G16.4426/EL5823/BE6203 Medical Imaging

Physics of Magnetic Resonance Imaging

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Outline

- Brief introduction to MRI
- Properties of the nuclear spins
- Precession and equilibrium magnetization
- RF excitation
- Relaxation
- Contrast mechanism

VALUE OF INNOVATION

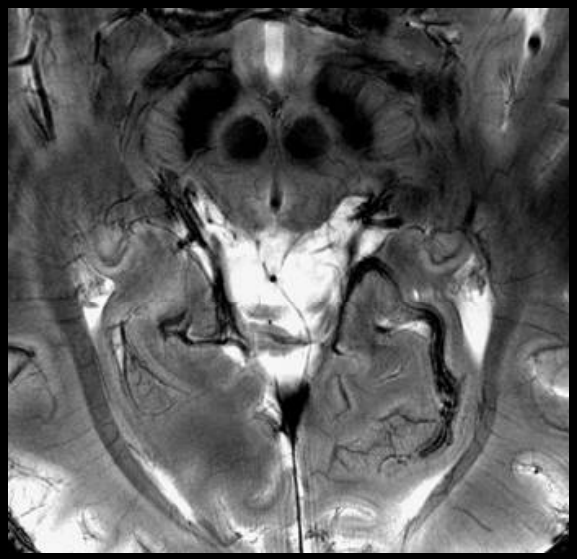
Physicians' Views Of The Relative Importance Of Thirty Medical Innovations

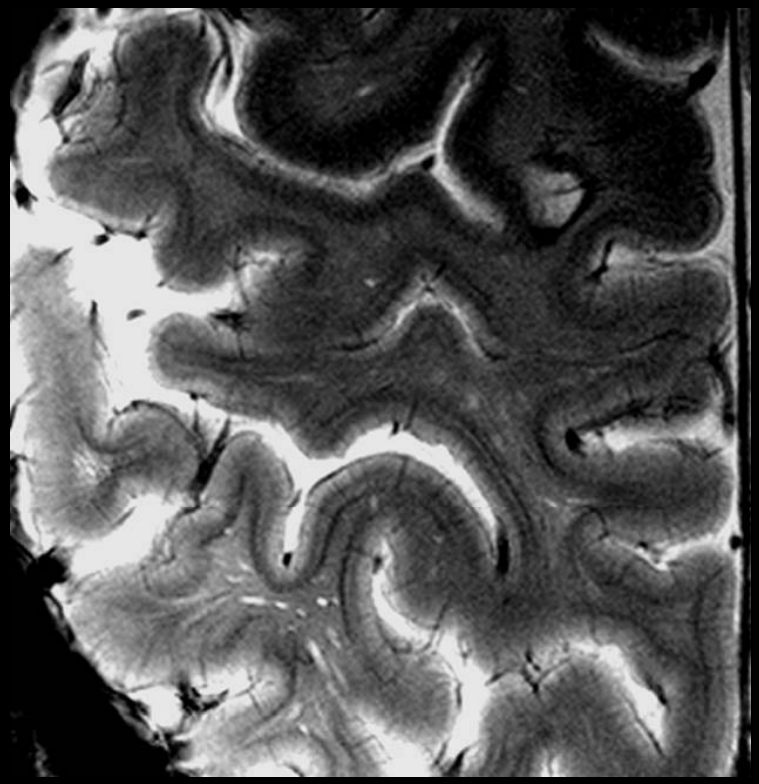
A survey of leading general internists provides a useful consensus on the relative importance of innovations to their patients.

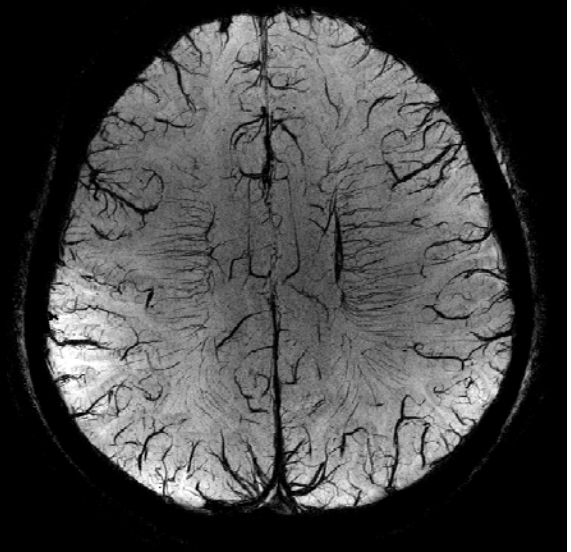
by Victor R. Fuchs and Harold C. Sox Jr.

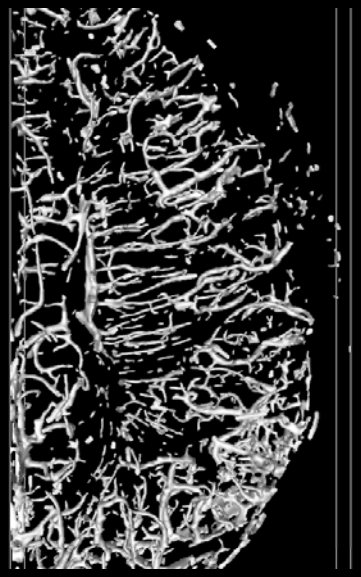
HEALTH AFFAIRS - Volume 20, Number 5

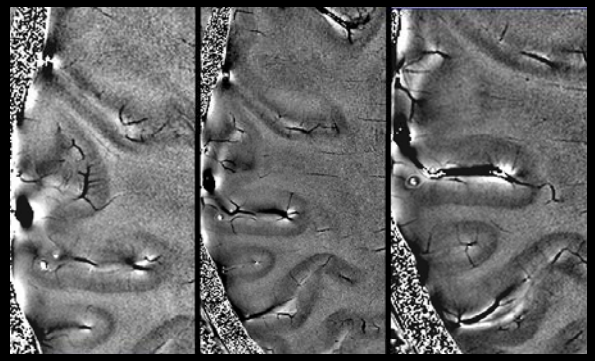
Rank	Innovation	Mean score[a]	Most	Not most or least	Least
1	MRI and CT scanning	0.878	75.6%	24.4%	0.0%
2	ACE inhibitors	0.767	54.2	44.9	0.9
3	Balloon angioplasty	0.758	53.8	44.0	2.2
4	Statins	0.736	48.0	51.1	0.9
5	Mammography	0.733	47.6	51.6	0.9
6	CABG	0.693	40.4	57.8	1.8
7	Proton pump inhibitors and H2 blockers	0.687	40.0	57.3	2.7
8	SSRIs and recent non-SSRI antidepressants	0.678	39.6	56.4	4.0
9	Cataract extraction and lens implant	0.651	38.2	53.8	8.0
10	Hip and knee replacement	0.649	31.6	66.7	1.8
11	Ultrasonography	0.647	31.1	67.1	1.8
12	Gastrointestinal endoscopy	0.624	28.0	68.9	3.1
13	Inhaled steroids for asthma	0.591	23.6	71.1	5.3
14	Laparoscopic surgery	0.558	20.9	69.8	9.3
15	NSAIDs and Cox-2 inhibitors	0.531	14.2	77.8	8.0
16	Cardiac enzymes	0.498	7.1	85.3	7.6
17	Fluoroquinolones	0.487	6.7	84.0	9.3
18	Recent hypoglycemic agents	0.478	12.9	69.8	17.3
19	HIV testing and treatment	0.444	15.6	57.8	26.7
20	Tamoxifen	0.440	3.1	81.8	15.1
21	PSA testing	0.438	12.9	61.8	25.3
22	Long-acting and parenteral opioids	0.376	8.4	58.2	33.3
23	H. Pylori testing and treatment	0.351	1.8	66.7	31.6
24	Bone densitometry	0.344	4.0	60.9	35.1
25	Third-generation cephalosporins	0.329	1.8	62.2	36.0

