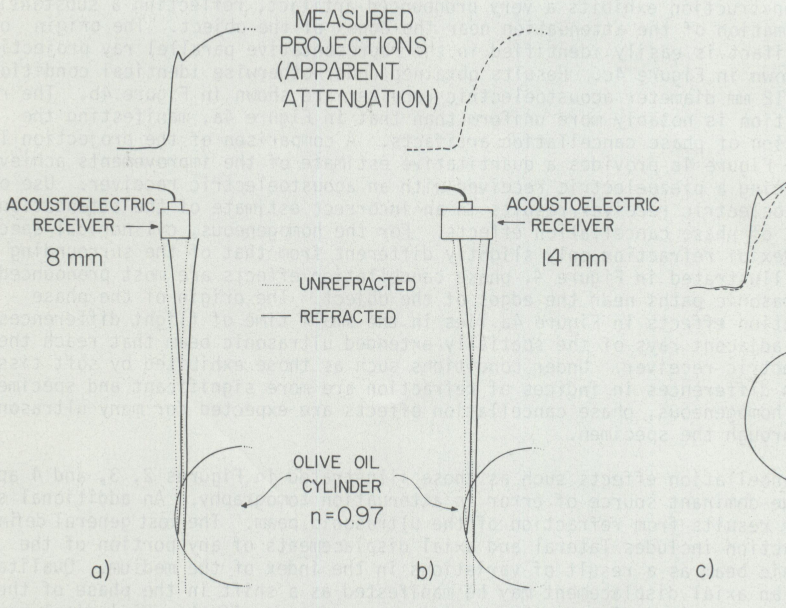


Figure 5. Results of measurements carried out on a cylindrical specimen of olive oil using acoustoelectric receivers of 8 mm and 14 mm lateral dimension. The apparent attenuation estimated from measurements made with the smaller receiver are presented in panel a), and with the larger receiver are presented in panel b). In panel c) the measured parallel ray projections are superimposed.

Reflection losses represent an additional source of error in ultrasonic attenuation tomography. Reflection losses refer to a reduction in the amplitude of the received ultrasonic pulse arising from non-unity transmission coefficients associated with impedance discontinuities encountered along a particular ray. The fact that these transmission coefficients depend upon the angle of incidence is significant, since each impedance discontinuity is viewed from many angles in a tomographic scan. Artifacts due to reflection can be minimized by the following approach. The attenuation coefficients of soft tissue and test objects such as common oils exhibit substantial frequency dependences in the low megahertz frequency range. In contrast, transmission coefficients are independent of frequency (in the absence of standing wave effects). Thus reconstruction of the frequency derivative of the attenuation coefficient estimated from measurements made over a range of frequencies eliminates reflection losses.

This approach to the elimination of reflection losses is illustrated in Figure 6. Two reconstructed images of an olive oil specimen contained in a 3.5 cm condom are presented, one based on the apparent attenuation at a single frequency (panel a) and the other based on the frequency dependence of the attenuation (panel b). A 12 mm diameter acoustoelectric receiver and a planar transmitting transducer were used for both reconstructions. Results of reconstructing the apparent attenuation coefficient at the frequency 4 MHz are presented in panel a) of Figure 6. A marked halo artifact is visible in the reconstruction. Transmission coefficients between media of slightly different refractive indices are expected to differ markedly from unity at angles of incidence that approach 90°, i.e., for paths near the edges of the specimen, presumably giving rise to some of the artifact in panel a) of Figure 6. The reconstruction shown in panel b) of Figure 6 is based on the frequency dependence of the attenuation estimated from the slope of a least squares line fit to the attenuation coefficient versus frequency data measured at 0.5 MHz steps over the range 3.5 to 7.0 MHz. Since transmission coefficients are independent of frequency, a reconstruction based on the frequency derivative is expected to be essentially free from reflection losses, accounting in part for the improved image in panel b) of Figure 6. The use of the slope of the attenuation also reduces the magnitude of artifacts due to refraction. The magnitude of artifacts arising from refraction at a single frequency depends directly upon the extent of lateral beam shift and the aperture of the receiving transducer, as illustrated in Figure 5. In contrast, artifacts arising from refraction in the slope of the attenuation depend only upon differences in the extent of lateral beam shift over the limited range of frequencies used to compute the slope. Thus the reconstruction of panel b) of Figure 6 is improved over that of panel a) by the reduction of refraction artifacts, as well as by the minimization of reflection effects. Artifacts remain due to differences between the actual path traveled by the refracted beam and the straight line path assumed in the reconstruction algorithm.

