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MRI Physics Review

Theoretical and practical concepts

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Acronyms

Generic	Siemens	GE	Philips	Hitachi	Toshiba
Pulse sequences					
Conventional spin-echo (SE)	SE	SE	SE	SE	SE
Turbo spin-echo (TSE)	TSE	FSE	TSE	FSE	FSE
Single-shot TSE (SS-TSE)	HASTE	SS-FSE	SS-TSE	SS-FSE	FASE
TSE (with restoration pulse)	RESTORE	FRFSE	DRIVE	driven equilibrium FSE	T2 Puls FSE
Inversion recovery (IR)	IR	IR/MPIR	IR	IR	IR
Fast inversion recovery	TIR	Fast IR	IR-TSE	IR	IR
Short tau IR (STIR)	STIR	STIR	STIR	STIR	fast STIR
Fluid-attenuated IR (FLAIR)	turbo dark fluid	FLAIR	FLAIR	FLAIR	fast FLAIR
Gradient-echo (GRE)	GRE	GRE	FFE	GE	field echo
Coherent gradient-echo	FISP	GRASS	FFE	rephased SARGE	SSFP
Incoherent gradient-echo	FLASH	SPGR	T1 FFE	spoiled SARGE	fast FE
Reverse-echo gradient-echo	PSIF	SSFP	T2 FFE	time-reversed SARGE	—
Balanced gradient-echo	true FISP	FIESTA	BFFE	balanced SARGE	true SSFP
Echo-planar imaging (EPI)	EPI	EPI	EPI	EPI	EPI
Double-echo steady state	DESS	—	—	—	—
Balanced dual excitation	CISS	FIESTA-C	—	phase balanced SARGE	—
Multi-echo-data-image-combination	MEDIC	MERGE	MFFE	—	—
Fast gradient-echo	turbo FLASH	fast GRE, fast SPGR	TFE	RGE	Fast FE
Hybrid sequence	TGSE	—	GRASE	—	Hybrid EPI

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Acronyms

Generic	Siemens	GE	Philips	Hitachi	Toshiba
Contrast parameters					
Repetition time (TR)	TR	TR	TR	TR	TR
Time to echo (TE)	TE	TE	TE	TE	TE
Time from inversion (TI)	TI	TI	TI	TI	TI
Flip angle	flip angle	flip angle	flip angle	flip angle	Flip angle
Number of echoes (in TSE)	turbo factor	ETL	turbo factor	shot factor	ETL
b factor/value	*b* factor	*b* factor	*b* factor	*b* factor	*b* factor
Geometry parameters					
Field of view (FOV)	FOV (mm)	FOV (cm)	FOV (mm)	FOV (mm)	FOV (mm)
Rectangular FOV	FOV phase	PFOV	rectangular FOV	rectangular FOV	rectangular FOV
Slice gap	distance factor	gap	gap	slice interval	gap

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Acronyms

Data acquisition parameters					
Averages	average	NEX	NSA	NSA	NSA
Bandwidth	bandwidth (Hz/pixel)	receive bandwidth (KHz)	fat water shift (pixel)	bandwidth (KHz)	bandwidth (KHz)
Variable bandwidth	optimized bandwidth	variable bandwidth	optimized bandwidth	variable bandwidth	matched bandwidth
Partial averaging	half Fourier	fractional NEX	half scan	half scan	AFI
Partial echo	asymmetric echo	partial echo	partial echo	half echo	matched bandwidth
Parallel imaging (image based)	mSENSE	ASSET	SENSE	RAPID	SPEEDER
Parallel imaging (*k*-space based)	GRAPPA	ARC	—	—	—

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Acronyms

Artifact reduction techniques					
Radial *k*-space filling	BLADE	PROPELLOR	multiVane	RADAR	JET
Gradient moment rephasing	GMR/flow comp	flow comp	flow comp/ FLAG	GR	FC
Presaturation	pre SAT	Sat	REST	Pre SAT	Pre SAT
Moving sat pulse	travel SAT	walking SAT	travel REST	Sequential pre SAT	BFAST
Fat saturation	fat SAT	chem SAT	SPIR	Fat Sat	MSOFT
Out-of-phase imaging	DIXON	IDEAL	ProSET	Water excitation	PASTA
Respiratory compensation	respiratory gated	respiratory compensation	PEAR	MAR	respiratory gated
Antialiasing (frequency)	oversampling	antialiasing	frequency oversampling	frequency oversampling	frequency wrap suppression
Antialiasing (phase)	phase oversampling	no phase wrap	fold-over suppression	antiwrap	phase wrap suppression

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Acronyms

Generic	Siemens	GE	Philips	Hitachi	Toshiba
Special techniques					
Volume TSE variable flip angle	SPACE	CUBE	VISTA	—	—
Volume gradient-echo	VIBE	LAVA-XV	THRIVE	TIGRE	—
Dynamic MRA	TWIST	TRICKS-SV	keyhole (4d Trak)	—	—
Noncontrast MRA gradient-echo	NATIVE – true FISP	inhance inflow IR	B-TRANCE	VASC ASL	TIME-SLIP
Noncontrast MRA spin- echo	NATIVE-SPACE	—	TRANCE	VASC FSE	FBI
Susceptibility weighting	SWI	SWAN	Venous BOLD	—	—
High-resolution breast imaging	VIEWS	VIBRANT-XV	BLISS	—	RADIANCE
Diffusion-weighted imaging	DWI	DWI	DWI	DWI	DWI
Diffusion tensor imaging	DTI	DTI	diffusion tensor imaging	—	DTI
Body diffusion imaging	REVEAL	—	DWIBS	—	body vision

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Nomenclature

S	spin quantum number	
N^+	number of spins in the high-energy population (Boltzmann)	
N^-	number of spins in the low-energy population (Boltzmann)	
ΔE	energy difference between high- and low-energy populations (Boltzmann)	J
k	Boltzmann's constant	J/K
T	temperature of the tissue	K
ω_0	precessional or Larmor frequency	MHz
γ	gyromagnetic ratio	MHz/T
B_0	external magnetic field strength	T
E	energy of a photon	J
h	Planck's constant	J/s
θ	flip angle	°
ω_1	precessional frequency of B_1	μT

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Nomenclature

B_1	magnetic field associated with the RF excitation pulse	mT
τ	duration of the RF excitation pulse	ms
ϵ	emf	V
N	number of turns in a coil	
$d\Phi$	changing magnetic flux in a single loop	V/s
dt	changing time	s
Mz_t	amount of longitudinal magnetization at time t	
Mz	full longitudinal magnetization	
Mxy_t	amount of transverse magnetization at time t	
Mxy	full transverse magnetization	
SI	signal intensity in a tissue	
ΔB_0	variation in magnetic field	ppm
G	gradient amplitude	mT/m
δ	gradient duration	ms
Δ	time between two gradient pulses	ms

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Nomenclature

b	b value or b factor	s/mm^2
ST	scan time	s
E_S	echo spacing in turbo spin-echo (TSE)	ms
t	time from inversion (TI)	ms
Ernst	Ernst angle	°
TE_{eff}	effective TE	ms
TE_{act}	TE set at the console	ms
B_p	magnetic field strength at a point along the gradient	T
Sl_t	slice thickness	mm
TBW	transmit bandwidth	KHz
$\omega_{sampling}$	digital sampling frequency	KHz
ΔT_s	sampling interval	ms
$\omega_{Nyquist}$	Nyquist frequency	KHz
RBW	receive bandwidth	KHz
W_s	sampling window	ms

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Nomenclature

M(f)	frequency matrix	
M(p)	phase matrix	
N_s	number of slice locations	
G(p)	max amplitude of the phase encoding gradient	mT/m
ϕ	incremental step between each *k*-space line	
G(*f*)	amplitude of the frequency encoding gradient	mT/m
FOV(f)	frequency FOV	cm
σ	standard deviation of background signal or noise	
S_p	separation between ghosts due to motion p	pixels
T_m	period of motion of something moving in the patient	ms
Re	Reynolds number	
d	density of blood	g/cm^3
v	velocity of flow	cm/s
m	diameter of a vessel	cm
vis	viscosity of blood	g/cm s
f_p	perceived frequency	KHz
f_t	actual frequency	KHz
ω_{csf}	chemical shift frequency difference between fat and water	Hz

