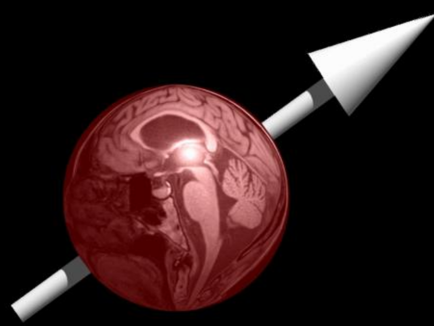
The Physics of Magnetic Resonance Imaging (MRI) Lecture 3



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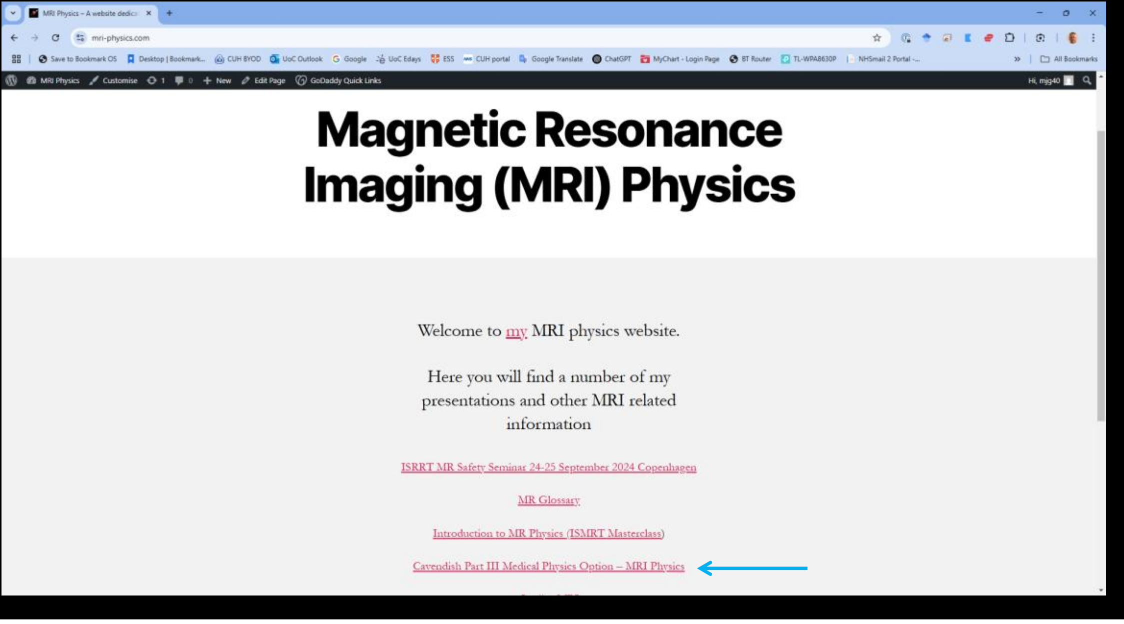
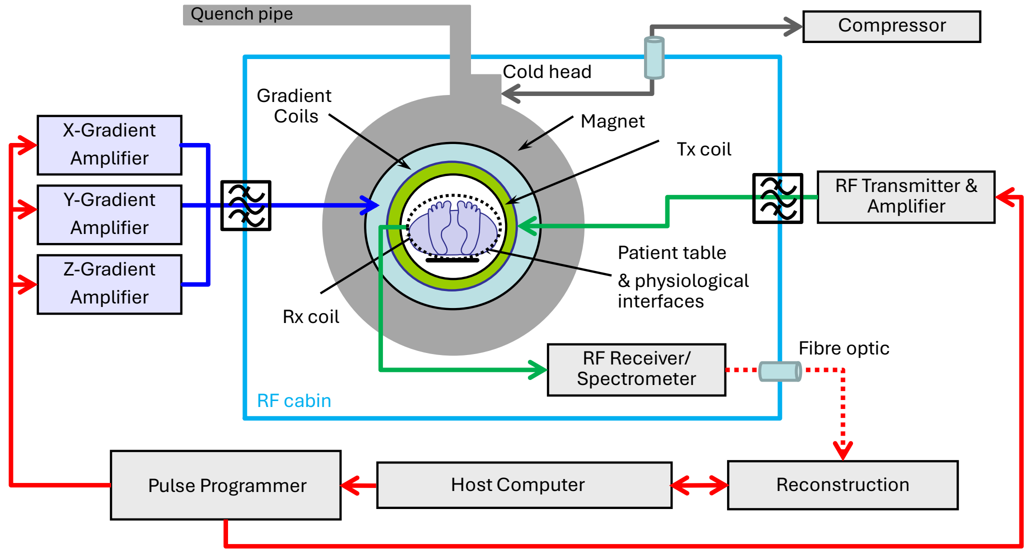


Learning Outcomes

- After these lectures you should be able to:
  - Explain how nuclear spin gives rise to magnetic resonance
  - Understand the principles of T_1, T_2 and T_2^*relaxation
  - Explain the principles of MR image formation
  - Describe the spin echo and gradient echo pulse sequences
  - Outline the basic components of an MRI system
  - Understand the safety issues related to MRI

Hardware Overview



Here is an overview of a typical superconducting MRI system. We start with the superconducting magnet here which we are seeing in cross section. The magnet contains windings of a special alloy of niobium and titanium which when it is kept at the temperature of liquid helium 4.2K or -269°C it becomes superconducting which means it has zero electrical resistance. When the service engineer first installs your magnet, he will inject several hundred amps into those windings and that current will continue to circulate unimpeded because of the zero resistance until the magnet needs to be deenergised for some reason. Because liquid helium is quite expensive there is normally a cold head or cryocooler mounted on the magnet. This is a very high-performance refrigerator that works to keep internal radiation shields very cold to minimise the helium boil. The cold head is connected, via helium lines through a waveguide, into a compressor in a separate equipment room.

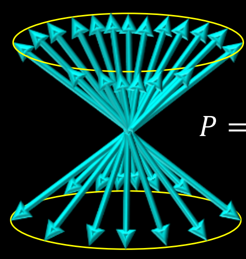
Should the magnet cease to be superconducting than the large amounts of energy that is stored in the windings will be converted to heat, the liquid helium will be vaporised and will increase the pressure within the magnet. There is a burst disc which will rupture when the pressure exceeds a certain value and then the case is helium will be vented, through the quench pipe, to the outside of the building. This is important because helium can potentially asphyxiate people since it displaces oxygen, and it is also extremely cold.

Inside the room temperature bore of the magnet we have the gradient coils connected to high power amplifiers for the X, Y and Z gradients and they have appropriately filtered electrical connexions through the penetration panel in the RF cabin. Inside the gradient coil have the whole body transmit coil, typically of a birdcage design and although it can be used reception as well it is mainly used to create a relatively homogenous transmit files inside the magnet. The transmit coil is connected to the RF transmitter and amplifier again through filtered connexions in the RF cabin Then we have the receiver coils which are generally arrays of small elements that are positioned as close as possible around the anatomy of interest on the patient and each of the individual coil elements are then connected to a RF receiver and spectrometer system where the analogue signals from the coils get digitised and put out on fibre optic playing through waveguides into the separate reconstruction computer. I have mentioned this cabin couple of times and waveguides that go through it because we are dealing with very small amplitude signals from the patient in the radio frequency part of the EM spectrum, we need to ensure that the system is isolated from any external radio frequencies which may be around about the same frequency range. The scanner itself is enclosed in the six-sided metal box or RF cabin, sometimes called a Faraday cage, that isolates the sensitive receiver coil inside from any external effects of radio frequencies and it also means that the high energy radio frequency pulses that we generate internally cannot leak out into the rest of the hospital. We have the patient table that moves the patient to the centre of the magnet known as the isocentre where the uniformity is greatest. The system also supports physiological interfaces, for example we may want to do an ECG synchronised scan, or a scan synchronised to the patient's respiratory motion. We have the host computer where the operator enters the patient demographics and scan parameters. The host computer communicates with a device called the pulse programmer that takes the geometric information like the FOV, slice thickness and calculates the desired gradient amplitudes for those parameters and handles the synchronisation between the gradient pulses and the shapes of the RF wave forms. The pulse programmer sends the gradient waveforms and controls the timings to the gradient amplifiers to create the high voltages and current required to drive the gradient coils. It also sends the digital RF waveforms to the RF transmitter and amplifier that generates the high intensity RF pulse through filtered connections through the RF cabin to the body transmit coil. The signals that are detected by the receiver coil go to an RF receiver and spectrometer that digitises the signals and converts them to an optical signal that is passed back through the RF cabin to the reconstruction computer which will perform tasks such as the fast Fourier transformation that is used to reconstruct the images, which are then sent to the host computer for display.

Magnet

- Role:
  - Nuclear polarisation (P)
  - Nuclear precession ($\omega_0 = \gamma B_0$)
- Performance specifications:
  - Field strength
    - B_0 (T)
  - Field uniformity
    - $\Delta B_0(x, y, z)$ (Hz/ppm)
  - Field stability
    - $\Delta B_0(t)$ (ppm/hr)
  - Spatial gradient
    - $\frac{\partial B_0}{\partial x}, \frac{\partial B_0}{\partial y}, \frac{\partial B_0}{\partial z}$ (T/m)
  - Fringe field (0.5mT/5G) footprint

$N_\uparrow$



$N_\downarrow$

$$P = \frac{N_\uparrow - N_\downarrow}{N_\uparrow + N_\downarrow} \approx \frac{1}{2}\frac{\hbar\gamma B_0}{k_b T} \approx 0.001\%$$

$SNR \propto B_0$ [neglecting a lot of other effects]

We will now look at the role of the individual components that make up the MRI system. Starting with the magnet. Its main role is to create this nuclear polarisation talked about before, so we know that is about a .001% difference in the spins in the low energy state compared to the high energy state that is what contributes to our net magnetisation or polarisation. We also know that the magnetic field causes nuclear precession that allows us to create transverse magnetisation that can be detected and used to form the MR image. The strength of the static magnetic field is very important to a first order approximation if we neglect a lot of other affects the signal to noise ratio in images is proportional to the strength of the static magnetic fields hence the interest from going from 1.5 tesla to 3 tesla and potentially for going to higher fields such as 7 tesla although there are lots of challenges associated with higher field imaging.

