

# Principles of MRI



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### Table 1: Materials that are of high signal intensity on T1 weighted MRI

Lipid and cholesterol
Subacute blood product (Met-Haemoglobin)
Gadolinium
Melanin, mucin and other proteins
Calcification (rarely)
Copper in Wilson's disease

## Basic MRI physics

### Frequency and Phase

The rotation of protons can be described in terms of frequency and phase. In a magnetic field, the protons spin around the axis of the magnet at a given frequency which is proportional to the field strength. This is governed by the Larmor equation: frequency = Larmor constant x magnetic field strength.

In addition to frequency, protons rotating in a magnetic field have a property known as phase. In the resting state, the protons are not in phase (Figure 3a). However, when particular radiofrequency pulses are used, the protons can be brought into phase (Figure 3b). When the radiofrequency is turned off, they dephase again. The ability to manipulate the phase of protons is important for differentiating tissues (see T2 weighted imaging) and for the spatial localisation of signal intensity (see spatial localisation).

### Understanding spatial frequencies

MR images are acquired and processed in terms of spatial frequency rather than pixel by pixel. This is necessary to overcome the difficulty in localising the origin of MR signal intensity (see spatial localisation). An image is made up of pixels that have amplitude (degrees of greyness) and position. One row of pixels can be represented by a waveform which plots amplitude on the y-axis and distance (position) on the x-axis (Figure 6). A mathematical operation known as a Fourier transform breaks this wave form down into constituent sine waves of differing amplitudes and frequencies. Fourier mathematics can be used to encode a 2D image entirely in terms of spatial frequencies rather than for each row of pixels in turn. This is difficult to conceptualise, but the spatial frequency representation of an image is referred to as k-space.

### Spatial localisation

#### (i) Slice selection

A magnetic gradient is applied along the axis of the magnet. Since protons resonate at frequencies depending on the applied field strength, slices can be selected by varying the radiofrequency energy used to acquire data (Figure 7).

#### (ii) In slice localisation

We learnt earlier that MR images are obtained in terms of spatial frequency. This followed a breakthrough in spatial localisation known as phase encoding. The phase of protons was discussed earlier, as well as the ability to manipulate phase using magnetic gradients and radiofrequency. Phase encoding is a brilliant way of localising signal intensity. We have seen that data is being obtained slice by slice by varying the magnetic field along the length of the patient and using a radiofrequency to which only protons in the slice of interest will resonate (Figure 7). How can

we localise signal intensity within the slice? Imagine applying a gradient along the x-axis of the slice which causes phase to vary by  $3 \times 360$  degrees across the region of interest (Figure 8). If there is a uniform distribution of protons in this area, signal intensity will cancel out to zero. However, if there are three blocks of protons evenly spaced out, signal intensity from these protons will summate giving a strong signal. This gradient can be said to be sensitive to spatial frequencies of three per unit length. The gradient can be varied to detect any spatial frequency required.

### T1 and T2 relaxation times

T1 and T2 are rate constants governing the return of protons to a resting state following excitation by a combination of applied radiofrequency energy and magnetic field changes.

T1 refers to the recovery of the protons' magnetic field along the axis of the magnet. When radiofrequency is applied, the protons are deflected away from the magnetic field and when it is turned off, the protons re-align with the field. During this process, electromagnetic radiation is emitted. The rate constant governing the return of longitudinal magnetisation is called T1. T1 is governed by the interaction of protons with large molecules such as cell membranes, lipids and myelin and is sometimes referred to as spin-lattice relaxation. Protons that are heavily bound have a rapid recovery of magnetisation and a short T1. There are a few substances that have a particularly short T1 (see Table 1) so that lesions that are high intensity on T1 can be easily characterised. Protons that are free to move, for example in solution, have a long T1.

T2 governs the loss of phase that occurs when a radiofrequency pulse is switched off. T2 is dominated by interaction with other protons (spin-spin interactions). Protons that are closely packed influence each other more than those in free solution. Thus water (very loosely packed) has a long T2 and tissues with little water content have a short T2. T2 is particularly useful in detecting the water content of tissues and is sensitive to pathologies that generate oedema (inflammation, infection, ischaemia, neoplasia etc).

### Artefacts

#### Ghosting

If errors are made in the detection of spatial frequency data, ghosts of the image are generated usually to the left and right side of the main image in the phase encoding axis. This can occur because of patient motion or from flowing blood which has a different phase to the surrounding tissues. The mathematical explanation for this phenomenon is beyond the scope of this article but can be found in an MR physics text.

