Basics of MRI

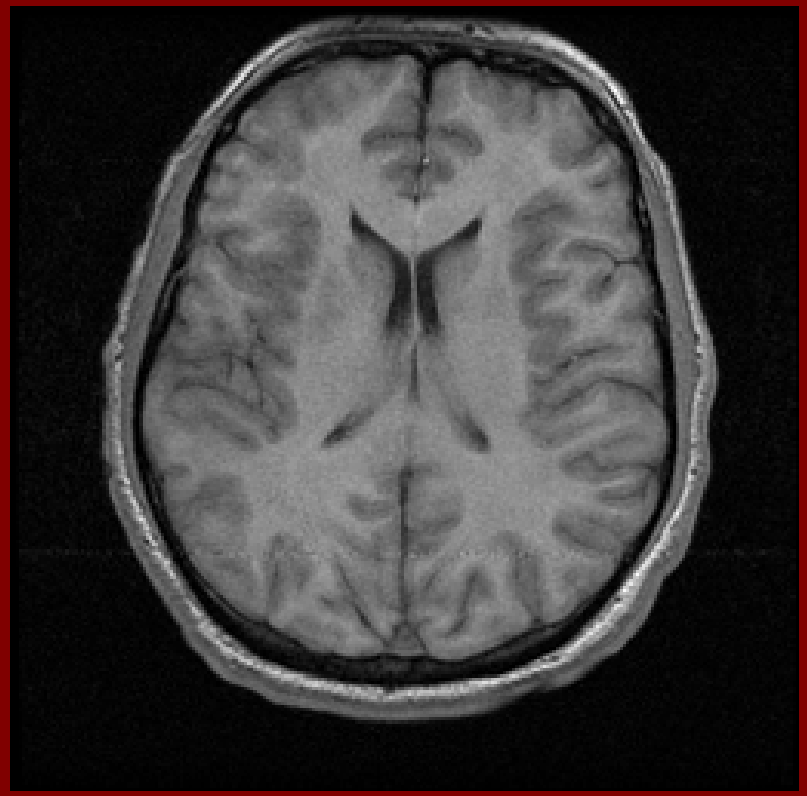
Professor Sir Michael Brady FRS FREng
Department of Engineering Science
Oxford University

Michaelmas 2004

Lecture 1: MRI image formation

- Basic nuclear magnetic resonance – NMR
- NMR spectroscopy
- Magnetic field gradients – spatial resolution
- Extension to 2D, 3D – images and volumes

T1 Weighted Image



SPGR, TR=14ms, TE=5ms, flip=20º

MR image of a horizontal slice through the brain.

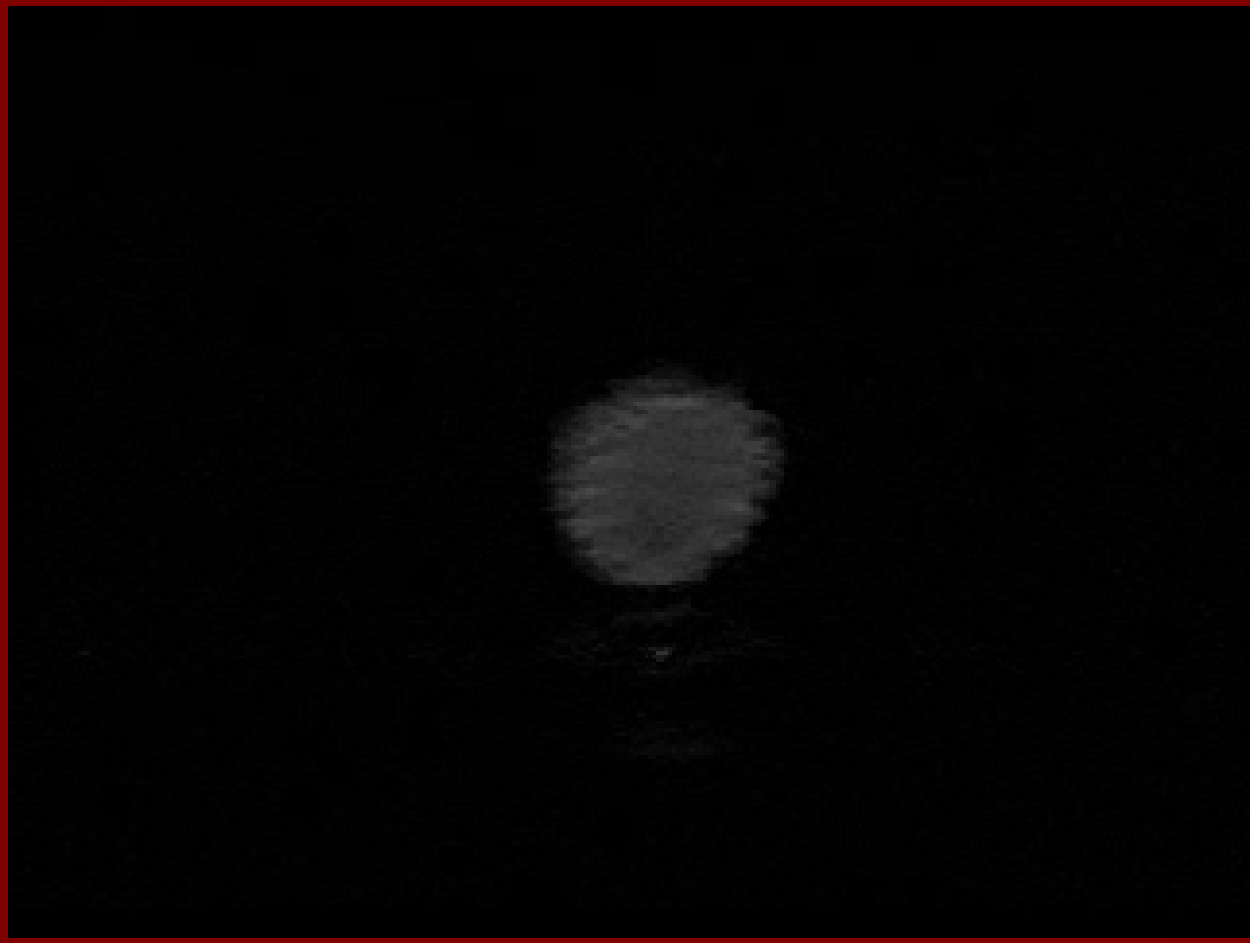
In this T1-weighted image, grey matter is lightly coloured, while white matter appears darker.

	T_1/s	R_1/s^{-1}
white matter	0.7	1.43
grey matter	1	1
CSF	4	0.25

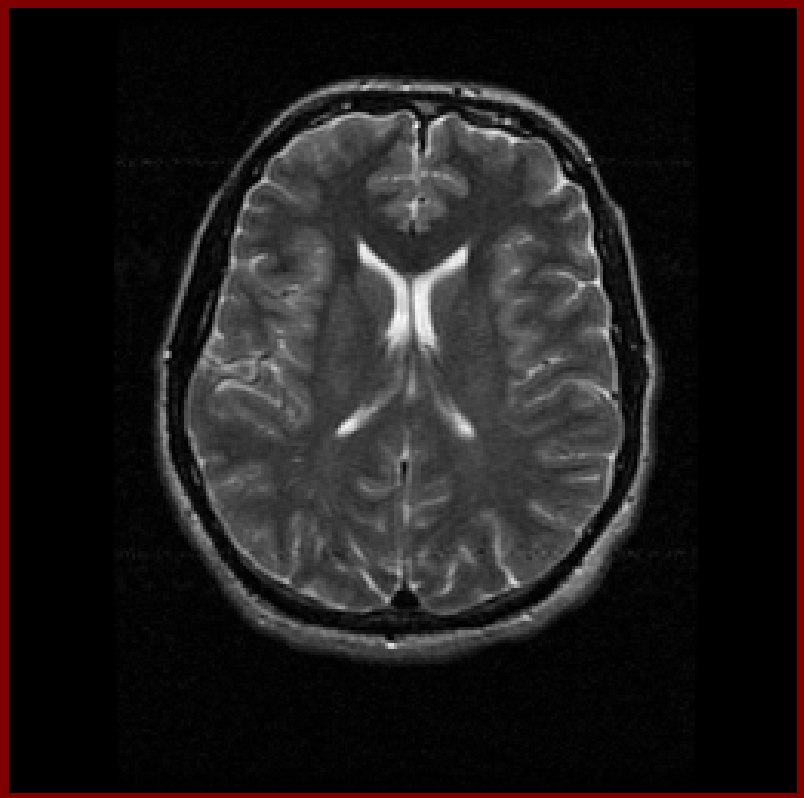
1.5T

Brain image analysis

- Accurate shape modelling and measurement



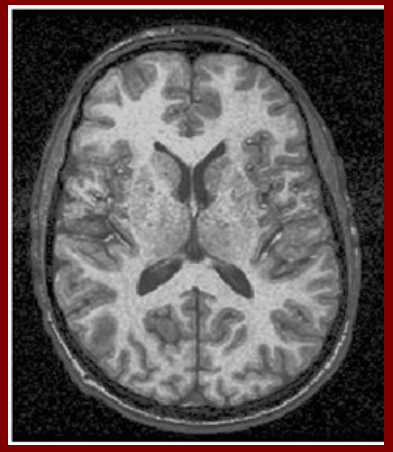
T2 Weighted Image



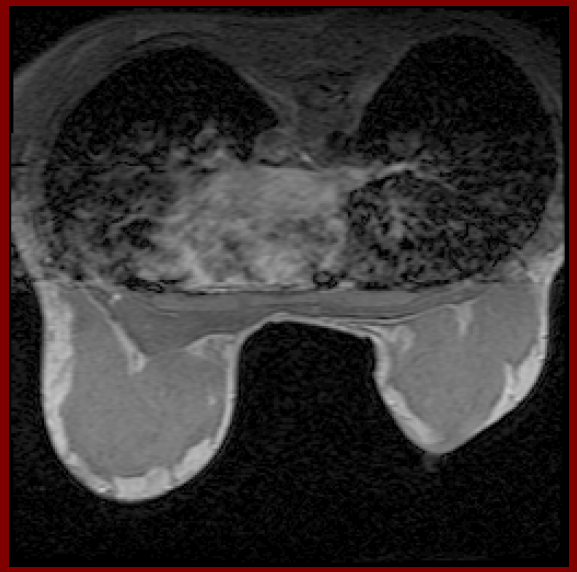
SE, TR=4000ms, TE=100ms

	T_2/ms
CSF	500
grey matter	80–90
white matter	70–80

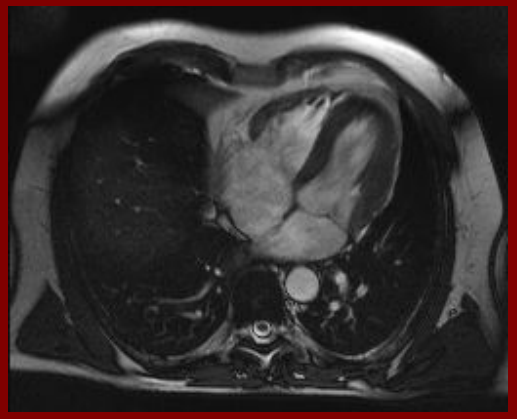
1.5T

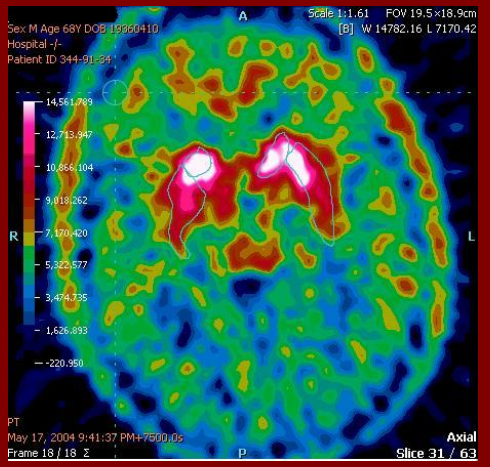


Over the past 20 years, we have developed new ways to image anatomy, new ways to see inside the body, non-invasively