The uniformity of the magnetic field is also very important in terms of the dephasing of the transverse magnetization, so we need to ensure that the field is as uniform as possible. The field also needs to be quite stable over time so whatever the method of generating the magnetic field it needs to have a good level of temporal stability.

We also need to consider the spatial gradient of the static magnetic field. This is nothing to do with the gradients that create the images, this is that variation in magnetic field that falls off with distance from the magnet and it is this variation in field with distance that is responsible for the projectile effect of ferromagnetic objects that may be inadvertently bought into the MR environment.

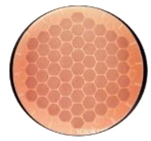
We also need to consider the fringe field from the magnet. Current MR safety guidelines state that the fringe field must not exceed 0.5 mT or 5 G in any adjacent publicly accessible areas, such as corridors or rooms adjacent to the MRI magnet. The space that contains the 0.5 mT field is known as the MR Environment. This may mean that additional static magnetic field shielding is may be required to be installed in the wall or possibly the ceiling and floor of the MRI Environment, depending on adjacent areas which may also contain equipment that is sensitive to magnetic fields, such as photomultiplier tubes in nuclear medicine gamma cameras.

Types of Magnets

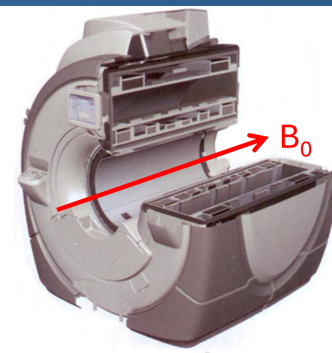
- **Field strengths**
  - Ultralow field < 0.1 T
  - Low field < 0.6 T
  - Mid-field ~ 1.5 T
  - **High field ~ 3.0 T**
  - Ultrahigh field ≥ 7 T
- **Technologies**
  - Permanent magnets
    - Neodymium Iron Boron
  - Iron-cored electromagnets
    - Copper
  - Superconducting magnets
    - Niobium Titanium (Tc ≈ 10K)
    - Magnesium Diboride (T_c = 39 K)
    - Niobium Titanium (T_c = 10 K)

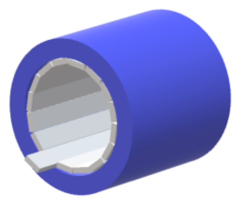
Superconducting Magnet

- Generates strong, highly uniform and stable static magnetic field (B_0)
  - 0.5 T to 10.5 T and above
  - ΔB_0 (x,y,z) < 1 ppm (40cm DSV)
  - ΔB_0 (t) < 0.01 ppm/hour
  - Actively shielded
- Steel cryostat containing approx. 2,000 L liquid helium at 4.2 K (-270 °C)
- Cryocooler minimises helium boil-off
- I = 600 A E = 15 MJ
- 3 T has approx. 60 km of superconducting niobium titanium (Nb_3Ti) filaments in a copper matrix
- Magnet is physically 'shimmed' using small steel 'shims' placed in trays by the vendor engineers when magnet in situ

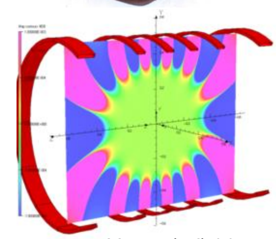


Nb_3Ti filaments in a copper matrix





Magnet showing shim trays inside



Magnetic field calculations

Looking at a bit more detail at superconducting magnets that are the only way to generate relatively strong, very uniform, and stable magnetic fields in the range from around 0.5T to 11.7 T. They have very good spatial uniformity < 1 ppm over a 40 cm diameter spherical volume (DSV). They also have very good temporal stability. To minimise the extent of the 0.5 mT fringe field the magnets are also actively shielded. The internals of a typical superconducting magnet are shown here. The magnet windings sit in a bath of liquid helium at 4.2K. There are several internal radiation shields that are mechanically cooled by the cryocooler as well as a vacuum layer and outermost there are the active shielding coils, also sitting in liquid helium. The current runs in the opposite direction in these coils and that helps to cancel out the fringe field from the main magnet coils Active shielding means that the fringe field falls off much more rapidly which makes siting the magnets easier, but it does mean that there are quite steep spatial field gradients around the magnet which increase the safety concerns.

Older magnet designs may require up to 2000 litres of liquid helium whilst newer designs which may only require a few litres. Helium is created as a by-product of radiation decay, so it is mined in a few places around the world. It is also a non-renewable resource and is quite expensive per litre. The cryocooler helps to minimise the helium boil off

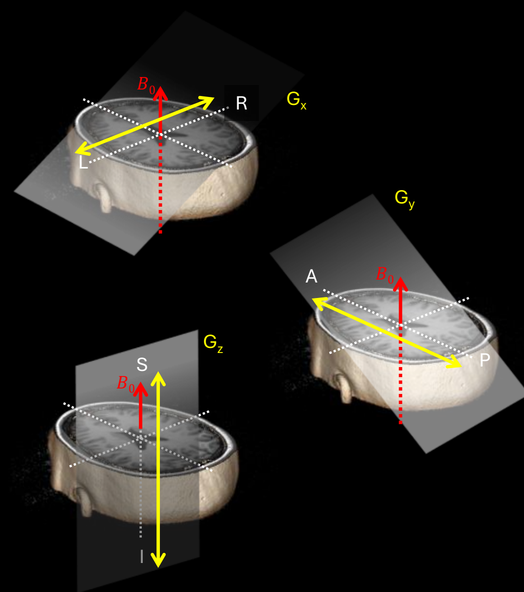
When an MRI system is installed the service engineer arrives with a power supply and slowly injects current into the windings to ramp the field. A 3 T magnet typically requires around 600 A, providing a stored energy of around 15 MJ. The power supply will then be disconnected since the windings have zero resistance the current will continue to circulate indefinitely providing the magnet remain superconducting. So, the magnets are generally never turned off except in an emergency or for some serious maintenance work.

A 3 T magnet typically has about 60 km of niobium titanium filaments in a copper matrix. The copper is there in the event of the magnet ceasing to be superconducting the stored energy will be converted to heat and the copper will conduct the heat hopefully without melting the windings.

To ensure that the magnet uniformity is as good as possible the magnet is physically shimmed by the engineer after it is ramped so that the effects of any steel within the walls or girders above or below the magnet can be taken into consideration. Inside the room-temperature bore of the magnet there will be many shim trays where shims, small steel pieces of steel, are put into appropriate locations within the shim trays. A system is used to iteratively plot the field in three-dimensions and calculate the amount and position of the shims inside the trays.

Gradients

- Role:
  - Spatial encoding, $\frac{\partial B_z}{\partial x}, \frac{\partial B_z}{\partial y}, \frac{\partial B_z}{\partial z}$
  - Motion encoding, e.g. diffusion weighted imaging
  - Unwanted signal dephasing
  - Dynamic linear shimming
- Performance specifications:
  - Peak amplitude G_{max} (mT/m) ~ 50–80 mT/m
  - Rise time $\frac{dG_{x,y,z}}{dt}$ (μs) ~ 300–250 μs
  - Linearity (%) ~ 2%
  - Eddy currents (%)
- Limitations:
  - Heat dissipation (kW) & acoustic noise (dBA)
  - Peripheral nerve stimulation

