

Magnetic resonance imaging in medicine

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Abstract

Over the past twenty years, magnetic resonance imaging (MRI) has become one of the most important imaging modalities available to clinical medicine. It offers great technical flexibility, and is free of the hazards associated with ionizing radiation. In addition to its role as a routine imaging technique with a growing range of clinical applications, the pace of development in MRI methodology remains high, and new ideas with significant potential emerge on a regular basis. MRI is a prime example of the spin-off benefits of basic science, and is an area of medicine in which physical science continues to play a major role, both in supporting clinical applications and in developing new techniques. This article presents a brief history of MRI and an overview of the underlying physics, followed by a short survey of current and emerging clinical applications.

Historical introduction

In the 1940s, in a laboratory somewhere in the depths of Stanford University, a physics researcher did something that would now be regarded as highly irresponsible and ill-advised. Felix Bloch put his finger into the apparatus that he had built to study a newly discovered phenomenon known as nuclear magnetic resonance (NMR). Because of its high water content, Bloch's finger produced a strong NMR signal. This observation is now regarded as the first step on the path towards biomedical magnetic resonance imaging (MRI): a technique based on fundamental nuclear physics that has spawned a multibillion pound industry and made tremendous contributions to the health and quality of life of millions of people.

NMR was anticipated on theoretical grounds, and was searched for unsuccessfully by the Dutch physicist Cornelius Gorter in 1936. Isidor Rabi observed the phenomenon in molecular beams in

1937, but it was 1946 before it was observed in liquids and solids, resulting in the award of the Nobel Prize for Physics to Bloch and Edward Purcell of Harvard in 1952. Since then, NMR has become a routine tool for physicists and chemists to probe molecular structures. These applications, fascinating though they may be, are beyond the scope of this article. However, the samples studied soon came to include biological materials. These were not imaging studies: their aim was to characterize tissue samples according to water content and nuclear relaxation times (see below). In 1960 a report from the US National Heart Institute noted that NMR experiments might lead to 'a scanning technique [that] will give coarse pictures... of muscle, arteries and other structures.' This was the first suggestion that NMR might be the basis of a new medical imaging modality, although the idea was not pursued at the time.

In 1971, writing in the journal *Science*,

Raymond Damadian reported differences in NMR signals obtained from cancerous and normal tissues. These findings played a key role in stimulating interest in NMR as a potential diagnostic technique. Damadian developed the first whole-body NMR imaging system in 1977 (using himself as the first guinea pig) and the first commercial scanner in 1980. His work was regarded by most of his contemporaries as at best a curiosity, but he had patented his ideas and in 1997 received \$128 million from the General Electric Corporation for patent infringements. Meanwhile, Paul Lauterbur developed the idea of using magnetic field gradients (see below) to map out signals from within the body. Using this approach, he published the first NMR image in *Nature* in 1973. Many key developments over the next few years took place in the UK. In 1974, Peter Mansfield (now Sir Peter) of the University of Nottingham produced the first NMR images of the human finger, and in 1978 Ian Young at EMI produced the first head images. In 1980 John Mallard's group at the University of Aberdeen, building on work by Richard Ernst in Zurich, developed the spin-warp imaging technique that forms the basis of almost all modern MRI.

By the early 1980s, major medical imaging equipment manufacturers were beginning to recognize the potential of NMR, and the first prototype systems were installed in hospitals. Because the word 'nuclear' had unfortunate connotations for the public, the technique was marketed as 'magnetic resonance imaging' to avoid misconceptions. By 1988, 1300 MRI systems had been sold; today, there are perhaps 17 000 systems in hospitals worldwide, with around 2500 more scanners (worth around £1.5 billion) being installed each year. The capabilities and applications of these systems have grown to an astonishing extent, and there is no sign of decline in the rate of innovation.

The physics of NMR

A rigorous description of NMR requires the use of sophisticated quantum mechanics. Fortunately, for our purposes, a classical or semiclassical approximation is adequate, although it is important to keep in mind the fact that the analogies used here are not exact!

