Physics and applications of medical imaging

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Medical imaging is the principal method for noninvasively obtaining anatomic and physiologic information about the human body. Imaging has experienced a quantum leap in technology and clinical applications over the past 25 years. This leap includes x-ray computed tomography (CT); emission computed tomography (SPECT and PET); magnetic-resonance imaging (MRI) and spectroscopy (MRS), including functional MRI (fMRI), and the networking of images in digital networks (PACS and IMACS). Even traditional projection x-ray imaging is undergoing a major change with the advent of digital x-ray image receptors. Images are important not only to the detection and diagnosis of disease and injury, but also to the design, delivery, and monitoring of treatment. The evolution of medical imaging is the product of physicists working in collaboration with engineers and physicians. Further advances are limited only by the creativity and imagination of these individuals. [S0034-6861(99)03702-2]

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I. PROJECTION RADIOGRAPHY

A. Introduction

The discovery of x rays by Wilhelm Rontgen in 1895 opened new pathways for the detection and diagnosis of disease in humans. Before this discovery, most diagnoses were made from a verbal description of the patient's history and symptoms, combined with the use of the physician's senses to detect peculiar odors, unusual visual signs, abnormal sounds, unnatural physical sensations, and, occasionally, odd tastes. X rays provided a novel approach to patient examination whereby the physician could study the internal anatomy and physiology of the patient. Today, x rays are used in hospitals, clinics, offices, and emergency facilities worldwide, and contribute essential information for the detection and diagnosis of a wide spectrum of illnesses and injuries in millions of patients each year.

A ''projection image'' is formed by x rays transmitted through a region of the body following their release from an x-ray tube. Each point in the projection image reveals the intensity of x rays directed towards the point, modulated by differences in density and atomic number of various tissue constituents in the path of the x-ray beam. The projection image is a two-dimensional depiction of a three-dimensional distribution of tissue constituents, with the third (depth) dimension of the body represented as overlapping shadows in the image plane. Mentally reconstructing the third dimension of the image is one of the challenges of learning radiology, the science and art of image interpretation in medicine.

B. Analog x-ray image receptors

The projection image formed by transmitted x rays is usually captured on photographic film. Film alone may be used as the receptor when images of exquisite spatial resolution are desired, such as for detection of hairline fractures in bones of the extremities. However, the film's thin layer of photographic emulsion is a relatively inefficient x-ray absorber. Film alone as a receptor requires long exposure times and, for most applications, yields unacceptable image blurring caused by voluntary and involuntary patient motion. In most cases, the film is sandwiched between fluorescent ''intensifying'' screens that absorb x rays with up to 50% efficiency, and in turn emit visible light that exposes the photographic emulsion on the film.

The first intensifying screens were developed by Edison and contained calcium tungstate ($CaWO_4$) as the light-emitting ingredient. In the 1970s, these were replaced by screens containing a rare-earth element such as gadolinium, lanthanum, or yttrium complexed with oxysulfide or oxybromide crystals embedded in a plastic matrix. Compared with $CaWO_4$, rare-earth screens are better absorbers of x rays, and some emit more light for each x ray absorbed. Today, most x-ray images are captured on x-ray film sandwiched between rare-earth intensifying screens in an imaging cassette that is tightly sealed to provide intimate contact and reduce geometric blurring between the film and screens. Cassettes are available for film of different sizes ranging from about $10\times10\,cm^2$ to approximately $35\times45\,cm^2$ used principally for chest imaging. Once the film has been exposed to x rays or light from the intensifying screens, it contains a ''latent image'' that can be made visible by chemical

processing. In the final image, darker areas represent anatomic regions penetrated by a greater number of x rays, whereas lighter areas depict regions where fewer x rays have been transmitted. Often, the lighter areas reveal bony structures or regions where a contrast agent is present, because the higher density and atomic number of the bone or contrast agent causes greater x-ray absorption.

C. X-ray fluoroscopy

Projection radiography yields exquisite two-dimensional images that present a ''snapshot'' of the patient's anatomy at a particular moment in time. Although these images are often sufficient to detect an abnormal condition and lead to a diagnosis of the cause of the abnormality, they are limited in their ability to depict rapid changes in the anatomy caused by underlying physiologic processes. For this purpose, a continuous x-ray image is required. The technique that yields such a continuous image is termed ''fluoroscopy'' because it captures the image instantaneously on a fluorescent screen and displays the image in real time to the viewer.