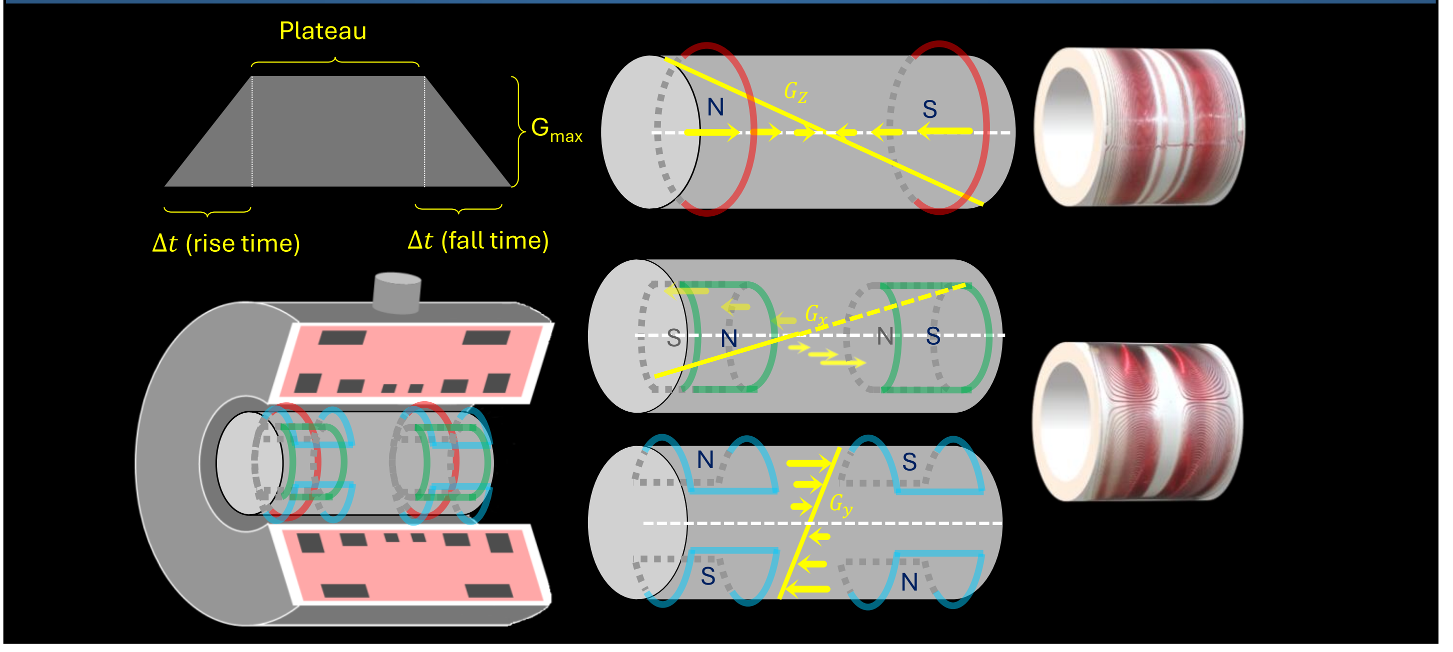


The main role of the gradients is spatial encoding, so the gradient fields need to point in the z-direction but spatially vary in the X, Y and Z directions. Gradients can also be used to encode motion, for example in diffusion weighted imaging gradient and they can also be used to dephase unwanted signals. Gradients are also used for dynamic shimming to further improve the uniformity of the magnetic field on a per patient basis.

In terms of performance specifications, we are looking at the peak amplitude achievable in mT/m, the rise time from zero to the maximum value in microseconds, the linearity of the gradients over a certain field of view and we are also looking to have minimal interactions other conducting structures such as the magnet cryostat. The gradients induce eddy currents in the cryostat that distort the waveforms.

In terms of limitations, high performance gradients require substantial currents and voltages that generate lots of heat that must be dissipated so we need a chilled water supply to the amplifiers and gradient coils themselves. Gradient switching can generate quite high levels of acoustic noise so there is a requirement to ensure that patients and anybody else who happens to be in the MR Environment during the scan has appropriate hearing protection. Finally, gradients can cause peripheral nerve stimulation that can range from a tingling sensation to, albeit very rarely, painful muscle contractions.

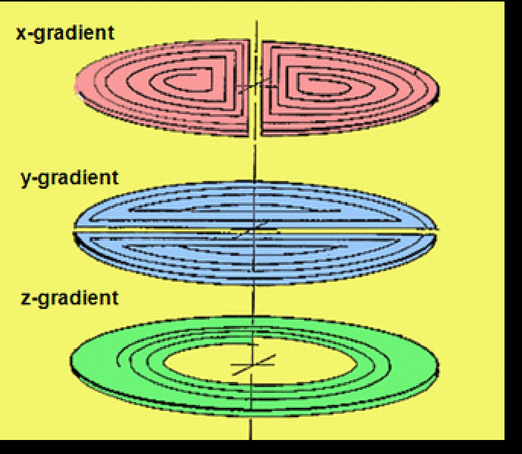
Gradient Coils



Gradient pulses are generally trapezoidal in shape and the gradient coils necessary to create them sit inside the room temperature bore of the magnet. The Z gradient is created using a Maxwell pair, whilst the X and Y gradients are created using a quadrupolar design, otherwise known as a Golay configuration, each rotated 90° with respect to each other. Modern gradient coils are considerably more complex in terms of their design, so our simple Maxwell pair are overwound at either end, whilst the Golay configuration has these typical kind of fingerprint designs. These designs improve the linearity and efficiency of the gradient coils.

We need very high-power amplifiers to generate the gradient pulses and we are typically looking for amplifier power on the order of one to three megawatts. We need very high voltages to overcome the back-emf in switching the gradient on so that the rise time is as short as possible and large currents to maintain high amplitude pulses. The gradient amplifiers and coils generate large amounts of heat, so we need to dissipate somewhere between 25 and 75 kW, so the amplifiers and gradient coils are water cooled. The gradient conductors are tubular so the coolant can flow directly through the windings improving the heat dissipation.

Vertical Field System Gradients



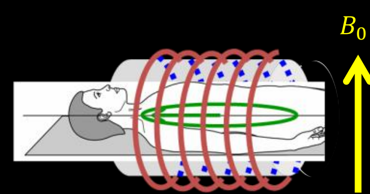
http://mri-q.com/open-scanner-gradients.html

Vertical or horizontal field systems have planar gradients built into the pole pieces at the top and the bottom, So the vertical Z-gradient is a Maxwell pair with the currents running in opposite directions and the X and Y gradients are created using these planar Golay designs. The gradient performance for these systems is generally inferior to what can be achieved on superconducting systems.

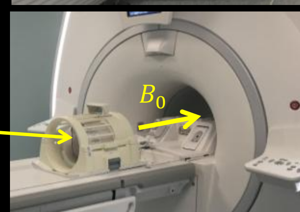
RF Transmit Coils

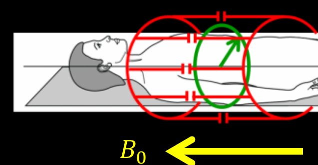
Coil design depends upon the orientation of B_0 compared to the subject

Vertical field magnet
solenoid coil
e.g. solenoid head coil



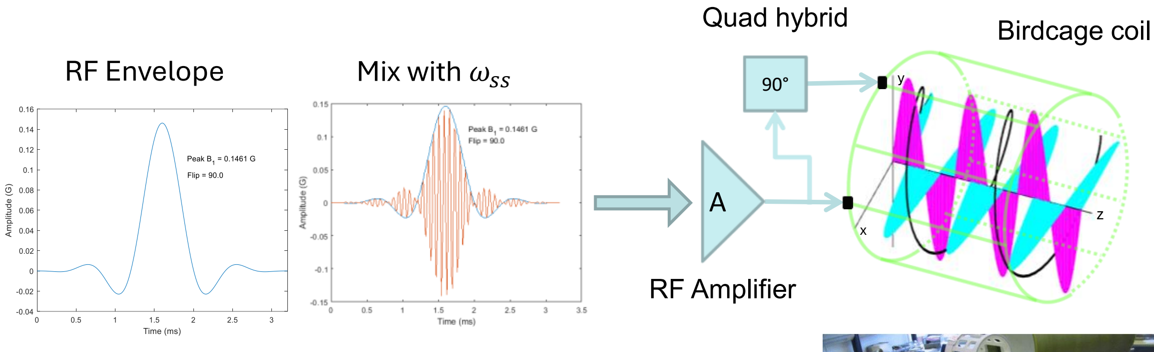
Horizontal field magnet
birdcage coil
e.g. birdcage head coil

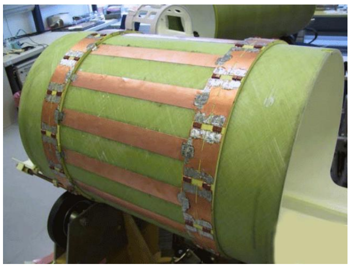




Looking at the RF frequency side of things the design of the transmit coil depends very much on the orientation of B_0 compared to the subject. Remember that B_1 needs to be orthogonal to B_0. For example, if we have a vertical field magnet then we can generate the rotating B_1 field using a solenoidal coil whereas in a horizontal field superconducting magnet where the coil is in the same plane as the patient then we need to generate our rotating B_1 field using a so-called birdcage coil.

RF Excitation





This is the RF excitation arrangement for a birdcage coil inside a superconducting magnet. The RF envelope dictates the bandwidth of the associated slice sickness. The envelope is then multiplied by the Larmor frequency for the desired slice. This pulse is applied to the input of the RF amplifier. To create a circularly polarised B_1 field the output of the amplifier drives the birdcage at two points 90° physically apart and with a 90° phase shift performed using the quad hybrid component. The picture shows a typical 16 rung birdcage body coil that is positioned inside the gradient coil assembly.

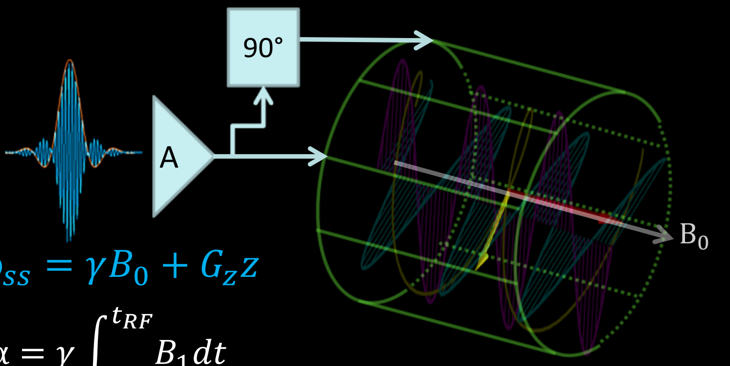
Radiofrequency B_1^+ (Transmit)

- Role:
  - Nutate (tip) net magnetisation
- Performance specifications:
  - Peak amplitude

    B_{1max} (μT)
  - Transmit field uniformity

    ΔB_1^+ (%)
- Limitations
  - Power deposition

    SAR (W/kg)



