

The growth in the applications of physics to medicine is perhaps best illustrated by the developments in medical imaging. When AAPM was founded, medical imaging consisted primarily of x-ray fluoroscopy (direct images) and radiography (exposures on film). Although these techniques still dominate the work of a radiology department, there are a number of new ways of making images of the interior of the living body. (See, for example, the cover of this issue and figure 1.) Many of these new imaging methods depend on computers to perform the enormous amounts of data-analysis required—computed tomography or nmr imaging, for example, would not be possible without computers—and in many other cases computers serve to provide enhanced images that may allow more accurate diagnoses.

In this article we will discuss several of these new medically useful imaging techniques. Digital subtraction angiography and computed tomography both make more sophisticated use of the information available from x rays. Nuclear magnetic resonance was a well-known research field in 1958; in the last few years it has become useful in medical imaging. The techniques of computed tomography have made it possible to make images of the distribution of radioactive tracers in the living body. The use of ultrasound to form images, in its infancy in 1958, is now a fairly common diagnostic tool.

While some of the contributions that physicists have made to medical imaging are eminently newsworthy, other, more mundane contributions—such as the development of quality-control devices to improve medical images—may have a more significant effect on health care. Thus, for example, unnecessary x-ray exposure has received considerable attention in the press; however, a poor x ray that causes a radiologist to miss an early diagnosis of a breast cancer may cost the patient her life, and the amount of radiation is of minor concern.

Digital subtraction angiography

One of the drawbacks of conventional film radiography is that material other than bone shows up poorly. Even when one introduces an absorbing material into the organ or vessel one wants to examine (to enhance the

The physics of medical imaging

Images computed from x-ray absorption, from positron emission, from nuclear magnetic resonance and from reflections of ultrasound are showing tissues and details unimagined 25 years ago.

Paul R. Moran, R. Jerome Nickles and James A. Zagzebski

contrast with surrounding material) the additional absorption may be hard to see against the background of bone or other tissues. The conventional technique also fails to use all the information that may be available in the transmitted x-ray beam, such as the absorption spectrum of the material and time variations in its behavior.

The technique of "subtracting the background" is well known in other areas of physics. It can be applied to diagnostic x-ray images to provide information not available from single broadband projection images. For example, from an image obtained using an absorbing marker one can subtract an image obtained without the marker, thus making clear the contribution of the absorbing material. Alternatively, one can subtract images obtained at two different energies from each other to bring out the presence of materials whose absorption changes between these two energies—that is, materials that have an absorption edge in that energy range.

Because the subtraction technique is most commonly used to examine blood vessels and related structures, and because the images are generally obtained using digitized fluoroscopy apparatus, the technique is referred to as digital subtraction angiography.

The subtraction of background features from an image allows the clinician to work with much smaller amounts of absorbing material. One common application (shown in figure 2) is the examination of arteries; iodine injected into the bloodstream is the absorbing medium. The sensitivity to changes in absorption allows one to inject the iodine into a vein and follow its progress into the arterial system as the blood is pumped out of the heart again after passing through the lungs.

Not only is venous injection much less invasive than arterial injection, but the concentration of iodine is typically twenty times lower than in direct angiography. Typically, one follows the flow of the injected iodine as it makes its first pass through the arterial system by making a sequence of images—about 1 to 30 images per second—and subtracting the pre-injection background from each. Because the iodine signals are weak, the image must be amplified, and because noise is amplified as well, the system designers must take care to provide for adequate x-ray exposure, and to use tv cameras and amplifiers with high signal-to-noise ratios. Depending on the degree of patient motion one can expect, one may also use signal-integration techniques to reduce the effects of noise.

Digital subtraction angiography is just one example of a generalized technique¹ in which data obtained under slightly different conditions are subtracted to obtain clinically useful information about the attenuation x rays. In its passage through the body (or whatever else is being examined) the intensity I of the transmitted beam is a function of its energy, of time, and of position. By examining the changes in the function $I(E, t, \mathbf{r})$ with respect to small changes in E , t , or $\mathbf{r}$, one is, in effect, looking not at I itself but at its partial derivatives—or rather, at the first few terms in the Taylor series for the function, because the intervals one uses in practice can never be truly infinitesimal.