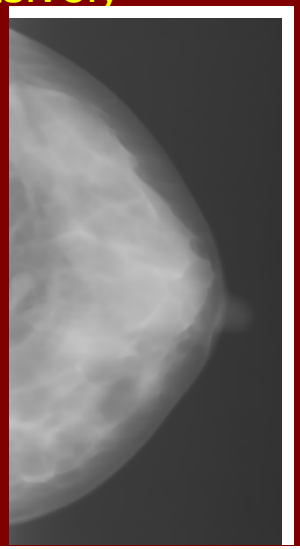


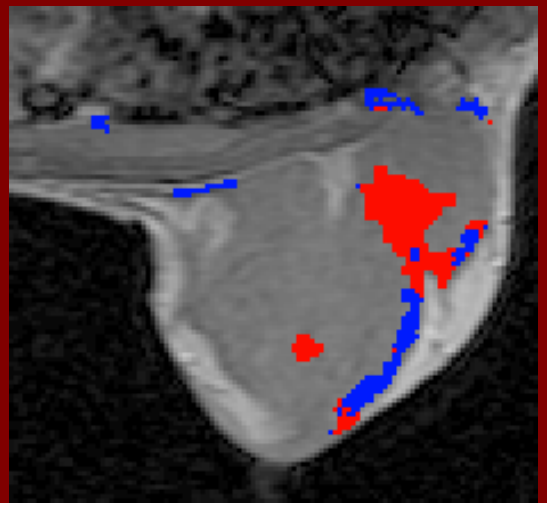
We can watch the body in action, as it responds to the injection of a drug or contrast agent, to highlight aberrant physiology

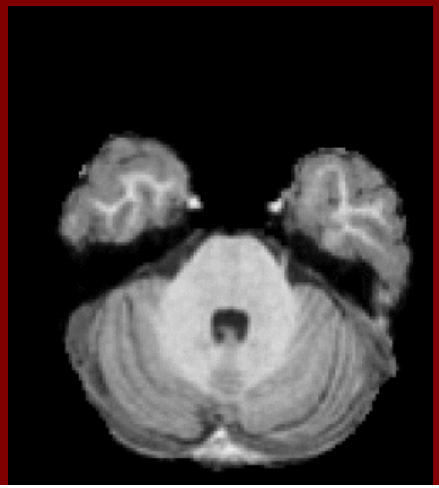




We can watch the body functioning in a whole range of ways – the brain thinking, degradation in white matter, and the pulsing of the heart

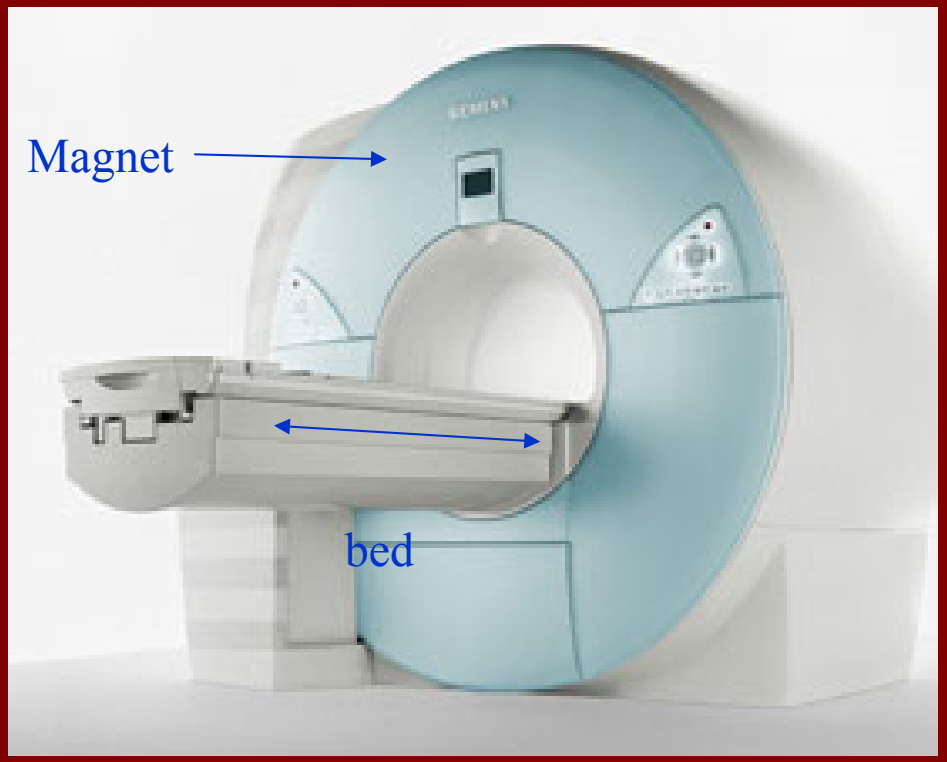






Now we are beginning to image cellular and molecular processes – the convergence of molecular biology and image analysis

MRI machines

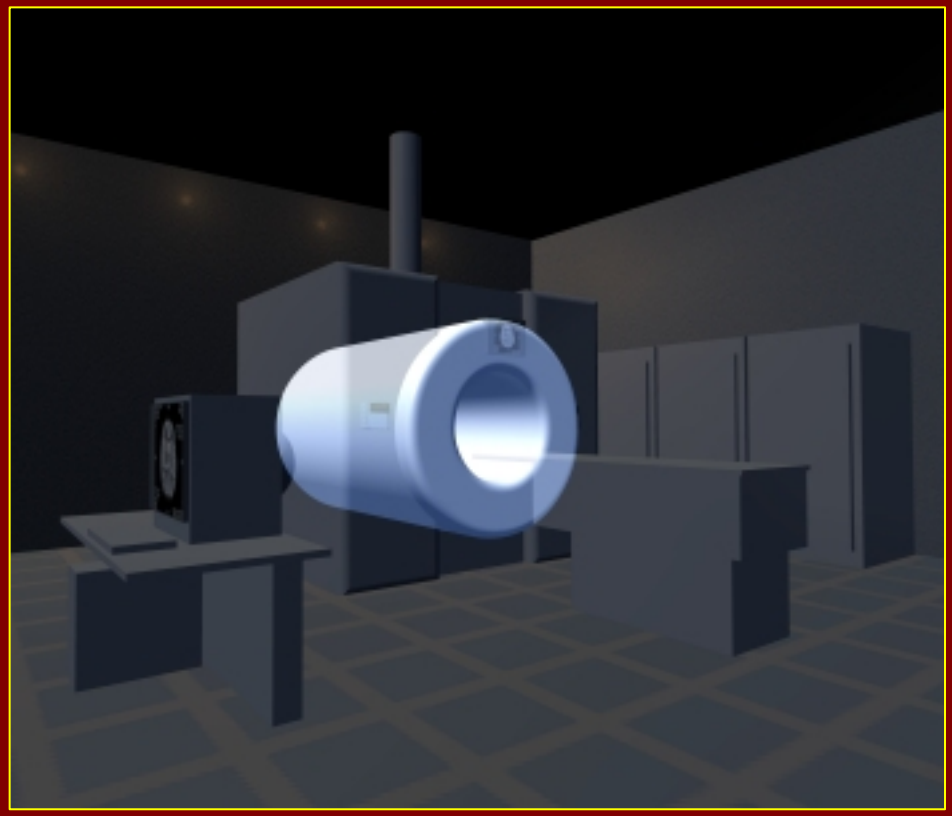


Siemens Avanto 1.5T state-of-the-art MRI machine

The Magnet

Magnets field strength:

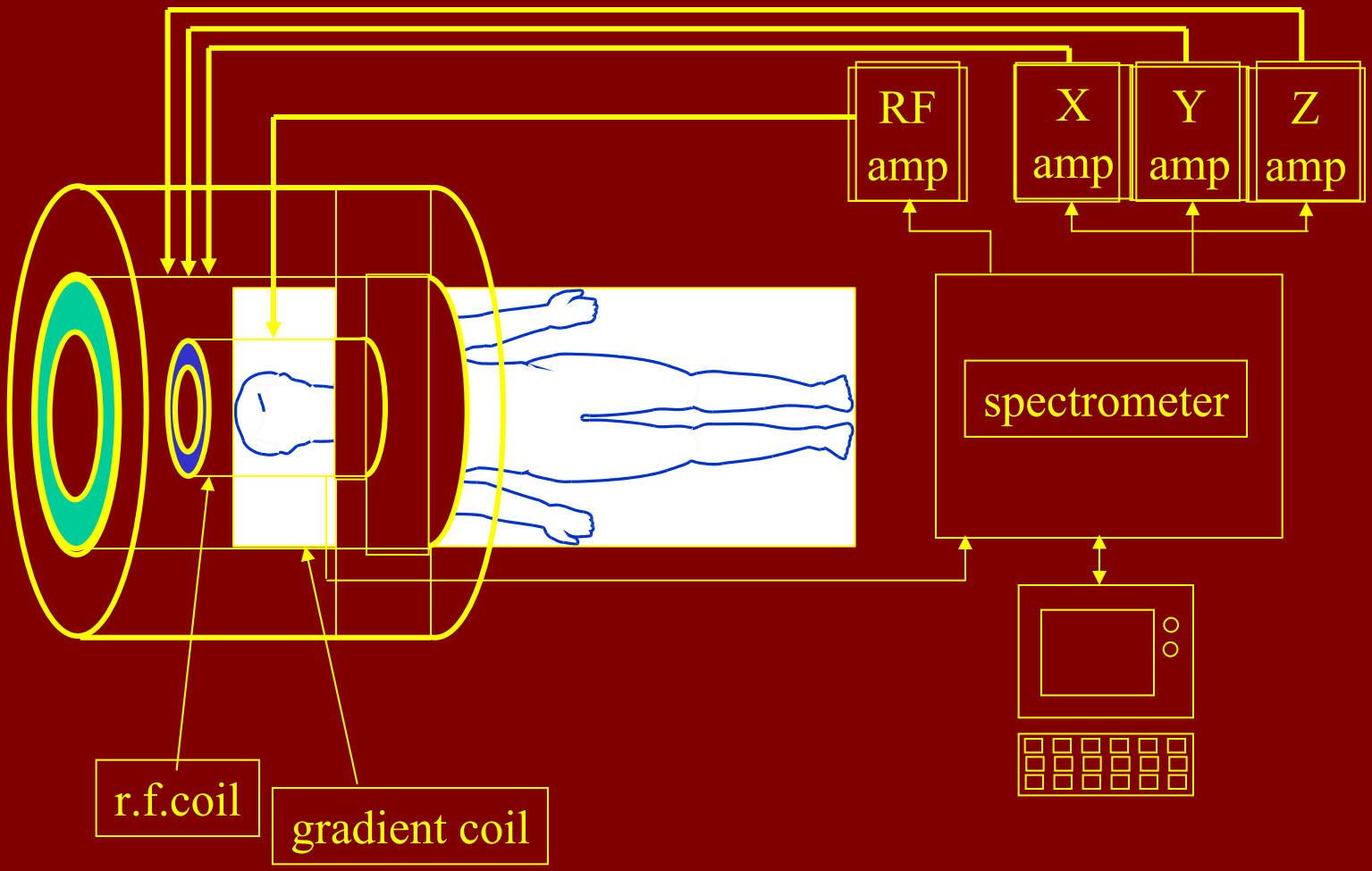
imaging - 0.2T to 2.0T
spectroscopy- 2.0T to 7.0T



- Low field 0.2- 0.5T
- Intermediate 0.5- 1.5T
- High field 1.5- 4.0T
- Ultra high > 4.0T

Earth's magnetic field = 5×10^{-5} Tesla

MRI System Block Diagram



MRI is considered ideally suited for soft tissue problems

*** MRI is to soft tissue as x-ray is to dense tissue (bone)***

- Diagnosing multiple sclerosis (MS)
- Diagnosing brain tumours
- Diagnosing spinal infections
- Visualizing torn ligaments in the wrist, knee and ankle
- Visualizing shoulder injuries
- Evaluating bone tumours, and herniated discs in the spine
- Diagnosing strokes in their earliest stages

Some disadvantages of MRI

- Extreme precautions must be taken to keep metallic objects out of the room where the machine is operating
- People with pacemakers can't safely be scanned
- Some people suffer from claustrophobia, and find the confinement discomforting
- The machine makes a very loud continuous hammering noise when operating
- Some people are too big to fit inside the magnet
- MRI scans require patients to hold very still for long periods of time ... up to 90 minutes or more in some cases
- MRI systems are expensive to buy – and run – so are currently beyond most DGHs. An MRI scan costs about £500 to the NHS

