In Situ Calibration for Quantitative Ultrasonic Imaging

Mel Siegel and Tom Ault

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,

1053-5888/98/$10.00©1998IEEE

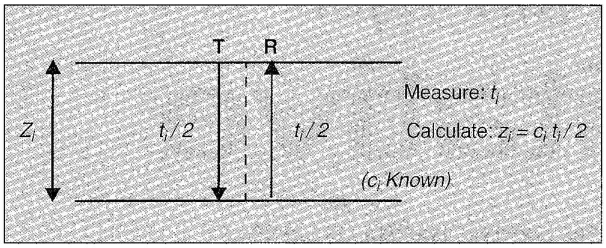


Fig. 1. Basic range measurement. Transmitter T and receiver R are displaced for clarity, but in practice may be one transducer.

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.

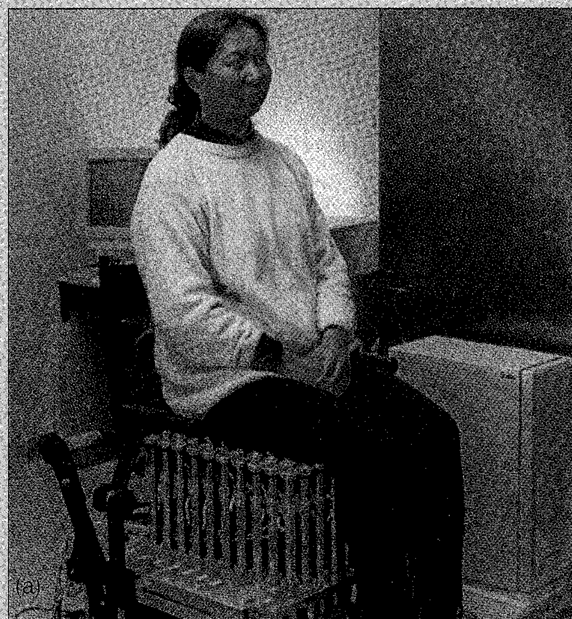
If modest dimensional distortions can be tolerated then nominal values can be used for the speed of sound; these are tabulated in standard reference books for common engineering materials and for typical human tissue types [5]. However, individual differences, and additional fine details (such as the 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



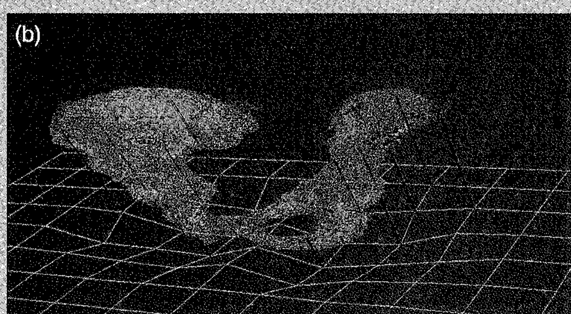


Fig. C. (a) The subject sits on the CASS. In actual use, to obtain good ultrasonic coupling, the subject's buttocks would be exposed to the surface of the CASS. The actuators are raised and lowered to create the seating contour. The soft tissues are given enough time to come to a resting state. (b) A 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

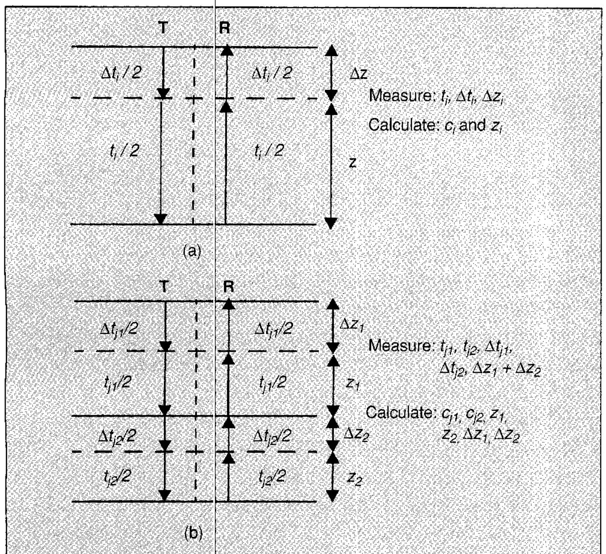


Fig. 2. (a) Differential time of flight method for one layer. (b) Differential time of flight method for multiple layers. The stiffnesses of the individual layers must be known.

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

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

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:

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

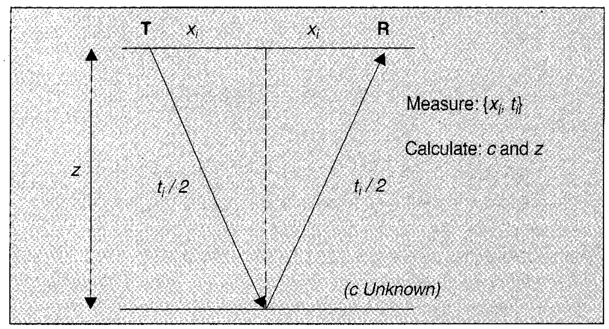


Fig. 3. Differential between two or more oblique paths.

