Faculty of Technology and Science
Physics

Stig E. Forshult

Magnetic Resonance Imaging MRI – An Overview

Karlstad University Studies
2007:22

Stig E. Forshult

Magnetic Resonance Imaging MRI – An Overview

Karlstad University Studies
2007:22

Stig E. Forshult. *Magnetic Resonance Imaging – MRI – An Overview*

Research Report

Karlstad University Studies 2007:22
ISSN 1403-8099
ISBN 978-91-7063-125-2

Distribution:
Karlstad University
Faculty of Technology and Science
Physics
SE-651 88 Karlstad
SWEDEN
Phone +46 54 700 10 00

www.kau.se

Printed at: Universitetstryckeriet, Karlstad 2007

Magnetic Resonance Imaging – MRI – An Overview

by

Stig E. Forshult
Department of Chemistry and Biomedical Science
Karlstad University, SE 651 88 Karlstad, Sweden

Magnetic Resonance Imaging – MRI – An Overview

Contents

Magnetic Resonance Imaging – MRI – An Overview

Stig E. Forshult
Department of Chemistry, University of Karlstads universitet
SE 651 88 Karlstad, Sweden

Abstract

Magnetic Resonance Imaging or MRI is a modern diagnostic technique for acquiring information from the interior of a body. Usually this is a human body or an animal, but MRI is also used in the industry for more technical purposes. The greatest advantage of MRI is that it can create three-dimensional images of the object under study without hurting the object in any way and without using any ionizing radiation.

The body to be imaged is placed in a strong magnetic field; more than ten thousand times as strong as the magnetic field of the Earth. A radio signal in the form of one or more short pulses is sent into the body, where it is absorbed by nuclei of hydrogen atoms. These are also called protons. The radio signal is of the same kind as radio waves used by TV and FM radio stations. The hydrogen nuclei of the body respond by creating a slowly decaying radio signal in a receiver coil. The strength of this signal mirrors the amount of protons; i.e. the concentration of hydrogen in various parts of the imaged body.

When creating an image of an organ within a body, the signal must be acquired from every part of the organ point by point by a scanning procedure. To accomplish this, the magnetic field is varied with successive gradients in three dimensions. The monitored signal is the sum of signals from every unique volume element within the body. In this way the instrument receives thousands or even millions of data creating a set of equations from which the signal magnitude of every single volume element can be calculated.

Most parts of the body have a roughly equal concentration of hydrogen. Thus, the radio signals from different tissues have similar strengths resulting in an image with low contrast. However, signals from protons decay with unequal speeds depending on their various environments. Hydrogen atoms exist in

different compounds and in different tissues. This fact can considerably influence the so-called relaxation time. Therefore the magnitude of the induced radio signal is monitored some time after the end of the primary radio frequency pulses which had started the whole process. Now various tissues show different signal strengths, from which it is possible to build an image with desired contrast. This is often excellent, even for soft tissues. Of special interest is that hydrogen in lesions and tumors may have other relaxation times than surrounding tissues and therefore can be detected in the image.

The contrast of the image is created by the experimental procedure and is not inherent in the imaged body. Thus different experimental routines will result in unequal images – not pictures – of the same object. Therefore it is crucial that the experimenter learns how to use different RF-pulse sequences and how to interpret the result.

Magnetic Resonance Imaging – MRI – an Overview

1. Introduction

Magnetic resonance imaging or MRI, first demonstrated in 1973[1] is now a mature analytical modality, which is extensively used as a diagnostic tool within clinical medicine and also in research. It has many advantages compared to imaging with *X*-rays and similar diagnostic techniques. It is non-invasive and it uses no ionizing radiation. The disadvantages are few. However, as MRI utilizes radio waves and strong magnetic fields, people with pacemakers cannot be imaged and various kinds of metallic implants within the body may also prohibit imaging. In addition, the instruments are still very heavy and expensive and they need a lot of specially designed space. Commonly they are set up in separate buildings.

MRI is in almost all cases sensitive only to one single element – **hydrogen**, which is a main constituent of every biological organ; more than 60% on an atomic basis. Thus there are almost no limitations to what samples of biological origin can be imaged. Only bone tissue, which has less and more rigidly bound hydrogen than most other parts of the body, gives a signal with inherently low amplitude. Other elements will not be seen, but some can influence the imaging procedure. The contrast in an MRI image is due to the fact that hydrogen atoms in different tissues and compounds have a slightly different chemical and magnetic environment[2]. As a consequence they will respond somewhat differently to radio waves in the shape of short (radio frequency) ***RF*-pulses**, which are sent into the studied object. This makes it possible to detect pathological changes deep within an organ[3]. Here MRI has its greatest advantage over most other imaging techniques.

The first human imaging with MRI took place more than a quarter of a century ago[4]. At that time it took several hours to produce an image, an operation, which today can be performed within minutes or even seconds. The very much faster and more reliable machines now commercially available have given physicians a very valuable diagnostic tool. However, there is still need for the professional judgment and skill when evaluating the images and also when choosing the imaging parameters in order to achieve the desired contrast. And developments are ongoing. New pulse sequences for special tasks and enhanced contrast are continuously being invented. Faster scanning routines make it

possible to image dynamic processes as blood flow and drug metabolism within the body in real time. This is also the basis of **fMRI**[5], where subtle changes in brain metabolism caused by various mental or physical activities, can be studied. Even the magnets are being improved but more slowly. There is a trend towards lighter and more shielded magnets making the stray field weaker in the vicinity of the apparatus in spite of an increased working field. The next breakthrough could be the construction of "high temperature" super conducting magnets, which would need only liquid nitrogen at 77 K and not expensive liquid helium at 4 K in order to be cooled to super conducting condition, i.e. to conduct an electric current without any resistance at all.

1.1 Imaging

Imaging can be done in many ways. Usually, but not always, electromagnetic radiation is utilized. Here the golden rule is that it is not possible to image objects with a dimension less than the wavelength used. However, this is not valid for MRI, which will be shown later.

Most photographic imaging uses ordinary visible light (wavelength ca 0.5 μm), which is reflected from the surface of an object. Visible light can also be used in transmission mode as in the common microscope and slide projector. To get additional or more detailed information from the surface of a body one can switch to longer wavelengths as in *IR*-imaging (the heat-camera) or to shorter ones and use *UV*-light (wavelengths from 0.4 μm and less) or even *X*-rays with wavelengths in the order of nanometers. Much longer wavelengths (mm–cm) in reflection mode come to use in Radar. To image very small objects (nm–μm) beams of electrons can be used as in electron microscopes, where both transmission and reflection modes are possible. A tunnel microscope, on the other hand, examines a surface in much the same way as one can do with ones fingertips, though almost on an atomic scale. Also reflection of ultra-sound with wavelengths in the mm-range is a commonly used technique in both medicine and engineering.

Common to all these imaging modalities is the fact that they result more or less in only two-dimensional images. The object under study is passive, merely reflecting or attenuating the radiation, which is directed onto it or through it. In contrast, MRI is a true three-dimensional imaging technique, where the object

itself is active, responding in various ways to the radiation, which is sent into it – not only onto it. In this way it has some resemblance to computed tomography (CT, CAT) and positron emission tomography (PET), where one also can get information from small specific volumes deep inside a human body.

1.2 Background

The foundation of MRI – at the start called NMR imaging – is the physical phenomenon **nuclear magnetic resonance, NMR**, which was detected before[6] and developed just after WWII[7] as proton (hydrogen) magnetic resonance. In the beginning it was a spectroscopic technique among others used mostly for structure identification in organic chemistry, even though gradient technique[8] and the NMR of biological samples[9] were in fact tested very early. The major difference compared to other kinds of spectroscopy is that the sample being studied must be positioned in a very strong magnet field, in the order of one Tesla = 10,000 Gauss. This is more than ten thousand times the strength of the magnetic field of the Earth, which is about 0.5 Gauss. However, because of the need for an extremely homogeneous magnetic field, in the order of 10 ppm, and even better for high-resolution NMR spectroscopy, the instrument must be shielded from all kinds of external magnetic fields and electromagnetic radiation.

With the introduction of **pulsed *RF*-technique** (instead of the earlier continuous wave) in combination with computerized **Fourier transformation**, **FT**[10] in the beginning of the seventies, NMR took a giant step forward making many new applications available. The next almost simultaneous major step was the utilization of **super conducting magnets**[11], which can create magnetic fields that are a magnitude higher than ordinary resistive coils. NMR instruments with magnetic fields of more than 20 Tesla (T) are now commercially available. The stronger the magnetic field the higher the frequency of the *RF*-pulse. The proportionality factor is called the **gyro-magnetic ratio**, which is $\gamma = 42.58$ MHz/T for protons. Therefore the NMR-instruments are often labeled 200 MHz or 800 MHz etc, showing the resonance radio frequency, which they use for proton NMR spectroscopy. For human imaging however, magnetic fields in the order of 0.1–4 T are commonly used[12]. This means that the *RF*-pulses will have frequencies up to about 170 MHz –

not far from commercial TV and FM radio stations, which can interfere with the imaging and vice versa.

Table 1 **Some biologically important nuclei**[13]

Element name	**Nucleus symbol**	**Natural abundance** %	**Spin I**	**Nuclear moment*** μ_N	**Gyromagnetic ratio, γ #** MHz/T
Hydrogen	^{1}H	99.985	1/2	2.79284	42.5775
Deuterium	^{2}H	0.015	1	0.85743	6.5359
Helium	^{3}He	1.38 ppm	1/2	–2.12762	–3.2436
Carbon	^{12}C	98.89	0	—	—
	^{13}C	1.11	1/2	0.70241	10.7084
Nitrogen	^{14}N	99.63	1	0.40376	3.0777
	^{15}N	0.37	1/2	–0.28319	–4.3173
Oxygen	^{16}O	99.76	0	—	—
	^{17}O	0.038	5/2	–1.89379	–5.7743
Fluorine	^{19}F	100	1/2	2.62897	40.0776
Sodium	^{23}Na	100	3/2	2.21752	11.2695
Phosphorus	^{31}P	100	1/2	1.13160	17.2514
Potassium	^{39}K	93.26	3/2	0.39146	1.9895
Calcium	^{40}Ca	96.94	0	—	—
	^{43}Ca	0.135	7/2	–1.3173	–2.8697
Xenon	^{189}Xe	26.44	1/2	–0.7768	11.8604

* The nuclear moment μ, is expressed as the number of nuclear magnetons, μ_N, where $\mu_N = e\hbar/2 \cdot m_p = 5.050783 \cdot 10^{-27}$ J/T. Nuclei with spin, I>1/2 also possess quadrupole moment, which make them less useful for MRI studies.

\# The gyromagnetic ratio γ, is the radio frequency at which the nucleus absorbs energy in a magnetic field of 1 Tesla.

Today NMR is probably the most versatile of all spectroscopic techniques with a very rapidly growing use in biochemistry, biology, and even in medicine, despite the fact that the instruments are still rather large and expensive. NMR is no longer limited to hydrogen. Also carbon, in the form of the rare nuclide carbon-13, fluorine, phosphorus, and several other elements (with magnetic nuclei, i.e. nuclei with spin, $I \neq 0$) are now easily available for routine study.

However, in MRI still almost only hydrogen is utilized. In special cases one can make use of Phosphorous-31[14] or of Fluorine-19[15], Helium-3 and Xenon-189 (see section **5.1**), which do not occur naturally in humans.

1.3 Spectroscopy

Spectroscopic techniques exist in many shapes. The best knowns are absorption and emission spectroscopy. In the first kind electromagnetic radiation of various wavelengths (frequencies) is continuously directed towards a sample, where some wavelengths are attenuated more than others. Commonly *UV*-, *VIS*- or *IR*-wavelengths are utilized. The absorbed energy lifts some molecules into a higher energy state. They get **excited**. However, they immediately de-excite again, resulting in a steady state concentration during irradiation. The transmitted wavelength pattern can be used for analytical purposes – both quantitative and qualitative. Also *X*-ray imaging works in approximately this way and continuous wave NMR, which was used before about 1970.

In emission spectroscopy the sample is heated or energized with an electrical discharge or in some other way – also here more or less continuously. Some molecules get excited and when they de-excite, they emit their excess energy in the form of electromagnetic radiation. The wavelength pattern of this emitted radiation can be measured and analyzed in much the same way as in absorption spectroscopy. In fact, a molecule, which can absorb radiation of a certain wavelength, can usually emit the same wavelength after being excited.

Fluorescence spectroscopy is a combination of absorption and emission. The sample is continuously irradiated, but only with one wavelength, which is absorbed by the compound being studied. Other compounds will be very little affected. The absorbed electromagnetic energy makes some of the molecules excited. They quickly re-emit their excess energy as electromagnetic radiation, which again is measured and analyzed. Where this technique can be used, it is usually both sensitive and very selective.

All the spectroscopic varieties discussed in the above paragraphs are based on the fact that excited molecules relax very fast, i.e. they return almost instantaneously to their energy ground states. This takes usually only about 10^{-8} seconds, even if there are examples from fluorescence, where the relaxation can

last much longer, for even seconds and more. Then one usually talks about phosphorescence. In most spectroscopy however, the very fast ongoing excitation – de-excitation process results in a steady state situation with very few excited molecules, which is stable throughout the whole analytical procedure.

NMR and MRI are based on a quite different principle. As in fluorescence spectroscopy the sample is energized with electromagnetic radiation. However, in NMR the radiation is sent into the sample from a *RF*-emitter as one or more very short pulses of linearly polarized radio waves of a suitable single (almost) frequency. These pulses last less than a tenth of a millisecond. Hydrogen nuclei in the sample get excited into their higher energy state from which they begin to **relax** back to the ground state. But they do so without emitting any radiation. This process is many magnitudes slower compared to most other relaxation processes mentioned earlier. It takes usually from milliseconds up to several seconds – depending on the chemical and magnetic environment of the protons – before the sample is back into its ground state again. During the relaxation process however, the excited protons will induce a weak *RF*-signal in a suitably positioned detector coil, which can be the same as the emitter, a so-called volume coil. However, in most cases specially designed detection coils are used as surface coils or array coils. This *RF*-signal, the so-called **free induction decay**, **FID**, contains all the information that can be extracted from the system studied. As the *RF*-wavelength is about five meters, no image can be directly projected. It has to be reconstructed using the information in the FID. The pulse technique makes it possible to vary the excitation situation in an almost unlimited number of ways and thus to get much more information from the system than with other spectroscopic alternatives.

2 Spatial Information

In imaging the most important thing besides measuring the signal amplitude is to know from where in the test object the signal originates. For this the magnetic resonance procedure is superior as it can give rise to three-dimensional images in situations where most other techniques can show only two. This is achieved because the *RF*-pulses excite exclusively hydrogen atoms that are at **resonance**, i.e., which feel exactly the right magnetic field and therefore oscillate with the same frequency as the radio waves. To get spatial information the magnetic field is varied linearly through the test object with the

aid of **magnetic field gradients**[16] in three dimensions. These gradients are active only during short and different parts of the scanning procedure. This places great demands on the electronic system of the MRI instrument, which very quickly has to switch on and off strong electric currents with high precision. Rising and declining magnetic gradients will also make a lot of **noise**, which can be very disturbing and even harmful[12].

2.1 Magnetic Gradients

The gradients divide the object studied into a great number of volume elements, called **voxels**, each of which has a volume of about one cubic millimeter or less. Every voxel gives rise to one signal with a unique amplitude. Calculating the amplitudes of all these signals and their space coordinates in the studied body is more or less what MRI is about. However, all the millions of voxel signals are added together and the MRI instrument can only monitor their sum. Thus one must utilize a scanning procedure and acquire a large number of aggregated signals during a series of consecutive experiments. These signals are Fourier-transformed and used to build an image of the object with a spatial resolution equal to the voxel volume. The three-dimensional image can be presented as a 3D illusion, which may be turned and twisted, but more often as a series of two-dimensional images. The plane for these can usually be freely chosen.

When imaging a patient the main magnetic field of a few Tesla (up to 8 T is allowed by FDA[12]) will in most cases run along the lying body (see figure 1). This is by convention the **longitudinal** z-direction, even if it is often drawn vertically. A magnetic gradient (G) of about 5 mT/m (0.5 Gauss/cm) is switched on in this direction for a short moment (τ), about 10 ms (followed by a reverse gradient for half the time to compensate for the phase shift introduced). Halfway into the magnetic gradient a *RF*-pulse is sent into the body. It should have a duration of much less than one millisecond and contain only a narrow band of closely lying frequencies around the desired one. The *RF*-pulse excites protons within a thin **axial** cross-section of the body, i.e. in an axial **slice** with a thickness of about one millimeter or less[17]. After the pulse only the excited protons will relax and induce a sinusoidal signal with gradually decaying amplitude in the receiving coil. This signal is the free induction decay, the **FID** (see figure 15). A shallower gradient or a broader frequency band will

result in a thicker slice, i.e. lower resolution. The slice thickness, d, can be approximately calculated by equation (1).

$$d = (\gamma \bullet G \bullet \tau)^{-1} \approx 0{,}5 \text{ mm}, \quad \text{with figures in the paragraph above} \qquad (1)$$

where the gyromagnetic ratio for protons is, $\gamma = 42.58 \bullet 10^6$ Hz/T, according to table 1.