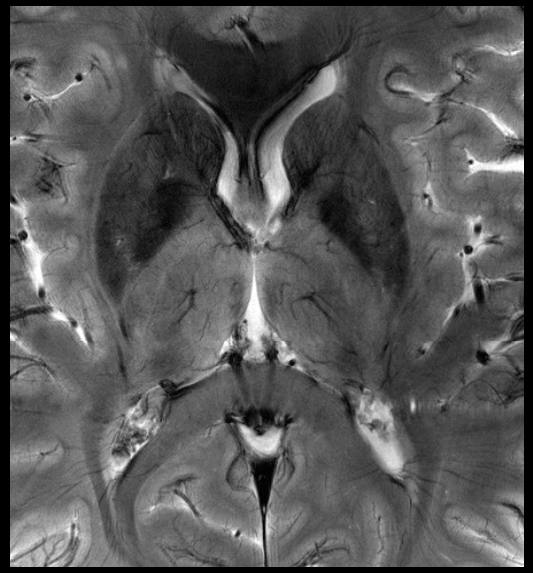












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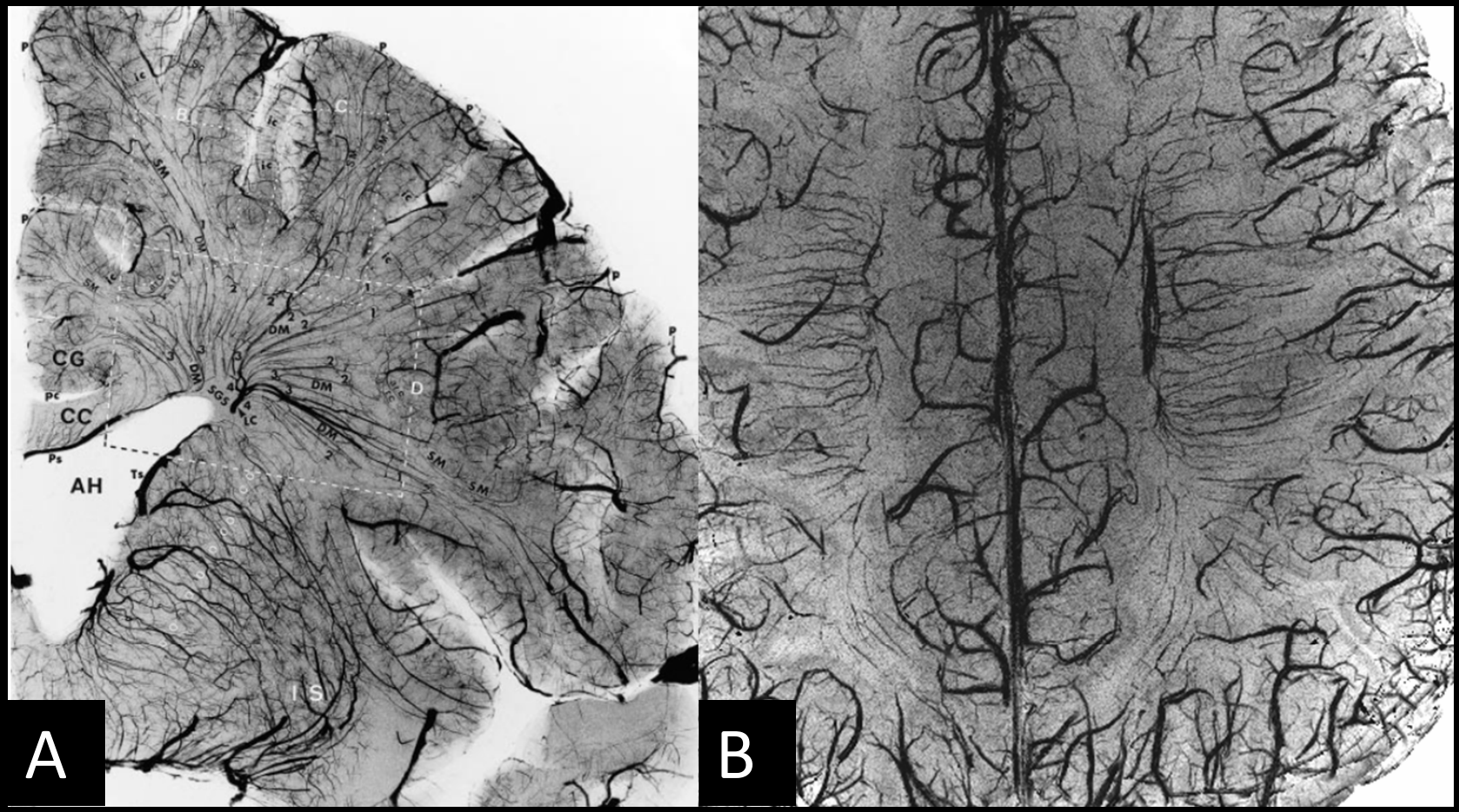
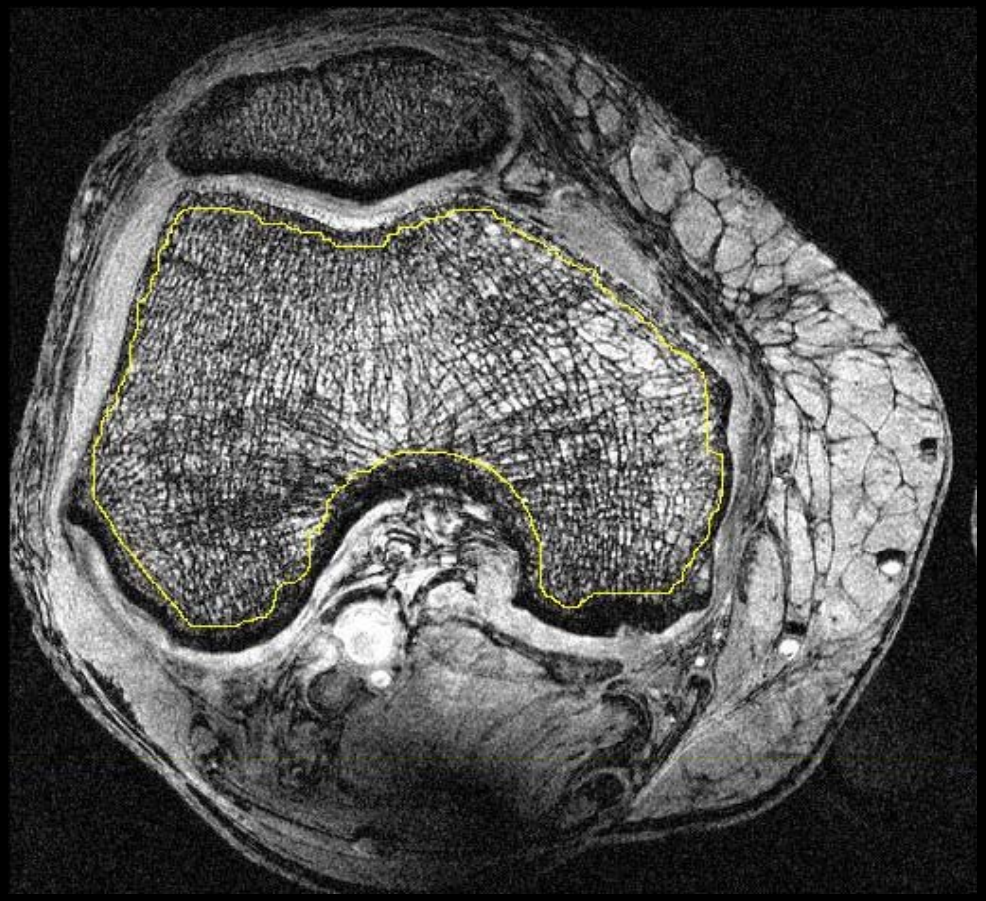
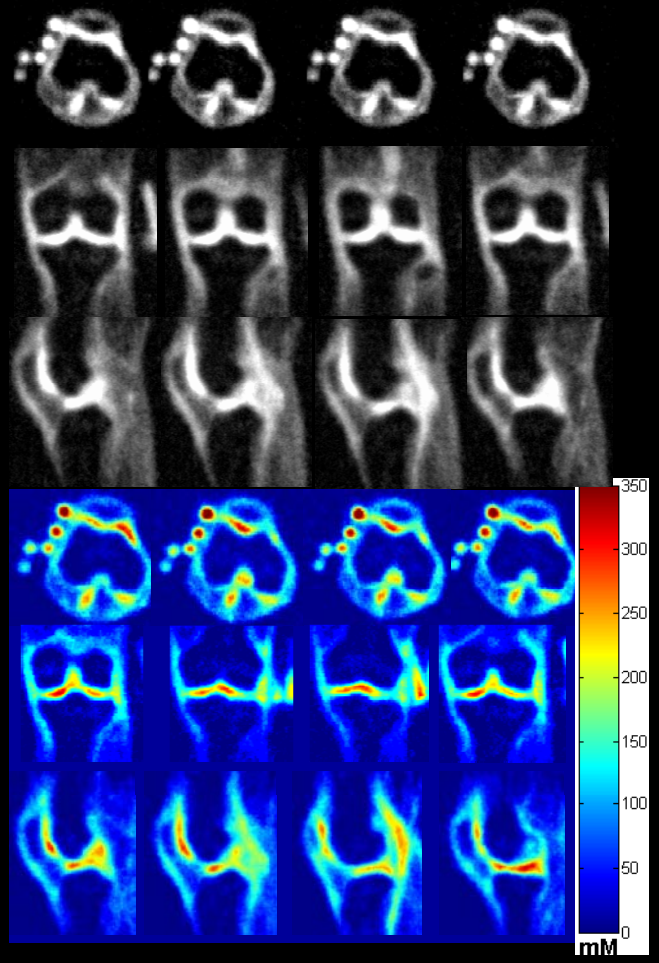
Ex-Vivo Pathology or In-Vivo Imaging?



High-Field MRI and Arthritis



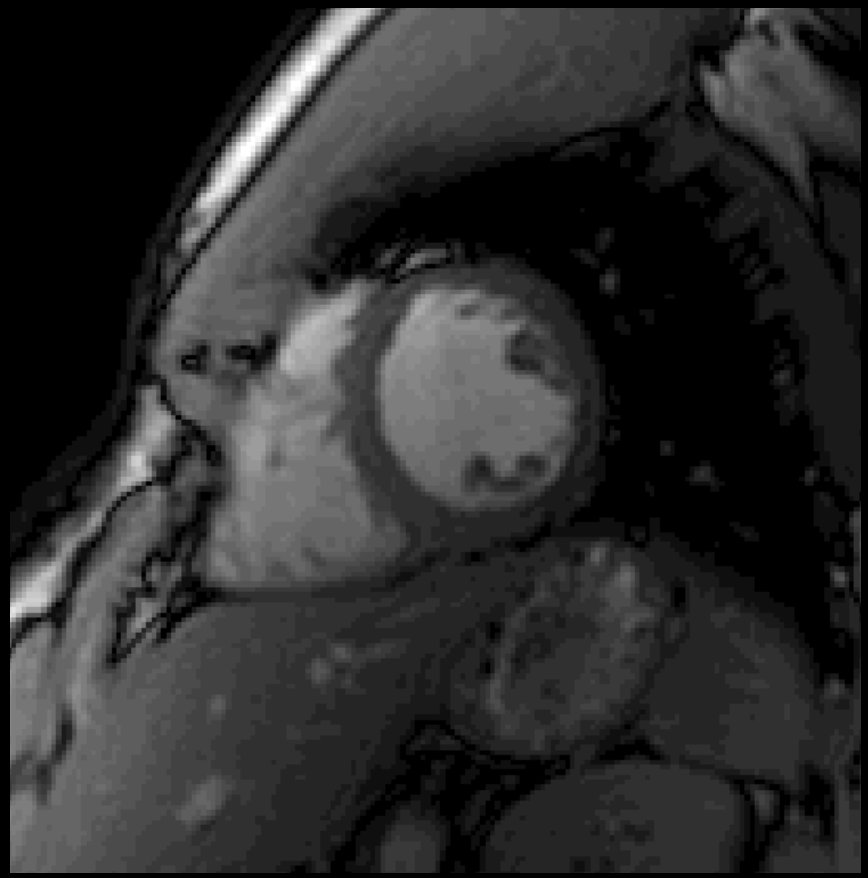
High-resolution knee imaging at 7T



3D sodium imaging for osteoarthritis

Ravinder Regatte, CBI

High-Resolution Cardiac MRI



Li Feng, CBI

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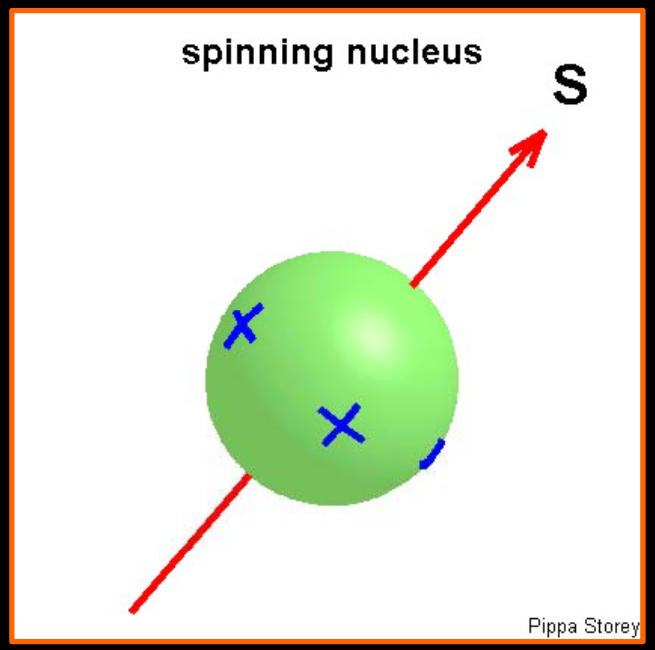
Magnetic Resonance Imaging

- Provide high resolution anatomic structure
- Provide high contrast between different soft tissues (X-Ray and CT cannot!)
  - Contrast can be manipulated
- No exposure to harmful ionizing radiations
- Increased complexity
- Long scan times
  - Uncomfortable for patients
  - Susceptible to patient motion

Nuclear Spin

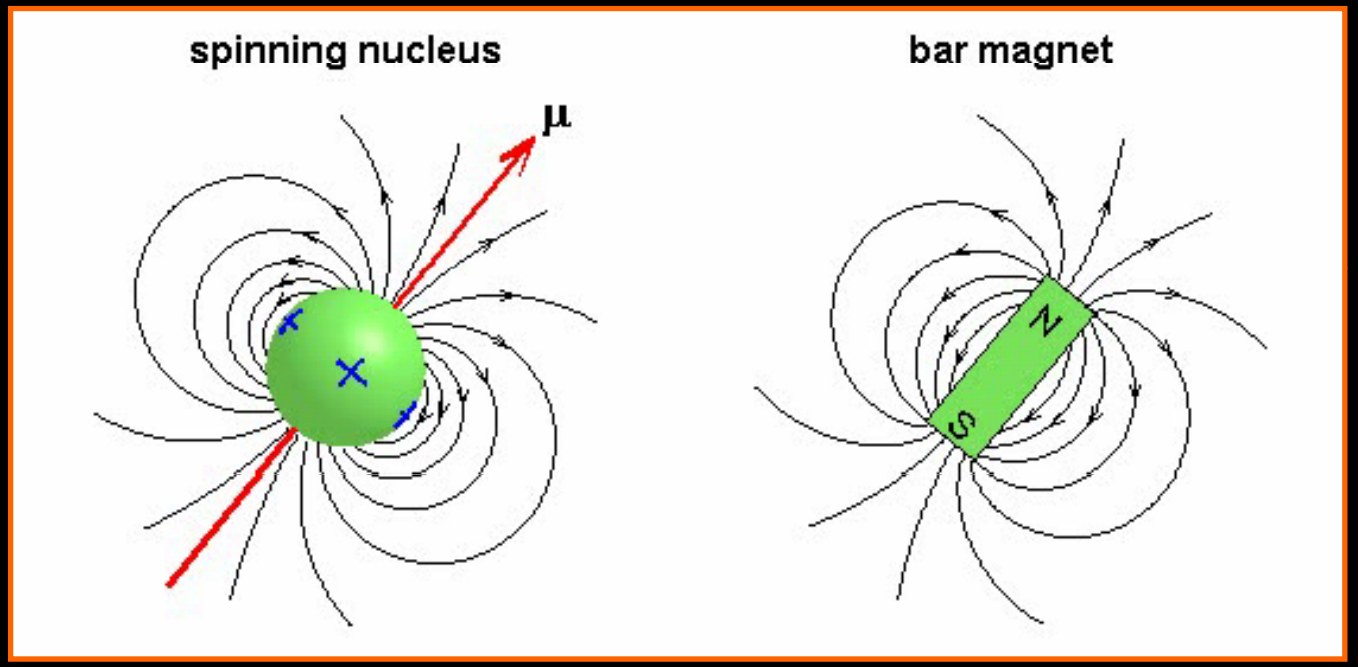
- Nuclei consist of protons and neutrons, so are positively charged
- Nuclei with an odd number of protons and/or neutrons have intrinsic spin
  - Spin is a fundamental property of nature like electrical charge or mass. Spin comes in multiples of 1/2 and can be + or -.



