In Situ Calibration for Quantitative Ultrasonic Imaging

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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 required 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 to acquire as an integral part of the measurement process, accurate knowledge of the speed of sound in skin, fat, muscle, bone, and other live 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 article 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 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.

Each of the results described in this article is already well known in another specialized field, e.g., medical imaging, submarine navigation, or oil prospecting. We present them here in a consistent form and hierarchical order with the hope that will be a useful tutorial and reference for students who entered the field from disciplines that did not expose them to the principles and practice of physical measurements.

The Quantitative Ultrasound Problem and Its Approach

At the literal "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 live 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 any 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 (see the two subsections "For One Homogeneous Layer" and "For Several Parallel Homogeneous Layers").
2. Coping with a speed of sound gradient in a layer (see the subsection "For Parallel Layers With a Gradient").
3. Coping with a tapered layer (see the subsection "For Nonparallel Layers).

We 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.

Assumptions and Context For Ultrasonic Imaging

Ultrasonic imaging with accurate dimensional calibration requires mechanically accurate scanning capability for the raster,

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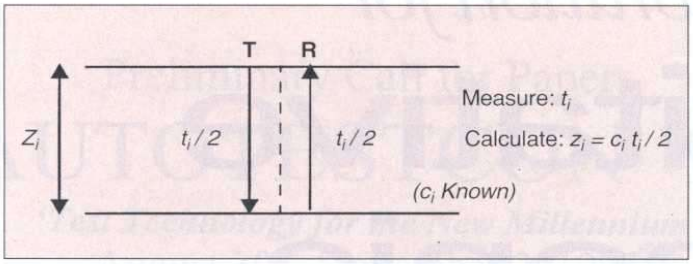

Fig. 1. Basic range measurement. Transmitter T and receiver R are displaced for clarity, but in practice may be one transducer.

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 article 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 can be measured, even if the absolute thickness of the aluminum cannot be. In other words, 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.

Review of Basic Ultrasonic Measurement Techniques

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. (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.) 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 speed of sound; these are tabulated in standard reference books for common engineering materials and for typical human tissue types [5]. However, individual differences, and additional fine details (such as the inhomogeneity and geometry issues discussed in subsequent sections, and the signal-processing complications of the sort mentioned previously) all frustrate rendering highly accurate images.

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 how the total compression is distributed 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

Motivation: The Computer-Aided Seating System

When a person is confined to a hospital bed or wheelchair, the long-term action of the forces arising from normal sitting or reclining can cause the exposed soft tissues to break down. These regions of breakdown are called pressure ulcers or pressure sores, and are commonly known as bed sores (Fig. A). Pressure ulcers are a serious problem to people confined to beds or wheelchairs and are especially serious for patients over 70 and for the spinal-cord injured [8]. The treatment of pressure ulcers is expensive—according to some accountings, an average of $120,000 per sore in 1987 [9]. An ounce of prevention is clearly worth a pound of cure in this case.

Although many factors putting a patient at risk for the development of pressure ulcers are well-identified, some individuals resist classification, being either especially resistant or especially succeptible to developing pressure ulcers despite what their risk factor profiles might indicate.

While many risk factors are well known, the mechanisms underlying the formation of pressure ulcers are not well understood. To say that sustained pressure leads to pressure ulcers is an oversimplification. Rather, the current model postulates that the distortion of the soft tissues arising from the shape of the seating or reclining surface is the determining factor in the development of pressure ulcers [10].

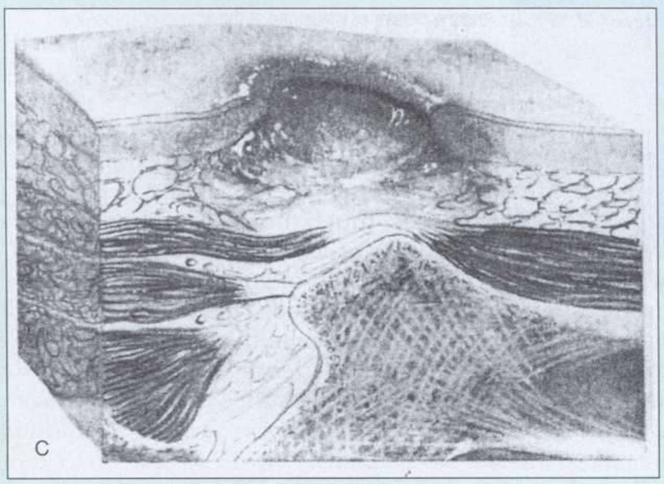

Fig. A. A drawing of a pressure ulcer (from [11]).

Consequently, to develop better preventative techniques for patients at risk of developing pressure ulcers, we need a better understanding of how the tissues of the body deform when the patient reclines or sits. While MRI or CT imaging the soft tissues of a reclining patient is relatively straightforward, imaging a sitting patient by these methods is not practical. Consequently, new systems are needed to measure the soft tissue deformation of sitting subjects.