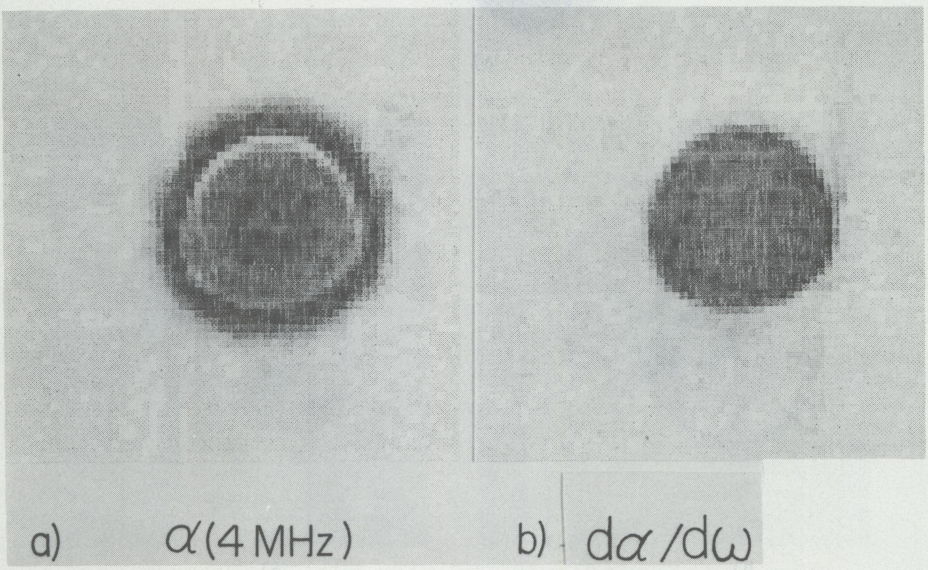
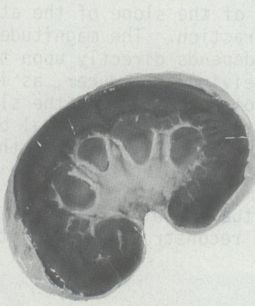
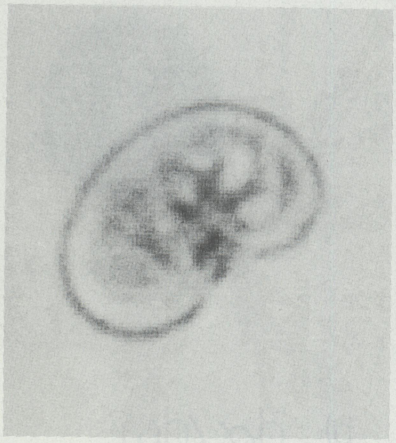


Figure 6. Reconstructed images of a cylindrical specimen of olive oil made with a 12 mm diameter acoustoelectric receiver. Results of reconstrucing the apparent attenuation coefficient at 4 MHz are presented in panel a). The reconstruction shown in panel b) is based on the frequency dependence of the attenuation estimated from the slope of a least squares line fit to the attenuation versus frequency data measured at 0.5 MHz intervals over the range 3.5 to 7.0 MHz.