Nuclear spin is the key concept in NMR. As its name implies, spin is related to the angular

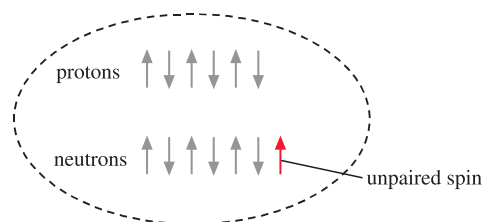


Figure 1. Origin of nonzero spin in the carbon-13 nucleus. The six protons provide no net spin, but the unpaired neutron endows the nucleus as a whole with a spin quantum number $I = \frac{1}{2}$.

momentum of the atomic nucleus, which in turn originates from that of the constituent nucleons. Protons and neutrons each have spin quantum number $I = \frac{1}{2}$. When a nucleus contains an even number of protons and neutrons, the individual spins of these particles pair off and cancel out and the nucleus is left with zero spin. However, in a nucleus containing an odd number of protons or neutrons, pairing is incomplete and the nucleus has a net spin of $\frac{1}{2}$ (see figure 1). All such nuclei undergo NMR, but in medical MRI the hydrogen nucleus, consisting of a single proton, is used because of its high NMR sensitivity and natural abundance, and because of the very high concentrations of water found in the body.

In a simple model, a nucleus with spin is pictured as a charged sphere rotating on its axis. According to classical electromagnetism, a spinning sphere of charge generates a magnetic dipole moment, μ . The fundamental requirement for NMR to occur is application of a magnetic field to the tiny nuclear magnetic moments. This is generated using a powerful magnet: for clinical MRI the field strength is usually 4000–60 000 times that of the Earth. Quantum mechanics stipulates that the magnetic moment of a nucleus with $I = \frac{1}{2}$ will adopt one of only two possible orientations relative to the direction of the field—conventionally defined as the z -axis. These orientations are shown in figure 2, which also defines the coordinate axes to be used.

Each magnetic moment also experiences a torque due to the field, resulting in precession about the z -axis so that the moment describes a cone in the xy plane as illustrated in figure 2. The angular frequency of precession, known as the Larmor frequency, is given by

$$\omega_0 = \gamma B_0$$

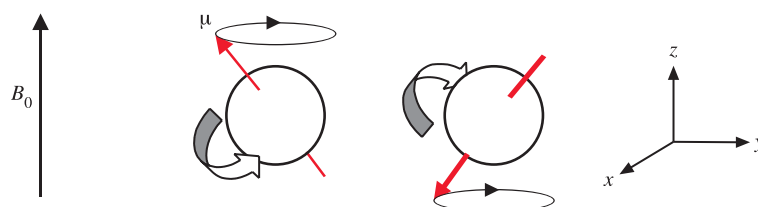


Figure 2. Possible orientations of a nucleus with $I = \frac{1}{2}$ in a static magnetic field. The size of the magnetic moment is well defined, and only two values of the z -component are possible—corresponding to alignment with or against the applied magnetic field. Nuclei precess about the z -axis, and the x and y components of the moment of a given nucleus at a given time are random.

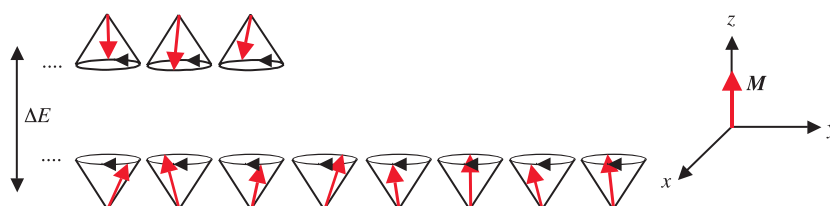


Figure 3. Origin of the bulk magnetization vector in a macroscopic sample. Excess magnetic moments aligned in the positive z -direction sum to give M , while the random x and y components cancel. The excess of a few nuclei per million has been greatly exaggerated in this diagram.

where B_0 is the flux density of the applied magnetic field and the gyromagnetic ratio, γ , is characteristic of the nucleus (for hydrogen, $\gamma = 42.57 \text{ MHz T}^{-1}$, so in a magnetic field of strength 1 T hydrogen nuclei precess at a rate of 42.57 million revolutions per second). Thus the z -component of the magnetic moment is well-defined and may adopt one of only two values while, in the classical model, the x and y components vary rapidly (ω_0 lies in the range from megahertz to hundreds of megahertz) and the position (or phase) of the magnetic moment on the precessional cone at a given time is random. This can also be understood in quantum mechanical terms as a manifestation of Heisenberg's uncertainty principle.

Spins in the two orientations have different energies, the difference being $\Delta E = \hbar\omega_0$, where $\hbar = h/2\pi$ (h is Planck's constant). Because of this, the populations of the two states differ slightly, with a few more nuclei per million aligned with B_0 than against. This is sufficient to endow the sample as a whole with a small bulk magnetization, M , lying along the positive z -axis (see figure 3). There is no net magnetization in the xy plane because of the random phases of the moments.