- The circulating charge is equivalent to a current loop
  - Produces its own magnetic field
  - Experiences a torque in the presence of an external magnetic field

Nuclei As Tiny Magnets

- Nuclei with intrinsic spin therefore behave like tiny magnets



- Each nucleus:
  - Has a magnetic moment μ which is parallel to its spin
  - Produces its own tiny magnetic field
  - Experiences a torque in the presence of an external magnetic field

Spin and Magnetic Moment

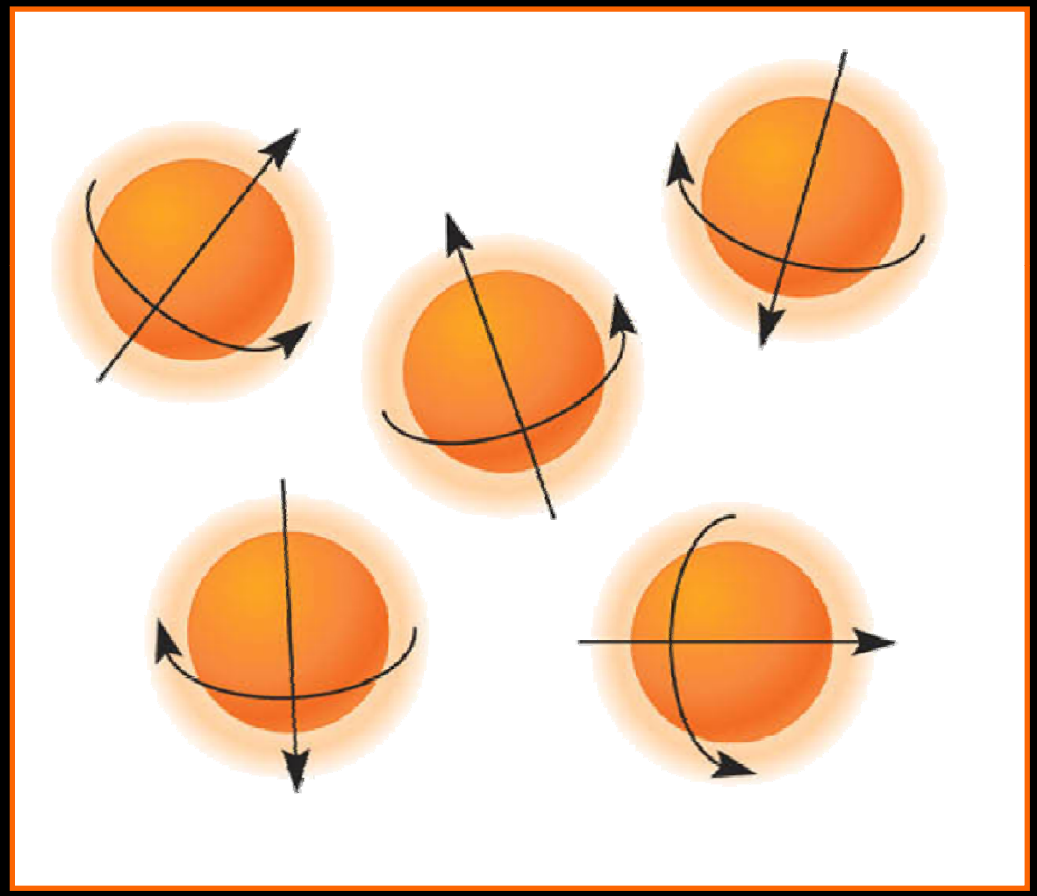
- The magnetic moment of a nucleus arises from circulation of charge due to the nuclear spin
- The magnetic moment is therefore parallel to the spin (angular momentum) axis:

$$\boldsymbol{\mu} = \gamma \mathbf{S}$$

($\boldsymbol{\mu}$ and $\mathbf{S}$ are vector quantities)

- The constant of proportionality γ is known as the gyromagnetic ratio and depends on the nucleus

Nuclei in the Absence of an External Magnetic Field



In the absence of an external magnetic field, the spins are oriented randomly

Magnetic dipole in an external magnetic field

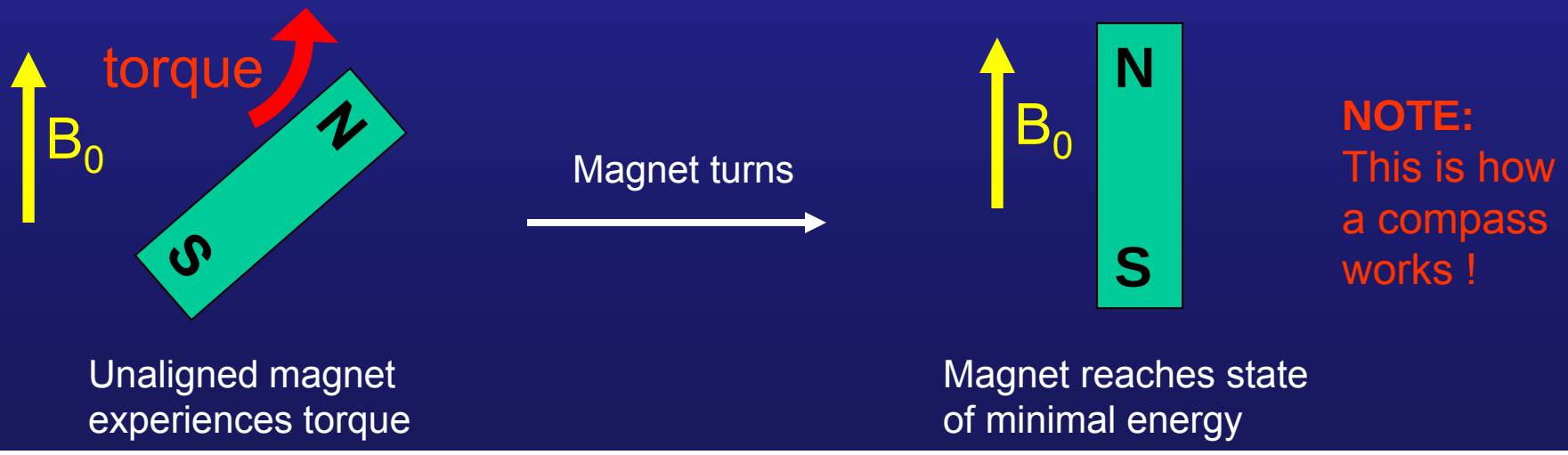
The energy of magnetic dipole is lower when it is aligned with the external magnetic field

$$E = -\mu \cdot \mathbf{B}$$ (dot product)

The magnetic dipole therefore experiences a torque that tries to turn it in the direction of the field

$$\tau = \mu \times \mathbf{B}$$ (cross product)

Case of simple bar magnet :



Behavior of nuclei in an external magnetic field

Nuclei behave differently from simple bar magnets when exposed to an external magnetic field

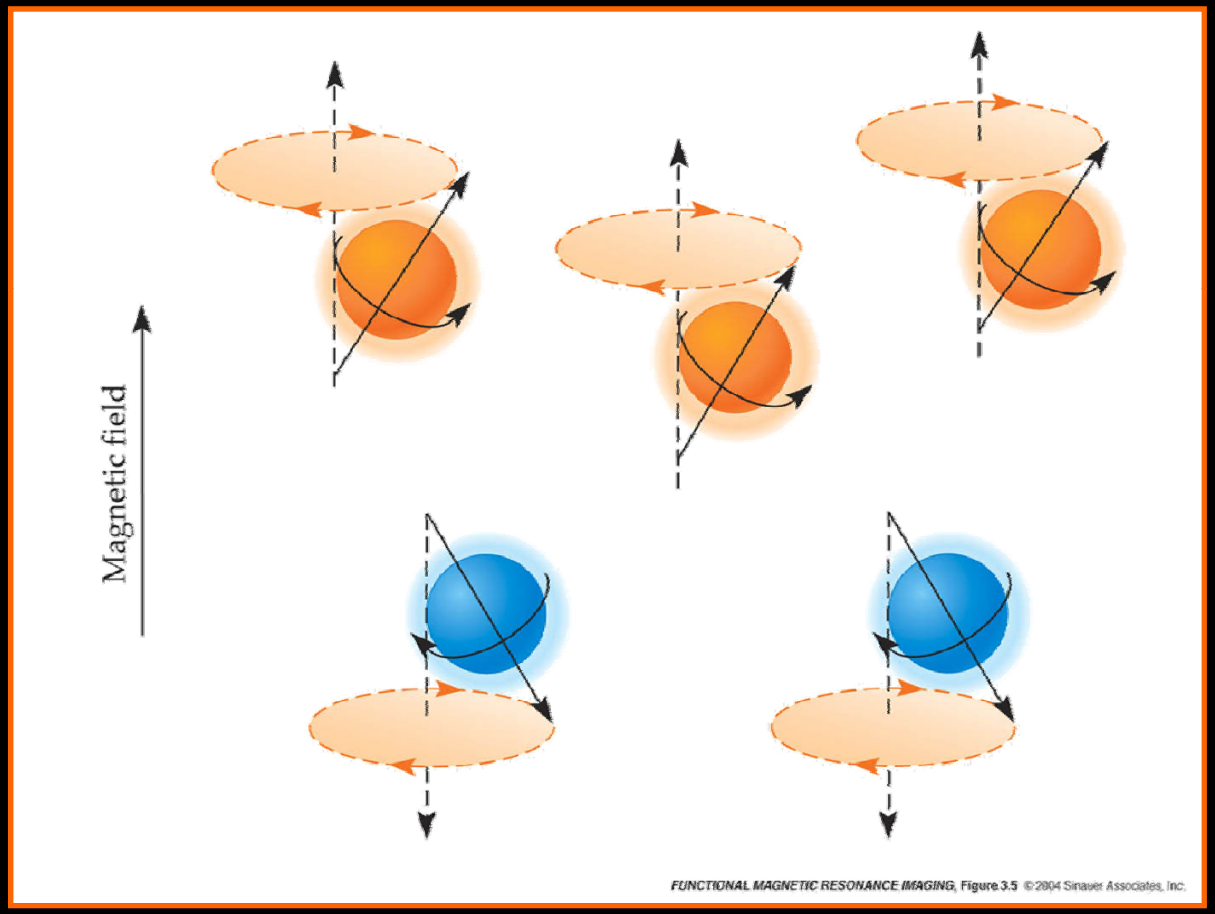
This is because nuclei with magnetic moments also have angular momentum (their spin)

Since the magnetic moment is locked to the spin axis, the nucleus does not simply tip into alignment with the magnetic field, but instead precesses around it

This is analogous to the behavior of a gyroscope in a gravitational field

Precession

Nuclei in an External Magnetic Field



FUNCTIONAL MAGNETIC RESONANCE IMAGING, Figure 3.5 ©2004 Sinauer Associates, Inc.