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Nomenclature

C_s	chemical shift (3.5 ppm or 3.5×10^{-6})	ppm
CS_p	pixel shift	mm
H_0	magnetic intensity	A/m
q	charge of a particle	C
F	Lorentz force (total emf on a charged particle)	V
E	electric field vector	
B	magnetic field vector	

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Abbreviations

A	Anterior	DE prep	Driven equilibrium magnetization preparation
AC	Number of acquisitions	DTI	Diffusion tensor imaging
ADC	Apparent diffusion coefficient	DWI	Diffusion weighted imaging
ADEM	Acute disseminating encephalomyelitis	ECG	Echocardiogram
ASIS	Anterior superior iliac spine	EPI	Echo planar imaging
AVM	Arteriovenous malformation	ETL	Echo train length
AVN	Avascular necrosis	FA	Fractional anisotropy
BFFE	Balanced fast field echo	FAT SAT	Fat saturation
BGRE	Balanced gradient echo	FC	Flow compensation
BOLD	Blood oxygenation level dependent	FDA	Food and Drugs Administration
CE-MRA	Contrast-enhanced MRA	FFE	Fast field echo
CNR	Contrast to noise ratio	FID	Free induction decay signal
CNS	Central nervous system	FIESTA	Free induction echo stimulated acquisition
CSE	Conventional spin echo	FISP	Fast imaging with steady precession
CSF	Cerebrospinal fluid	FLAIR	Fluid-attenuated inversion recovery
CT	Computer tomography	FLASH	Fast low angled shot
CVA	Cerebral vascular accident	fMRI	Functional MRI

Abbreviations

FOV	Field of view	L	Left
FSE	Fast spin echo	MP RAGE	Magnetization prepared rapid gradient echo
GFE	Gradient field echo	MR	Magnetic resonance
GMN	Gradient moment nulling	MRA	Magnetic resonance angiography
GMR	Gradient moment rephasing	MRCP	Magnetic resonance cholangiopancreatography
GRASS	Gradient recalled acquisition in the steady state	MRI	Magnetic resonance imaging
GRE	Gradient echo	MS	Multiple sclerosis
GRE-EPI	Gradient echo EPI	MT	Magnetization transfer
HASTE	Half acquisition single-shot turbo spin echo	NEX	Number of excitations
I	Inferior	NSA	Number of signal averages
IAM	Internal auditory meatus	P	Posterior
IM	Intramuscular	PC	Phase contrast
IR	Inversion recovery	PC-MRA	Phase contrast MRA
IR-FSE	Inversion recovery FSE	PD	Proton density
IR prep	Inversion recovery magnetization preparation	Pe	Peripheral
IV	Intravenous	PEAR	Phase encoding artefact reduction
IVC	Inferior vena cava	PSIF	Reverse FISP

Abbreviations			
R	Right	SSFP	Steady-state free precession
RC	Respiratory compensation	SS-FSE	Single-shot FSE
REST	Regional saturation technique	STIR	Short TAU inversion recovery
RF	Radio frequency	SW	Susceptibility weighted
ROI	Region of interest	TE	Echo time
RR	R to R interval	TFE	Turbo field echo
S	Superior	TI	Inversion time
SAR	Specific absorption rate	TIA	Transient ischaemic attack
SAT	Saturation	TLE	Temporal lobe epilepsy
SE	Spin echo	TMJ	Temporomandibular joint
SE-EPI	Spin echo EPI	TOF	Time of flight
SNR	Signal to noise ratio	TOF-MRA	Time of flight MRA
SPAMM	Spatial modulation of magnetization	TR	Repetition time
SPGR	Spoiled GRASS	True FISP	Siemens version of BGE
SPIR	Spectrally selective inversion recovery	TSE	Turbo spin echo
SS	Single shot	VENC	Velocity encoding
SS-EPI	Single-shot EPI		

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- Introduction
- Atomic structure
- Motion in the atom
- MR-active nuclei
- The hydrogen nucleus
- Alignment
- Net magnetic vector (NMV)

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- Precession and precessional (Larmor) frequency
- Precessional phase
- Resonance
- MR signal
- The free induction decay (FID) signal
- Pulse timing parameters

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Introduction

There are essentially two ways of explaining the fundamentals of MRI: classically and via quantum mechanics.

Classical theory (accredited to Sir Isaac Newton and often called Newtonian theory) provides a mechanical view of how the universe (and therefore how MRI) works. Using classical theory, MRI is explained using the concepts of mass, spin, and angular momentum on a large or bulk scale.

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Introduction

Quantum theory (accredited to several individuals including Max Planck, Albert Einstein, and Paul Dirac) operates at a much smaller, subatomic scale and refers to the energy levels of protons, neutrons, and electrons. Although classical theory is often used to describe physical principles on a large scale and quantum theory on a subatomic level, there is evidence that all physical principles are explained using quantum concepts.

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Atomic structure

All things are made of atoms. Atoms are organized into molecules, which are two or more atoms arranged together. The most abundant atom in the human body is hydrogen, but there are other elements such as oxygen, carbon, and nitrogen. Hydrogen is most commonly found in molecules of water and fat. The atom consists of a central nucleus and orbiting electrons.

This mass comes mainly from particles called nucleons, which are subdivided into protons and neutrons.

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Atomic structure

Atoms are characterized in two ways:

- ✓ **The atomic number is the sum of the protons in the nucleus. This number gives an atom its chemical identity.**
- ✓ **The mass number or atomic weight is the sum of the protons and neutrons in the nucleus**