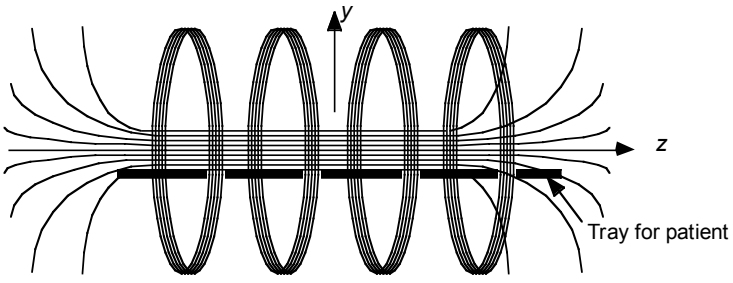


Figure 1 Coils creating a homogenous horizontal magnetic field along the *z*-axis. On the periphery of the big coils are smaller so-called saddle coils (see figure 2), which create magnetic gradients in the *x*- and *y*-directions

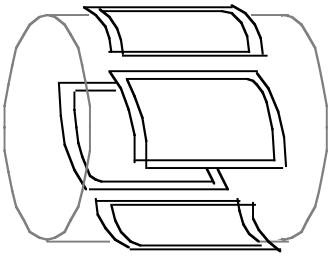


Figure 2 Pair-wise arrangement of saddle coils within the main ones for creating magnetic gradients in the *x*- and *y*-directions. For the *z*-gradient there are additional coils concentric with the main ones.

2.2 Three Dimensional Encoding

Information from the two other dimensions can be extracted with so called **phase** and **frequency decoding**[18]. This is because the frequency of the sinusoidal radio signal depends on the strength of the magnetic field in a way that will be described later – the stronger the field the higher the frequency. However, as all excited protons experience (almost) the same magnetic field

they will also create a coherent radio signal with (almost) a single frequency. If, when the relaxation process has started, another magnetic gradient is switched on along the *y*-axis, perpendicular to the first one, protons at different levels of the axial slice will oscillate with different speeds. Thus, when the gradient is switched off again, the signals from the front side of the body will be slightly ahead of those from the rear side (or opposite). They will have different **phases**, when the read-out starts. These phases describe how far from the front of the body the signal comes. Usually 128 or 256 phase encoding gradients with increasing steepness are used.

During read-out a third gradient along the *x*-axis – perpendicular to the other two – is switched on. Then signals from the left and right sides of the body will have different frequencies and can be distinguished from one another with the aid of a Fourier transform of the FID signal[19]. Even here the axis is digitized into 128 or 256 steps for every step of the phase encoding gradient. In this way imaging of several axial slices can be made – one after the other. However, it is equally possible to create images of coronal or sagittal slices or in any desired direction and even curved slices by the proper choice of magnetic gradient sequences. In fact most slices are slightly curved as it is almost impossible to make the magnetic field perfectly homogenous.

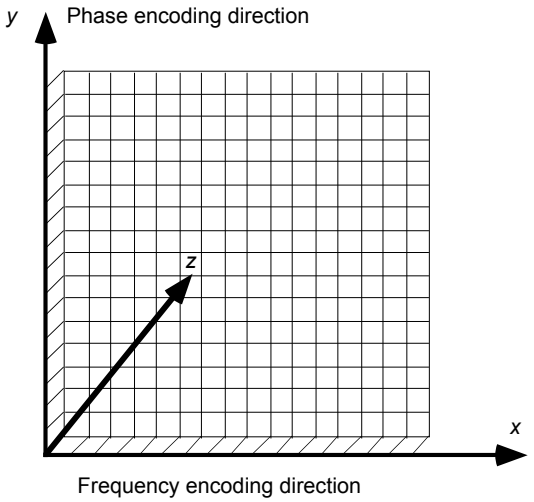


Figure 3 Slice with 16x16 voxels. After phase and during frequency encoding voxels in the *x*-direction have signals with different frequencies and along the *y*-axis they have different phases. Thus every voxel shows a unique frequency-phase combination.

Depending on the desired resolution a single slice may consist of 256x256 = 65,536 voxels. To calculate the signal amplitudes from all these voxels one needs to solve at least an equal number of equations, thus the great number of magnetic gradients. Every signal the detection coil picks up is a sum of signals from all voxels with different space-related frequencies, phases and amplitudes. It must be sampled with at least twice the signal frequency[20]; i.e. at about 200 MHz. During a 100 μs readout this will result in some 20,000 data points (times 256 read-out cycles). The sampled data belong to the *spatial frequency domain* (inverted distance) and are stored in the computer in an array called ***k*-space**. Every *line* in *k*-space holds the data sampled after one phase-encoding gradient. However, there is no direct connection between the cells in *k*-space and the voxels of the object. Instead the spatial frequency domain data in *k*-space must be two-dimensionally Fourier transformed into the *real space* domain. These data can be converted into an image of the object, as every combination of frequency and phase is related to a single point in space, to a single voxel. In reality the situation is slightly more complicated as the detection coils have different sensitivities in different directions and distances and because the signal will often decay during read-out etc. Corrections for such things are built into the standard sampling routines. There are also other ways to sample the data.

The image produced will consist of a large number of points with various amplitudes, which usually are presented as different shades of gray or occasionally color-coded. It is important to understand that this is not a picture of the examined object. It is a coded image, which has to be interpreted by an experienced user. Primarily the amplitudes reflect the number of mobile and fairly loosely bound (water and fat) protons in the voxel where the signal originates. However, as most biological tissues have similar proton (hydrogen) densities (see table 2, which shows average values as the literature is not consistent), the contrast in such an image is not as good as desired. To enhance the contrast one can use the fact that hydrogen atoms in various chemical environments have a broad range of different **relaxation** speeds. This can be used in an almost unlimited number of ways to create images with lots of information. One example of this has already been mentioned. Because of very short relaxation time for hydrogen atoms in bone tissue, these will already have relaxed when the read-out procedure starts. Thus bone will give a very weak signal in contrast to soft tissue. Also the lungs will give weak signals because of their very low content of protons (and other nuclei). On the other hand, tumors

are often vascular and thus give rise to rather strong signals.

Table 2 **Relative effective mobile hydrogen densities of some tissues**[21]

Blood	93	Liver	81
Bone	12	Lung	5
Cerebrospinal fluid	96	Muscle	82
Fat	88	White matter	70
Gray matter	84	Water	100

3 Theory

3.1 The Larmor Frequency

Even a very small piece of biological tissue contains billions of billions of hydrogen atoms. Actually 1 cm^3 of any organ contains more than 10^{22} hydrogen atoms. The nuclei of hydrogen atoms – the protons – act as small compass magnets, which normally have a totally random orientation and equal energy. For every magnet pointing in an arbitrary direction there is another one pointing in the opposite direction. Thus they average out the magnetic moments of one another and no external magnetic effect is seen.

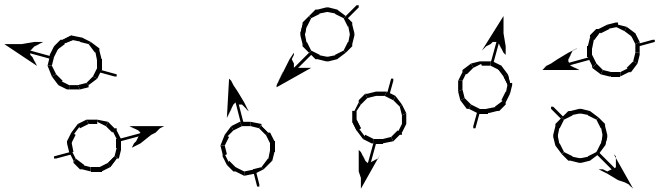


Figure 4 Protons in no external field, symbolized by magnets pointing in random directions

However, if one places a sample containing hydrogen in a magnetic field, B_0, a new situation arises. The magnetic field defines a direction in space, which by convention is the longitudinal z-axis. The magnetic moments of the protons line up along this axis, half of them parallel to the field (as compasses) and the other half (almost) anti-parallel to the field. The protons with magnetic vectors parallel to the field ($m_I = 1/2$) will have a slightly lower energy than the others ($m_I = -1/2$) and because of this, there is a small but very important difference in population between the two energy levels. Thus, the magnets no longer cancel out exactly. At equilibrium a small net magnetization is left along the

positive *z*-axis. This is called the **net magnetization vector**, the **NMV**, often symbolized by $\mathbf{M_z}$ (see figure 5). In the *xy*-plane the magnetic moments of the protons still cancel out. The magnitude of $\mathbf{M_z}$ and accordingly, the amplitude of the MR-signal are proportional to the proton density, the external magnetic field, and to the square of the gyromagnetic ratio, γ. This has resulted in a move towards ever-higher fields.

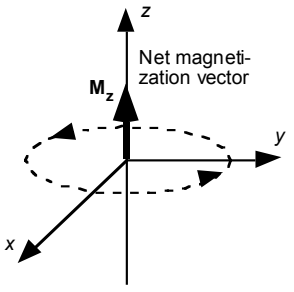


Figure 5 Net magnetization of protons in an external field

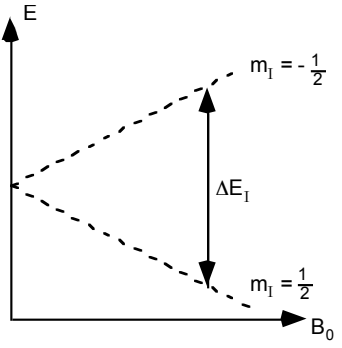


Figure 6 The energy difference, ΔE_I, depends on the magnetic field, B_0

The energy difference between the two levels can be calculated by means of equation (2).

$$\Delta E = h \bullet \gamma \bullet B_0 \qquad (2)$$

In this formula h is Planck's constant, $h = 6.626 \bullet 10^{-34}$ Js, and γ the gyromagnetic ratio for protons (see table 1). With $B_0 = 1$ Tesla the energy difference is $\Delta E = 2.821 \bullet 10^{-26}$ J, which is a very small energy indeed. Nevertheless, it is the energy needed to "lift" or excite protons from their lower energy level to their higher one. This can be done with the aid of electromagnetic radiation of a suitable frequency, *f*, which must have photon energy of ε according to equation (3). The photon energy and the energy difference, ΔE, between the two proton levels must equalize for resonance to occur.

$$\varepsilon = h \bullet f = \Delta E \qquad (3)$$

$$f = \gamma \bullet B_0 \qquad (4)$$

As can be seen in equations (2) and (4) both ΔE and *f* are linearly dependent on B_0. At a magnetic field of 1 Tesla the resonance frequency, is $f = 42.58$ MHz. This is called the **Larmor frequency**, f_L. The Larmor frequency should be compared to FM radio, which uses frequencies around 100 MHz, and

television, which broadcasts between 50 and 90 MHz and from 200 MHz and up, i.e. in the same range as MRI.

In most other spectroscopic techniques the energy difference is much larger, leaving the higher energy level virtually unpopulated. This is not at all the case in NMR. The population ratio between the two levels can be calculated with Boltzmann's distribution law, equation (5), where Boltzmann's constant is $k = 1.381 \cdot 10^{-23}$ J/K.

$$N_{low}/N_{high} = exp(-\Delta E/kT) \quad (5)$$

Here the indices refer to the parallel and the anti-parallel states respectively. At body temperature, i.e. at T = 37° C = 310 K and a field of 1 Tesla one gets $N_{low}/N_{high} \approx 1.000007$. This means that per one million protons in the higher energy state there are one million and seven in the lower state, i.e. an excess of only about 7 ppm. Nevertheless, this small excess is what creates the important net magnetization vector.

3.2 Longitudinal relaxation

If electromagnetic radiation with frequency $f = 42.58$ MHz is sent into a sample in a field with $B_0 = 1$ Tesla, there is a statistical probability that some of the protons will absorb one photon each and raise their energy states. It should be noticed however, that according to Einstein's absorption/emission law, there is an exactly identical probability for protons in the higher energy state to be pushed down to the lower state. With a pulse of suitable magnitude and duration, a so-called **90-degree** pulse along the *x*-axis, it is possible to force exactly half of the protons on each level to change states. Now the populations are equalized and no net magnetization vector exists in the *z*-direction. The system is **saturated** and NMV is zero, $\mathbf{M_z} = 0$.

Immediately after the end of the pulse the system starts to relax back to its equilibrium state by transferring excess energy into the lattice to which the protons are bound. Accordingly the *z*-magnetization recovers and the NMV is reestablished. Exactly the same thing happens, when a fresh sample is put into a magnetic field (see figure 7). This process can take from less than milliseconds up to several seconds (see table 3) depending on the proton environment and the magnetic field. Usually the process is well described by equation (6), which is a first order exponential function.

$$\mathbf{M}_{z,t} = \mathbf{M}_{z,eq} \bullet (1 - exp\ (-R1 \bullet t)) \qquad (6)$$

Here R1 is the longitudinal relaxation rate constant or spin-lattice relaxation rate constant. However, it is more common to use the time constant, T1, which is the reciprocal of R1. T1 is the time, when $1/e$ of the excited protons has not yet relaxed. This is close to one third (e = 2.718). After two T1s about 15% non-relaxed protons remain and after five T1s less than 1%. Thus one has to wait for a time equal to several T1s before the system is back to equilibrium again and one can do another experiment without it being affected by the previous one.

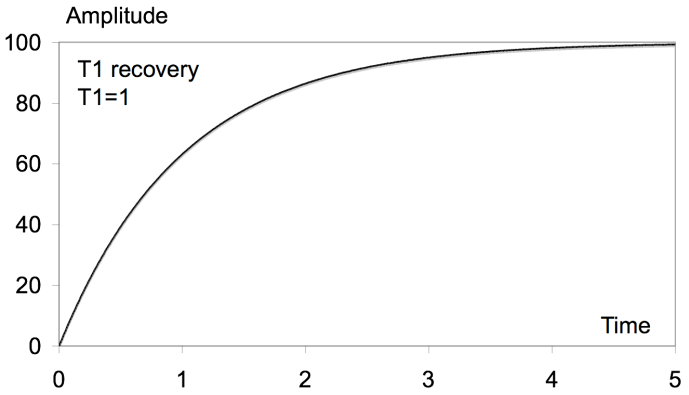


Figure 7 Recovery of the z-magnetization after a 90-degree pulse

Occasionally the half-life of the system is used instead of T1. This is a term more common in radioactive decay, which obeys exactly the same mathematical laws as most NMR relaxation processes. One half-life, $T_{1/2}$ = T1•*ln* 2, is the time constant multiplied by *ln* 2 = 0.693. When one half-life has elapsed, half of the excited protons have relaxed.

Table 3 **Approximate values of T1 (ms) of some tissues at field strength of 1,5 Tesla**[22]

Blood	1000	Liver	500
Bone	short	Lung	–
Cerebrospinal fluid	2000	Muscle	850
Fat	250	White brain matter	800
Gray brain matter	900	Water	3000

Table 3 shows that hard and more or less solid tissues show short T1s, whilst tissues containing much water have longer relaxation times but always shorter than pure water. Vascular tumors and edema usually have rather long relaxation times. T1 is somewhat dependent on the magnetic field and values in the literature vary considerably between sources.

Depending on the strength of the RF-pulse more or less than half of the protons will change states. If one doubles the pulse energy a **180-degree** pulse is created, which will turn all the magnetic moments of the protons upside down. The NMV is now **inverted** and aligned along the negative z-axis, $\mathbf{M_z}<0$. It will relax back to equilibrium with exactly the same time constant as above according to equation (7).

$$M_{z,t} = M_{z,eq} \bullet (1 - 2 \bullet exp\,(-t/T1)) \qquad (7)$$

Here R1 from equation (5) has been exchanged for 1/T1. After one half-life, i.e. when t = 0.693•T1, the z-magnetization goes through zero – the null point – before increasing again (see figure 6). This can be used to null desired signals during imaging and is called **inversion recovery**, **IR**[23]. In this case a 180-degree pulse is sent into the sample at time, **TI**, the **inversion time**, before the 90-degree excitation pulse. With carefully chosen TI certain tissues, usually CSF or fat, can be made almost invisible in an MRI image.

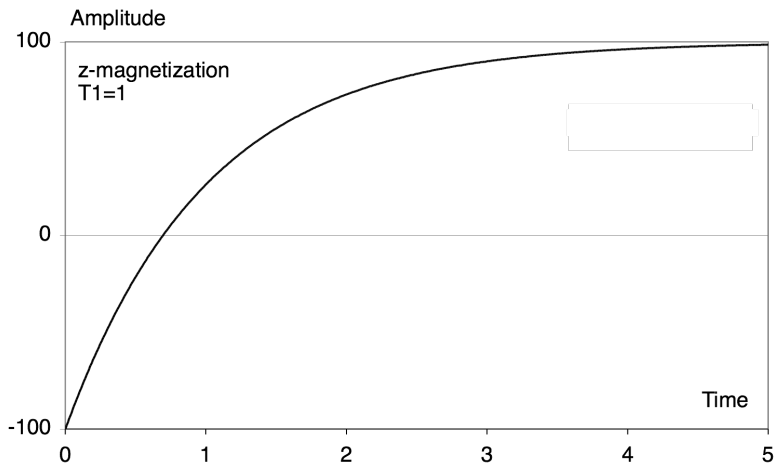


Figure 8 Graph of the recovery of the z-magnetization after a 180-degree pulse

An ideal RF-pulse should be rectangular with zero frequency bandwidth. However, to achieve this is theoretically impossible and every RF-pulse has

inevitably a certain non-zero bandwidth; the shorter the pulse the broader the bandwidth. In MRI this is commonly about 500 Hz. The pulse-form is usually a truncated sinc-pulse $\left(\text{sinc} = \frac{\sin x}{x}\right)$, which is symmetric in time (see figure 9). However, if the pulse length is increased to "infinity", the bandwidth can be made very narrow. Thus one can irradiate a sample with a very precise frequency during a specified time. Then hydrogen nuclei at resonance with this frequency – and only these – will continuously change states. Their net magnetic moment is now zero and they will not affect the environment in any way. With this technique one can more or less specifically wipe out signals from protons in fat or other tissue of choice because the Larmor frequency varies slightly – only a few ppm – between protons in different compounds.