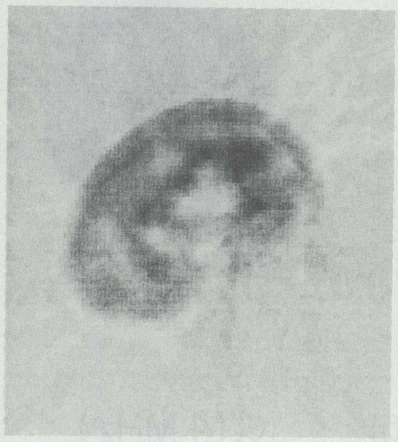
The potential application of the methods described above to quantitative imaging of soft tissue is illustrated in Figure 7. Images of an excised dog kidney are presented. A photograph of the approximate plane imaged is shown in panel a). A reconstructed image of the apparent attenuation at the frequency 5 MHz is presented in panel b). A reconstructed image based on the slope of the attenuation versus frequency curve over the range 5.0 to 6.5 MHz is presented in panel c). The acoustoelectric receiver with 14 mm lateral dimension and the focused transmitting transducer were used in these measurements. The reconstruction at a single frequency shown in panel b) exhibits pronounced edge artifacts. The reconstruction based on the frequency dependence of the attenuation shown in panel c) illustrates the elimination of frequency independent artifacts arising from reflection and refraction. The slope reconstruction of panel c) was accomplished with the use of only four frequencies spaced at 0.5 MHz intervals over the range 5.0 to 6.5 MHz.



a)



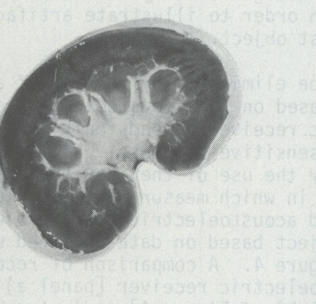
b)



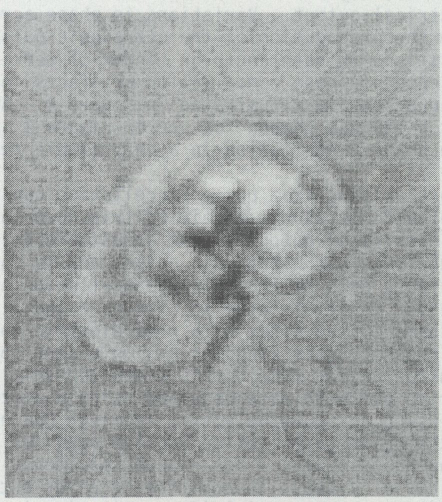
c)

Figure 7. Images of an excised dog kidney. Panel a) is a photograph of the approximate plane imaged. The image in panel b) is the reconstruction of the apparent attenuation coefficient at 5 MHz. The image in panel c) is the reconstruction of the slope of the attenuation versus frequency estimated from the slope of a least squares line fit to measurements carried out at 5.0, 5.5, 6.0, and 6.5 MHz.

The use of certain frequency independent parameters associated with reflection and refraction to generate an image is illustrated in Figure 8. Reflection effects are associated with variations in the acoustic impedance and refraction effects are associated with variations in the velocity of sound. The specimen imaged was the same excised dog kidney shown in Figure 7. A photograph of the approximate plane imaged is presented in panel a) of Figure 8. The reconstruction presented in panel b) is based on the (zero frequency) intercept of a least squares line fit to the measured attenuation versus frequency data. The extrapolation of this line to zero frequency is based on the approximately linear relationship between ultrasonic attenuation coefficient and frequency exhibited by soft tissue. Reconstructions based on zero frequency intercepts may be useful in studies requiring good border definition.



a)



b)

Figure 8. Images of an excised dog kidney. Panel a) is a photograph of the approximate plane imaged. The image in panel b) is the reconstruction of the (zero frequency) intercept of the least squares line fit to the apparent attenuation over the range 5.0 to 6.5 MHz.