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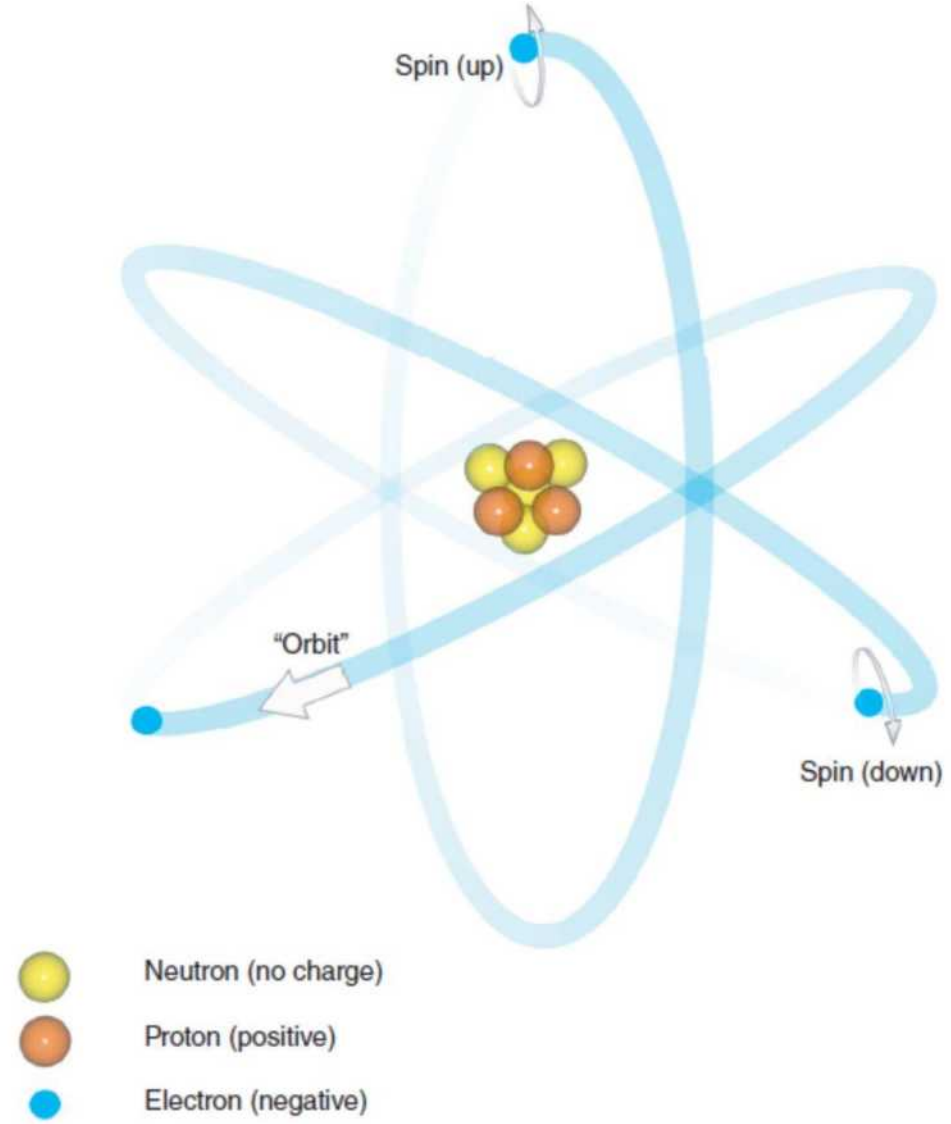
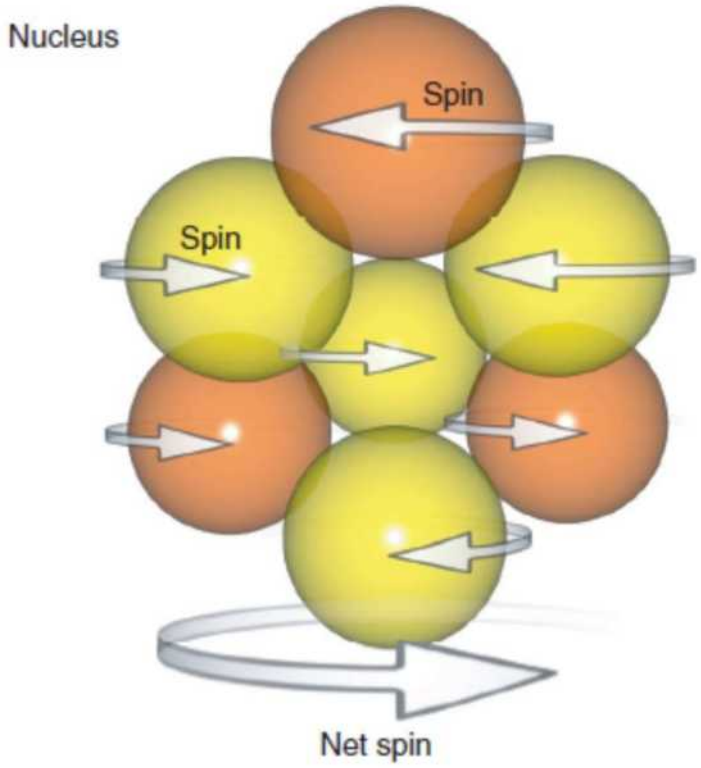
Motion in the atom

Three types of motion are present within the atom:

- **Electrons spinning on their own axis**
- **Electrons orbiting the nucleus**
- **The nucleus itself spinning about its own axis.**

NOTE: The principles of MRI rely on the spinning motion of specific nuclei present in biological tissues.


Spin (up)
"Orbit"
Spin (down)
Neutron (no charge)
Proton (positive)
Electron (negative)


Nucleus
Spin
Spin
Net spin

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However,

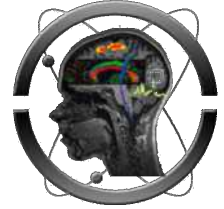
in nuclei with an odd number of protons,

an odd number of neutrons,

or an odd number of both protons and neutrons,

the spin directions are not equal and opposite,

so the nucleus itself has a net spin or angular momentum.



This means that their spin has a half-integral value, e.g. ½, 5/2.

However, this phenomenon also occurs in nuclei with an odd number of both protons and neutrons resulting in an even mass number.

This means that it has a whole integral spin value, e.g. 1, 2, 3.

Examples are lithium (which is made up of three protons and three neutrons) and nitrogen (seven protons and seven neutrons).

However, these elements are largely unobservable in MRI so, in general, only nuclei with an odd mass number or atomic weight are used. These are known as MR-active nuclei.

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What makes a proton spin and why is it charged?

On a subnuclear level,

individual protons are made up of quarks,

each of which possesses the characteristics of alignment and spin.

NOTE: The net charge and spin of a proton are a consequence of its quark composition.

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What makes a proton spin and why is it charged?

The proton consists of three spinning quarks.

Two quarks spin up and the other spins down.

The net spin of the proton (1/2) is caused by the different alignment of the quarks.

The net charge of the proton is caused by each spin-up quark having a charge of + 2/3, while the spin-down quark has a charge of – 1/3 (total charge + 1).

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MR-active nuclei

MR-active nuclei are characterized by their tendency to align their axis of rotation to an applied magnetic field.

This occurs because they have angular momentum or spin and, as they contain positively charged protons, they possess an electrical charge.

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MR-active nuclei Faraday's law

The law of electromagnetic induction: (Michael Faraday in 1833)

the connection between electric and magnetic fields and motion

Faraday's law determines that a moving electric field produces a magnetic field and vice versa.

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MR-active nuclei

MR-active nuclei have a net electrical charge (electric field) and are spinning (motion)

therefore, automatically acquire a magnetic field.

In classical theory, this magnetic field is denoted by a magnetic moment.

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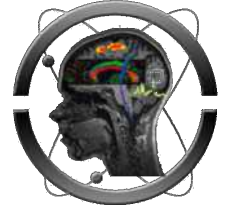
MR-active nuclei

The magnetic moment of each nucleus has vector properties, i.e. it has size (or magnitude) and direction.

The total magnetic moment of the nucleus is the vector sum of all the magnetic moments of protons in the nucleus.



Important examples of MR-active nuclei:

- 1H (hydrogen)
- 13C (carbon)
- 15N (nitrogen)
- 17O (oxygen)
- 19F (fluorine)
- 23Na (sodium)

Table 1.1 Characteristics of common elements in the human body.

Element	Protons	Neutrons	Nuclear spin	% Natural abundance
^{1}H (protium)	1	0	1/2	99.985
^{13}C (carbon)	6	7	1/2	1.10
^{15}N (nitrogen)	7	8	1/2	0.366
^{17}O (oxygen)	8	9	5/2	0.038

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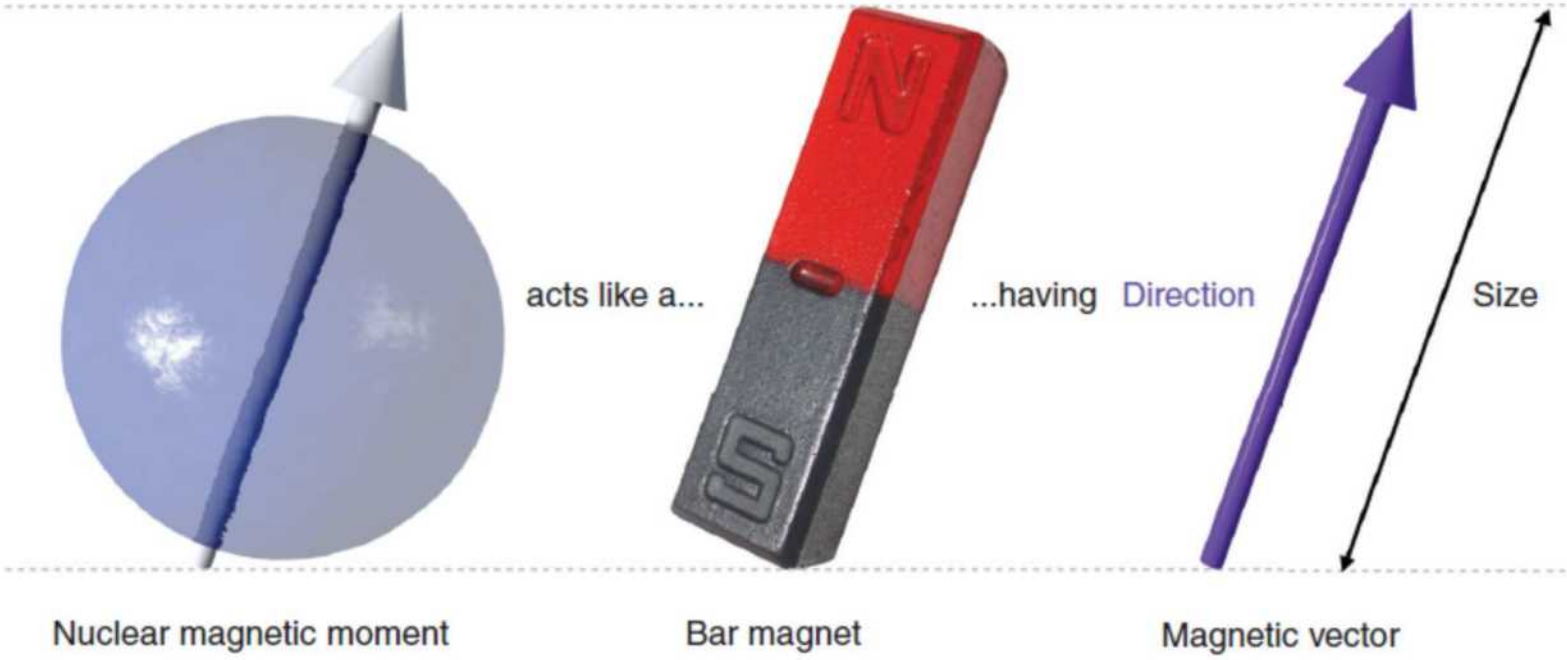
The hydrogen nucleus