One such system is the Computer-Aided Seating System (CASS), under development by David Brienza's group at the Rehabilitation Engineering Research Center of the University of Pittsburgh. This system (shown in Fig. B) consists of a chair with 132 actuators in an 11×12 grid. The actuators can move up and down under computer control, and the heads of the actuators swivel about a ball joint, allowing the CASS to create a continuum of seating contours. Sensors that measure the orientation and tilt angle of the

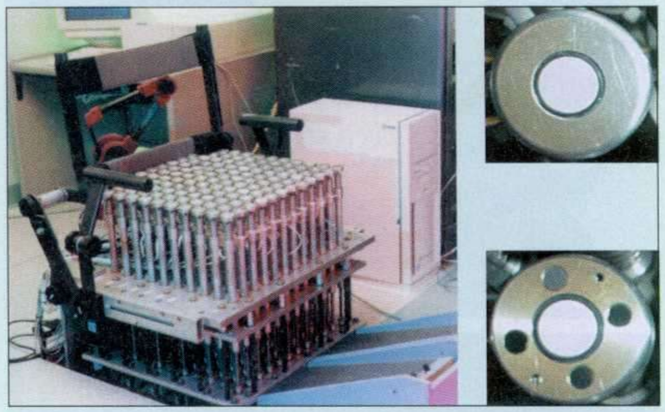

Fig. B. The Computer-Aided Seating System.

actuator heads are now being installed on the CASS.

Each actuator has a pressure sensor for measuring the pressure at the buttocks/seating surface interface. In addition, a 3×3 group of actuators in the center of the CASS array each have four ultrasonic transducers mounted on them. The operating frequency of 7 MHz was chosen as a good compromise between measurement resolution and penetration depth.

The process of measuring the shapes of the soft tissues of the buttocks is illustrated in Fig. C. A 3-D position wand with one ultrasonic sensor is used to measure the position relative to the CASS of several protuberances of the pelvis over which the skin is quite thin. The positions of the CASS actuators relative to the CASS itself are also known. With these data and a CT or MRI 3D model of the pelvic bone, the overall soft tissue thicknesses between actuators and bone can be computed. The ultrasonic transducers on the actuators themselves are used to measure the thicknesses of the individual layers of soft tissue that make up this overall thickness. While the ultrasonic transducers on the CASS actuators could also be used to measure points on the pelvic surface, the position wand produces much more reliable measurements of such points because the ultrasonic signal has to pass through only a very thin layer of tissue.

From these measurements of soft tissue shape, it is hoped that a greater understanding of the role of soft tissue distortion in the development of pressure ulcers can be gained. It is also hoped that a relationship between external pressure distribution and in-

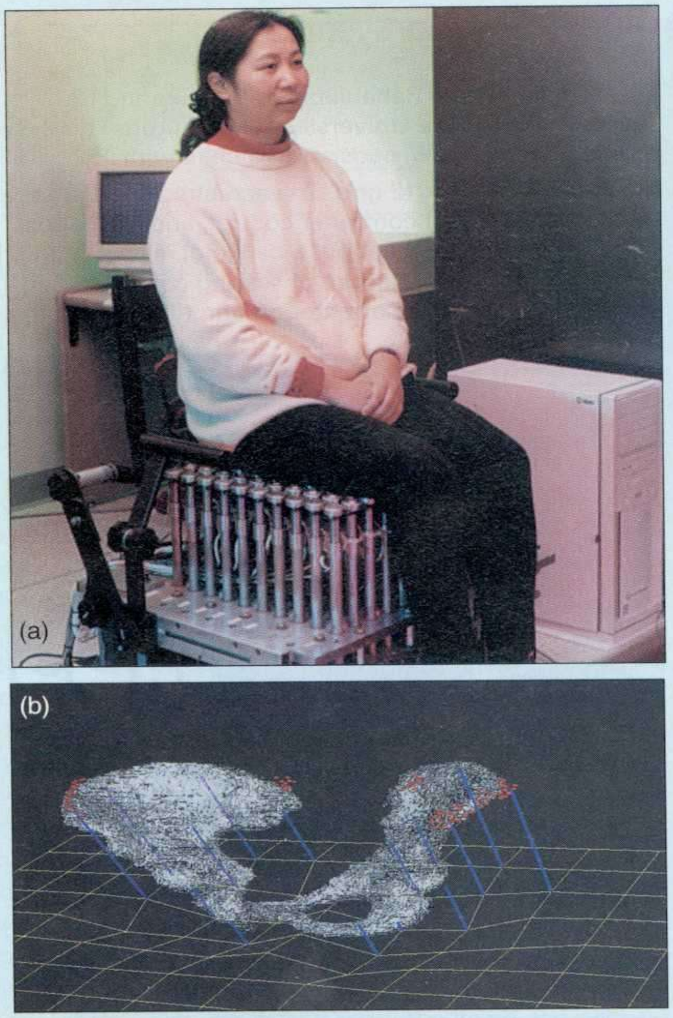

Fig. C. (a) The subject sits on the CASS. In actual use, to obtain good ultrasonic coupling, the subject's buttocks would be exposed to the surface of the CASS. The actuators are raised and lowered to create the seating contour. The soft tissues are given enough time to come to a resting state. (b) A 3D position wand with an ultrasonic transducer on the tip is used to measure points on the anterior and posterior spinae of the pelvis. The 3D position wand is an "inverse robot arm," i.e. you move the tip to some location, and it tells you the joint angles. (c) The points measured on the surface of the pelvis are matched to a surface model of the pelvis ("registration"), allowing the position and orientation of the pelvis relative to the actuators of the CASS to be computed. The nodes of the yellow grid represent the positions of the CASS actuators; the red dots are the points measured by the position wand; the cylinder heights are the thicknesses of the soft tissue between the pelvis and skin surface, and the colors of the cylinders are the pressures measured at the actuators (blue=low, red=medium, yellow=high).

ternal soft tissue shape can be obtained. This, in turn, will lead to the development of better preventative measures for wheelchair-bound individuals.