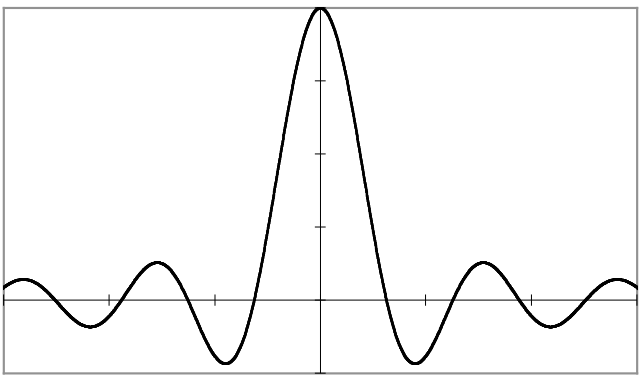


Figure 9 The sinc graph

A similar way to accomplish this is with a weak and tuned 90-degree RF-signal, which should be on for ten milliseconds or so in order to presaturate certain protons immediately before the 90-degree excitation pulse. Then these protons will be silent throughout the signal acquisition. The method is called **spectral selective inversion recovery**, **SPIR**[24] and is often used to suppress signal from fat in MR angiography, when imaging the spine etc.

Protons in proteins and other macromolecules have such a very short T1 and broad resonance spectrum that they normally are invisible in MRI as are protons in water tightly bound to proteins. The proteins, however, easily exchange energy with protons in the surrounding free water via the bound water. This can shorten T2 of free water and lower the contrast in the image. However, a RF-pulse tuned slightly – a few kHz – off-resonance, will be absorbed by the proteins and via **magnetization transfer**[25] protons in bound

water are saturated. They are no longer able to act as an energy sink for surrounding free water. In this way it is possible to enhance the contrast, especially when imaging the brain.

3.3 Transverse relaxation

Longitudinal magnetization along the *z*-axis is the prime requirement for the NMR phenomenon. However, $\mathbf{M_z}$ will give no signal in the receiving coil, which is positioned along the *y*-axis. To understand how the NMR-signal appears another effect has to be taken into account, the transverse magnetization in the *xy*-plane, $\mathbf{M_{xy}}$, perpendicular to the B_0 field.

When a sample is put into the magnetic field B_0, its protons do not line up exactly parallel and anti-parallel to B_0. Instead they will be at an angle to the *z*-axis creating magnet vectors also in the *xy*-plane (see figure 10). Despite this there will be no net *xy*-magnetization. This has a quantum mechanical background, but it can be partly visualized if one assumes the proton magnetic vectors to be "spinning tops". These rotate or precess in an uncoordinated way but with the same speed around the *z*-axis making the *x*- and *y*-magnetic vectors to cancel. Their precessing speed is exactly the Larmor frequency as calculated in equation (4). This is the resonance condition. Only *RF*-pulses with electromagnetic radiation at the Larmor frequency are able to exchange energy with the rotating magnetic vectors.

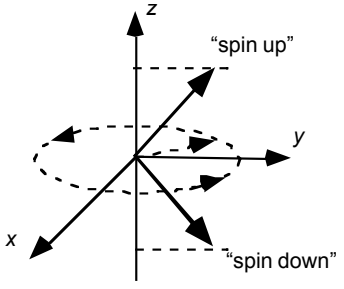


Figure 10 Symbolic representation of rotating proton spins

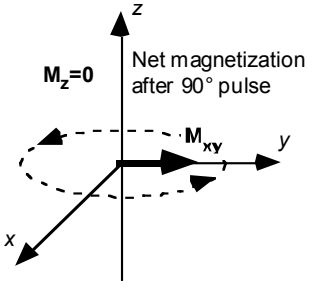


Figure 11 Net magnetization directly after a 90° pulse

The *RF*-pulse is sent into the sample along the *x*-axis. It has its magnetic field vector polarized in the *y*-direction. In addition to changing the *z*-magnetization as described in the previous paragraph, all *xy*-magnetization is now forced onto

the *y*-axis. The net effect is that the *z*-component of the NMV is tipped 90 degrees from the *z*-axis down to the *y*-axis (see figure 11). However, the magnetic vectors will continue to precess around the *z*-axis, but now in the *xy*-plane. The *z*-magnetization, $\mathbf{M_z}$, has been transformed into an equally large and coherently rotating *xy*-magnetization, $\mathbf{M_{xy}}$, which will induce a signal, the FID, in a receiver coil situated outside the sample along the *y*-axis (or *x*-axis). When the pulse is over, the coherence of this transverse magnetization will successively decay via a process called **transverse** or spin-spin relaxation (see figure 12). As the name indicates, this has to do with magnetic coupling to neighboring spins. The process can mathematically be described with an equation (8) similar to equation (6) for longitudinal relaxation.

$$\mathbf{M_{xy,t}} = \mathbf{M_{xy,0}} \bullet exp\,(-R2 \bullet t) = \mathbf{M_{xy,0}} \bullet exp\,(-t/T2) \qquad (8)$$

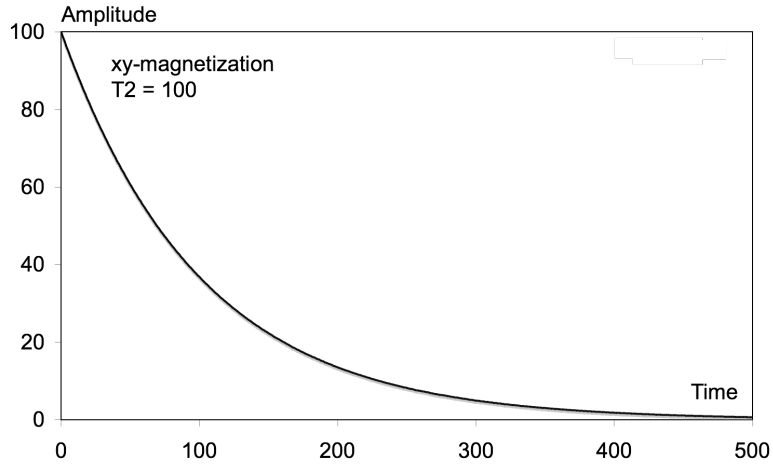


Figure 12 Decaying coherence of the transverse magnetization

Here R2 and T2 are the transverse relaxation rate and transverse relaxation time constants respectively. T2 is always shorter than T1 (see table 4). However, the two kinds of relaxation always happen simultaneously, which is symbolically shown in figures 13 and 14.

Equation (8) shows how the envelope of the FID will develop. The FID is a sinusoidal signal with exponentially decaying amplitude. This signal is induced in the receiver coil, when the rotating hydrogen magnetic vectors pass the coil twice a period. Every other time it gives rise to a positive voltage and the next time to a negative one. Thus the FID will oscillate with the Larmor frequency, f_L, as shown in figure 15.

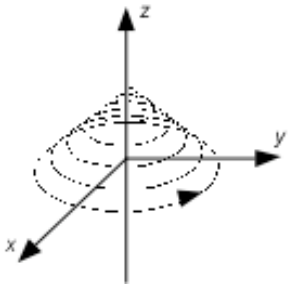


Figure 13 Symbolic representation of the NMV spiraling from $\mathbf{M_{xy}}$ back to $\mathbf{M_z}$

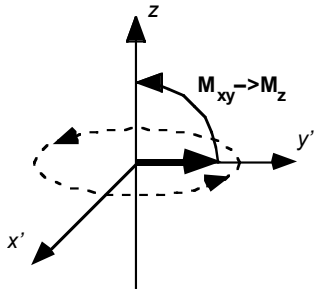


Figure 14 Showing how the NMV relaxes back from the *xy*-plane into the *z*-direction

Table 4 **Approximate values of T2 (ms) of some tissues**[22]

Blood	180	Liver	40
Bone	very short	Lung	—
Cerebrospinal fluid	250	Muscle	45
Fat	80	White brain matter	90
Gray brain matter	100	Water	2500

Besides the pure spin-spin relaxation process, there are also other phenomena which will contribute to the **dephasing** of the *xy*-magnetization. The most obvious is that the effective magnetic field is never perfectly homogenous throughout the whole sample. One reason is small inhomogeneities in the applied field. To cope with this, the machine specifications usually are about 10-50 ppm for MRI-instruments and less than 0.1 ppm for high-resolution NMR spectrometers, where much effort (and money) is used to make the field as homogenous as possible. It is also common practice to regularly shim the magnet, that is to check the homogeneity of the field and to make corrections if necessary. This can often be done automatically.

However, not even the best magnets can compensate for the small magnetic field differences which are caused by the non-identical chemical environment felt by the protons. These will give rise to so-called **chemical shifts**[26], as differently bound protons have slightly different Larmor frequencies. This forms the basis of NMR spectroscopy.

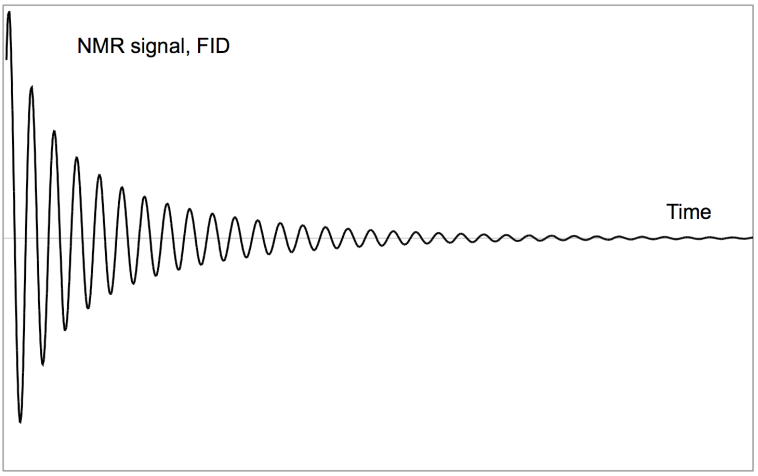


Figure 15 A schematic single frequency free induction decay curve. Not to scale.

With the aid of Fourier transformation (FT) the different frequencies can be extracted from the FID, creating an NMR-spectrum with signal intensity as a function of frequency (see figures 16 and 17). Of special interest in this case is that the Larmor frequencies for fat and water differ by about 3.5 ppm, resulting in a chemical shift[27] of 0.035 Gauss or 150 Hz at a main field of 1 Tesla (= 10,000 Gauss) and more at higher fields. This may give rise to a small spatial mis-interpretation of the origin of the MRI signal, but it can also be used to enhance the image contrast as described in the last paragraphs of **3.2**.

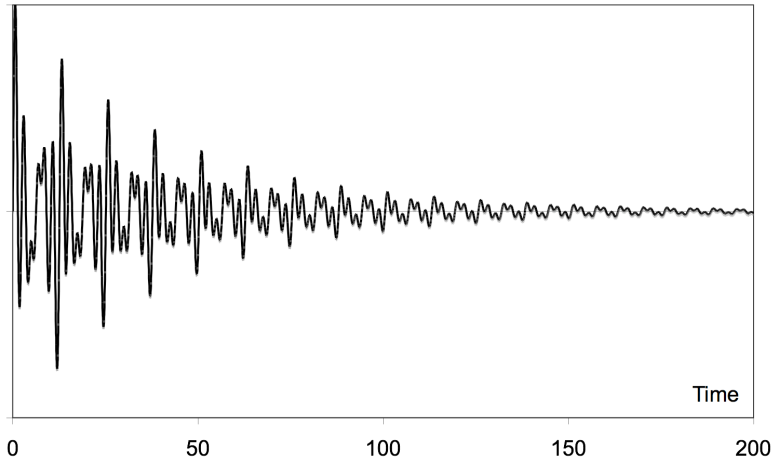


Figure 16 A simulated FID for a simple compound

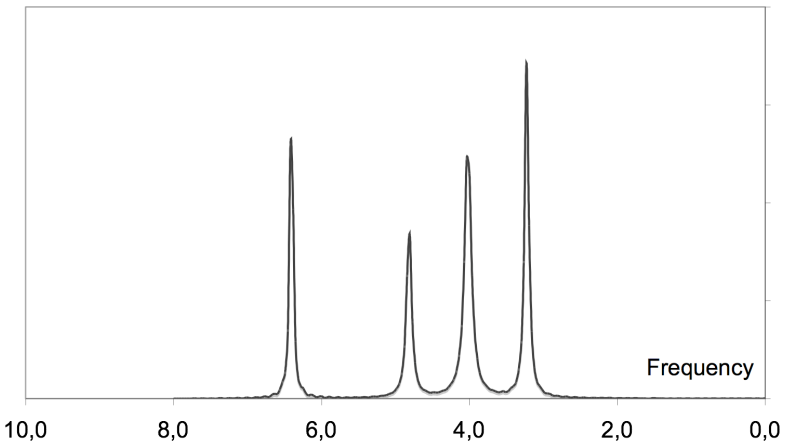


Figure 17 Low resolution NMR-spectrum corresponding to the FID in figure 16. The frequency *x*-axis shows relative units.

3.4 The spin echo

In MRI all kinds of field inhomogeneities, whatever the background, make proton magnetic vectors at various locations precess with slightly different speeds. They will therefore dephase with an effective time constant **T2***, which can be 2-50 times shorter than T2; the better the magnet the less difference between T2 and T2*. Also the FID signal will decay accordingly.

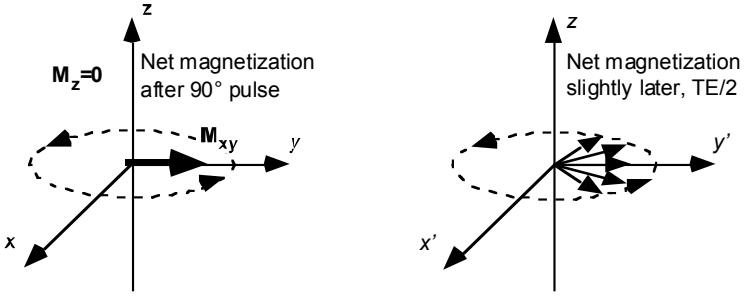


Figure 18 a) Net magnetization immediately after a 90° pulse, when *xy* and *x'y'* coincide and
b) at time TE/2 later observed in the *x'y'*-frame rotating with the Larmor frequency

However, there is a means to get around this by utilizing a pulse train of one or more 180-degree pulses after the initial exciting 90-degree pulse. These 180-degree pulses cause every magnetic vector to turn around independent of its actual phase (see figures 18 and 19, where *x'* and *y'* are axes in a coordinate frame rotating around the *z*-axes with the medium Larmor frequency).

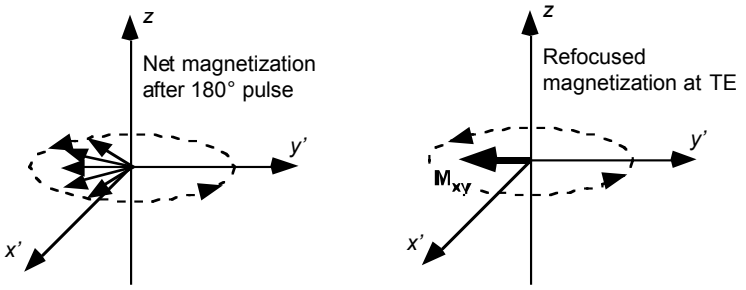


Figure 19 a) The net magnetization in figure 18 b immediately after a 180° pulse and
b) after refocusing along the *y'*-axes

However, as every spin vector still has its own magnetic environment, neither its precession speed nor direction will change. This means that the faster precessing vectors now are behind the slower ones and will eventually catch up with them. The vectors will rephase at time TE/2 after the 180-degree pulse, which is the same as time TE after the initial 90-degree pulse (see figure 19 b). This will happen simultaneously for all vectors, resulting in a strong signal in the receiving coil, situated along the *y*-axis or somewhere in the *xy*-plane. The signal is called a **spin-echo**, **SE** or a conventional spin echo, **CSE**[28].

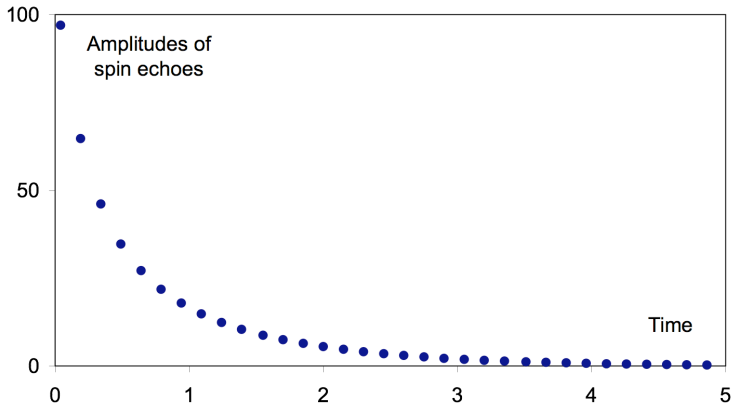


Figure 20 A chain of spin echoes created by a cpmg pulse train

The 180-degree pulse may be repeated many times generating new echoes with decaying amplitudes, which will mimic the true T2 (see figure 20).