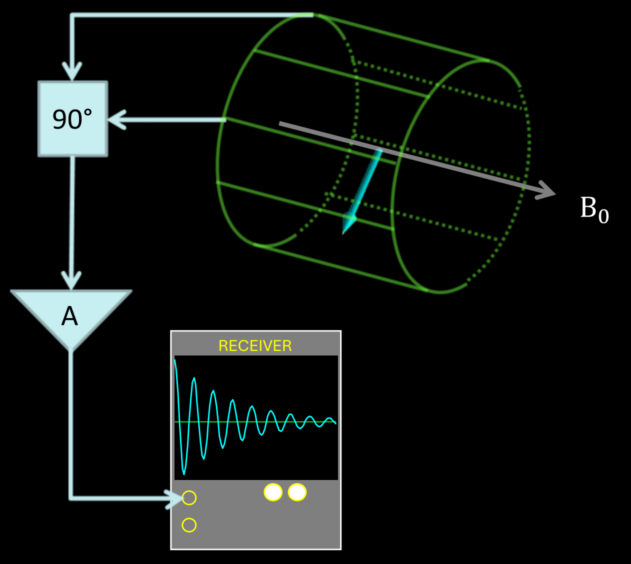
$$\omega_{ss} = \gamma B_0 + G_z z$$

$$\alpha = \gamma \int_0^{t_{RF}} B_1 dt$$

The requirements of the RF transmit side or B_1^+ is to nutate or tip the net magnetisation. We establish a circularly polarised B_1 field inside our birdcage coil, shown as the yellow arrow in the laboratory frame of refence. Because that frequency is resonant with the precessional frequency at the desired slice location z, then whilst the slice select gradient is applied, we can tip the net magnetization vector into the transverse plane. The nutation angle is given by the integral of the RF pulse envelope. In terms of performance specifications, the system is characterised by the peak RF amplitude that is available terms of G/μT and the uniformity of the transmit field within the coil, that can be expressed as a percentage of the nominal flip angle. The main limitation of the RF system is the maximum power, that can be deposited in the patient, because of the eddy current induced heating in the conducting tissues caused by the B_1 field. This power deposition is known as the specific absorption rate (SAR) and is limited as part of the MRI system design specification. We shall discuss SAR in more detail later.

Radiofrequency (Receive)

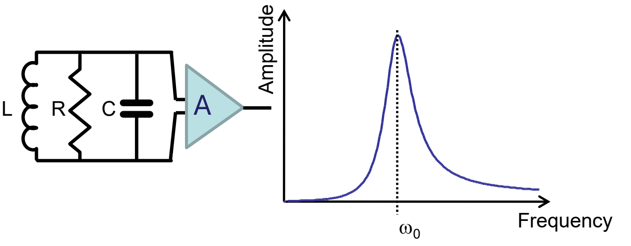
- Role:
  - Detect magnetisation
  - Demodulate/digitise signal
- Performance specifications:
  - Filling factor
  - Receive field uniformity
  - Number of receiver channels
  - Noise amplification

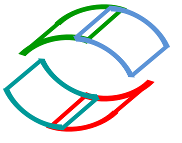


The role of the radiofrequency receive system is to detect the precessing transverse magnetisation that induces a small voltage in the coil. This signal is then amplified. In this case we are displaying the free induction decay signal. The performance specifications of the receiver coils are firstly that they need to have a very good filling factor. The coil needs to fit the anatomy of interest as closely as possible to maximise the signal and therefore maximise the signal to noise ratio. Ideally the received field should be uniform but the trend these days is to have multi element array coils that are very good at detecting the signal very close to the elements but not quite so sensitive to signals deeper in the body. However, there are several other advantages from having dense arrays of smaller sized coil. However, each coil element must be connected to a separate receiver channel, and the images from each receiver are finally combined. The receive chain should also minimise the amount of noise amplification. This has led to manufacturers digitising the signals closer to the coil elements, and in some cases including the analogue-to-digital converters into the coil design. and then converting to optical signals for transfer to the reconstruction computer.

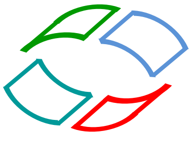
Coil Basics

- Coils are resonant tuned circuits
- A coil has a depth sensitivity approx. equal to its radius
- Arrays are combinations of individual coils that cover a large area/volume but minimally interact
- Array geometries are optimised for parallel imaging





Original phased array design

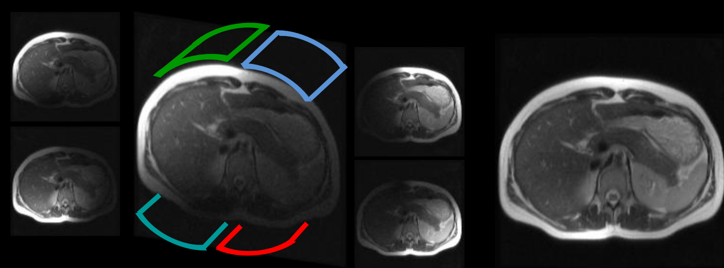


Parallel imaging optimised array

The basics of any coil whether that is for transmit or receive is that the coils are resonantly tuned circuits so the coil itself will have inductance and resistance but we the add additional capacitance to ensure that it has a frequency response near the Larmor frequency. The capacitance also ensures that the transmit coil has a matched 50 Ω input impedance to allow proper power transfer. Coils will have a relatively low Q to ensure that they are still resonant for a range of patient loadings. Receiver coils are generally made up of several small individual coil elements. Each coil element is sensitive to a small volume of interest thereby minimising the amount of noise that is detected but the use of multiple elements ensures that we have the desired patient coverage. The original coil arrays known as phased array designs required that the coil elements be appropriately overlapped so there was minimum mutual interaction between coil element so that the response from the individual coils just did not appear to come from one large single coil. As the number of elements in coils has increased then there is generally some spatial separation between them. This variation in spatial sensitivity profiles allows for advanced acquisition and reconstruction methods that are used to accelerate image acquisition.

Types of Receive Coil

- Example multi-channel coils
- Head
- Neck
- Spine
- Knee
- Foot/ankle
- Wrist/hand
- Shoulder
- Cardiac
- Breast
- Torso
- Abdominal
- Endo-rectal
- Flexible

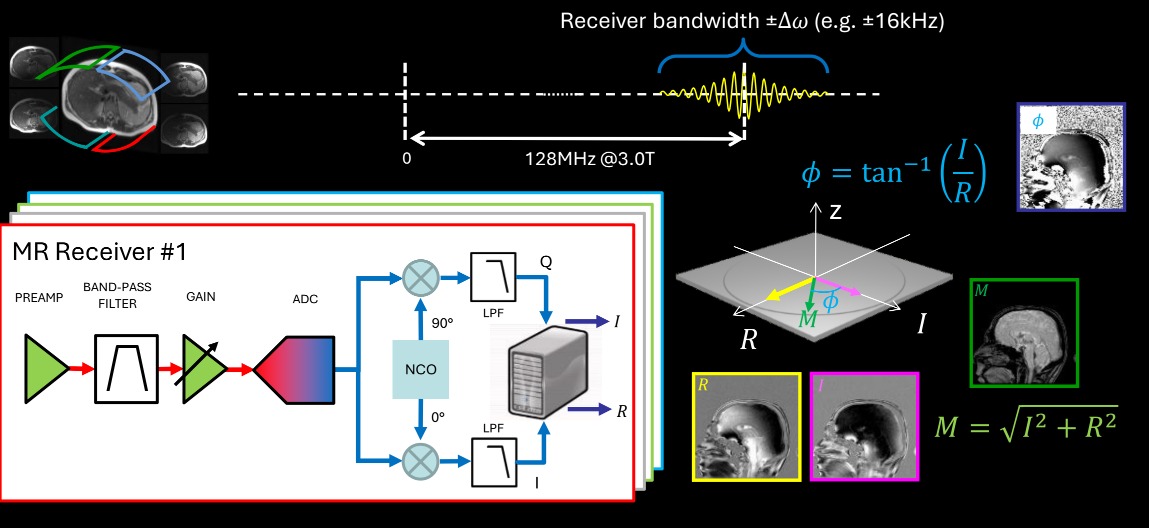


Individual coil images

Sum-of-squares combination

There are a wide variety of different receiver coils with different numbers of channels. Here is a list of potential coils that you could purchase from your MRI vendor. This is an 8-channel breast coil, an eight-channel brain coil, 12 channel abdominal coil and a 32-channel cardiac coil Each coil is connected to its own receiver system, so images are reconstructed from each coil and then combined using an algorithm like a root sum of squares. In this diagram of a four-channel array we can see the individual coil images and the final combined image. Note how the individual coil images have a spatial sensitivity that falls off with distance from the coil, but this is compensated for during the image reconstruction process.

Receiver (x N-Channels)



This is the typical architecture of an MR receiver channel, and we have one of these receiver channels for every element that makes up our coil array. In this case we have only four elements and 4 receiver channels.

Following low noise preamplification, bandpass filtering and variable gain the analogue signal is digitised. The range of useful frequencies that are encoded in the MRI acquisition process is in this small audio frequency range, e.g., ±16 kHz. However, to make MRI work these frequencies are centred on the much higher Larmor frequency, e.g., 128 MHz at 3 T. Therefore, the signal must be demodulated to eliminate this carrier frequency. We are then left with a ±16 kHz audio-frequency signal. The demodulation is performed digitally in quadrature, so we end up with complex raw-data known as in-phase and quadrature. Following Fourier transformation, we also get complex data known as the more conventional real and imaginary components.

It is perfectly possible to calculate real, imaginary and phase images however we generally just deal with the magnitude since as you can see there are phase errors that would also require correcting.