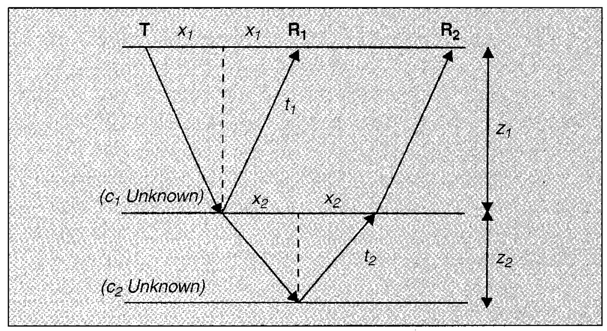


Fig. 4. Extending the differential method of Fig. 3 to multiple layers.

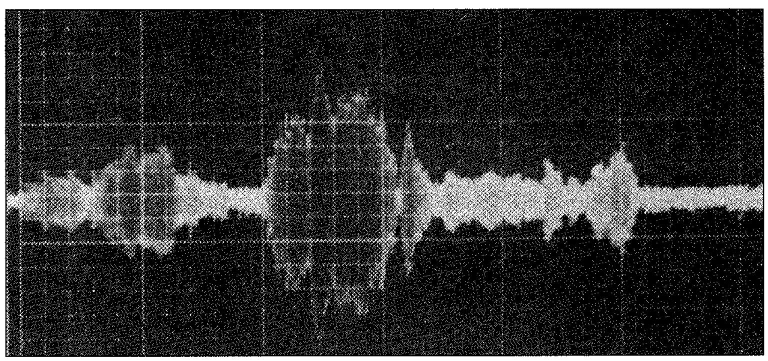


Fig. 5. Paths when layers are not of uniform thickness.

where $4a_i^2 = c^2t_i^2$ 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

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

The authors thank Joyoni Dey for stimulating conversations concerning experimental approaches to some of the issues discussed here, Dave Brienza for his many technical contributions and for giving us the opportunity to work with the CASS group, and Jue Wang for designing and implementing the hardware for the ultrasonic transducers on the CASS.

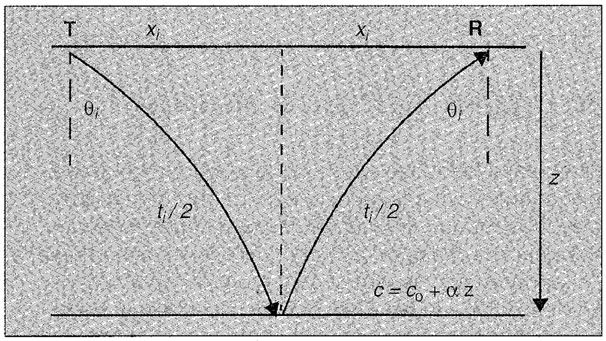


Fig. 6. Trajectories when the speed of sound is not constant.

References

[1] Contents and synopses of articles in the *Journal of Image Guided Surgery* 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," Journal of Image Guided Surgery, vol. 1, no. 2, 1995.

[3] D. M. Brienza, K. C. Chung, C. E. Brubaker, J. Wang, and P. E. Karg, "A system for the analysis of seat support surfaces using surface shape control and simultaneous measurement of applied pressures," *IEEE Trans. Rehabilitation Engineering*, vol. 4, no. 2, 1996.

[4] W. N. McDicken, *Diagnostic Ultrasonics: Principles and Use of Instruments*, 3rd ed., Churchill Livingstone, 1991.

[5] R. B. Lindsay, and R. T. Beyer, *"Acoustics," A Physicist's Desk Reference: The Second Edition of Physics Vade Mecum*, pp 52-64, H. L. Anderson, ed., American Institute of Physics, New York 1989.

[6] K. H. Waters, Reflection Seismology, A Tool for Energy Resource Exploration, 3rd ed., Wiley, 1987.

[7] R. J. Urick, *Principles of Underwater Sound*, 3rd ed., McGraw-Hill, 1983.

[8] P. E. Karg, D. M. Brienza, C. E. Brubaker, J. Wang, and C. T. Lin, "A system for the design and analysis of seat support surfaces," *Proc. 19th Annual RESNA Conf.*, Salt Lake City, Utah, June 7-12, 1996.

[9] J. Sanders, B. Goldstein, and D. Leotta, "Skin response to mechanical stress: adaption rather than breakdown- a review of the literature," *Journal of Rehabilitation Research and Development*, vol. 32, no. 3, 1995.

[10] S. Levine, R. Kett, and M. Ferguson-Pell, "Tissue shape and deformation versus pressure as a characterization of the seating interface," *Assistive Technology*, vol. 2, 1991.

[11] J. A. Makleburst, "Pressure ulcer assessment," *Clinics in Geriatric Medicine*, vol. 13, no. 3, 1997.

[12] Y. Sajio, H. Sasaki, T. Naganuma, and M. Tanaka, "Ultrasonic tissue characterization of diseased myocardium by scanning acoustic microscopy, *Journal of Cardiology*, vol. 25, no. 3, 1995.

[13] B. Oosterveld, J. Thijssen, P. Hartman, and G. Rosenbusch, "Detection of diffuse liver disease by quantitative echography: dependence on a priori choice of parameters, *Ultrasound in Medicine and Biology*, vol. 19, no. 1, 1993.

[14] J. Serup, J. Keiding, A. Fullerton, M. Gniadecka, and R. Gniadecki, *High-Frequency Ultrasound Examination of Skin: Introduction and Guide in Non-Invasive Methods and the Skin*, CRC Press, 1995.

***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.*