IV. DISCUSSION

The present work was motivated by reports from this and other laboratories that quantitative indices based on ultrasonic attenuation may be capable of differentiating normal and pathological tissue.[1,2] Results of investigations from our laboratory summarized in Figure 1 suggest that the frequency dependence of the ultrasonic attenuation may be potentially useful in differentiating normal, ischemic, and scarred myocardium.[3,4] Computed ultrasonic tomography would appear to be a means of providing local, quantitative values of ultrasonic parameters in a format suitable for tissue characterization. The Mayo group demonstrated the principle of reconstructions based on both ultrasonic attenuation and velocity, and identified specific methodological problems.[11,12] Subsequent papers by the groups from Mayo,[13] General Electric,[14] and Colorado[15] have further demonstrated the potential of the technique.

Emphasis in the present work is on those aspects of the method which are dominated by front-end limitations, i.e., by the propagation and detection of the ultrasonic signal. The combined effects of phase cancellation, refraction, and reflection give rise to significant artifacts in attenuation tomography. Although post-processing of data containing these artifacts may improve image quality, this approach may not be consistent with the goal of identifying tissue pathologies on the basis of quantitative indices. The requirement for quantitative results stimulated our interest in front-end improvements.

The dominant artifact in ultrasonic attenuation tomography appears to be phase cancellation effects that are associated with the use of piezoelectric receiving transducers. Substantial errors in estimates of the apparent attenuation of myocardial tissue determined using piezoelectric receivers are demonstrated in Figures 2 and 3. Since the electrical output of a piezoelectric receiver is proportional to the instantaneous integral of the pressure across the face of the receiver, any variation in phase of the received signal results in a degraded electrical output. Variations in phase across the face of the receiver result from the propagation of the ultrasonic pulse through tissue because of small variations in pathlength or index of refraction. Some consequences of the presence of phase cancellation artifacts in reconstructions based on apparent attenuation are illustrated in Figure 4. A homogeneous, cylindrical specimen of castor oil was chosen for use in Figure 4 in order to illustrate artifacts arising from phase cancellation in a very simple test object.

Errors arising from phase cancellation can be eliminated by the use of a newly developed ultrasonic receiving transducer based on the acoustoelectric effect. The electrical output of the acoustoelectric receiver depends only on the intensity of the received ultrasonic pulse, and is insensitive to phase. Elimination of artifacts arising from phase cancellation by the use of the acoustoelectric receiver is illustrated in Figures 2 and 3, in which measurements carried out on the same tissue sites with piezoelectric and acoustoelectric receivers are compared. The reconstructed image of a simple test object based on data obtained with the acoustoelectric receiver is presented in Figure 4. A comparison of reconstructions carried out with data obtained using a piezoelectric receiver [panel a) of Figure 4] and using an acoustoelectric receiver [panel b) of Figure 4] indicates the severity of the phase cancellation problem even for the case of a relatively simple, homogeneous object. For tissue specimens, which are inherently inhomogeneous, the use of a phase insensitive receiver such as the acoustoelectric device appears to be essential if quantitative images based on attenuation are required.

Additional errors in ultrasonic attenuation tomography may arise from the combined effects of refraction and reflection, as illustrated in Figures 5 and 6. A reduction in the magnitude of lateral refraction errors can be achieved by the use of a receiving transducer of relatively large lateral dimension. A comparison of results obtained with an 8 mm and a 14 mm lateral dimension acoustoelectric receiving transducer is presented in Figure 5. Resolution need not be compromised by the use of a relatively large area receiving transducer provided that a relatively well collimated ultrasonic beam is provided by the transmitting transducer. Reflection effects can be minimized, and refraction effects further reduced, by the use of an approach based on the frequency derivative of the attenuation. The attenuation coefficient of soft tissue exhibits an approximately linear dependence on frequency in the low megahertz range. In contrast, reflection and refraction effects are substantially independent of frequency. Thus reconstructions based on the slope of a least squares line fit to the apparent attenuation measured over a modest range of frequencies provide images which are notably improved over single frequency images, as illustrated in Figure 6.