The next step in the NMR experiment is to irradiate the sample with radiofrequency (RF)

electromagnetic waves. The effect of this irradiation can be described in either quantum mechanical or classical terms. In the quantum model, if the energy of the RF photons is equal to the difference between the two energy levels, nuclei in the lower level may gain energy by absorbing photons and those in the higher level may be induced to emit photons and so lose energy. These processes cause nuclei to flip between the two orientations, modifying the population distribution and hence the magnitude and direction of M .

However, considering the colossal number of nuclei present in a macroscopic sample, it is convenient to adopt an entirely classical model. In this approach, NMR is described in terms of interactions between the bulk magnetization and the magnetic field component of the RF wave, B_1 . If the RF frequency, and hence the frequency at which B_1 rotates in the xy plane, matches the precessional frequency of the magnetic moments, ω_0 , a resonant condition is achieved in which M tips, or nutates, from the z -axis towards the xy plane. Nutation continues until B_1 is removed, so the amplitude and duration of B_1 can be tailored to nutate M through specific angles. An important special case is that in which M is nutated entirely into the transverse plane, leaving no magnetization along the z -axis. An RF wave of sufficient duration

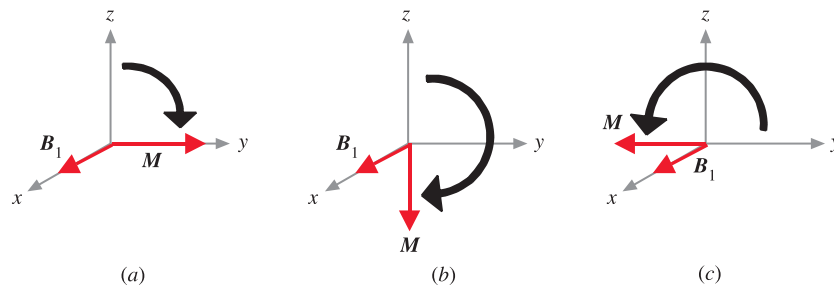


Figure 4. Special cases of an RF pulse: (a) 90° pulse, (b) 180° inversion pulse, (c) 180° refocusing pulse. Rotation of B_1 and precession of M about B_0 have been omitted for clarity.

Table 1. Relaxation times of water protons in brain tissues measured in a healthy volunteer at 1.5 T (courtesy of Dr Marcus Newbold, GKT School of Medicine, King's College London).

Tissue	T_1 (mean \pm SD) (ms)	T_2 (mean \pm SD) (ms)
Grey matter	1078.5 ± 52.6	100.4 ± 12.3
White matter	684.2 ± 20.8	79.2 ± 5.0
Cerebrospinal fluid (CSF)	3958.9 ± 305.5	914.6 ± 422.4

and amplitude to achieve this is known, for obvious reasons, as a 90° pulse. Similarly, a 180° pulse inverts M or, if applied after a 90° pulse, rotates M through 180° in the transverse plane (figure 4). These pulses are the basic ‘building blocks’ of NMR, and by stringing them together in different combinations an essentially limitless range of experiments can be performed.

Whichever model is adopted, the RF frequency required to induce NMR is equal to the nuclear Larmor frequency, $\omega_0 = \gamma B_0$.

A 90° pulse generates net transverse magnetization, which precesses about the direction of B_0 . This precessing magnetization can be detected by means of electromagnetic induction, by placing a tuned antenna close to the sample. In the context of biomedical NMR, such an antenna is usually known as an RF coil, and similar coils are used to transmit RF and to detect the emitted NMR signal.

Following excitation, nuclei return to their initial population distribution via a variety of relaxation processes, categorized according to whether they cause loss of energy from the spin system or simply exchange of energy between spins. These two phenomena are known as spin–lattice and spin–spin relaxation, and are characterized by the relaxation times T_1 and T_2 , respectively. T_1 relaxation results in exponential recovery of z -magnetization, M_z , while T_2 processes cause exponential decay of the

precessing transverse magnetization, M_{xy} :

$$M_z = M(1 - e^{-t/T_1})$$

$$M_{xy} = M e^{-t/T_2}.$$

Thus the NMR signal detected by the RF coil rapidly decays due to T_2 relaxation. Relaxation behaviour is strongly dependent on the physico-chemical environment of the nucleus, and hence on the tissue type in which the nuclei are located, and this is an important source of image contrast in biomedical MRI. Table 1 shows typical *in vivo* relaxation time values for brain tissues.

From NMR to MRI

The NMR experiment described thus far results in acquisition of an undifferentiated signal, with no means of incorporating spatial information so as to produce an image. Study of a specific tissue can only be achieved by excising the tissue from the body, or—if the tissue is superficial—by placing a small RF coil, known as a surface coil, against the body.