The so-called **cpmg**-train of RF-pulses[29] should be: 90° – TE/2 – 180° – TE – 180° – TE – 180° – etc, where **TE** is called the **echo time**. TE is both the time between successive 180-degree pulses and from one echo to the next one, which appear half-ways between the *RF*-pulses. TE must always be very much less than T2 (see figure 21).

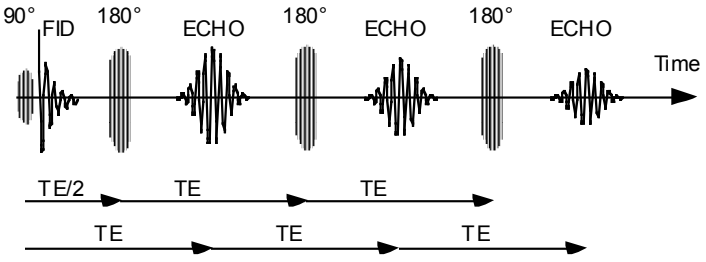


Figure 21 Sequence of *RF*-pulses,  , for Spin Echo and the resulting FID and echoes,  .

4. Constructing the MR image

4.1 Creating contrast

As can be seen from the discussion above, the spin magnetization must have a net *y*-component to give rise to a signal in a receiving coil and subsequently to an image of the studied object. The simplest experiment is to send a 90-degree pulse into the object, while the *z*-gradient of the magnetic field is active, and then immediately read the FID. This is called **saturation recovery**, **SR**[30]. As explained earlier, this has to be done with phase and frequency decoding line by line in the slice in order to calculate unique signals from every voxel within the selected slice. For every line (i.e. for every phase encoding gradient) a new 90-degree excitation pulse must be fired. Before doing so one must wait three to five times the longest T1 in the sample. This is the **repetition time**, **TR**. Then another slice is selected and so on. In fact new slices may be excited while waiting for the *z*-magnetization in one slice to recover, thus shortening the total acquisition time. To increase the signal-to-noise ratio, **SNR**, the whole

experiment can be repeated an arbitrary number of times. The SNR will increase with the square root of this number, called the number of excitations, **NEX**.

Immediately after the excitation pulse the first part of the FID is usually much disturbed and not very suitable for quantitative measurements. Therefore it is common practice to add a 180-degree inversion pulse shortly after the excitation and to read the signal at the spin echo as described in **3.4**.

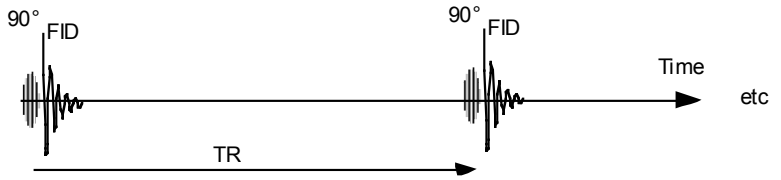


Figure 22 Schematic *RF*-pulse sequence for Saturation Recovery MRI.

An MR image created with a long repetition time, TR > 2 s, and a short echo time, TE < 25 ms, is said to be **spin density weighted** or **proton density weighted**. It has good SNR but usually a rather low contrast. However, this can be enhanced in various ways with the aid of specially designed pulse sequences. The most common of these have already been mentioned; **inversion recovery**, **spin echo** and **saturation recovery**. The spin echo sequence uses the fact that tissues have various transverse relaxation times, T2. If the read-out is delayed, the signal from some tissues may have died out, resulting in dark areas, whilst others still have significant amplitude, which will show up as bright areas in the image. Such an image, created with a long TR > 2 s and a long TE > 80 ms, is said to be **T2-weighted**. Here fat appears dark and water bright. The opposite is valid for a **T1-weighted** image. In this case one uses a short TR < 200 ms, and a short TE < 25 ms, allowing only protons with enough short T1 get time to recover their z-magnetization. Protons with long T1 will be continuously saturated. They have very small z-magnetization to flip into the xy-plane by the excitation *RF*-pulses and thus they give no signal or a very weak one in the image. The short TE will decrease the influence of T2 on the image. Also the inversion recovery pulse sequence can be used for specific weighting the image. With a short inversion time, TI ≈ 150 ms, the signal from fat will be very small and with long TI ≈ 1–2 s, the signal from body fluids will almost disappear. In this way one can choose the signals to be attenuated. These pulse sequences are called **short time inversion recovery**,

STIR[31], and **fluid attenuated inversion recovery**, **FLAIR**[32], respectively, the latter being especially useful in brain imaging, where CSF will be dark.

In practical work all these weightings are present more or less at the same time. However, with carefully chosen values of TR, TE, and TI, where applicable, it is possible to enhance the desired weighting and to achieve the contrast needed for a certain investigation. Usually this demands a lot of trial and error, providing the experience needed to choose the optimum parameters. In most cases the experiment starts with acquiring a low-resolution spin density image. With the aid of this the **field of view**, **FOV**, is defined and a number of T1- and T2-weighted scans are performed with the resolution desired. If needed, more scans can be done with other pulse sequences depending on the goal of the experiment.

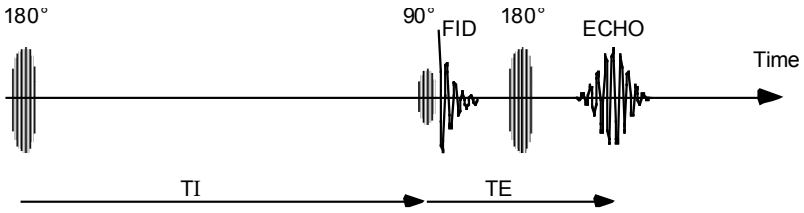


Figure 23 Schematic *RF*-pulse sequence for Inversion Recovery MRI

4.2 Fast Acquisition MRI Methods

It takes a rather long time to create an image with any of the pulse sequences discussed above. For 128 slices and 128 phase encoding gradients the required time will be $128^2 \approx 16{,}000$ times TR. Even for a rather short TR ≈ 1 s, such a scanning cycle will take about four and a half hours! This is, of course, a prohibitively long time for clinical imaging. Thus, many methods have been invented to shorten the time needed, the most obvious being to reduce the number of data-points. Instead of 128 slices, only 16 or 32 may be needed. This will shorten the acquisition time accordingly. However, for the same volume studied, a general reduction of the number of data-points will reduce the resolution of the image.

Another method to shorten the acquisition time, **multi-slice imaging**, has been briefly mentioned in the previous section. It relies on the fact that, after having acquired the signals originating from one phase decoding gradient, one

must wait for several T1s until a new excitation pulse and a new read-out is possible. This may take several seconds. However, during the waiting time it is possible to excite and read several more slices, usually 16 or 32, parallel to the first one. As these slices experience other magnetic fields than the first one, the frequencies of the excitation pulses must be adjusted according to equation (4). As B_0 is a continuously varying entity and as short *RF*-pulses contain a narrow band of frequencies, there will always be a minor overlap between adjacent slices. Therefore, it is most common to firstly measure every other slice and then to go back and read the omitted ones, if needed. Thus one can drastically reduce the scanning time with more than a magnitude down to about half an hour. Nevertheless it will still be long.

Another way to use the "dead-time" is to utilize not only the first but several of the spin-echoes that a cpmg pulse train gives rise to. This appeared as **Rapid Acquisition** with **Relaxation Enhancement**, **RARE**[33], but is nowadays commonly called **turbo spin echo**, **TSE**, or **fast spin echo**, **FSE**[34]. Here the number of used echoes is called the turbo factor, which can be eight or sixteen or even more. The acquisition time will be reduced with the turbo factor. The TSE-method may be used in combination with multi-slice imaging. However, as one uses a greater part of the repetition time for the read-out, fewer slices may be addressed. Thus the factors for time reduction are not multiplicative, but scan times can be decreased to ten minutes or less.

The TSE-method gives more or less T2-weighted images. However, the later echoes will be more T2-weighted than the earlier, which are spin density weighted, resulting in a non-equal contrast and also a varying signal-to-noise ratio. With a limited number of echoes it is not usually a major problem. Instead the signals can be used for producing both a T2-weighted image (with an effective TE a few times longer than the one applied) and a spin density weighted one in the same scan.

A third way to reduce the acquisition time is to make the echo appear as early as possible. Instead of using an inverting 180-degree pulse, on can deliberately speed up the transverse dephasing by the aid of a magnetic gradient followed by a reversed gradient, which can be the readout *x*-gradient. It will rephase the spins, again resulting in an echo. This is the **gradient refocused echo**, **GRE**[35] (or GRASE), which makes the echoes appear much faster than a 180-degree *RF*-pulse. However, this way to rephase the nuclear spin vectors does not

compensate for magnetic inhomogeneities or chemical shifts. Thus the amplitudes of the echoes will depend on T2* instead of on T2 as in the CSE case.

To speed up the MRI scan even more TR should be decreased as much as possible. This can be done, but at the expense of less recovery of the z-magnetization after the next excitation pulse for tissues with long T1 and accordingly to a smaller *xy*-magnetization. This is the same as a strong T1-weighting of the image. If this is not desired, one can decrease the duration of the excitation pulse making the flip angle, α, less than 90 degrees, maybe 10 degrees or even smaller. The *xy*-magnetization will be less than with a 90-degree pulse, resulting in a worse signal-to-noise ratio, but the relaxation time will also decrease and the acquisition time likewise. This is called **fast low angle shot**, **FLASH**[36]. Here the first excitation pulses will create a non-stable *xy*-magnetization, but soon a steady state is established, with constant magnetization, which can be used for reading the signal. For every chosen TR there is a flip angle, the so-called Ernst angle, α_E, which gives the maximum magnetization in the *xy*-plane and thus the highest signal amplitude according to equation (9). The shorter the TR, the smaller the flip angle should be.

$$\cos \alpha_E = \exp(-TR/T1) \qquad (9)$$

For really short TRs in the order of T2 it may be necessary to destroy (spoil or crush) the remaining *xy*-magnetization with uncontrolled phase before the next excitation pulse is fired[37]. This can be achieved with a non-polarized *RF*-pulse or with an additional magnetic gradient in the z-direction. However, with very short TR<T2*, a steady state magnetization may be achieved both in the z- and *xy*-planes. To achieve this, a "rewinding" gradient has to be applied in the phase-encoding direction after read-out. Such pulse trains are called **fast imaging** with **steady state precession**, **FISP**[38] or only **steady-state free procession**, **SSFP**[39].

Depending on how and when the dephasing and rephasing (reading) gradients are used within these pulse sequences images with either predominantly T1- or T2- or mixed weighting can be produced.

The magnitude of the flip angle is determined by the strength of the RF-pulse according to equation (10). Both the duration, τ, and the amplitude can be varied. The amplitude is related to the magnetic field, B_1, of the electromagnetic

wave of the *RF*-pulse.

$$\alpha \text{ (radians)} = 2\pi \bullet \gamma \bullet \tau \bullet B_1 \qquad (10)$$

As before γ is the Larmor frequency, 42.58 MHz. Usually it is desirable to make τ short, 10 – 100 microseconds. For a 90-degree pulse (= $\pi/2$ radians) this means a magnetic field strength of B_1 = 0.6–6 Gauss (60–600 μT) and a pulse power of several kilowatts.

For a pulse with longer duration the magnetic field strength should be lower according to equation (10). This will be the case, when a very narrow frequency band is desired. In such an experiment the pulse length should be 10 – 100 milliseconds as described in **3.2**.

4.3 Receiver Coils and Parallel Imaging

Some of the most important components in MRI experiments are the receiver coils. In principle a receiver coil is only a wire-loop which can be built into the equipment together with the *RF*-coils. These are usually used in pairs to cover the whole FOV. However, in most cases separate coils are used because specifications for transmitter and receiver coils are not the same. The receiver coils should be shaped to fit the anatomy studied and be as small as possible. It is very important that the very best coil is chosen for every single experiment. Thus different coils are designed for different studies and a great number of varieties exist as different solenoids, various surface coils, saddle coils, birdcage coils, mouse coils etc. For enhanced sensitivity they should be put as close to the studied object as possible. However, the coils must never touch the patient as they may be warmed up by the *RF*-energy.

There is also a possibility to use several receiver coils and even a phased array of coils. This is called **phased parallel imaging**, **PPI**[40]. In this way a large number of signals from the studied object can be recorded simultaneously, either to shorten the acquisition time, as the number of phase encoding gradients can be reduced, or to create a larger FOV, e.g. the whole spine. In combination with echo planar imaging (see next section) it is also possible to greatly enhance the SNR. However, the technique is not without its problems and several acquisition methods have been developed. The first human PPI imaging took place 1997 with a technique called **sensitivity encoding**, **SENSE**[41] but others soon followed[42].

4.4 Echo Planar Imaging

The ultimate way to do a really fast MRI scan would be to complete all read-outs and produce the whole image after one single *RF*-excitation, i.e. during about a tenth of a second or less. This is called single-shot **echo planar imaging**, **EPI** (SS-EPI), and a way to perform this was in fact published very early during the history of MRI[43]. Here no magnetic gradient is used during the excitation pulse. Thus protons in the whole body situated in the homogeneous magnetic field are excited. During read-out the x-gradient is switched between opposite directions, while phase encoding gradients are on for short moments during every read-out period in both the y- and z-directions. In this way the xy-magnetization is continuously dephased and rephased giving rise to a large number of echoes, which are used for creating the image. Thus the whole imaging can be acquired in about 0.1 s, though not without trade-offs. This means that read-out of some 128^3 ($\approx$ 2 million) voxels have to be done during 128^2 ($\approx$ 16,000) readout periods, while the gradients have to be switched on and off incredibly rapidly. The ramp time must be less than 0.1 ms, for which an electric effect of about 100 kW is needed! The rapidly changing fields can also give rise to nerve stimulation in the examined patient[44]. However, it is impossible to make gradients as square formed as desired. Special, very effective sampling schemes must also be utilized, e.g. forth and back or spirally with simultaneous x- and y-gradients 90 degrees out of phase etc. These must take into account that also parts of the body outside the FOV may give rise to signals. They may result in ghosting and blurring of the image. Accordingly, the demands on the hard- and software are very high and for a long time no instruments could meet the specifications. The first successful EPI imaging of a moving object in real time was not done until 1986[45]. Even now, not all instruments are designed to perform single-shot EPI.

A less demanding procedure is multi-shot EPI, where all information from one slice is achieved after a single shot, i.e. during 0.1 s or less and a three dimensional image in a few seconds. For ordinary MRI-imaging multi-shot EPI or less fast procedures are usually sufficient. However, for fMRI experiments and also for angiography, diffusion studies, and other imaging in real time, signal acquisition must be very fast and here only single-shot EPI will do.

Thus the EPI technique – especially single-shot EPI – is very useful but not without its problems. Resolution is limited and the weighting will usually not be

uniform during the whole acquisition, which can give rise to uneven contrast and images that are difficult to interpret. Usually T2-weighting dominates in EPI, but even T1-weighting can be achieved by putting additional *RF*-pulses before the excitation pulse. This will saturate signals from desired tissues. Because of steep and non-square magnetic gradients, about 2,5 Gauss/cm (25 mT/m) and limited sampling time, other artifacts as aliasing (ghosting) and geometrical displacement can appear in the image. Especially fat tissue may be misplaced (a few cm) because of its different chemical shift compared to water. Therefore it is common to suppress the fat signal by using a tuned *RF*-signal (see section **3.2**) or in some other way

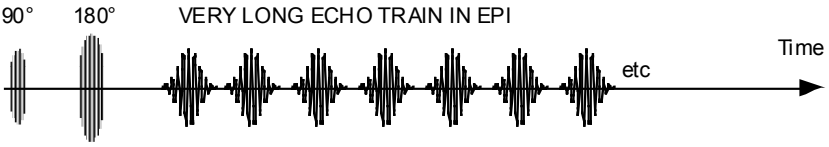


Figure 24 Gradient echo train created by continuously switched read-out gradients, while phase-encoding gradients are active shortly within each read-out cycle

4.5 Magnetic Resonance Angiography

A common artifact in MRI is caused by motion. As imaging may take many seconds or even minutes, it happens that the object is not totally immobilized during the acquisition time. Especially breathing and cardiac motion are difficult to control. This may give rise to ghost images and to misinterpretation of the spatial information[46]. Therefore imaging is usually performed during a breath-hold if possible or synchronized, gated with the heartbeats or pulse flow[47]. Nevertheless, patient movement is often a problem in MRI and only very fast acquiring techniques as described in **4.1** and **4.3** can solve this. Blood flow is a special case, where venous blood shows up bright in T2 weighted spin echo examinations, while rapidly flowing arterial blood appears dark. Oxygen content is also important as discussed in **4.6**.