The CASS is a good example of the difficulties researchers face in making quantitative ultrasonic measurements. Multiple, uncharacterized layers of soft tissue lie between the surface of the buttocks and the pelvis. Furthermore, both the mechanical and biological (healthy vs. deteriorating) state of these tissues affects the velocity of sound within them. While an average velocity of 1540 m/s could be used to compute tissue thicknesses, much more accurate and meaningful measurements could be obtained if in situ calibration of the velocity of sound in different tissue layers could be performed. This would not only lead to more accurate tissue thickness measurements, but would also aid in identifying different tissue types and their states of health.

experimental verification of their utility is warranted (and is currently underway in our lab).

Our Approaches to the Real Problem

For 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. (While diffuse reflection is also reported in the medical imaging literature, in the present article we consider only specular reflection.) In each measurement i the speed of sound c and the layer thickness z are intertwined in

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

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

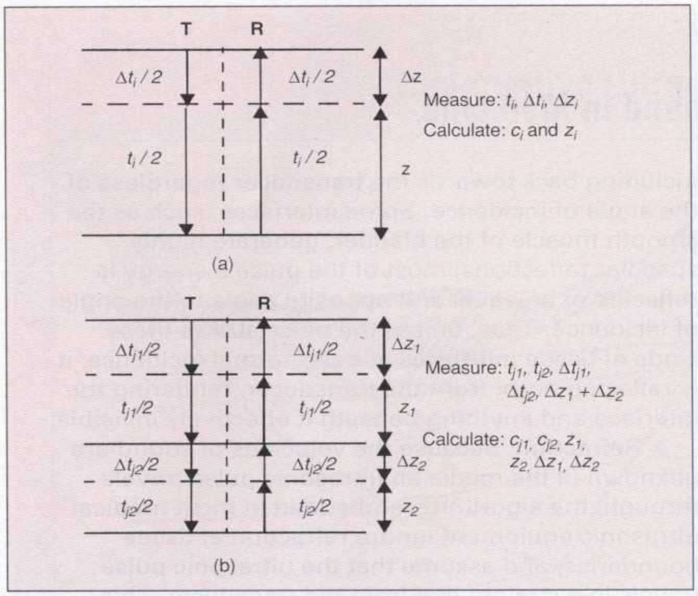

Fig. 2. (a) Differential time of flight method for one layer. (b) Differential time of flight method for multiple layers. The stiffnesses of the individual layers must be known.

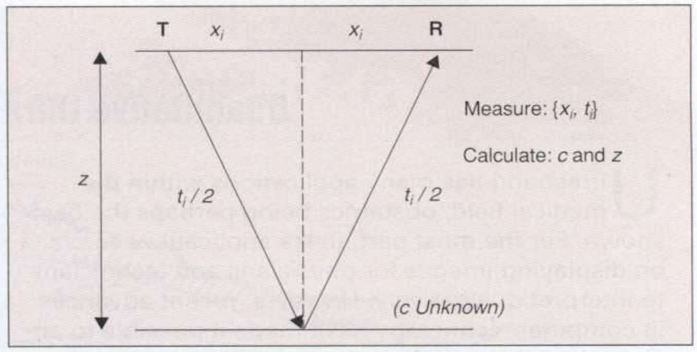

Fig. 3. Differential between two or more oblique paths.

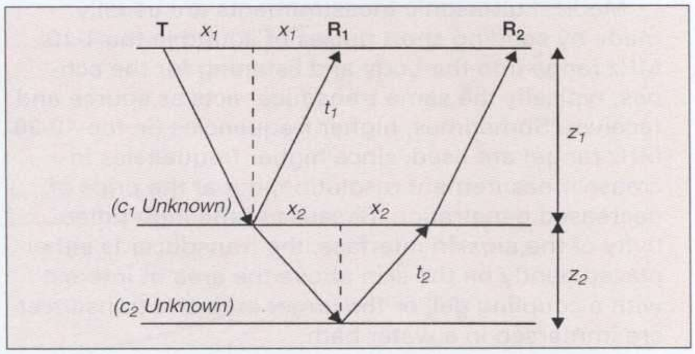

Fig. 4. Extending the differential method of Fig. 3 to multiple layers.

$$c = \frac{2\sqrt{x_2^{\,2} - x_1^{\,2}}}{\sqrt{t_2^{\,2} - t_1^{\,2}}}$$

$$z = \sqrt{\frac{x_2^{\,2} \cdot t_1^{\,2} - x_1^{\,2} \cdot t_2^{\,2}}{t_2^{\,2} - t_1^{\,2}}}$$

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

For 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. (It is presumed that the signals at receivers $\mathbf{R_1}$ and $\mathbf{R_2}$ can be distinguished by their relative amplitudes.) The approach can be applied to an arbitrary number of layers. This technique is the mainstay of geoacoustics [6], where, in oil prospecting, for example, it is routinely necessary to characterize multiple complex rock layers (strata).

