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Measurement Issues in Quantitative Ultrasonic Imaging

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Abstract* - Ultrasonic imaging of "flesh and blood" is vulnerable to the natural variability of these media: the speed of sound is not known, not constant, and not amenable to calibration using simply shaped manufactured samples. When images of high dimensional accuracy are needed, as for image-guided surgery, the "average" or "typical" values used in diagnostic ultrasound may not be good enough. In this paper we identify the main sources of uncertainty, and we suggest and model experimental approaches to in situ calibration.

I. INTRODUCTION

Applications recently proposed for ultrasonic imaging, for example, in "image guided surgery" [1][2] and in the custom design of wheelchair cushions [3], demand much higher dimensional *measurement* accuracy than is demanded by any of the traditional diagnostic *visualization* applications of ultrasound [4]. To render dimensionally accurate ultrasonic images it is necessary either to have a priori, or alternatively to acquire as an integral part of the measurement process, accurate knowledge of the speed of sound in the skin, fat, muscle, bone, and other living media whose interfaces spawn echoes. To correctly interpret the intensities of the echoes it is also necessary to have accurate knowledge of the acoustic impedances of these media.

In contrast with engineering materials, whose acoustic properties can usually be measured off-line on manufactured artifacts of known dimensions, for dimensionally accurate imaging of living tissue there is no apparent alternative to in situ calibration. In this paper we pose the *quantitative ultrasonic imaging problem*, and we propose and model some apparently workable approaches to in situ measurement of the speed of sound and, where appropriate, its gradient. We defer to future papers discussion of acoustic impedances, and thus of signal intensities. Because a method's accuracy cannot be modeled in the absence of knowledge of the signal and noise intensities, quantitative treatment of this subject is also deferred.

Section II is a description of the problem, including an outline of our approach and its scope. Section III describes the underlying assumptions of the following work and places it in the context of the anticipated applications. Section IV reviews the basic principles of time-of-flight based ranging, and presents a simple differential approach to measuring the speed of sound in situ. Section V approaches this subject with additional sophistication; it discusses an oblique path geometry for single- and multi-layer characterization, coping with gradients in the speed of sound, and coping with non-parallel layers. Section VI briefly draws some conclusions and outlines anticipated future work.

II. PROBLEM AND APPROACH

At the cutting edge of their instruments, practitioners in the emerging field of image guided surgery would like to have navigational accuracy under 1 mm, and even better endpoint precision. Related application areas, e.g., our own collaboration with clinical practitioners who design custom seat cushions to prevent pressure sores in wheelchair-bound patients, make less stringent but in principle quite similar demands on the accuracy of ultrasonic dimensional measurements and image generation based on those measurements.

Unfortunately the uncertain speed of sound in an individual patient's living skin, fat, muscle, etc, frustrates the surgeon's (and the seat cushion designer's) desire for these levels of accuracy and precision. The solution is to find measurement techniques that do not require a priori knowledge of the acoustic properties of the media traversed. That is, we need to find experimental techniques that measure the speed of sound in situ, through the very regions of "flesh and blood" whose dimensions we seek. These techniques must function though we are denied access to media samples in manufactured shapes, denied access to both sides, and denied all access except through whatever overlaying and underlaying strata constitute the natural structures.

We proceed by separating the general problem into three measurement "modules" for each of which we propose an apparently robust experimental solution:

1. in situ speed of sound and layer thickness measurement given multiple parallel homogeneous layers (Section V, A. and B.);
2. coping with a speed of sound gradient in a layer; (Section V, C.)
3. coping with a tapered layer (Section V, D.).

Present time and space limitations require us to defer assembling the modules into a comprehensive system. A future integrated system, with experimental confirmation, will satisfy the medical imaging requirements posed herein, as well as corresponding requirements for nondestructive inspection of engineering structures where the materials are analogously unavailable for off-line measurement of their acoustic properties.

III. ASSUMPTIONS AND CONTEXT

Ultrasonic imaging with accurate dimensional calibration requires mechanically accurate scanning capability for the raster, and accurately calibrated ultrasonic ranging capability to each surface of interest. The "surfaces of interest" are the interfaces between layers of different but nominally homogeneous materials, e.g., skin, fat, muscle, and bone. A viable system will also need to detect and compensate for inhomogeneities within layers.

Current mechanical, optical, magnetic, etc, tracking technologies are assumed in this paper to be of sufficient accuracy and precision to meet the application's requirements for dimensional calibration of the raster.

When imaging multilayered *engineering* specimens, the materials comprising the individual layers can usually be characterized off-line as to, e.g., speed of sound, dispersion, attenuation vs frequency, acoustic impedance, etc, sufficiently well that accurate gauging is straightforward. Even when there is insufficiently detailed prior knowledge of these material properties, with man-made specimens it is often sufficient to obtain *relatively* accurate measurements, for example, the fraction of initial aluminum thickness lost in a small corroded spot on an airplane's skin, and accurate relative gauging is insensitive to the material properties.