Early applications of fluoroscopy employed a fluorescent screen that was viewed directly by the physician. The image was so dim that it could be seen only after the observer's vision had been ''dark adapted,'' and even then only gross features in the image could be distinguished. Fluoroscopy was improved in the 1950s with invention of the image intensifier. In this device, transmitted x rays are captured on a CsI intensifying screen and converted instantaneously into a two-dimensional distribution of electrons ejected from a photocathode juxtaposed to the CsI screen. The electrons are accelerated through 25–35 kV onto an output screen that emits a small image in response to the impinging electrons. The gain in image brightness (on the order of 50 000×) in the image intensifier alleviates the need for dark adaptation. The image on the output screen may be viewed directly through an optical system of lenses and mirrors, or captured by a television camera and transmitted electronically to a remote television monitor for viewing. This process converts the light image into an electronic signal, which can then be digitized for integration into an imaging network.

Fluoroscopy is an important part of x-ray imaging, especially for studies of the gastrointestinal tract and for angiographic studies of the central and peripheral circulatory system. It is used for image-guided therapeutic approaches such as interventional radiology and minimally invasive surgery that are growing in popularity because of their reduced patient morbidity and, usually, lower cost compared with alternative therapeutic approaches.

D. Digital x-ray image receptors

The combination of intensifying screens and film has many advantages as an image receptor for projection radiography. It is simple, portable, inexpensive, and yields images with excellent anatomic detail. The method has some disadvantages as well. Its use is restricted to a narrow range of exposures, and sometimes repeat examinations are necessary because films are over- or under-exposed. Film images are bulky to store and easy to misplace, and they must be physically transported from one location to another. Replacing film-based x-ray image receptors with digital x-ray detectors reduces these problems, and offers other advantages as well such as (1) image processing to improve contrast, sharpen edges, and reduce image noise, (2) integration of x-ray images with those from other digital imaging methods such as computed tomography, nuclear medicine, and magnetic-resonance imaging, (3) electronic transmission of x-ray images within the institution to provide immediate access for individuals caring for patients, (4) electronic transmission of images to and from distant locations to improve the care of patients in remote areas (teleradiology), and (5) more effective use of algorithms for computer-assisted diagnosis.

Three approaches to digital x-ray imaging are currently available commercially. One approach is to capture the projection image with an image intensifier, and to digitize the resulting signal from a video camera optically coupled to the intensifier's output screen. This approach yields a continuous digital image, but has limited spatial resolution compared with images on x-ray film. Another approach is to use a phosphor screen that forms a latent image by trapping electrons excited by the absorption of the incident x rays. By subsequently illuminating the phosphor with a scanning laser beam, the trapped electrons escape and release blue light that is captured by a photomultiplier tube to yield an electronic signal that can be digitized. This approach provides single images like those obtained with x-ray film and intensifying screens, except with reduced spatial resolution. The third commercial approach uses an amorphous selenium (aSe) photoconductive screen to convert incident x rays directly into a distribution of charge carriers on the screen's surface. This technique employs the photoconductor (aSe) widely used in earlier times for photocopying, and resembles xeroradiography, a method used in the 1970s for breast imaging. In the present case, however, the charge distribution is converted into a digital readout. The commercial version of the aSe approach uses a large rotating drum that is too large to be housed in most existing examination rooms (Neitzel, Maack, and Guenther-Kohlfahl, 1994).

Although still experimental, flat-panel display technologies are rapidly improving and show considerable promise as digital x-ray receptors capable of providing immediate images for radiography and fluoroscopy with exquisite spatial resolution. One approach is the use of an x-ray sensitive, light-emitting phosphor that is optically coupled to an array of photodetectors (e.g., aSi:H) to yield digitizable electronic signals (Fig. 1). A more direct approach is the use of a photoconductor such as aSe combined with an array of flat-panel thin-film transistors (termed an active matrix array) to yield a ''direct conversion'' image receptor in which the projection

