PHASE CANCELLATION, REFLECTION, AND REFRACTION EFFECTS IN QUANTITATIVE ULTRASONIC ATTENUATION TOMOGRAPHY

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ABSTRACT. This study was undertaken to evaluate the potential of computed ultrasonic tomography for providing local, quantitative values of an ultrasonic index of attenuation, on the premise that such an index may be capable of characterizing the state of tissue. The combined effects of artifacts arising from phase cancellation, reflection, and refraction are known to substantially degrade reconstructions based on attenuation. In this paper we carefully segregate the contribution of each of these sources of artifact, illustrating the results with single scan lines (i.e., projections) and two-dimensional reconstructions. Phase cancellation artifacts are eliminated by the use of an intensity sensitive ultrasonic receiving transducer based on the acoustoelectric effect. Reflection artifacts are minimized and refraction artifacts reduced by reconstructing the frequency dependence of the attenuation coefficient, estimated by measurements carried out at eight frequencies. Results of reconstructions of test specimens and excised dog hearts suggest that an approach based on the slope of the attenuation coefficient estimated over a range of frequencies using data obtained with a large diameter acoustoelectric receiver offers promise for quantitative attenuation tomography.

I. INTRODUCTION

The potential use of ultrasound for the non-invasive identification of tissue pathology provides the overall motivation for evaluating computed ultrasonic tomography as a means of providing local, quantitative values of ultrasonic parameters. Greenleaf and his collaborators demonstrated the principle of reconstructions based on both ultrasonic attenuation[1] and velocity[2] and identified significant methodological problems which are discussed below. Subsequent papers by the groups from Mayo,[3] General Electric,[4] and Colorado[5] have further demonstrated the potential of the technique, with primary emphasis on time-of-flight reconstructions which were demonstrated to be immune from some of the methodological problems that degrade the results of reconstructions based on attenuation. The present report addresses the potentials and limitations of reconstructions based on attenuation for characterizing the state of soft tissue. Emphasis 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. Specifically, the role of phase cancellation, reflection, and refraction effects in degrading the results of attenuation tomography is investigated. Artifacts arising from phase cancellation are eliminated by the use of an acoustoelectric receiver. Reflection effects are minimized and refraction artifacts reduced by reconstructing the frequency derivative of the attenuation coefficient.

II. MODEL FOR ULTRASONIC PROPAGATION EFFECTS

To apply the method of reconstruction from projections, results of measurements are appropriately linearized to yield the line integral along a ray between transmitter and receiver of the parameter of interest. In attenuation tomography the goal is to reconstruct local values of the ultrasonic attenuation coefficient α. Under idea conditions the local attenuation coefficients $\alpha(z)$ for an ultrasonic pulse propagating in the z-direction is defined by the equation

$$P(z) = P_0 \exp\left\{-\int_0^z \alpha(z')dz'\right\} \qquad (1)$$

where P_0 is the pressure amplitude at the transmitting transducer and $P(z)$ is the pressure amplitude at the receiving transducer. The logarithm of the received signal yields the line integral of the local attenuation coefficient in a form suitable for reconstruction.

The present work focuses on those features of the propagation and detection of the ultrasonic pulse which complicate the simple relationship expressed in Eq. (1) and thus degrade the results of reconstructive tomography based on attenuation. Specifically we examine three sources of artifact: i) phase cancellation, ii) reflection, and iii) refraction. Phase cancellation effects occur when an ultrasonic wave exhibiting distorted phasefronts is received by a spatially extended piezoelectric receiver. Since the electrical output of a piezoelectric receiver is proportional to the instantaneous integral of the pressure across the face of the transducer, any variation in phase of the received signal produces a degraded electrical output. Variations in the phase of the ultrasonic pulse across the face of the receiver usually result from propagation through media exhibiting variations in the index of refraction. Nevertheless, phase cancellation effects are purely instrumental in character since they arise because of the phase sensitive nature of piezoelectric receivers. Reflection losses refer to a reduction in the amplitude of the received ultrasonic pulse arising from impedance discontinuities encountered along a particular ray. The effect of reflections is to introduce a product of n non-unity transmission coefficients into Eq. (1), where n is the number of impedance discontinuities encountered between transmitter and receiver. The fact that these transmission coefficients depend upon the angle of incidence is significant, since each discontinuity is viewed from many angles in a tomographic scan. The most general definition of refraction refers to 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.