Signal-to-noise Ratio (SNR)

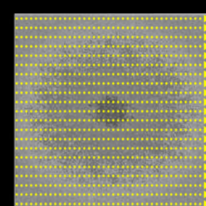
$$SNR \propto \frac{\Delta x \cdot \Delta y \cdot \Delta z \cdot \sqrt{N_x \cdot N_y \cdot NSA}}{\sqrt{RBW}}$$

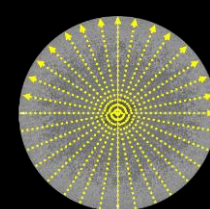
- $\Delta x, \Delta y, \Delta z$ are the pixel sizes in x, y and z directions. Δz is usually the slice thickness
- N_x, N_y are the number of samples in the x and y directions
- NSA is the number of signal averages
- *RBW* is the receiver bandwidth
- Remember that some parameters are interdependent, e.g., changing the matrix size for a fixed field-of-view will change the acquisition time as well as the pixel size

The signal-to-noise ratio in MR images is dependent on several factors including the relaxation times and pulse sequence used to acquire the data, but this equation shows some of the other acquisition factors. The SNR is proportional to the size of the imaging voxel, the square root of the number of samples in the frequency and phase encoding direction and the square root of the number of signal averages, i.e., simply repeating the acquisition, adding the data and dividing by the number of averages. The SNR is also related to $1/\sqrt[2]{\text{Receiver bandwidth}}$

Reconstruction

- Role:
  - Manage raw data
  - Reconstruct images
  - Inline post-processing
- Performance specifications:
  - Raw data capacity
  - Reconstruction and post-processing algorithms
  - Reconstruction speed





The final component of the MR system hardware is the reconstruction engine, and its role is to manage the raw data particularly as high-resolution multi-coil data may generate large volumes of data. The reconstruction software will be optimised to run multidimensional fast Fourier transforms, as is needed for the multi-coil 3D acquisition of over 200 slices. The recon system will also have to perform the multi-channel image combination as well as any other additional image processing that is required. Some of the performance specifications we should be looking are the capacity to hold large volumes of raw data, the range of reconstruction algorithms that are available including not only conventional Cartesian, line by line, acquisitions but the increasing use of non-Cartesian trajectory's such as this radial acquisition where the data requires regridding onto a regular grid prior to Fourier transformation. The reconstruction and post-processing speed is governed by whatever hardware processors are incorporated into the system.

Active Hardware Development

- Magnets
  - Low cryogen volumes, e.g. ~ 0.7 L
  - Ultra-high field strengths (11.4 T +)
  - Ultra-low field strength (64 mT)
- Gradients
  - Higher performance with wider bore
  - Very high-performance head-only systems
- RF
  - Higher number of receiver channels
  - Lighter and more flexible coils
  - High-order shims integrated into coils
  - Wireless coils
- Acquisition and Reconstruction
  - More sophisticated acquisition and reconstruction algorithms including ML/DL
  - Open-source reconstruction tools
- Combined/hybrid modalities
  - PET/MR
  - US/MR
  - MR-Linac
  - Intraoperative MRI

This slide is to give you some indication of the current trends and developments in commercial MRI systems. One vendor is now offering 1.5T superconducting magnets that only requires approximately 7 L of liquid helium to remain superconducting. Which makes siting the system somewhat easier as you do not need to have a quench pipe. There are also research and development activities to develop and build ultrahigh field magnets at 10.5 T and above. Conversely an ultra-low field (64 mT) portable MRI system has recently come on to the market allowing the MRI scanner to be taken to critically ill patients that cannot be moved. In terms of gradients vendors are working on providing higher performance gradients with wider bore size scanners. Since the power required to drive a gradient coils scales with the 5th power of the radius these energy requirements are substantial. One way to obtain higher performance imaging just for the brain is through the development of head only magnets and gradient coils which can achieve extremely high gradient performance but over a limited field-of-view. RF coils are increasing in channel density and vendors are now developing very light and highly flexible coils almost like blankets, or potentially wearable coils. Optimisation of the static magnetic field may be improved, particularly in the head, by also using the coil elements for high-order active shimming. There is also active work on developing wireless coils without the bulky cables and connectors that are currently used.

In terms of acquisition pulse sequences are becoming more sophisticated in terms of image contrast and reducing acquisition times. These methods also require more sophisticated reconstruction algorithms using iterative algorithms and machine learning and deep learning to improve SNR and reduce artefacts. There has also been a significant amount of work in recent years in the development of open-source reconstruction tools. MRI is starting to be combined with other techniques to create hybrid modalities. such as simultaneous PET/MR using solid-state PET detectors that are insensitive to the magnetic field. Combined US/MRI systems have been developed to leverage the speed of ultrasound with the soft tissue contrast of MRI. MRI is also becoming part of therapy with several combined MRI/linear accelerator systems in clinical use enabling organs to be tracked in near real time to allow even more precise radiotherapy. Finally, the use of MRI for neurosurgical intra-operative imaging is now well established. An MRI system is installed in the operating theatre and the surgeons can rapidly move the patient from the surgical area into the scanner to check that their resection margins are appropriate at the time of surgery.

Safety of Magnetic Resonance Investigations

"MRI" has killed patients and staff

An important part of MR Physics is understanding the safety issues associated with MRI. MRI scanners have killed both patients and staff as well as causing life changing injuries. It is unfair to say that MRI scanners have killed people obviously it has been staff members who have misused the systems or allowed unsafe persons, or patients with unsafe implants to enter the MR Environment, who are responsible for these deaths and injuries.

Oxygen Cylinder vs. Magnet

Boy, 6, Dies Of Skull Injury During M.R.I.

By DAVID W. CHEN
Published: July 31, 2001

Outside of the X-ray, perhaps no other medical examination is as well known or as safe as the magnetic resonance imaging test, which is conducted eight million times a year in the United States on patients ranging from people with brain tumors to famous athletes with knee injuries.

But today, officials at the Westchester Medical Center announced that something went horribly wrong on Friday with an M.R.I. test on a boy, 6, who had just undergone surgery. Even though no metal objects are supposed to be in the testing area, because they will be pulled toward the 10-ton machine by its powerful electromagnet, a metal oxygen tank somehow made it into the examination room.

The tank, about the size of a fire extinguisher, became magnetized, then flew through the air at 20 to 30 feet per second and fractured the boy's skull.

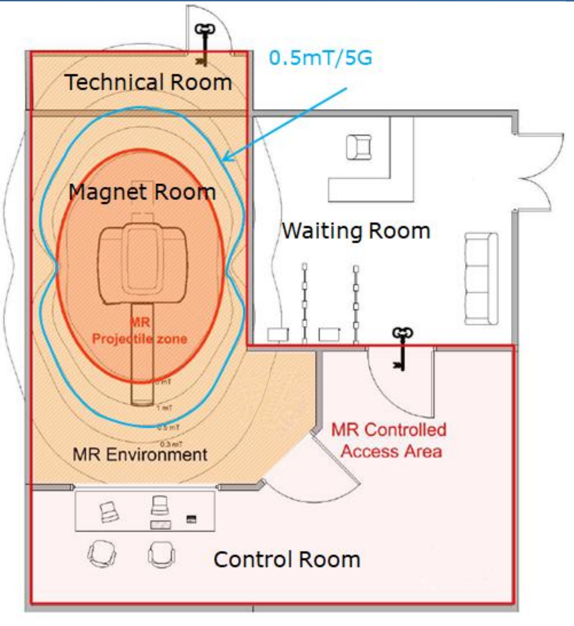
$2.9 Million Settlement Closes Colombini MRI Death Case

This week the settlement documents were released — closing the chapter on the lawsuit that arose from the seminal event in MRI safety, the 2001 oxygen tank fatality of then-six-year-old Michael Colombini.

Nearly nine years after the accident, the lawsuit was settled for $2.9 million, a settlement that was likely both diminished by, and made possible by, a pre-trial motion which excused GE Healthcare as a defendant to the suit.

The seminal event for MRI safety was the death of 6-year-old Michael Colombini over 20 years ago . Michael had had a brain tumour resected and was brought down to the MRI scanner for a post-operative scan under sedation. During the scan Michaels oxygen saturation started to drop and the anaesthetist called for oxygen. Unfortunately, the piped oxygen in the room was not working and the MR technologists went off to reset the system. whilst they were away a nurse entered the MRI facility and heard the anaesthetist calling for oxygen. she saw an oxygen cylinder in an adjacent room and bought it into the magnet room. As the article says the tank about the size of a fire extinguisher became magnetised then flew at 20 to 30 ft/s and fractured the boy’s skull. Unfortunately, Michael subsequently died from his injuries. Nine years later the case was settled for $2.9 M dollars. The movie shows the effect of an oxygen cylinder being taken close to a 1.5 tesla magnet.

MR Suite Layout



MR Environment
A volume containing the full extent of the 0.5 mT (5 G) magnetic field around the MRI scanner.

MR Controlled Access Area
A locally defined area of such a size to contain the MR ENVIRONMENT. Access shall be restricted and suitable warning signs should be displayed at all entrances.

MR Projectile Zone
A locally defined volume containing the full extent of the 3 mT (30 G) magnetic field contour, or other appropriate measure, around the MRI scanner

In the UK MRI safety guidance from the Medicines And Healthcare products Regulatory Agency (MHRA) recommends that the MR suite be demarcated by three areas. The MR environment is a volume that contains the full extent of the 0.5mT magnetic field around the MRI scanner. Note that this volume can extend into other rooms such as the MR technical room at the back. A MR controlled access area is also defined that contains the MMR environment. Access to the MR controlled access area is restricted and only accessible through a minimum number of self-locking doors and suitable warning signs should displayed at all entrances. there is also I defined Mo projectile zone which contains the full extent of the 3mT magnetic field contour. This is the field at which you would expect a ferromagnetic object to start feeling some force of attraction or torque. Only certain equipment that is deemed MR safe or MR Conditional with the specific conditions stated all allowed inside the MR environment.