However, the sensitivity of MRI to movement can also be turned to advantage when studying moving objects like blood flow. This is **magnetic resonance angiography**, **MRA**[48]. The proton spins in an object, moving through the field of view along the slice selective gradient, will continuously change their

precession speeds, when the gradient is on. Thus they are out of phase with stationary objects within the same voxel. This can be used for identifying moving objects. In fact, gradient NMR was used for the study of diffusion and blood flow very long before the invention of MRI[28,49].

One angiographic method utilizes the fact that blood which flows into an excited slice is unaffected by the excitation pulse. A new excitation pulse shortly after the first one will show a bright signal from the blood, whilst the background is relatively dark because of saturation. This is the method of **time of flight**, **TOF**. Here the slice thickness, the presaturation of the slice, and the tip angle should be optimized in a way that the inflowing blood has almost filled the slice, whilst the background tissue is still almost saturated. It is also possible to carry out the experiment the other way round, i.e. to saturate protons on either side of the chosen slice beforehand. Some of these protons will flow into the slice, where they are unaffected by the *RF*-pulse and will be dark against a brighter background. In this way flow-related artifacts can be drastically reduced.

Another way is to make two or three measurements of the same area with a double gradient in the direction of the flow. Half of the magnetic gradient should be in the direction of the flow and the other half in the opposite direction. Thus the precession of the spins will first speed up and then slow down again to their normal velocity, leaving stationary spins unaffected. Protons in moving blood, however, will change phase and can be detected. This is the method of **phase contrast** angiography, **PC**.

Even more ways to study movement have been developed and this is a rapidly growing area of research, where very fast acquisition techniques are used making it possible to study flow and moving objects in real time.

4.6 Contrast agents

In *X*-ray imaging it is very common to use contrast agents, which directly interfere with and attenuate the radiation. They will in this way enhance the contrast of the image. The contrast agents used in MRI work in a quite different way[50]. They do not at all interfere with the *RF*-pulses or directly with the FID signal. Instead they shorten the relaxation times, T1 and T2, in the

tissues, which they have entered. There they change the local magnetic environments of the protons, whose relaxation parameters also change accordingly. This can be turned to advantage in two ways.

Firstly, shortening the longitudinal relaxation time makes it possible to reduce the time between consecutive exciting *RF*-pulses. This will also reduce the time for the overall experiment, which is usually a goal in itself. Thus a decrease in T1 is of more use than a decrease in T2.

Secondly, if a contrast agent unevenly enters adjacent tissues, as is often the case for tumors, they will appear differently on an image, which may enhance the contrast. This is especially helpful, when the blood-brain-barrier is broken by malign tumors.

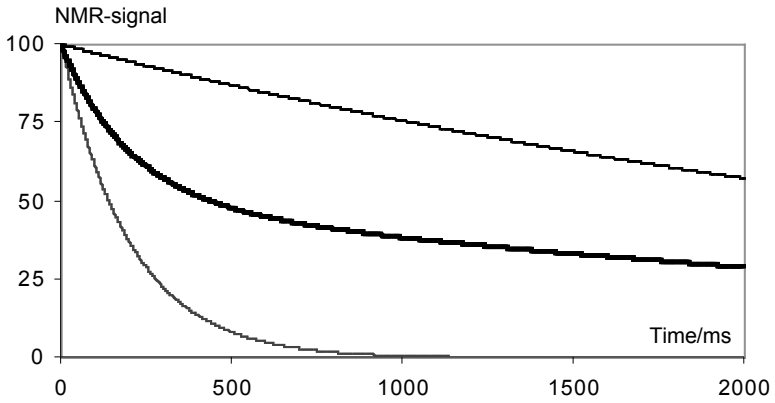
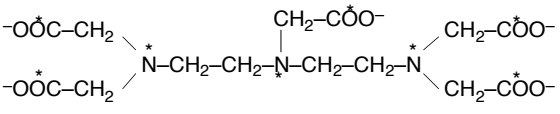


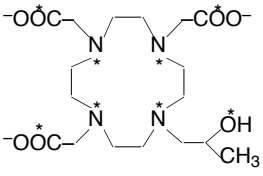
Figure 25. Simultaneous acquisition of the FID envelope from pure water and water containing 5 mM of copper sulphate in the middle (thick line) with signal from pure water above and the solution below.

Both T1 and T2 are affected. A simple example of the latter is water, for which T2 can decrease more than tenfold, from 3600 to 200 ms, when a very small amount of paramagnetic copper ions, in this case 5 mM is added (see figure 25), where the FIDs from pure and doped water are overlapped. This also means that tissues containing much water have longer relaxation times than more compact tissues. Thus, increasing the water content can both increase the relaxation time and increase the spin density, i.e. the number of protons.

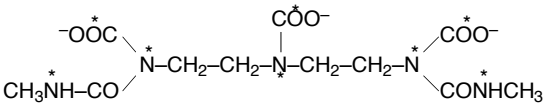
Contrast agents in MRI are paramagnetic or ferromagnetic substances containing unpaired electrons. The most obvious of these are oxygen and iron. Both appear in blood and it has long been known that the magnetic susceptibility varies between diamagnetic oxygenated and the paramagnetic deoxygenated blood[51]. An effect of this is that protons in arterial blood show longer relaxation times than those in venous blood[52], which forms the basis for **blood oxygen level dependent contrast**, **BOLD**[53]. Hemoglobin will also change its influence on relaxation times if it is oxidized to methemoglobin. Here the heme iron has been transformed from its ferrous, Fe^{2+}, to its ferric form, Fe^{3+}, changing the number of unpaired electrons from four to five.



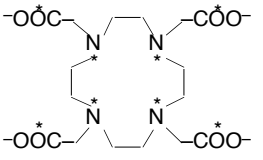
diethylene triamine penta acetic acid
DTPA



10-(2-hydroxypropyl)-
1,4,7,10-tetraaza-cyclo-
dodecane-1,4,7-triacetic
acid, HP-DO3A



diethylene triamine pentaacetic acid bis-methylamide,
DTP-BMA



1,4,7,10-tetraaza-cyclo-
dodecane-1,4,7,10-tetra-
acetic acid, DOTA

Figure 26 Chelating agents for complex formation with Gadolinium. They are sold under the commercial names: Magnevist®, ProHance®, Omniscan®, and Dotarem® respectively. The asterisks indicate electron excess points, where bonding between the metal and the complex agent occur.

However, most MRI contrast agents now in routine use, contain compounds with rare earth metals, usually gadolinium, Gd^{3+}, which has seven unpaired electrons and primarily affects T1. These agents have been allowed in clinical use since 1988 at a concentration of about 0.1 mmol/kg. The metal ion is in a complex with DTPA (diethylene triamine penta acetic acid) or some other

chelating agent. Here the metal is very tightly bound, more or less locked into a cage, and it is therefore almost non-toxic despite the fact that gadolinium is a heavy metal and thus toxic in itself[54] (see figure 26, where the binding sites are marked with asterisks).

Usually the contrast agent has left the body after a few hours. The half-life is less than half an hour[54]. Other examples of MRI contrast agents are ferric ammonium citrate, ferric porphyrines[55], micro particles of iron oxide[56], manganese compounds, and stable free radicals, usually nitroxide radicals. The latter are rather toxic and can be used only in very low concentrations.

5 Imaging with hyperpolarized gases

It is almost impossible to image the lungs as they contain air and carbon dioxide, which have a very low concentration of protons. Even filling the lungs with hydrogen gas would give a proton concentration of less than one per mille compared to tissue, besides being an extremely dangerous experiment. Nevertheless, imaging of lungs was done rather early in the history of MRI[15]. In this case gases of perfluorinated hydrocarbons were used, as the sensitivity to fluorine is almost the same as to hydrogen. However, the resolution was poor.

A quite different way is to have the patient inhale a **hyperpolarized inert gas**, **HPG**. This is usually a rare isotope of helium, He-3[57], or xenon, Xe-129[58], which have been given a very large NMV. This is now an established technique[59]. During imaging the *RF*-emitter and receiver coils must be tuned to the Larmor frequency of the inert gas instead of hydrogen (see table 1). Instead of having only 6-7 ppm excess magnetization for protons in a magnetic field, it is possible to transfer almost all atoms of He-3 or Xe-129 into one spin state, thus increasing the sensitivity about one million times. Because both helium and xenon are inert gases with filled electron shells, they do not form compounds and are not bound to any lattices. Thus the longitudinal relaxation time, T1, in the free gas is several hours for Xenon and somewhat shorter for Helium. Because of this the hyperpolarization can and must be carried out beforehand. This is performed indirectly with the so-called optical pumping of rubidium (or potassium) atoms in gaseous state with a powerful laser (795 nm) at 150–200° C (see figure 27). The valence electrons of the alkali metals are polarized and this polarization is transferred to the nuclei of the inert gases via

atomic collisions[57]. This will take a few hours. The method is somewhat different for He-3 and Xe-129.

When imaging the lungs, the patient must inhale hyperpolarized gas, which can then be excited by an *RF*-pulse in the same way as in ordinary imaging. However, this can only be done once, as the z-magnetization is not restored. Instead the gas successively relaxes to a normal equilibrium condition and loses its hyperpolarization. This process is much faster in tissue than in the isolated gas, but is still often in the order of several seconds. If more excitations are to be done a new batch of gas must be inhaled.

Xenon is easily dissolved in body fluids such as blood, why it can be used for imaging not only the lungs. In this case more ways to administer the hyperpolarized gas may be used.

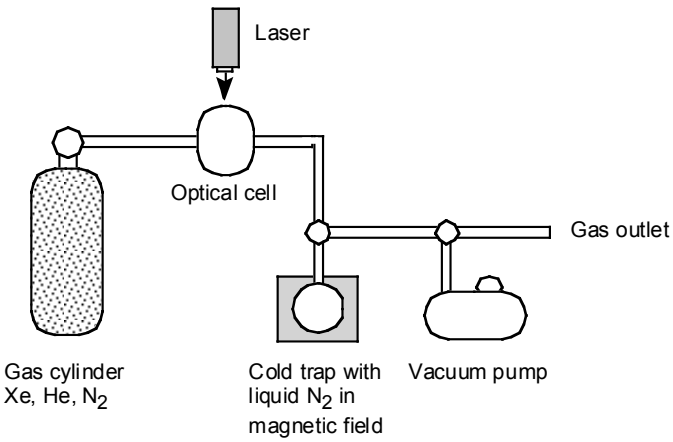


Figure 27 Schematic drawing of device for making hyperpolarized xenon gas for MRI

6. Use of MRI

MRI has found more and more applications within clinical diagnosis although early ideas that NMR relaxation would be a convenient way of detecting vascular tumors have not proved true. Nevertheless, the use of MRI for tumor diagnosis is important. However, this is more dependent on that one with MRI can create a useful contrast between various kinds of tissue, even tumors, than on a simple difference in relaxation time. For designed areas the resolution can be increased very much and it is even possible to image a single cell[60].

From the beginning imaging the brain was an important use as MRI is not hampered by the skull and thus gives a superb contrast. Also today brain imaging at suspected stroke or multiple sclerosis is a very common diagnosis task for MRI, including fMRI. However, computed tomography may sometimes be the preferred tool for head injuries as it is faster than MRI.

Imaging the spine and joints is also an area where MRI usually gives a better contrast than *X*-rays, again depending on the fact that bone does not obscure the soft tissue. In general MRI has a much better contrast than *X*-rays, when soft tissues are imaged. Thus it is possibly the case that MRI will take over an increasing number of such diagnosis tasks that are done with *X*-rays today. However, imaging with *X*-rays is still faster and cheaper and there are also situations, where imaging by MRI is impossible (see Chapter **7**).

With echo-planar MRI imaging can be done in a fraction of a second and this has recently opened the way for imaging in real time. Images may be hard to interpret though, as the contrast can have a mixture of T1- and T2-weighting. There is also a trade-off in resolution. Anyway, it is now possible to image a beating heart[45] and detect arterial sclerosis, to carry out functional MRI, and to study the metabolism of drugs and other ingested compounds. Not least for neuroscience this is a valuable tool in for instance, the treatment of patients recovering from stroke or for schizophrenia. The perspectives seem more or less unlimited and as development continues many more applications will be possible.

7. Safety and Risks

7.1 Safety in MRI imaging

MRI is in most cases assumed to be a very safe diagnostic modality. It is not invasive, contrast agents can often be avoided and it does not use ionizing radiation. When used properly no harmful effects have become apparent. However, there are drawbacks and risks[61]. The strong magnetic field has probably no impact on the human body[12], but many metallic implants and pacemakers are not compatible with MRI and deaths have occurred[62,63]. Ferromagnetic objects of any kind, within or outside the patient's body (beds, wheel chairs, gas cylinders, fire extinguishers, tools, etc) are of course strictly

forbidden. "**The magnet is always on!**" They can act as deadly projectiles and if they happen to be pulled into the magnetic bore, they cannot be taken away without quenching the magnet.

Objects of iron, cobalt, nickel and steel should always be considered ferromagnetic, which can easily be tested with a small permanent magnet. However, other metallic objects close to the imaged site can also disturb the homogeneity of the magnetic field and hamper the interpretation of the image. Many metals such as aluminum, copper, gold, lead, mercury, silver, tin, and zinc are safe, but chromium, platinum, vanadium, and even non-ferro-magnetic stainless steel should be avoided and also titanium if possible[64].

A few physiological effects have been noticed during MRI examinations: 1) a slight increase of body temperature, a few tenths of degrees, caused by the *RF*-pulses. This is a non-resonance effect and cannot be used to selectively heat a desired tissue, 2) nerve stimulation in limbs, when the magnetic fields are switched rapidly in EPI, 3) the switching gives also rise to quite a lot of noise, from which the patient must be protected, and noise-damping systems must be built in from the beginning. FDA has also limited a number of variables mentioned above as maximum field to 8 Tesla, maximum ramp speed to 3 T/s, maximum specific absorption rate to 4 W/kg, and maximum noise to 99 dB[12].

7.2 Risks of MRI

The risks of MRI can be summarized in the following way:

1) External Projectile Effects
2) Internal Projectile Effects
3) Other magnetic Effects
4) Radiofrequency Energy
5) Gradient field changes
6) Acoustic Noise
7) Quenching of the Magnetic Field
8) Fringe fields
9) Claustrophobia

7.2.1 *External Projectile Effects*

It is absolutely necessary to avoid ferromagnetic objects close to the MRI machine. Routines must be established to ensure no such objects will ever be in the MRI room. Nevertheless, a number of deaths have occurred[62], when various things have been pulled into the magnet bore and hit the patient with great power. As "the magnet is always on", it is just as important to avoid non-safe metallic objects in its vicinity, when the magnet is empty. If a wrench, a pair of scissors or another tool has happened to get into the magnet, it cannot easily be taken away unless the magnet field is quenched. This will take several hours or more.

7.2.2 *Internal Projectile Effects*

Various prostheses, implanted medical devices, or other metallic objects within the body may prohibit MRI examinations[65]. Tooth fillings and eye lenses are OK and also several kinds of prostheses, but this always has to be checked beforehand. The same is valid also for piercing and sometimes even for tattoos, which often contain iron oxide (see also below). Metallic fragments, which by accident may have entered a body, can cause serious harm. A couple of cases of blindness have occurred in this way[66].

7.2.3 *Other magnetic effects*

Even if an implant is not ferromagnetic and moved by the magnetic field, the field may cause it to stop working. People with pace makers can usually not be imaged – a few deaths have occurred[63] – and other medical devices such as insulin pumps may also be prohibitive. Non-ferromagnetic stuff may be allowed outside the magnet bore. However, both paramagnetic and diamagnetic objects may slightly disturb the homogeneous magnetic field within the field of view, which will result in a poor resolution of the image.

A number of patients have complained about vertigo, nausea, or metallic taste, especially at high fields. Some have even seen flashing lights. In addition, it has been observed that blood cells can change shape during an MRI examination. It is not known, whether anything of this can cause any harm, but it is always important to listen to what patients say.

7.2.4 *Radiofrequency Energy*

The *RF*-pulses have high power, but as they last only for micro- or milliseconds, they should generally not transfer much energy to the patient. A maximum increase in body temperature of one degree is usually accepted, but only a few tenths of a degree is more common. However, wire loops can act as *RF*-antennas and result in local overheating and even burns. ECG and other cables cause the most common problems, but also necklaces, earrings, and various kinds of piercing. Even tattoos can act as sites for enhanced RF-absorption and burns.

7.2.5 *Gradient field change*

Especially with echo planar imaging these magnetic gradients can be very steep and stimulate peripheral nerves. The effect is usually only irritating, but it can even cause cramp and be painful. However, so far there is no evidence of any cardiac stimulation through gradient field changes.

7.2.6 *Acoustic noise*

This is something, which is connected with changes of the gradient field, as these will cause the magnet to vibrate. This vibration is the source of the very loud noise, up to 100 dB and sometimes even more, which MRI scanners emit. Therefore patients should always wear earplugs, in order not to harm the ears, even if they are sedated. Modern MRI equipment is more firmly built, which results in lower noise, but it seems impossible to lower it far enough for a comfortable examination.

7.2.7 *Quenching the magnetic field*

If the MRI scanner runs out of liquid helium, the magnet coils will heat up and lose their superconductivity. The coil current and the magnetic field will rapidly decrease and release much energy, which probably will damage the scanner. At the raised temperature the liquid coolants, helium and nitrogen will boil off and fill the scanner room, displacing the air. This is of course life threatening to somebody in the room. At the same time evacuation can be hampered by condensation of moisture, caused by the cool nitrogen gas.