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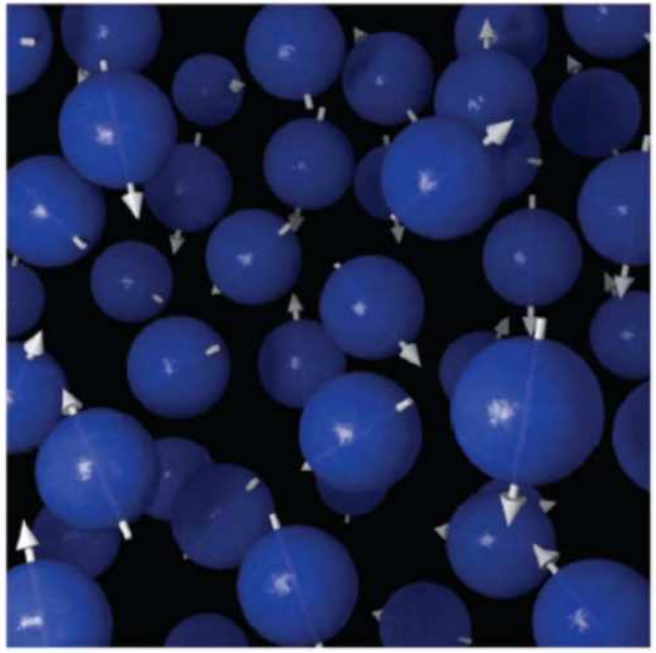
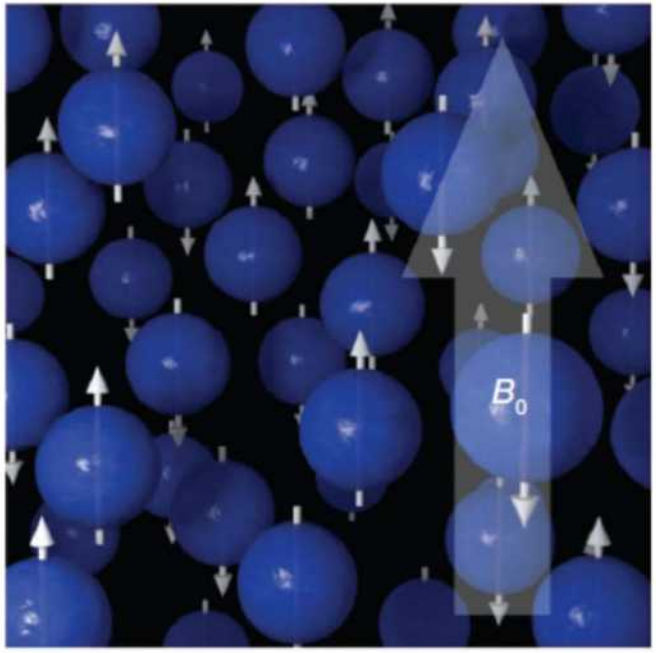
Alignment

Classical theory uses the direction of the magnetic moments of spins (hydrogen nuclei) to illustrate alignment



Random alignment
No external field



Alignment
External magnetic field

Parallel alignment:
Alignment of magnetic moments in the same direction as the main B0 field
(also referred to as spin-up).

Antiparallel alignment:
Alignment of magnetic moments in the opposite direction to the main B0 field (also referred to as spin-down)

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Alignment

After alignment,

There are always more spins with their magnetic moments aligned parallel than antiparallel.

The net magnetism of the patient (termed the net magnetic vector, NMV) is therefore aligned parallel to the main B0 field in the longitudinal plane or z-axis.

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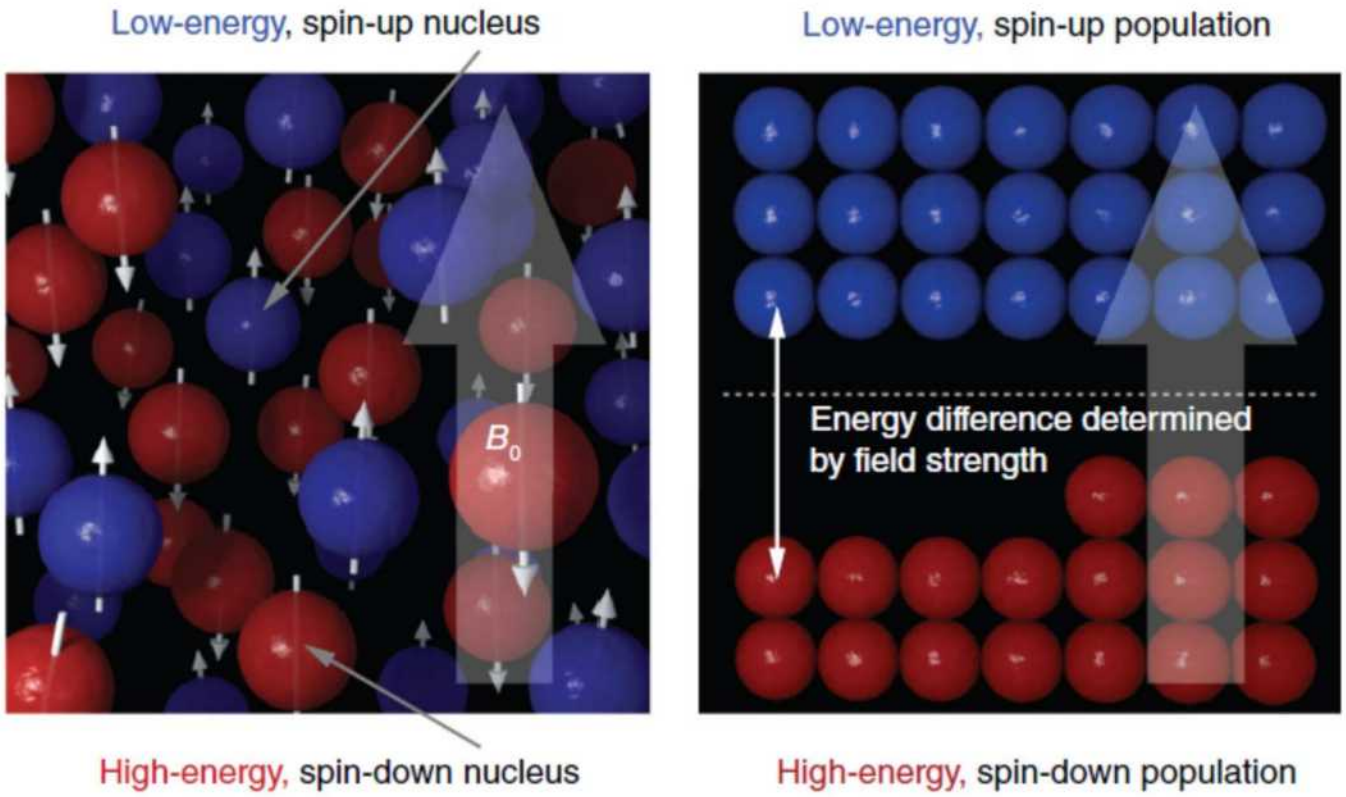
Zeeman interaction