For Nonparallel Layers

When layers are tapered, as illustrated (in a two dimensional cross-section) in Fig. 5, 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:

$$\frac{x^2}{a_i^{\,2}} + \frac{z^2}{b_i^{\,2}} = 1$$

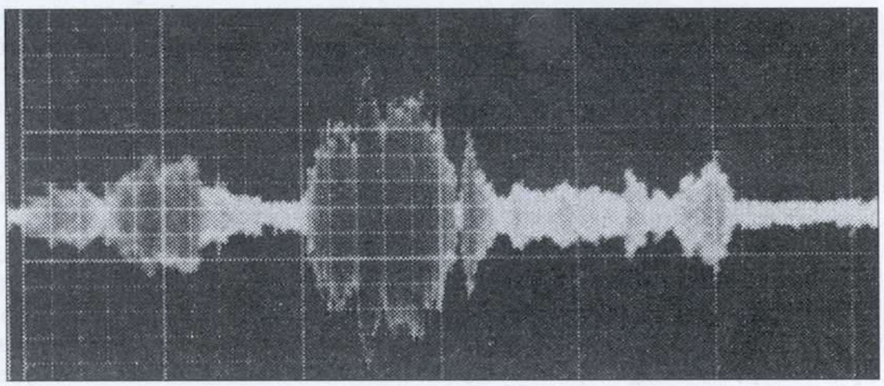

Fig. 5. Paths when layers are not of uniform thickness.

where $4a_i^{\,2} = c^2 t_i^{\,2}$ 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.

For 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. (If c is *continuous* there is refraction but no reflection.) 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. 6. The linear approximation to arbitrary gradients is particularly useful because the trajectories

Quantitative Ultrasound in Medicine

Ultrasound has many applications within the medical field, obstetrics being perhaps the best known. For the most part, these applications focus on displaying images for physicians and technicians to interpret qualitatively. However, recent advances in computer technology have made it possible to analyze these images digitally, leading to more accurate quantitative as well as qualitative analysis.

Medical ultrasonic measurements are usually made by sending short pulses of sound in the 1-10 MHz range into the body and listening for the echoes; typically the same transducer acts as source and receiver. Sometimes, higher frequencies (in the 10-20 MHz range) are used, since higher frequencies increase measurement resolution, but at the price of decreased penetration. Because of the high reflectivity of the air/skin interface, the transducer is either placed gently on the skin above the area of interest with a coupling gel, or the target area and transducer are immersed in a water bath.

There are three principal imaging modes used in medical ultrasound: the A-mode, the B-mode, and Doppler imaging.

- The **A-mode** is the simplest: it is just a display of the amplitude of the return echoes vs. time (Fig. D (a), and Fig. E).
- **B-mode** images are constructed from data collected via an array of multiple transducers or by mechanically scanning a single-element transducer. Transducer position is shown from left-to-right in the image, time of flight from top to bottom, and echo intensity as a function of transducer position and time-of-flight is represented by the gray level of the corresponding pixel (Fig. D (b)).
- **Doppler** images are usually rendered as B-mode images in which the echo's Doppler shift (which is proportional to the target's velocity) is represented by the color and intensity of that pixel (Fig. D (c)).
- Medical ultrasonic imaging suffers from many complications, including:
- **Variability of Sound Velocity**: The speed of sound in living tissue varies with tissue type, ranging between 1450 and 1650 m/s for most soft tissues. Furthermore, sound velocity can vary greatly between tissues nominally of the same type, depending on their states of health and the mechanical stresses to which they are being exposed.
- **Variability of Target Strength**: Different tissue boundaries reflect ultrasound in different ways. Most tissue boundaries are rough compared to the wavelength of the ultrasonic pulse; they therefore tend to scatter acoustic energy in all directions, including back towards the transducer regardless of the angle of incidence. Some interfaces, such as the smooth muscle of the bladder, generate highly specular reflections; most of the pulse's energy is reflected at an equal and opposite angle to the angle of incidence. Thus, unless the pulse strikes these kinds of tissue interfaces at near-normal incidence, it is reflected away from the transducer, rendering the interface and anything beneath it effectively invisible.
- **Refraction**: Because the velocities of sound are unknown in the media an ultrasonic pulse travels through, the algorithms embedded in most medical ultrasonic equipment ignore refraction at tissue boundaries, and assume that the ultrasonic pulse travels in a straight line from the transducer. This assumption can lead to errors in distance measurements and velocity measurements in Doppler imaging.
- **Backscattering**: Living tissue has many small structures whose size is on the order of the wavelength of the ultrasonic pulse. These small structures act as "scattering centers," sending a portion of the pulse's energy in all directions. This backscattered radiation produces coherent interference, which manifests itself in ultrasonic images as the time-varying fluctuation in echo strength called *speckle.* Once considered undesirable noise, backscatter is now being analyzed to identify different types of tissue (a process called tissue characterization), because the characteristics of the backscattered signal are strongly related to the density and effective cross-section of the scattering centers, which are, in turn, characteristic of a particular tissue type. Sometimes, such analysis can be done "by eye," as in Fig. E, but quantitative methods are expected to be more effective. Ultrasonic tissue characterization is used not only to distinguish between tissue types, but also to distinguish healthy and diseased tissues of the same type. Examples include attempts at identifying damaged heart muscle [12] and diffuse liver disease [13].