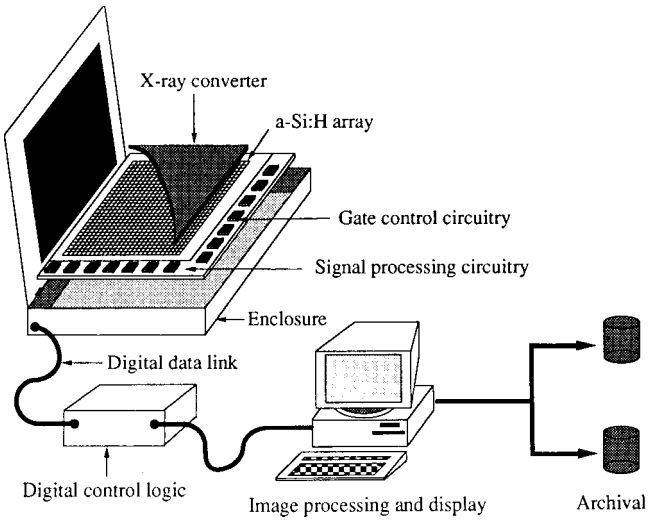

FIG. 1. A flat-panel phosphor/photodiode detector for digital x-ray projection imaging (from Antonuk *et al.*, 1995).

x-ray image is converted directly into a digitizable charge distribution over a large-area plate. The five requirements of an ideal x-ray photoconductor are well met by aSe: (1) the relatively high atomic number Z of aSe provides efficient x-ray absorption; (2) only a small amount of energy is required to create an electron-hole pair in aSe; (3) aSe has negligible dark current; (4) the charge carriers can migrate a considerable distance along the applied electric field in aSe without being trapped; and (5) since most of the impinging x-ray energy is absorbed in aSe, images are produced at relatively low doses (Rowlands and Kasap, 1997). Other possible photoconductors include lead iodide, thallium bromide, and cadmium-zinc telluride. Although the relatively low Z of organic photoconductors limits their x-ray absorption efficiency, it is conceivable that they could be used in an organic binder containing high-Z x-ray absorbing particles (Wang and Herron, 1996).

A major advantage of the direct-conversion approach is its structural flexibility that allows fabrication of large-area detectors for medical x-ray imaging. This approach to digital x-ray imaging has yielded spatial resolutions on the order of 150 μm, and has the potential to achieve resolutions as fine as 50 μm coincident with the most exacting requirements of x-ray imaging. With continued research and demonstration of efficacy through clinical applications, direct-conversion, flat-panel receptors could be the pathway over which x-ray projection imaging progresses into the digital era.

II. X-RAY TOMOGRAPHY

A. Introduction

In projection radiography, the third (depth) dimension of tissue is represented as overlapping shadows in a two-dimensional image. As a result, an anatomic structure of interest is frequently obscured by shadows of objects above or below it in the patient. Removing these shadows often improves the delineation of the shape and composition of the structure. Two approaches, analog tomography and computed tomography, can be used to remove the shadows. Over the past two decades, computed tomography has grown widely in acceptance, and analog tomography has declined in popularity.

B. Analog tomography

In analog tomography, anatomic structures in a specific image plane (actually an image section of specific thickness, often a few mm) in the patient are kept in focus in the image, while structures above and below the image plane are blurred. The blurring is accomplished by moving the x-ray tube and the image receptor in synchrony during exposure around a pivot (fulcrum) in the image plane. The depth of the image plane in the patient can be altered by moving the patient up or down with respect to the fulcrum. Analog tomography is limited by the presence of image artifacts (tomographic ghosts) that interfere with image clarity, and by low image contrast caused by scattered radiation and the inability of conventional x-ray image receptors to detect differences in x-ray intensity of less than a few percent. Because of its low cost and simplicity, analog tomography is still used today in certain applications such as the acquisition of supplemental information about suspicious areas seen in chest radiographs. However, most applications of analog tomography have been superseded by computed tomography.

C. Computed tomography

The first commercial computed-tomography (CT) unit was announced in 1972, and reflected the pioneering work of the Austrian mathematician Radon, the South African physicist Cormack, and the English engineer Hounsfield. The latter two individuals shared the 1979 Nobel Prize in medicine for their contributions to CT. The first unit employed a narrow beam of x rays scanned across the patient in synchrony with a scintillation detector moving on the patient's opposite side (Hounsfield, 1973). The intensity I of x rays measured by the detector is

$$I = I_0 \exp[-\Sigma \mu_i x_i],$$

where μ_i represents the linear attenuation coefficient of each of "i" structures in the path of the narrow x-ray beam, and x_i represents the thickness of each of the "i" structures.