We have already mentioned energy subtraction, which may be used to provide images selectively displaying particular organs or tissues. Iodine, for example, has an abrupt discontinuity in its absorption spectrum at 33 keV (its K-edge). Subtraction images

Paul Moran, Jerome Nickles and James Zagzebski are in the department of medical physics at the University of Wisconsin in Madison; Moran is currently visiting professor of radiology at Bowman Gray School of Medicine, Lake Forest University, Winston-Salem, N.C.

formed from data obtained with nearly monoenergetic beams with average energies above and below 33 keV will emphasize tissues containing iodine—whether naturally or artificially introduced—while suppressing the effects of bone and other tissues. Because one is not relying on a change in iodine concentration with time, one can allow the iodine to reach a temporary equilibrium distribution before one obtains the image. Heavily filtered beams from conventional x-ray sources as well as beams from monoenergetic sources have been used to investigate this technique, and it is currently being studied with finely tuned synchrotron radiation.²

First-order spatial derivatives yield not only the data for traditional image-processing techniques such as edge

enhancement, smoothing, and various operations on the spatial-frequency content of the image, but also for more recently introduced techniques of computed tomography.

Image processing of single films is severely limited by the fact that many variables have already been averaged over in the production of the image. They are also constrained by the fact that the trained eye and brain are already an excellent image-processing system, so that one must take care not to degrade subtle features that an observer may use to analyze an image. Because of this, computer processing of conventional radiograms has had little impact on clinical practice. However, there is considerable information in a single x-ray film that can be brought out by appropriate processing. Spatial

filtration, for example, can eliminate large variations in overall brightness in different areas of the image; alternatively, it can pick out features having a particular spatial frequency. A group at Wisconsin has used the latter in an attempt to show coronary arteries (high spatial frequencies) against the left ventricle (low spatial frequency); the technique, in fact, involves a second-order derivative, in that the spatial filtration is performed on a subtraction angiogram of the sort we described earlier. So far this technique has not proved as useful as the more invasive cardiac catheterization techniques.

Other second-order techniques have also been investigated. These include:

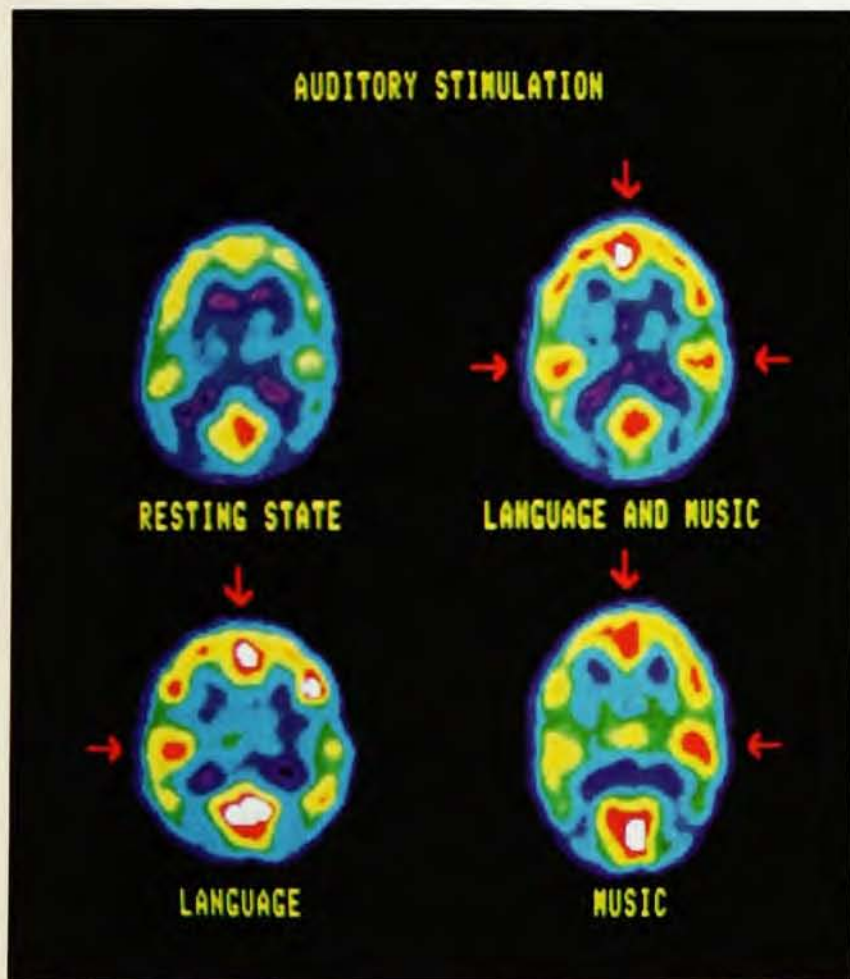
- ▶ hybrid subtraction—the derivative of the intensity with respect to time and energy
- ▶ tomographic subtraction angiography—the derivative of the intensity with respect to spatial variables and time
- ▶ tomographic energy subtraction—the derivative of the intensity with respect to energy and depth
- ▶ multiple K-edge subtraction—the second derivative of the intensity with respect to energy

Each of these techniques combines some of the advantages of the separate subtractions involved.