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Boltzmann equation

Equation 1.1		
Number of energy states = 2S + 1	S is the spin quantum number. The value of S for hydrogen is ½	This equation explains why hydrogen can only possess two energy states. If S = ½, then the number of energy states is 2 × ½ + 1 = 2

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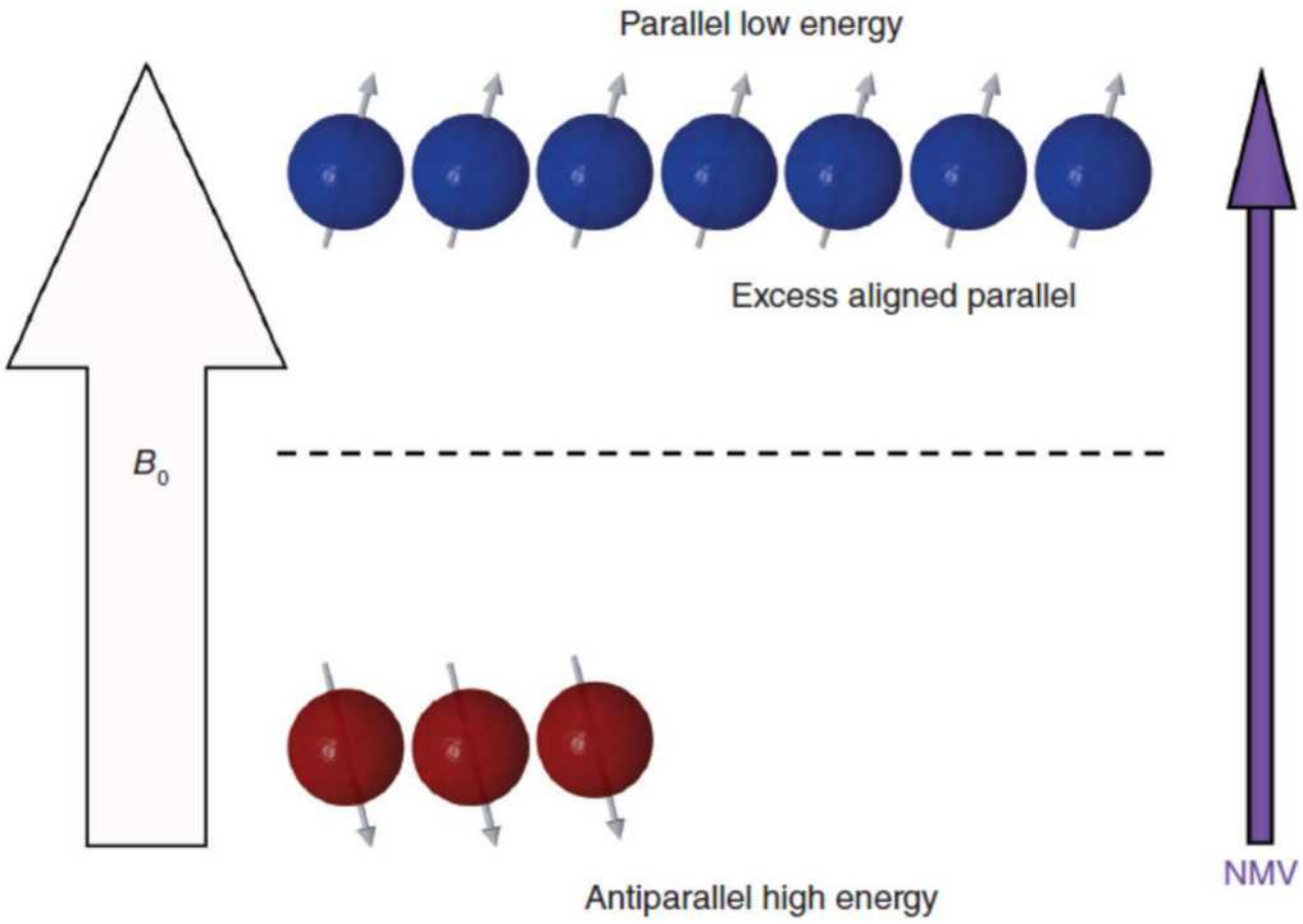
Net magnetic vector (NMV)



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Net magnetic vector (NMV)

NOTE: the NMV also increases in size and is one of the reasons why thesignal-to-noise ratio (SNR) increases at higher field strengths.

In thermal equilibrium, the strength of the external field also determines the relative quantities of spin-up to spin-down nuclei because this also affects the difference in energy levels between the two energy states.

Equation 1.2		
$N^+ / N^- = e^{-\Delta E / kT}$	N^+ and N^- are the number of spins in the high- and low-energy populations, respectively. ΔE is the energy difference between the high- and low-energy populations in Joules (J) k is Boltzmann's constant (1.381×10^{-23} J/K) T is the temperature of the tissue in Kelvin (K)	This equation enables prediction of the number of spins in the high- and low-energy populations and how this is dependent on temperature. In MRI, thermal equilibrium is presumed in that there are no significant changes in body temperature in the scan room

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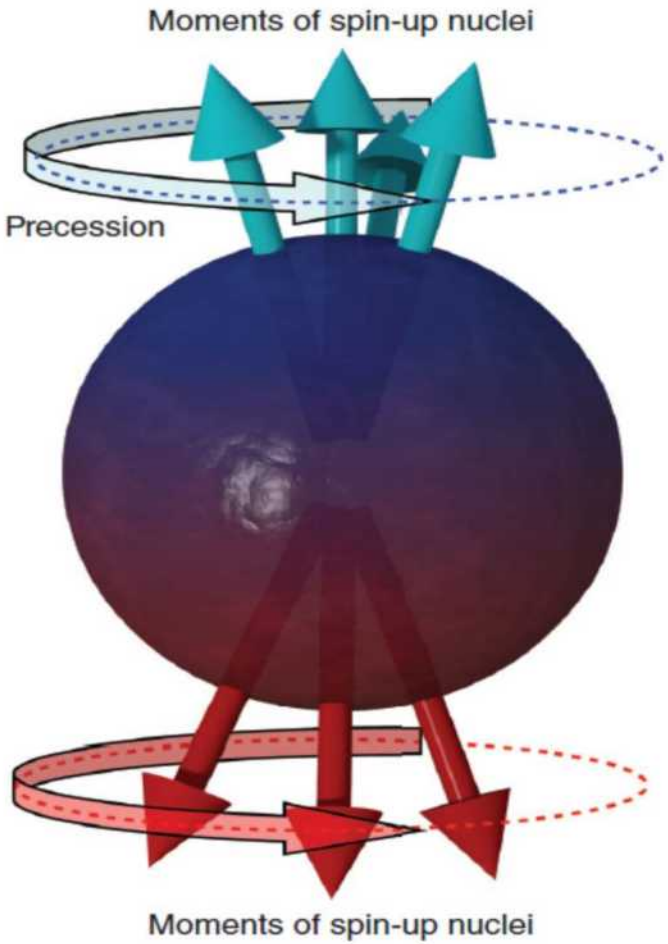
Precession and precessional (Larmor) frequency



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Precession and precessional (Larmor) frequency

Equation 1.3		
$\omega_0 = \gamma B_0/2\pi$ simplified to $\omega_0 = \gamma B_0$	ω_0 is the precessional or Larmor frequency (MHz) γ is the gyromagnetic ratio (MHz/T) B_0 is the strength of the external magnetic field (T)	This is the Larmor equation. The 2π function enables the conversion of ω_0 from angular to cyclical frequency. As γ is a constant, for a given MR-active nucleus ω_0 is proportional to B_0

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Gyromagnetic Ratio

The gyromagnetic ratio expresses

the relationship between angular momentum and the magnetic moment of each MR-active nucleus.