#### Gibbs or truncation artefact

We saw earlier that MRI has difficulty in representing straight lines and sharp edges. When there is a

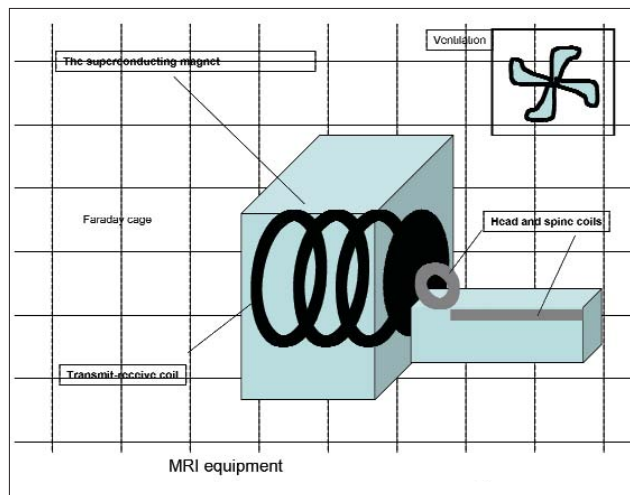


Figure 1 - MRI equipment

- A superconducting magnet made of millions of turns of wire consisting of Niobium-Titanium alloy embedded in copper. This is maintained at a temperature below 77K by liquid helium which in turn is kept cold by a refrigeration unit (cold head). Helium is slowly lost from the system and must be refilled every 2-3 years.
- Transmit-receive coils made of copper which emit radiofrequency energy when an electrical current is passed through them. When radiofrequency passes through them an electrical current is induced and this is how magnetic resonance signal intensity is detected.
- Motorised table
- Faraday cage – copper wires are built into walls and ceiling to prevent extraneous radiofrequency energy entering the room.
- Ventilation is required to maintain oxygen levels, which can become depleted if helium slowly leaks into the room even during normal operation of the magnet. If cooling of the magnet fails, the coils will become resistive causing a sudden rapid rise in temperature, a process known as quenching. This will result in a large release of helium gas, and facilities must be in place to allow this to be safely vented to the outside of the building.

straight line, such as the interface between the cord and CSF multiple lower intensity lines may be repeated on either side of the interface. This can be mistaken for an intrinsic cord abnormality. The artefact is named after Willard Gibbs (1839-1903) an American physicist who first described this phenomenon in relation to Fourier transforms. It is also referred to a truncation artefact because the problem would not arise if an infinite number of spatial frequencies were used (in practice the number of frequencies is truncated to allow a reasonable imaging time).

**Phase wrap**

If the field of view selected is smaller than the slice, signal from outside the field of view may be projected on top of the desired image. This artefact can be avoided by selecting a larger field of view or nulling the signal from areas that are not of interest.

**Susceptibility**

If there is material in the region of interest which modifies the magnetic field, signal in adjacent tissue is modified and spatially misrepresented. This leads to distortion of the image, usually with loss of signal in adjacent tissues.

**Chemical shift**

Protons in fat resonate at slightly different frequencies to protons in water when the same magnetic field strength is applied. This results in slight spatial misregistration of signal from fat. ♦

**REFERENCE**

McRobbie DW, Moore EA, Graves MJ, Prince MR. MRI: From picture to proton. Cambridge University Press, 2003.

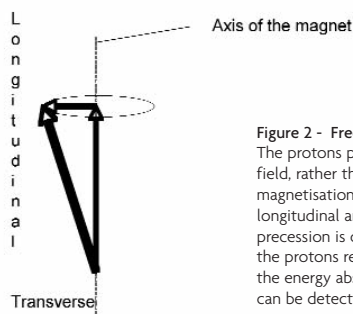


Figure 2 - Frequency

The protons precess around the axis of the magnetic field, rather than being precisely aligned with it. The magnetisation of the proton can be resolved into longitudinal and transverse components. The frequency of precession is crucial to magnetic resonance imaging, because the protons resonate to radioenergy of that frequency, and the energy absorbed is later re-emitted in radiowaves that can be detected.