With a single measurement of x-ray transmission, the separate attenuation coefficients cannot be determined. However, these coefficients can be distinguished if enough transmission measurements are obtained at different orientations through the patient, with the aid of calculations using some type of back-projection algorithm. The result of such calculations is a two-dimensional map of linear attenuation coefficients distributed across the imaging section with a thickness

defined by the width of the scanning x-ray beam. These coefficients can be converted into CT numbers with the formula

$$\text{CT number} = 1000[\mu - \mu_w]/\mu_w ,$$

where μ is the linear attenuation coefficient at a specific location in the section, and μ_w is the coefficient of water for the x-ray energy employed for CT scanning. The distribution of CT numbers across the section can be displayed in various shades of gray to yield an image of the distribution of tissues in the section, each with its own CT number. This image is referred to as a CT image.

In early CT units, the x-ray tube and scintillation detector scanned the patient along a linear path perpendicular to the axis of rotation of tube and detector. Transmission data were subjected to an iterative method for computing attenuation coefficients. The process of acquiring transmission data and producing a gray-scale image was too time consuming for a busy clinical environment. The computational problem was solved by developing faster algorithms using a convolution (filtered back projection) model for image reconstruction. This approach subjects the transmission data to a Fourier transform into frequency space, permitting use of ramp and cutoff-frequency filters to improve image quality and enhance subtle features in the image. The acquisition time for x-ray transmission data was shortened markedly by development of purely rotational CT units that complete the entire scanning process in a few seconds. Combining this motion with simultaneous movement of the patient along the axis of rotation permits accumulation of many cross-sectional images during one relatively short examination period. This process, known as ''spiral'' or ''helical'' scanning, has significantly expanded the applications of CT, especially in the thorax and abdomen. Spiral scanning yields a three-dimensional array of CT numbers, and images parallel (''sagittal'' and ''coronal'' slices), perpendicular (''transaxial'' slices), or at any angle to the long axis of the patient, by compiling arrays of attenuation coefficients across the corresponding planes. The three-dimensional database can be configured to yield images that appear to be three-dimensional, and windowed to provide images of specific ranges of CT numbers corresponding to selected tissues.

Although rotational CT units can produce images in a few seconds, they are unable to acquire data quickly enough ($<$0.1 s) to capture images of the heart and other blood-perfused organs without significant blurring caused by motion. For these images, a way is needed to acquire x-ray transmission data from various angles without mechanical motion of the scanner. A scanner designed for this purpose employs an electron gun that scans a stationary metal annulus to generate x-ray beams along different projections. The resulting examination times are as short as 50–100 ms (Hendee and Ritenour, 1992).

III. EMISSION TOMOGRAPHY

In nuclear imaging, a small amount of a radioactive pharmaceutical is administered to the patient. The pharmaceutical carries the radioactivity to different organs or tissues according to its biokinetic properties. As the radioactive ''tag'' decays, it emits γ rays or, in the case of positron imaging, annihilation photons produced during annihilation of positrons released by the tag. As the emitted radiation escapes from the body, it is detected by one or more scintillation detectors positioned near the patient. In conventional nuclear imaging, the signals from the detectors are processed to yield a two-dimensional planar image of the three-dimensional distribution of radioactivity inside the patient. In emission tomography, two-dimensional cross-sectional images are reconstructed from multiple projections obtained at different angles around the patient. Emission tomography with radioactive pharmaceuticals emitting γ rays is referred to as ''single-photon emission computed tomography (SPECT).'' When annihilation photons are imaged following positron decay of radioactive pharmaceuticals, the technique is termed ''positron tomography (PET).'' Until recently, PET had the advantage of using coincidence measurement of annihilation photons to yield much higher radiation-detection efficiencies compared with SPECT.

An exciting recent development in nuclear imaging is operation of multiple-detector SPECT cameras in coincidence mode to yield images of positron-emitting radioactive pharmaceuticals. This potential is reinforced by regional supplier networks for ^{18}F-labeled deoxyglucose that obviate the need for an on-site cyclotron to produce positron-emitting ^{18}F. These developments are enhanced by growing recognition of the usefulness of positron imaging for detecting and staging cancer in a variety of anatomic sites, including brain, breast, and lung. It is conceivable that patients at high genetic risk for cancer will someday be administered ^{18}F-labeled deoxyglucose and scanned at periodic intervals for early detection of cancer.