In contrast, with *living anatomical* specimens the subject-to-subject variation in material properties is problematic, yet individual off-line characterization of these properties is obviously impossible. Even for an individual subject, when dimensional accuracy is a critical issue, the possibility needs to be considered that a nominal tissue type in fact has locally inhomogeneous acoustical properties, it has globally different acoustical properties in different parts of the body, and it has temporally variable acoustic properties due to diet, muscle tone, etc. Furthermore the possibility must be considered that the vital state of tissue, e.g., whether muscles are tense or relaxed, whether limbs or buttocks are mechanically loaded or unloaded, etc, may affect acoustic properties and thus the dimensional accuracy of ultrasonic images.

In the past these difficulties in principle have rarely been a practical impediment because the applications of anatomical ultrasonic imaging have been primarily diagnostic, requiring only qualitative or semi-quantitative dimensional accuracy, i.e., enough to allow the physician to assess the normalcy of anatomical structures, to observe the approximate location, size, and shape of organs, etc. Recently, however, the applications mentioned have been hampered by the need for dimensional accuracy and precision beyond any that can be expected from only "generic" speed of sound estimates.

IV. REVIEW OF THE BASIC TECHNIQUE

The basic single sided ultrasonic measurement is a recording, versus time, of multiple echo amplitudes. Each interface between two layers of different acoustic impedance spawns an echo. The time delays between the transmitted pulse and the first echo, and between successive echoes, combined with a priori knowledge of the speed of sound in each layer, gives the layer thicknesses. This is illustrated for one layer in Fig. 1. Multiple layers are handled straightforwardly providing the number and nature of up and down segments constituting each echo can be surmised[1]. For measured time t_i in layer i with speed of sound c_i the layer thickness is

$$z_i = c_i \cdot \frac{t_i}{2}$$

Images are built up by raster scanning of pencil sensors, by linear scanning of one dimensional array sensors, or in areal patches by two dimensional array sensors.

If modest dimensional distortions can be tolerated then nominal values can be used for the speeds of sound; these are tabulated in standard reference books for common engineering materials and for typical human

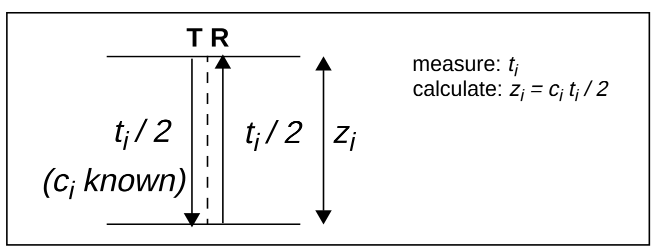

Fig. 1: Basic range measurement. Transmitter **T** and receiver **R** are displaced for clarity, but in practice may be one transducer.

1. Important signal processing issues, such as how to define time-of-flight when dispersion and frequency-dependent attenuation distort the reflected pulse shape, are omitted from the present discussion.

tissue types [5]. However individual differences, and additional fine details (such as the inhomogeneity and geometry issues discussed in subsequent sections, and complications of the sort mentioned in Footnote 1) all frustrate rendering highly accurate images.

A. Differential methods

Given a homogeneous layer of well defined mechanical properties (perfectly fluid, or perfectly elastic, or otherwise precisely characterized), simple differential measurements suffice, at least in principle, to measure both the thickness of the layer *and* the speed of sound in it. As illustrated in Fig. 2, changing the thickness of a layer slightly, by Δz_i, and measuring the change in echo time-of-flight, Δt_i, suffices to find the speed $c_i = 2\ \Delta z_i/\Delta t_i$, and thus the thickness $z_i = c_i\ t_i/2$.

It is, however, doubtful that this approach will be adequate in applications of interest to us. While skin, fat, and muscle layers can be depressed enough to make the method at first seem feasible, in the absence of complete mechanical models for these media we are at a loss to predict the distribution of the total compression among the various layers, how the densities and bulk moduli of the various layers may change (and thus how the speed of sound in them may change) due to incomplete fluidity, how compression may introduce directional dependencies into the speed of sound, etc.

It is thus not immediately obvious, given multilayered subjects of complex and incompletely known mechanical properties, that ultrasonic imaging with the accuracy demanded by the anticipated applications is actually possible. In the next section we outline approaches that appear to have sufficient promise that experimental verification of their utility is warranted (and is currently underway in our lab).