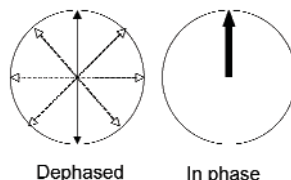


Figure 3 - Phase

Phase is a separate property of precessing protons which may either rotate asynchronously or in synchrony (in phase). The phasing and dephasing of protons is an important contributor to tissue contrast in MRI, particularly for T2 weighted and flow related imaging (MR angiography).

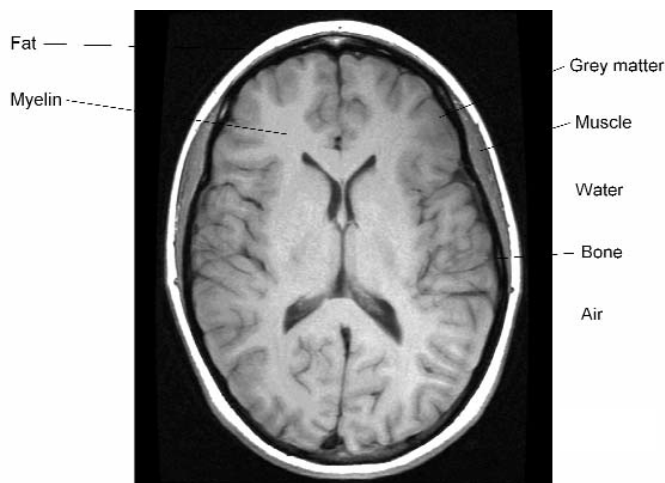


Figure 4 - T1 weighting

This is sometimes referred to as spin-lattice imaging as it is influenced by the interaction of protons with large molecules. These molecules tend to shorten T1 and this effect is displayed as high intensity on MR images. On this T1 weighted image, subcutaneous fat and myelin are shown as high intensity; grey matter and muscle are intermediate intensity; water is low intensity; bone and air are very low intensity.

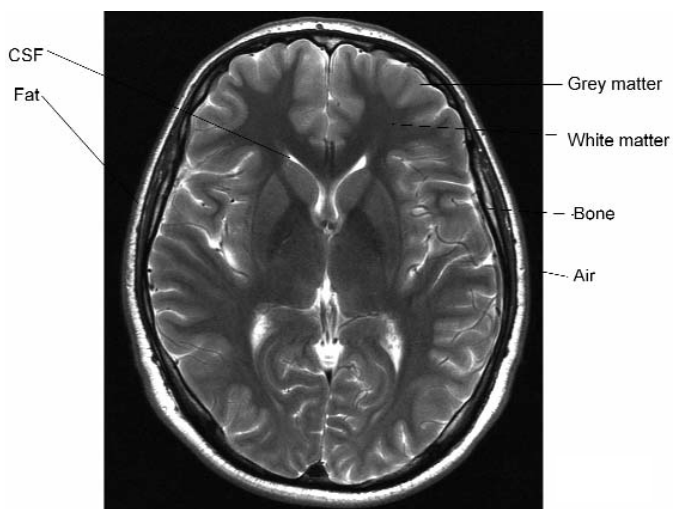
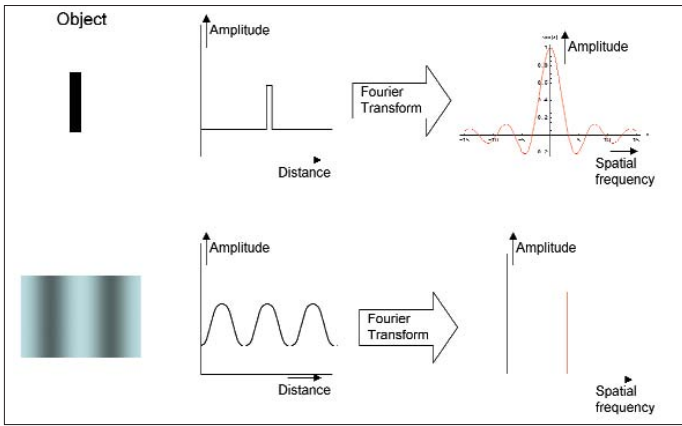
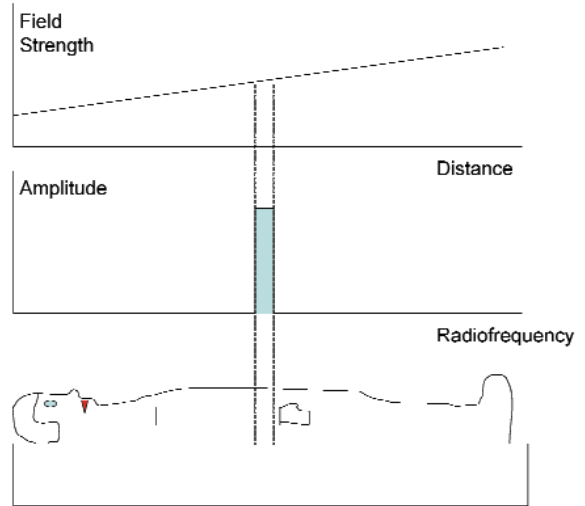