It is constant and is expressed as the precessional frequency of the magnetic moment of a specific MR-active nucleus at 1 T.

The unit of the gyromagnetic ratio is therefore MHz/T. The gyromagnetic ratio of hydrogen is 42.58 MHz/T.

Table 1.4 Magnetic characteristics of common elements.

Element	Nuclear spin	Gyromagnetic ratio (MHz/T)	Larmor frequency at 1.5 T (MHz)
^{1}H (hydrogen)	1/2	42.5774	63.8646
^{13}C (carbon)	1/2	10.7084	16.0621
^{15}N (nitrogen)	1/2	4.3173	6.4759
^{17}O (oxygen)	5/2	5.7743	8.6614

In addition, magnetic moments of MR-active nuclei have different precessional frequencies at different field strengths. For hydrogen, for example:

- At 1.5 T, the precessional frequency is 63.87 MHz (42.58 MHz × 1.5 T).
- At 1.0 T, the precessional frequency is 42.57 MHz (42.58 MHz × 1.0 T).
- At 0.5 T, the precessional frequency is 21.29 MHz (42.58 MHz × 0.5 T).

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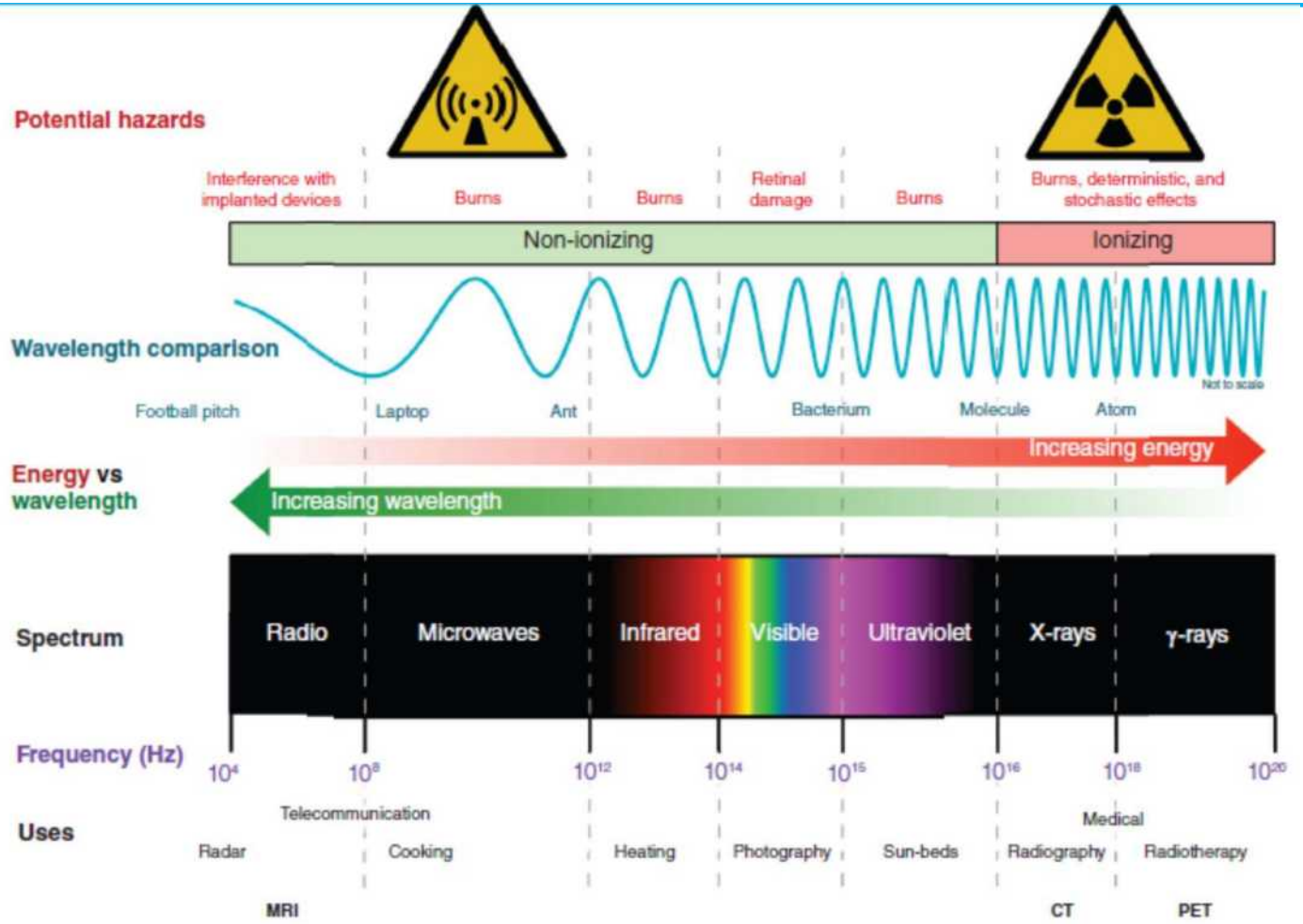
Radiofrequency (RF) band of the electromagnetic spectrum



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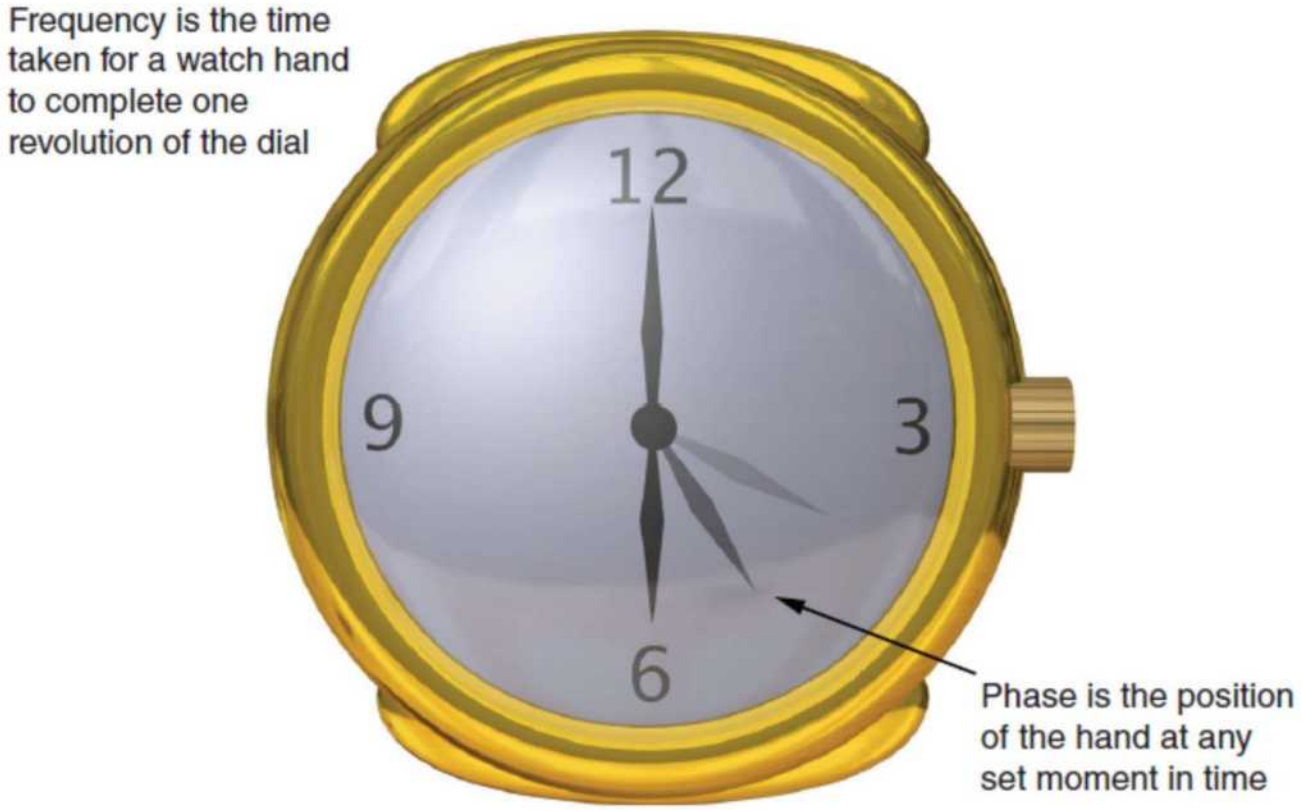
Precessional phase

Phase refers to the position of magnetic moments on their precessional path at any moment in time.