MRI installed base

- 1990 – unit sales of MRI systems, tens to hundreds of MRI scans
- 2004 installed base is 12,000 MRI systems
- 75-80 million scans per year (£400/scan)
- Market growth at 10% pa
- Major growth is in high field MRI ($\geq$ 3T)
- Despite much excitement about open magnet systems, take-off is slow (Hitachi dominate this sector)
- Special purpose MRI systems have not had much impact (sigh)

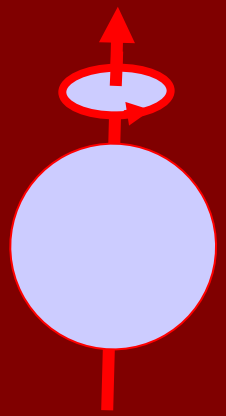
Energy (spin) states

Long before Tony Blair, Quantum Mechanics invented the concept of *spin*

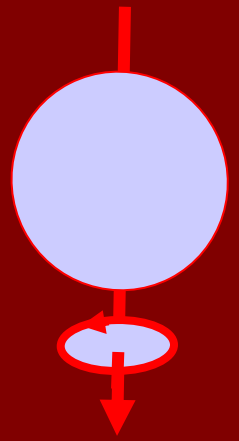
In a strong magnetic field, nuclei act like tiny dipole magnets that align with – or, amazingly, against – the field

Strong magnetic field

B_0



Aligned with the field: lower energy state



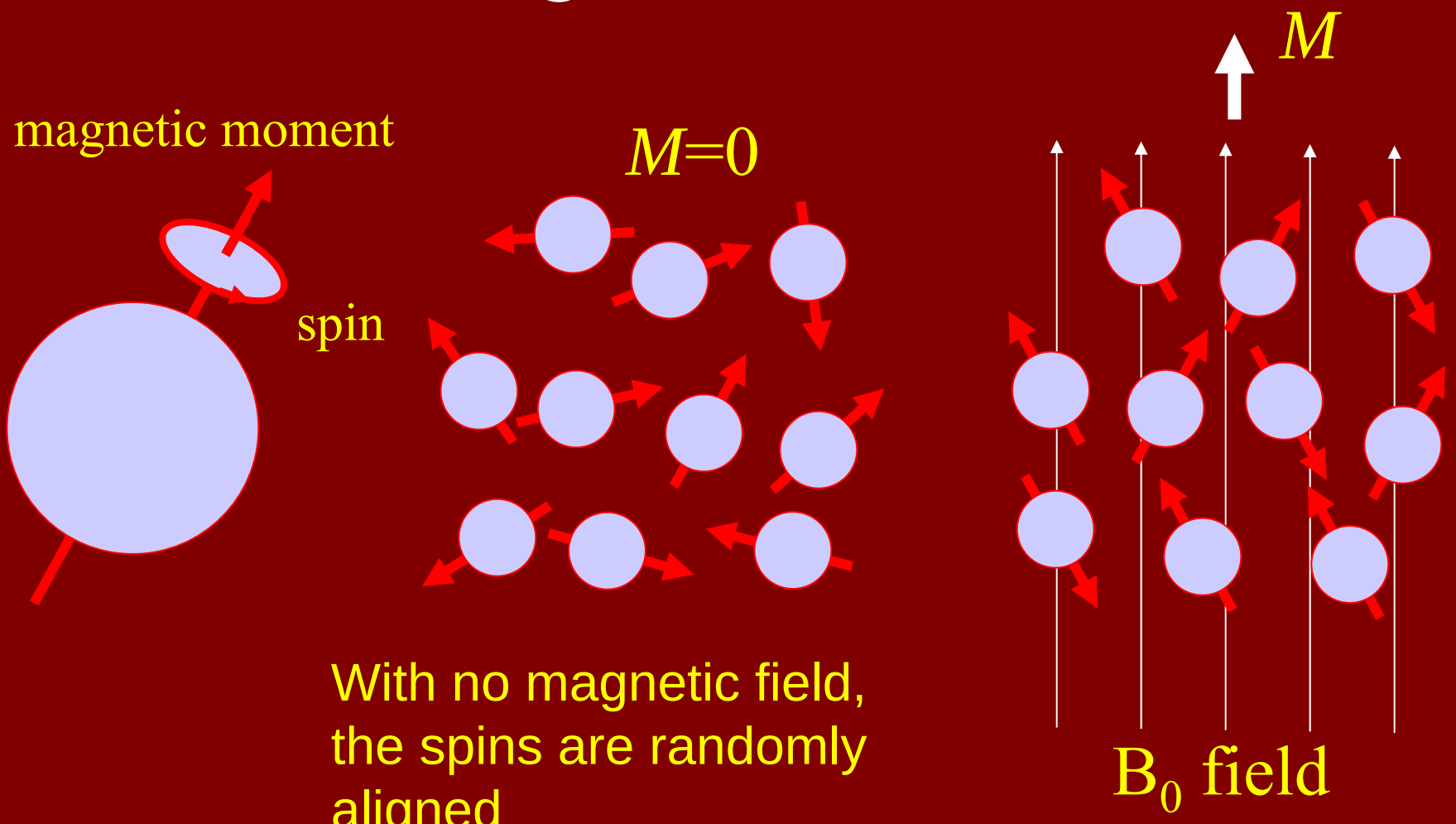
Aligned against the field: higher energy state

Protons (= hydrogen nuclei) have two spin states. They act like miniature tops and "precess" about the field direction

Spin, moment

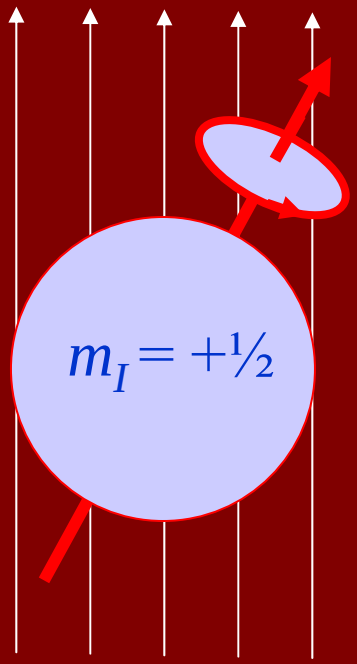
- All nuclei have spin – multiples of ½
- Combined with charge ➔ moment
- Nucleus with odd spin acts like a small dipole magnet
- If nucleus has *S* spin states, the moment (magnet) has $2S+1$ stable states in an external magnetic field
- Hydrogen (proton): $S = ½$ ➔ 2 states

Alignment of Spins in a Magnetic Field



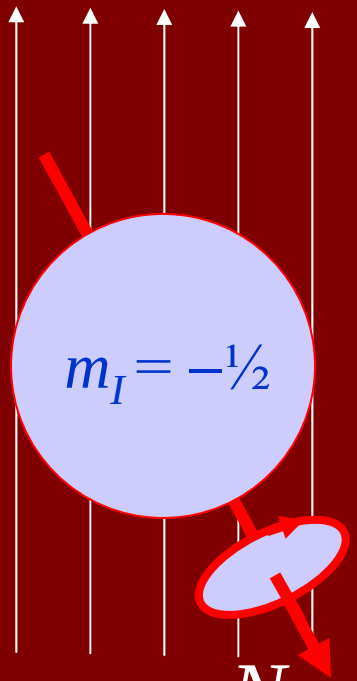
With no magnetic field, the spins are randomly aligned

Energy in a Magnetic Field (Zeeman Splitting, Spin ½)



$E_{+1/2} = -\gamma\hbar B_0/2$

$P_{+1/2} = 0.5000049$



$E_{-1/2} = +\gamma\hbar B_0/2$

$P_{-1/2} = 0.4999951$

1.5T, T=310K, $P(E) \propto \exp(-E/kT)$

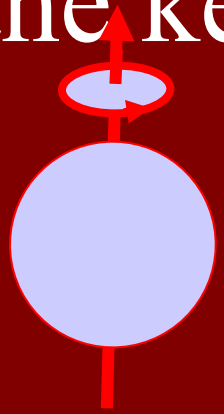
$$\frac{N_{up}}{N_{down}} \approx \exp\left(\frac{\gamma\hbar B_0}{kT}\right)$$

Excess of protons aligned with field

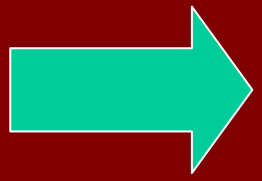
- For an external magnetic field of 3T, there are only about 10 per million more protons parallel to the field than anti-parallel!!
- Nevertheless, there are millions of protons, so this is enough to give a useful magnetic field
- The smaller the field, the fewer the excess, the poorer the SNR – so use a very large magnetic field

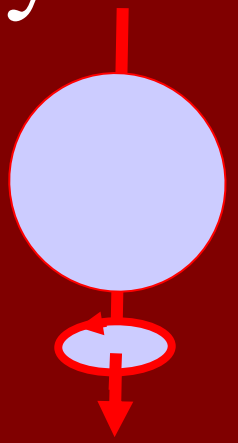
NMR – the key discovery

The nucleus is bombarded with Radiofrequency (RF) energy



An unsuspecting low energy nucleus





At certain resonant frequencies of the RF energy, the proton flips to the high energy state

When the RF energy is turned off, the newly high energy nucleus may revert to its low energy state, giving off RF energy in the process

So what?

The resonant frequency at which this happens is called the Larmor frequency:

$$\omega = B_0 \times \gamma$$

In this equation, γ is the "gyromagnetic constant" for the stuff that is being energised

Critically, this constant depends on the biochemical nature of the stuff and its surroundings: ***Chemical Shift***

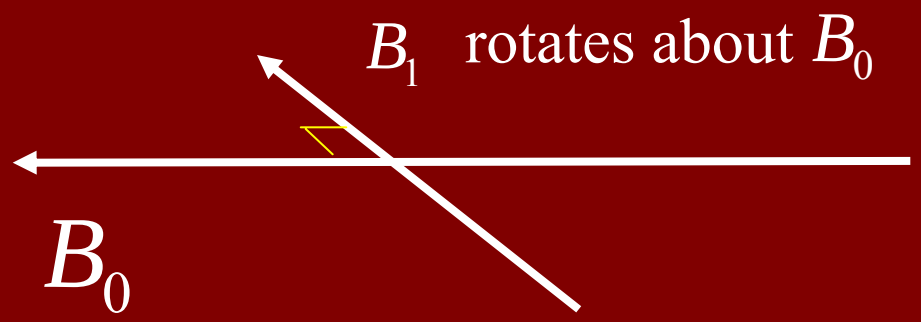
A typical field strength B0 used in MRI is 1.5 Tesla

At this field strength, the Larmor frequencies ξ for Hydrogen and Carbon 13 (the atoms most relevant in medical imaging) are 63.9 MHz and 16.1 MHz respectively.

