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Principles and Applications of Magnetic Resonance Imaging (MRI) in Neurology and Neurosurgery

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Magnetic Resonance Imaging has evolved from NMR Spectroscopy, a chemical analysis technique developed 40 years ago, into a sophisticated imaging modality which is rapidly becoming a major factor in the diagnosis of brain disease. The technique uses the properties of the interaction of certain spinning nuclei with applied magnetic fields to observe the behavior of these nuclei after being stimulated by radio-frequency radiation. Application of secondary (gradient) magnetic fields allows the spins to be frequency encoded with respect to their position in an object. The radio frequency signals produced by the spinning neclei are detected and analysed by a computer to form images. The sophisticated high field-strength magnets used in these systems dictate that special provisions must be made for their installation in a hospital environment. Specific examples of MR images are presented with respect to stereotactic surgery planning, diagnosis of Multiple Sclerosis and spinal cord lesions, correlation with other imaging modalities, and in-vivo phosphorous spectroscopy.

Magnetic resonance imaging (MRI) is a technique which monitors the behaviour of the nuclei of various atoms within the volume being studied. It does so by detecting effects generated by the spin associated with these nuclei. Magnetic resonance, as a chemical analysis technique, was pioneered by Bloch (1946) and Purcell, Torrey, and Pound (1946). For the next 25 years, Nuclear Magnetic Resonance Spectroscopy became a technique which was employed for the chemical analysis of a wide variety of compounds. Its use in in-vivo situations had its beginnings in the work of Lauterbur (1973) who proposed an imaging technique based on the already well established algorithms for resconstructing CT images, and the work of Damadian, Minkoff, Goldsmith, Stanford, and Koutcher (1976) who investigated the NMR relaxation times of tumourous tissues in-vivo.

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The technique of magnetic resonance imaging has been increasing in use in medicine over the past few years, and today there are in excess of 500 clinical units installed world wide. It is used for imaging all parts of the body, but most extensively in the brain and spinal cord.

The present paper is divided into two sections. The first section gives a brief non-mathematical outline of the principles behind the imaging technique, while the second section focuses on some of the exciting clinical applications to which the technique has been adapted.

Basic Principles

Nuclei containing an odd number of protons, an odd number of neutrons or both, have a spin. It is a fundamental fact of physics that any charged particle in motion has a magnetic field associated with it. Since the nuclei are also charged particles (by virtue of the positively charged protons) the spinning action causes an extremely small magnetic field (magnetic moment) to be set up, in the same way as a magnetic field is generated by an electric current travelling in a wire surrounding an electro-magnet. The body's most abundant nucleus with these properties is that of hydrogen, the element which is generally investigated with MRI.

If we place some hydrogen nuclei (protons) in a magnetic field, just over half of them will tend to line up with the direction of the magnetic field, as a compass needle aligns itself with the earth's field. However, since protons are also spinning, instead of aligning themselves directly with the main field, they also *precess* around the direction of the field like a gyroscope or a spinning top. This behaviour is called *Larmor precession* (Figure 1). The frequency of this precession is proportional to the strength of the applied magnetic field, just as the gyroscope would precess more rapidly if the earth's gravity became stronger.

Suppose a water sample is placed in a magnetic field. After a short time, a little over half the hydrogen nuclei will be aligned with the magnetic field in the manner described. The excess nuclei, which are directed in the same sense as the magnetic field, have a resultant which behaves like a spinning magnet. If the axis of this magnet is tipped away from the direction of the main field, then it will begin precessing or wobbling in the same manner as the spinning top. We may induce the axis tip and resultant wobbling by irradiating the sample with energy in the form of a radio frequency (RF) wave of the same frequency as the nuclear precession. While the wobbling continues, the nuclei emit their own radio signals, which are picked up by a sensitive radio receiver, and analysed by a computer to form an image.

The degree of tip of the axis of the spinning "magnet" formed by the nuclei varies with the length of time for which we apply the radio signal. If the tip angle happens to be 90° we say that we have given the spins a 90° RF pulse.

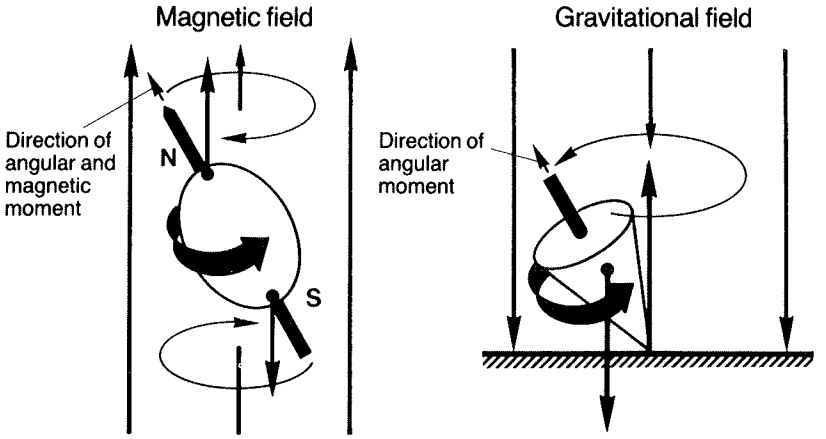

Figure 1: The charged spinning nucleus precesses in a magnetic field in much the same manner as a spinning top precesses in the gravitational field of the earth. Reproduced with the permission of Philips Medical Systems, The Netherlands. © 1984.

Under these conditions, the amount of induced wobbling and the generated signal is maximum. An RF pulse applied for twice as long as was required for a 90° tip, results in a tip of 180°. Under these conditions the spins are pointing in the opposite direction to the main magnetic field and the precession ceases. Since only precessing nuclei generate radio signals, there is no signal generated after a 180° RF pulse.

The rate with which the magnetization recovers to normal after each of these situations is important in imaging. After a 90° pulse, the signal generated by the nuclei decays away with a time constant of T2 (that is, after T2 seconds, the signal level will be down to 37 percent of its original value). T2 is known as the transverse relaxation time. After a 180° pulse, the nuclei return to their original state with a time constant T1, the longitudinal relaxation time. T1 is the time taken for 63% of the nuclei to return to equilibrium. T1 and T2 vary with different biological materials and it is the differences in these values between white/grey matter, for example, which gives us the remarkable contrast we see in typical images. For pure water both T1 and T2 are approximately 5 seconds. However, in most substances T1 is much longer than T2.