Figure 5 - T2 weighting

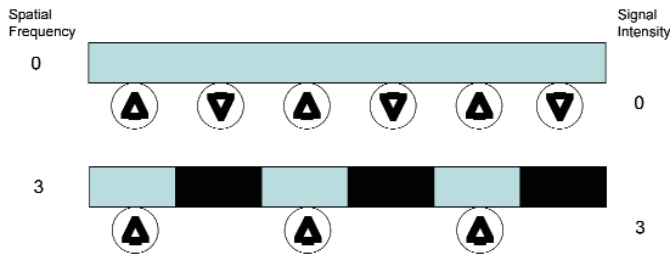
T2 weighted imaging may be referred to as spin-spin imaging which describes the effect of neighbouring protons on each other. T2 refers to the rate at which the phase of protons is lost over time. In practice, T2 hyperintensity is strongly influenced by water content, although fat is also hyperintense. On this T2 weighted image, CSF and subcutaneous fat are high intensity; grey matter is higher intensity than white matter; bone and air are very low intensity.



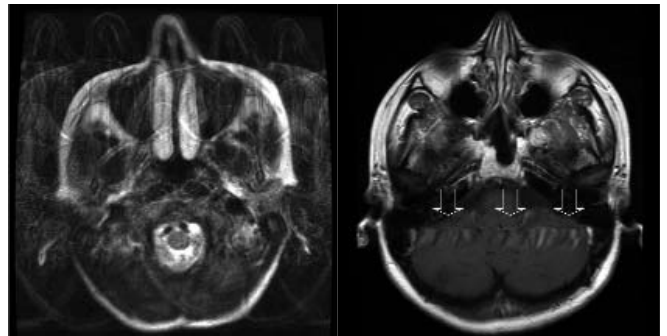
**Figure 6 – Spatial frequencies**  
 For simplicity, the objects chosen on the left vary only in one dimension. The first object, a line, has a single area of increased intensity on the amplitude/distance plot. However, on the amplitude/spatial frequency plot, an infinite number of spatial frequencies are required to capture the imaging data precisely. In particular, smaller and smaller amplitudes are required at higher spatial frequencies to represent perfectly the sharp edges of the line. The second object, a continuously varying background has a sinusoidal pattern on the amplitude/distance plot. This is easily represented in the frequency domain with only a single spatial frequency required. In general, few spatial frequencies are required to define objects with blurred margins whereas a large number are required to reproduce sharp margins.