The method of spatial encoding used universally in MRI is due to Lauterbur. As with many of the great ideas of science, it is at once simple and extremely powerful. Temporary imposition of an additional static magnetic field, which lies parallel to B_0 but varies linearly in strength with position along the x , y or z -axis

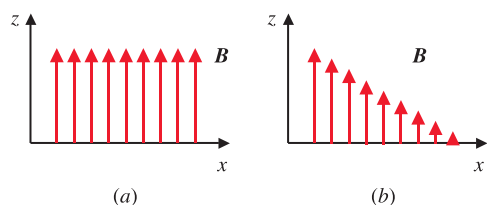


Figure 5. Effect of a magnetic field gradient along the x -axis. (a) Without the gradient, B is independent of x . (b) In the presence of the gradient, B varies linearly with x . Note that the direction of the B vector is the same in both cases; only the amplitude varies in case (b).

(or indeed some direction oblique to these axes) (figure 5) results in a linear variation in the Larmor frequency as a function of position:

$$\omega(x) = \gamma B = \gamma(B_0 + G_x x)$$

where the strength of the gradient (in mT m^{-1}) is G_x (in this case applied along the x -axis). This frequency variation can be used in three ways to achieve spatial localization.

1. *Slice selection.* If the gradient is applied at the same time as the initial RF pulse, and the pulse is modified to contain a narrow band of frequencies rather than a single frequency, a thin slice of spins can be selectively excited. Using this approach, MRI becomes a two-dimensional tomographic imaging technique (figure 6).
2. *Frequency encoding.* If the gradient is applied during signal acquisition, the resonance frequency of spins contributing to the signal becomes a linear function of their position along the gradient direction. The signal can be decomposed using Fourier methods to determine the amount of signal at each frequency and hence at each position, resulting in a projection through the sample (figure 7).

At this point, it is perhaps worth eliminating a common misconception. Many people who are learning about MRI imagine that images are formed by scanning line by line through the object, obtaining signal from each point in space sequentially. We can see from the preceding description that this is not the case. The signal acquired originates from the entire object under study, and spatial information is obtained only by breaking that signal down into its different frequency components.

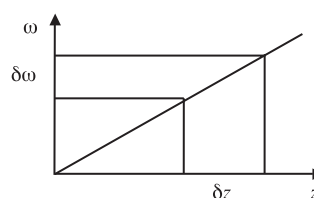


Figure 6. Use of a magnetic field gradient for selective excitation of a transaxial slice. If a gradient is applied along the z -direction together with an RF pulse of bandwidth $\delta\omega$ magnetization will be excited within a slice of thickness $\delta z = \delta\omega/\gamma G_z$. If performance is ideal, magnetization outside this slice will be left along the z -axis and will produce no signal.

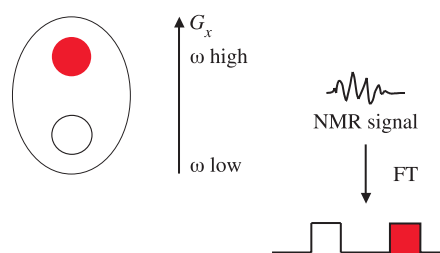


Figure 7. Frequency encoding. If a gradient is applied during signal acquisition, the resulting NMR signal will contain components with a range of frequencies. Fourier transformation of this signal yields a projection through the object. A greatly simplified object is shown in this diagram.

3. *Phase encoding.* By applying a gradient for a short time between excitation and signal acquisition, and repeating the experiment a number of times with different gradient amplitudes, it is possible to generate a set of data that may be Fourier transformed to yield spatial information along the direction perpendicular to the frequency encoding axis. Phase encoding is one of the most difficult concepts in MRI, and a full explanation is beyond the scope of this article. Suffice it to say that the process is mathematically equivalent to frequency encoding, and is really a ‘trick’ to get around the fact that frequency encoding cannot be performed along two axes simultaneously.

In conventional MRI, all three methods are applied to generate multiple two-dimensional slices through the object or patient under study. To allow time for this process, acquisition of the NMR signal is delayed for up to a few hundred milliseconds after excitation. This is achieved by