The hybrid subtraction technique, for example, was introduced³ to reduce the effects of patient motion in digital subtraction angiography. Instead of taking single exposures before and after injecting the absorbing material, one takes a pair of exposures at widely separated energies. Subtraction of the energy pair produces images in which no tissues except the carriers of the opaque medium show up; subtraction of the resulting time-separated images then removes the remaining effects of bone and displays clearly the effects of the injected contrast material.

Some of the arteries carrying the injected iodine may well overlap in the two-dimensional image obtained in a digital subtraction angiogram; this overlap often makes diagnosis difficult. One possible solution⁴ to this problem is to use tomographic techniques in conjunction with subtraction angiography. To obtain such three-dimensional information one can either move the source rapidly or use multiple sources to illuminate the patient from several different angles. The resulting data can then be processed to provide images of tissues containing iodine viewed from several different angles, giving the clinician an effectively three-dimensional view of the arteries.

Energy subtraction combined with tomography can provide information



Variations in brain metabolism as a function of stimulation. These positron-emission tomograms show the distribution of F^{18} -labeled 2-deoxyglucose in a transverse slice of a living brain with various kinds of auditory stimulus. (The eyes are open for all four images.) In the resting state the ears are plugged; in the other images the stimulation consists of a segment of the radio program "The Shadow" (language) and a Brandenburg Concerto by Johann Sebastian Bach (music). The frontal lobes are indicated by the vertical arrows; the auditory cortex (temporal lobes) are indicated by horizontal arrows and the visual cortex (occipital lobes) is at the bottom of the images. (Courtesy of Michael Phelps, UCLA.)

Figure 1

on the chemical composition of various regions of the body. One can also obtain information about materials by making use of the different variations with energy of the photoelectric and Compton-effect contributions of x-ray scattering. The ratio of these two contributions at widely separated energies can be used to get some idea of the effective atomic number. In a tomogram, one can thus get more information about the chemical composition of each region than from just the absorption data at a single energy.

The first-order techniques, specifically computed tomography (equivalent to depth subtraction) and digital subtraction angiography (time subtraction) have thus far clinically had the most importance. Their impact is probably due less to the computer processing, which can enhance the display of data, than it is due to the interesting data that can be generated by subtraction processes in general. By combining data from several images obtained under slightly different conditions, one can exploit much more fully the physical, geometrical and physiological information obtainable from what would otherwise be "just an x-ray."

Computed tomography

Computed tomography consists of producing an image of a slice (*tomos* in Greek) of a three-dimensional object by combining a large set of scans through the object at different angles, each scan producing the equivalent of strip of a standard x-ray image. The information from all these scans is then used to reconstruct the two-dimensional variation of x-ray absorption through the entire slice.

The basic techniques of producing and detecting the x-rays is over 30 years old. What is new is the information-processing capacity. Allan Cormack and Godfrey Hounsfield—as well as others—independently developed⁶ the mathematical techniques for reconstructing tomograms from the integrated absorption of beams passing through the absorber. Hounsfield and Cormack won the Nobel Prize for Physiology and Medicine in 1979 for this work (see *PHYSICS TODAY*, December 1979, page 19). The commercial development of scanners for computed x-ray tomography followed rapidly, and most major hospitals by now have access to CT scanners.

Modern x-ray CT scanners in general incorporate a large number (500 to 1500) highly collimated detectors mounted on a circular gantry that surrounds the patient in the region from which one wants the tomogram. An x-ray tube is also mounted on a gantry and rotates around the patient, producing some 500 to 2000 views, each a different angular projection of the



Digital subtraction angiogram of the carotid arteries that provide blood to the brain. The photo shows an x-ray image of the aorta branching, among others, into the two carotid arteries (indicated by arrows). The images are read off a fluoroscope by a tv camera, digitized and then subtracted. Note the absence of bones or other structures in the resulting image. Figure 2

patient, in the course of a 360° rotation. Each view produces about 500 measurements of the x-ray transmission across the subject's diameter, and a typical view takes about 5 to 10 milliseconds. The average power input to the x-ray tube required for this speed is about 20 to 30 kW. The x-ray beam is collimated axially and is variable between 1.5 and 10 mm; the thinner the illuminated slice, the better is the axial resolution. To remove the effects of variations in x-ray output and detector sensitivity, the individual measurements are normalized to reference detector readings and scaled to a reference material. The result is the total attenuation along the path the x-ray beam takes through the sample.