In the presence of an external magnetic field, the spins align with its direction and precess about it

Precession of nuclei in an external **B** field

Torque causes a change in angular momentum (in this case the spin)

$$\frac{d\mathbf{S}}{dt} = \boldsymbol{\tau}$$

However the torque is perpendicular to the magnetic moment

$$\boldsymbol{\tau} = \boldsymbol{\mu} \times \mathbf{B}$$

and the magnetic moment is locked to the spin axis

$$\boldsymbol{\mu} = \gamma \mathbf{S}$$

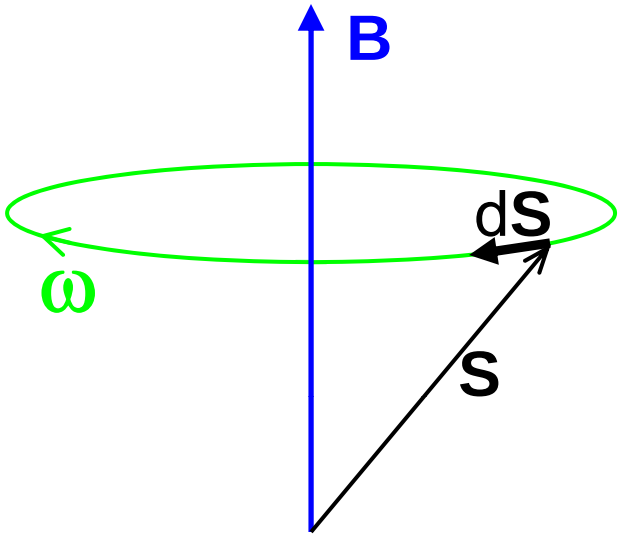
Hence the change in spin is perpendicular to the spin itself

$$\frac{d\mathbf{S}}{dt} = \gamma \mathbf{S} \times \mathbf{B}$$

This means that the spin changes direction but not magnitude

i.e. The spin precesses about the external field

Geometrical representation



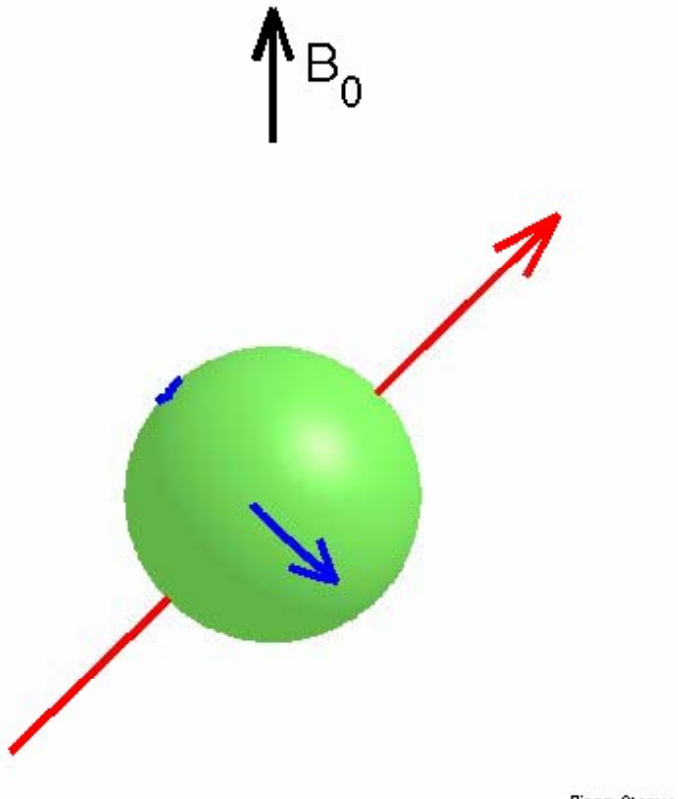
The frequency of precession is given by

$$\omega = \gamma B$$

and is known as the **Larmor frequency**

(Note that ω and B in the above equation are both scalar quantities. The direction of precession actually obeys a left-hand rule. Hence the vector form of the above equation would be: $\omega = -\gamma \mathbf{B}$.)

Precession of a nucleus



In MRI, the external static magnetic field is generally denoted $\mathbf{B_0}$

Larmor frequency: $\omega_0 = \gamma B_0$

The field strengths typically used in MRI range from about

$B_0 \sim 0.5 – 11$ tesla (T)

where 1 tesla = 10^4 gauss

Some of the nuclear isotopes used in MRI

Nuclear isotope	Natural abundance [%]	Net spin [ħ]	$\gamma/2\pi$ [MHz/T]
^{1}H	99.98	½	42.58
^{2}H	0.015	1	6.53
^{3}He	0.00014 ‡	½	32.44
^{13}C	1.11	½	10.71
^{19}F	100	½	40.05
^{23}Na	100	3/2	11.26
^{31}P	100	½	17.23
^{129}Xe	26.44	½	11.84

‡ The helium-3 used in magnetic resonance studies is derived from tritium decay

Natural abundance is the percentage of an element that is present in the form of a particular isotope e.g. 1.11% of carbon occurs in the form of ^{13}C.

The gyromagnetic ratios are of the order of megahertz (MHz) per tesla (T). Hence at the field strengths typically used in MRI the Larmor frequencies are in the MHz (radiofrequency) range.

Signal in MRI

Each nucleus generates its own tiny magnetic field, and as it precesses, that magnetic field will oscillate at the Larmor frequency.

The net magnetic field oscillations generated by all the nuclei in the sample can be detected by an RF receiver coil and constitute the MR signal.

However:

At equilibrium the protons have random phases

$\Rightarrow$ There is no coherent oscillation and no signal

To generate a signal, we first need to make the nuclei precess in phase. This is achieved through a process known as radiofrequency (RF) excitation.

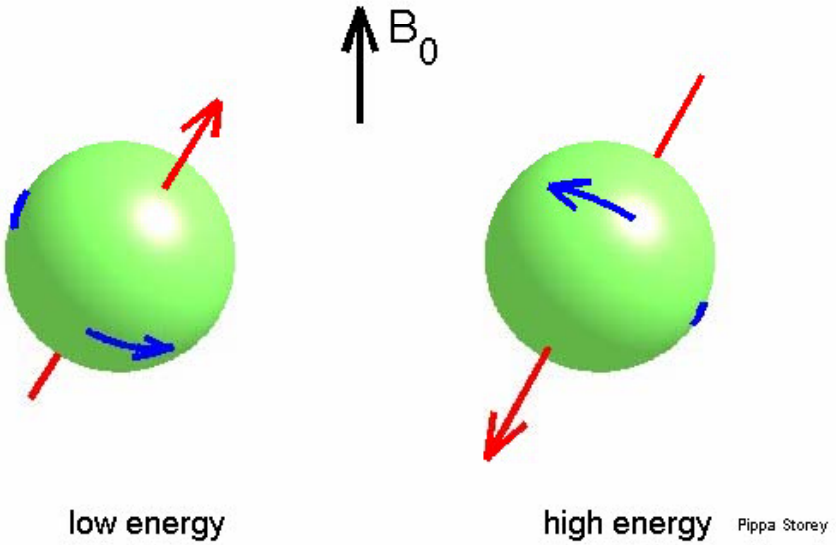
RF excitation (preliminaries)

The term ‘RF excitation’ derives from the fact that it involves imparting energy to the nuclei, or ‘exciting’ them.

As noted earlier, the energy of a nucleus depends on the orientation of its magnetic moment with respect to the external field

$$E = - \mu \cdot \mathbf{B}_0$$ (dot product)

At equilibrium, more spins are oriented towards the magnetic field (low energy) than away from it (high energy).



This produces a net longitudinal magnetization along the direction of the field.

Equilibrium magnetization

The amplitude of the MR signal is ultimately limited by the magnitude of the equilibrium magnetization $\mathbf{M}_0$.

M_0 is determined by the vector sum of the nuclear magnetic moments. It therefore depends on the relative populations of the various spin states. These depend in turn on the energies of the states through the Boltzmann distribution.

In general, it can be shown that M_0 is given by

$$M_0 = \frac{\rho\gamma^2\hbar^2 I(I+1)B_0}{3kT}$$

where ρ is the number of nuclei per unit volume, I is their total spin, *k* is Boltzmann's constant and *T* is the absolute temperature.

Observations about the equilibrium magnetization

From the formula for the equilibrium magnetization

$$M_0 = \frac{\rho \gamma^2 \hbar^2 I(I+1) B_0}{3kT}$$

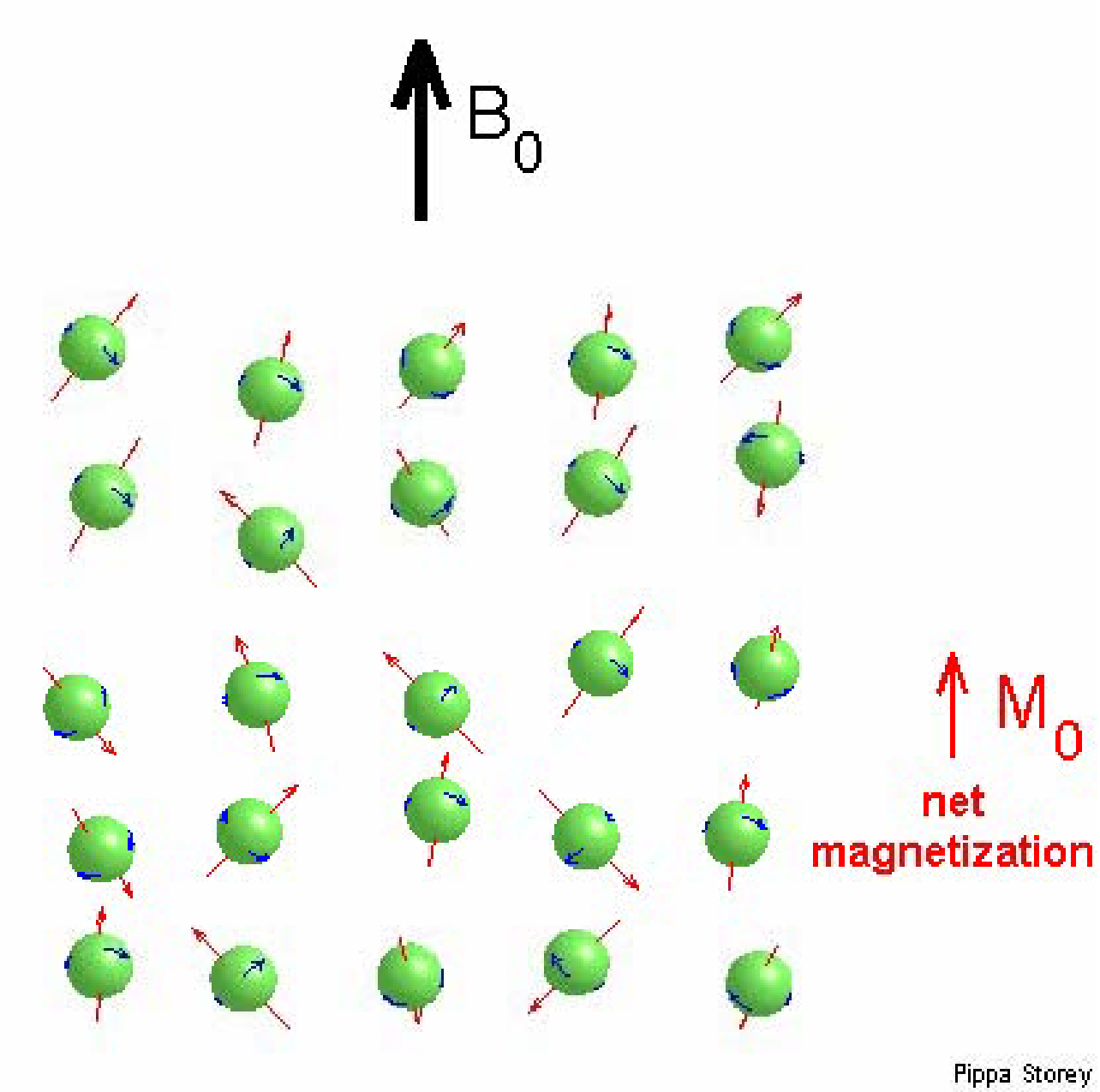
it can be seen that M_0

- depends linearly on the field strength B_0
- depends quadratically on the gyromagnetic ratio γ

Hence the signal is stronger

- at higher field strength
- from a nucleus with higher gyromagnetic ratio

Equilibrium magnetization



RF excitation (continued)

To obtain an MR signal we must tip the net magnetization away from the $\mathbf{B}_0$ direction.

Since the magnetization **M** is just the vector sum over all the nuclear magnetic moments

$$\mathbf{M} = \frac{1}{V}\sum_i \boldsymbol{\mu}_i$$

it obeys the same equation of motion as the spins

$$\frac{d\mathbf{M}}{dt} = \gamma \mathbf{M} \times \mathbf{B}$$

(assuming non-interacting spins)

Therefore, when we tip the magnetization away from the B_0 direction, it will rotate at the Larmor frequency.

On a microscopic level, this is equivalent to exciting the nuclei and making them precess in phase.

Nuclear Magnetic Resonance

- Nuclei have natural frequency of precession (Larmor frequency)
- Their energy depends on their alignment with B_0

  (small angle = low energy)
- Energy is transferred to the nuclei by applying a rotating magnetic field B_1 in the transverse plane whose frequency ω equals the Larmor frequency

$\omega = \omega_0$ 'Resonance condition'

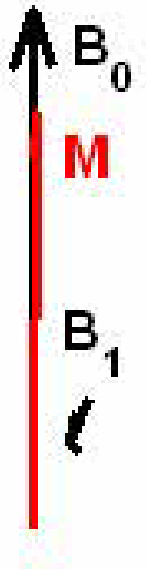
- As the nuclei absorb energy, the magnetization tips away from B_0

  (bigger angle = higher energy)
- The resulting transverse magnetization rotates around B_0

$\Rightarrow$ MRI signal

Excitation and signal detection

Excitation



Pippa Storey

To summarize so far…

- Apply a transverse magnetic field B_1 at the Larmor frequency
- Magnetization is tipped away from direction of B_0
- Spins precess in phase
- When switch off field, magnetization continues to rotate
- Signal comes from transverse component of magnetization
- Spins eventually dephase and the signal dies away
  (transverse relaxation)
- Spins lose energy and return to equilibrium
  (longitudinal relaxation)

Relaxation processes are important for producing signal contrast and will be discussed in more detail later

Rotating frame

An alternative, and very useful, way of visualizing RF excitation is in a reference frame rotating with B_1.