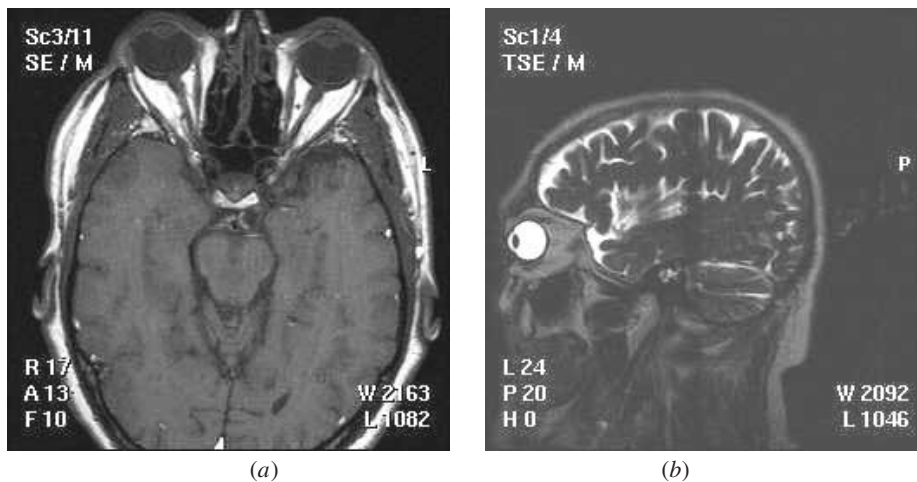


Figure 8. Flexibility of imaging plane and image weighting in MRI. (a) Transaxial T_1 -weighted image of the head at the level of the orbits. There is good soft tissue contrast, and fluid-filled structures, such as the eyes and the ventricles in the brain, appear dark. (b) Heavily T_2 -weighted parasagittal image of the head. Note the intense signal from CSF around the surface of the brain and from the contents of the eye, except for the dark lens (pictures courtesy of Guy's and St Thomas' NHS Trust).

using magnetic field gradients and RF pulses to produce what is known as an echo at the desired time. Unlike x-ray CT, MRI can image slices in any desired plane by appropriate use of gradients. Alternatively, the technique can be modified to produce three-dimensional volumes of data that can be viewed using special software or post-processed to generate arbitrary slices.

The images generated in MRI are made up of voxels (three-dimensional pixels), and the brightness of each voxel is dependent on the intensity of signal from the corresponding location in the object. A typical image consists of 256×256 or, increasingly, 512×512 voxels. This means that the in-plane resolution is typically 0.5–2.0 mm. The voxel is usually considerably larger in the third dimension, as the slice thickness is typically 1–5 mm. For specialist (usually non-medical) applications, NMR microscopes are available that can achieve a resolution of a few micrometres.

As described so far, an MR image of the body is essentially a map of water distribution. Such an image would be of limited clinical value, since water density varies relatively little between tissues. One of the major strengths of MRI is the ability to manipulate image contrast. The two simplest ways of doing this follow readily from the discussion above, and involve weighting the image according to tissue relaxation times.

During the interval between excitation and

acquisition of an echo, magnetization in different parts of the sample will be undergoing T_2 decay at different rates depending on its environment (i.e. the tissue in which it is located). Thus by varying this interval, known as the echo time, T_E , it is possible to vary the degree of T_2 weighting—i.e. the extent to which differences in T_2 affect the appearance of the image. In addition, as noted above it is necessary to repeat acquisition a number of times to allow phase encoding. During the interval between repetitions, known as the repetition time, T_R , magnetization undergoes differential T_1 recovery. By varying this interval, the user can alter the extent of T_1 weighting.

T_1 and T_2 weighting result in very different image appearances. T_1 -weighted images provide excellent soft tissue contrast, and material with long T_1 , such as cerebrospinal fluid (CSF) in the ventricles of the brain, appears dark. This type of imaging is often used to depict anatomical structures. On T_2 -weighted images, tissues with increased water content appear bright. This includes CSF, but also pathological processes such as neoplasm, inflammation, ischaemia and degenerative changes. Thus T_2 -weighted images are often used to highlight areas of disease. Typical T_1 - and T_2 -weighted images are shown in figure 8.

Many other types of image contrast may be introduced by appropriate adaptation of the

sequence of magnetic field gradients and RF pulses. For example, images may be produced that highlight flowing blood, diffusion of water molecules, blood perfusion, temperature or tissue magnetic susceptibility. All of these types of image have or are finding a role in clinical medicine.

When contrast manipulation is inadequate, artificial contrast agents may be used to highlight particular structures of interest. These are usually based on metal ions with large magnetic moments and have a dramatic effect on the relaxation properties of water protons and hence on signal characteristics. The active component of the most commonly used agents is the gadolinium ion, Gd^{2+} . These agents are usually given by intravenous injection and distribute in the body according to blood flow. Thus they are of most use for highlighting highly vascular tissues, such as tumours, or for imaging blood vessels themselves. Orally administered agents are also available, often consisting of macroscopic magnetic particles, and targeted agents have the potential to enhance signal from specific tissues or even from regions in which specific genes are being expressed.