It is useful to generalize Eq. (1) in a formal way to include explicitly some of the complicating features of the problem,

$$P(z,\omega)=P_0(\omega)\exp\left\{-\int_0^z \alpha(z',\omega)dz'\right\}\Phi(\omega)\left(\prod_{i=1}^{n} T_i(z_i,\theta_i)\right)F(\vec{r},\omega) \ . \tag{2}$$

In Eq. (2) the term $\Phi(\omega)$ accounts for possible phase cancellation effects, with $\Phi(\omega)=1$ for waves with uniform phasefronts across the face of a piezoelectric receiver or for a wave with arbitrarily distorted phasefronts across the face of an acoustoelectric receiver. The product of transmission coefficients $T_i(z_i,\theta_i)$ accounts for reflection effects. The factor $F(\vec{r},\omega)$ represents the spatial character of the ultrasonic field in the vicinity of the receiver, including refraction effects.

For purposes of normalization, each scan line begins and ends with segments corresponding to paths containing only water between transmitter and receiver. Under these conditions the phase factor $\Phi(\omega)$ is unity, the term in the product of transmission coefficients is unity, and the factor accounting for the spatial character of the ultrasonic beam in the vicinity of the receiver takes on its reference value $F_0(\vec{r},\omega)$. The received signal under these reference conditions is thus

$$P_{ref}(z,\omega) = P_0(\omega)e^{-\alpha_0(\omega)z}F_0(\vec{r},\omega) \tag{3}$$

where $\alpha_0(\omega)$ is the attenuation coefficient of water. After logarithmic processing, the attenuation relative to that of water is obtained by subtracting the result obtained from the water path from the results obtained at all points along the scan line,

$$\begin{aligned}\ln P'(z,\omega) &\equiv \ln P(z,\omega) - \ln P_{ref}(z,\omega)\\ &= -\int_0^z \alpha'(z',\omega)dz' + \ln \Phi(\omega) + \sum_{i=1}^{n} \ln T_i(z_i,\theta_i)\\ &\quad + \ln[F(\vec{r},\omega)/F_0(\vec{r},\omega)] \ .\end{aligned} \tag{4}$$

Here $\alpha'(z',\omega)$ is the attenuation coefficient measured with respect to that of water, i.e., $\alpha'(z',\omega) \equiv \alpha(z',\omega) - \alpha_0(\omega)$. In Eq. (4) the line integral of the attenuation coefficient measured relative to that of water is the quantity upon which the reconstruction is to be based. The second, third, and fourth terms on the right hand side of Eq. (4) represent sources of error arising from phase cancellation, reflection, and refraction, respectively. One goal of the present study was to isolate the contribution of each of these three sources of error.

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 transducers.[6] Since the factor $\Phi(\omega)$ involves the integral of the instantaneous pressure across the face of the piezoelectric receiver, phase cancellation effects are especially severe when large diameter receivers are used. Some reduction in phase cancellation effects was achieved in previous studies by the use of a small diameter receiver. 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. In the present study, the contribution of phase cancellation artifacts to reconstructive tomography based on attenuation is evaluated by a comparison of results obtained with piezoelectric and acoustoelectric transducers.

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). We are thus led to consider the results of measurements made at two frequencies ω_1 and ω_2.[7] If the measurements are processed to yield results in the form of Eq. (4) and subsequently subtracted, the frequency independent transmission coefficients cancel leaving

$$\begin{aligned}\ln P'(z,\omega_1) - \ln P'(z,\omega_2) &= -\int_0^z [\alpha'(z',\omega_1) - \alpha'(z',\omega_2)]dz'\\ &\quad + \ln[F(\vec{r},\omega_1)/F_0(\vec{r},\omega_1)] - \ln[F(\vec{r},\omega_2)/F_0(\vec{r},\omega_2)] \ .\end{aligned} \tag{5}$$

This approach also results in reduction of artifact arising from lateral refraction. To the extent that the field patterns are similar at frequencies ω_1 and ω_2, the last two terms of Eq. (5) partially cancel since the extent of lateral shift is independent of frequency. Some artifact remains because the path traveled through refractive media deviates from the straight line path assumed in the reconstruction algorithm.