In this frame

- B_1 appears stationary
- B_0 can be ignored (providing B_1 is exactly on resonance) since its effect is already accounted for in the rotation of the reference frame itself

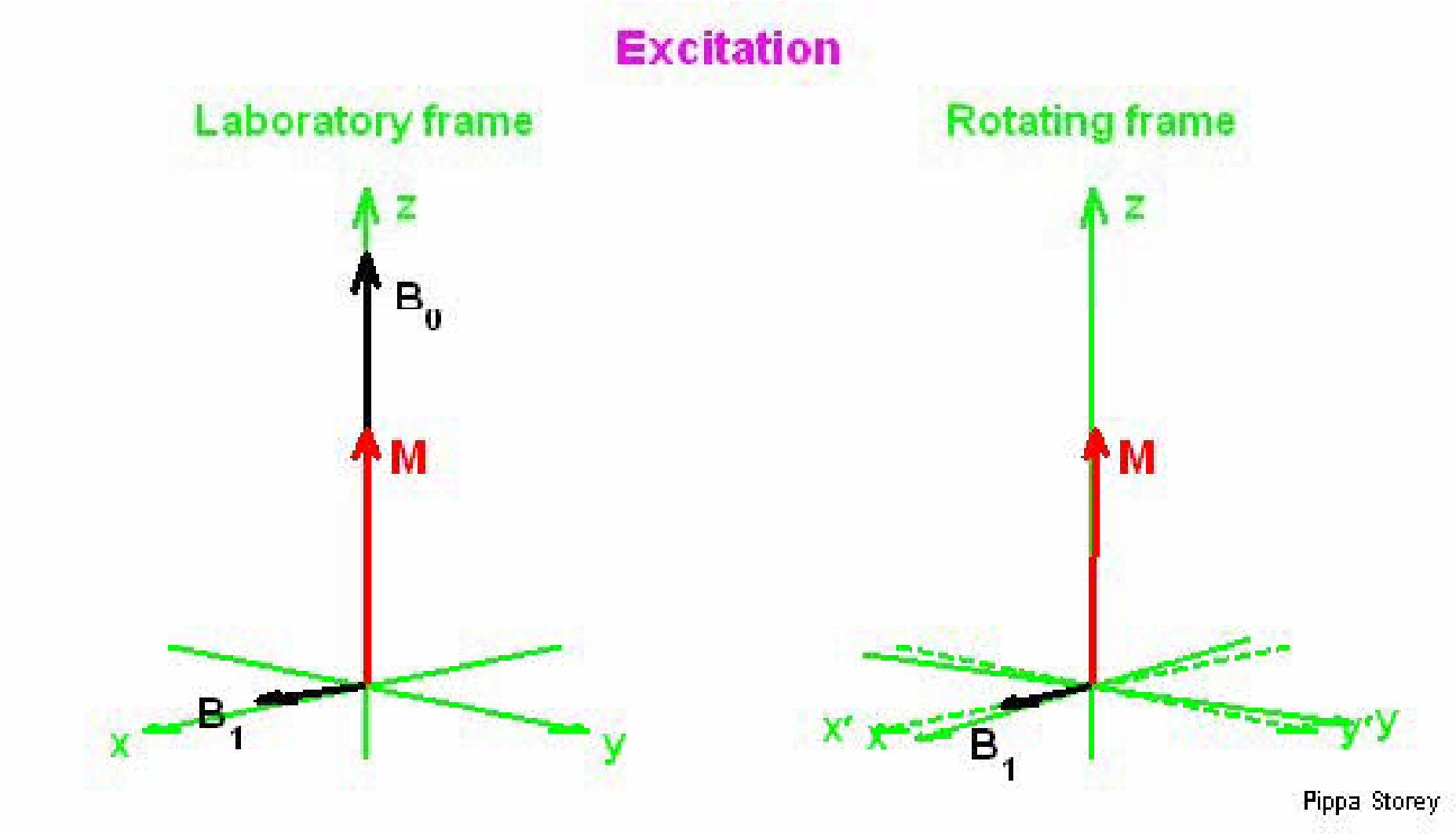
The effect of B_1 in the rotating frame is therefore identical to the effect of B_0 in the laboratory frame:

The magnetization rotates about B_1 with a frequency

$$\omega_1 = \gamma B_1$$

in analogy to Larmor precession.

Excitation and relaxation in the rotating frame



RF pulses and flip angle

In MRI the B_1 field is applied in short intense bursts known as RF pulses.

The amplitude and duration of each pulse is chosen to tip the magnetization through a desired angle, known as the flip angle (FA).

Assuming the RF pulse is exactly on resonance, we have

$$\omega_1 = \gamma B_1$$

from which the flip angle can easily be calculated

$$FA = \gamma B_1 \tau$$

where τ is the duration of the RF pulse.

Equation of motion in rotating frame

The previous arguments can be justified (and generalized to off-resonance situations) by deriving the equation of motion for the magnetization in the rotating frame.

For non-interacting spins

$$\frac{d\mathbf{M}}{dt} = \gamma \mathbf{M} \times \mathbf{B}$$ (laboratory frame)

In a reference frame rotating with angular velocity ω_{rot} this is

$$\frac{d\mathbf{M}_{rot}}{dt} = \gamma \mathbf{M}_{rot} \times \mathbf{B}_{eff}$$ (rotating frame)

where

$$\mathbf{B}_{eff} = \mathbf{B}_{rot} + \frac{\boldsymbol{\omega}_{rot}}{\gamma}$$

and the subscripts 'rot' indicate vectors in the rotating frame.

If we choose $\omega_{rot} = -\omega \mathbf{k}$ (the direction of precession for $\mathbf{B}_0 = B_0 \mathbf{k}$)

$$\frac{d\mathbf{M}_{rot}}{dt} = \mathbf{M}_{rot} \times \left(\gamma \mathbf{B}_{rot} - \omega \mathbf{k} \right)$$ (rotating frame)

Interpretation of equation of motion in rotating frame

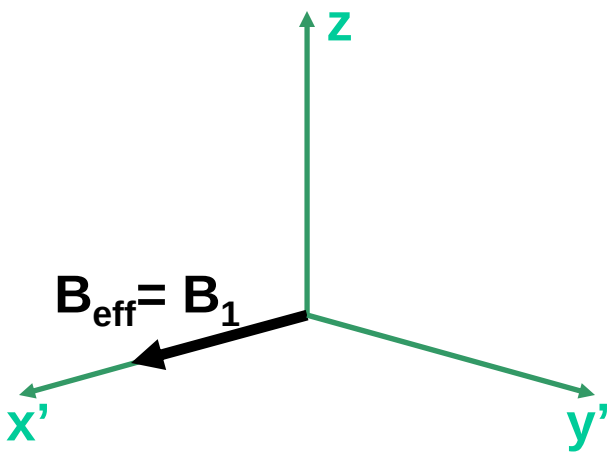
Suppose we have a static magnetic field $\mathbf{B}_0 = B_0\mathbf{k}$ and an RF field $\mathbf{B}_1(t)$ rotating in the x-y plane with rotational velocity $\omega_{rot} = -\omega\mathbf{k}$.

We choose a frame of reference rotating ***with the*** $\mathbf{B}_1(t)$ ***field***. In this frame $\mathbf{B}_1$ appears ***stationary***.

Then:

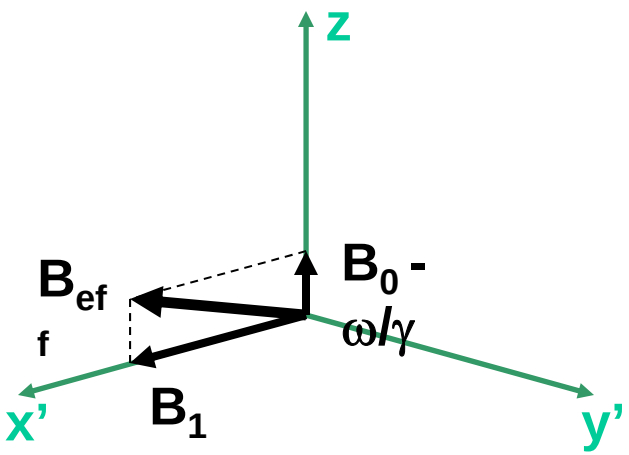
on resonance

$\omega = \omega_0$



off resonance

$\omega \neq \omega_0$



where x' and y' denote axes in rotating frame

Relaxation processes

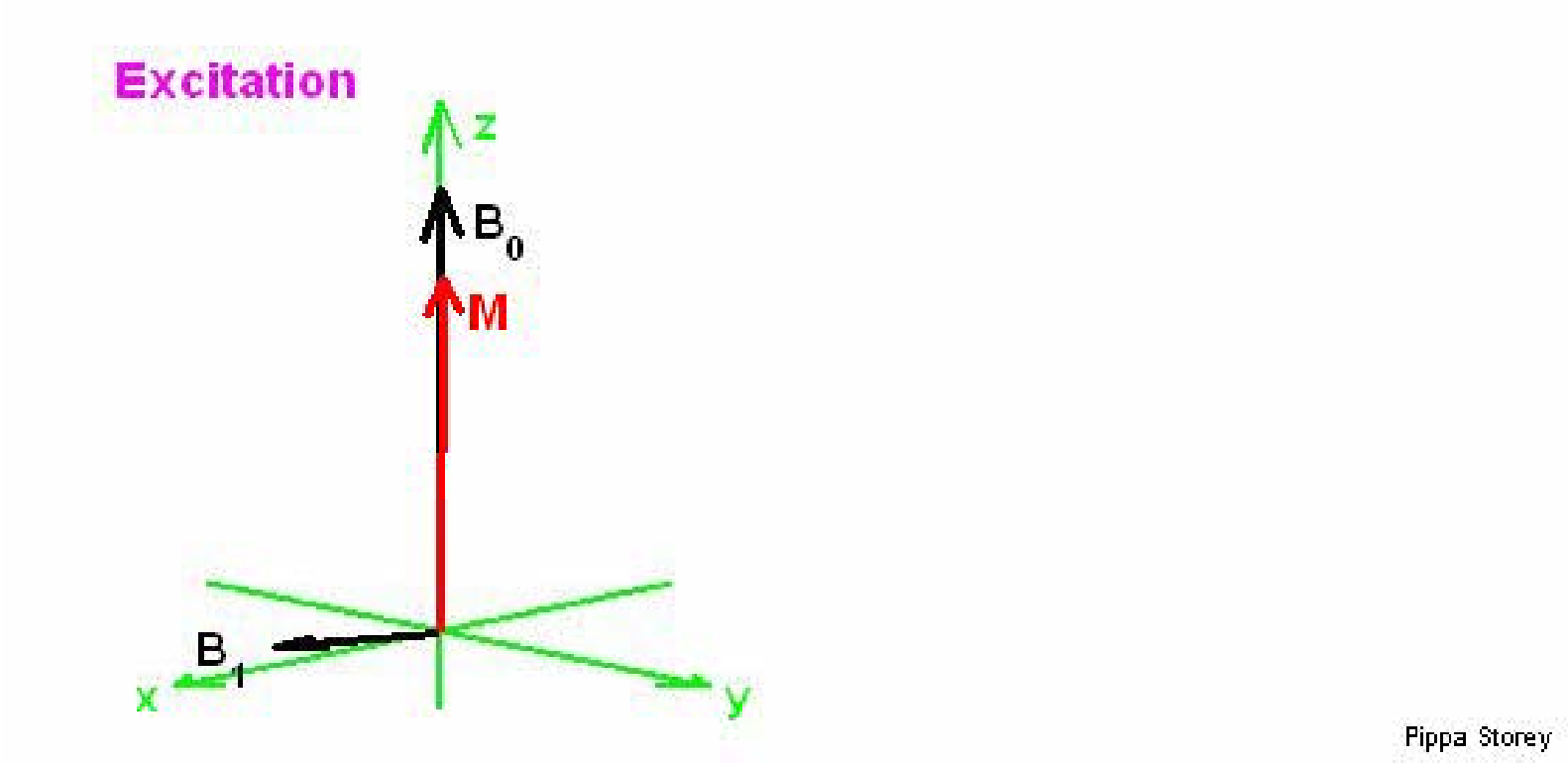
To produce an MRI signal, the spins are excited using a transverse RF field at the Larmor frequency

The spins then precess in phase, creating a rotating magnetic field that can be detected with an RF coil

The signal will not persist indefinitely due to relaxation processes

- Dephasing (transverse relaxation)
  - Different spins precess at slightly different frequencies and gradually lose their phase coherence
  - Causes the signal to decay
  - Relaxation times T2 and T2*
- Energy loss (longitudinal relaxation):
  - Spins lose energy to their environment
  - Causes the magnetization to return to equilibrium
  - Relaxation time T1

Free induction decay (FID)



Following excitation, the magnetization relaxes

- **Dephasing** attenuates the transverse component of the magnetization and causes the signal to decay
- **Energy loss** causes the longitudinal component of the magnetization to recover to its equilibrium value

These processes occur simultaneously, but dephasing occurs faster than energy loss.