The unit of phase is a radian.

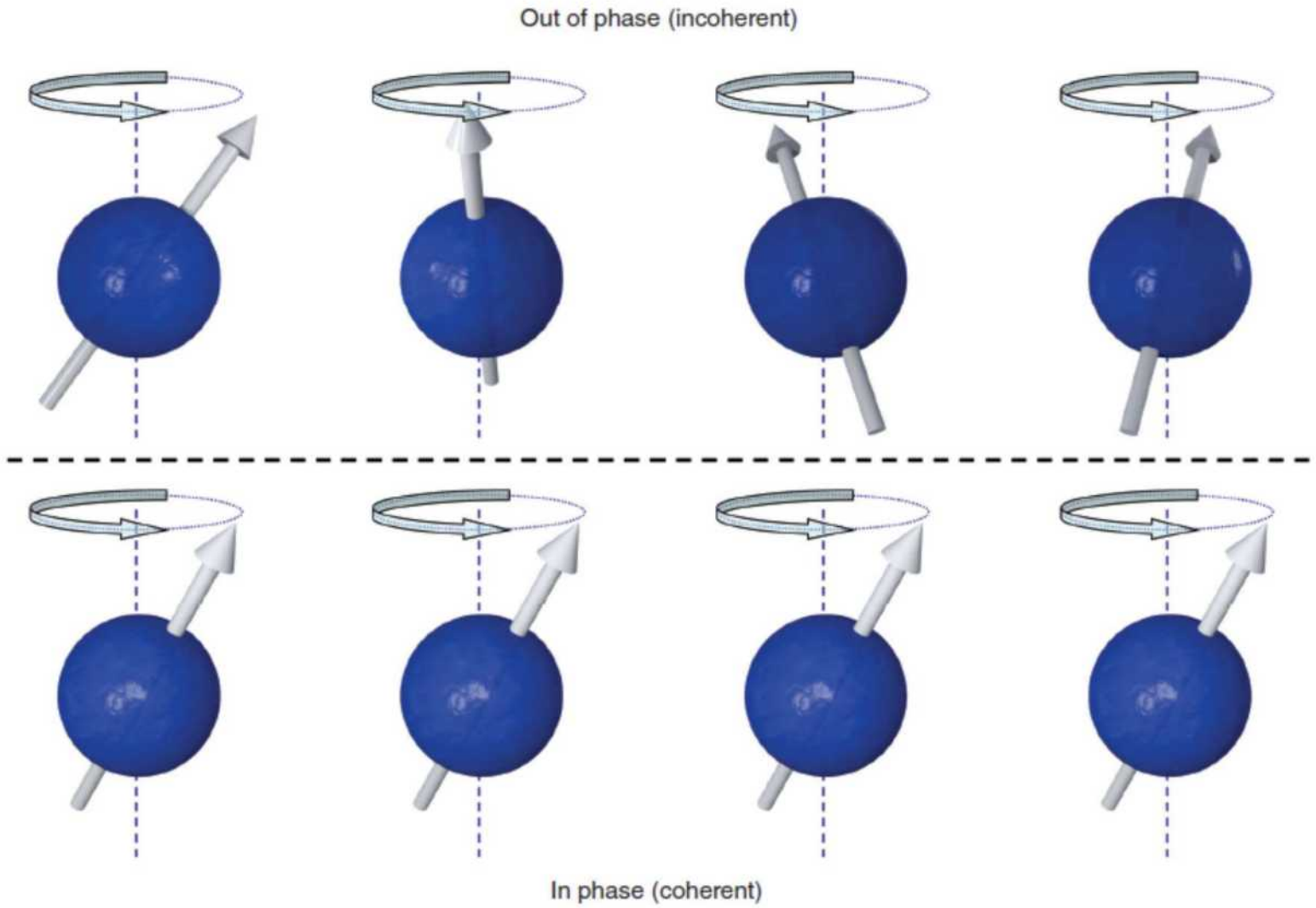
A magnetic moment travels through 360 rad or 360° during one rotation.

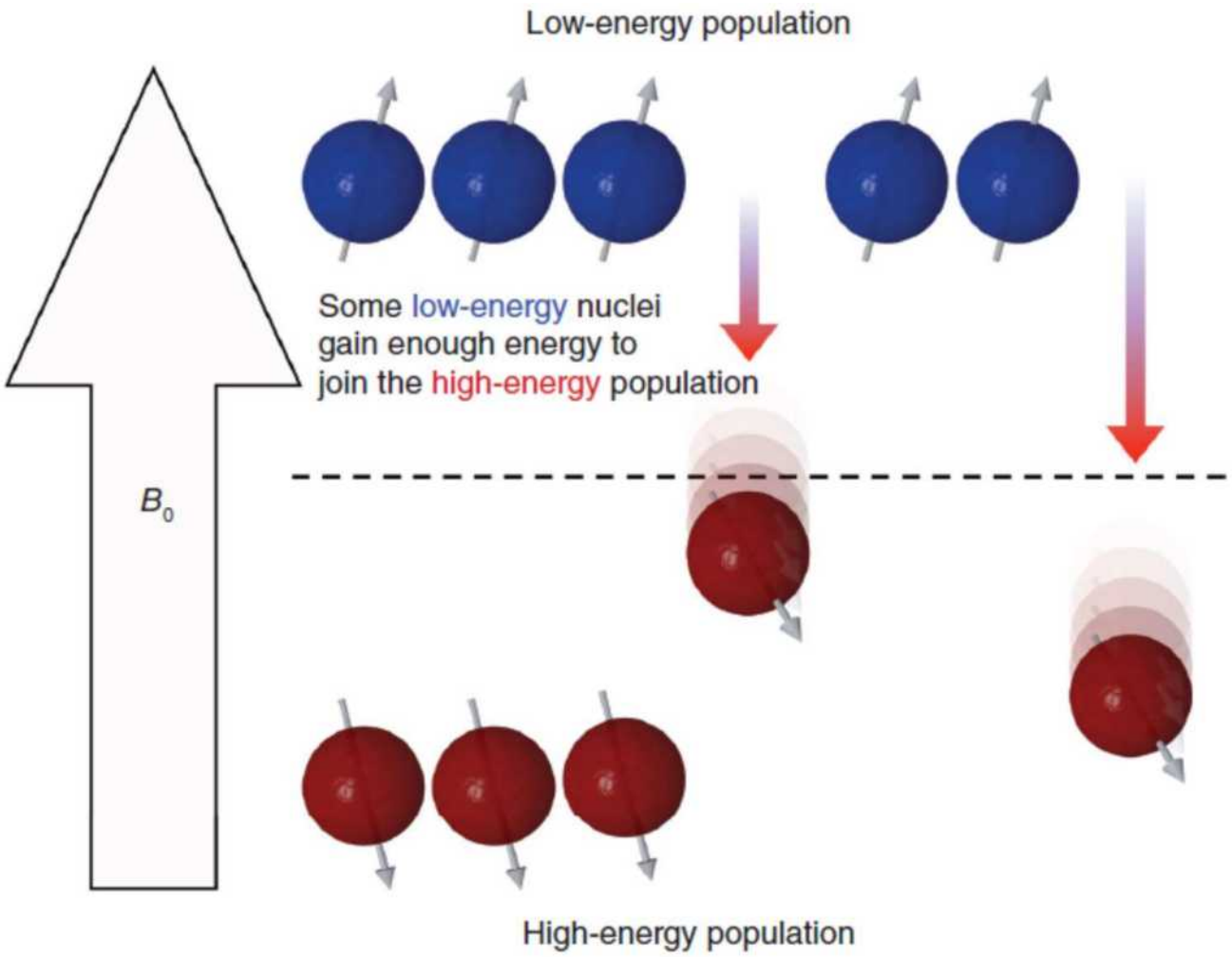
In this context, frequency is the rate of change phase of magnetic moments, i.e. it is a measure of how quickly the phase position of a magnetic moment changes over time.