We have shown how a resonance can be set up in the nuclei of the hydrogen atoms. How then do we detect this resonance? Suppose a sample of water in a vial is surrounded by a coil consisting of several turns of wire with its axis at right angles to the static magnetic field B. A voltage is then induced in the coil by the precessing nuclei in exactly the same way that a voltage is induced by a

magnet moving inside a coil. This induced voltage alternates at the same frequency as the nuclear precession, and is called the *nuclear induction* signal.

Of course, if we were measuring the induced voltage at the same time as we were applying the rotating magnetic field, we would not be able to tell whether the signal came from the nucleus or the applied field. However, when we turn the applied radio signal off, the nuclei continue to precess. This behaviour does not last indefinitely but dies away after a time. The generated signal is called free induction decay (FID) and has a time constant T2 as previously described. While for pure water T2 is a few seconds, for most physiological substances it is much less than this.

When dealing with protons in a magnetic field of 1 Tesla (10^4 gauss) the frequency of precession is 42.6 MHz. In practice MR imaging always uses frequencies in the short-wave radio frequency range (between 6 MHz and 80 MHz). The precession frequency is directly proportional to the strength of the magnetic field.

A simple apparatus for detecting an NMR signal in a sample of water is shown in Figure 2. In Figure 2, the RF generator supplies a pulse of RF current at the resonant NMR frequency, to a small coil surrounding the vial of water, placed between the poles of the magnet. After the pulse ceases, the FID signal is picked up by the same coil and displayed on an oscilloscope.

Imaging with MR

The experiment described above measures the number of nuclei in the sample (proportional to the height of the signal), and the relaxation time T2 (the time constant of the signal decay). However, it does not measure the

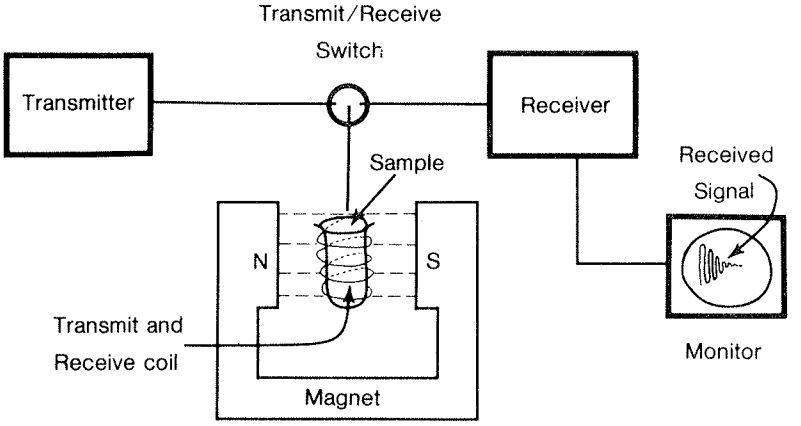

Figure 2: Simplified block-diagram showing the components of a basic system necessary to perform a simple NMR experiment.

distribution of the nuclei within the sample, which of course we must be able to do if we are at all interested in imaging. To achieve spatial localization within the sample we introduce a field gradient, which is a spatially varying magnetic field that we can superimpose on the main field. These magnetic fields are created by a set of "gradient coils" which are located within the bore of the magnet, just outside the patient "tube." The current for these coils is supplied by high power amplifiers under the control of the system computer. There are separate sets of coils for each of the X, Y and Z directions. The effect of activating a Z gradient, for example, is to decrease the magnetic field at one end of the magnet by a fraction of a percent, and to increase it by the same factor at the opposite end. The net result is to cause the protons in the imaging volume to precess with different Larmor frequencies depending on their positions.

The application of a gradient field allows us then to determine the distribution of the nuclei along the Z direction as we see in Figure 3. In the absence of the gradient, all of the nuclei precess at the same frequency. However, with the gradient switched on, those in the higher field precess with a higher frequency, those in the centre with a medium frequency, and those in the lower field with a lower frequency. The radio signal detected after an excitation pulse is rather complex since it contains a combination of each of these frequencies. However, a Fourier Transform sorts out the relative amounts of each of the frequencies present in the signal and allows us to express the result as a plot of intensity versus frequency. In the presence of a gradient, frequency is proportional to magnetic field which is itself proportional to distance, so by calculating the Fourier Transform of the signal, we may produce a map of proton density versus distance.

At this point we have been able to determine the number of nuclei in each plane along the length of our sample. However, we must now try to map the distribution of the nuclei within a plane. To achieve this we must first excite a single plane within our object. We can do this by first switching on a gradient (in the Z direction, for example) and irradiating the sample with an RF pulse which contains only frequencies appropriate to a slice of a particular position and thickness (recall that the nuclei in the presence of the Z gradient will precess with different frequencies depending on their position). This is analogous to placing a tuning fork on the sounding board of a piano. Only the strings having the same resonant frequency as the tuning fork will vibrate. In this way we may excite an individual slice within the volume.

Once such a slice has been selected, we may change the direction in which the gradient is applied so that the protons which were originally precessing with a common frequency, now precess with increasing fequencies as we cross the diameter of the slice. However, each nucleus on any line perpendicular to this diameter will precess with the same frequency and thus emit the same frequency radio signal. Nevertheless, the total signal emitted by the slice contains a range of frequencies spanning the width of the slice. Following the

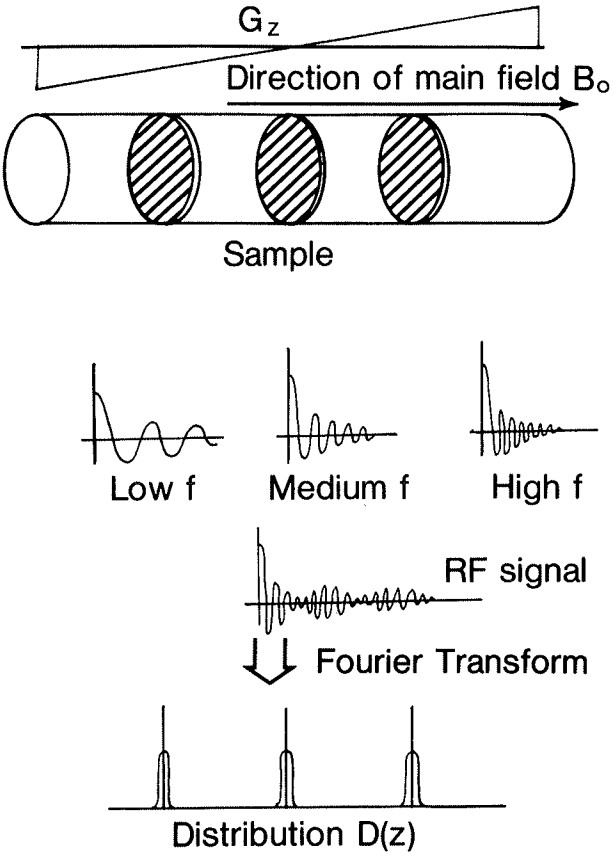

Figure 3: Calculation of spatial distribution of nuclei using an applied magnetic field gradient. Reproduced with the permission of the Society of Nuclear Medicine. © 1984.

illustration below (Figure 4), we see that the Fourier Transform of the signal this time gives us not just 3 points, but a map of the distribution of the proton density taken in the direction of the applied gradient. We have, in effect, a "projection" of the hydrogen density, within the slice.