Of these problems, the first three issues (variability of sound velocity, variability of target strength, refraction) could be addressed through the techniques developed in this paper. These techniques make use of separate transducers for transmission and reception, while medical ultrasonic imaging systems almost always use the same transducer for transmitter and receiver. Extending medical ultrasonic equipment to use separate transmitters and receivers in a fashion that is acceptable to the medical community is a new and exciting opportunity.

Quantitative Ultrasound in Seismology

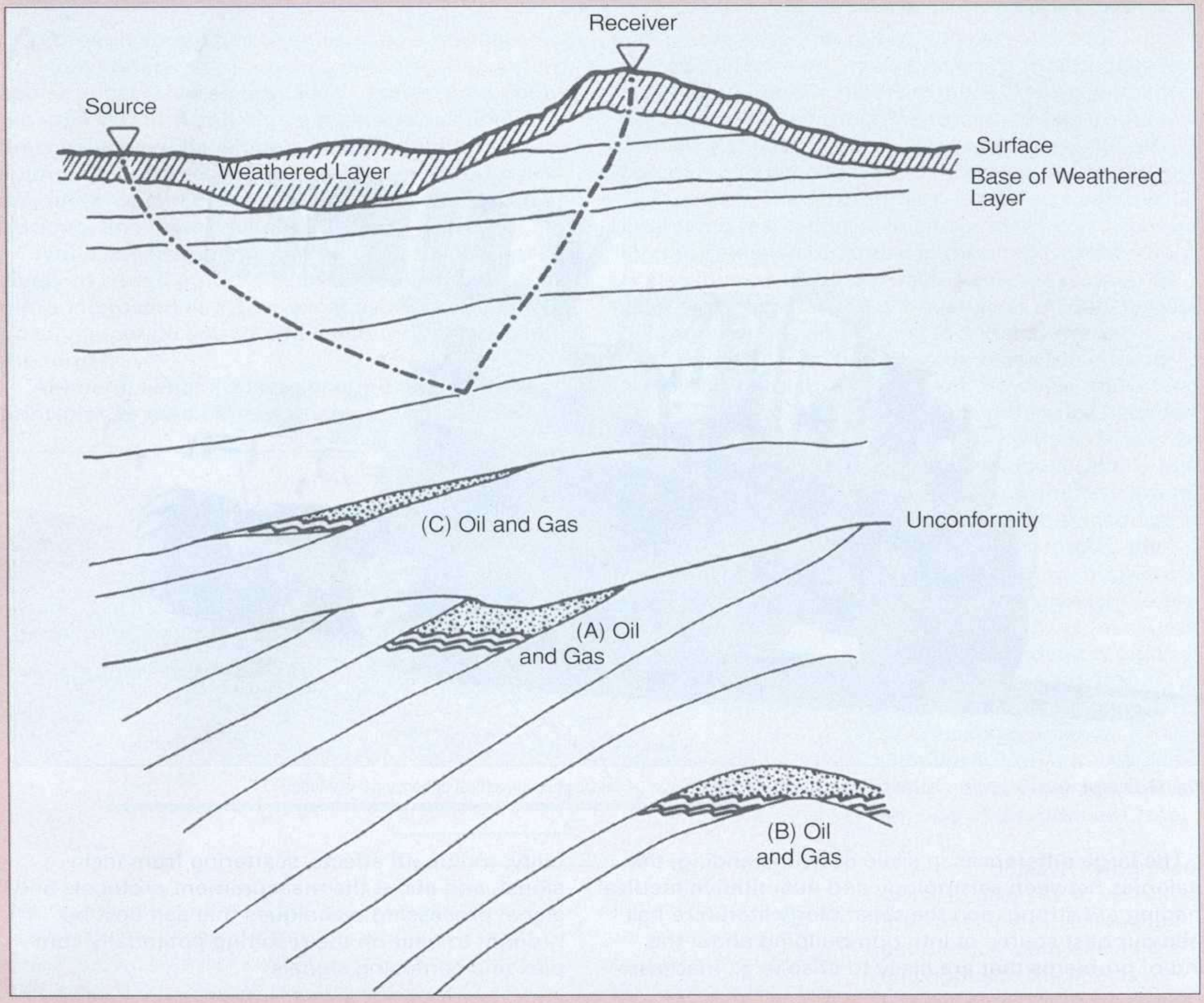

Fig. F. A diagrammatic cross section shows trapping modes for oil and gas in the pores of rocks (from [6]).