Advances in molecular biology and genetics are yielding new knowledge at an astonishing rate about the molecular and genetic infrastructure underlying human health and disease. New knowledge about receptor sites, metabolic pathways, and ''antisense'' molecular technologies promises to yield increasingly specific agents that can be tagged with radioactive markers to permit visualization of normal and abnormal tissue structure and function at microscopic levels. These possibilities, referred to collectively as ''molecular medicine,'' have the potential to enhance the contributions of nuclear imaging to clinical medicine.

IV. MAGNETIC RESONANCE

A. Introduction

Damadian used nuclear magnetic resonance (NMR) in 1971 in an effort to distinguish normal from cancerous tissue in rats, and extended these studies to humans in 1973 (Damadian, 1973). That same year, Lauterbur published the first magnetic-resonance images (Lauterbur, 1973). The first human images were acquired in 1977

(Hinshaw, Bottomley, and Holland, 1977). The first magnetic-resonance image of the human brain was demonstrated in 1980 (Holland, Moore, and Hawkes, 1980). Unlike x-ray imaging, magnetic-resonance imaging does not depend on the transmission through tissue of radiation from an external source. Instead, the tissue itself is the source of imaging signals that arise from the macroscopic spin magnetization **M** of polarized water protons in tissue. Far less frequently, the signals may originate from other nuclei such as phosphorus or sodium. The motion of the magnetization vector **M** of uncoupled spins of water protons is given by the Block equation

$$d\mathbf{M}/dt = \gamma\mathbf{M}\times\mathbf{H} - M_{xy}/T_2 - (M_z - M_0)/T_1$$

where γ is the gyromagnetic ratio of hydrogen, **H** is the effective magnetic field, T_1 is the spin-lattice relaxation time defined as the time constant for the longitudinal magnetization M_z to return to its equilibrium value M_0 following receipt of a radio-frequency (rf) pulse that orients it at 90°, and T_2 is the spin-spin relaxation time defined as the time constant for decay of the coherent magnetization M_{xy} in the transverse plane to the equilibrium value $M_{xy}=0$. Both T_1 and T_2 are affected by interactions of water with tissue molecules. By depicting these effects in magnetic-resonance images through selected sequences of rf pulses, T_1 and T_2 can be made to contribute independently by varying degrees to the contrast among different tissues in magnetic-resonance images.

B. Magnetic-resonance imaging

At the heart of a magnetic-resonance-imaging (MRI) system is a magnet that provides a highly stable and uniform magnetic field for nuclear polarization. Although resistive and permanent magnets have been used, most MRI units employ superconducting magnets. These magnets provide field strengths typically between 0.3 and 2 tesla, and the NbTi alloy conductors must be cooled to superconducting temperature (~4 K) with liquid helium. A natural next step in the evolution of MRI is the use of high-temperature superconductors such as bismuth-strontium-calcium-copper oxide/silver (BSCCO/Ag) that can accommodate greater currents at higher temperatures (National Research Council, 1996).

A radio-frequency coil serves two purposes in MRI. First, by sending a brief rf pulse into a region of tissue, it misaligns the magnetization vector of water protons with respect to the applied magnetic field. Second, it receives the weak rf signal emitted by the tissue as the magnetization vector realigns with the field according to the relaxation times T_1 and T_2. The design of rf coils has been largely experimental and focused primarily on the goals of improving signal quality and data-acquisition rate. Future developments may include improved computational design, cooled or superconducting rf coils to reduce noise, and multiple coils operated in parallel to improve data-acquisition efficiency.

Spatial localization of the rf signals emitted from tissue is accomplished by spatially encoding the signals through use of gradient magnetic coils that impose linear magnetic gradients along three-dimensional orthogonal axes in the tissue. This gradient approach is also used to encode dynamic information related to studies of blood flow and diffusion. Factors to be considered in the design of gradient coils include geometry, size, anatomic region of interest, desired gradient strength, efficiency, inductance, eddy currents, gradient uniformity, forces and torques on the coils, heat dissipation, and nerve stimulation in tissue (National Research Council, 1996).