III. EXPERIMENTAL METHODS

The mechanical scanning apparatus consisted of a carriage for linear motion supported from above in a mounting that permitted rotation about a central axis. Transmitting and receiving transducers were mounted coaxially on arms that extended down from the carriage into a water bath in which the specimen was mounted. The separation between transmitting and receiving transducers was usually 18.5 cm. 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 1.6 or 1.4 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.

The tomographic measurement system operates under control of a computer which provides commands for mechanical scanning, monitors the analog electronics and provides commands for adjustment of analog circuitry, manages data acquisition, and performs the reconstruction. All reconstructions were carried out by the filtered back projection technique using a standard sharp cutoff filter.[8] No smoothing or other post-processing of the data was carried out.

To generate the ultrasonic pulse, the output of a continuous wave rf oscillator was gated phase coherently to generate a narrow band pulse of approximately 3 μsec duration. The frequency of the voltage tunable cw oscillator was selected under computer control and was in the range 3 to 7 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 (ENI model 240L). 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 autoranging scheme, the amplitude of the rf pulse applied to the transmitting transducer was adjusted over a wide dynamic range (>50 dB) to compensate for variations in the total attenuation. Thus the receiver was required to operate only over a modest dynamic range (~20 dB) to achieve an overall dynamic range of more than 70 dB.

The output of the receiving transducer was amplified and the leading edge of the received signal was used to trigger circuitry in the receiver system. An appropriate segment of the received signal was gated into a peak detector, the output of which was sampled and fed to the computer. Using this signal, the computer adjusted the control voltage to the preamplifier, thus varying the amplitude of the transmitted ultrasonic pulse to bring the received signal into the linear range of the detector. Once the computer verified that a valid signal has been received, the value of that signal and the value of the corresponding transmitter gain level were stored as data. At each point in the scan, data at as many as 8 frequencies were collected, with the computer selecting each frequency sequentially by providing the appropriate command to the voltage tunable cw oscillator.

In order to permit the collection of data over a range of frequencies and to assure a reasonably well collimated beam for all frequencies used, a broadband, planar transmitting transducer of 12.7 mm diameter (Panametrics V309, nominal center frequency 5 MHz) was used for all investigations. Thus data taken with different receiving transducers or data processing combinations were obtained under identical conditions with respect to the generation of the ultrasonic pulse and its propagation through the specimen of interest. Data are presented for three broadband, planar ultrasonic receiving transducer arrangements: i) a 12 mm diameter piezoelectric, ii) a 12 mm diameter acoustoelectric, and iii) 3 mm diameter piezoelectric. The 12 mm piezoelectric and 12 mm acoustoelectric receiver signals were obtained from the same single crystal 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.[6] 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. Selection of either the piezoelectric or the acoustoelectric 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. Data are also presented using a commercially available 3 mm diameter piezoelectric receiver (Panametrics V3066, nominal center frequency 5 MHz).

Objects scanned and reconstructed in these investigations included approximately cylindrical specimens of castor oil and olive oil in negligibly thin-walled rubber containers. Excised dog hearts were selected as biological test specimens. Physical properties of the specimens investigated are summarized in Table I.

IV. EXPERIMENTAL RESULTS

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. As indicated in Table I, the characteristic impedance and index of refraction of castor oil are nearly equal to those of water, so the magnitude of reflection losses is small and lateral refraction effects are modest. Furthermore, the ultrasonic attenuation coefficient of castor oil is substantial, reducing the relative importance of reflection and refraction losses. In Figure 1a the results of scanning the castor oil specimen at a frequency of 4 MHz using the 12 mm diameter piezoelectric receiver are illustrated. The reconstruction exhibits a very pronounced ring artifact, reflecting a substantial overestimation of the attenuation near the edges of the object. This overestimation is easily identified in the projection line shown above the reconstruction in Figure 1a. Results obtained under otherwise identical conditions using the 12 mm diameter acoustoelectric in Figure 1b are notably more uniform than those in Figure 1a, manifesting the elimination of phase cancellation artifacts. A comparison of the projection lines shown above the reconstructions in Figure 1 provides a qualitative view of the improvements achieved by replacing a piezoelectric receiver with an acoustoelectric receiver. (The vertical scales for the projection lines shown in all figures were adjusted to make the scan lines approximately equal in height. Similarly, the full gray scale is used in each panel, with black representing the most attenuating regions and white representing the least attenuating regions in each reconstruction.)