Relaxation and signal contrast

Relaxation processes limit the available acquisition time and broaden the spectroscopic linewidths

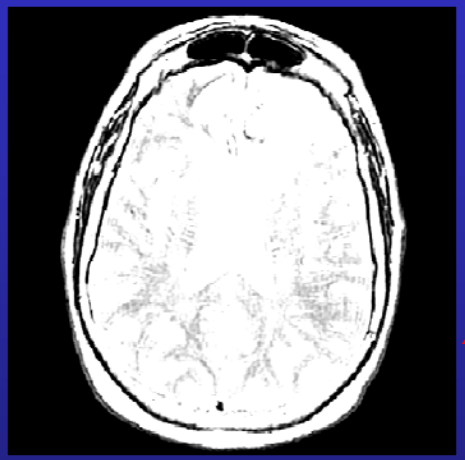
However, because they involve inter-nuclear and intermolecular forces, their rates vary with the molecular environment

This means that they can be exploited to achieve signal contrast among different tissues

Importance of signal contrast

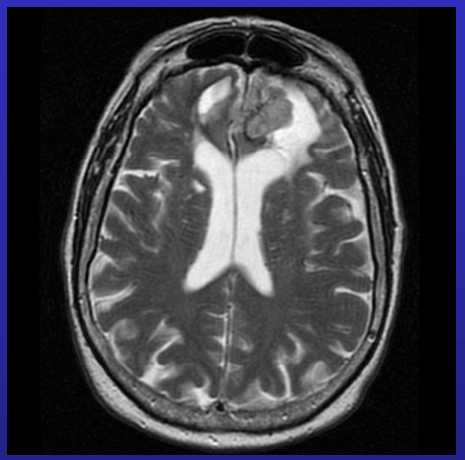
We are familiar with the importance of achieving high signal (specifically high signal-to-noise ratio or 'SNR')

However, in imaging we don't want high signal everywhere



High signal everywhere
$\Rightarrow$ not diagnostic

X



Good signal contrast

It is equally important to achieve good signal contrast (specifically good contrast-to-noise ratio or 'CNR')

An important source of signal contrast is relaxation processes, which vary among different tissues

Sources of signal contrast in MRI

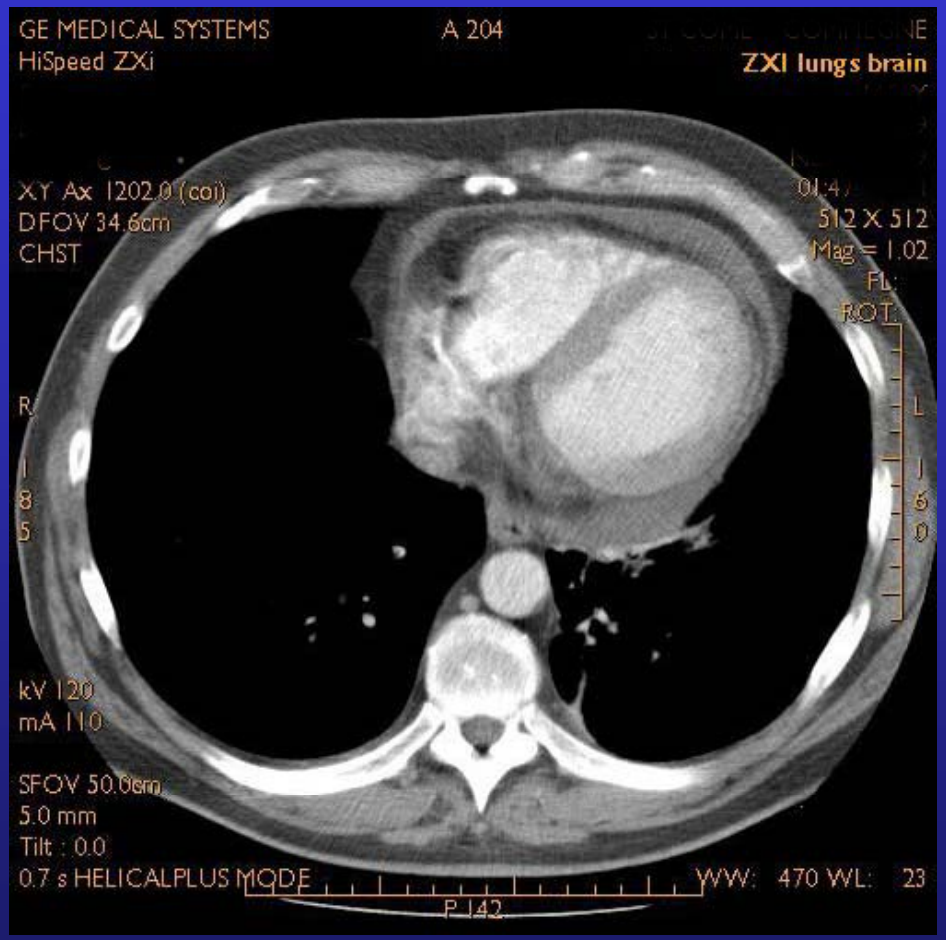
There are many different mechanisms that can be exploited to produce signal contrast

Relaxation processes account for some of them, but there are many others, which we will encounter later in the course

The allow the MRI technique to be adapted to optimize conspicuity of structures of interest

$\Rightarrow$ **This makes MRI extremely versatile**

Comparison with CT



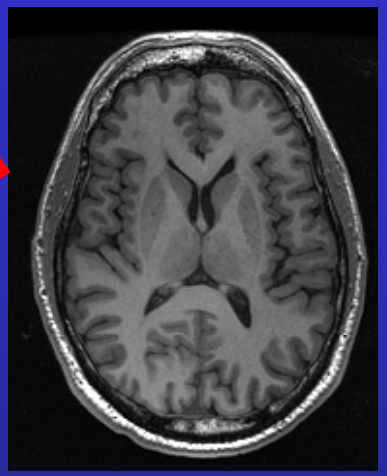
In CT there is only one contrast mechanism

signal intensity $\propto$ tissue density

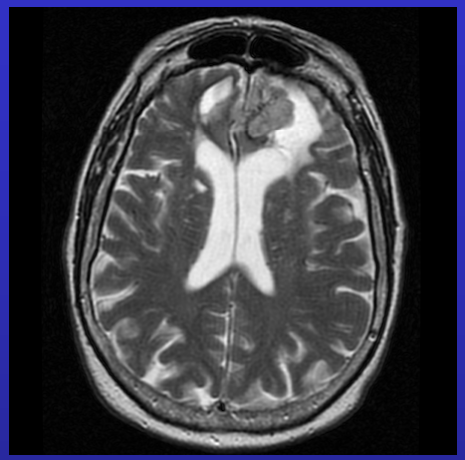
What causes the different signal contrast?

- Relaxation times T_1, T_2 and T_2^*

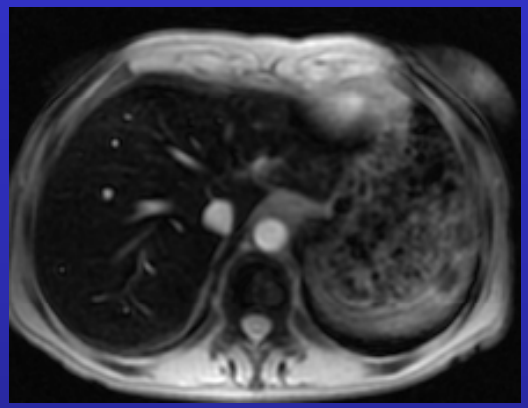
TODAY



T_1



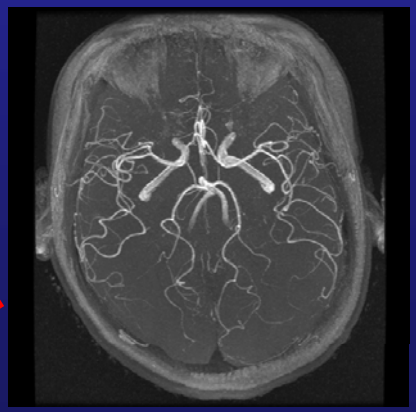
T_2



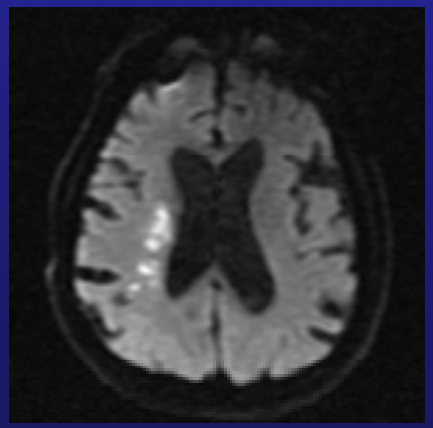
T_2^*

- Flow
- Diffusion

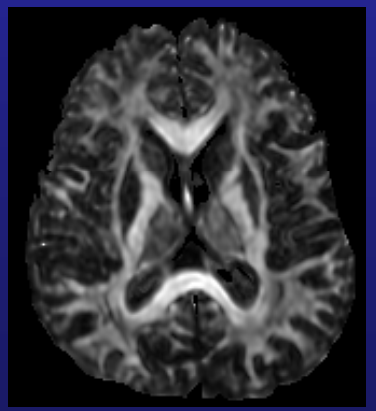
LATER



flow

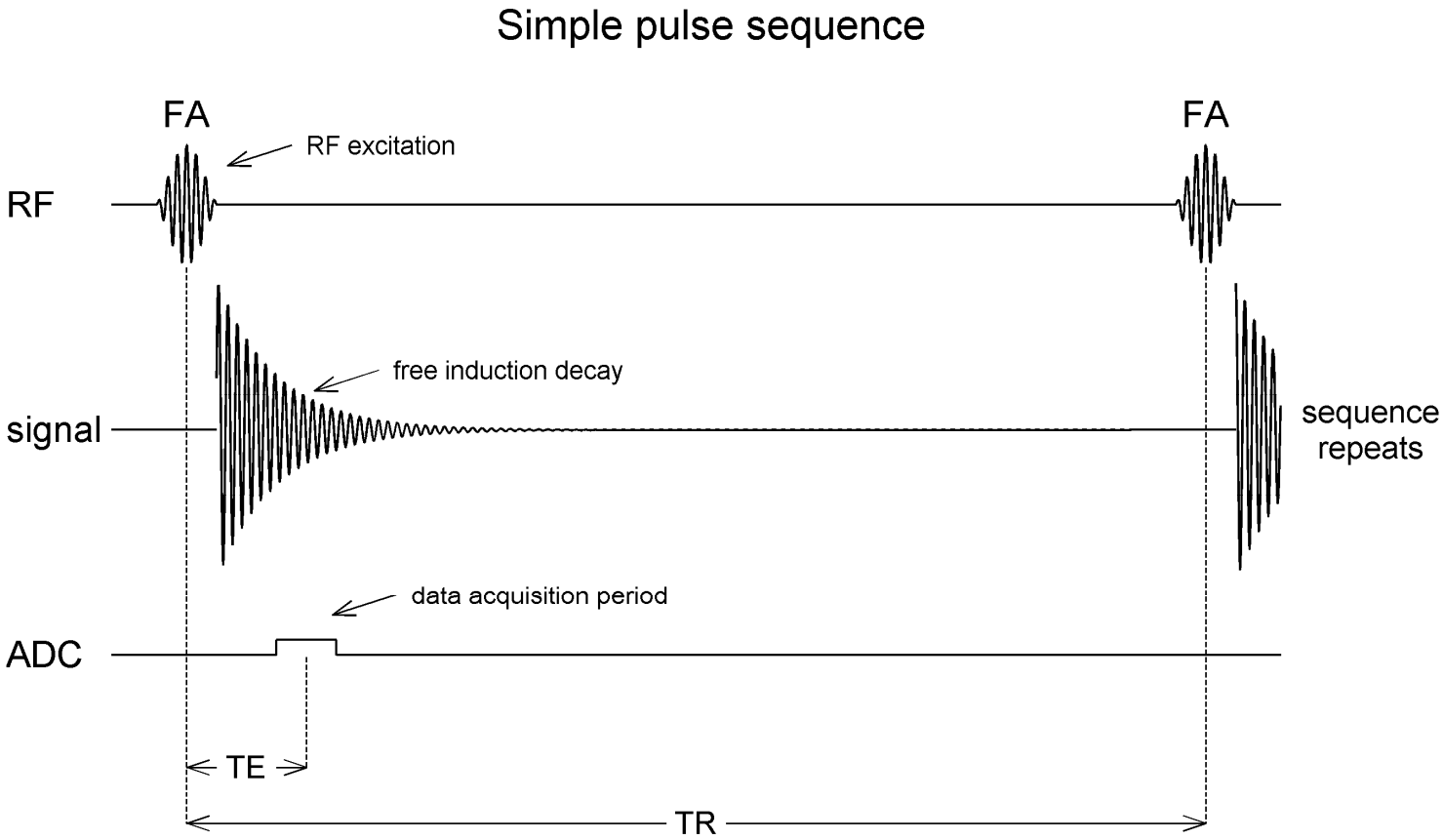


diffusion



diffusion anisotropy

In imaging we must excite the spins and acquire the emitted signal many times in succession