This is the method originally used by Lauterbur (1973) and is analogous to the x-ray projections which are made with a CT scanner. The images may be reconstructed using a mathematical algorithm to combine projections from a large number of different directions. There are other methods of reconstructing the MR Image, the most popular being the "Two-dimensional Fourier" or "Spin-Warp" method (Edelstein, Hutchinson, Johnson, and Redpath, 1980; Kumar, Welti, and Ernst, 1975) now employed by most commercial scanners.

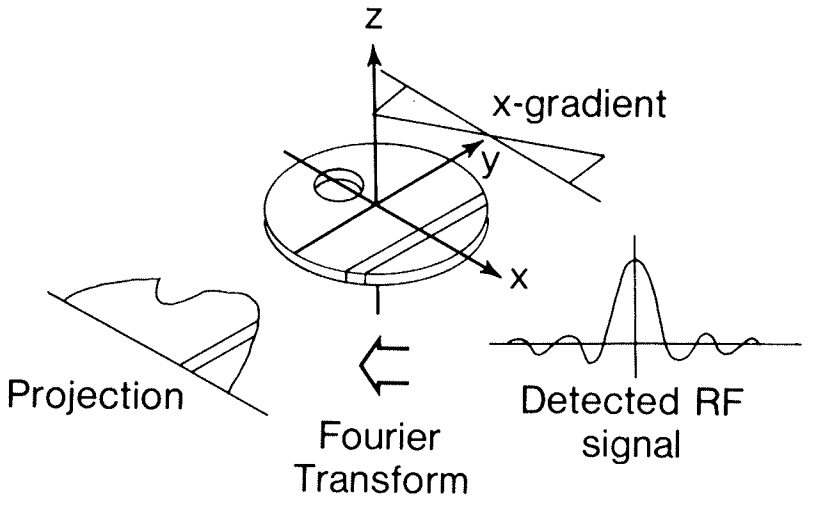

Figure 4: Acquisition of "projections" of proton density in a slice using magnetic field gradients. Reproduced with the permission of the Society of Nuclear Medicine. © 1984.

The mechanics of this technique are similar to the method described above (Peters and Sanctuary, 1984) but the details are more complex and beyond the scope of this discussion.

Contrast Observed in MR images

The most commonly used method of imaging is the so-called "spin-echo" technique which can exhibit both T1 and T2 contrast variations by changing imaging parameters. We discussed earlier the decay of the signal characterized by the parameter T2. The spin-echo technique enables us to take a "snapshot" (echo) of the signals at various times after the initial excitation pulse is delivered. These snapshots are commonly made at multiples of 30 or 50 ms. Each snapshot gives us a picture whose brightness at any point is related to the signal emitted by nuclei at that position at the time of the measurement. Thus, a substance with a long T2 value (cerebral spinal fluid for instance) will remain bright in several echos. On the other hand, white matter may be bright on the first, but would show up darker on subsequent echo images. The computer can analyse the exact nature of this inter-echo decay and assign T2 values to each element of tissue. More generally, the parameters of the spin-echo technique are varied to best demonstrate a particular pathology in the image.

However, we often experience a situation where structures which are initially (e.g., the first echo) darker than their surroundings, are brighter on a later echo. To understand this phenomenon, we must look more closely at the nature of a repeated pulse sequence.

Recall that after excitation, the longitudinal component of the magnetization recovers with a time constant T1 while the transverse component decays with a time constant T2. At the very beginning of an experiment, all of the nuclei are in equilibrium and aligned with the field. After the first 90° pulse, we expect a maximum signal which is related to the proton density only. However, the second and subsequent pulses will result in a signal whose strength is determined by the extent of the longitudinal relaxation between 90° pulses. If a long pulse repetition time (TR) is used (greater than several times the maximum T1 in the sample) full recovery of all nuclei will occur. If this is not the case, then long T1 substances will not recover fully and the signal level after the next 90° pulse will be low. Thus, substances with a long T1 and long T2 can appear darker than their surroundings on an early echo, but brighter on a subsequent echo. Examples of this are found when imaging cerebrospinal fluid in the brain or aqueous humor in the eye.

Of course, if the proton densities of two substances are different, then this will also affect the measured signal strength. Then a tissue with a low proton density but long T2 could exhibit the same behaviour as a tissue with high proton density but short T2. This is not a significant factor in the brain since most tissues have roughly the same proton density.

Because the degree of relaxation of a nucleus depends upon its T1, a short TR (TR less than 700 ms) sequence is often used to generate a "T1 weighted" image (an image where the brightness is dominated by T1). In these images, nuclei with a short T1 will have almost fully recovered by the time the next 90° RF pulse comes along and will appear bright in the image, whereas a substance with a longer T1 will not have recovered to the same extent and thus the available signal following another 90° pulse will be lower.

We note in passing that a T1 weighted image displays long T1 regions darker than their surroundings, while T2 weighted images show long T2 areas as brighter. Since T2 generally increases with T1 (although there are exceptions), the same slice imaged with a T1 weighting can exhibit almost the inverse contrast relationship shown in a T2 weighted image (see Figure 5).

While various sequences are referred to as being T1 or T2 weighted, we may calculate true T1 or T2 images by extracting this information from multiple images of the same slice made with different pulse sequences. Typical T1 and T2 values of biological tissues at 20 MHz (magnetic field of approximately 0.5 Tesla) are given below in Table 1.

An alternative T1 weighted method is the so-called "Inversion Recovery" sequence, where the spins are inverted with a 180° pulse prior to being read with a 90° pulse. In this sequence, the degree of T1 relaxation between the two RF pulses affects the strength of the signal in the image.