MR Safety Definitions

- ASTM F2503-08
  - MR safe = safe in the MR Environment under all conditions
  - MR conditional = safe in the MR Environment providing specific conditions can be met
    - Must be labelled with conditions by manufacturer
    - e.g. static field 3 T and a static field gradient <7.2 T/m
  - MR unsafe otherwise

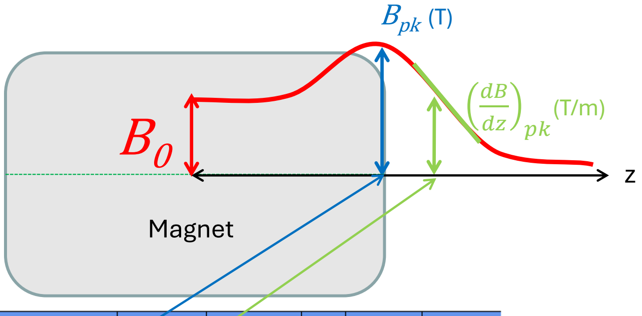
Any equipment that needs to be taken into the MR Environment needs to be labelled as MR Safe or MR Conditional, where the conditions for use can be met in that facility. All other items in the Controlled Access Area that are potentially moveable should be labelled as MR unsafe.

This patient monitoring system has an MR conditional label affixed to the base. Such labelling also extends to devices implanted in the patient whether they be passive such as a hip replacement or active such as a neuro stimulator.

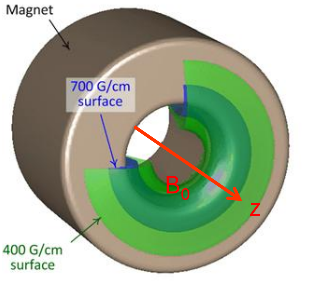
B_0 Spatial Field Gradient $\frac{dB}{dz}$

F = translational force on object
χ = magnetic susceptibility of object
V = volume of object
B = static field

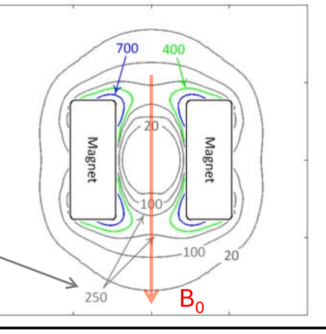
$$F \propto \chi V B \frac{dB}{dz}$$



Parameter	Radial Location R(m)	Location along Z(m)	B(T)	Grad(B) (T/m)	max(B)* grad(B) (T^2/m)
Peak B	0.35	0.64	3.9	7.2	28.2
Peak Gradient	0.51	0.92	1.8	12.4	22.6



In this example, there is no path into the magnet that does not pass through the “entry SG” contour of 250 G/cm (2.5 T/m)



The main risk for ferromagnetic projectile incidents s is the spatial gradient field or how the magnetic field decreases with distance from the magnet. The use of actively shielded magnets means that this field can reduce quite rapidly with distance. The ferromagnetic attraction force is proportional to the magnetic susceptibility of the object, its volume, the static field strength if it is magnetically unsaturated and most importantly the rate of change of the static magnetic field with distance from the magnet. MRI manufacturers are now required to provide plots of this spatial gradient with position and measurements of the distance from the centre of the magnet to the peak field, the peak gradient, and the product of the peak field and peak gradient. Therefore, it is this spatial gradient that dictates the safety of certain MRI conditional devices or patients with implants going into the magnet .

Recent MRI Safety Incidents*

July 2025

Man dies after weight-training chain around neck pulls him into MRI machine

Keith McAllister had approached machine after wife called for help, and was sucked in by device's magnetic force

May 2017

A woman dies after a resonance 'blows' her morphine pump

The patient had the device for the administration of this opiate four years before because of her back pains

January 2018

Man dies after being sucked into MRI scanner at Indian hospital

Man was carrying oxygen cylinder which was pulled by machine's magnetic force and then thought to have punctured

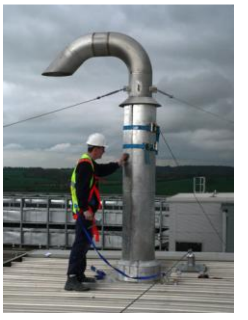
October 2021

South Korean man dies after oxygen cylinder sucked into MRI machine

The 60-year-old patient was hit in the head by a 60kg oxygen cylinder while undergoing a magnetic resonance imaging scan at a hospital in Gimhae

Magnetic Field Quench

- A superconducting magnet can contain up to 2,000 litres of liquid helium (-269°C)
- If the magnet ceases to be superconducting the huge energy stored (10+MJ) is converted into heat. This is why the Nb_3Ti filaments are embedded in copper to avoid destroying the magnet
- The liquid helium is vaporised with a 754:1 expansion ratio (1.5M litres)
- The very cold gas is vented to the outside through the quench pipe
- In a double fault situation, the gas could theoretically fill the magnet room
- Risk of cold contact burn, asphyxia and explosion risk from liquified oxygen



Quench pipe

We have already discussed magnetic field quench this is this can happen when the magnet suddenly ceases to be superconducting. the energy he stored energy is converted to heat and the liquid helium he is vaporised with a 754:1 expansion ratio. When this happens the pressure inside the magnet rises and the burst disc ruptures, and the helium vapour is vented to the outside through the quench pipe. There have been situations where the helium gas has vented into the magnet room and in at least one situation the pressure overload in the room prevented the inward opening magnet door from being opened. The helium gas is extremely cold therefore there are risks of cold contact burns, helium displaces oxygen so there is also a risk of asphyxia and there is also a potential explosion risk because the temperature is cold enough to liquefy oxygen from the air.

Gradient Safety

- 100 times weaker than the static magnetic field
- Switch about 0-5kHz
- Peripheral nerve stimulation
  - Ranging from a tingling sensation to painful muscle spasms
  - Avoid creating large loops
- Acoustic noise
  - Can exceed 100dBA
  - Hearing protection is necessary

The magnetic field gradients are about 100 times weaker than the static magnetic field and switch in the region from 0 to 5 kHz. Gradient scanning juice peripheral nerve stimulation which can range from a mild tingling sensation too painful muscle spasm a bit like a little light tasering. Patients should therefore not have their hands clasped so as to create large loops In which currents can be induced. The other safety concern with gradient switching is the acoustic noise that is generated, which for certain pulse sequences can exceed 100 dB(A). It is therefore necessary to ensure that anybody inside the magnet room during imaging has suitable hearing protection.

Radiofrequency Safety*

- RF magnetic fields are 10,000 times weaker than the static magnetic field
- Switch between 8.5 to 130 MHz depending upon B_0
- Ampère −Maxwell law $\nabla \times \mathbf{B} = \mu_0(\mathbf{J}_c + \mathbf{J}_d) = \mu_0\left(\sigma\mathbf{E} + \varepsilon\frac{\partial\mathbf{E}}{\partial t}\right)$
- Major heating arises from conduction currents ($\mathbf{J}_C = \sigma\mathbf{E}$) and P = $\mathbf{J}_C \cdot \mathbf{E} = \sigma|\mathbf{E}|^2$
- Significant heating from dielectric losses P = $\omega\varepsilon''|\mathbf{E}|^2$ Complex permittivity $\varepsilon = \varepsilon' - i\varepsilon''$
- Minor heating due to induction, Maxwell-Faraday equation $\left(\nabla \times \mathbf{J}_{eddy} = -\sigma\frac{\partial\mathbf{B}}{\partial t}\right)$ and P= $\mathbf{J}_{eddy} \cdot \mathbf{E}$
- Whilst tissue heating is strictly controlled, anything electrically conducting can heat up significantly. Need to be very careful with passive and active implants

The other major risk of serious injury in MRI is associated with the radio frequency field. Although the RF magnetic fields are 10,000 times weaker than a static magnetic field, they do switch between 8.5 to 130 MHz depending upon the strength of the static magnetic field.

1. Conduction (Resistive) Heating is caused by electric fields driving free charge carriers through a resistive medium. In MRI this is the dominant contributor to SAR and strongly depends on tissue conductivity σ.

2. Dielectric heating arises because polar molecules, especially water, are forced to reorient by the RF electric field. Their response lags the field, and that continual lag dissipates energy as heat. The energy loss arises from the imaginary part of permittivity $\varepsilon = \varepsilon' - i\varepsilon''$. Dielectric heating increases strongly with field strength
3. Inductive heating is caused by electric fields induced by time-varying magnetic fields, even when no voltage is applied directly. In MRI this occurs during gradient switching and RF transmission. It is responsible for peripheral nerve stimulation and additional SAR. It is not a separate heating mechanism but is a secondary route to conduction heating. The expression $P = \mathbf{J}_{eddy} \cdot \mathbf{E}$ gives the local power density converted into heat by eddy currents induced by time-varying magnetic fields, describing inductive heating in conductive structures such as the RF shield, gradient hardware, and implants.

Specific Absorption Rate (SAR)

- Faraday's law gives $EMF \propto \frac{dB}{dt}$
- Electric fields produced in conducting tissues will drive electric currents and result in tissue heating
- For a volume conductor with induced current density $\mathbf{J}$ and induced electric field $\mathbf{E}$ the power is given by $P = \mathbf{J} \cdot \mathbf{E} = \sigma E^2$, where σ is the conductivity (Sm^{-1})
- The specific absorption rate (SAR) (Wkg^{-1}) is defined as $SAR = \frac{\sigma|E|^2}{\rho}$, where ρ is the tissue density
- Assuming a spherical object of radius R with a uniform conductivity
- Integrating over the spherical volume

$$\mathrm{SAR} = \int_{r=0}^{R} \int_{\theta=0}^{\pi} \int_{\phi=0}^{2\pi} \frac{\sigma|E|^2}{\rho} r \, dr d\theta d\phi$$

$$\mathrm{SAR} = \frac{\pi \sigma \omega_o^2 B_1^2 R^5}{15}$$

- Importantly,
  - SAR increases with ω_o^2 i.e., static field strength squared (B_0^2)
  - SAR increases with B_1^2 i.e., flip angle squared
  - SAR increases with patient size

Less than 1% of the energy dissipated in a patient is used to tip the nuclear magnetization. The vast majority produces nothing but undesirable tissue heating. The specific absorption rate or SAR in W/kg is defined as the conductivity multiplied by the induced electric field squared divided by the tissue density. The SAR for a particular pulse sequence is estimated by the MRI system manufacturer, whilst the limits are defined in an international standard IEC-60601-2-33. The right-hand side of the slide shows an estimate of the power deposition in a spherical object with a uniform conductivity. The important relationships or that the power deposition increases with the square of the static field strength. The power deposition also increases with the square of the B_1 field or flip angle, and with the size of the patient. Since SAR is measured in W/kg it is important that the correct patient weight is entered into the scanner when the patient is registered.