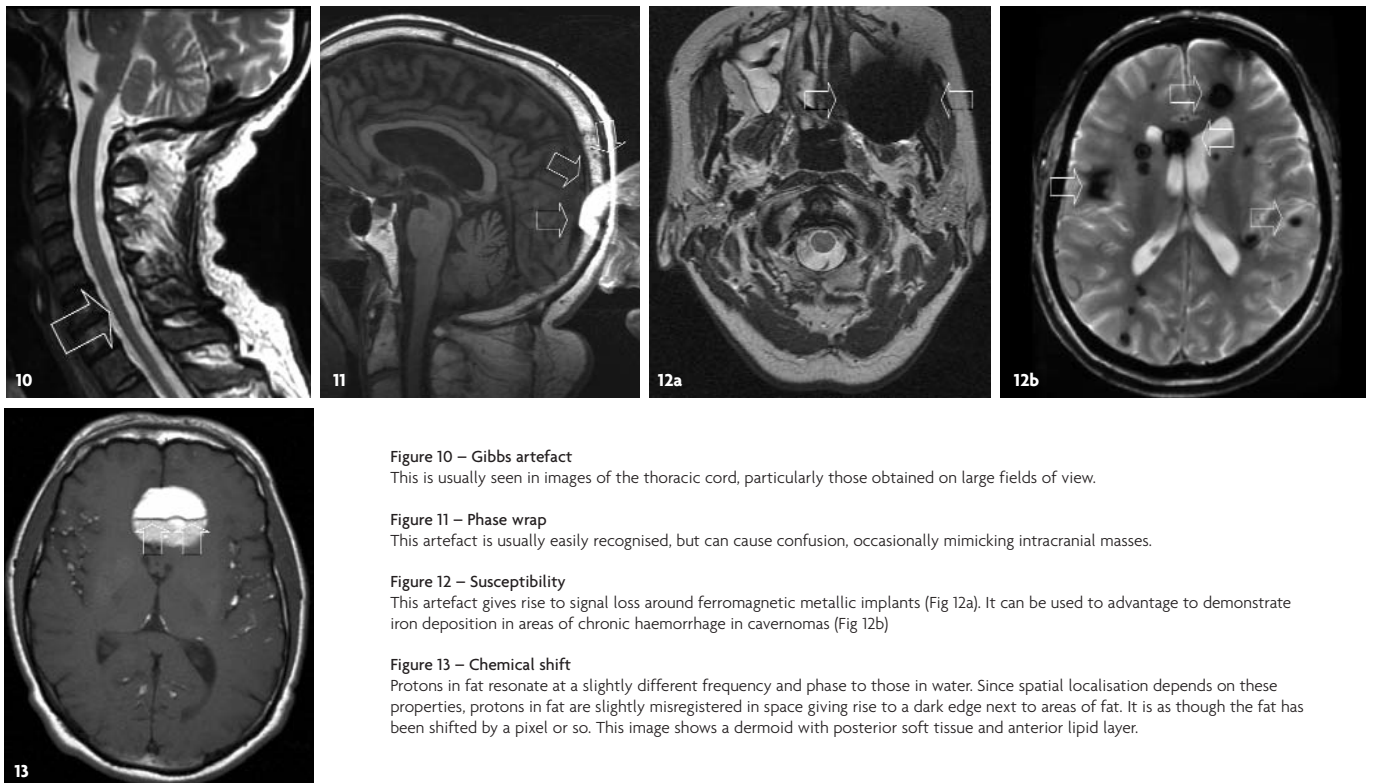
**Figure 7 – Slice selection**  
 The diagram illustrates how a combination of a magnetic gradient and a radiofrequency pulse can be used to excite a particular slice of the body. The thickness of the slice depends on the steepness of the magnetic gradient and on the range of radiofrequencies used (bandwidth).



**Figure 8 – In slice localisation**  
 This diagram illustrates the principle of localisation by phase-encoding. A magnetic gradient applied across a slice of tissue will alter the phase of protons along the plane of the gradient. In this case, a gradient has been applied that varies the phase by  $3 \times 360$  degrees per unit length. In the upper section, protons are evenly distributed and do not generate any signal because of phase cancellation. In the lower section, protons are distributed at a spatial frequency of 3 per unit length. Their signal intensities summate perfectly so that the gradient has sensitivity to spatial frequencies of 3 per unit length. The gradient must be varied to detect all the spatial frequencies required for a complete image (typically 256 frequencies for MRI).



**Figure 9 – Ghosting**  
 Ghosting artefacts are common in MRI and take the form of repeated representations of high intensity areas across the image. Motion is one of the frequent causes (Fig 9a left). Flowing blood also produces ghosts across the image which are more obvious on gadolinium enhanced acquisitions and may be mistaken for enhancing lesions (Fig 9b right).



**Figure 10 – Gibbs artefact**  
 This is usually seen in images of the thoracic cord, particularly those obtained on large fields of view.

**Figure 11 – Phase wrap**  
 This artefact is usually easily recognised, but can cause confusion, occasionally mimicking intracranial masses.

**Figure 12 – Susceptibility**  
 This artefact gives rise to signal loss around ferromagnetic metallic implants (Fig 12a). It can be used to advantage to demonstrate iron deposition in areas of chronic haemorrhage in cavernomas (Fig 12b)

**Figure 13 – Chemical shift**  
 Protons in fat resonate at a slightly different frequency and phase to those in water. Since spatial localisation depends on these properties, protons in fat are slightly misregistered in space giving rise to a dark edge next to areas of fat. It is as though the fat has been shifted by a pixel or so. This image shows a dermoid with posterior soft tissue and anterior lipid layer.