TE is how long we wait before measuring the signal

TR is how long we wait before the next excitation

Choice of TE and TR determines the signal contrast

T_2^* weighting

T_2^* is the transverse relaxation time of the FID

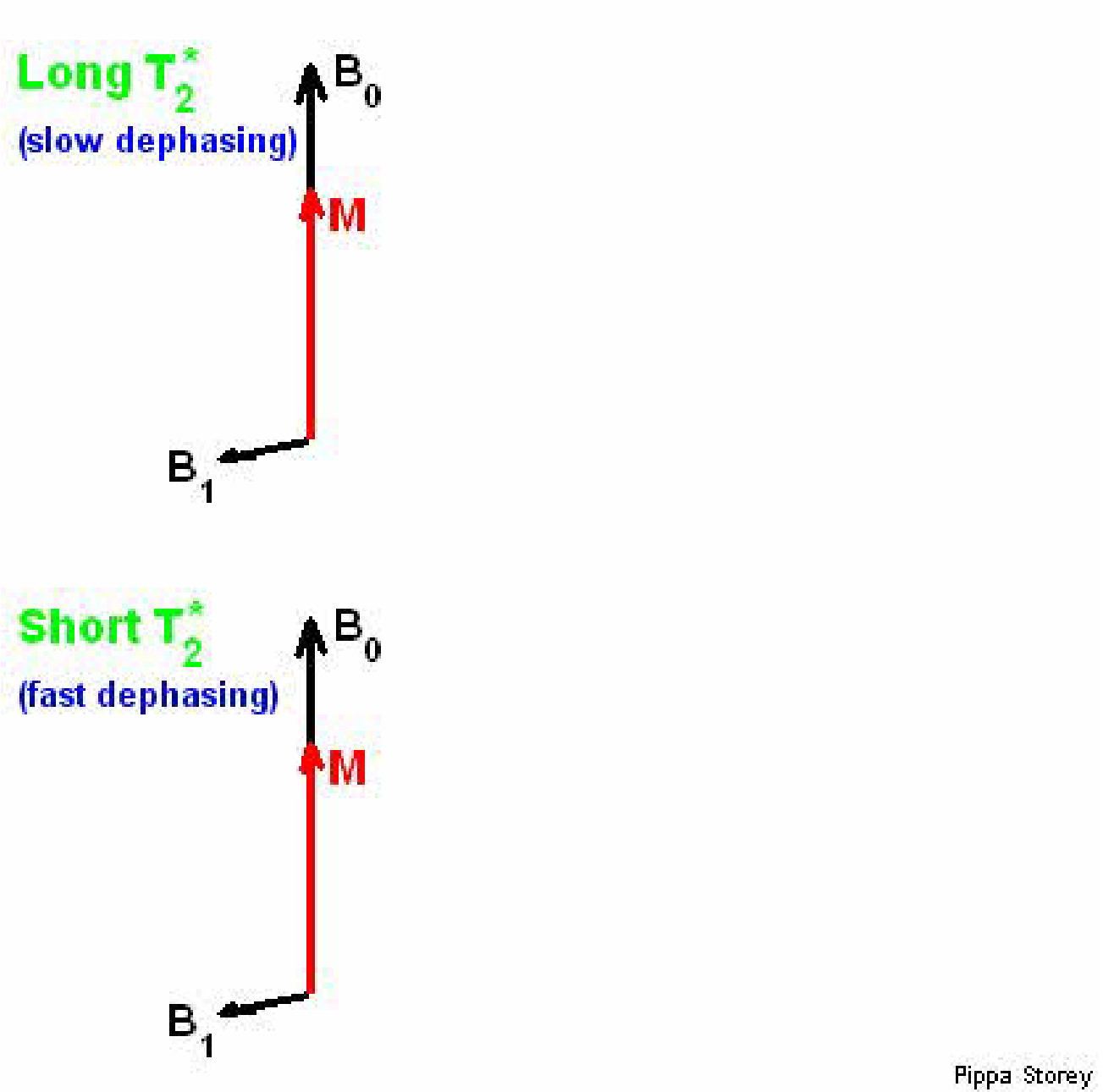
If we choose a long TE then:

Tissues with: long T_2^* have high signal (bright)

short T_2^* have low signal (dark)

The resulting image is then said to be 'T_2^* weighted', since the signal depends heavily on T_2^*.

T_2* weighting



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T_1 weighting

T1 is the longitudinal relaxation time

If we choose short TR:

Tissues with long T_1 do not fully recover to equilibrium between successive excitations

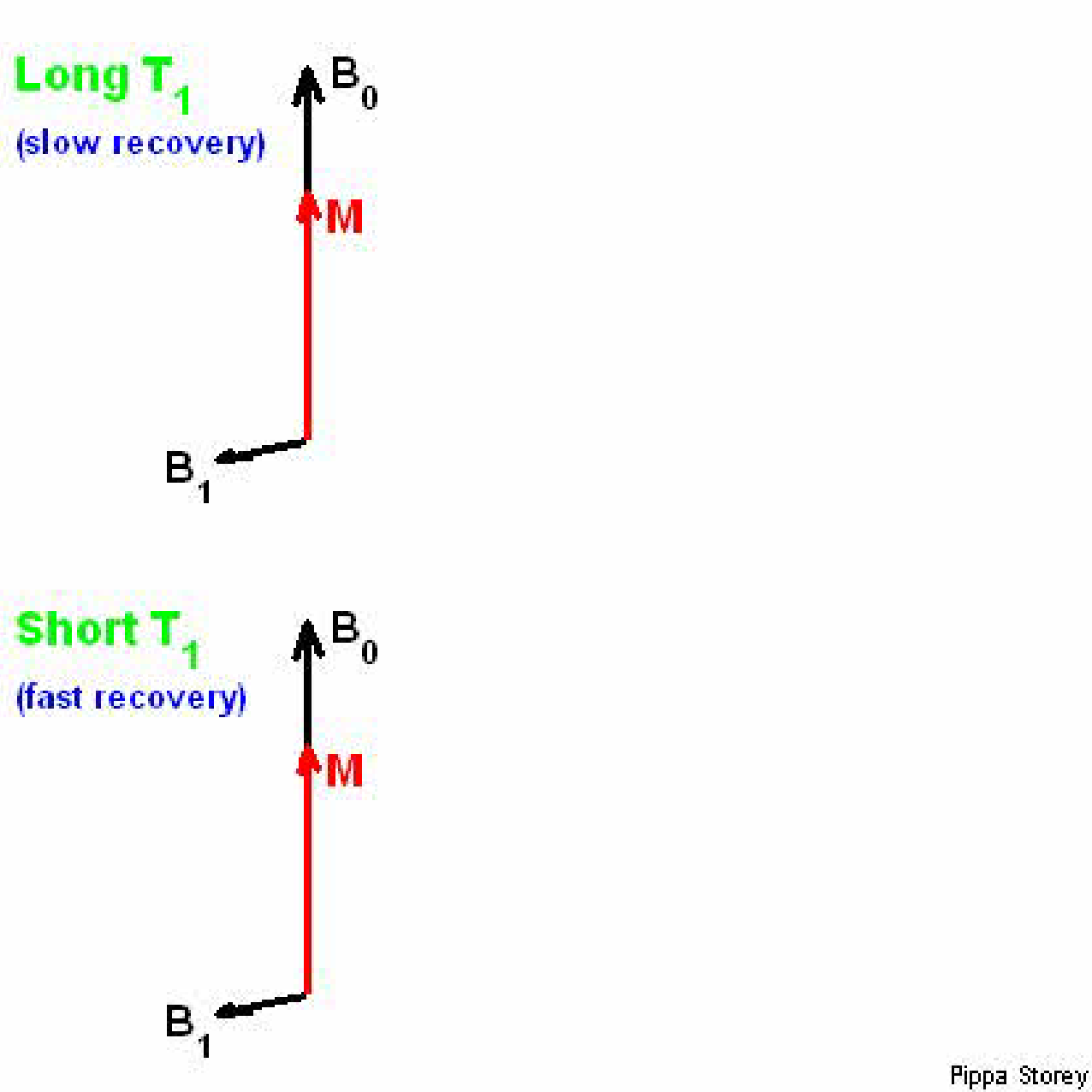
$\Rightarrow$ Low signal

Tissues with short T_1 recover more quickly

$\Rightarrow$ High signal

The resulting images are said to be ‘T1 weighted’ since the signal depends heavily on T1.

T_1 weighting



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Origin of relaxation processes

Relaxation processes result from a range of mechanisms

- Microscopic interactions between nuclei and molecules
- Mesoscopic & macroscopic variations in magnetic field strength

Microscopic mechanisms:

Each nucleus experiences the tiny magnetic fields of other nuclei and molecules in its environment.

As the molecules move and tumble they produce magnetic field fluctuations at the site of the nucleus

Two types of fluctuations are important

Fluctuations in the x-y plane at frequency $\omega \cong \omega_0$

Induce transitions between higher and lower energy states and bring the spins into thermal equilibrium with their environment

Variations in the z-component at low frequency ($\omega \cong 0$)

Cause the spins to precess at slightly different frequencies and produce dephasing

T_1 relaxation

T_1 relaxation is also known as 'longitudinal relaxation'
It describes the loss of energy from the spins to their environment

It causes recovery of the longitudinal component of magnetization back to its equilibrium value M_0

$$\frac{dM_z}{dt} = -\frac{M_z - M_0}{T_1}$$

which has the solution

$$M_z = M_0 + \left[M_z(0) - M_0\right]e^{-t/T_1}$$

where $M_z(0)$ is the initial value of the longitudinal magnetization (e.g. its value immediately after an RF excitation)

T1 relaxation is caused by magnetic field fluctuations in the x-y plane at frequency $\omega \cong \omega_0$

Dependence of T_1 on tissue type and field strength

Longitudinal relaxation occurs most efficiently when the molecular tumbling rate is close to the Larmor frequency

The tumbling rate is closest to the Larmor frequency for small molecules such as lipids

$\Rightarrow$ fat has short T_1 (~250ms at 1.5T)

The tumbling rate of free water in body fluids (e.g. blood and CSF) is much faster than the Larmor frequency

$\Rightarrow$ CSF and blood have long T_1 (>1s at 1.5T)

The tumbling rate of water is reduced in solid tissues where its motion is more restricted

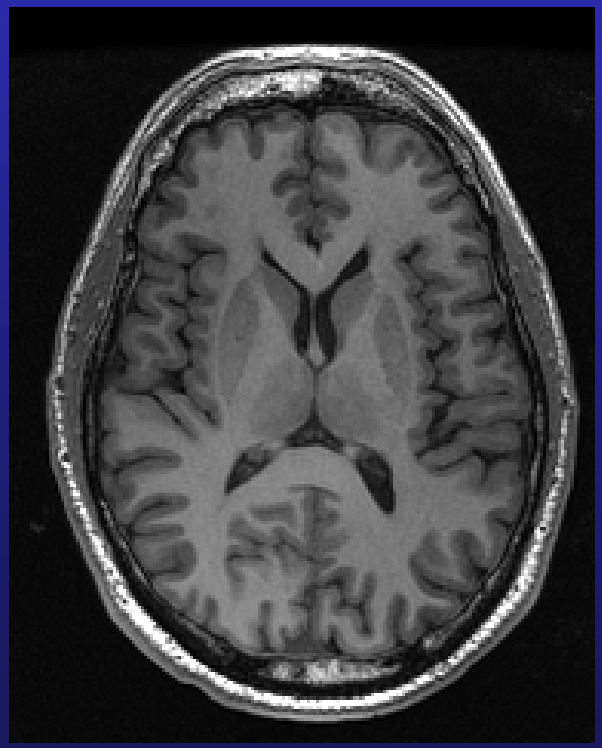
$\Rightarrow$ solid tissues have shorter T_1 than fluids

Since the Larmor frequency depends on field strength, so does the value of T_1

In general, T_1 increases with increasing B_0

T_1 weighted image

Tissues with: long T_1 appear dark
short T_1 appear bright



Fluid tends to have longer relaxation times than tissue

$\Rightarrow$ CSF appears dark

T_2 relaxation

T2 relaxation is the **irreversible** dephasing among spins due to **microscopic** interactions between molecules and nuclei

T2 relaxation causes a decay of the transverse component of magnetization according to the equation

$$\frac{dM_\perp}{dt} = -\frac{M_\perp}{T_2}$$

which has the solution

$$M_\perp = M_\perp(0)e^{-t/T_2}$$

where $M_\perp(0)$ is the initial value of the transverse magnetization (e.g. the value immediately after an RF excitation)

The mechanisms that cause T_1 relaxation also contribute to T_2 relaxation.