One way to understand the computations involved is to consider each scan as consisting of parallel rays. The Fourier transform of a single angular view is then, it turns out, basically the two-dimensional Fourier transform of the x-ray attenuation density in the slice, for values of the wavevector perpendicular to the view angle. (See the articles by William Swindell and Harrison Barrett, *PHYSICS TODAY*, December 1977, page 32, and by Rowland Redington and Walter Berninger, *PHYSICS TODAY*, August 1981, page 36.) The complete scan thus produces a complete Fourier transform, for wavevectors in all directions. (It is, essentially,

the Fourier transform given in polar coordinates.) To reconstruct the tomogram, the data for a complete set of views are appropriately filtered ("apodized") and entered into a Fourier inversion algorithm. Other geometries of illumination and other ways of looking at the problem lead to different computational schemes, with other integral transforms replacing the Fourier transform.

The output of the calculations is typically displayed on a tv monitor divided into an array of 512×512 pixels, the brightness at each point being the result of an inversion computation that involves on the order of a million computer operations. The display is typically supported with image-array buffers and electronics so that the operator can select the mid-level brightness and the contrast range of the display. The computations are sufficiently sensitive that the x-ray quantum statistics limit the precision. Image noise, at spatial resolutions of 5 line pairs per centimeter, is typically less than 1/2% the attenuation of an equal volume of water. The range of attenuations from air to very dense bone is about 8000 times as much. Figure 3 shows a computed tomogram.

Images for nmr

The frequency and decay rate of an nmr signal depend on the local magnetic—and hence chemical—environment of the nucleus being probed. One can thus produce images—of the interior of a body, for instance—that display the variations in nmr signals and thus display anatomical features. As in x-ray computed tomography, the measured data are given as a Fourier transform of the spatial image desired.

The initial work for producing nmr images was also done in the early 1970s. (See *PHYSICS TODAY*, May 1978, page 17.) Paul Lauterbur announced⁶ the underlying principles of nmr "zeugmatography" in 1973, about the same time that the work on x-ray computed tomography was taking place. However, while CT scanners are now in place at many hospitals and are being actively used in diagnoses, nmr scanners are only now moving into research medical centers for clinical evaluation. The chief reason for the difference is probably historical: X-ray technology has had a long history in medicine, and computed tomography can be viewed as a more efficient use of information for which the interpretation was clear, the clinical efficacy was obvious and the engineering and technological problems were known and soluble. For nmr no such history exists, and clinicians must now begin to establish a tradition of interpreting nmr images.

Because nmr signals are intrinsically weak, data must be collected over long

times to obtain reasonable signal-to-noise ratios. Thus, for example, while it takes about 2 seconds to complete an x-ray scan of a head with sufficient detail to study the morphology of the ventricles in the brain, a proton-nmr scan requires several tens of minutes to produce data for a comparable image. (Protons produce the largest intrinsic nmr signal and are the most abundant nuclei in the body; other nuclei, with smaller gyromagnetic ratios and smaller abundances, would require even longer times for scans.)

The physical appearance of the apparatus to produce an nmr scan is very similar to that used for x-ray computed tomography: A large circular gantry surrounds the patient, with the axis of the gantry usually along the long axis of the patient. The gantry houses coils to produce a strong static axial magnetic field, coils to produce gradients in this field, and an rf coil to transmit the excitatory pulses and to pick up the nmr signal.

The strong static, homogeneous base magnetic field—ranging from 0.07 T up to 1.5 T in some applications—produces a net equilibrium nuclear magnetization. From a semiclassical point of view, one can think of the nuclei as precessing when they are displaced from the equilibrium; the resonant frequency—the Larmor frequency—depends on the magnetic moment of the nuclei and on the magnetic field at the nucleus. The decay of the precession depends on local dephasing processes as well as thermodynamic relaxation processes. When one applies a short pulse of rf at the Larmor frequency, the spins are made to precess, and one can observe the resulting “ringing” as an induced emf at the Larmor frequency, which can be picked up by the same tuned coil that transmitted the excitation pulse. The signal is amplified and demodulated by phase-coherent single-sideband detectors, which use the exciting oscillator for the reference signal.

To obtain information about spatial position in the nmr signal, one adds a gradient to the applied field. The spatial variation in the field gives rise to a frequency modulation of the rf signal. A single measurement cycle with a particular gradient applied thus gives a spatial Fourier transform of the excited nuclear resonance. A series of such cycles with the gradient varied in direction from cycle to cycle generates the full two- or three-dimensional Fourier transform of the nmr signal. Computer algorithms then select particular delay times (to get consistent intensities), apodize the data and perform the Fourier transforms that yield the sort of images shown on the cover and in figure 4.

When a proton scan is made sensitive only to density, a superbly uninterest-

An x-ray computed tomogram of a head.

The brain tissue is clearly differentiated from the surrounding bone as well as the (darker) ventricles, filled with cerebrospinal fluid. The light area to the right of the ventricle indicates a diseased condition.