The use of the phase cancellation insensitive acoustoelectric receiver to image soft tissue in vitro is illustrated in Figures 7 and 8. A comparison of the photograph of the approximate plane imaged and the reconstructed images indicates that reconstructions based on attenuation are capable of providing reasonably good resolution of anatomical detail. In previous work from this laboratory values of the slope of the attenuation taken from reconstructed images of excised dog hearts were demonstrated to agree quantitatively with independent measurements of the slope of the attenuation in dog myocardium.[10] Thus reconstructions based on estimates of the frequency dependence of the attenuation obtained using the acoustoelectric receiver appear to hold promise for tissue characterization based on quantitative indices.

REFERENCES

Supported in part by NIH grants HL19537 and RR00396, and NSF grant APR77-09776. Contributions by D. M. Kramer and T. Shoup are gratefully acknowledged. Pranoat Suntharothok-Priesmeyer was responsible for production of the text and illustrations.

[1] Proceedings of the First International Symposium on Ultrasonic Tissue Characterization, M. Linzer (editor), NBS Publication 453, (1977).

[2] Proceedings of the Second International Symposium on Ultrasonic Tissue Characterization, M. Linzer (editor), NBS (1978) (in press).

[3] "Ultrasonic Attenuation in Normal and Ischemic Myocardium," M. O'Donnell, J. W. Mimbs, B. E. Sobel, and J. G. Miller, Proceedings of the Second International Symposium on Ultrasonic Tissue Characterization, M. Linzer (editor) (1978) (to be published).

[4] "The Role of Collagen in the Ultrasonic Attenuation of Myocardial Tissue," J. G. Miller, M. O'Donnell, and J. W. Mimbs, Proc. Third International Symposium on Ultrasonic Tissue Characterization, M. Linzer (editor), NBS (1979).

[5] "The Use of Colloidal Carbon as a Tracer for Vascular Injury," R. S. Cotran, E. R. Suter, G. Majno, Vasc. Dis. 4, 197-127 (1967).

[6] "The 'no-reflow' Phenomenon After Temporary Coronary Occlusion in the Dog," R. A. Kloner, C. E. Ganote, R. B. Jennings, J. Clin. Invest. 54, 1496-1508 (1974).

[7] "Depressed Myocardial Creatine Phosphokinase Activity Following Experimental Myocardial Infarction in Rabbit," J. K. Kjekshus, B. E. Sobel, Cir. Res. 27, 403-414 (1970).

[8] "Collagen as a Determinant of Ultrasonic Attenuation in Myocardial Infarcts," M. O'Donnell, J. W. Mimbs, B. E. Sobel, and J. G. Miller, in Ultrasound in Medicine, Vol. 4 (1978), (in press).

[9] "Phase cancellation Effects: A Source of Attenuation Artifact Eliminated by a CdS Acoustoelectric Receiver," L. J. Busse, J. G. Miller, D. E. Yuhas, J. W. Mimbs, A. N. Weiss, and B. E. Sobel, in Ultrasound in Medicine, Vol. 3, D. White, ed. (Plenum Press, New York, 1977) pp. 1519-1535.

[10] "Phase Cancellation, Reflection, and Refraction Effects in Quantitative Ultrasonic Attenuation Tomography," John R. Klepper, Gary H. Brandenburger, L. J. Busse, and J. G. Miller, Proc. IEEE Ultrasonic Symposium 77CH 1264-1SU, pp. 182-188 (1977).

[11] "Algebraic Reconstruction of Spatial Distributions of Acoustic Absorption within Tissue from Their Two-Dimensional Acoustic Projections," J. F. Greenleaf, S. A. Johnson, S. L. Lee, G. T. Herman, and E. H. Wood, in Acoustical Holography, Vol. 5, New York, Plenum Press, pp. 591-603 (1974).