V. APPROACHES TO THE REAL PROBLEM

A. One homogeneous layer

The approach hypothesized in the previous section and illustrated in Fig. 2 attempts to measure the unknown speed of sound by measuring time-of-flight over a path that is under the experimenter’s control; we reject it because we are not confident that its application leaves unchanged the material properties, and thus the speed of sound we are trying to measure. A differential approach that is free of this concern is illustrated in Fig. 3. It involves two or more measurements over different oblique paths[2]. In each measurement i the speed of sound c and the layer thickness z are intertwined in:

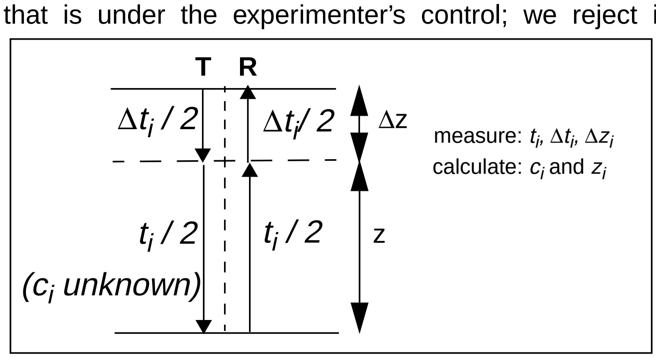

Fig. 2: Differential time-of-flight method.

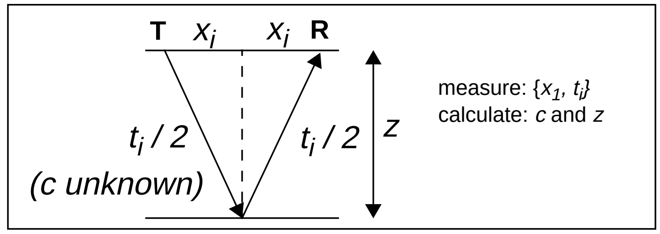

Fig. 3: Differential between two or more oblique paths.

$$c^2 \cdot \frac{t_i^2}{4} = z^2 + x_i^2$$

Their simultaneous solution when $i = \{1,2\}$ yields:

$$c = \frac{2\sqrt{x_i^2 + z^2}}{t_i}$$

$$z = \sqrt{\frac{x_2^2 \cdot t_1^2 - x_2^2 \cdot t_1^2}{t_2^2 - t_1^2}}$$

If $i>2$ a least-squares solution will optimize accuracy.

B. Several parallel homogeneous layers

The single-layer method of Fig. 3 is easily extended to multiple layers, as illustrated in Fig. 4. Selecting two values of x_1 and measuring the two corresponding values of x_2, the two speeds of sound c_1 and c_2, and the two depths z_1 and z_2 are measured[3]. The approach can be applied to an arbitrary number of layers. This tehcniques the mainstay of geoacoustics [6], where in,

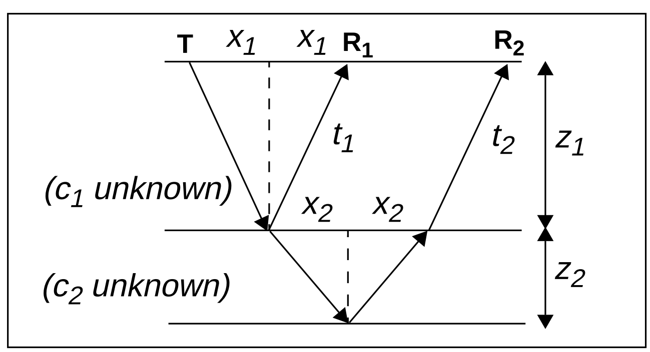

Fig. 4: Extending the method of Fig. 3 to multiple layers.

2. While diffuse reflection is also reported in the medical imaging literature, In the present paper we consider only specular reflection.
3. It is presumed that the signals at receivers $\mathbf{R_1}$ and $\mathbf{R_2}$ can be distinguished by their relative amplitudes.

e.g., oil prospecting, it is routinely required to characterize multiple complex rock layers (strata).

C. Parallel layers with a gradient

Snell's Law of refraction in the form $\sin\theta/c = k$, where k is a constant, holds even if the speed of sound c is a function of position within a medium. The angle θ, measured between the local tangent to the trajectory and the local gradient of c is then also a function of position,therefore the trajectory is curved[4]. A linear gradient, $c = c_0 + \alpha z$, where c_0 is the baseline speed, α is its gradient, and z is the depth, is illustrated in Fig. 5. The linear approximation to arbitrary gradients is particularly useful because the trajectories are then simply circular arcs; this result is well known in underwater acoustics [7]. Three transmitter/receiver separations suffice to measure c_0, z, and the three launch angles $\{\theta_1,\theta_2,\theta_3\}$ corresponding to the $\{x_1,x_2,x_3\}$. The radius of the circular arc is given by

$$R_i = -\frac{c_i}{\sin\theta_i}$$

where the index *i emphasize*s that the radius R_i, the speed of sound c_i, and the angle θ_i are all measured at the same point. The center of the arc is at distance R_i along the perpendicular to the trajectory on its concave side.