The image was obtained with a Siemens DR3 scanner at the University of Wisconsin. Figure 3



ing image results. All tissues have just about the same nmr-observable density of hydrogen. However, the decay times of the nmr signal turn out to depend strongly on physical and chemical factors that differ from tissue to tissue.⁷ The biology of these differences is not as yet well understood.

As we mentioned, there are several kinds of processes that affect the decay of the nmr signal:

- Processes that contribute to the establishment of thermal equilibrium of the spins with the rest of the body; one can think of these processes as analogous to the slow collapse of the spinning top, except that here, because the spin itself never decreases, the decay is to the aligned state. The characteristic time for these processes is called T_1 .

- Processes that contribute to the loss of coherence among the various precessing spins. This, considerably shorter, relaxation time is generally called T_2 .

By appropriately timing sequences of excitation pulses and selecting delay times for making the measurements, one can thus modulate the density-dependent nmr intensity by factors that depend on the local physical and chemical environment. Because the timing sequences as well as the field gradients can all be controlled, the operator not only acquires the data to produce an image, but in a profound sense also produces the specific subject to be imaged.

It even appears possible to observe directly the shifts in the Larmor frequency that arise from changes in the chemical environment, and thus obtain

direct information about metabolic processes—for example, the relative levels and distribution of metabolic products incorporating phosphorus-31, say—in a living subject. The physician thus will be able to use nmr images to study normal and diseased tissues from a wide variety of viewpoints, anatomical, morphological, physiological or functional. Each of these uses of nmr data will of course be guided by—and contribute to—an already substantial knowledge of normal and pathological states. Thus, for example, while x-ray tomography yields highly detailed images of morphology and sensitivity to high-Z elements, an nmr scan can yield images that depend on physiology or functional properties. Most x-ray studies provide no distinction between a living subject and a cadaver; nmr scans can look totally different in the two cases.

Positron tomography

The information obtained from the imaging techniques we have discussed so far leads only indirectly to information about the chemical composition of the tissues imaged. Much disease, however, has a chemical origin; and changes in metabolism and biochemistry often come early, while changes in organ size, position or density generally occur late in the disease process. The use of radioactively labeled compounds to trace biochemical pathways has been a well-known research tool for quite some time. Applying the techniques of computed tomography allows one to map the distribution of the radionuclides.⁸ To do this successfully, one must use radionuclides that meet two requirements: They must mimic the

behavior of elements in the metabolic pathways and they must produce emanations whose paths one can determine. Positron emitters provide the natural choice for localization because the two 511-keV photons produced when the positron annihilates allow for easy reconstruction of the path. Metabolic equivalence then leads one to the short-lived positron emitters C^{11} , N^{13} , O^{15} and F^{18} .

Most positron tomography systems today look from the outside very much like x-ray tomography or nmr-imaging systems. Their discrete scintillators are arranged in rings; typically, there are about one hundred detectors per ring, with up to five rings in the gantry. A coincidence event between two detectors across the axis of the subject defines a line along which the positron annihilation must have occurred. The collection of millions of coincidence counts along the thousands of possible projection rays permits the reconstruction through the now-standard backprojection techniques we described above. One can improve the resolution by spacing the detectors closer together. A group at UCLA, for example, is studying closely packed bismuth germanate detectors, with each detector identified by cleverly adapted mercurous iodide photodiodes. Determining where along the ray the positron decayed can provide another improvement in resolution. This can be done with time-of-flight techniques and very fast counters.⁹ Fast scintillators such as CsF and BaF₂ can produce time resolutions better than 300 picosec and can greatly help in suppressing image noise.

The amount of radioactive tracer material required to produce useful images is minuscule. One can use C^{11} -labeled carbon monoxide to trace blood flow to detect, for example, motion abnormalities of the cardiac walls by observing the heart contractions in a cine loop. A few millicuries of radiocarbon (its half-life is 20 minutes) will suffice for this purpose. This amount of radioactivity corresponds to 2×10^{11} molecules of carbon monoxide, about $\frac{1}{3}$ of a picomole. Carbon monoxide is a potent poison, but a factor of 10^{10} times as much as required to produce a detectable physiological response.

Nuclear medical procedures—that is, those that depend on introducing radioactive tracers—differ from other radiologic imaging techniques in that they are fundamentally dynamic. They are also sensitive to extremely small concentrations and can pinpoint highly specific receptors. The measurement of cardiac volume we mentioned above is not the best example of the sensitivity of positron emission tomography. The fluid volume of the heart is a

simple mechanical property that can be obtained by many other imaging techniques; the receptors for carbon monoxide are simply hemoglobin, and are ubiquitous in the bloodstream.