However, T2 relaxation also results from low frequency magnetic field variations in the z-direction which cause dephasing

Hence $T_2 < T_1$

Dependence of T_2 on tissue type and field strength

T2 relaxation occurs most efficiently when the molecular tumbling rate is low

Rapid motion tends to inhibit T2 relaxation by averaging out the effects of microscopic interactions over time (a phenomenon known as 'motional narrowing')

The tumbling rate of free water in body fluids is very fast, so T2 relaxation is inefficient

$\Rightarrow$ CSF and blood have long T_2 (>100ms at 1.5T)

The tumbling rate of water is reduced in solid tissues where its motion is more restricted

$\Rightarrow$ solid tissues have shorter T_2 than fluids

Since T_2 relaxation is determined largely by low frequency fluctuations, T_2 is largely independent of field strength

Approximate T_1 and T_2 values at 1.5T

tissue	T1 (ms)	T2 (ms)
gray matter	950	100
white matter	600	80
muscle	900	50
CSF	4500	2200
fat	250	60
blood	1200	100 – 200*

*** Lower value is for venous blood, higher value for arterial blood**

Macroscopic and mesoscopic effects on transverse relaxation

Transverse relaxation is due to dephasing

Different nuclei experience different magnetic fields, which means their Larmor frequencies are slightly different

Microscopic component:

Fields due to neighboring nuclei and molecules

Time scale T_2

Macroscopic/mesoscopic component:

Fields due to: - B_0 inhomogeneity

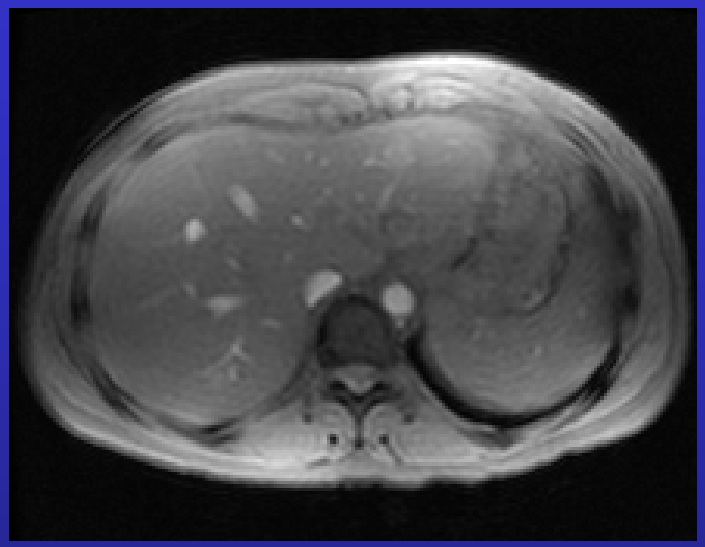
- susceptibility differences in tissue

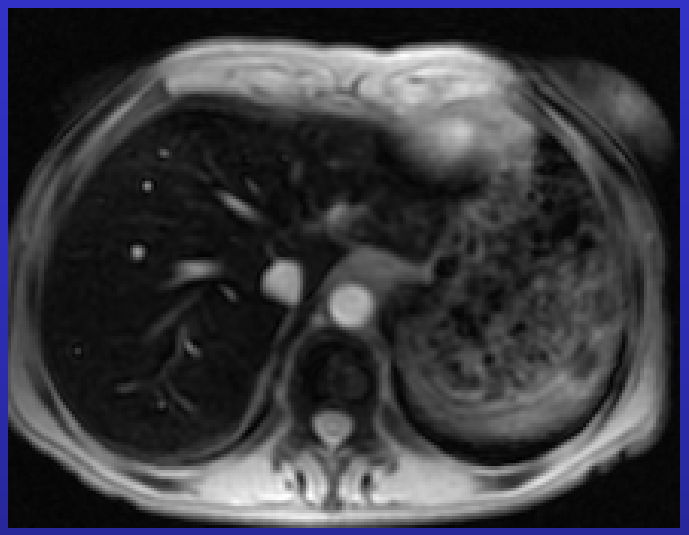
T_2^*

Relaxation time of FID

T_2^* involves both microscopic and macroscopic components $\Rightarrow$ $T_2^* < T_2$

T_2* weighted images





TE = 9ms

Control subject Iron overload

Iron is stored in liver as ferritin and hemosiderin (paramagnetic)
They produce local variations in the magnetic field
Causes dephasing i.e. short T_2*
$\Rightarrow$ Liver with high iron load looks dark

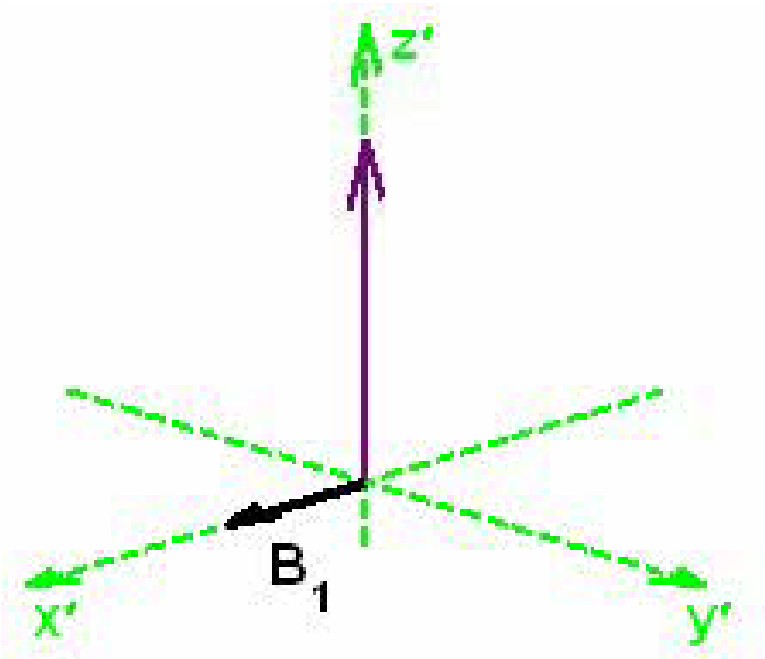
T_2-weighting

- Magnetic field differences due to macroscopic factors vary slowly with time
- It is possible to 'undo' the dephasing from such factors with the aid of a 180º RF pulse
- The 180º pulse reverses the phases of the spins
  - Slow spins are given a 'head start'
  - Fast spins are given a handicap
- The spins come back into phase, forming a 'spin echo'
- Only the dephasing due to microscopic factors remains

$\Rightarrow$ T_2-weighting

Formation of a spin echo

(nb. spin diagram is shown in reference frame rotating at Larmor frequency)



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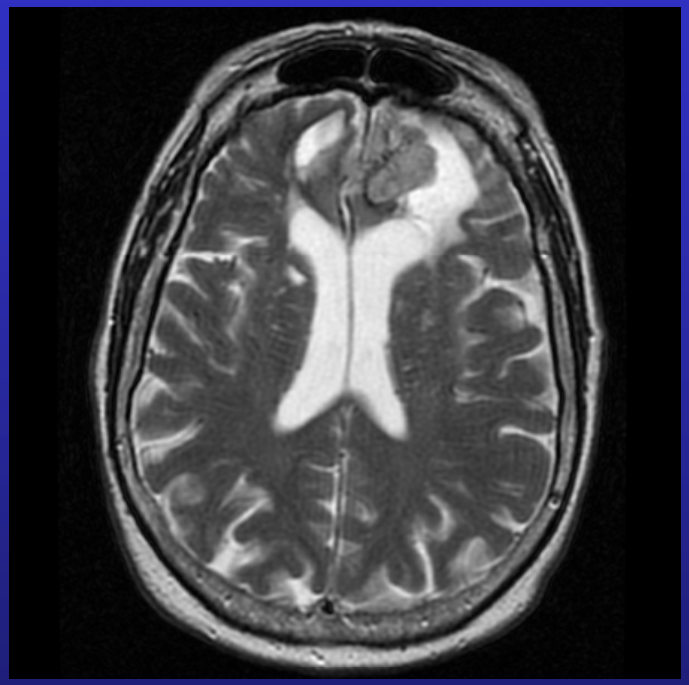
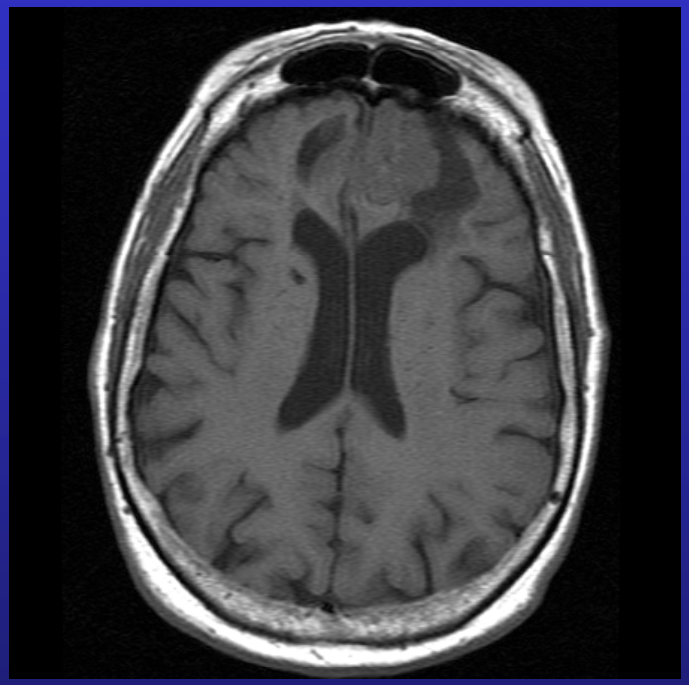
Spin-echo imaging

- Long TE $\Rightarrow$ T_2-weighting
- Short TR $\Rightarrow$ T_1-weighting



T_2-weighted



T_1-weighted

Fluids tend to have longer relaxation times than tissue

CSF and edema are bright on T_2-weighted images, and dark on T_1-weighted images

Summary chart

	Long TR	Short TR
Short TE	Proton density-weighted	T1-weighted
Long TE	T2- or T2*-weighted (for spin echo or gradient echo respectively)	Not used

Exploiting T_1, T_2 and T_2^* differences

Morphology: Fluids have longer relaxation times than tissues

Lesions: Many lesions (e.g. tumors) are edematous
$\Rightarrow$ Relaxation times similar to fluids

Iron overload: Ferritin is paramagnetic
$\Rightarrow$ Shortens relaxation times

Hemorrhage: Many blood derivatives containing iron are paramagnetic

Blood oxygenation: Deoxyhemoglobin is paramagnetic $\Rightarrow$ fMRI

Macroscopic effects on dephasing

Large-scale field inhomogeneities can arise from

Poor shimming
The magnet does not produce a uniform field

Susceptibility differences within the tissue
The magnetic field is different in diamagnetic substances (e.g. water) from paramagnetic substances (e.g. blood derivatives) and ferromagnetic substances (e.g. many metals)

Any questions?

Acknowledgments

- I am extremely grateful to Prof. Pippa Storey (NYU, Radiology) who provided most of the slides used in this presentation

Homework

- Reading:
  - Prince and Links, Medical Imaging Signals and Systems, Chap. 12
  - Note down all the corrections for Ch. 12 on your copy of the textbook based on the provided errata (see Course website or book website for update)
- Problems
  - P12.1
  - P12.2
  - P12.4
  - P12.5
  - P12.7
  - P12.10
  - P12.11
  - P12.12

See you next week!