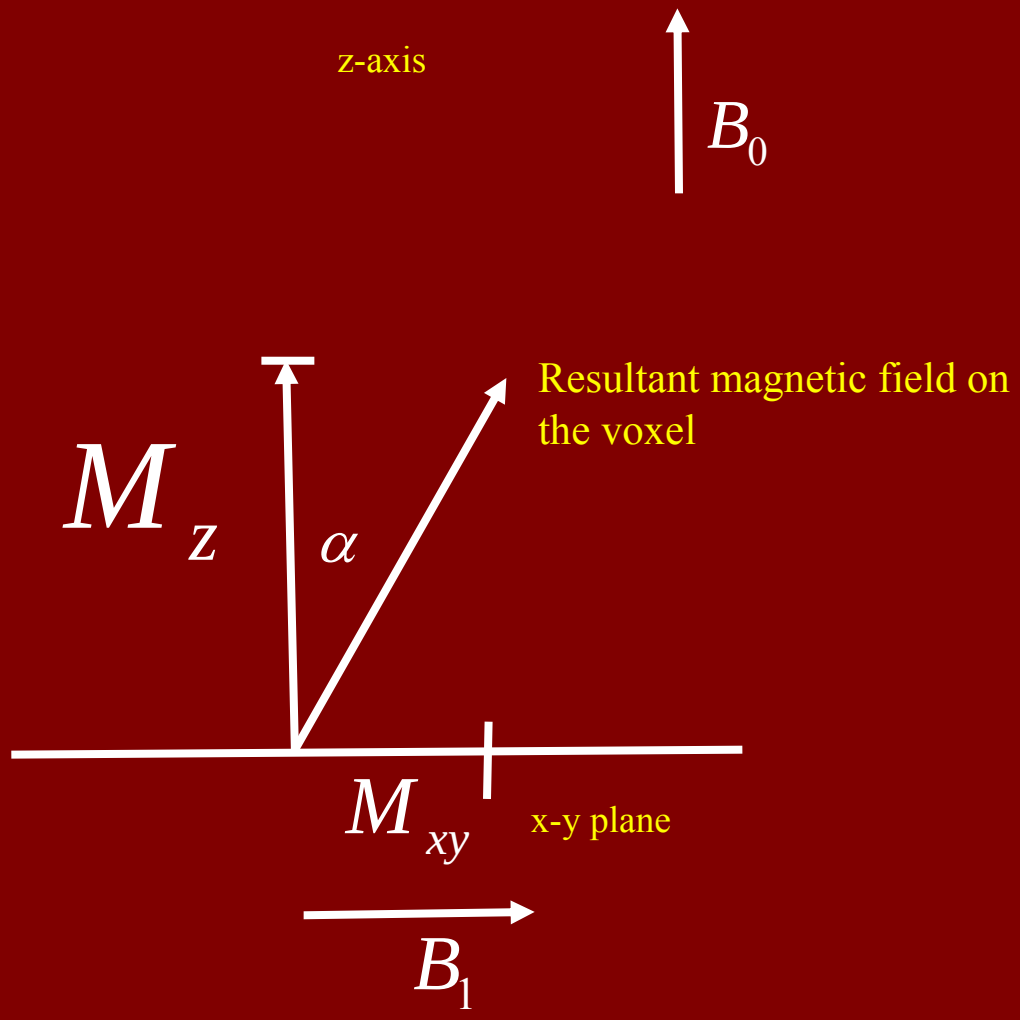
Probing with different frequencies of RF energy enables us to build a spectrum of what is in the sample

Exciting a spin system

- subject it to a short period of high intensity radiowaves at a frequency close to the Larmor frequency
- This is called the B_1 field, orientated in a direction perpendicular to, and rotating about, the B_0 field. The magnitude of $B_1 \approx 10^{-5}\ B_0$
- In a co-ordinate system rotating at or close to the Larmor frequency, this results in rotation of the magnetization away from the direction of the external magnetic field



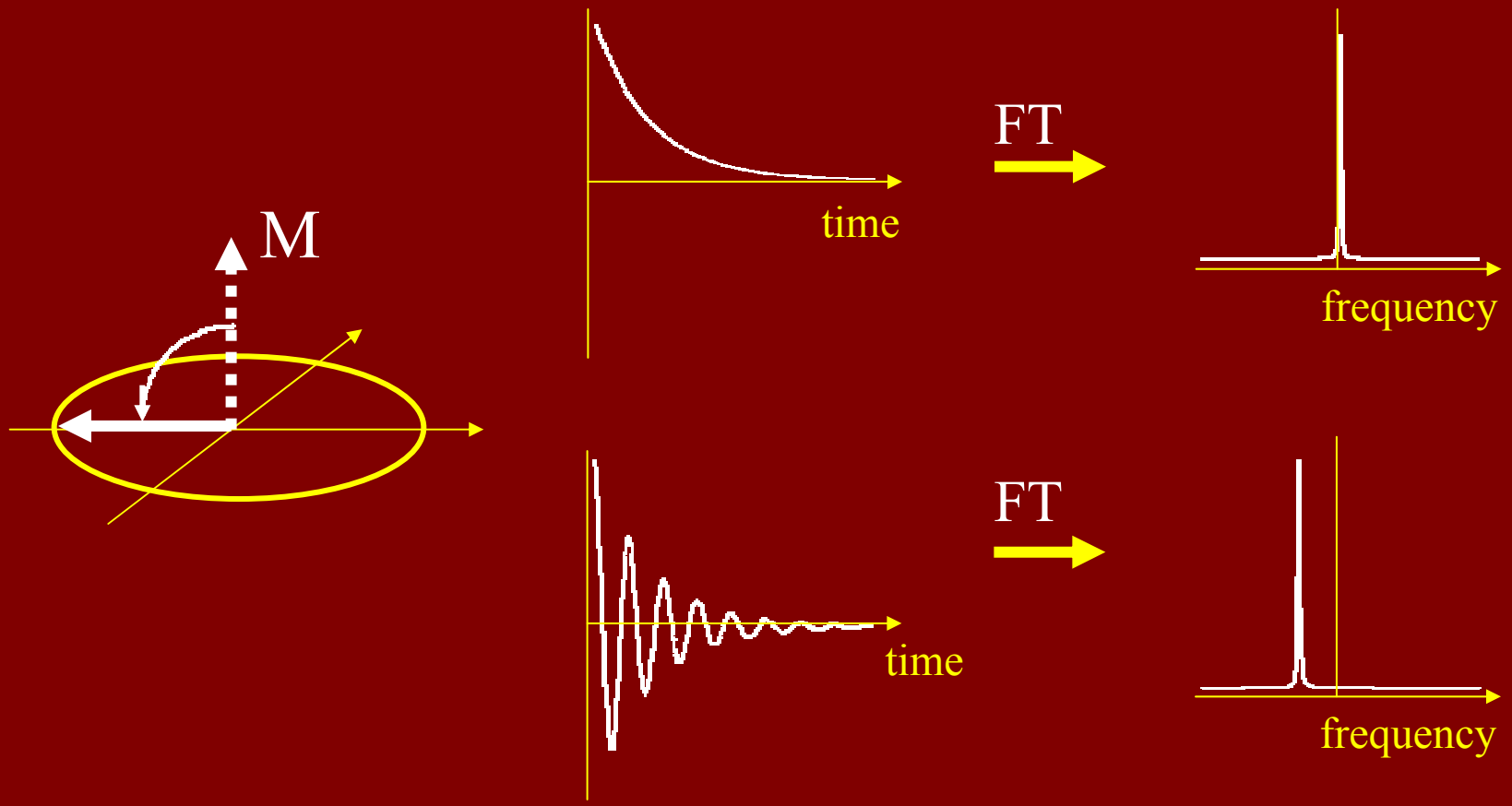
Applying a pulse – B_1



The longer the RF pulse is applied, and the stronger it is, the bigger the deflection of the net magnetic field, that is, the bigger the angle α.

It can reach 90, or even 180 degrees. The bigger α, the longer it takes to recover when the RF is turned off.

Free Induction Decay



The relaxation constant T_1

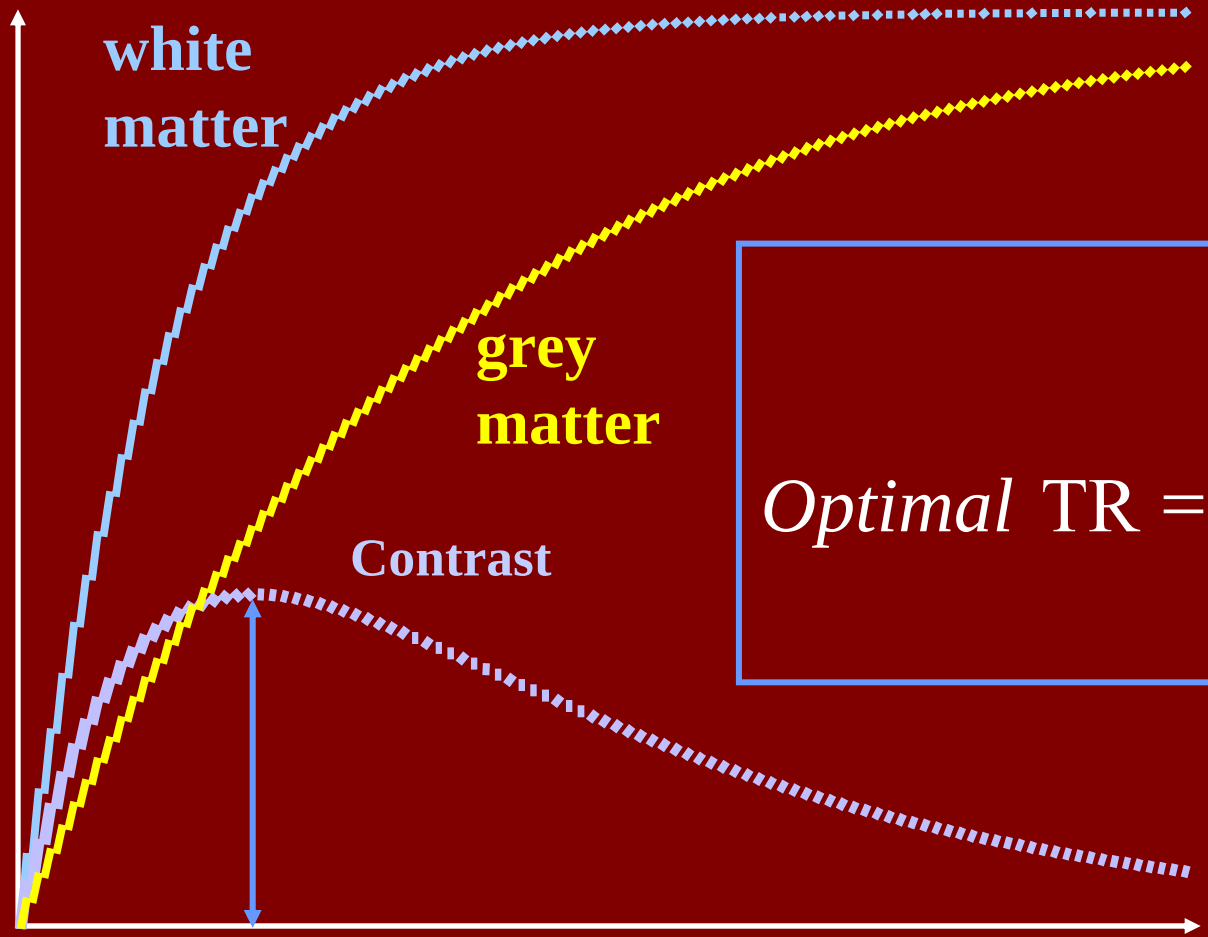
For the nuclei to return to their initial energy states by emitting energy (the MR signal), the excited spin system must be exposed to an electromagnetic field oscillating with a frequency at or close to the Larmor frequency. This process can occur by the nuclei being `stimulated' by surrounding nuclei and is assumed to occur in a simple exponential manner

$$M_z(t) - M_0 = \left(M_z(0) - M_0\right)e^{-\frac{t}{T_1}}$$

T_1 is called the *spin-lattice relaxation time*

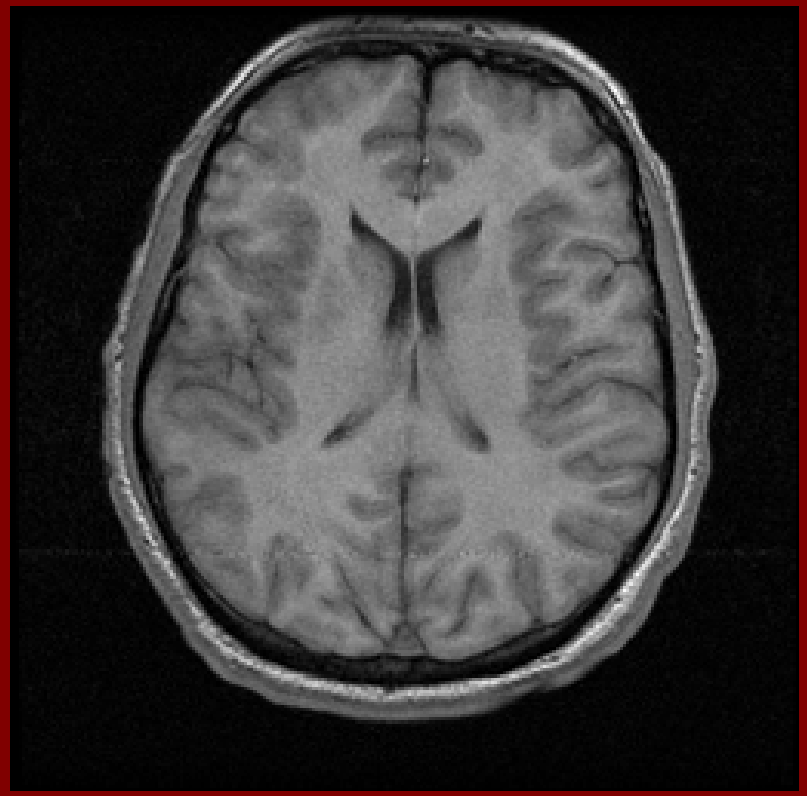
It corresponds to the time required for the system to return to 63% of its equilibrium value after it has been exposed to a 90˚ pulse