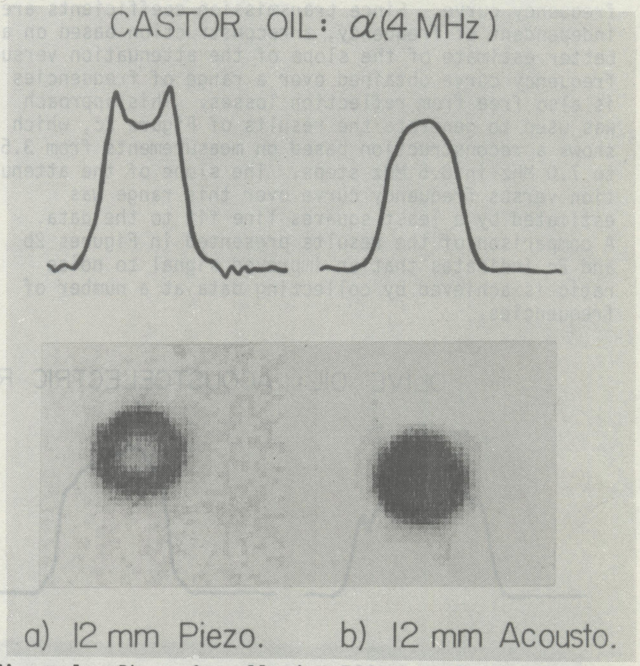

Figure 1. Phase Cancellation Effects: Projections and reconstructions of the attenuation of a cylinder of castor oil from measurements obtained using a) a piezoelectric receiving transducer and b) an acoustoelectric receiving transducer.

Table I. Material Properties of Test Objects

Substance	Density (gm/cm^3)	Index of Refraction (relative to water)	Characteristic Impedance (relative to water)	α(5 MHz) (cm^{-1})	$d\alpha/d\omega$(3-7 MHz) (cm^{-1} MHz^{-1})
Water	1.00	1.00	1.00	-	-
Castor Oil	0.96	1.01	0.97	1.4	0.6
Olive Oil	0.92	0.97	0.89	0.22	0.1
Dog Left Ventricle	1.06	1.05	1.11	0.27	0.07

The effects of reflection losses were studied by reconstructing the image of a specimen of olive oil contained in a 3.5 cm diameter condom. Olive oil is markedly less attenuating than castor oil (see Table I), making the effects of reflection losses relatively more significant. The impedance discontinuity at an olive oil--water interface is larger than at a castor oil--water interface, further increasing the importance of reflection effects. To eliminate artifacts arising from phase cancellation, the acoustoelectric transducer was used. Refraction artifacts were minimized by the use of a relatively large aperture (12 mm diameter) receiver, so that little of the ultrasonic beam was lost as a result of small lateral displacements. To illustrate the role of reflection effects, three reconstructions and corresponding projection lines of the olive oil specimen are presented in Figure 2. In Figure 2a the results of reconstructing the attenuation coefficient at the frequency 4 MHz are presented. A marked halo artifact is visible in the reconstruction and its origins can be identified in the projection line. Results of the model presented in Section II suggest that artifacts arising from reflection can be eliminated by reconstructing the difference in the attenuation coefficient at two frequencies. Figure 2b illustrates the results of reconstructing the difference between the attenuation coefficient at 6 MHz and the attenuation coefficient at 4 MHz. The reconstructed image of Figure 2b is more homogeneous than that of Figure 2a, manifesting the elimination of the halo artifact. The difference between the attenuation coefficients at two frequencies represents a crude approximation to the slope of the attenuation versus frequency curve. Since transmission coefficients are independent of frequency, a reconstruction based on a better estimate of the slope of the attenuation versus frequency curve obtained over a range of frequencies is also free from reflection losses. This approach was used to generate the results of Figure 2c, which shows a reconstruction based on measurements from 3.5 to 7.0 MHz in 0.5 MHz steps. The slope of the attenuation versus frequency curve over this range was estimated by a least squares line fit to the data. A comparison of the results presented in Figures 2b and 2c indicates that an improved signal to noise ratio is achieved by collecting data at a number of frequencies.