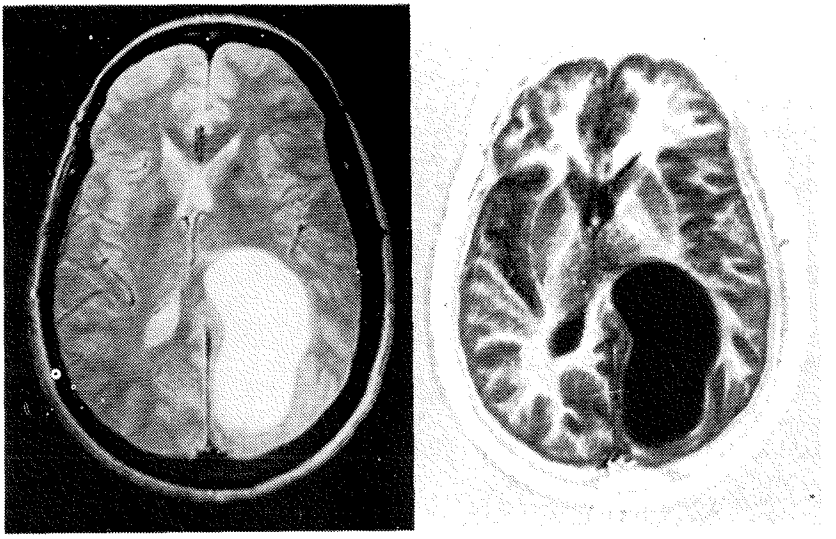

Figure 5: Dramatic difference in contrast observed with T1 weighted (left) and T2 weighted (right) images of the same tumour.

Equipment

An MR unit for whole body imaging consists typically of a superconducting magnet with a bore approximately 1 meter in diameter and operating at a field strength of 0.5–1.5 Tesla. A superconducting magnet carries its electrical current flowing through conductors which are held at –269° C, at which temperature the wires exhibit zero resistance. Once the current flow is started, the source is removed and the current is maintained without outside assistance. The windings, which are well insulated from outside heat sources, must nevertheless be kept at this supercold temperature, and this is achieved by immersing the coils in a bath of liquid helium, which boils at –269° C. The helium boils off at the rate of approximately 0.2–0.5 litres per hour, and must be topped up every 4–8 weeks. Helium consumption may be minimized by using an external refrigerator (cyrogenerator), the use of which adds further cooling to the inside of the magnet, as well as an outer cylinder filled with liquid nitrogen at a temperature of –196° C. The newly discovered high temperature superconductors may change this situation in the future and result in cheaper magnets.

Table 1

Typical NMR Relaxation Times
of Human Tissue at 20MHz*

TISSUE	T1 (msec)	T2 (msec)
Bladder	850	110
Blood	500	200
Brain		
White matter	690	107
Grey matter	825	110
Heart		
Pericardium	310	110
Carcinoma	1200	200
Kidney		
Cortex	700	100
Medulla	1000	130
Papilla	800	140
Liver		
Normal	300	60
Hepatoma	400	70
Lung		
Normal	630	70
Carcinoma	800	100
Spine		
Annulus	400	25
Nucleus	800	100
Herniated disc	1000	90
Neurofibroma	1200	170

*NOTE: Relaxation times normally vary with frequency. The data are accurate to approximately ± 10%

(Data courtesy of Frank Q.H. Ngo, PhD [Personal Communication], Department of Molecular and Celluar Biology, Cleveland Clinic Foundation, Cleveland, Ohio.)

The radiofrequency energy is transmitted to, and received from, the patient via RF antennas (in the form of coils) within the magnet bore. The most efficient signal is obtained by having the coils close to the patient, so separate systems are used when imaging the head or body. While the head and body coils are used both for excitation of the nuclei and reception of the signal, in some instances it is useful to employ a specially designed "surface coil" which is used for reception of the signal only (transmission of the RF pulse being via the body

coil). The use of surface coils results in more efficient collection of signals from specific organs (and hence less noisy images). Their application for spine imaging is further discussed later in this paper.

Shielding

Since the magnet behaves like any bar-magnet, its lines of force extend well beyond the bore of the magnet itself. The strength of the magnetic field falls off as $1/(\text{distance})^3$. Because of the interaction of the magnetic field with sensitive electronic devices such as gamma cameras, CT scanners, TV monitors and heart pacemakers, it is often necessary to provide some kind of magnetic shielding on the site. This is conveniently achieved by using a "cage" made from longitudinally placed steel bars which form a "tunnel" in which the magnet rests. These effectively provide a return path for the lines of magnetic flux, and reduce the field outside the magnet by 5–10 times in the transverse direction and 2–3 times in the longitudinal direction. This shielding is more than sufficient for most installations.

The MR system picks up very weak radio signals from inside the body. We must therefore ensure that signals from outside are not allowed into the magnet. This may be achieved either by placing the entire imaging system within a radio-frequency screened room—a very expensive solution, or by building this shield into the magnet, effectively shielding only the patient from external radio frequency noise. This latter solution is quite effective in most instances, and is much cheaper than installing a fully RF screened room at a cost of up to $200,000. However, before making a decision one way or the other the prospective purchaser should have an RF field site survey performed by the manufacturers of the equipment. A typical installation showing a high field (1.5 Tesla) imaging system surrounded by a magnetic shield is shown in Figure 6. Part of the internal RF shield, resembling an extension to the patient tunnel, can be seen protruding from the bore of the magnet.

Resolution and Imaging Times

Resolution of MR images varies from approximately 0.5 mm for high resolution surface coil studies, to 2 mm for standard body scans. Scanning times may vary from 30 secs for a low resolution and relatively noisy image to 10–15 minutes for typical studies. When using relatively long pulse repetition times, (greater than 600 ms) it is possible to record the data from many slices simultaneously. Therefore, in a typical scanning session 16 slices may be obtained in 15 minutes or so of scanning time. The imaging period is not the only time taken up by the procedure. After the patient is positioned, the system enters an automatic set-up mode whereby the system is "tuned" to the specific characteristics of the individual patient. Recent advances in pulse-

programming techniques have resulted in fast imaging methods which allow the scanning of a single slice in 2–4 seconds (de Roos, Bluem, Bloem, Glover, and Kressel, 1986).