T1 Weighted Imaging



$$Optimal\ \mathrm{TR} = \frac{\ln\left(\frac{T_{1b}}{T_{1a}}\right) T_{1a} T_{1b}}{T_{1a} - T_{1b}}$$

T1 Weighted Image



SPGR, TR=14ms, TE=5ms, flip=20º

	T_1/s	R_1/s^{-1}
white matter	0.7	1.43
grey matter	1	1
CSF	4	0.25

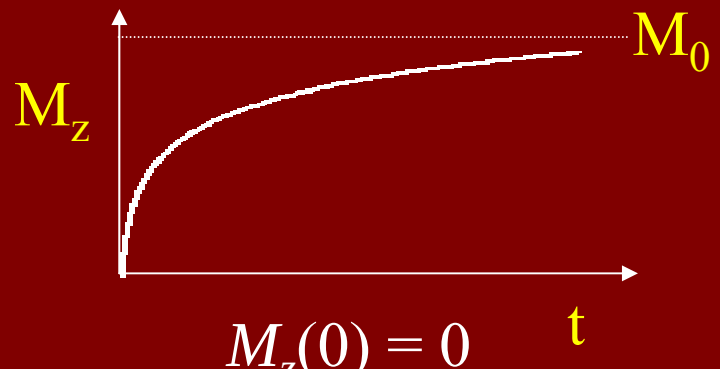
1.5T

T1 Relaxation

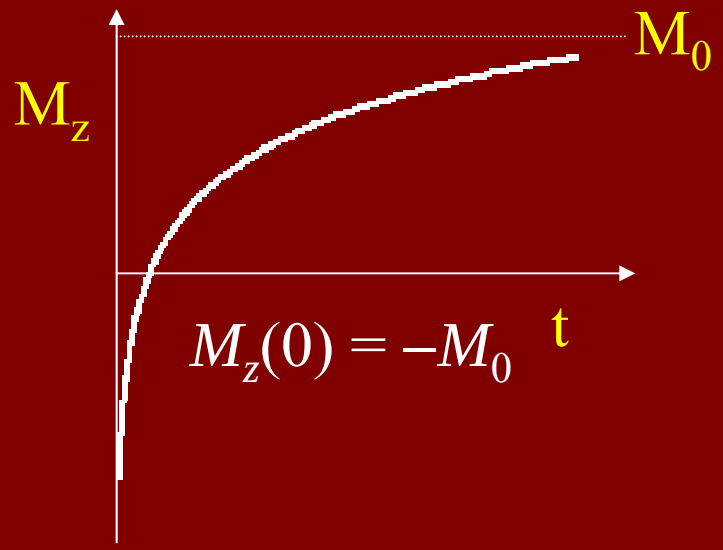
$$\frac{dM_z(t)}{dt} = -\frac{\left[M_z(t) - M_0\right]}{T1}$$

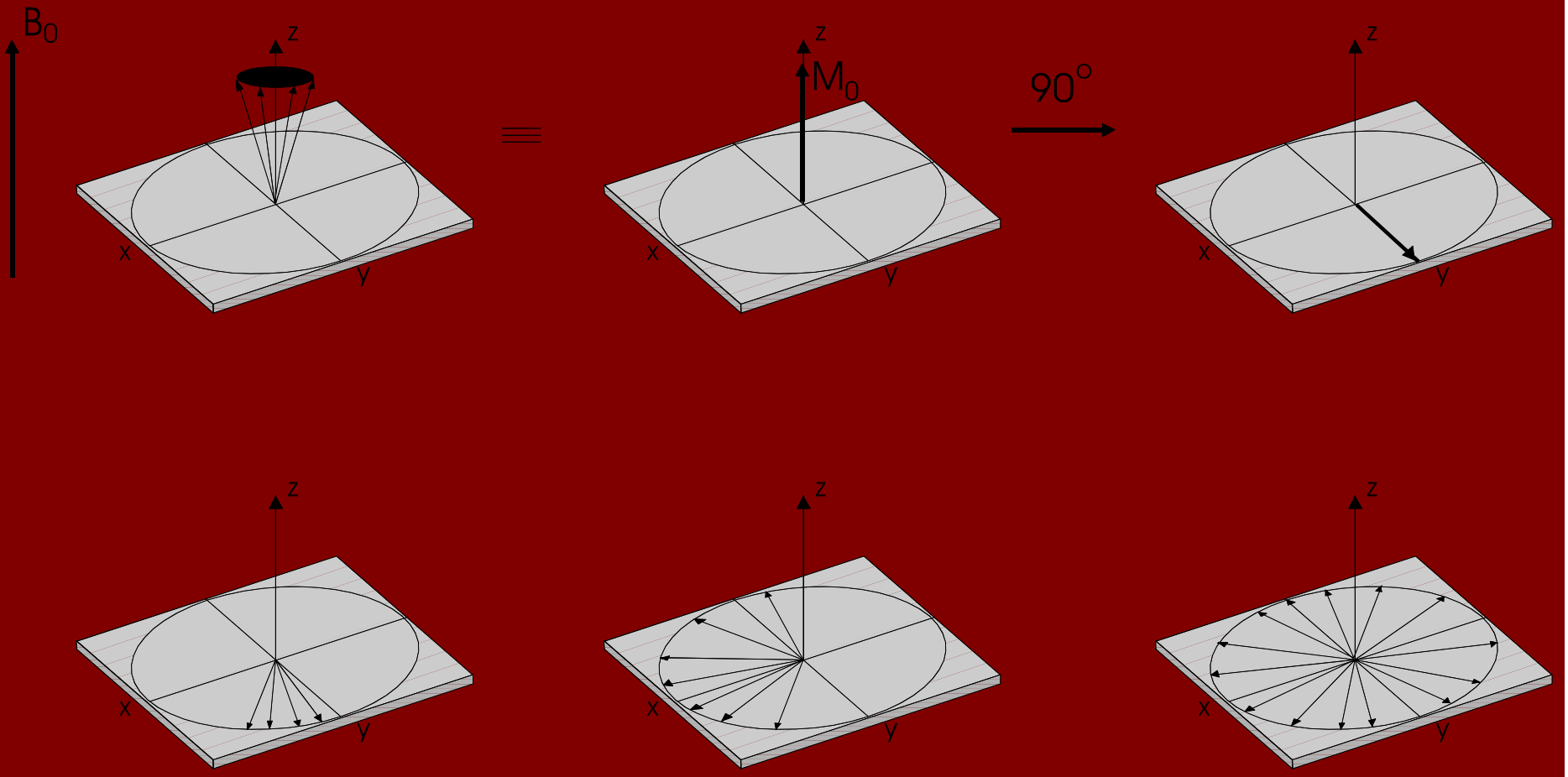
$$M_z(t) = M_0 + \left\{M_z(0) - M_0\right\}\exp(-t/T1)$$

saturation–recovery



inversion–recovery





The contribution of all the spins precessing around the external magnetic field B_0 produces a net magnetisation M_0. When a 90º RF pulse is applied, this net magnetisation is tipped onto the x,y-plane. Dephasing of the spins results in a quick decrease of the net magnetisation in the x,y-plane. The dephasing is exponential and characterised by T_2.

The relaxation constants T_2 T_2^*

Immediately after a pulse is applied, all of the nuclei precess around the magnetic field in phase. As time passes, the spins begin to dephase and so the observed signal decreases. They do so according to:

$$M_T(t) = M_T(0)e^{-t/T_2}$$

T_2 is called the *spin-spin relaxation time*

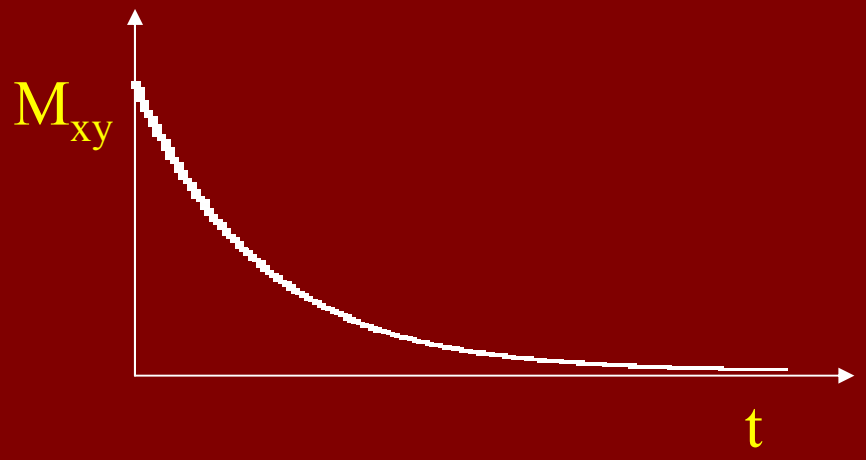
- T_2 values are 40-200ms depending on the tissue
- T_2 is approximately ten times smaller than T_1.
- Different scan sequences show up differences in these relaxation times generating what are referred to as T_1, T_2 or proton density (the concentration of protons) weighted images.