Artifacts which result from small lateral shifts in the position of the ultrasonic beam due to refraction were reduced in the results presented in Figure 2 by the use of a relatively large aperture receiving transducer. This approach for reducing refraction effects is feasible only with the use of a receiver which is insensitive to the effects of phase cancellation. This point is illustrated in Figure 3 for the olive oil specimen, where three receiving transducer arrangements were used under the same conditions as those in Figure 2. The fact that phase cancellation effects preclude the use of a large diameter piezoelectric receiver to minimize refraction effects can be seen by comparing the results of Figure 3a obtained with a 12 mm piezoelectric receiver with those of Figure 3c obtained with a 12 mm acoustoelectric receiver. Previous studies from our laboratory indicated that artifacts arising from phase cancellation effects can be reduced by the use of a small aperture receiver. Results obtained using a 3 mm diameter piezoelectric receiver are presented in Figure 3b. A comparison of Figures 3a and 3b serves to illustrate the trade-off between phase cancellation effects and effects arising from lateral beam shifts associated with refraction. The results of Figure 3b exhibit a substantial overestimation of the attenuation at the edges of the specimen, presumably due to the fact that a substantial fraction of the beam missed the small aperture receiving transducer as a result of refraction. Thus any decrease in phase cancellation effects achieved by the use of the small diameter receiver appears to be overshadowed by the corresponding increase in artifacts arising from refraction.

The methods developed to cope with phase cancellation, reflection, and refraction were used to image an excised dog heart in a plane approximately 2.5 cm from the apex. Data was taken with the 12 mm acoustoelectric receiver at 7 frequencies spaced at 0.5 MHz intervals over the range 4 to 7 MHz. After the scan, a 12 mm thick segment corresponding to the region imaged was sliced from the heart and photographed to reveal the apical and basal views shown in Figure 4a. In Figure 4b a reconstruction based on data at 6.5 MHz is presented. Because the attenuation at this frequency is moderately high, artifacts due to reflection have only a modest effect on image quality although they do

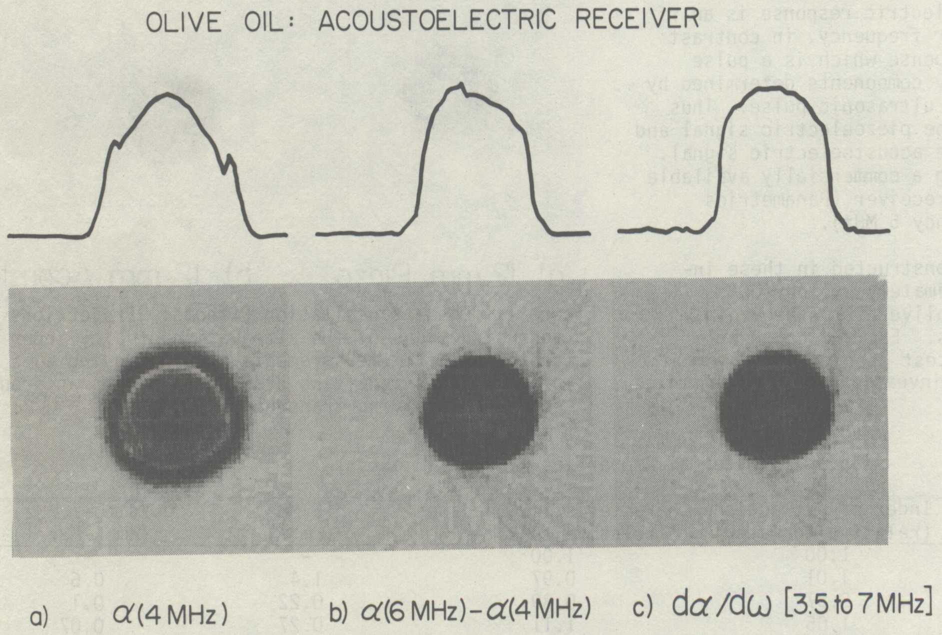

Figure 2. Reflection Effects:
a) the attenuation coefficient at a single frequency,
b) the difference between the values of the attenuation coefficient at two frequencies, and
c) the slope of the attenuation versus frequency curve.