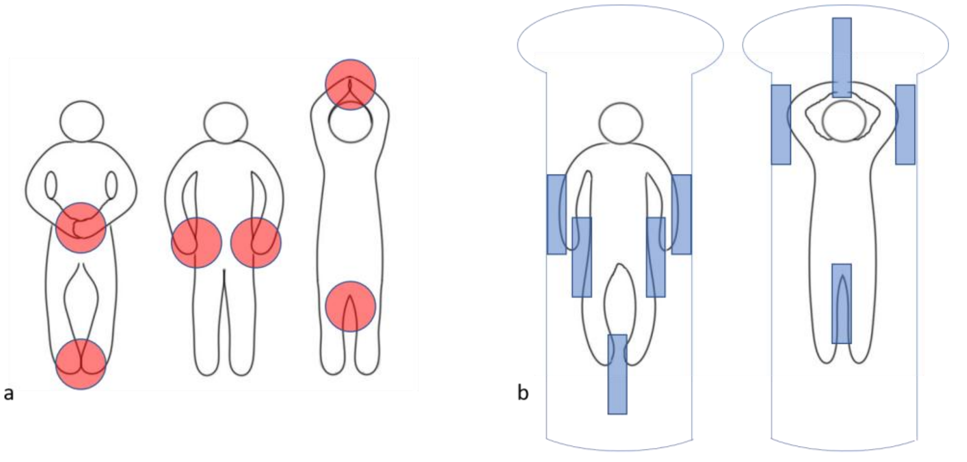
RF Exposure Limits (IEC 60601-2-33)

	Whole body SAR	Partial body SAR	Head SAR	Local SAR (a)		
Body Region →	whole body	exposed body part	head	head	trunk	extremities
Operating Mode ↓	(W/kg)	(W/kg)	(W/kg)	(W/kg)	(W/kg)	(W/kg)
Normal	2	2 - 10 (b)	3.2	10 (c)	10	20
1st Level Controlled	4	4 - 10 (b)	3.2	20 (c)	20	40
2nd Level Controlled	>4	>(4 - 10) (b)	>3.2	>20 (c)	>20	>40
Short duration SAR	The SAR limit over any 10 s period shall not exceed two times the stated values					

The SAR is averaged over a 6-minute period

IEC 60601 which is part of a series of international standards for medical electrical equipment covering the basic safety and essential performance for both equipment and systems. Part 2 - 33 covers the particular requirements for the safety of magnetic resonance equipment for medical diagnosis. The table shows the or if exposure limits for MRI systems. They can be operated in either the normal, first level or very rarely second level-controlled modes. For example, in normal mode the whole-body SAR cannot exceed 2 W/Kg. The manufacturers will provide some form of SAR monitoring which shows both the 10s short duration and standard 6-minute averaged SAR. if the acquisition exceeds either of these SAR values then the scan is immediately stopped.

Skin-skin and bore contact burns



(a) Potential location of skin-to-skin contact burns. (b) Thick insulated pads of at least 2cm thickness should be used to avoid skin-to-skin points of contact as well as contact with the bore of the MR system

RF burns have been particularly noted where there is a small area of skin-to-skin contact. The burns are generally due to the limbs creating large diameter loops and where the loop is closed becomes the point of greatest electrical resistance. Interestingly, unlike touching a hot object, these burns are not happening at the skin but in the fatty tissue immediately blow the skin where there are no pain sensors. Patients could be burned but until that burn damage has an opportunity to propagate through to the skin there is no sensation of pain and no initial observable evidence of a burn. Therefore, it is recommended that thick insulated pads should be used to avoid skin to skin points of contact. In addition, the capacitors in the body coil can also create very strong electric fields so the patient should have padding to avoid direct contact with the bore of the MR system

An MR "Warming" Event

Several hours after MRI

One week later

Here is an example of an MR Warming incident that was not reported by the patient at the time of the scan but several hours later. One week on we can see that this has now turned into a first-degree burn. It is likely that this patient was not properly padded as shown in the previous slide.

Metallic Microfibre Clothing

Published March 1, 2012 as 10.3174/ajnr.A2827

TECHNICAL NOTE

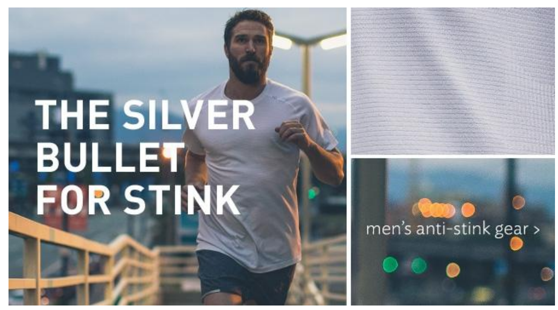
Invisible Metallic Microfiber in Clothing Presents Unrecognized MRI Risk for Cutaneous Burn

J.A. Pietryga
M.A. Fonder
J.M. Rogg
D.L. North
L.G. Bercovitch

SUMMARY: We report a case of a thermal burn that occurred during MR imaging likely caused by invisible silver-embedded microfibers in the fabric of an undershirt. As the prevalence of fabric containing nondetectable metallic microfiber increases in athletic and "tech" clothing, the importance of having patients change into safe facility-provided garments before MR imaging is emphasized.

ABBREVIATIONS: ACR = American College of Radiology; SAR = specific absorption rate; SMF = silver microfiber

Fig 1. A linear erythematous blistering eruption is noted on the patient's right flank minutes after completion of the MR imaging of her brain and spine.



Burns are now starting to be reported from patients who attend for an MRI wearing their fitness clothing that has invisible silver microfibers woven into the material. Silver is reported to stop the odour causing bacteria that occur when you sweat from reproducing, however it also nicely conducts the MRI induced eddy currents resulting in injury.

FDA Adverse Events

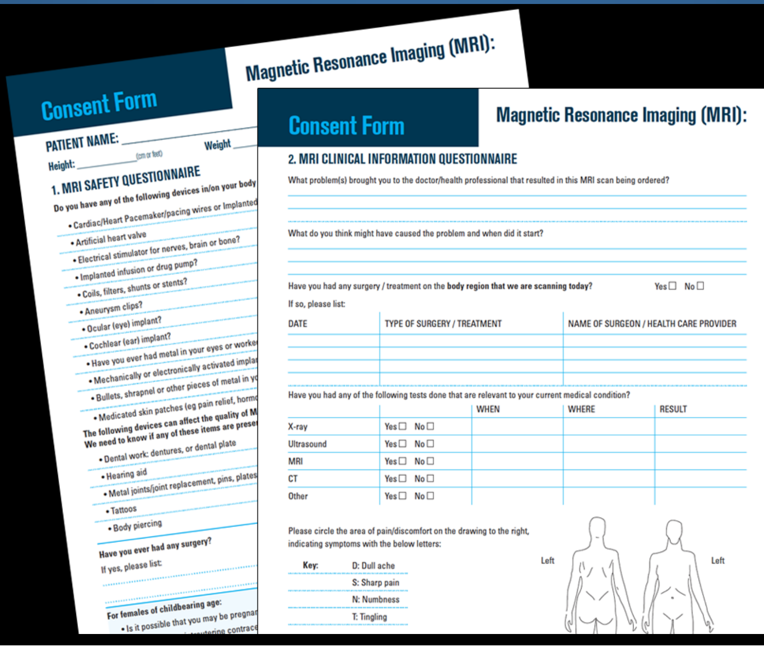
MRI Adverse Event Category	Number (%)
Thermal	906 (59)
Mechanical	170 (11)
Projectile	133 (9)
Miscellaneous	109 (7)
Image Quality	89 (6)
Acoustic*	86 (6)
Unclear	55 (4)
Peripheral Nerve Stimulation	0 (0)

* Hearing loss and/or tinnitus (temporary or permanent)

Delfino, J. et al MRI-related FDA adverse event reports: A 10-yr review. Medical Physics 2019

This slide summarises a 10-year review of all MRI related adverse events reported to the US Food and Drug administration. As you can see just under 1,000 thermal injuries were reported with projectile injuries only accounting for 133 events. Eight six events were related to acoustic injuries which included both temporary and permanent hearing loss or tinnitus. This data should be put into the context of the millions of MRI scans that are performed every year throughout the world.

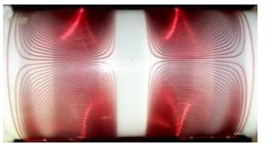
Safety Questionnaire

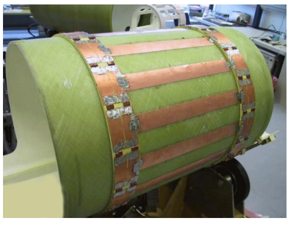


Before undergoing any MRI scan patients are given a safety questionnaire and consent form where they are asked various questions to determine whether they may have any metal on the surface or inside their bodies. Burns have for example also been reported in patients who have tattoos due to the ferromagnetic content of black inks. Several drug skin patches have metallic backing which can also cause burns. Any non-ferromagnetic implanted metal object has the potential to heat up so it is vital that the questions are answered as accurately as possible to ensure that the MRI examination in undertaken safely or in a few cases the patient is advised that they should not undergo an MRI examination.