$$\frac{1}{T_2^*} = \frac{1}{T_2} + \frac{1}{T_{2inhomogeneous}}$$

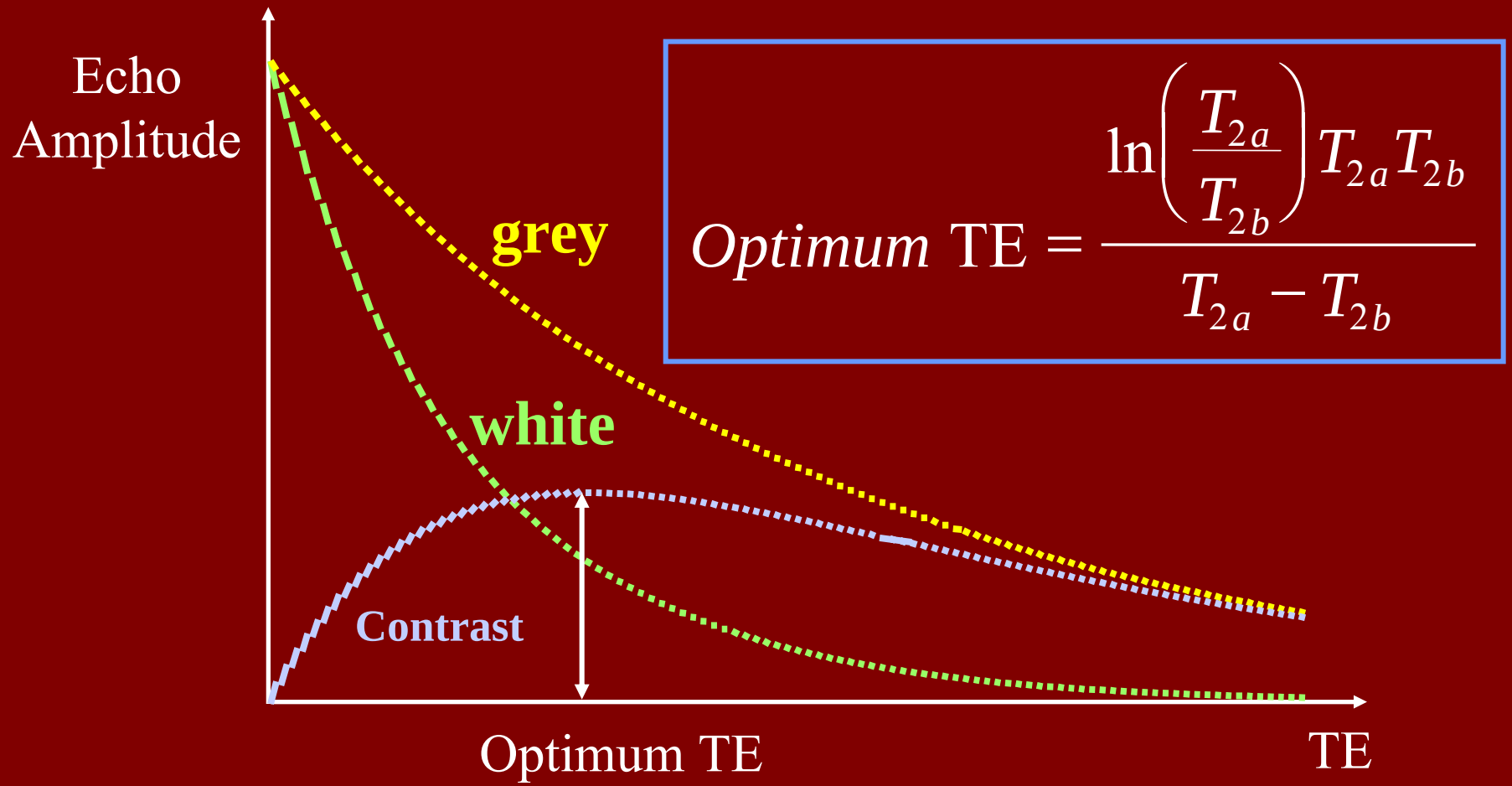
T2 Relaxation

$$\frac{dM_{xy}(t)}{dt} = -\frac{M_{xy}(t)}{T2}$$

$$M_{xy}(t) = M_{xy}(0)\exp(-t/T2)$$

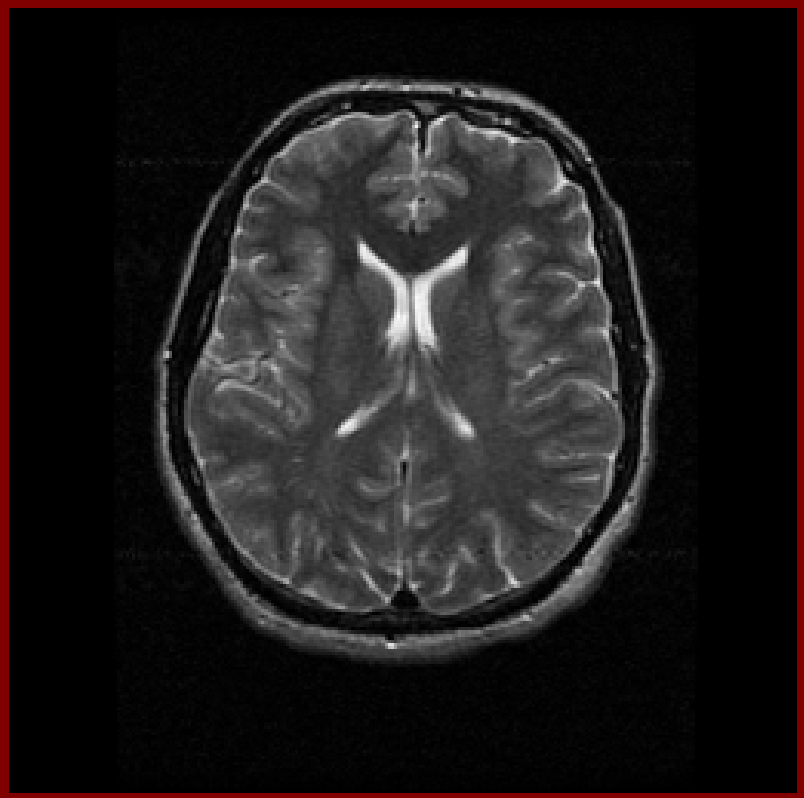


T2 Weighted Imaging



$$\textit{Optimum}\ \mathrm{TE} = \frac{\ln\left(\frac{T_{2a}}{T_{2b}}\right) T_{2a} T_{2b}}{T_{2a} - T_{2b}}$$

T2 Weighted Image

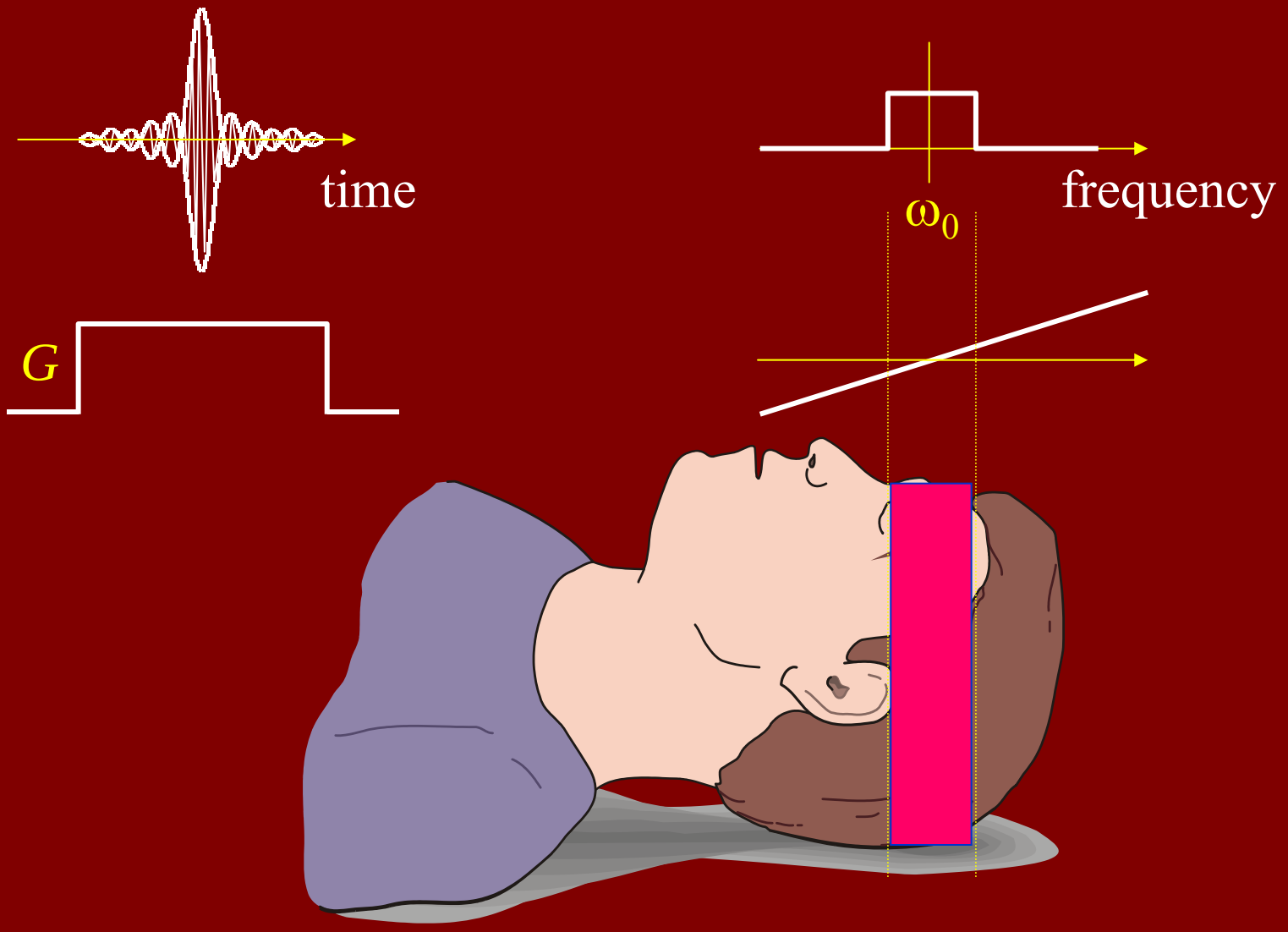


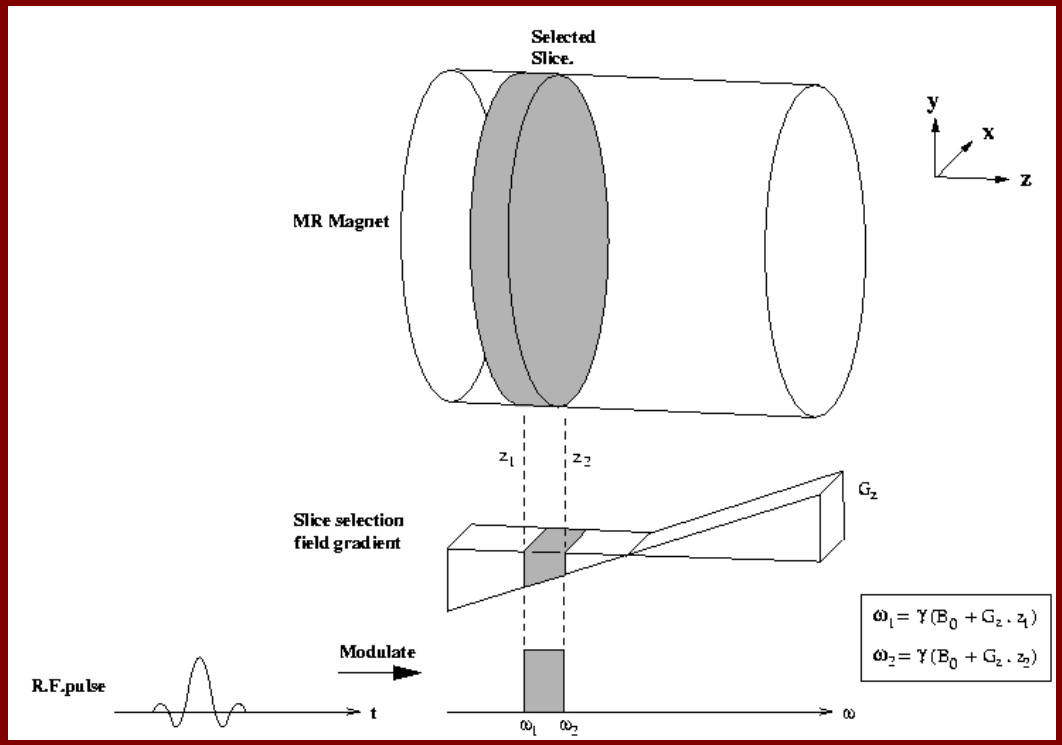
SE, TR=4000ms, TE=100ms

	T_2/ms
CSF	500
grey matter	80–90
white matter	70–80

1.5T

Slice Selection





To image a slice of material requires a method of exciting only material within that slice. This is achieved by superimposing a small spatially varying magnetic field, G_z, called a gradient field. The field is applied in the same direction as while the RF pulse is applied.