MRI instrumentation

The appearance of a clinical MR scanner is dominated by the magnet. Until recent years these were almost invariably large and forbidding, with a bore that totally enclosed the patient and frequently disappeared into the wall of the examination room. Now, developments in magnet technology have led to designs that increase both patient acceptance and the range of potential applications. Also, whereas most MR magnets once operated at a field strength of 0.5–1.5 T, the market has now polarized between low (0.2–0.5 T) and high (1–3 T) field strength systems (with a few research systems at 4–8 T). The lower field systems are cheaper and, being more open and more patient-friendly, are arguably more suitable for paediatric and interventional work. Higher field systems produce better quality images and are appropriate for more specialized work in areas such as neurology and cardiology.

High fields are achieved using superconducting electromagnets: once the magnet is on field, current flows without loss and no power supply

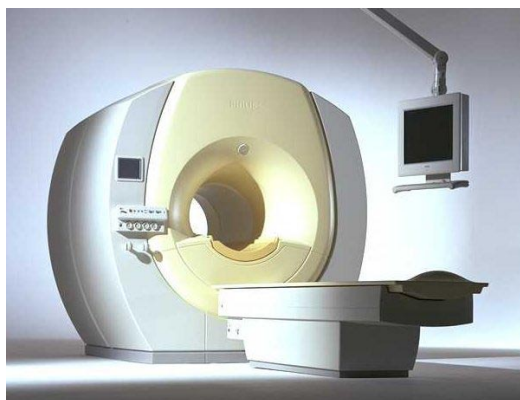


Figure 9. Superconducting MR system. This particular design is available in 0.5, 1.0, 1.5 and 3.0 T versions (picture courtesy of Philips Medical Systems).

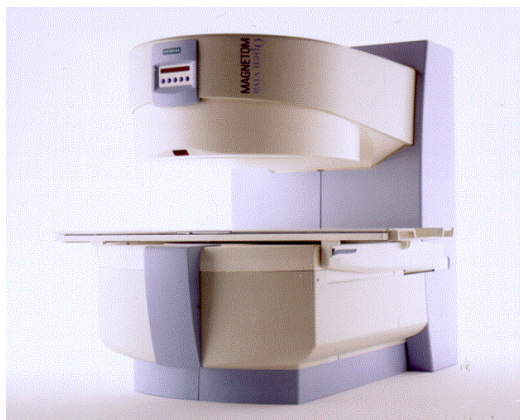


Figure 10. Low-field (0.2 T) open MR system using a permanent magnet (picture courtesy of Siemens Medical Solutions).

is needed. Newer high-temperature superconductors have had a limited impact on MRI thus far, and superconductivity is maintained by immersing the turns of the magnet in a bath of liquid helium (at 4.2 K). Modern superconducting MR systems are far more compact and patient-friendly than older designs (figure 9).

Lower field systems are usually based around resistive electromagnets or permanent magnets. These are typically more open than superconducting systems, allowing more interaction with the patient (figure 10).

As well as the magnet, the performance of an MR scanner depends on the design of the gradient and RF systems, and increasingly on the performance of the computer system used to control data acquisition and image processing.



Figure 11. Sagittal MR image of the knee. The plane has been selected to show the cruciate ligaments (the black 'X'-shaped structure behind the knee joint). Damage to these ligaments due to sporting injuries is a major indication for MRI of the knee (picture courtesy of Guy's and St Thomas' Hospital NHS Trust).

Innovations in all these areas are increasing the speed of MRI data acquisition and opening up new possibilities in real-time imaging.

Clinical applications

Because of its excellent depiction of soft tissues, MRI has long been the modality of choice for examination of the central nervous system (brain and spine), with indications including suspected tumours, slipped disks, multiple sclerosis and degenerative diseases (see figure 8). It now also has a major role in orthopaedics and musculoskeletal imaging (figure 11).

Cardiac MRI is an area of increasing importance, including investigation of both congenital abnormalities and acquired conditions such as coronary heart disease. The high resolution and contrast between tissues possible using MRI allow image processing techniques to be applied to extract features of interest (figure 12).

Improvements in the speed of MRI in recent years have made abdominal imaging possible. Figure 13 shows an MR angiogram of the kidneys, acquired using an injection of contrast agent to highlight flowing blood. Images such as this can be used to investigate narrowing of the arteries due to fatty plaques.