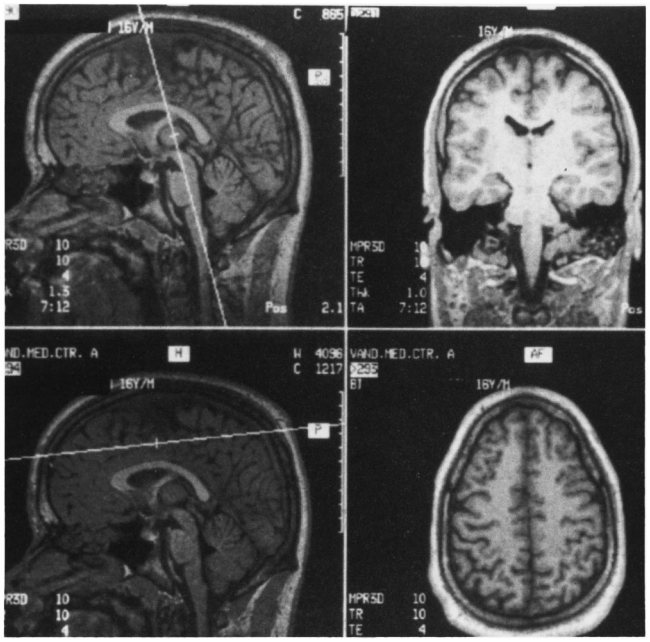

FIG. 2. MRI anatomic images through different planes from a 128×256×256 three-dimensional dataset (from Price, 1995).

Although relatively new, MRI is an exceptionally powerful imaging technique in clinical medicine. To date it has been used principally for anatomic imaging (Fig. 2), although flow and diffusion images are growing in popularity. Functional MRI (fMRI) is a rapidly developing area with significant clinical potential. This technique exploits the paramagnetic behavior of deoxyhemoglobin in red blood cells as an intrinsic intravascular contrast agent. When in a magnetic field, a blood vessel containing deoxyhemoglobin distorts the field in its immediate environs, with the degree of distortion increasing with the concentration of deoxyhemoglobin. This distortion affects the behavior of water protons in the environs and, consequently, the magnetic-resonance signal arising from these protons.

Neural activation of a region of the brain stimulates increased arterial flow of oxygenated blood, thereby decreasing the concentration of deoxyhemoglobin in the region. This decrease affects the immediate magnetic field and changes the intensity of the magnetic-resonance signals from the region. Changes in the magnetic-resonance signal can be detected and displayed as functional-MRI images (Fig. 3). These images, termed BOLD (blood-oxygen-level dependent) images,

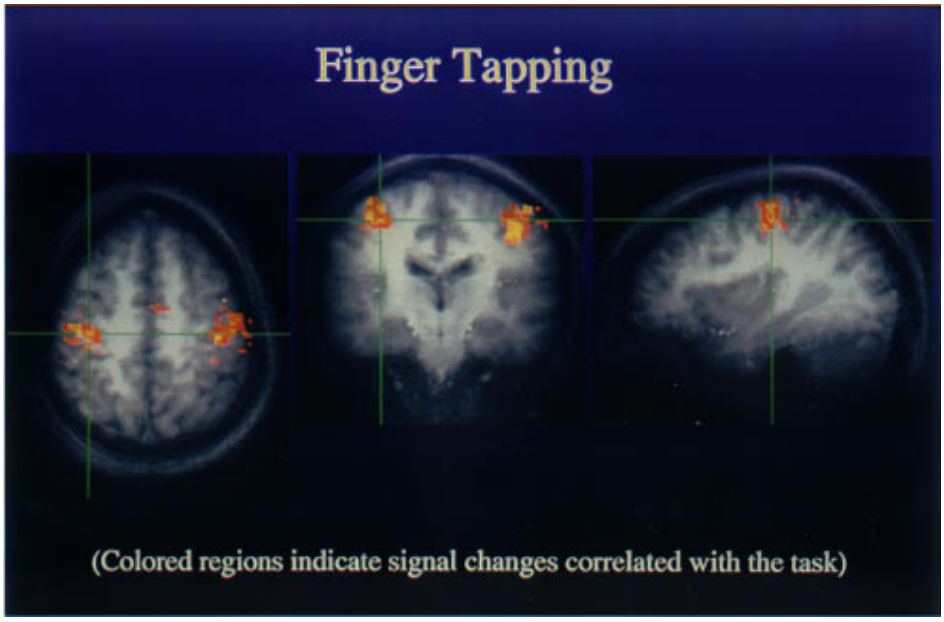

FIG. 3. (Color) Orthogonal fMRI images of the human brain averaged over five subjects showing neural activation of the motor cortex (light areas in the image) associated with finger tapping (Birn, 1998).