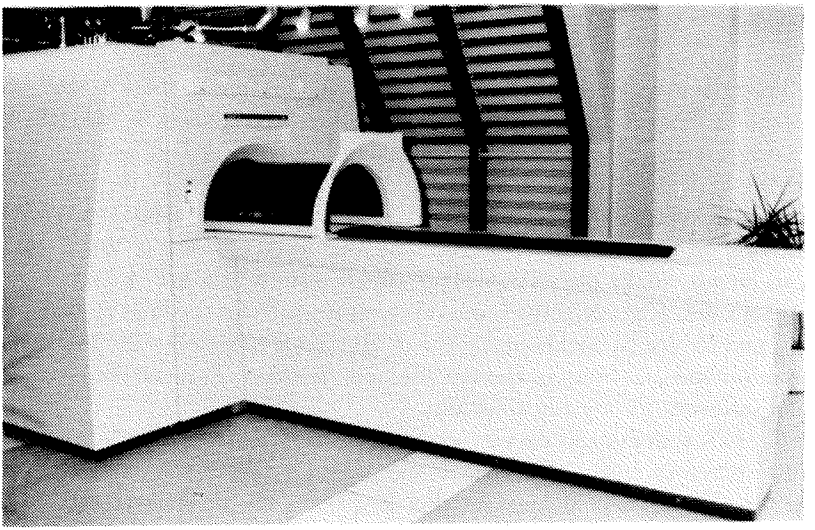

Figure 6: Typical clinical MRI installation.

Safety

Since MR does not use ionizing radiation, it is considered to be safer than medical diagnostic techniques using X- or gamma rays or heavy ions. Nevertheless, the effect on the human body of strong static or varying magnetics field, and high intensity RF energy, must be considered.

The magnetic field may interact with cardiac pacemakers, and partly for this reason most sites are designed such that the stray field outside the imaging area is less than 5–10 gauss (1 Tesla = 10,000 gauss), or 10–20 times the ambient magnetic field of the earth. Generally speaking, small prostheses do not present a problem except for disturbance effects (artifacts) that may result in the image. Patients with intracranial aneurysm clips should be excluded however, in case the material used for the clip is ferromagnetic causing it to move during the imaging procedure.

The greatest danger of the high magnetic field associated with an imaging system is the strong attraction of ferromagnetic objects towards the magnet. Precautions must be taken to prevent such objects from entering the room, since accidents causing considerable damage to the magnet have been reported. Even small items such as scissors or coins may become dangerous

projectiles if inadvertently introduced into the magnet. Ferromagnetic material in artificial limbs is a definite hazard, and such prostheses must be removed from the patient prior to scanning. Credit-cards and analog watches are quick to fall victim to the magnetic field, which will erase magnetically encoded data on the cards and permanently magnetize the delicate movements of watches. The effect of stray magnetic fields on hospital equipment must also be considered. CRTs and nuclear medicine instrumentation are particularly affected by even low fields (1–2 gauss) so great care must be exercised in planning a facility in relation to other equipment in the hospital.

Clinical Studies

As a diagnostic technique in the brain, MRI has shown itself to perform the imaging function better than most other modalities. One of the most dramatic demonstrations of MR imaging in the brain is in the assessment of Multiple Sclerosis (MS) (see Figure 7). MRI has consistently demonstrated its superiority in this area compared with CT (Runge, Price, Kirshner, Allen, Partain, and James, 1984; Sheldon, Siddharthan, Tobias, Sheremata, Soila, and Viamonte, 1985).

For other areas of the body, the merits of MRI are less well established. The forte of MRI is its ability to obtain excellent soft-tissue contrast resolution, and

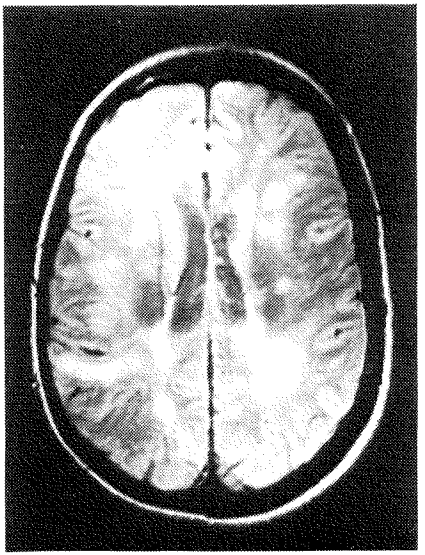

Figure 7: Typical MRI scan of patient afflicted with Multiple Sclerosis. Plaques are demonstrated by the bright areas distributed across the area of the scan.

for structures which do not move during the scanning sequence, spatial resolution may approach that available with high resolution CT. These factors make MRI particularly useful in imaging the central nervous system, as well as bones and joints. (In the latter case it is not the bone which is being imaged directly—since there is little unbound water in bone—but the water/fat component of the bone marrow which does exhibit a high NMR signal).

One of the problems with MR imaging is that in spite of its excellent level of sensitivity, there is often a problem with specificity. Thus, certain tumours are often indistinguishable from surrounding edema unless contrast agents are employed. These agents, (for example, Gd-DTPA) do not in themselves emit a stronger NMR signal, but being paramagnetic, tend to decrease the T1 relaxation time of the hydrogen nuclei in the immediate neighborhood. Gd-DTPA selectively crosses the blood-brain barrier in such a way that the tumourous mass can be highlighted and separated from the surrounding edema which remains unaffected. One of the distinct advantages of MRI over CT is its ability to image sections, not only in transverse planes, but in sagittal and coronal planes as well (see Figure 8).

In the following sections, all MR images and spectra displayed were made on a Philips Gyroscan S-15 system at the Montreal Neurological Hospital and Institute. This unit operates at a magnetic field strength of 1.5 Tesla and is equipped to perform integrated imaging and spectroscopy.

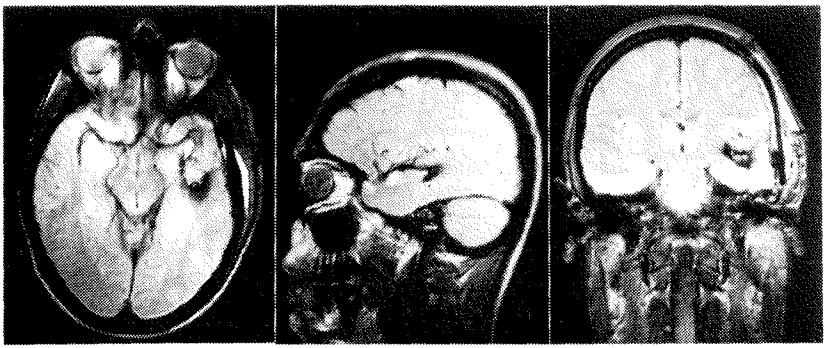

Figure 8: Brain tumour (venous angioma) imaged with transverse (left), sagittal (centre) and coronal (right) views.

Planning for Stereotactic Surgery

Stereotactic Neurosurgery is a technique used for approaching lesions with a probe of some kind from a small hole in the skull, obviating the need for a craniotomy. The location of the lesion is accurately determined by imaging the

brain while a rigid frame is fixed to the patient's head. The frame has calibrated markings which establish a three-dimensional coordinate system. Attached to the frame is a series of plastic plates which contain reference markers that may be easily identified in the image. The positions of such markers in the images allow the exact scale and orientation of any imaged slice to be obtained. Computer software can be used to analyse these images and allow the coordinates of lesions to be determined to an accuracy of ± 1 mm (Brown 1979; Lundsford, Latdraw, and Vrier, 1983).