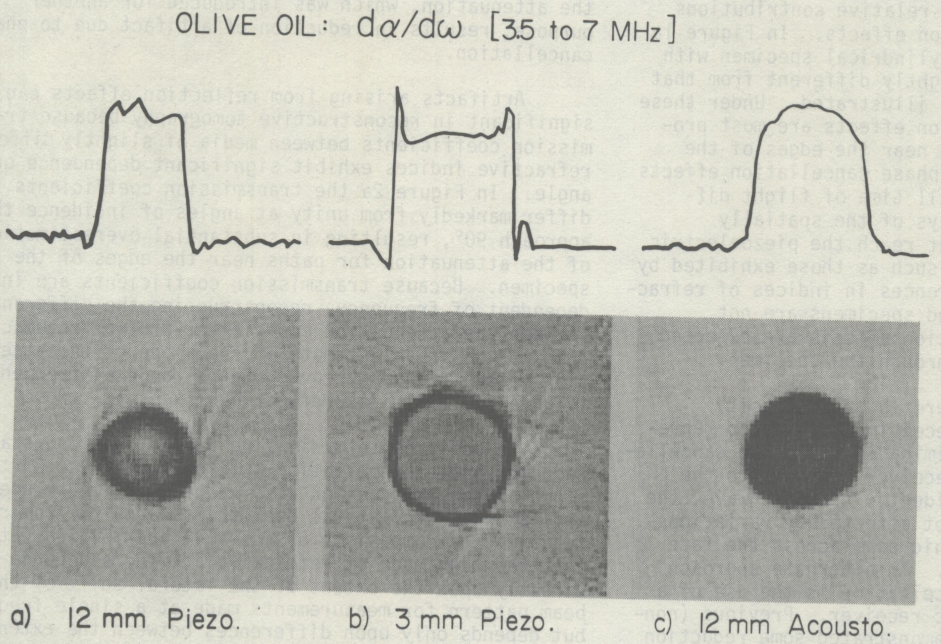

Figure 3. Refraction and Phase Cancellation Effects: slope of the attenuation using
a) a 12 mm piezoelectric receiver
b) a 3 mm piezoelectric receiver, and
c) a 12 mm acousto-electric receiver.

invalidate the quantitative results. In figure 4c a reconstruction based on the slope of the attenuation from 4 to 7 MHz is presented. Values of the slope of the attenuation in the region of the left ventricle of the reconstruction shown in Figure 4c were compared with corresponding values obtained in previous measurements on about 250 regional sites from dog left ventricle using a different technique.[9] Results from Figure 4c indicate a mean value of the slope of 0.08 cm^{-1} MHz^{-1} with a standard deviation of 0.02 cm^{-1} MHz^{-1} ($N = 54$), which compares favorably with the mean value 0.073 cm^{-1} MHz^{-1} and standard deviation 0.012 cm^{-1} MHz^{-1} for the slope of the independent measurements computed over the range 4 to 7 MHz.

V. DISCUSSION

Motivation for the present work was provided by indications that quantitative indices based on ultrasonic attenuation may be capable of differentiating normal and pathological tissue. 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. Previous investigations indicated that the combined effects of phase cancellation, reflection, and refraction gave 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. Segregating effects arising from phase cancellation, reflection, and refraction is a necessary first step in a program to evaluate techniques for reducing errors from these sources.

To illustrate phase cancellation effects that arise from the use of piezoelectric receiving transducers,