Geologists, particularly those employed in oil prospecting business, use the separated transmitter/receiver model shown in Fig. F. They are looking, of course, for pockets of petroleum, or for layers of petroleum-bearing porous rock, between layers ("strata") of various kinds of less interesting rock.

The distances, and correspondingly the energies per pulse, are naturally much larger in geological than in medical imaging: the source is likely to be a quantity of dynamite, the receiver an array of seismic monitoring instruments, and the distances involved several kilometers.

For more sustained excitation than is provided by dynamite, and to be able to control the excitation mode (e.g., to generate shear vs. longitudinal motion) large all-terrain vehicles may be equipped with various configurations of thumping and shaking machines, adding new meaning to the term "earth moving equipment" (see Fig. G and http://www.cgg.com/equipment/seismic.html).

Sophisticated arrays of sources are used to achieve directionality, as with diffraction gratings and directional antenna arrays. Sophisticated arrays of receivers are used to make the most of the return signal, e.g., to calibrate out unknown sound velocities and other parameters by taking advantage of multiple transmitter-receiver separations, as described in the main text.

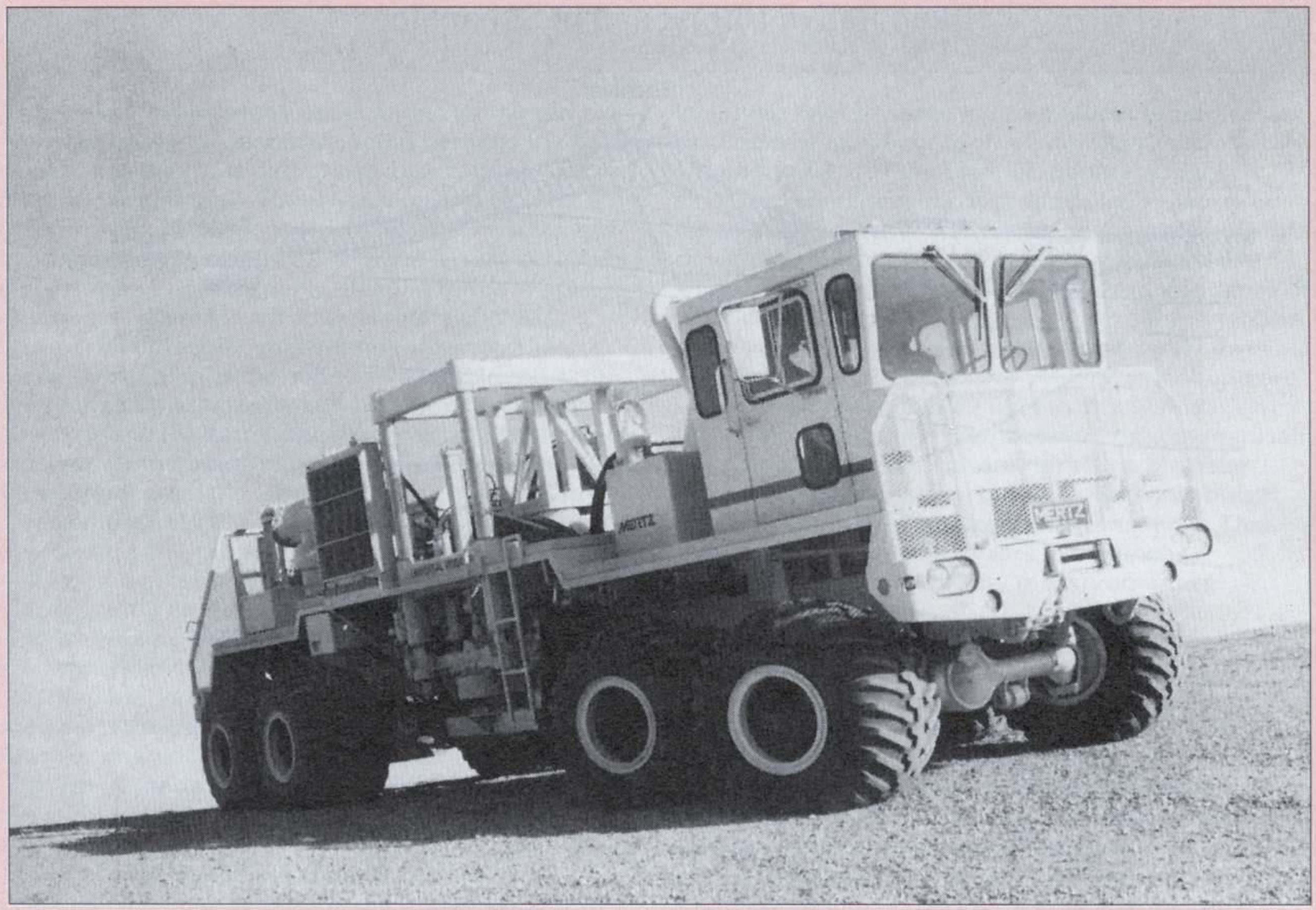

Fig. G. Seismic wave transmitter. The Mertz Universal vibrator #26/219 can be fieldadapted for vertical or horizontal vibration.