D. Nonparallel layers

When layers are tapered, as illustrated (in a two dimensional cross-section) in Fig. 6 the acoustic time-of-flight defines an elliptical locus to which the reflecting discontinuity is tangent. For each transmitter $\mathbf{T_i}$ - receiver $\mathbf{R_i}$ separation $2x_i$, we thus have the equation of an ellipse:

$$\frac{x^2}{a_i^2} + \frac{z^2}{b_i^2} = 1 \text{ where } a_i^2 = c^2 t_i^2 \text{ and } b_i^2 = a_i^2 - x_i^2$$

Usually only one physically reasonable line will be tangent to two such ellipses. Thus if c is known, two $\{x_i,t_i\}$ pairs fix the depth and slope of the reflecting plane. If necessary, an additional pair will resolve any ambiguity. When c is not known in advance, an additional pair is sufficient to find both c and the correct reflecting plane.

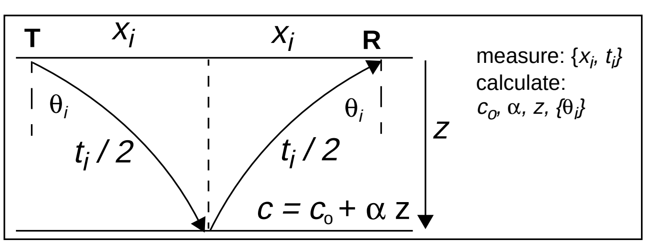

Fig. 5: Trajectories when the speed of sound is not constant.

4. If c is <u>continuous</u> there is refraction but no reflection.

VI. CONCLUSIONS AND FUTURE WORK

We have considered the necessity of integral in situ calibration of acoustic properties for precision dimensional measurements and image rendering using ultrasound echo time-of-flight methods on living subjects. We have shown that by combining time-of-flight measurements over several paths with external measurements of transmitter and receiver locations generating those paths the relevant acoustic parameters and dimensional measurements can be extracted. It remains to verify these results experimentally in the simple geometries considered. It also remains to show, theoretically and experimentally, that these methods can be successfully combined in a system that addresses more general geometries that combine the impediments that we have so far addressed only separately.

ACKNOWLEDGMENTS

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REFERENCES

[1] Contents and synopses of articles in the <u>Journal of Image Guided Surgery</u> can be found on the WWW at http://www.igs.wiley.com.

[2] Thomas Ault and M. W. Siegel, "Feasibility Study of a System for Frameless Patient Registration Using Ultrasonic Imaging", <u>Journal of Image Guided Surgery</u>, Vol 1 No 2, 1995.

[3] D. M. Brienza, K. C. Chung, C. E. Brubaker, J. Wang, P. E. Karg, "A System for the Analysis of Seat Support Surfaces Using Surface Shape Control and Simultaneous Measurement of Applied Pressures", <u>IEEE Transactions on Rehabilitation Engineering</u>, Vol 4 No 2, 1996.

[4] McDicken, W. N., <u>Diagnostic Ultrasonics: Principles and Use of Instruments</u> (Third Edition), Churchill Livingstone, Edinburgh, 1991. ISBN 0 443 041326.

[5] Lindsay, R. B., and R. T. Beyer, Ch. 2, "Acoustics", pp 52-64, in <u>A Physicist's Desk Reference: The Second Edition of Physics Vade Mecum</u>, Herbert L. Anderson, Editor-in-Chief, American Institute of Physics, New York 1989. ISBN 0-88318-629-2.

[6] Waters, Kenneth H., <u>Reflection Seismology, A Tool for Energy Resource Exploration</u> (Third Edition), Wiley, New York, 1987. ISBN 0-471-01144-4.

[7] Urick, Robert J., <u>Principles of Underwater Sound</u> (Third Edition), McGraw-Hill, New York, 1983. ISBN 0-07-066087-5.

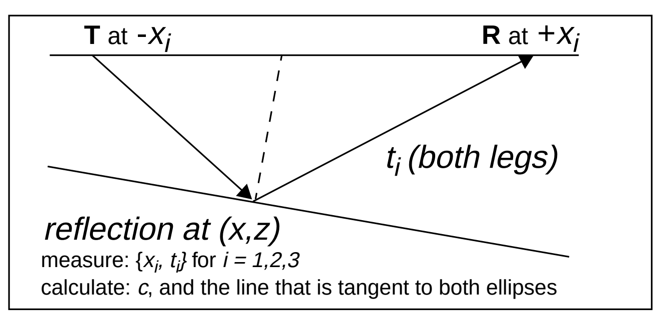

Fig. 6: Paths when layers are not of uniform thickness.