A quenching is a most unlikely event, which might be necessary to initiate voluntarily if an object pulled into the magnet has hurt a patient.

7.2.8 Fringe fields

Depending on the strength of the magnet there is always a rapidly decreasing fringe field around the magnet. Despite attempts to shield the magnet the field can extend several meters in some directions, which apart from the points mentioned may also erase information on credit cards. Therefore the area in the vicinity of the magnet, where the field may exceed 0.5 mT, must have restricted access, as 0.5 mT is considered safe for pacemakers[67]. However, the extension of the fringe field varies very much between different types of magnets. Superconducting magnets have the largest fringe fields, whilst they are very small for permanent magnets (see also Chapter **8**).

7.2.9 Claustrophobia

This is a very common problem. For some 10 % of the patients it is such a great problem that they cannot go through an MRI examination. Small children and animals should always be sedated to avoid similar problems.

8. Equipment for Magnetic Resonance Imaging

With few exceptions the manufacturers of hardware for MRI have limited background in NMR. Most of them are manufacturers of various kinds of medical equipment, such as *X*-ray machines and CAT, and some of them are electronic companies. As an MRI machine may cost a few million dollars this is a multi billion industry with many vendors looking for profit.

8.1 Resistive Magnets

The earliest MR images were produced with the aid of magnets with resistive coils. However, even very high power instruments cannot produce stronger fields than a few tenths of a Tesla. To get superior resolution higher fields are needed and even more stable fields than can easily be achieved. Resistive coils also consume much electric power and need extensive cooling. Therefore resistive magnets are not much used nowadays.

8.2 Superconducting Magnets

Today these are the most common magnets on the market. In the shape of NMR magnetic fields of up to about 20 T are used, but in MRI the limit is usually 2–4 T. Instruments at 1.5 or 3 T, with a superior field stability and homogeneity are among the most common. These instruments need liquid helium at 4 K for the coils of commonly Nb-Ti alloy to be brought down to superconducting condition. This is expensive, but new electronic cooling systems have decreased the demand for helium and today superconducting coils are the most economic ones for high fields and the only ones available for very high fields. These instruments are usually very large because of the need for much insulation to keep the cryogen at low temperature. Another drawback is the large fringe field area, which can extend several meters in many directions. However, this can be partly overcome by passive or active shielding of the magnet[68]. For passive shielding the NMR room should have (heavy) iron plates in walls, ceiling and floor. In active shielding an additional coil carrying a counter current is situated outside the main coils minimizing the magnetic fields outside the bore.

8.3 Permanent Magnets

For special tasks very heavy and expensive permanent magnets of iron, alnico alloys, or rare earth metals can be used. However, they are not common. These magnets usually have an open design with a vertical magnetic field. Therefore, fairly large objects may fit. However, the maximum field is low – 0.2–0.5 T (although increasing) – and this is not always sufficient.

8.4 Design

Every MRI-magnet should have an extremely homogenous magnetic field. Even the best magnets, however, often have small differences in field strength between the centre and the periphery. Along the field lines the variations are less. Thus it is important to put the imaged object close to the center of the magnet and, when this is not possible, make the geometry of the object lie parallel to the field lines.

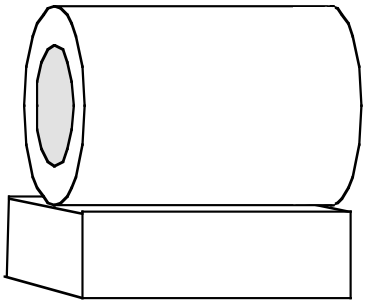


Figure 28 Simple MRI magnet design with horizontal bore and magnetic field

Most magnets are constructed as a horizontal tube into which the object to be imaged must fit. The diameter of the bore is usually about 60–80 cm, which limits the size of objects (patients) to be imaged. The effective length of the bore can be almost any size but it is usually limited to less than one meter. A magnet of this size may be used for imaging any part of a human being, as long as he/she is not too large, or an animal (see figure 28 above).

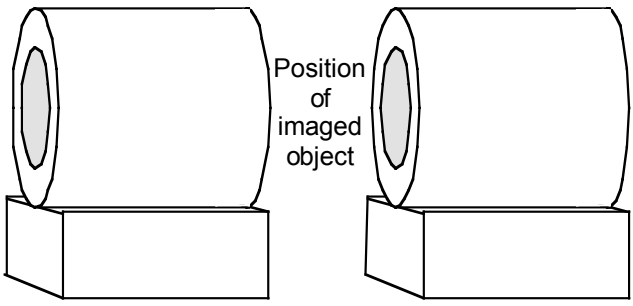


Figure 29 Sketch of two MRI magnets in series for imaging larger objects

To image larger people or animals other magnetic designs must be used. Widening the magnetic bore is possible, but at a cost, either in field strength, field homogeneity, economy or all. Another way is to put two horizontal magnets in a row at a desired distance, the north pole of one of them pointing towards the south pole of the other (see figure 29). Now it is possible to put even very large objects into the gap between them. These may be animals or sitting or standing people, who may suffer from claustrophobia if put into the narrow bore of a machine with conventional design. As it is not quite easy to construct systems of this kind with enough field homogeneity, poor resolution may be the result. However, when imaging large animals resolution is not always crucial. Anyway this design is expensive containing as it does two

synchronized magnets.

As mentioned in **8.3**, permanent magnets can also be built with an open design, usually with one magnetic pole in the form of a table and the other extending down from the ceiling (see figure 30). These kinds of design, which may look patient friendly, are only used for special purposes.

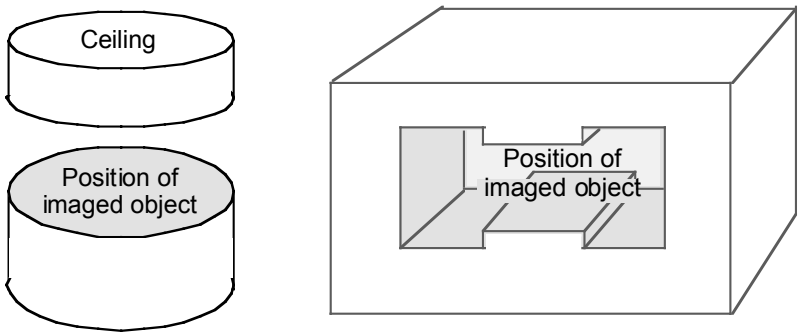


Figure 30 Principle design of open setups of permanent magnets with vertical magnetic field

8.5 RF- and detector coils

These have been partly dealt with in **4.3**. The *RF* coils and a basic pair of detector coils, the volume coils, are usually pre-built into the magnet design. However, in most cases specially designed detector coils are used depending on which organ is to be imaged. These should be put as close as possible to the imaged object but without direct contact with a patient. Flat coils should be positioned with their long axes parallel to the magnetic field. Phased array coils should also as a rule be positioned along the magnetic field.

8.6 Computers

Without computers no magnetic resonance imaging would have ever been possible. The development of MRI has very much paralleled the development of computers, both with respect to speed and memory capacity. The most obvious examples are decreased scan time and image manipulation. What was only theoretically possible, when echo planar imaging was suggested in 1976, is now routine. Every MRI machine needs a powerful computer both to perform the MRI scans with exact magnetic gradients, which have to switch both

direction and magnitude in microseconds, and various *RF*-pulse sequences and to receive and handle all the data during the scans. In addition the computer should be able to present the data in a useful graphic way. This means that powerful graphic software is needed for manipulating the images in three dimensions for best interpretation. Finally the need for the data storage quickly enters the Terabyte level. Thus a computer for use with an MRI scanner can hardly be too powerful.

9. Acknowledgements

This text has been written with the aid of a grant from The Natural Sciences and Engineering Research Council of Canada, which is greatly acknowledged. I also want to thank Mrs. Elaine Smith at Ontario Veterinary College and the librarians at University of Guelph for very much much practical help, Dr. H. Dobson and Dr. E. G. Janzen at University of Guelph, ON, Dr. P. Gareau at Robarts Research Institute, London, ON, Dr. R. Towner, Oklahoma Medical Research Foundation, OK, for much support and valuable discussions, and Dr. M. Cooper, University of Karlstad, Sweden for revising the English manuscript.

Figure 31 1.5 T MRI scanner during installation at Ontario Veterinary College, University of Guelph, Canada

10. References

[1] P. Lauterbur, "Image Formation by induced Local Interactions: Examples employing Nuclear Magnetic Resonance", *Nature*, **242**, 190-191 (1973)
P. Mansfield, P.K. Grannell, "NMR Diffraction in solids", *Journal of Physics C, Solid State Physics*, **6**, L422-26 (1973)
A. Kumar, D. Welti, R.R. Ernst, "NMR Fourier Zeugmatography", *Journal of Magnetic Resonance*, **18**, 69-83 (1975)

[2] P. Mansfield, A.A. Maudsley, "Medical Imaging by NMR", *British Journal of Radiology*, **50**, 188-94 (1977)

[3] J.R. Hansen, "Pulsed NMR study of water in muscle and brain tissue", *Biochimica et Biophysica Acta*, **230**, 482-486 (1971)
R.V. Damadian, "Tumor Detection by Nuclear Magnetic Resonance", *Science*, **1171**, 1151-53 (1971)

[4] P. Mansfield, I.L. Pykett, P.G. Morris, "Human whole body line-scan imaging by NMR", *British Journal of Radiology*, **51**, 921 (1978)
J.M.S. Hutchison, W.A. Edelstein, G Johnson, "A whole-body NMR imaging machine", *Journal of Physics E, Scientific Instruments*, **13**, 947-55 (1980)

[5] C.T.W. Moonen, P.C.M. van Zijl, J.A. Frank, D. Lebihan, E.D. Becker, "Functional magnetic resonance imaging in medicine and physiology", *Science*, **250**, 53-61 (1990)
J. Belliveau, D. Kennedy, R. McKinstry, B. Buchbinder, R. Weisskoff, M. Cohen, J. Vevea, T. Brady, B. Rosen, "Functional mapping of the human visual cortex by magnetic resonance imaging", *Science*, **254**, 716-719 (1991)

[6] I.I. Rabi, "Space quantization in a gyrating field", *Physical Review*, **51**, 652-54 (1937)
I.I. Rabi, J. Zacharias, R. Millman, P. Kusch, "A New Method of Measuring Magnetic Moments", *Physical Review*, **52**, 318 (1938)

[7] F. Bloch, W. W. Hansen, M. Packard, "Nuclear induction", *Physical Review*, **69**, 127 (1946)
E. Purcell, H. Torrey, and R. Pound, "Resonance absorption by nuclear magnetic moments in a solid", *Physical Review*, **69**, 37-38 (1946)

[8] M.R. Gabillard, "Résonance nucléaire mesuré du temps de relaxation T2 en présence d'une inhomogenéité de champ magnétique supérieur à la largeur de raie", *Comptes rendus hebdomadaires des séances de l'Académie des Sciences*, **232**, 1551 (1951)
M.R. Gabillard, "A steady state transient technique in nuclear resonance", *Physical Review*, **85**, 694 (1952)

[9] E. Odeblad, B.N. Bhar, G. Lindström, "Proton magnetic resonance of human red blood cells in heavy-water exchange experiments", *Archives of Biochemistry and Biophysics*, **63**, 221 (1956)

[10] J.B.J. Fourier, "On the Propagation of Heat in Solid Bodies", The Paris Institute 1807

R.R. Ernst, W. A. Andersson, "Application of Fourier Transform Spectroscopy to Magnetic Resonance", *Review of Scientific Instruments*, **37**, 93 (1966)

[11] C.G. Masi, "Lodestone technology comes to NMR imaging", *Diagnostic Imaging*, **5**, (1983)

[12] FDA, "Criteria for Significant Risk Investigations of Magnetic Resonance Diagnostic Devices", July 15, 2003, http://www.fda.gov/cdrh/ode/guidance/793.pdf

[13] *Handbook of Chemistry and Physics*, ed. D.R. Lide, 80th ed, The Chemical Rubber Co 1999

M. Winter, http://www.webelements.com/, The University of Sheffield, UK, 2003

D. Morris, http://bmrl.bmrf.uiuc.edu:8080/MRITable/, Biomedical Magnetic Resonance Laboratory, University of Illinois, 1997

[14] M. Brauer, L.M. Foley, M.J. Yau, and R.T. Thompson, "Dynamic response of livers within chronic ethanol-treated rats to physiological challenge studied by functional MRI and ^{31}P MRI", *Alcoholism: Clinical & Experimental Research*, **26**, 366 (2002)

[15] P.A. Rinck, S.B. Petersen, E. Heidelberger, V. Acuff, J. Reinders, M.L. Bernardo, L.K. Hedges, P.C., Lauterbur, "NMR Ventilation Imaging of the Lungs Using Perfluorinated Gases", *Magnetic Resonance in Medicine*, **1**, 237 (1984)

[16] A.N. Garroway, P.K. Grannell,, P. Mansfield, "Image formation in NMR by a selective irradiative process", *Journal of Physics C, Solid State Physics*, **7**, L457-62 (1973)

P.C. Lauterbur, "Magnetic Resonance Zeugmatography", *Pure and Applied Chemistry*, **40**, 149-57 (1974)

[17] P. Mansfield, A.A. Maudsley, "Medical imaging by NMR", *British Journal of Radiology*, **50**, 188-94 (1977)

[18] A. Kumar, D. Welti, R.R. Ernst, "Imaging of macroscopic objects by NMR Fourier Zeugmatography", *Naturwissenschaften*, **62**, 34 (1975)

[19] R.R. Ernst, W.A. Andersson, "Application of Fourier Transform Spectroscopy to Magnetic Resonance", *Review of Scientific Instruments*, **37**, 93 (1966)

[20] H. Nyquist, "Certain Factors Affecting Telegraph Speed", *Bell Systems Technology Journal.*, **3**, 324 (1924)
C.E. Shannon, "A mathematical theory of communication", *Bell Systems Technology Journal,* **27**, 379 (1948)

[21] G Brix, L R Schad and W J Lorenz, "Evaluation of proton density by magnetic resonance imaging: phantom experiments and analysis of multiple component proton transverse relaxation", *Physics in Medicine and Biology*, **35**, 53-66 (1990)
S. Clare, "Functional Magnetic Resonance Imaging, Principles and applications", *Thesis*, University of Nottingham, (1997)

[22] J. Szumowski, J. H. Simon, "Proton Chemical Shift Imaging" in D.D. Stark and W.G. Bradley, eds., *Magnetic Resonance Imaging*, 2nd ed, p 482 (479-521), Mosby Yearbook, St Louis, 1992
V. Kuperman, *Magnetic Resonance Imaging*, p 58, Academic Press, San Diego 2000
J.T. Bushberg, J.A. Seibert, E.M. Leidholt Jr, J.M. Boone, *The Essentials of Medical Imaging*, 2nd ed, p. 391, Lippincott Williams & Wilkins. Philadelphia 2002

[23] W.A. Edelstein, P.A. Bottomley, H.R. Hart *et al.*, "Signal, Noise, and Contrast in Nuclear Magnetic Resonance Imaging (NMR Imaging)", *Journal of Computer Assisted Tomography*, **7**, 391-401 (1983)

[24] A. Haase, J. Frahm, W. Hänicke, D. Matthaei, "1H NMR Chemical Shift Imaging", *Physics in Medicine and Biology*, **30**, 341-44 (1985)

[25] H.T. Edzes, R.T. Samulski, "Spin Diffusion and Cross Relaxation in the Proton NMR of Hydrated Collagen", *Nature*, **265**, 521 (1977)
R.N. Muller, M.J. Marsh, M.L. Bernardo, P.C. Lauterbur, "True 3-D imaging of limbs by NMR zeugmatography with off-resonance irradiation", *European Journal of Radiology*, **3**, 286-290 (1983)
S D. Wolf, R.S. Balaban, "Magnetization transfer contrast (MTC) and tissue water proton relaxation in vivo", *Magnetic Resonance in Medicine*, **10**, 135-44 (1989)

[26] W.G. Proctor and F.C. Yu, "The Dependence of a Nuclear Magnetic Resonance Frequency upon Chemical Compound", *Physical Review*, **77**, 717 (1950)

[27] N. Bloembergen, "Fine Structure of the Proton Magnetic Resonance in $CuSO_4 \bullet 5H_2O$", *Physical Review*, **75**, 1326 (1949)

[28] E.L. Hahn, "Spin Echoes", *Physical Review*, **80**, 580 (1950)

[29] H.Y. Carr, E. M. Purcell, "Effects of Diffusion on Free Precession in Nuclear Magnetic Resonance Experiments", *Physical Review*, **94**, 630 (1954)
S. Meiboom, D. Gill, "Modified Spin-echo Method for Measuring Nuclear Relaxation Times", *Review of Scientific Instruments*, **29**, 688 (1958)

[30] F.S. Buonanno, I.L. Pykett, T.J. Brady, *et al.*, "Clinical Applications of Nuclear magnetic Resonance (NMR)" *DM*, **21**, 1-83, (1983)