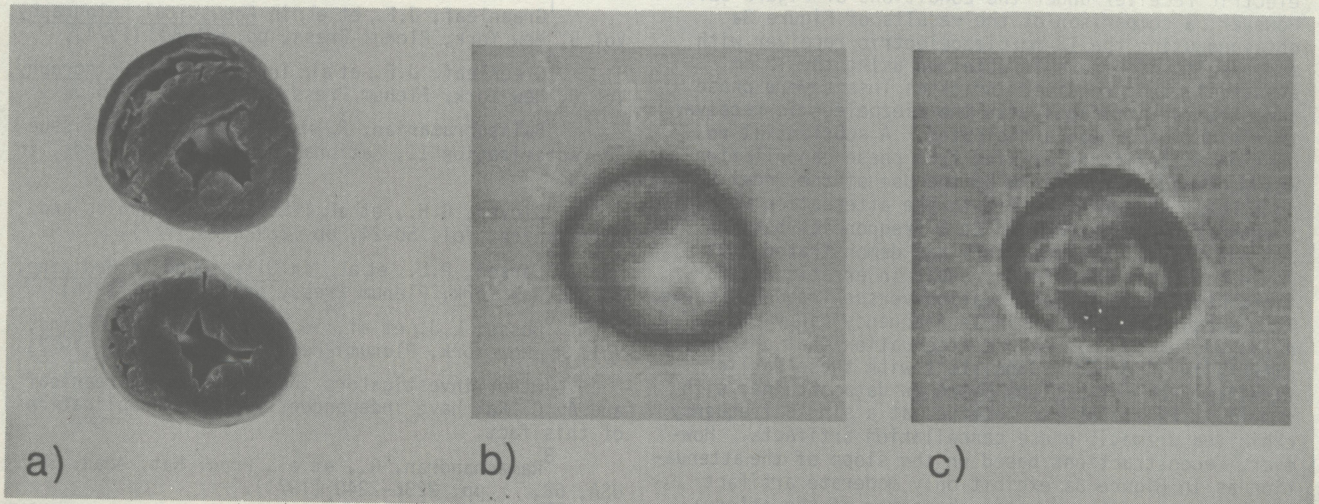

Figure 4. Cross sections of an excised dog heart showing a) photographs of apical and basal views of the imaged region, b) a reconstruction of the attenuation coefficient at 6.5 MHz using the 12 mm acoustoelectric receiver, and c) reconstruction of the slope of the attenuation from 4 to 7 MHz using the 12 mm acoustoelectric receiver.

a test specimen (castor oil) was selected with physical properties that minimize the relative contributions from reflection and refraction effects. In Figure 1a the case of a homogeneous, cylindrical specimen with index of refraction only slightly different from that of the surrounding medium is illustrated. Under these conditions, phase cancellation effects are most pronounced for ultrasonic paths near the edges of the objects. The origin of the phase cancellation effects in Figure 1a 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.

As illustrated in Figure 1b, an intensity sensitive acoustoelectric receiving transducer represents one approach to the elimination of phase cancellation effects. Since this receiver responds to the momentum carried by the incident ultrasonic wave, the acoustoelectric output is not affected by variations in the phase of the ultrasonic beam across the face of the receiving transducer. An alternate approach to the problem of phase cancellation is the use of a small diameter piezoelectric receiver. Previous (non-tomographic) studies have demonstrated some reduction in artifacts with the use of a small aperture receiver. The results shown in Figure 3b indicate that the use of a single small diameter piezoelectric receiver to reduce phase cancellation is impractical in tomography because lateral displacements of the beam due to refraction produce substantial artifacts with this configuration. An array of small diameter piezoelectric receivers would reduce the susceptibility to refraction and provide partial reduction of phase cancellation effects. Provided that acoustoelectric receivers can be improved to provide adequate sensitivity, the complete elimination of phase cancellation effects with an acoustoelectric receiver of sufficiently large diameter to minimize refraction effects would appear to be possible.

Reduction of phase cancellation effects that arise from the use of piezoelectric receivers has been achieved using another approach, as illustrated in Figure 3a. Substantial artifact from phase cancellation would be expected at all points along a scan line with the use of the relative large aperture piezoelectric receiver under the conditions of Figure 3a. However, a comparison of the results of Figure 3a obtained using the 12 mm piezoelectric receiver with the results of Figure 3c obtained using the 12 mm acoustoelectric receiver indicates less severe phase cancellation artifacts with the piezoelectric receiver than might have been anticipated. A substantial reduction in artifacts arising from phase cancellation was achieved in Figure 3a by the use of the index based on a least squares line fit to the attenuation versus frequency curve determined at 8 frequencies between 3.5 and 7 MHz. Previous work has demonstrated that phase cancellation effects result in erratic alterations in the expected attenuation versus frequency curve. Thus measurements at a single frequency can yield estimates of the ultrasonic attenuation that are substantially in error. Consistent with these earlier findings, reconstructions based on data obtained with the 12 mm piezoelectric receiver at a single frequency exhibited dramatic phase cancellation artifacts. However, reconstructions based on the slope of the attenuation as in Figure 3a exhibit only moderate artifact because some of the erratic character of the attenuation versus frequency curve is averaged out by the process of estimating the slope from a least squares fit to the data over a range of frequencies. Thus the technique of reconstructing the frequency dependence of the attenuation, which was introduced for another purpose, results in reduction of artifact due to phase cancellation.