Until recently, CT had been the major imaging technique used in stereotactic planning. However, we have recently shown that it is possible to use MRI in this manner also (Peters et al., 1986). Because of the ability of MRI to show accurate differentiation between white matter, grey matter, and CSF, it is much better than CT for depicting anatomical structures within the brain. A second advantage is that MR imaging is not confined to transverse sections but may equally well display cuts in sagittal or coronal planes.

The stereotactic frame used at the Montreal Neurological Institute is shown in Figure 9, fixed to a patient just prior to an MRI scan. Fastened to the sides can be seen the plate containing the reference markers ('z'bars) which show up in images as seen in Figure 10. In this example, the coordinates of a target site within a cyst are being determined prior to aspiration. The scale on the image is derived from the positions of the markers seen surrounding the image. After

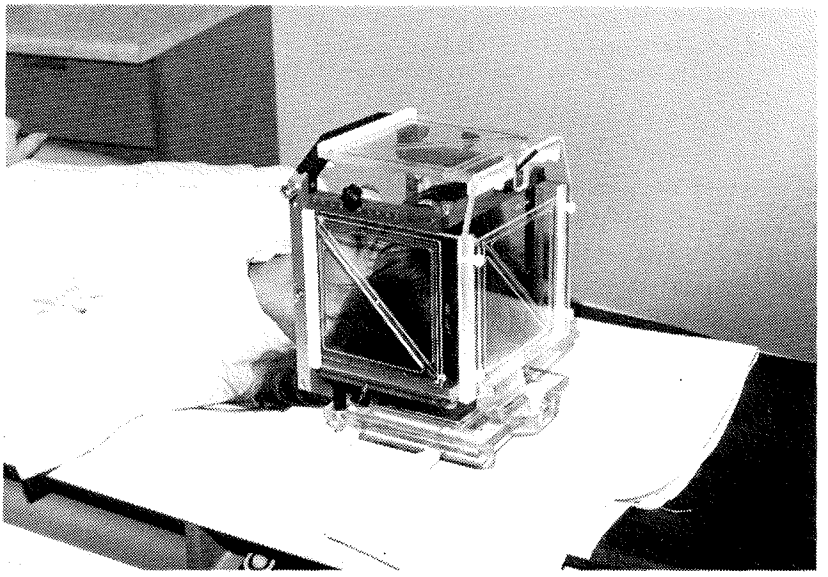

Figure 9: Sterotactic frame attached to patient's head prior to MRI scan.

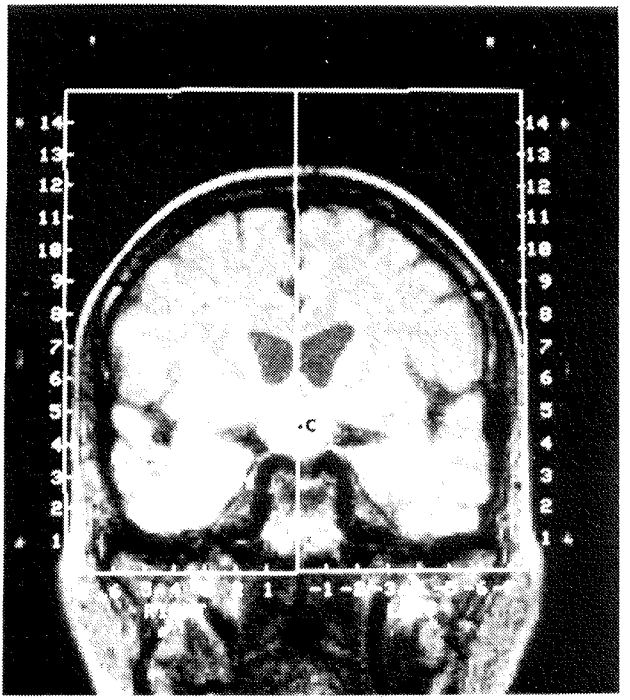

Figure 10: Stereotactic anaylsis of lesion. The scale markings are calculated by computer software based on the positions of markers associated with the frame.

the scan and image analysis have been performed the patient is taken directly to the operating room where special instruments attached to the frame approach the target at the defined coordinates via a small drill-hole in the skull. This technique is being used for biopsy of tumours, aspiration of cysts, implantation of radioactive seeds, planning of radiation therapy and planning of depth electrode implantations.

Integration with Other Imaging Modalities

MRI gives superb visualization of anatomy and anatomical abnormalities. However, a total view of the brain includes knowledge of the metabolic activity and also the vascular structures within the brain. Because other techniques are able to do better than MRI in these areas (e.g., Positron Emission Tomography [PET], and Digital Subtraction Angiography [DSA] respectively), we must be able to easily relate images from one modality with those of another.

For example, consider the images seen in Figure 11 which represent the sagittal MR image and the DSA image of the same patient. In their original state they have different scales and orientations, and are also represented in the same colour schemes.

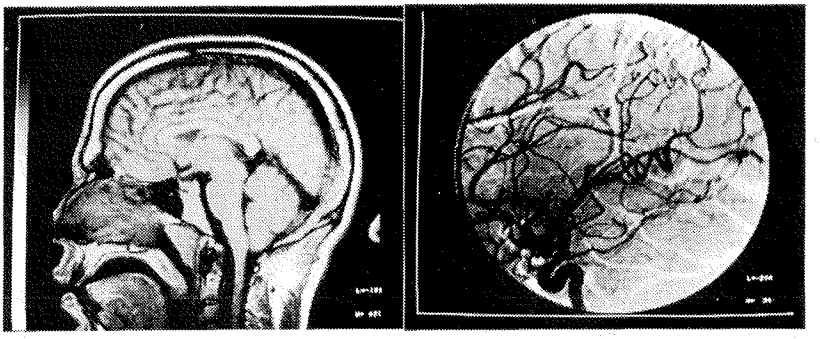

Figure 11: Sagittal MR image (left) and arterial (black) and venous (white) subtracted angiogram of same patient (right).