[31] A.J. Dwyer, J.A. Frank, V.J. Sank, J.W. Reinig, A.M. Hickey, and J.L. Doppman, "Short-T1 inversion-recovery pulse sequence: Analysis and initial experience in cancer imaging", *Radiology*, **168**, 827-36 (1988)

[32] J.N. Rydberg, C.A. Hammond, M.S. Grimm, B.J. Erickson, C.R. Jack, J. Hunston *et al.*, "Initial clinical experience in MR imaging of the brain with a fast fluid-attenuated inversion-recovery pulse sequence", *Radiology*, **193** 173-80 (1994)

[33] J. Hennig, A. Nauerth, H Friedburg, "RARE Imaging. A Fast Method for Clinical MR", *Magnetic Resonance in Medicine*, **3**, (1986)

[34] L.E. Crooks, D.A. Ortendahl, L. Kaufman, J. Hoenninger, M. Arakawa, J. Watts, C.R. Cannon, M. Brant-Zawadzki, P.L. Davis and A.R. Margulis, "Clinical efficiency of nuclear magnetic resonance imaging", *Radiology*, **146**, 123-128 (1983)

[35] K. Oshio, D.A. Feinberg, "GRASE (Gradient and Spin) Imaging, A Novel Fast Imaging Technique", *Magnetic Resonance in Medicine*, **20**, 344-49 (1991)

[36] A. Haase, J. Frahm, D. Matthaei, W. Hänicke, D.K. Merbolt, "FLASH Imaging. Rapid NMR Imaging using low flip-angle pulses", *Journal of Magnetic Resonance*, **67**, 258 (1986)

[37] Y. Zur, M.L. Wood, L.J. Neuringer, "Spoiling of transverse magnetization in steady-state sequences", *Magnetic Resonance in Medicine*, **21**, 251 (1991)

[38] A. Oppelt, R, Grauman, H. Barfuß, H. Fisher, W. Schajor, "FISP – a new fast MRI sequence", *Electromedica*, **54**, 15 (1986)

[39] H.Y. Carr, "Steady-State Free Precession in Nuclear Magnetic Resonance", *Physical Review*, **112**, 1693–1701 (1958)

[40] J.W. Carlsson, "An algorithm for NMR imaging reconstruction based on multiple receiving coils", *Journal of Magnetic Resonance*, **74**, 376-80 (1987)
M. Hutchinson, U. Raff, "Fast MRI data acquisition using multiple detectors", *Magnetic Resonance in Medicine*, **6**, 87-91 (1987)

[41] D.K. Sodickson, W.J. Manning, "Simultaneous acquisition of spatial harmonics (SMASH): Fast imaging with radiofrequency coil arrays", *Magnetic Resonance in Medicine*, **38**, 591-603 (1997)

[42] K.P. Preussmann, M. Weiger, M.B. Scheidegger, P. Boesiger, "SENSE: Sensitivity encoding for fast MRI", *Magnetic Resonance in Medicine*, **42**, 952-62 (1999)

[43] P. Mansfield, "Multi-planar image formation using NMR Spin Echoes", *Journal of Physics*, **C10**, L55-58 (1977)
R.J. Ordidge, P. Mansfield, R.E. Copeland, "Rapid biochemical imaging of NMR", *British Journal of Radiology*, **54**, 850-55 (1981)

[44] J. Reilly, "Peripheral nerve stimulation by induced currents: Exposure to time-varying magnetic fields", *Medical and Biological Engineering and Computing*, **27**, 101-10 (1989)

[45] B. Chapman, R. Turner, R.J. Ordidge, M. Doyle, M. Cawley, R. Coxon, P. Glover, P. Mansfield, "Real-time movie imaging from a single cardiac cycle", *Magnetic Resonance in Medicine*, **5**, 246 (1987),
http://www.geocities.com/CapeCanaveral/9469/project_01.html

[46] M. Wood, M. Henkelman, "MR image artifacts from periodic motion", *Medical Physics*, **12**, 143-51 (1985)

[47] P. Lanzer, E. Botvinick, E. Lorenz, "Cardiac imaging using gated nuclear magnetic resonance", *Radiology*, **150**, 121 (1984)

[48] D.G. Nishimura, A. Macovski, J. Pauli, "Magnetic resonance angiography", *IEEE Transactions on Medical Imaging*, **5**, 140-51 (1986)
R.G. de Graaf, J.P. Groen, "MR Angiography with pulsatile flow", *Magnetic Resonance Imaging*, **10**, 25-34 (1992)

[49] R.J. Singer, "Blood-flow rates by NMR measurements", *Science*, **13**, 1652-1655 (1959)

[50] V.M. Runge, R.G. Stewart, J.A. Clanton *et al.*, "Work in progress: Potential oral and intravenous paramagnetic NMR contrast agents", *Radiology*, **143**, 789-91 (1983)

[51] R.C. Shettle, "Experiments Showing the Paramagnetic Condition of Arterial Blood, as Compared with the Diamagnetic Condition of Venous Blood", *Proceedings of the Royal Society of London*, **23**, 116-120 (1874)
L. Pauling, C. Coryell, "The magnetic properties and structure of hemoglobin", *Proceedings of the National Academy of Sciences*, **22**, 210 (1936)

[52] J.M. Gomori, R.I. Grossman, H.I. Goldberg, R.A. Zimmerman, and L.T. Bilaniuk, "Intracranial hematomas: Imaging by high-field MR", *Radiology*,

Kenneth Buckwalte, Indiana University, School of Medicine, *MRI*, 1998,
http://www.indyrad.iupui.edu/public/lectures/mri/iu_lectures/spatial_encoding_part_1/sld001.htm

Norbert Retti, Radiodiagnostik I der Universitätsklinik Innsbruck, Austria, *MR online*
http://members.telering.at/mri/index.html
Nicely presented and explained pulse sequences, In German

European Magnetic Resonance Forum 2003, *A short history of magnetic resonance imaging from a European point of view*
http://www.emrf.org/FAQs%20MRI%20History.html
An overview with pictures of early researchers and equipment

Andrei Volkov, *Contrast agents in Magnetic Resonance Imaging*, 1997
http://www.cc.utah.edu/%7Eav6a51/mri.htm

Some textbooks on NMR

A. Abragam, *Principles of Nuclear Magnetism*, Oxford University Press, 1978, ISBN: 0 19 851236 8, 1983 (pbk) 0 19 852014 X
The classic. Originally published 1961, reprinted 1978 and in paperback 1983.

J.W. Akitt, Brian E. Mann, *NMR and chemistry, An introduction to modern NMR spectroscopy*, Stanley Thornes, 4th ed 2000, ISBN: 0 7487 4344 8

Ray Freeman, *Magnetic resonance in chemistry and medicine*, Oxford University Press, Oxford, UK, 2003, ISBN: 0 199 26061 3 and 0 199 26225 X (pbk)

Roger S. Macomber, *A complete introduction to modern NMR spectroscopy*, Wiley 1998, ISBN: 0 471 15736 8

Jeremy K.M. Sanders, Brian K. Hunter, *Modern NMR Spectroscopy*, Oxford University Press, 2nd ed. 1993, ISBN: 0 19 855567 9
Very straight forward and the first parts are not hard to read

Daniel Canet, *Nuclear Magnetic Resonance, Concepts and Methods*, Wiley 1996, ISBN: 0 471 96145 0
This book has a rather theoretical approach including a quantum physical treatment. However, it contains much useful information and the latter parts may be skipped.

Brian P. Cowan, *Nuclear Magnetic Resonance and Relaxation*, Cambridge University Press 1997, ISBN: 0 521 30393 1
Rather theoretical, but covers much of the relaxation background of MRI not mentioned in most other books, Has a part about MRI

Nobel Prizes on NMR and MRI

Otto Stern, 1888-1969, Prize 1943 in physics "for his contribution to the development of the molecular ray method and his discovery of the magnetic moment of the proton"
Isidor I. Rabi, 1898-1988, Prize 1944 in physics "for his resonance method for recording the magnetic properties of atomic nuclei"
Felix Bloch, 1905-1983, and **Edward M.Purcell**, 1912-1997, Prize 1952 in physics "for their development of new methods for nuclear magnetic precision measurements and discoveries in connection therewith"
Nicolaas Bloembergen, 1920-, Prize 1981 in physics "for his contribution to the development of laser spectroscopy" (Theory of NMR relaxation)
Richard Ernst, 1933-, Prize 1991 in chemistry "for his contributions to the development of the methodology of high resolution nuclear magnetic resonance (NMR) spectroscopy" (Pulsed techniques and Fourier transform methods)
Kurt Wütrich, 1938-, Prize 2002 in chemistry "for his development of nuclear magnetic resonance spectroscopy for determining the three-dimensional structure of biological macromolecules in solution"
Paul C. Lauterbur, 1929-, and **Sir Peter Mansfield**, 1933-, Prize 2003 in physiology or medicine "for their discoveries concerning magnetic resonance imaging"

157, 87-93 (1985)

[53] S. Ogawa, R.S. Menon, D.W. Tank, S.G. Kim, H. Merkle, J.M. Ellermann, K. Ugurbil, "Functional brain mapping by blood oxygenation level-dependent contrast magnetic resonance imaging. A comparison of signal characteristics with a biophysical model", *Biophysics Journal*, **64**, 803-12 (1993)

[54] N. Krasnow, "Effects of lanthanum and gadolinium ions on cardiac sarcoplasmic reticulum", *Biochimica et Biophysica Acta*, **282**, 187-194 (1972)

[55] G.A. Mercier, "On the molecular spin density and the electrostatic potential as determinants of the relaxivity of metalloporphyrins", *Magnetic Resonance Imaging*, **13**, 807-817 (1995)

[56] V.M. Runge, T.H.P. Rijcken, A. Davidoff, J.V. Wells, D.D. Stark, "Contrast-enhanced MR imaging of the liver" *Journal of Magnetic Resonance Imaging*, **4**, 281-9 (1994)

[57] M.A. Bouchiat, T.R. Carver, C.M. Varnum, "Nuclear polarization in He-3 gas induced by optical pumping and dipolar exchange", *Physical Review Letters*, **5**, 373 (1960)

[58] M.S. Albert, G.D. Cates, B. Driehuys, W. Happer, B. Samm, C. Springer Jr, A. Wishnia, "Biological magnetic resonance imaging using laser-polarized ^{129}Xe", *Nature*, **370**, 199 (1994)

[59] J.R. McFall, H.C. Charles, R.D. Black, H. Middleton, J.C. Swartz, B. Saam, B. Driehuys, C. Erickson, W. Happer, G.C. Cates, G.A. Johnson, C.E. Ravin, "Human Lung Air Spaces. Potential for MR Imaging with hyperpolarized He-3", *Radiology*, **200**, 553-8 (1996)

[60] J.B. Aguayo, S.J. Blackband, J. Schoeninger, M.A. Mattingly, M. Hinterman, "Nuclear magnetic resonance imaging of a single cell", *Nature*, **322**, 190-191 (1986)

L. Ciobanu, C.H. Pennington, "3D Micron-scale of single biologic cells", *Solid State Nuclear Magnetic Resonance*, **25**, 138-41 (2004)

[61] J.F. Schenck, "Safety of strong, static magnetic fields", *Journal of Magnetic Resonance Imaging*, **12**, 2-19 (2000).

A. Kangarlu, P-M.L. Robitaille, "Biological Effects and Health Implications in Magnetic Resonance Imaging", *Concepts in Magnetic Resonance*, **12**, 321-59 (2000), and references therein.

[62] "Child Dies in MRI Machine", *The Associated Press*, filed at 2:42 p.m. ET, July 30, 2001

[63] J. Calvert, T. Taylor, "Death Link to Hospital Scan", *The Herald Sun News Corporation*, Melbourne Australia, Thursday April 13, 2000, p 10

[64] J.F. Schenck, "The role of magnetic susceptibility in magnetic resonance imaging: MRI magnetic compatibility of the first and second kinds", *Medical Physics*, **23**, 815 (1996)

[65] R.P. Klucznik, D.A. Carrier, R. Pyka, R.W. Haid, "Placement of intracerebral aneurysm clip in magnetic field with fatal outcome", *Radiology*, *187*, 855-6, (1993)

[66] W.M. Kelly, P.G. Paglen, J.A. Pearson, A.G. San Diego, M.A. Soloman, "Ferromagnetism of intraocular foreign body causes unilateral blindness after MR Study", *American Journal of Neuroradiology*, **7**, 243-5 (1986)

[67] M.H. Repacholi *et al.*, "Guidelines on limits of exposure to static magnetic fields", *Health Physics*, **66**, 100-106 (1994)

[68] P. Mansfield, B. Chapman, "Active magnetic screening of gradient coils in NMR imaging", *Journal of Magnetic Resonance*, **66**, 573-576 (1986)

Appendix I

Weighting of the MR image

	TR	TE	TI
Saturation recovery			
Spin (proton) density weighting	long	—	—
Conventional spin echo			
Spin density weighting	long	short	—
T1-weighting	short	short	—
T2-weighting	long	long	—
Inversion recovery			
T1-weighting, most common	short	short	
	depending		
T2-weighting	long	long	
	depending		

Approximate numbers

	Long	Intermediate	Short	Very short	
TR	> 2000	1000	< 700	250	ms
TE	> 80	50	< 30	10	ms
TI	≈ 1500	500	≈ 150	100	ms

Compare with tables 3 and 4

Appendix II

Some acronyms used in MRI

There are several more acronyms used than those in the table below, most of them dependent of the vendor. Some may be found at the following websites or in textbooks.

http://www.scmr.org/technologists/crossvendorlexicon.pdf,
Society for Cardiovascular Magnetic Resonance, Mt. Royal, N, 2004

http://www.emrf.org/FAQs%20Acronyms.html,
European Magnetic Resonance Forum 2004

http://fonar.com/glossary.htm,
MRI Glossary, Fonar Corporation, Mellville NY, 2004

α	Tip angle
α_E	Ernst angle
B_0	Static magnetic field, measured in Tesla (T)
BOLD	Blood Oxygen Level Dependent contrast
CE	Conventional Echo
CNR	Contrast to Noise Ratio
cpmg	Carr, Purcell, Meiboom, Gill – pulse sequence
CSE	Conventional Spin Echo
CSF	Cerebrospinal Fluid
CT	Computerized X-ray Tomography
DTPA	Diethylene Triamine Penta Acetic acid
DWI	Diffusion Weighted Imaging
E	Energy, measured in Joules
ECG	Electrocardiogram
EEG	Electroencephalogram
EPI	Echo Planar Imaging
f (or ν)	Frequency, measured in Hz, 1 Hz = 1 cycle/s
FDA	Food and Drug Adminstration
FE	Field echo
FFE	Fast Field Echo
FFT	Fast Fourier Transform
FID	Free Induction Decay
FISP	Fast Imaging Steady state Precession
FLAIR	Fluid Attenuated Inversion Recovery
FLASH	Fast Low Angle SHot
f_L	Larmor Frequency
fMRI	Functional Magnetic Resonance Imaging
FOV	Field Of View (studied area)

FSE	Fast Spin Echo (=TSE)
FSPGR	Fast Spoiled Gradient Recall Echo
FT	Fourier Transform
γ	gyromagnetic ratio (or magneto-gyric ratio)
Gauss	1 Gauss = 0,1 mT
Gd	Gadolinium, transition metal used in contrast agents
GFE	Gradient Field Echo
GRASE	Gradient and Spin Echo
GRE	Gradient Refocused Echo
HP	Hyper Polarized
IR	Inversion Recovery
k-space	where primary imaging data are stored in computer
K	Kelvin, 0 K = -273.15°C
MRA	Magnetic Resonance Angiography
MRE	Magnetic Resonance Elastography
MRI	Magnetic Resonance Imaging
NEX	Number of EXitations
NMR	Nuclear Magnetic Resonance
NMV	Net Magnetization Vector
ν (or f)	Frequency, measured in Hz, 1 Hz = 1 cycle/s
ω	Angular frequency, measured in radians/s, $\omega = 2\pi \bullet \nu = 2\pi \bullet f$
PC	Phase Contrast
PET	Positron emission tomography
PFI	Partial flip angle
ppm	Parts Per Million = 10^{-6}
RF	Radio Frequency
rad	Radian, 1 rad = $180/\pi$ degrees
RARE	Rapid Acquisition with Refocused Echoes (=TSE)
ROI	Region Of Interest
SE	Spin Echo
SENSE	SENSitivity Encoding
SGE	Spoiled Gradient Echo
SL	Spin Lock
SMASH	Simultaneous Acquisition of Spatial Harmonics
SNR	Signal to Noise Ratio
SR	Saturation Recovery
SSFP	Steady State Free Precession
SPIR	Spectral Selective Inversion Recovery
STIR	Short Time Inversion Recovery
T	Tesla, 1 T = 10,000 Gauss
T1	Longitudinal (or spin-lattice) relaxation time
T2	Transverse (or spin-spin) relaxation time
$T_{1/2}$	Half life of excited system
T2*	Effective transverse relaxation time
TE	Echo Time
TI	Inversion Time
TOF	Time Of Flight
TR	Repetition Time
TSE	Turbo Spin Echo
VENC	Velocity ENCoding

Appendix III

MRI Literature

General books on MRI and imaging

Dominik Weishaupt, Victor D. Koechli, *How does MRI work?*, Springer Verlag, 2nd ed. 2006, ISBN: 3 540 30067 8
Short with just the bascis

Pottumarthi V. Prasad, ed., Magnetic Resonance Imaging, Humana Press, 2006, ISBN: 1 588 29397 1
For biological scientists and researchers with CD ROM

Catherine Westbrook, Carolyn Kaut, Jon Talbot, *MRI in practice*, Blackwell Science, 3rd ed. 2005, ISBN: 1 405 12787 2
Covers basic MRI. Fairly easy to read.