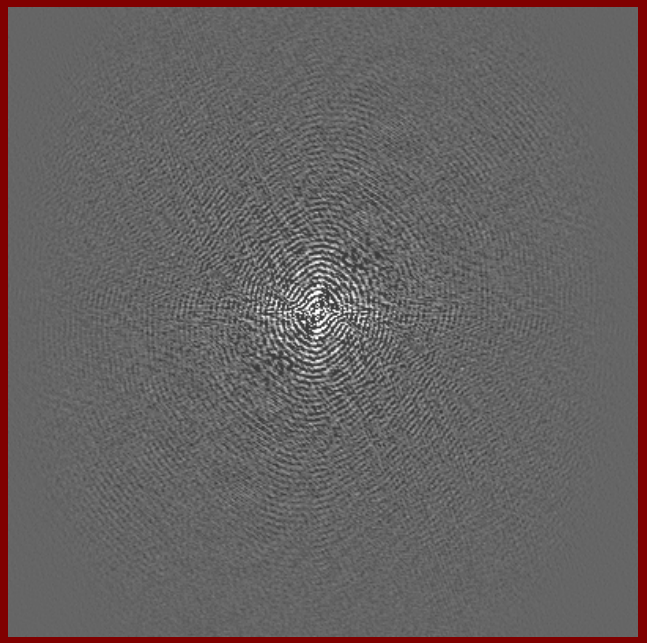
Recall Larmor's equation : $\omega = B_0 \times \gamma$

Now suppose there is a magnetic field gradient δB per metre

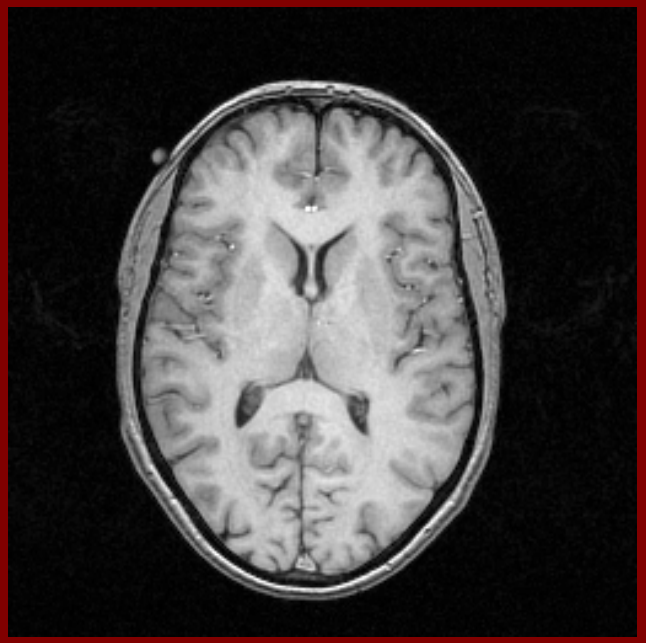
$\omega + \delta\omega(x) = (B_0 + x.\delta B) \times \gamma$, so knowing δB and the chemical shift $\delta\omega$ enables us to estimate x

Three mutually orthogonal gradient coils are used to localise a nucleus in x-, y-, and z-. Normally a slice is localised (z) and then a coil pair in x,y

MRI: the Fourier Transform



FFT

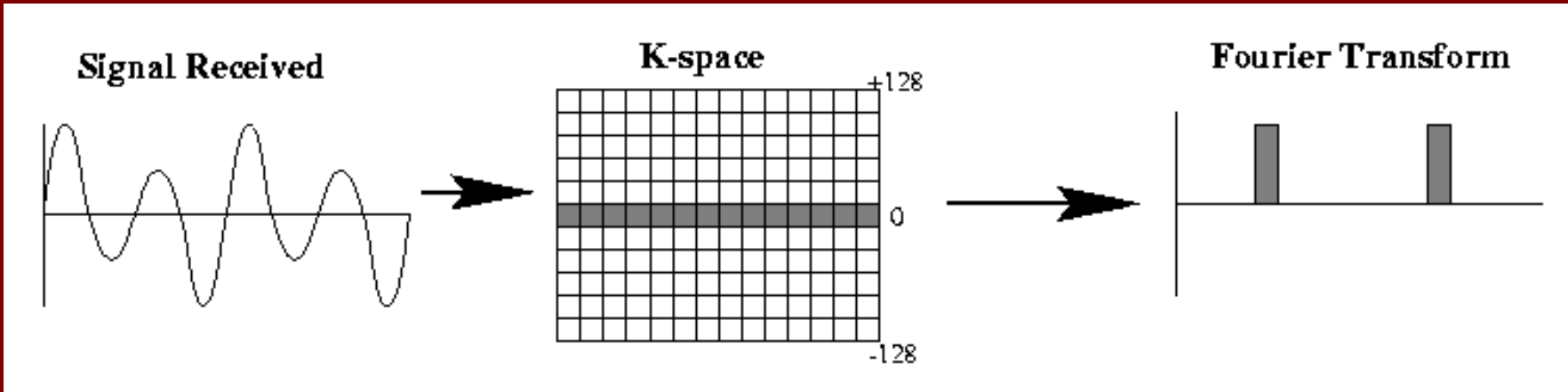


K-space – as measured in the MRI experiment

The medical image we inspect is the FT of k-space

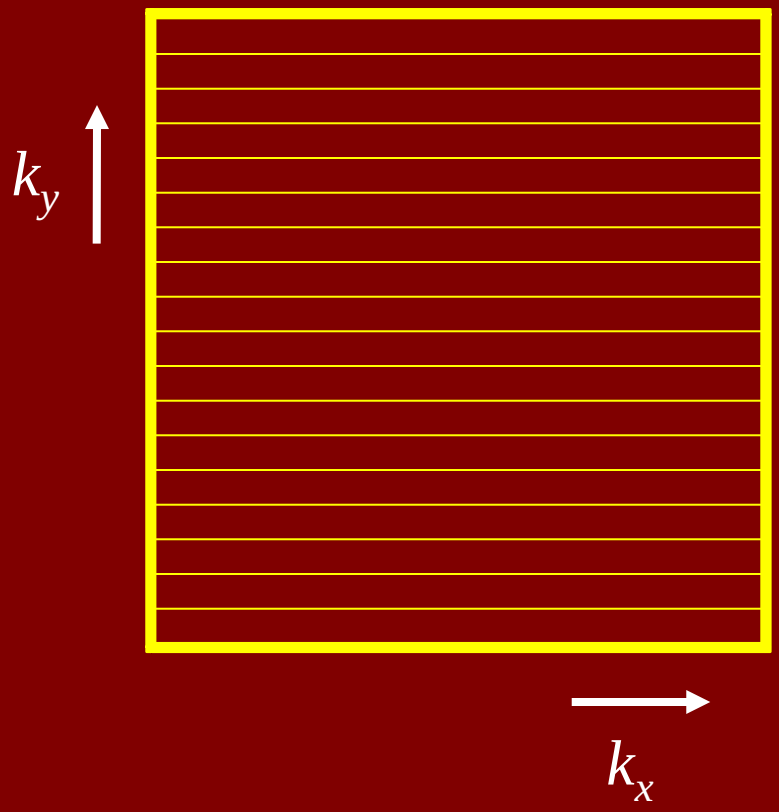
Note: motion of the subject will be local in k-space – so have a global effect on the image!

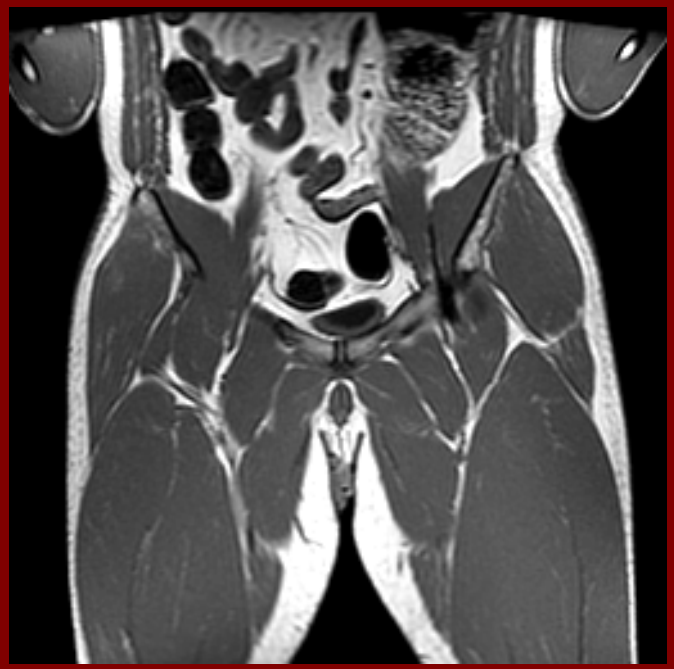
K-space



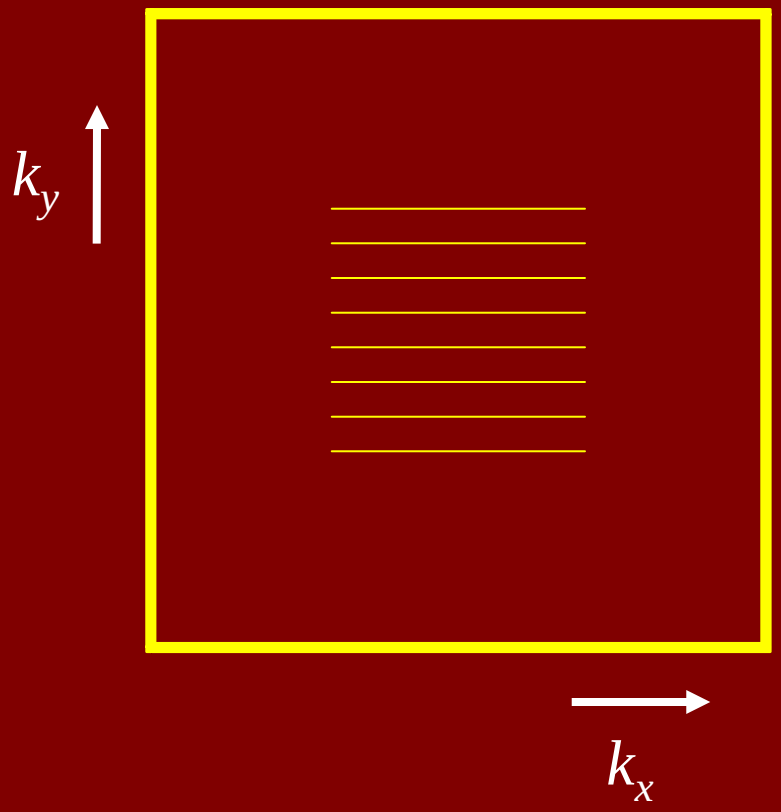
Each row of k-space contains the raw data received under a particular phase gradient, where the order in which the rows are recorded depends on the imaging sequence used; Once all of k-space has been assembled, it is Fourier transformed (2D FFT) to obtain the image