To ease the task of comparing and integrating these images we may use the computer to scale each image appropriately, and to display each in a unique colour set so that the components of one image are not confused with those of the other when the images are superimposed. In this case, in the original image we separated the arterial and venous phases of the angiogram and coloured the arterial phase red and the venous phase blue. These images are then combined on a computer display along with an appropriately scaled and rotated version of the MR picture (Figure 12.) [A black and white photograph is presented here instead of the original colour photo] The rotation and scaling of one image with respect to the other may be performed by matching external coordinates (as in Stereotaxy) or internal as in the example shown where the extremities of the corpus callosum were matched with the pericollosal artery and the vein of Galen. These vessels lie in the midline of the brain corresponding to the MR slice and can be seen to closely match the contour of the corpus callosum. Other vessels are not in the same plane, of course, but would show correlation with other structures in sagittal slices made more laterally.

Spine Imaging with Surface Coils

In spite of the generally high quality of MR images they are nevertheless degraded somewhat by the relative size of the receiving antenna compared to the size of the structure being imaged. For head studies, we therefore use a

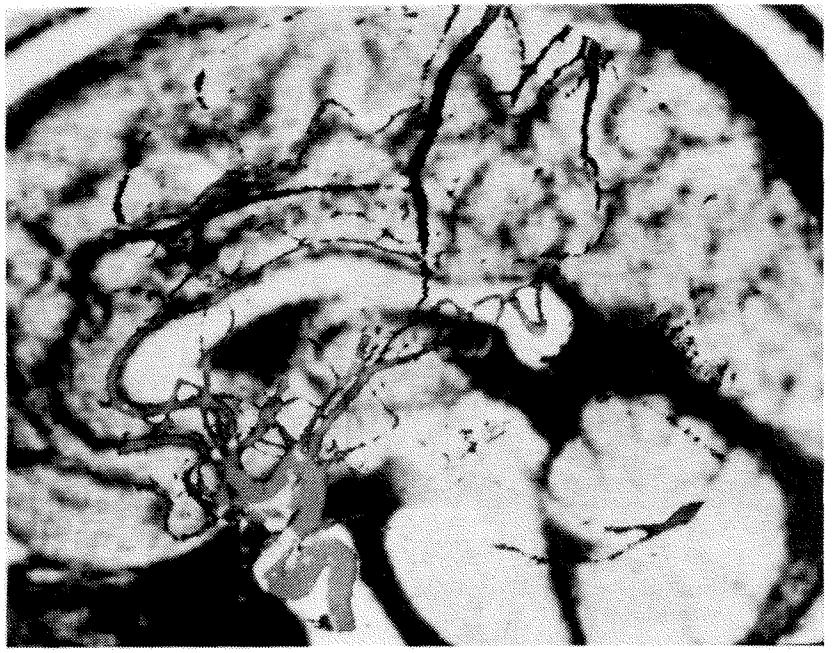

Figure 12: Combined image of MRI and DSA information.

transmit/receive antenna which is tailored to the size of the head, rather than using the larger body coil. In spinal imaging, where we are only interested in detail fairly close to the surface, we may use a specially designed "surface coil" which receives most of its signal from within a depth of about 10 cm from the coil surface. When such a coil is used to receive information from the spine the received signal intensity is greatly increased, and this is reflected in the resulting increase in image quality. A typical spine image made using a surface coil is shown in Figure 13. Surface coils are useful in other parts of the body, especially when the organs being imaged are close to the surface, e.g., the orbit, breast, knee and testes.

Spectroscopy

Well before imaging became a practical technique, MR spectroscopy was being used in chemical and biological research. Just as the frequency of precession may be changed by placing a sample in a field gradient, the local chemical environment in which a nucleus finds itself will also affect the frequency of precession. This is because chemical bonds between atoms of a molecule act somewhat like an electrical shield on the charged nucleus, which

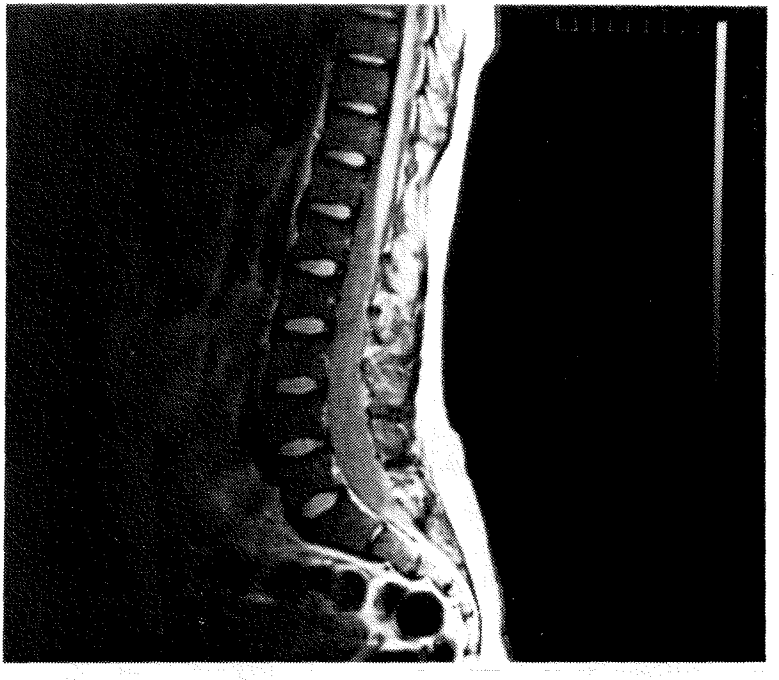

Figure 13: Spine image made with surface coil. Note the increase in signal in the sensitive region of the coil close to the surface of the patient.

"sees" a slightly modified field. For example, the precession frequency of a hydrogen nucleus is different, by several parts per million (ppm), from a hydrogen nucleus which is part of a fatty molecule. By carefully observing these small differences in resonance frequency, it is possible to characterize the structure of the molecule being studied.

The most common elements studied with in vivo spectroscopy techniques at this time are hydrogen and phosphorus, with the latter being considered the most important, since phosphorus plays an important part in many metabolic processes, and manifests itself in a variety of forms [e.g., inorganic phosphate, (pi), adenosine tri-phosphate (ATP), and phosphocreatine (PCr)]. The strengths of the signals produced by the phosphorus nuclei in each of these (and other) compounds can give an indication of diseased or abnormal tissue relative to normal tissue.

To perform spectroscopy adequately, it is necessary to use magnets of high field strength and stability, and to undertake satisfactory proton imaging as well as phosphorus spectroscopy on humans, we must use a magnet with a field strength of at least 1.5 Tesla, and a field uniformity of better than 0.1 ppm over

at least a 20 cm diameter spherical volume. Recent advances in technology (see Segebarth, Balériaux, Arnold, Luyten, and den Hollander, 1987) have resulted in integrated systems which allow spectroscopic examinations to be planned by first making a standard series of proton images, and then interactively defining a volume of 30–50 ml in the region of interest. This is immediately followed by a spectroscopy experiment where the phosphorus signals are received only from the volume selected.