are useful in mapping functional neural activities onto the cerebral cortex and studying such activities in response to various somatosensory and cognitive tasks. Functional MRI is an exciting and fast-evolving technology that could benefit immeasurably from improvements in areas such as (1) motion detection and compensation, (2) characterization of temporal response patterns, (3) characterization of physiological noise and its effects of functional MRI, and (4) display of volumetric functional-MRI images, especially in real time.[1]

C. Magnetic-resonance spectroscopy

Combining MRI and magnetic-resonance spectroscopy permits noninvasive acquisition of unique *in vivo* information about the chemical composition of human tissues. Through the technique of chemical-shift imaging (CSI), spatial and temporal changes in tissue function can be studied by examining the magnetic-resonance spectra usually of ^{1}H, but occasionally of other nuclei such as ^{13}C, ^{14}N, ^{15}N, ^{19}F, ^{23}Na, ^{31}P, ^{39}K, ^{35}Cl, and ^{37}Cl. Compared with ^{1}H, these other nuclei occur less frequently in tissue and yield much weaker magnetic-resonance signals that are difficult to separate from noise. Magnetic-resonance spectroscopy is being investigated especially for its ability to distinguish benign from malignant tumors and recurrent tumors from scar tissue caused by earlier radiation therapy. It also is promising as a method to monitor tumor regression during radiation therapy and chemotherapy.

[1]See the Kleppner article in this volume for a discussion of the use of optical pumping of He^{3} to enhance MRI imaging in lung diagnostics.

V. IMAGE NETWORKING

Most medical imaging systems (CT, MRI, ultrasound, nuclear medicine) present imaging data in digital form. The major exceptions to this rule are projection radiography and fluoroscopy. As described earlier, these applications are also becoming digital. Several advantages can be achieved by linking digital imaging systems electronically in a ''picture archiving and communications system (PACS)'' [referred to as an ''image management, archiving and communication system (IMACS)'' when integrated with other information networks such as hospital and clinic information systems]. These advantages include the capability to store images from many units electronically in one location, retrieve and assemble images for comparative studies from different techniques used to examine the same patient, and transmit images to remote locations for viewing by specialists without loss of the central data file. A PACS has the potential to expedite diagnoses, improve diagnostic accuracy, enhance information transmission to other physicians, and eliminate misplaced and lost films. Many imaging specialists believe that a fully integrated PACS will be essential to operation of a radiology service in the future. Others are convinced that the cost (several million dollars for a typical imaging service) will prevent many departments from converting completely to PACS, at least in the near future. They believe that conversion will most frequently occur incrementally, with film remaining as the preferred image receptor for certain applications for some time to come.

VI. MEDICAL IMAGES IN RADIATION TREATMENT PLANNING

Medical images are used to detect, diagnose, and stage many types of cancer. They also are used to de-

sign, guide, and monitor the treatment of cancer, and to follow the patient after treatment to detect possible recurrence of the disease. Cancer is a disease characterized by the uncontrolled proliferation of cells. Cancer is treated by removing the cancerous cells, either through surgical extraction or by killing them with poisons (chemotherapy) or ionizing radiation (radiation therapy). Although medical images are important to all three approaches, they are employed most widely in radiation therapy. Their application to radiation therapy is discussed here.