Val M. Runge, *The Physics of Clinical MRI taught through Images*, Thieme, 2005, ISBN: 1 588 90322 2
MRI physics for beginners with a practical approach

D.G. Mitchell, M. Cohen, *MRI Principles*, Saunders, Philadelphia, 2nd ed. 2004, ISBN: 0 721 60024 7

R.H. Hasheim, W.G. Bradley, C.J. Lisantic, *MRI, The Basics*, Lippincott Williams and Wilkinson, Philadelphia, 2nd ed. 2004, ISBN: 0 781 74157 2

Evan Siegelman, ed., *Body MRI*, Saunders, 2004, ISBN: 0 721 63740 X

Mark A. Brown, Richard C. Semelka, *MRI, Basic Principles and Applications*, Wiley Liss, Hoboken, N.J., 3rd ed. 2003, ISBN: 0 471 43310 1

Stewart C. Bushong, *Magnetic resonance imaging. Physical and biological principles*, Mosby Inc., 3rd ed. 2003, ISBN: 0 3223 01485 2
Covers very much about theory and practice, but not in depth, easy to read.

Peter Reimer, Paul M. Parizel, Falko-Alexander Stichnoth, eds., *Clinical MR-imaging, A practical approach*, Springer Verlag, 2nd ed. 2003, ISBN: 3 540 43467 4
A computer supported diagnostic manual.

Paul Suetens, *Fundamentals of medical imaging*, Cambridge University Press, Cambridge, UK, New York, 2002

Jerrold T. Bushberg, J. Anthony Seibert, Edwin M. Leidholdt Jr., John M. Boone, *The Essential Physics of Medical Imaging*, Lippincott Williams & Wilkins Publishers, 2nd ed. 2002, ISBN: 0 683 30118 7
Very nicely written. Two chapters are devoted to MRI.

Catherine Westbrook, *Handbook of MRI technique*, Blackwell Science 1999 (2002), ISBN: 0 632 05264 3
A clinical approach with not much theory. Much hands-on.

Gregory L. Wheeler, Kathryn E. Withers, Shappell, *Lippincott's Magnetic Resonance Imaging Review*, Lippincott Williams & Wilkins, 1996, ISBN: 0 397 55156 8

Allen D. Elster, Jonathan H. Burdette, *Questions & answers in magnetic resonance imaging*, Mosby, 2nd ed., 2001, ISBN: 0 323 01184 5

Felix W. Wehrli, Derek Shaw, J. Bruce Neland, *Biomedical Magnetic Resonance Imaging*, VCH Publishers 1988, ISBN: 0 89573 349 8
Covers the subject in great detail with much fundamental theory relevant for biological imaging.

Peter A. Rinck, *Magnetic resonance in medicine*, Blackwell Wissenschafts-Verlag, 4th ed. 2001, ISBN: 0 632 05986 9
"The Basic Textbook of the European Resonance Forum". Includes MRI and MRS.

Functional MRI

Scott H. Faro, Feroze B. Mohamed, Functional MRI, *Basic Principles and Clinical Applications*, Springer Verlag, 2006, ISBN: 0 387 23046 7
Clinical applications for the non-scientist physician

Scott A. Huettel, Allen W. Song, Gregory McCarty, *Functional Magnetic Resonance Imaging*, Sinauer, Sunderland, 2004, ISBN: 0 878 93288 7
Introductory textbook with CD ROM

Peter Jezzard, Paul M. Matthews, Stephen M. Smith, *Functional MRI, An Introduction to Methods*, Oxford University Press 2001, ISBN: 0 192 63071 7

Richard B. Buxton, *Introduction to functional magnetic resonance imaging, principles and techniques*, Cambridge University Press, 1st ed. 2001, ISBN: 0 521 58113 3,

C.T.V. Moonen, P.A Bandettini., eds., *Functional MRI*, Springer 2000, ISBN: 3 540 67215 X

MRI Safety

Frank G. Shellock, Emanuel Kanal, *Magnetic Resonances: Bioeffects, safety, and patient management*, Lippincott-Raven, 2nd ed., 1996, ISBN: 0 397 58437 7

Carolyn Kaut-Roth, Euclid Seram, *Rad Techs Guide to MRI: Imaging procedures, Patient Care and Safety*, Blackwell Science Inc, 2002, ISBN: 0 632 04507 8

Reference books

Robert R. Edelman, John R. Hesselink, Michael B. Zlathin, *Clinical Resonance Imaging*, vol 1 & 2, W.B. Saunders Co, 2nd ed. 1996, ISBN: 0 721 65221 2
Covers most every topic up to 1994.

F.R. Gutierrez, J.J. Brown, S.A. Mirowitz, eds., *Magnetic resonance imaging*, vol 1 & 2, Mosby year book, St. Louis 1992

MRI with a more theoretical approach

Marinus T. Vlardingerbroek, Jaques A. den Boer, *Magnetic resonance imaging, Theory and practice*, Springer, 3nd ed. 2003, ISBN: 3 540 43681 2
Has many theoretical parts, but also several practical details, (lacks index)

Vadim Kuperman, *Magnetic resonance imaging, Physical principles and applications*, Academic Press 2000, ISBN: 0 12 429150 3
Nicely written, covers rather much in a small volume

Zhi-Pei Liang, Paul C. Lauterbur, *Principles of magnetic resonance imaging: a signal processing perspective*, IEEE Press, New York 2000, ISBN: 0 780 34723 4

Walter Johannes Schempp, *Magnetic resonance imaging. Mathematical foundations and applications*, Wiley-Liss 1998, ISBN: 0 471 16736 3

E. Mark Haacke *et al.*, *Magnetic resonance imaging, physical principles and sequence design*, J. Wiley & Sons 1999, ISBN: 0 471 35128 8

Jian-Ming Jin, *Electromagnetic analysis and design in magnetic resonance imaging*, CRC Press 1999, ISBN: 0 849 39693 X
Rather theoretical, describes design of receiving coils, presents many images in colour

NMR-MRI Journals

Applied Magnetic Resonance, Springer Verlag, Wien, ISSN: 0937-9347
http://www.springeronline.com/sgw/cda/frontpage/0,11855,5-10054-70-1102827-0,00.html

Concepts in Magnetic Resonance, A and B, John Wiley & Sons, ISSN: 1546-6086
http://www3.interscience.wiley.com/cgi-bin/jhome/38226

Diagnostic Imaging, CMP Medica
http://www.diagnosticimaging.com, http://www.dimag.com/

European Radiology, European Congress of Radiology, Leuven, Belgium
http://www.ecr.org/pages/members/european-radiology.php

International Journal of Imaging Systems and Technology, John Wiley and Sons, ISSN: 0899-9457
http://eu.wiley.com/WileyCDA/WileyTitle/productCd-IMA.html

Journal of Biomolecular NMR, Springer, The Netherlands, ISSN: 0925-2738
http://www.kluweronline.com/issn/0925-2738/contents

Journal of Magnetic Resonance, Elsevier, ISSN: 1090-7807
http://www.sciencedirect.com/science/journal/10907807

Journal of Magnetic Resonance Imaging, John Wiley & Sons, ISSN: 1053-1807
http://www3.interscience.wiley.com/cgi-bin/jhome/10005199

Magnetic Resonance Imaging, Elsevier, ISSN: 0730-725X
http://www.sciencedirect.com/science/journal/0730725X

Magnetic Resonance in Chemistry, John Wiley & Sons, ISSN: 0749-1581
http://www3.interscience.wiley.com/cgi-bin/jhome/3767

Magnetic Resonance in Medicine, John Wiley & Sons, ISSN: 0740–3194
http://www.interscience.wiley.com/jpages/0740-3194
http://www3.interscience.wiley.com/cgi-bin/jhome/10005196

Magnetic Resonance Materials in Physics, Biology and Medicine (MAGMA), Springer, Berlin, ISSN: 0968-5243
http://www.springerlink.com/content/1352-8661/

Medical Image Analysis, Elevier, ISSN: 1361-8415
http://www.elsevier.com/wps/find/journaldescription.cws_home/620983/description

Medical Physics, American Institute of Physics, ISSN: 0094-2405
http://scitation.aip.org/medphys/

NMR in Biomedicine, John Wiley & Sons, ISSN: 0952-3480
http://www3.interscience.wiley.com/cgi-bin/jhome/13087

Progress in Nuclear Magnetic Resonance Spectroscopy, Elevier, ISSN: 0079-6565
http://www.sciencedirect.com/science/journal/00796565

Radiology, Radiological Society of North America, ISSN: 0033-840033-8419
http://radiology.rsnajnls.org/

Topics in Magnetic resonance Imaging, Lippincott Williams& Wilkins, ISSN: 0899-3459
http://www.topicsinmri.com/pt/re/tmri/home.htm

Review articles

J. Hennig, O. Speck, M.A. Koch, C. Weiller, "Functional magnetic resonance imaging: a review of methodological aspects and clinical applications". *Journal of Magnetic Resonance Imaging*, **18**, 1-15 (2003)

Norman F. Ramsey, "Early history of magnetic resonance", *Physics in Perspective*, **1**, 123-35 (1999)

Mark. S. Cohen "Echo-planar imaging (EPI) and functional MRI", in Bandettini, Moonen, *Functional MRI*, Springer, Berlin 2000

Stuart Clare, "Functional Magnetic Resonance Imaging: Methods and Applications", *Dissertation*, University of Nottingham, UK, October 1997
http://www.fmrib.ox.ac.uk/~stuart/thesis/

A. Oppelt, T. Grandke, "Magnetic Resonance Imaging", *Superconductor Science and Technology*, **6**, 381-95 (1993)

Mark S. Cohen, B.R. Rosen, T.J. Brady, "Breaking the speed limit in MRI", *Magnetic Resonance*, **2**, 26-37 (1992)

Peter Mansfield, "Imaging by nuclear magnetic resonance", *Journal of Physics E, Scientific Instruments*, **21**, 18-30 (1988)

Randall B. Lauffer, "Paramagnetic metal complexes as water proton relaxation agents for NMR imaging: Theory and design", *Chemical Reviews*, **87**, 901 (1987)

E. R. Andrew, "NMR Imaging", *Accounts of Chemical Research*, **16**, 114-22 (1983)

Texts published on line

Joseph P. Hornak, *The Basics of MRI*, Published on the web 1996-2004,
http://www.cis.rit.edu/htbooks/mri/
This is an on-line book, which covers just as much as an ordinary rather small book, animations.

J. Ray Ballinger, *Introduction to MRI*, Published on the web 1994-1996 and 1999-2003
http://www.mritutor.org/mritutor/index.html
Mostly fundamental topics, but has many entries for different parts of the NMR and MRI theory. An accompanied "MRI teaching file" includes many clinical examples

Wm Faulkner, *Basic MR Principles*, Published on the web 1996,
http://www.radscice.com/basic.html
Covers only the most basic principles but no clinical applications. It has some nice animations

Doug Fletcher, *MRI Physics, A nuts and bolts approach*, 1999
http://rad.usuhs.mil/rad/handouts/fletcher/fletcher/sld001.htm
Contains about 60 handouts for over-head projections accompanying an MRI course

University of Missouri Health Care, *MRI Lectures 1-7*
http://radiology.muhealth.org/new_Radiology_Web/TeachingFiles/MRI_Lectures.htm, http://radiology.muhealth.org and advices on that site
Very ambitious

Todd A. Gould, *How MRI works*, Published on the web, continuously updated
http://www.howstuffworks.com/mri.htm
A popular non-quantitative text for the general public

Gary P. Liney, *Magnetic Resonance Imaging (MRI)*, 2003,
http://www.hull.ac.uk/mri/lectures/Gpl%20web%20page/gpl_page

Kenneth Buckwalte, Indiana University, School of Medicine, *MRI*, 1998,
http://www.indyrad.iupui.edu/public/lectures/mri/iu_lectures/spatial_encoding_part_1/sld001.htm

Norbert Retti, Radiodiagnostik I der Universitätsklinik Innsbruck, Austria, *MR online*
http://members.telering.at/mri/index.html
Nicely presented and explained pulse sequences, In German

European Magnetic Resonance Forum 2003, *A short history of magnetic resonance imaging from a European point of view*
http://www.emrf.org/FAQs%20MRI%20History.html
An overview with pictures of early researchers and equipment

Andrei Volkov, *Contrast agents in Magnetic Resonance Imaging*, 1997
http://www.cc.utah.edu/%7Eav6a51/mri.htm

Some textbooks on NMR

A. Abragam, *Principles of Nuclear Magnetism*, Oxford University Press, 1978, ISBN: 0 19 851236 8, 1983 (pbk) 0 19 852014 X
The classic. Originally published 1961, reprinted 1978 and in paperback 1983.

J.W. Akitt, Brian E. Mann, *NMR and chemistry, An introduction to modern NMR spectroscopy*, Stanley Thornes, 4th ed 2000, ISBN: 0 7487 4344 8

Ray Freeman, *Magnetic resonance in chemistry and medicine*, Oxford University Press, Oxford, UK, 2003, ISBN: 0 199 26061 3 and 0 199 26225 X (pbk)

Roger S. Macomber, *A complete introduction to modern NMR spectroscopy*, Wiley 1998, ISBN: 0 471 15736 8

Jeremy K.M. Sanders, Brian K. Hunter, *Modern NMR Spectroscopy*, Oxford University Press, 2nd ed. 1993, ISBN: 0 19 855567 9
Very straight forward and the first parts are not hard to read

Daniel Canet, *Nuclear Magnetic Resonance, Concepts and Methods*, Wiley 1996, ISBN: 0 471 96145 0
This book has a rather theoretical approach including a quantum physical treatment. However, it contains much useful information and the latter parts may be skipped.

Brian P. Cowan, *Nuclear Magnetic Resonance and Relaxation*, Cambridge University Press 1997, ISBN: 0 521 30393 1
Rather theoretical, but covers much of the relaxation background of MRI not mentioned in most other books, Has a part about MRI

Nobel Prizes on NMR and MRI

Otto Stern, 1888-1969, Prize 1943 in physics "for his contribution to the development of the molecular ray method and his discovery of the magnetic moment of the proton"
Isidor I. Rabi, 1898-1988, Prize 1944 in physics "for his resonance method for recording the magnetic properties of atomic nuclei"
Felix Bloch, 1905-1983, and **Edward M.Purcell**, 1912-1997, Prize 1952 in physics "for their development of new methods for nuclear magnetic precision measurements and discoveries in connection therewith"
Nicolaas Bloembergen, 1920-, Prize 1981 in physics "for his contribution to the development of laser spectroscopy" (Theory of NMR relaxation)
Richard Ernst, 1933-, Prize 1991 in chemistry "for his contributions to the development of the methodology of high resolution nuclear magnetic resonance (NMR) spectroscopy" (Pulsed techniques and Fourier transform methods)
Kurt Wütrich, 1938-, Prize 2002 in chemistry "for his development of nuclear magnetic resonance spectroscopy for determining the three-dimensional structure of biological macromolecules in solution"
Paul C. Lauterbur, 1929-, and **Sir Peter Mansfield**, 1933-, Prize 2003 in physiology or medicine "for their discoveries concerning magnetic resonance imaging"

Magnetic Resonance Imaging MRI – An Overview

Magnetic Resonance Imaging or MRI is a modern diagnostic tool for acquiring information from the interior of a human body. MRI can create three-dimensional images of a human organ without hurting the organ in any way and without using any ionizing radiation.

The body must be placed in a strong magnetic field, more than ten thousand times the magnetic field of Earth. A radio signal is sent into the body, where it is absorbed by hydrogen atoms. The hydrogens in the body respond by sending back a signal to a detector. The strength of this signal mirrors the amount of hydrogen in various parts of the body.

When creating an image of an organ, the signal must be acquired from every part of the organ point by point by a scanning procedure. To accomplish this, the magnetic field is rapidly varied with gradients in three dimensions. The monitored signal is the sum of signals from every unique volume element within the body. The instrument receives a large number of data from which the signal magnitude of every volume element can be calculated.

Most parts of the body have a roughly equal concentration of hydrogen. However, signals from hydrogen atoms decay with unequal speeds depending on their various environments. Therefore the magnitude of the induced radio signal is monitored some time after the end of the initializing pulse. Now various tissues show different signal strengths, from which it is possible to build an image with desired contrast. This is often excellent, even for soft tissues. It is of special interest that the signal from hydrogen in lesions and tumors often decay slower than surrounding tissues and therefore can be detected in the image.

The contrast of the image is created by the experimental procedure and is not inherent in the imaged body. Thus different experimental routines will result in unequal images – not pictures – of the same object. Therefore it is crucial that the experimenter learns how to use different RF-pulse sequences and how to interpret the result.

Karlstad University Studies
ISSN 1403-8099
ISBN 978-91-7063-125-2