The large differences in scale notwithstanding, the analogies between seismology and quantitative medical imaging are strong, and the seismology literature has been our best source of intuition-building about the kind of problems that are likely to arise (e.g., inaccessibility, multipath effects, scattering from inclusions), and about the measurement protocols and signal processing techniques that can best be brought to bear on the resulting potentially complex and confusing signals.

are then simply circular arcs; this result is well known in underwater acoustics [7]. Four transmitter/receiver separations suffice to measure c_0, α, z, and the four launch angles $\{\theta_1, \theta_2, \theta_3, \theta_4\}$ corresponding to $\{x_1, x_2, x_3, x_4\}$. The radius of the circular arc is given by

$$R_i = \frac{c_i}{\alpha \sin\theta_i},$$

where the index i emphasizes 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.

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 be shown, theoretically and experimentally, that these methods can be successfully combined in a practical system that addresses natural geometries combining the impediments that we have herein addressed only separately.

Acknowledgments

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Quantitative Ultrasound Underwater: Lessons for Medical Imaging

Although sound has been used as a measurement tool underwater for over a century, the modern age of underwater sonics (a.k.a. "sonar") did not begin until World War II. Since then, various applications have been developed, including depth sounding, fish finders and counters, position markers, underwater communications, target location and tracking, and minesweeping.

While sonar systems can be used to listen passively to sound emitted from target itself, we are more interested in active sonar systems, which purposely generate sound and listen for echoes from the target.

Although sonar systems operate on the same principles as medical and seismological systems, they must be equipped to deal with the temperature gradient of different layers of the sea. Since the velocity of sound in water varies with temperature, this gives rise to a velocity gradient that causes straight-line paths to become curved arcs. (Sound velocity in water also varies with pressure and salinity. However, the values of these quantities, and hence their effect on sound velocity, are much more predictable than ocean temperature.) If not compensated for, these gradients can cause the position of the target to be mismeasured.

When combined with the reflective properties of the surface and floor of the ocean, the curved paths can create "shadow zones," places where, in principle, sound from an active system cannot penetrate (see illustration). In practice, there is some penetration, but the strength of the return echoes from these "shadow zones" is diminished typically by 60 dB compared to echoes from outside the shadow zone.

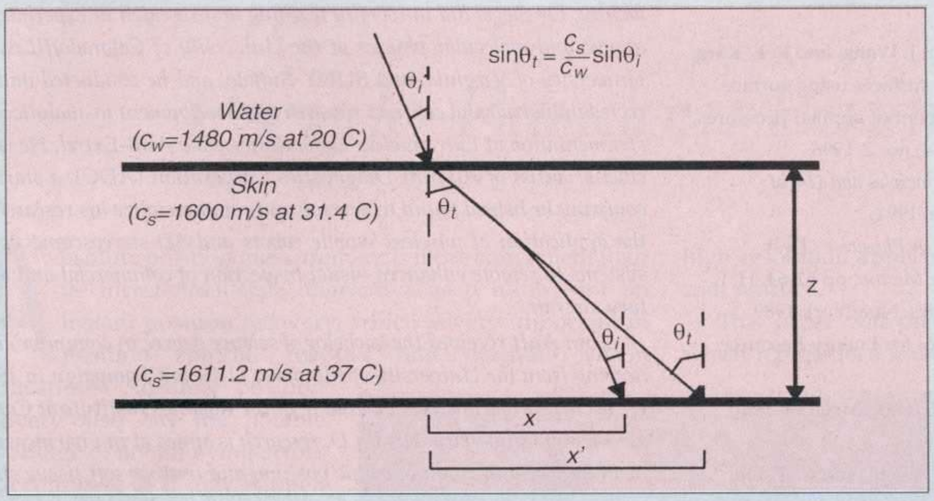

Fig. H. Illustration of the effect of a 5.6 C gradient over a depth z on the path of an ultrasonic pulse at a water/skin interface. The pulse is incident at an angle of θ_i to the normal to the surface of the skin, and is transmitted at an angle of θ_t to the same normal. It exits the thermal gradient region at x with an angle of θ to the normal. If the effect of the gradient is ignored, there will be an error in the computed position of x of (x-x') and an error in the computed angle of θ equal to (θ- θ_t). While (x-x') is small for a very large range of θ_i, (θ- θ_t) can become large for even modest values of θ_i.

Current research in sonar systems concentrates on improving resolution, increasing range, and improving the detection of targets in the shadow zone; there is also interest in "underwater stealth"—ways to decrease the detectability of underwater military targets (like submarines and mines) to sonar systems.

In living beings, homeostatis keeps the internal temperature relatively constant. However, there is a approximately 5-6C temperature difference between the surface of the skin and the inside of the body. While the effects of this gradient are not significant if the transducer is within 10 of normal to the skin surface, large angles of incidence can give rise to errors in position, as measured where the gradient region ends, and even modest non-normal angles of incidence can give rise to significant errors in the angle of transmission at the far side of the gradient (see Fig. H).

Aniostropies in living tissue can also give rise to velocity gradients in tissues whose geometrical path effects are analogous to the effects of temperature gradients underwater. So far, no research has been done on the extent to which ultrasonic pulse paths deviate from straight lines in living tissue.