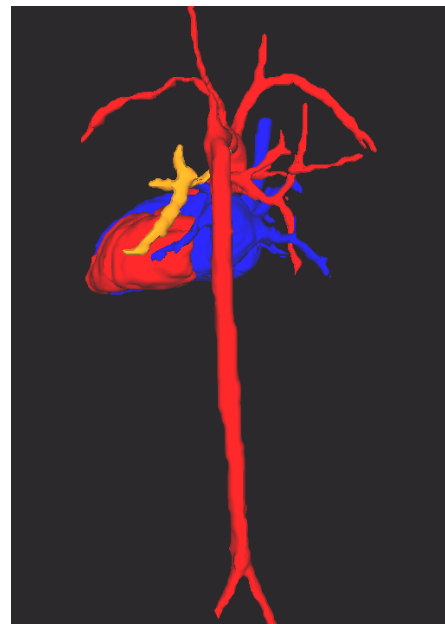


Figure 12. Rendered view of the heart and major vessels obtained by computerized post-processing of MR images in a patient with patent ductus arteriosus, a form of congenital heart disease (picture courtesy of Dr Marc Miquel, GKT School of Medicine, King's College London).



Figure 13. MR angiogram showing the abdominal aorta and its various branches, including the renal arteries that feed the kidneys with blood. The aorta forks into the right and left common iliac arteries, each of which in turn divides into external and internal branches supplying the legs and pelvis (picture courtesy of Guy's and St Thomas' Hospital NHS Trust).

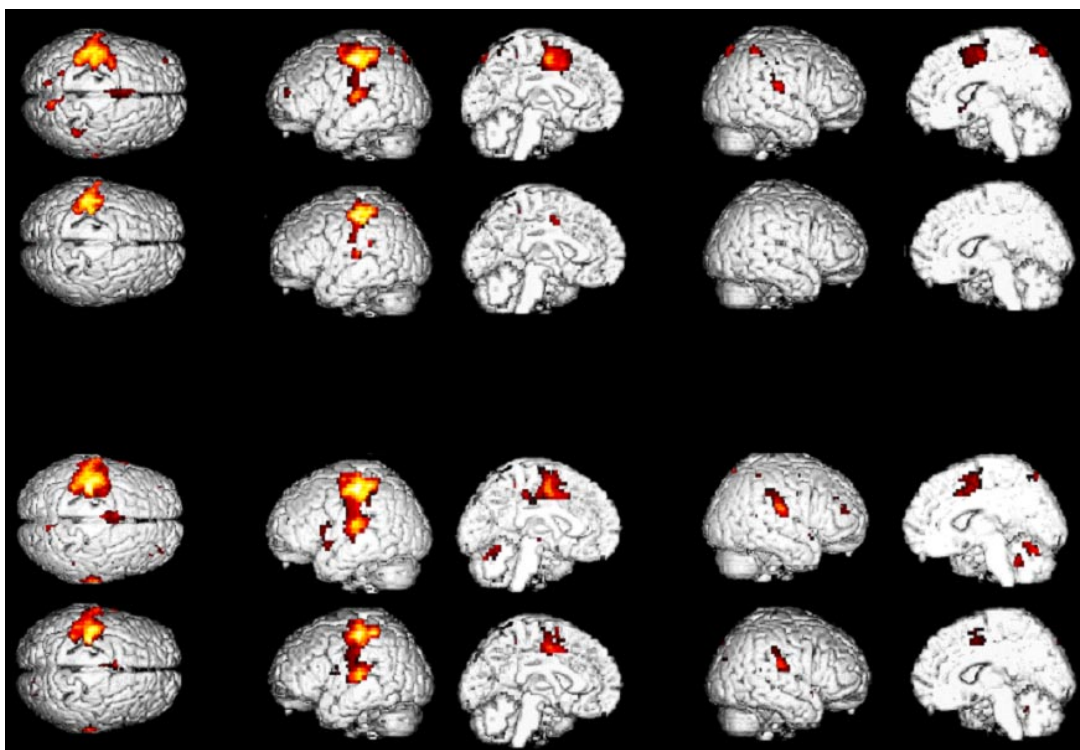


Figure 14. Brain activation due to a simple right-handed motor task investigated using fMRI. The images show activation in the part of the left hemisphere responsible for controlling the right hand (picture courtesy of Dr Andrew Simmons, Institute of Psychiatry, King's College London).

Functional MRI (fMRI) allows parts of the brain involved in processing sensory data (e.g. from the eyes and ears) or involved in motor tasks (such as moving the fingers) to be identified and studied. The technique uses the fact that activation of part of the brain results in increased delivery of oxygenated blood to that area, and the magnetic properties of this blood allow the area to be identified using MRI (figure 14). The fMRI technique is helping to improve our understanding of the basic functioning of the brain, and in individual patients it can be used to map the function of important parts of the brain prior to surgery.