An example of the sort of image unobtainable by other means is the use of rubidium-81 to trace the flow of blood to the heart muscle. In a normal heart this “myocardial perfusion” is about 50 ml/min for every 100 g of heart tissue. Rubidium chloride follows potassium pathways in the heart and is distributed by the blood flow. A dose of 16 mCi of Rb^{81} -labeled RbCl administered intravenously can provide a clear picture of the myocardial perfusion. Similarly, regional metabolic demands can be mapped through the kinetics of labeled fluorosugar analogs and fatty acids, probing the biochemical states of the heart and its fuel needs. Only nmr imaging can approach this type of information.

It is in the study of the brain, however, that positron tomography has made its greatest contributions. In the brain only oxygen and glucose serve as metabolic substrates. One can label glucose with carbon-11, but because the metabolic products of glucose (lactate and CO_2) are also mobile, the interpretation of an emission tomogram is difficult. However, a fluorinated analog of glucose, 2-fluoro-2-deoxyglucose can cross the blood-brain barrier but cannot be metabolized completely; it remains trapped¹⁰ as the phosphate, 2-FDG-6-phosphate. Labeling 2-FDG with fluorine-18 allows one to make positron-emission maps of glucose uptake by the brain. Figure 1 shows such a set of maps for different kinds of activity. The differences in brain meta-

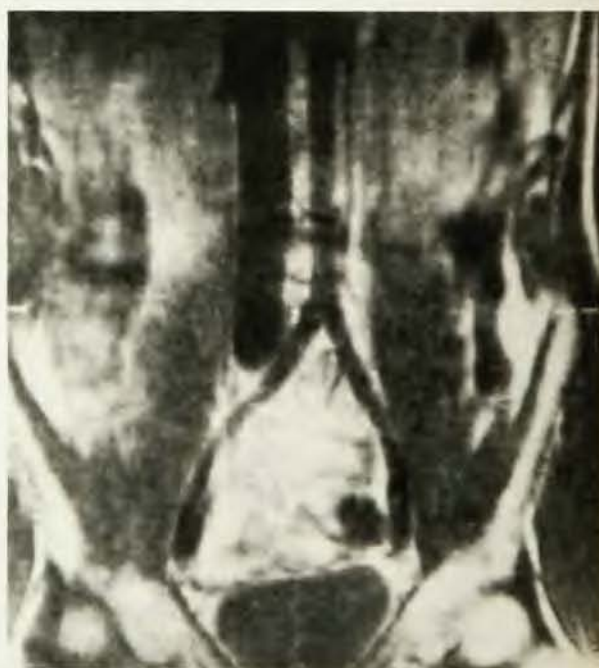
bolism associated with differences in information processing are evident.

Oxygen does not have an analog capable of being trapped. One can, however, set up a steady state in which the subject comes into equilibrium with O^{15} -labeled oxygen inhaled over a period of 5–10 minutes. (The half-life of O^{15} is 2 min.) The regional input-output differences are balanced by decay and the activity thus reflects local oxygen needs. Regional blood flow can simply be mapped with O^{15} -labeled water. An even clearer picture can be obtained by observing the decay of O^{15} activity after the subject inhales a brief pulse of labeled oxygen. Such maps can show detail comparable to that obtained with labeled glucose analogs.

Diagnostic ultrasound

The last of the techniques we would like to discuss offers perhaps the least invasive method of probing the soft tissues of the body to produce images or detect disease states: the use of brief pulses of ultrahigh-frequency acoustic waves as a sort of sonar to map tissues within the body. As it travels through the body, an ultrasound pulse is reflected and scattered by changes in density and elasticity at tissue interfaces. Plotting the echo strength as a function of time allows one to visualize tissues as a function of distance within the body. For one-dimensional information one can simply plot the echo strength as a function of delay (“A-mode” display). For two-dimensional information, obtained by sweeping the source and detector across the patient, one generally just plots a point on a screen for echoes above a threshold, the distances along the line corresponding to delays

An nmr image of a coronal slice of an abdomen. Differences in brightness of the image correspond to differences in the decay times of the proton spin resonances and thus reflect different chemical environments—differentiating, for example, fatty tissue from muscle. (Photo courtesy Anne Deery, Siemens) Figure 4