A typical volume selected on a proton image of the head is shown in Figure 14. Figure 15 shows a normal brain spectrum and a spectrum from a similar region of the brain containing a tumour. The changes in the peaks of the phosphorous spectrum between the two scans are clearly evident.

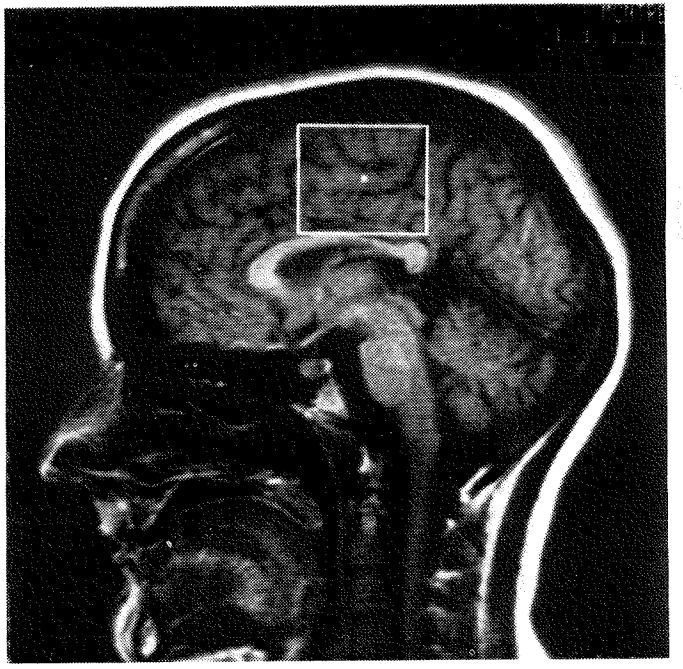

Figure 14: Region of interest for 31-P spectroscopy selected on proton image.

Conclusion

Magnetic Resonance Imaging is now a well-established diagnostic tool in medicine. It is gaining widespread use worldwide and is fast becoming the diagnostic imaging procedure of choice, particularly in neurology. MRI offers many advantges over other imaging techniques. It provides images with

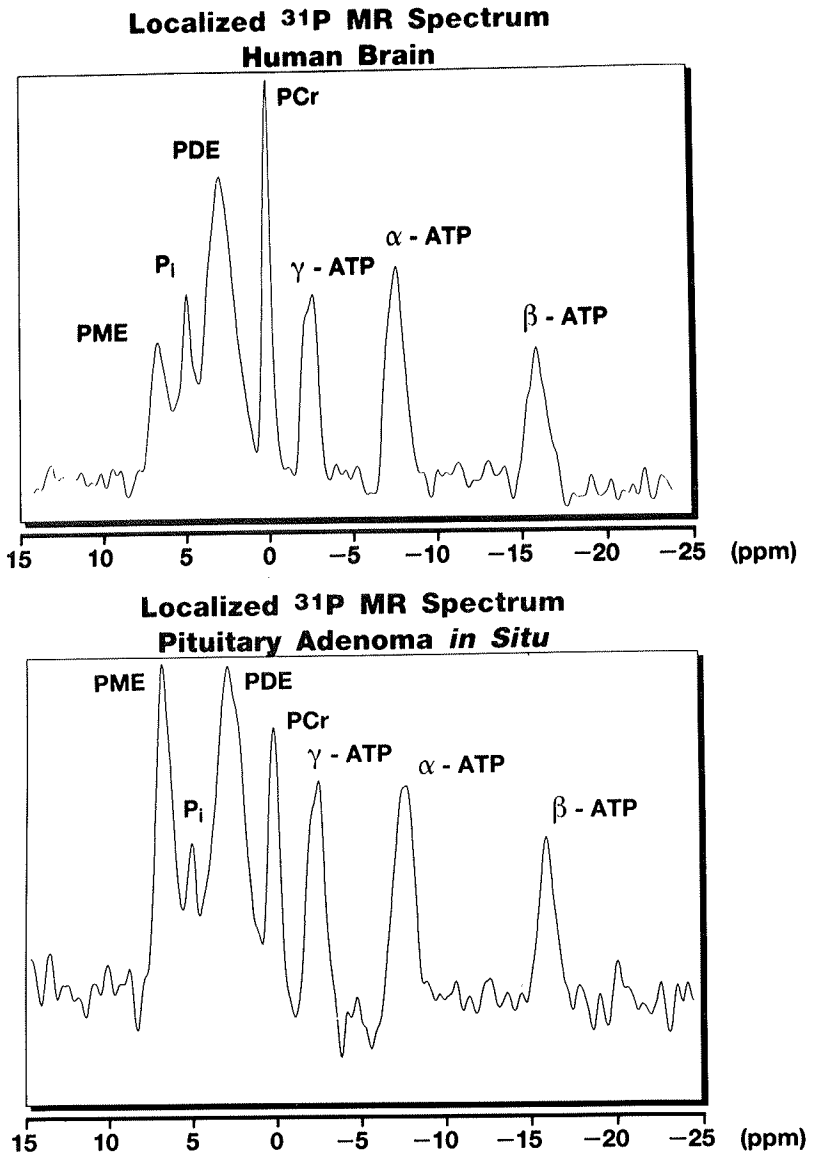

Figure 15: Normal phosphorus spectrum of brain (top).
Phosphorus spectrum of brain tumor (bottom).

The peaks on the spectra (in Figure 15) are as follows:
PME—Phosphomonoesters
PDE—Phosphodiesters
Pi—Inorganic phosphate
PCr—Phosphocreatine
ATP—Adenosine triphosphate (3 peaks)

excellent soft tissue contrast resolution, and does so with a spatial resolution comparable to that of X-ray CT. MRI does not use ionizing radiation and, therefore, does not subject the patient to risks associated with its use. Unlike CT, MRI offers the ability to image in any plane, namely transverse, sagittal, coronal or oblique without moving the patient. In addition to the role of MR in imaging, it is now also proving to be extremely valuable for in-vivo spectroscopic studies, especially when integrated with imaging techniques which enable the spectroscopic volume of interest to be interactively defined within the brain. Future advances in MRI can be expected to produce shorter imaging times, angiographic flow imaging and higher resolution images. The cost of systems can be expected to drop somewhat in the coming years, as electronic sub-systems become more integrated, and as recent developments in the field of higher temperature superconductors result in magnets which are less expensive to both build and maintain.

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