Proton magnetic resonance spectroscopy (MRS) uses minute differences in the resonance frequencies of protons in different chemical environments to perform noninvasive chemical analysis. NMR spectroscopy considerably predates MRI and has wide applications outside medicine. However, MRS is now starting to have a role in the diagnosis and understanding

of conditions such as epilepsy and Alzheimer's disease, in which the relative levels of the compound visible in the spectrum are found to change (figure 15).

Carrying out surgical procedures in or close to an MR scanner might seem a dangerous or at best an unpromising idea. The high magnetic field turns conventional surgical instruments into deadly missiles, while the restricted bore of conventional MR systems offers little access to the patient. However, the development of special tools and more open magnet geometries has led to the development of interventional MRI, in which minimally invasive procedures or even open surgery can be performed inside the MR scanner. Figure 16 shows an artist's impression of an interventional MR room, similar to a suite soon to be installed at the author's institution. Developments like this have led some commentators to speculate that MR scanners may have a prominent place in the operating theatre of the future.

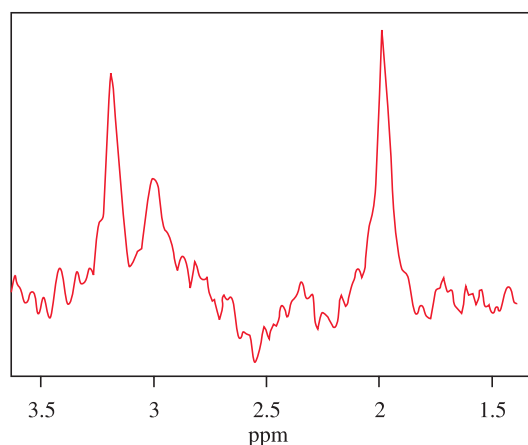


Figure 15. Proton magnetic resonance spectrum of the temporal lobe of the brain in a healthy volunteer. The three large peaks shown (left to right) are due to (1) compounds containing choline, (2) creatine and phosphocreatine, and (3) N-acetyl aspartate (picture courtesy of Dr Marcus Newbold, GKT School of Medicine, King's College London). The horizontal axis represents the resonance frequency of protons in each of the compounds in units of 'parts per million' relative to an agreed standard frequency. The height of each peak on the vertical axis is proportional to the amount of the corresponding compound present in the tissue under study.

Conclusion

This brief overview has described the basic physics of MRI and some of its current clinical applications. In the space available it has not been possible to give more than a cursory survey of the field, but hopefully this is sufficient to give an indication of the power and flexibility of MRI. New advances in techniques, instrumentation and applications are announced on a weekly basis, and we can be sure that MRI is still a long way from fulfilling its full potential in medical diagnosis and treatment.

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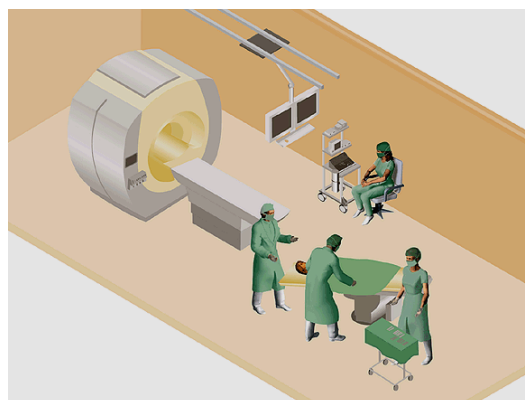


Figure 16. Artist's impression of an interventional MR facility based around a 1.5 T superconducting MR system (picture courtesy of Philips Medical Systems).

Suggestions for further reading

The physics of MRI is usually taught to physicists at postgraduate level. Many of the newer developments referred to in this article are as yet described only in research papers and highly specialized textbooks. Here is a small selection of more accessible books aimed primarily at radiography graduates.

- Elster A D and Burdette J 2001 *Questions and Answers in MRI* (St Louis: Mosby)
 Hashemi R and Bradley W G 1997 *MRI: the Basics* (Philadelphia: Lippincott Williams and Wilkins)
 Westbrook C 1998 *MRI in Practice* (Oxford: Blackwell Science)

The following website may also be of interest:

Hornack J P *The Basics of MRI*
<http://www.cis.rit.edu/htbooks/mri/>



Stephen Keevil holds a PhD in NMR spectroscopy from the University of London, and is currently senior lecturer in radiological sciences and joint director of the Research Centre for Magnetic Resonance Imaging and Intervention at King's College London. His research includes various aspects of MRI and he has taught magnetic resonance physics at all levels from A-level to PhD.