Radiological imaging systems

	Digital subtraction angiography	Computed tomography	Nuclear magnetic resonance	Positron computed tomography	Ultrasound
What is detected	transmitted x rays	transmitted x rays	rf radiation from induced emfs	511 keV annihilation photons	acoustic echoes
What is imaged	electron density of object	electron density of object	induced nuclear magnetization	<i>in vivo</i> positron emitters	discontinuities in speed of sound
What causes structure	variation of electron density with composition and density	variation of electron density with composition and density	variation of strength and decay of magnetization with composition	differential uptake of labelled compounds	variation of density and elasticity of tissues
What is inferred	location of tissues filled with absorbing material	sizes and shapes of organs	physical and chemical variations among tissues	flow and metabolism of tracer material	sizes and shapes of organs; acoustical properties
Typical application	examination of arterial narrowing	detection of brain tumors	imaging of brain tumors	mapping of glucose metabolism in brain	fetal growth; tumor detection; cardiology
Signal source	x-ray tube	x-ray tube	precession of nuclei	ingested labelled compounds	piezoelectric transducer
Signal detector	image intensifier	x-ray detector	rf pickup coil	scintillation detectors	piezoelectric transducer
Image plane	longitudinal	transverse	any	transverse	any
Spatial resolution	1/2 mm	1 mm	2 mm	10 mm	2 mm
Temporal resolution	10^{-2} sec	1 sec	10^{-1} sec to 10^2 sec	10^1 sec to 10^3 sec	10^{-2} sec
Typical radiation dose	2 rad (imaged field)	1 rad (imaged slice)	(not applicable)	10^{-2} rad (whole body)	(not applicable)
Cost (approximate)	\$0.2 million	\$1-2 million	\$1-2 million	\$1 million	\$0.1 million
Chief use	clinical diagnosis	clinical diagnosis	physiological and clinical research	physiological research	clinical diagnosis

of the echo ("B-mode" display).

The ultrasound beam is produced and echoes are detected by a piezoelectric transducer. Manual-scanning instruments have the transducer attached to an electromechanical scan arm. The arm constrains the motion of the transducer to a plane. It also provides signals used to compute the position and orientation of the transducer for determining the path of the ultrasound beam, and thus the reflector locations. More commonly, the beam is swept automatically by a small hand-held scanning assembly; the operator positions and orients the scanner in the desired plane. Most instruments have an image memory or scan converter that presents the image on a tv monitor, and one can usually observe the image as it is produced. There are several techniques available for performing the automatic scans: mechanical motion of the transducer, rotation of a sound-deflecting mirror, electronic switching between elements in an array, or phased excitation of a multi-element sector array.

The rapid scanning speed and the "real-time" images of the hand-held automatic scanners offer several advantages to the clinician. By providing flexibility and more or less instantaneous feedback, the sonographer can move the scanner to find landmarks, such as blood vessels, that can help pinpoint the location of the tissue or organ under investigation. Such landmarks are especially important when there is only subtle acoustic contrast between the structure of interest and the background. Once the structure is found, the sonographer can position the scanner to provide optimum contrast and clarity.

The maximum possible scanning speed of acoustic imaging systems is limited by the speed of sound in tissue (about 1540 m/sec), the maximum depth of tissues of interest and the detail required for the image, which determines the number of lines per scan. Rapid scanners are very useful in providing the real-time images and the flexibility we have mentioned. They are also essential in viewing moving objects such as the heart. One can now obtain ultrasound images of the heart at a rate of more than 30 scans per second. The images are clear enough that one can diagnose cardiac problems such as mitral valve stenosis and other valve diseases as well as congenital heart defects. Even faster scans are possible for examining superficial anatomic structures or for producing images with limited resolution.

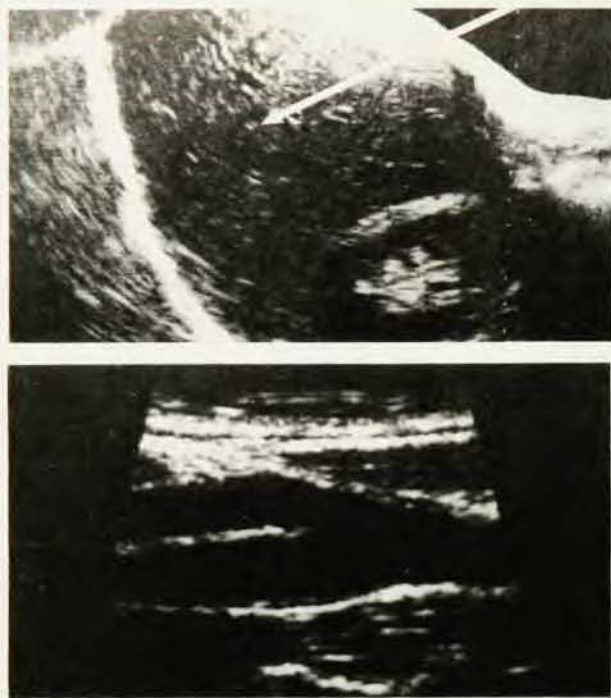
One can add a further dimension to ultrasonic scans by coupling measurements of Doppler shifts with the detection of echoes so that one can, for example, measure blood flow at the same time that one is imaging the vessels through which the blood is flowing. Such "duplex" transducers could be very valuable in examinations of the heart, to detect abnormally high blood velocities or abnormal flow directions through defective valves.