[12] "Algebraic Reconstruction of Spatial Distributions of Acoustic Velocities in Tissue from Their Time-of-Flight Profiles," J. F. Greenleaf, S. A. Johnson, W. F. Samayoa, and F. A. Duck, in Acoustical Holography, Vol. 6, New York, Plenum Press, pp. 71-90 (1975).

[13] "Quantitative Cross-Sectional Imaging of Ultrasound Parameters," J. F. Greenleaf, S. A. Johnson, R. C. Bahn, B. Rajagopalan, Proceedings IEEE Ultrasonics Symposium, 77 CH 1264-1SU, New York, pp. 989-995 (1977).

[14] "Reconstruction of Ultrasound Propagation Speed Distributions in Soft Tissue: Time-of-Flight Tomography," G. H. Glover and J. C. Sharp, IEEE Transactions on Sonics and Ultrasonics, Vol. SU-24, pp. 229-234 (1977).

[15] "Ultrasound Transaxial Tomography by Reconstruction," P. L. Carson, T. V. Oughton, and W. R. Hendee, in Ultrasound in Medicine, Vol. 2, New York, Plenum Press, pp. 391-400 (1976).

Discussion:

LECTURER: J.G. MILLER

q.
What is the sensitivity of your phase insensitive transducer?

J. MILLER:
It's substantially less sensitive than a conventional piezoelectric. In fact it has about the same sensitivity as the ultrasonovision that Dr. Mezrich talked about. We don't really know yet; it is too early in this business. If you ask me to compare it to other transducers then I should tell you one additional fact. Over the range of frequencies that we are interested in, which is roughly two to ten magahertz, the response characteristics of our receiver are really quite horizontal literally. In comparison a typical commercial piezoelectric over that same range has a shape like that. How bad is it? It is about two decades.

G.T. HERMAN:
Something puzzled me about your plots. The curves with your receiver were not only different qualitatively but quantitatively. The maximum of your measured integrated attenuation coefficient with your receiver, was below the minimum of the other one, so while qualitatively it is clear it is better, what can you say about which one is more correct quantitatively?

J. MILLER:
I can comment on that in some detail because of course we are very interested in the specific numerical values of attenuation. It appears very much as if our acousto-electric receiver gives you a value which is quite consistent with radiation pressure measurements which are completely phased insensitive and which represent a kind of gold standard for this kind of thing except of course that they operate on time scales of seconds instead of microseconds. Our receiver has a roughly microsecond-like response time. You raise an extremely important point. Many people say that when they get data, suppose that you have an attenuation coefficient versus frequency plot which you measured like that and then you say well, what we know is roughly a linear attenuation versus frequency, so this is some kind of noise; so let's do the best we can and fit some kind of straight line to it. Well, if you go ahead and measure that same site with our receiver as you correctly point out you'd almost surely find that a lower value was the right answer. You will find throughout the literature that the estimated values of attenuation are substantially high over what we think the correct values are. That is a discussion which as those of you in the ultrasound community know, is being actively pursued at Kit Hill's Lab, at Floyd Dunn's Lab and many other places. So it is not unique to St. Louis. But the concept is that this transducer may be a way to get numerically quantitative results. It is nice that the measured attenuation is lower in a sense that presumably you are overestimating some loss with conventional transducers.

G. WADE:
I am wondering if you have explored the possibility that the transducer might have an optic acoustic effect. Since it operates via the carrierelectrons in cadmium sulphide, it perhaps has an effect that if you shine light on it, it would be different than if you do not. Have you investigated that?

J. MILLER:
In fact, I did not mention this only because of the limited time. As Prof. Wade points out, the material cadmium sulphide can be photo excited to move characters from the valence to the conduction band. What we do in fact is to operate it in just the mode you suggested. We illuminate the crystal with an incandescent light source. Any wavelength shorter than the band gap which is roughly in the green will do. We adjust the intensity of the light incident on the crystal to just match the conductivity relaxation frequency. I'm sure you know and must have had in mind that's precisely the condition for maximum transfer. So we are operating in just the mode you suggest.