The successful treatment of cancer with radiation requires that the cancer is localized to a specific region of tissue, and that the cancer and its microscopic extensions into normal tissue receive a dose of radiation sufficient to kill the tumor cells, while keeping the dose to nearby normal tissues low enough to avoid serious complications. Achieving this balance between radiation doses to tumor and normal tissues demands careful planning and precise treatments, including accurate delineation of the margins of the cancer and identification of the location of nearby radiosensitive normal tissues. These needs frequently are met by incorporating CT and MRI into the treatment planning process. CT and MRI images provide not only a cross-sectional picture of the patient's anatomy, including the cancer and surrounding tissues, but also an accurate representation of the body contour and organs that are especially sensitive to radiation. The digital data from these imaging units can be entered directly into the treatment-planning computer, and proposed treatment plans can be superimposed onto the cross-sectional images. Images acquired over the course of treatment can be examined to monitor the regression of the cancer and to make adjustments in the treatment plan in response to changes in the patient's anatomy.

In most radiation-therapy services, commercial imaging units are used to generate cross-sectional information for treatment planning. To improve the alignment of the cross-sectional information with the actual geometry encountered during treatment, some physicists have built a CT scanner on a gantry identical to that supporting the linear accelerator used for radiation therapy. Although these CT units yield spatial resolution inferior to that of commercial CT units, they duplicate the treatment geometry and provide images that are good enough for treatment planning.

Many x-ray treatments consist of multiple fixed radiation fields converging on the cancer from different directions. This approach concentrates the radiation dose in the cancer while delivering much lower doses to surrounding normal tissues. In some cases an even better dose distribution can be achieved by rotating the treatment machine partially or completely around the patient during treatment so that only the cancer is always in the path of the radiation beam. Since cancers are asymmetrical, the size of the x-ray beam should be expanded and contracted continually during treatment in order to restrict the dose to normal tissues to the lowest possible level. Often the dose distribution can be improved even more by varying the dose rate during rotation. This approach is referred to as ''conformal therapy.'' Its applications require detailed three-dimensional knowledge of the anatomy of the irradiated tissue through accumulation of medical images, together with exquisite computerized control of the treatment unit and patient couch (Mageras *et al.*, 1994).

A further advance in conformal therapy would be the convergence of CT imaging and x-ray therapy into a single gantry, so that tomographic images could be used to monitor treatment alignment and dose distribution continually as the treatment progresses. This hybrid approach is being pursued by a few medical physicists (Convery and Rosenbloom, 1995; Mackie *et al.*, 1993); it presents formidable technical challenges as well as promises for improved radiation therapy.

VII. CONCLUSIONS

Medical imaging celebrated its centennial anniversary in 1995, and today it continues to push the frontiers of research and clinical applications forward. It is an excellent example of what can be done through a multidisciplinary effort, in this case involving physicists, engineers, and physicians. Many research opportunities are available, as much remains to be done in further improving the applications of medical imaging to reducing human disease and disability.

REFERENCES

Antonuk, L., *et al.*, 1995, Radiographics **15**, 999.
Birn, R., 1998, Medical College of Wisconsin (unpublished).
Convery, D., and M. Rosenbloom, 1995, Phys. Med. Biol. **40**, 979.
Damadian, R., 1973, Ann. (N.Y.) Acad. Sci. **222**, 1048.
Hendee, W. and E. Ritenour, 1992, *Medical Imaging Physics* (Mosby-YearBook, St. Louis).
Hinshaw, W., P. Bottomley, and G. Holland, 1977, Nature (London) **270**, 722.
Holland, G., W. Moore, and R. Hawkes, 1980, J. Comput. Assist. Tomogr. **4**, 1.
Hounsfield, G., 1973, Br. J. Radiol. **46**, 1016.
Lauterbur, P., 1973, Nature (London) **242**, 190.
Mackie, T., T. Holmes, S. Swerdloff, P. Reckwerdt, J. Deasy, J. Yang, B. Paliwal, and T. Kinsella, 1993, Med. Phys. **20**, 1709.
Mageras, G., Z. Fuks, J. O'Brien, L. Brewster, C. Burman, C. Chui, S. Leibel, C. Ling, M. Masterson, and R. Mohan, 1994, Int. J. Radiat. Oncol., Biol., Phys. **30**, 971.
National Reserve Council, 1996, Mathematics and Physics of Emerging Biomedical Imaging (National Academy Press, Washington, DC).
Neitzel, U., I. Maack, and S. Guenther-Kohlfahl, 1994, Med. Phys. **21**, 509.
Price, R., 1995, Radiographics **15**, 175.
Rowlands, J., and S. Kasap, 1997, Phys. Today **50**, 24.
Wang, Y., and N. Herron, 1996, Science **273**, 632.