A significant fraction of all ultrasound examinations performed involve studies of the abdominal organs (figure 5) or examinations of the fetus. In the abdomen, the images may help determine whether abnormal masses are likely to be malignant. Ultrasound provides a method of rapidly detecting stones and other obstructions in the gall bladder and related structures. Variations in the overall texture of the

image of an organ are sometimes an indication of a disease condition. For example, early signs of rejection of kidney transplants may be detectable with ultrasound.

Fetal maturity studies—using measurements of head size or other anatomical dimensions from ultrasound scans to indicate the development of a fetus—are routine in most obstetrics departments and even in some physicians' offices. The resolution available with some devices allows one to distinguish fetal structures with sufficient detail permit diagnoses of many types of fetal abnormalities.¹³ Most obstetricians now perform amniocentesis—in which a sample of the amniotic fluid is withdrawn for laboratory analysis, particularly to detect genetic problems—under ultrasonic guidance. By following the position of the probing needle on a screen, the obstetrician can avoid puncturing the fetus, placenta or umbilical cord.

The spatial resolution in an ultrasound image is related to the volume occupied by the sound pulse as it propagates through tissue. Ordinarily this volume can be divided into two components: one that depends on the lateral dimensions of the transducer beam and a second that depends on the time duration of the pulse, that is, its axial dimension. A typical instrument may use a 3.5 MHz transducer with a circular aperture 20 mm in diameter, focused at a depth of 12 cm. The diffraction-limited lateral resolution in the transducer's focal region is about 2.6 mm. The axial resolution for such a transducer is usually on the order of 1 mm. For a scanner whose focal length is fixed, the lateral resolution is, of course, less for reflectors located away



Ultrasound images.

Above, an image of the abdomen of an adult; the bright band at upper right indicates the skin; the curved band at the left and bottom is the diaphragm; the liver is indicated by an arrow. Below, an image of the carotid artery, showing its bifurcation (into the external and internal carotid arteries) in the neck; such an image can be combined with measurements of the Doppler shift to give further information about blood flow through the artery. (Lower image courtesy of J. Buschell, Diasonics.) Figure 5

from the focal region.

One can improve the resolution by using higher frequencies. Unfortunately, however, higher frequencies are more strongly absorbed; the attenuation of ultrasound in soft tissues increases approximately linearly with frequency. The short-wavelength path to improved resolution can thus be used only for superficial structures, where the beam need not penetrate deeply into the body. One can, for example, use sound frequencies as high as 15 MHz to examine the eye for tumors or detachments of the retina. The lower part of figure 5 shows an image of the bifurcation of the carotid artery (in the neck) obtained with a scanner operating at 8 MHz; the image was produced in "real time." Such images can show atheromatous plaque (which can tear loose and cause a stroke by blocking a smaller artery in the brain) early in its formation. Small calcifications in the arterial walls cast a recognizable acoustic shadow over deeper structures. Measurements of Doppler shifts, as we have mentioned, can provide additional information on the blood flow through the arteries, showing evidence of higher flow speeds through narrowed sections or turbulence associated with protrusions into the vessel.¹⁴

Recent technical improvements in diagnostic ultrasound equipment include incorporation of digital processing and digital image memories, improvements in the design of single-element and array transducers, fabrication techniques that increase the sensitivity and reduce off-axis radiation, and the development of special-purpose scanners for the breast, for

superficial examinations and for other applications. In spite of all the advances that have been made, ultrasonography currently uses only part of the potential information that might be extracted from echo signals. The contrast and detail on B-mode images are based on the magnitude of echo signals reflected by soft tissues. In the future it may be possible to extract additional diagnostic information by applying more sophisticated signal processing to the returning echo signals, using, for example, waves scattered in directions other than backwards to construct the image, or using more information from the backscattered wave in computerized reconstructions and to give quantitative characterizations of the tissues being examined.¹⁵

To produce such more sophisticated images, we will have to understand better the interactions of high-frequency sound waves and soft tissues. The human body is acoustically a very complex structure, with many sources of scattering, reflection, and refraction. The bulk of our knowledge of the quantitative transmission characteristics of human tissue, including speeds of sound, ultrasonic attenuation and ultrasonic scattering, stems from measurements carried out on excised human and animal organs.¹⁶ However, it is not always clear that the behavior of sound waves *in vivo* is always the same, and experimenters are now developing methods to make measurements of ultrasound characteristics in living subjects. It may become possible to observe specific tissues by their specific ultrasonic signatures. One useful tool¹⁷ in such studies is an acoustic

"phantom." Such phantoms are built to have precisely known acoustic properties at all points and to mimic to some extent the behavior seen in living organs; they are valuable in testing imaging and signal-processing algorithms and for checking¹⁸ theories on how beams of ultrasound interact with soft tissues.

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