The Main Risks

- **Static magnetic field B_0**
  - 0.2 – 3.0T clinically, 7.0T and above research
  - Ferromagnetic attraction (projectile) risk
  - Quench risk (asphyxia/cold contact burns)
- **Time varying magnetic field gradients $G_{x,y,z}$**
  - 10-50mT/m, approx 0-5kHz switching rate
  - Peripheral nerve stimulation and acoustic noise
- **Radiofrequency magnetic fields B_1**
  - 8-128MHz, 25kW+ RF amplifier
  - Induced electric currents can cause severe burns





In summary the main risks associated with MRI are firstly the static magnetic field is in the range 0.2 to 3 tesla clinically with 7 tesla and above used primarily for research. The main risk is ferromagnetic attraction or a projectile risk from ferromagnetic objects brought into the vicinity of the MRI magnets there is also a smaller risk associated with the magnet quenching and voiding helium gas into an enclosed space.

The risk associated with a time varying magnetic field gradient are primarily transient peripheral nerve stimulation and high levels of acoustic noise.

The main risk from the radio frequency magnetic fields are induced electric currents that can potentially cause severe burns

Learning Outcomes

- After these lectures you should be able to:
  - Explain how nuclear spin gives rise to magnetic resonance
  - Understand the principles of T_1, T_2 and T_2^*relaxation
  - Explain the principles of MR image formation
  - Describe the spin echo and gradient echo pulse sequences
  - Outline the basic components of an MRI system
  - Understand the safety issues related to MRI

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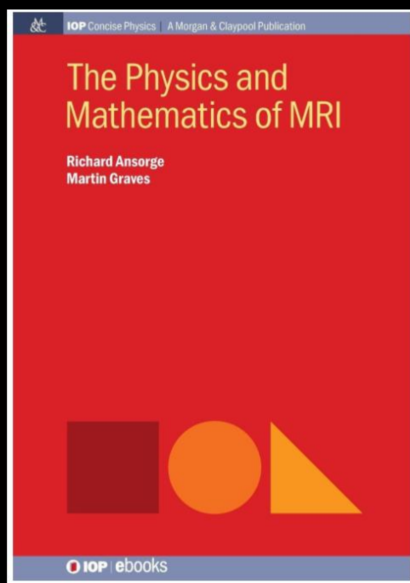
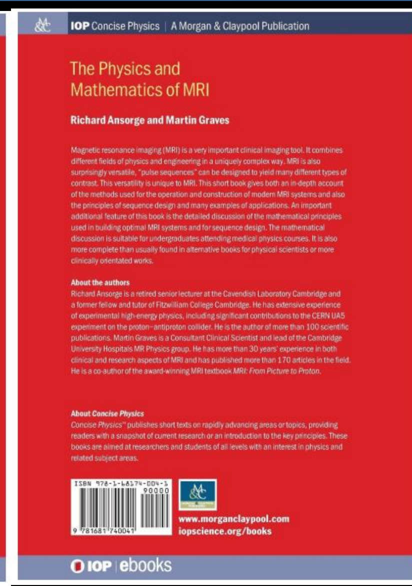
If you are interested in further reading about the subject, then this book that I co-authored with Richard Ansorge who has now retired from the Cavendish. It should be freely downloadable from the University via the link at the bottom

Further Reading

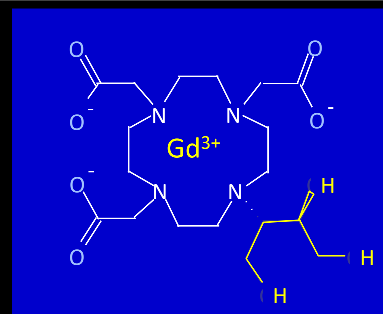


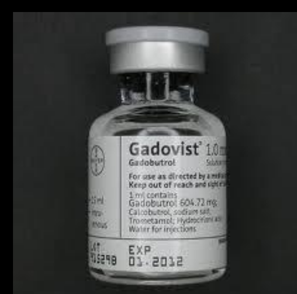


http://iopscience.iop.org/book/978-1-6817-4068-3

Gadolinium Based Contrast Agent (GBCA)

- Chelate of Gadolinium (Gd^{3+})
- Seven unpaired electrons in outer shell
- The electron has a 658 x larger magnetic moment than the proton*
- Increases T_1 (and T_2 relaxation) of nearby water molecules
- Demonstrates T1w hyper-enhancement in many pathological tissues





*https://physics.nist.gov/cgi-bin/cuu/Value?muemsmup

Contrast agents in MRI. These agents are based on the use of the rare earth element gadolinium (Gd). Since Gadolinium is toxic it is chelated to a compound that reacts with the metal ions to form a stable, water -soluble complex. A typical contrast agent is Gadovist (Bayer, Berlin, Germany) in which the gadolinium is chelated to dihydroxy-hydroxymethylpropyl- tetraazacyclododecane-triacetic acid (otherwise known as butrol). Gadolinium has seven unpaired electrons in its outer shell. Since the electron magnetic moment is about 700 times that of the proton, the magnetic moment of gadolinium causes a substantial reduction in the T1 relaxation time of water molecules that interact with the paramagnetic core of the contrast agent. On T1w imaging a reduction in T1 results in signal hyperintensity, and the effect of the gadolinium helps to amplify the signal and differentiate pathology from healthy tissue.

Contrast Agent Relaxation

$$(1/T_1)_{post} = [Gd]\, R_1 + (1/T_1)_{pre}$$

$$(1/T_2)_{post} = [Gd]\, R_2 + (1/T_2)_{pre}$$

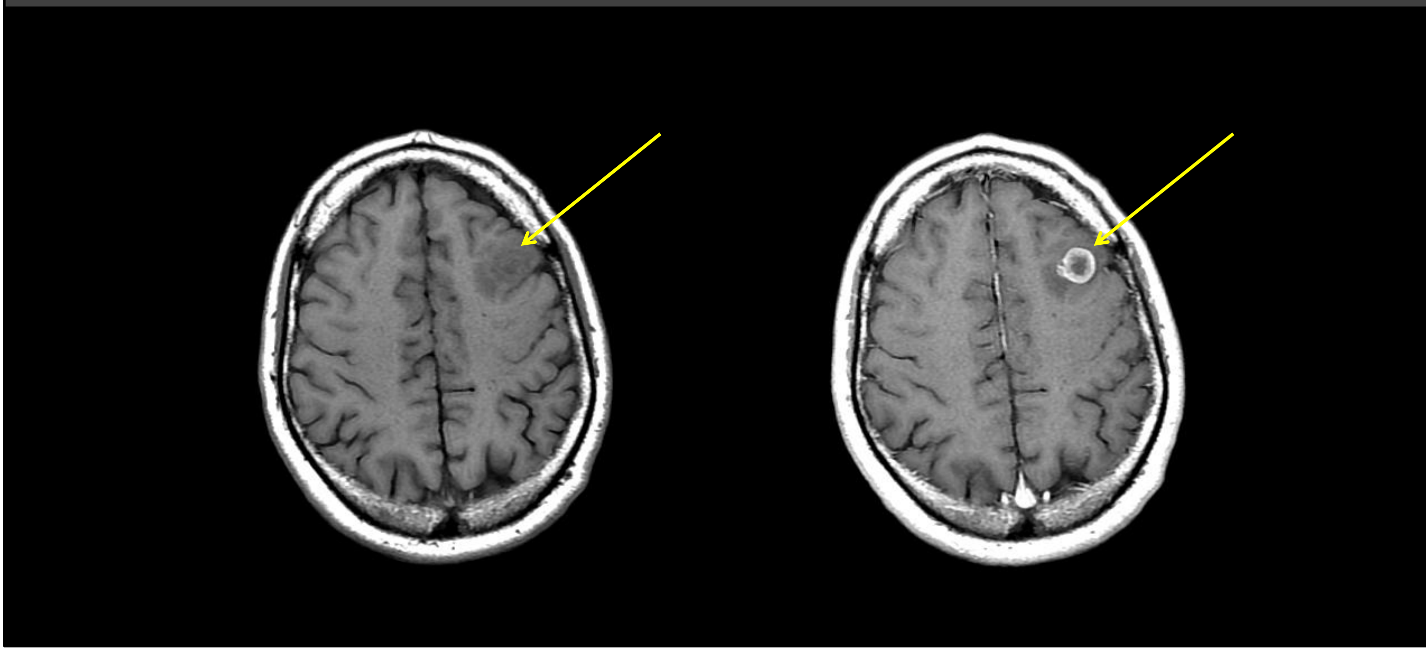
[Gd] is the concentration (mM L^{-1})

R_1 and R_2 are the T_1 and T_2 relaxivities of the contrast agent ($mM^{-1}\ s^{-1}$)

[N.B. The relaxivities are dependent upon B_0, temperature, surrounding material, e.g., water, plasma, blood]

To a good approximation, the change in relaxation rate 1/T1 is directly proportional to the contrast agent concentration at clinical doses. The proportionality constant is the relaxivity r1 of the agent measured in units of mM-1 s-1. Similar relationship applies to T2, but since T1 is generally much longer than T2 in biological tissue the effect on T1 is much bigger, as well as a reduction in T1 appearing as a signal hyperintensity.

Pre and Post Contrast T1w



Here is an example of a patient with a brain tumour. The effect of the contrast agent on the image on the right compared to the unenhanced study on the left can be clearly seen. The ring-enhancement indicates a tumour with a central necrotic region which is most likely a metastasis from a primary cancer elsewhere in the body.