Artifacts arising from reflection effects can be significant in reconstructive tomography because transmission coefficients between media of slightly different refractive indices exhibit significant dependence on angle. In Figure 2a the transmission coefficients differ markedly from unity at angles of incidence that approach 90°, resulting in substantial overestimation of the attenuation for paths near the edges of the specimen. Because transmission coefficients are independent of frequency, reconstructing the difference between the attenuation coefficient at two frequencies (Figure 2b) or an estimate of the slope of the attenuation versus frequency curve over a range of frequencies (Figure 2c) minimizes reflection losses.

In addition to minimizing reflection losses and reducing phase cancellation artifacts, the use of the slope of the attenuation also reduces the magnitude of artifacts due to lateral beam shifts arising from refraction. A comparison of Eqs. (4) and (5) indicates that the magnitude of refractive artifacts depends directly upon the extent of the lateral shift of the beam pattern for measurements made at a single frequency but depends only upon differences between the extent of lateral beam shift experienced at one frequency and that experienced at the second frequency for measurements carried out at two frequencies. A similar result applies for measurements made over a range of frequencies. Thus the magnitude of refractive effects for reconstructions based on the slope of the attenuation such as those illustrated in Figure 3 are less than would be anticipated for reconstructions based on results obtained at a single frequency.

In summary, results of this study suggest that an approach based on the slope of the attenuation coefficient estimated over a range of frequencies using data obtained with a large diameter acoustoelectric receiver offers promise for quantitative attenuation tomography.

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D. Dick - University of Colorado - I was very interested in your castor oil reconstruction with the conventional transducer and the power sensitive transducer. The predicted change was at the edge where you would have large edge effects that would be cancelling out and certainly your profile showed that. Yet the reconstructed image, both edges were black and the only thing that changed was the center which was clear or nearly clear in the conventional transducer became black in the power sensitive transducer. It seemed like the center changed more than the edges. Would you comment on that?

A: Well, actually, the edge changed more than the center. As I said in the talk, all reconstructions were shown in the full dynamic range. In other words, the lowest number in each reconstruction was given the bottom, the highest number in each reconstruction was given the top gray level black beam high attenuation. It just turns out that the reconstruction with the acoustic electric transducer is so uniform that it all appears as black in the background it all appears as white. The reconstruction with piezoelectric transducer, the edges were significantly greater attenuation than you might expect.

H. Barrett - University of Arizona - Could you give us some more details on your acustoelectric transducers. In particular, what material, what thickness, and what sensitivity they have relative to the piezoelectric transducer?

A: The question was, could I give more specifics about the acoustoelectric transducer. The acoustoelectric transducer is a single crystal of cadmium sulfide plated with indian electrodes. It's approximately a 1/2-inch apperture, it's about 1.4 centimeters thick and its modulator was the light source to produce induction electrons to maximize the acoustoelectric effect. Its sensitivity should probably be compared to a broad band piezoelectric transducer. In that case, for the same apperture transducer, its sensitivity is about 20 dB, less than that, that you would get with a piezoelectric transducer. However, you pick up so much in terms of right signal and no base cancellation that we don't think that's a big problem.

S. Wang - IBM Research Labs. - I think this was very interesting accumulating. We are also doing a similar study along this line and I wanted to give you some of our ideas. By using a reconstruction, it is the function ot the beam width, equipment the beam width has been changed, and also the banding of the acoustic has it also varies with the frequency. I wonder if you have ever in your reconstruction, have you ever looked into that area?

A: Let me first answer that first question which was about beam width, effective beam width. Certainly, when we take the slope of the attenuation versus frequency, the different beam width for the different frequencies introduces some error, particularly at the edge of the object in reconstructions mainly because the lower frequencies appear to see a larger object than the higher frequencies. And that shows up in our reconstructions as a small white disc or a small white ring around the object. There's also critical angle reflection that occurs and that may make the object smaller also.

My second question is, have you ever looked at the frequency dependent velocity reconstruction at the time of the flight?

A: We have not looked at it.

R. Kuc - Columbia - The slope on attenuation with frequency, I take it, in your talk, was done at several discreet frequencies. Have you tried doing that in single pulses?

A: We did some early work where we used a broad band pulse and did spectral analysis of the pulse. That works. It's more involved. It could not be done with the acoustoelectric transducer, however, because the acoustoelectric signal is a video signal as opposed to an RF signal. And there's no frequency information for changing that so we used gated RF to gain the frequency information that we need to determine the slope.