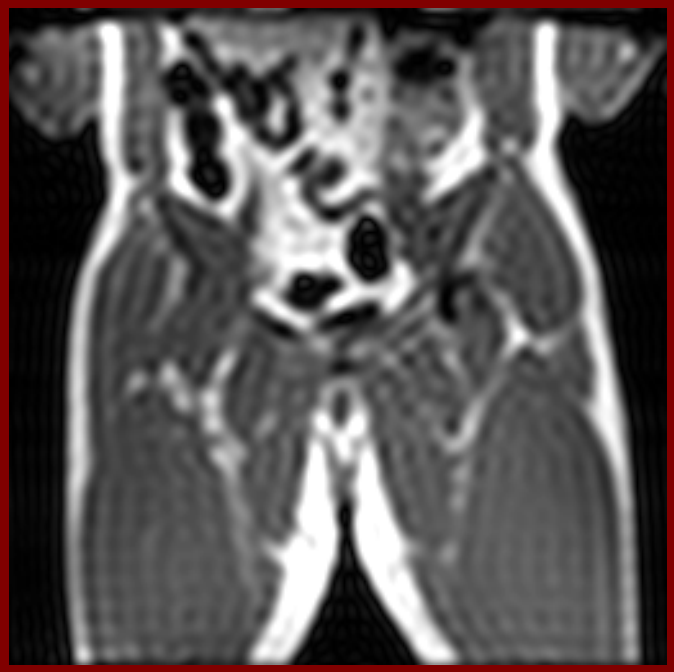
Full *k* Space Coverage



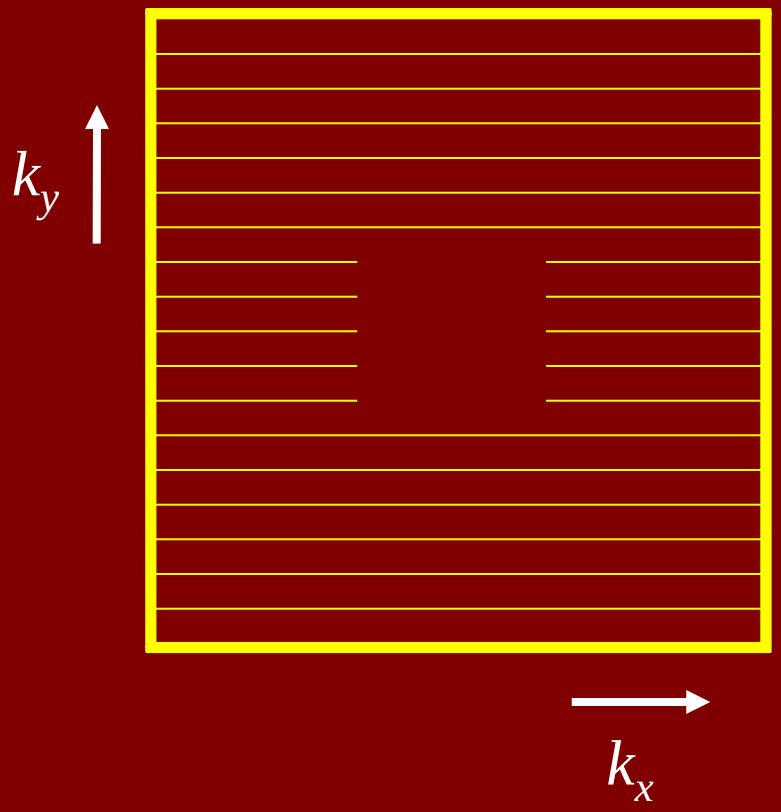


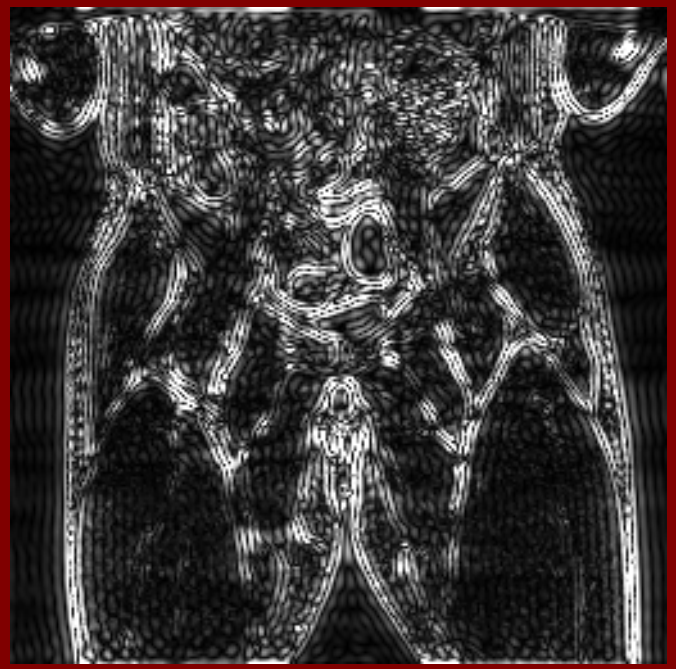
Only Centre of *k* Space





Only Edges of *k* Space





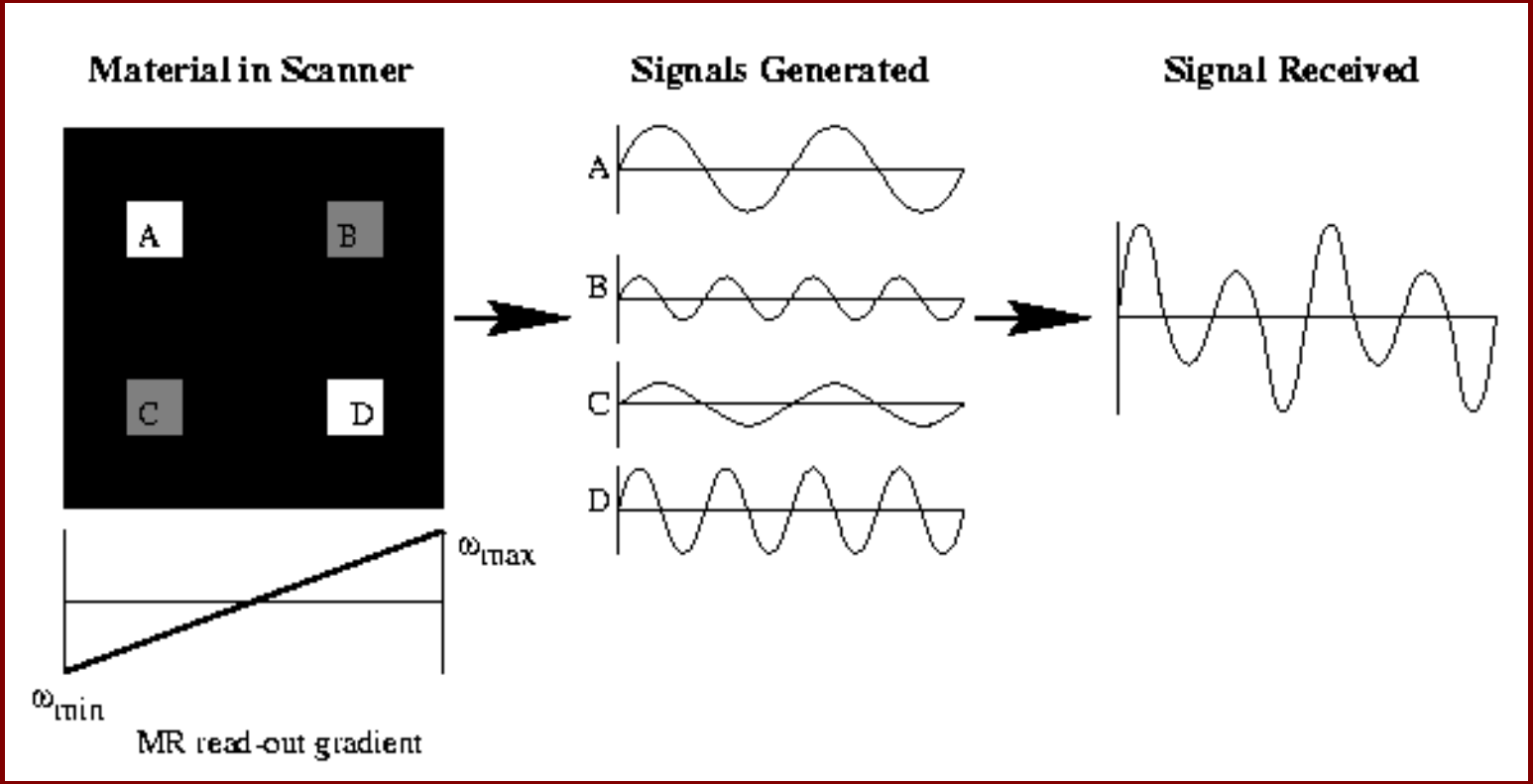
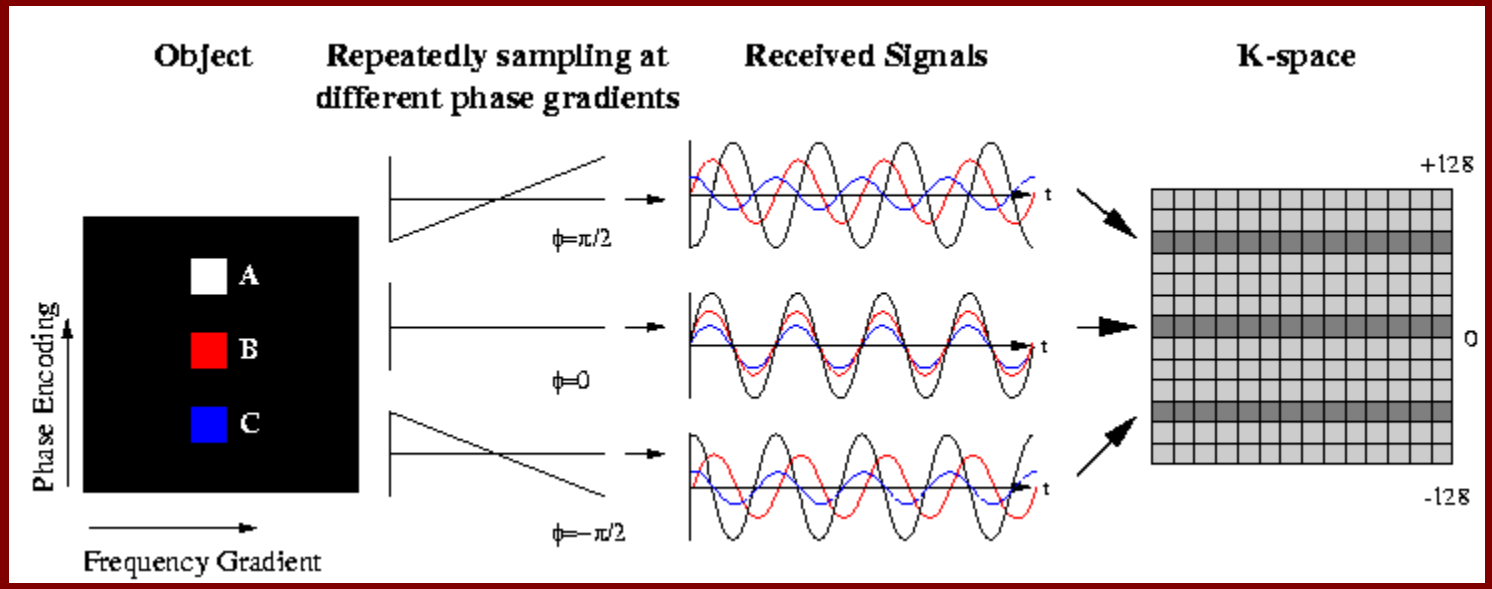


Illustration of the MR read gradient and signals generated at different spatial locations. The illustration shows all the signals in phase which corresponds to the zeroth row of k-space. The material at A and D has a high MR signal, the material at B and C has a low signal. A and C are towards the left so have a low frequency of precession; B and D are on the right so have a higher frequency.

phase-encoding gradients



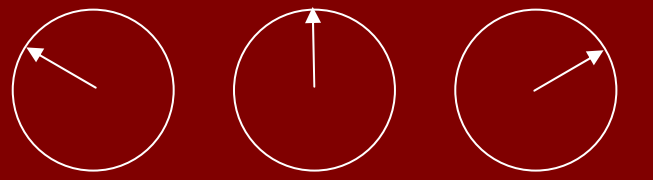
A, B and C have high, medium and low MR signals. They are at the same x-position so have the same frequency of precession.

Their y-positions are encoded by repeated scanning with different phase gradients. With a zero phase gradient (the central row), all the signals are in phase.

With a positive phase gradient (the top row), A has a phase lead, and C has a phase lag with respect to B. Recording the signals under all phase gradients allows the y-positions to be recovered by Fourier Transforming k-space.

Why Use *k* Space?

Larmor equation $\nu = \gamma B_0$



$$\nu(x,y) = \gamma B_0 + \gamma G_x x + \gamma G_y y$$

phase $\phi(x,y,t) = 2\pi \int \gamma B_0 dt + 2\pi \int \gamma G_x x dt + 2\pi \int \gamma G_y y dt$

elemental signal $\delta S(x,y,t) = \rho(x,y) \exp\{i\, \phi(x,y,t)\}$

total signal $S(t) = \iint \rho(x,y) \exp\{i\, \phi(x,y,t)\}\, dx dy$

A Few Substitutions

From: total signal $S(t) = \iint \rho(x,y) \exp\{i\, \phi(x,y,t)\}\, dxdy$

$$k_x(t) = \int \gamma G_x dt \qquad k_y(t) = \int \gamma G_y dt$$

In rotating frame:

To: total signal $S(t) = \iint \rho(x,y) \exp\{2\pi i(k_x x + k_y y)\}\, dxdy$

This is the standard Fourier Equation!

A potted history of MRI 1: NMR

- Felix Bloch & Edward Purcell (1946) NP'52
- Mostly uses oxygen and potassium nuclei – eg ATP consumption in the heart
- Oxford has made massive contributions … Rex Richards, Walter Bodmer, George Radda, …
- Continues for spectroscopy … currently 18T systems for protein structure (OI/Brucker)
- NMR says ***that*** there is a substance present in a compound but can't say where

Many of the terms and concepts in MRI originate in NMR:

T_1, T_2, flip angle, pulse sequence, …

A potted history of MRI

- 1973 Paul Lauterbur demonstrated MRI using the back projection ideas introduced in CT (the same year) NP ‘03
- 1975 Richard Ernst introduced the modern technique *k-space* NP ‘91
- 1977 Peter Mansfield introduced EPI – basis for all modern fast imaging & fMRI NP ‘03
- 1980 first whole body image, 1986 first clinical systems, 1987 first research cardiac motion sequence, 1989 Siemens – OI joint venture to make superconducting magnets,
- 1989-1997 competitive advantage based on technological superiority; 1998 ➔ MR system becomes commoditized – competitive advantage based on systems integration & software