Research on increasing the ability of sonar systems to image targets with low sonic profiles and targets within the shadow zone has a much more obvious application to medical ultrasonics. Techniques applied to these problems underwater could also be applied to imaging structures deep within the body, where attenuation greatly reduces the strength of a return echo by the time it reaches the transducer, or to imaging large, smooth structures, whose reflection is largely specular, at non-normal angles of incidence.

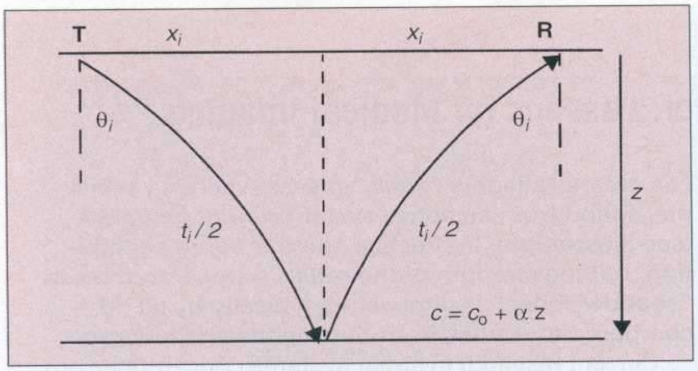

Fig. 6. Trajectories when the speed of sound is not constant.

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***Mel Siegel** founded and directs the Intelligent Sensor, Measurement, and Control Laboratory in the Robotics Institute, School of Computer Science, Carnegie Mellon University. Prior to joining the CMU faculty, Dr Siegel did university teaching and research in experimental atomic and molecular physics at the University of Colorado/JILA, the University of Virginia, and SUNY-Buffalo, and he conducted and directed industrial and contract research and development in analytical instrumentation at Extranuclear Laboratories, now ABB-Extrel. He is the chief scientist of Aircraft Diagnostics Corporation (ADC), a start-up company he helped found to develop and commercialize his research on the application of wireless mobile robots and 3D-stereoscopic vision systems to remote enhanced visual inspection of commercial and military aircraft.*

***Tom Ault** received the bachelor of science degree in computer engineering from the University of Illinois at Urbana-Champaign in 1992. He is currently a doctoral candidate in the Robotics Institute at Carnegie Mellon University. His Ph.D. research is aimed at in vivo measurement and visualization of pelvic position and buttock soft tissue shape with respect to variations in seating surface shape. In addition to medical applications of ultrasound, his research interests include neural networks, mobile robot motion planning, automated information retrieval, and artificial intelligence in general. He expects to graduate in the Spring of 1999.*

Although his English is still thick with a Russian accent, he now works for the same employer that I do, seated in the cubicle near mine. Quite by chance, we discovered that each had worked in the defense sector of his respective country. After the Cold War and the escalating collapse of the Soviet economy, my adversary journeyed to the U.S. with his family, hoping to find work. He did, as a computer programmer and architect.

We have chatted often, trading stories that confirm the universality of the engineering challenges faced by any modern technical endeavor. On one occasion, my comrade related an interesting tale that highlights how the severity of these challenges can directly affect the type of technology that is developed. His story concerned a U.S.-designed-and-built electronic system that was captured by the Soviets. Through reverse-engineering, the talented Soviet engineers and scientists were able to determine the function of every integrated circuit in the system, except for two pins on one of the newer (late 1970s) chips.

Try as they might, they could not discover the function served by these two pins. Only years later, at an international conference, was the mystery solved through a conversation with the original U.S. developers of the chip. The two pins in question were used in checking the proper functioning of the chip logic—to facilitate a built-in test! This new capability was unknown to the resource-strapped Soviets. The universal challenges of cost, time, and proper direction had prevented them from understanding this new approach to chip design.

While times have changed and the Cold War is officially over, the challenges encountered during the practice of good engineering remain the same: limited budgets, shrinking schedules, and poor focus on the part of management. Now, though, the Russian and American engineers can work together, strengthened by their subtly different approaches to engineering, to overcome these challenges. Only time will tell.

Erratum

On page 13 of the September 1998 issue of *IEEE Instrumentation and Measurement Magazine*, in the article "In Situ Calibration for Quantitative Ultrasonic Imaging," Figure 5 should have appeared as the figure below. We apologize for any inconvenience this may have caused.

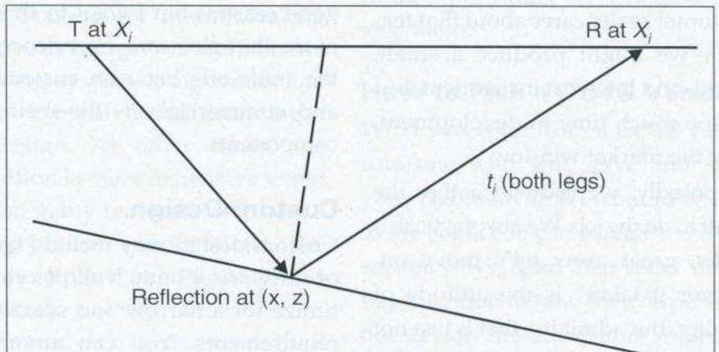

Figure 5. Paths when layers are not of uniform thickness.