Frequency is the time taken for a watch hand to complete one revolution of the dial
12
9
3
6
Phase is the position of the hand at any set moment in time

In MRI, the relative phase positions of all magnetic moments of hydrogen are important.

- **Out of phase or incoherent means that magnetic moments of hydrogen are at different places on the precessional path at a moment in time.**
- **In phase or coherent means that magnetic moments of hydrogen are at the same place on the precessional path at a moment in time. When the only influence is B0, the magnetic moments of the nuclei are out of phase with each other, and therefore the NMV does not precess.**


Out of phase (incoherent)
In phase (coherent)


Low-energy population
Some low-energy nuclei
gain enough energy to
join the high-energy population
B_0
High-energy population

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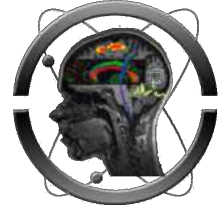
Resonance

Resonance is a phenomenon that occurs when an object is exposed to an oscillating perturbation that has a frequency close to its own natural frequency of oscillation.

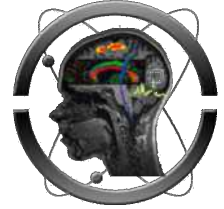
When a nucleus is exposed to an external force that has an oscillation similar to the natural frequency of its magnetic (its Larmor frequency), the nucleus gains energy from the external source.



Resonance is achieved by transmitting an RF pulse called an RF excitation pulse. This is produced by a transmit coil.

The RF excitation pulse produces an oscillating magnetic field, termed B1. The B1 field is applied at 90° to B0 at a narrow range or bandwidth of frequencies centered around a central frequency (termed the transmit bandwidth).

The magnetic field associated with the RF excitation pulse B1 is very weak compared with that of the main external field B0.



That a flip angle of 180° is caused when the RF excitation pulse is twice the magnitude of that used to produce a 90° flip angle.

In quantum mechanics, a 180° RF pulse produces an inversion of the spin populations, i.e. all the low-energy spins have enough energy to locate in the high-energy population and all the high-energy spins have been stimulated to give up their energy and locate in the low-energy population. This is called saturation and is caused when the spins are unable to absorb more energy or to be stimulated and release more energy. The amount of RF needed to produce a 90° flip angle is half of this value and relates to equalizing the high- and low-energy spins.

Equation 1.5		
$\theta = \omega_1 \tau$ Therefore from the Larmor equation $\theta = \gamma B_1 \tau$ $90° = \pi/2 = \gamma B_1 \tau$ $180° = \pi = \gamma B_1 \tau$	θ is the flip angle (°) ω_1 is the precessional frequency of B_1 (μT) B_1 is the magnetic field associated with the RF excitation pulse (mT) τ is the duration of the RF excitation pulse (ms)	This equation shows that the flip angle is determined by the strength of the B_1 field and the duration of the pulse. In trigonometry, a factor of 2π represents 360° A flip angle of 90° can therefore be written as $\pi/2$; a flip angle of 180° is π. Replacing θ with these values shows that an RF pulse producing a flip angle of 90° has either half the power or half the duration of an 180° RF pulse [9]

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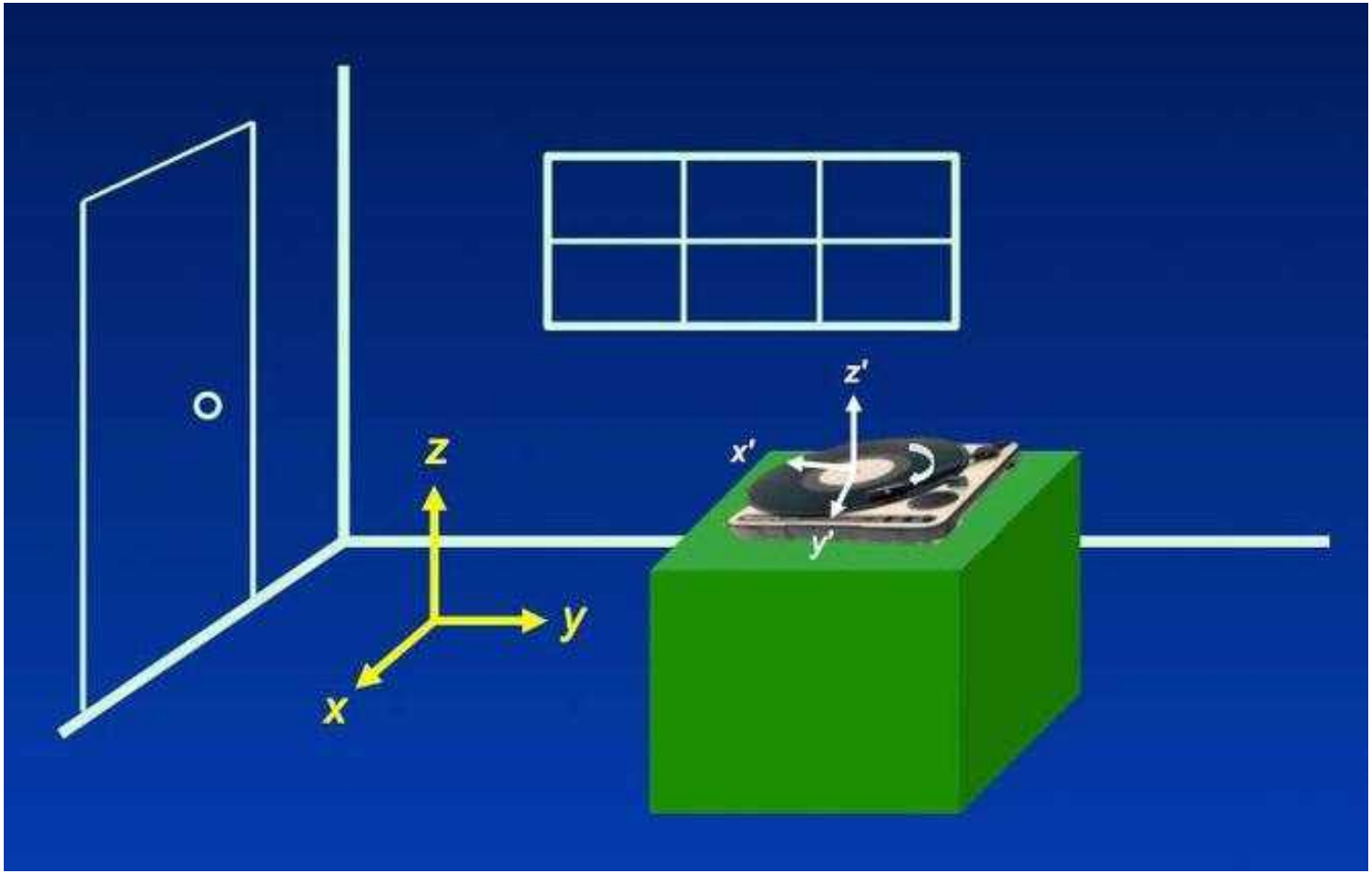
Stationary vs rotating frame of reference

- **The stationary frame** of reference refers to the observer (i.e. you) viewing something moving.

- **The rotating frame** of reference refers to the observer viewing this from a different perspective.


z'
x'
y'
z
y
x

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MR Signal

Because of resonance, in-phase or coherent magnetization precesses in the transverse plane.

This changing magnetic field generates an electric current.

Faraday's law of induction explains this phenomenon.

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MR Signal

The change of magnetic flux through a closed circuit induces an electromotive force (emf) in the circuit.

The laws of electromagnetic induction (Faraday) state that the induced emf:

- **is proportional to the rate of change of magnetic field and the area of the circuit**
- **is proportional to the number of turns in a coil of wire**
- **is in a direction so that it opposes the change in magnetic field that causes it (Lenz's law)**

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MR Signal

According to Faraday's law,

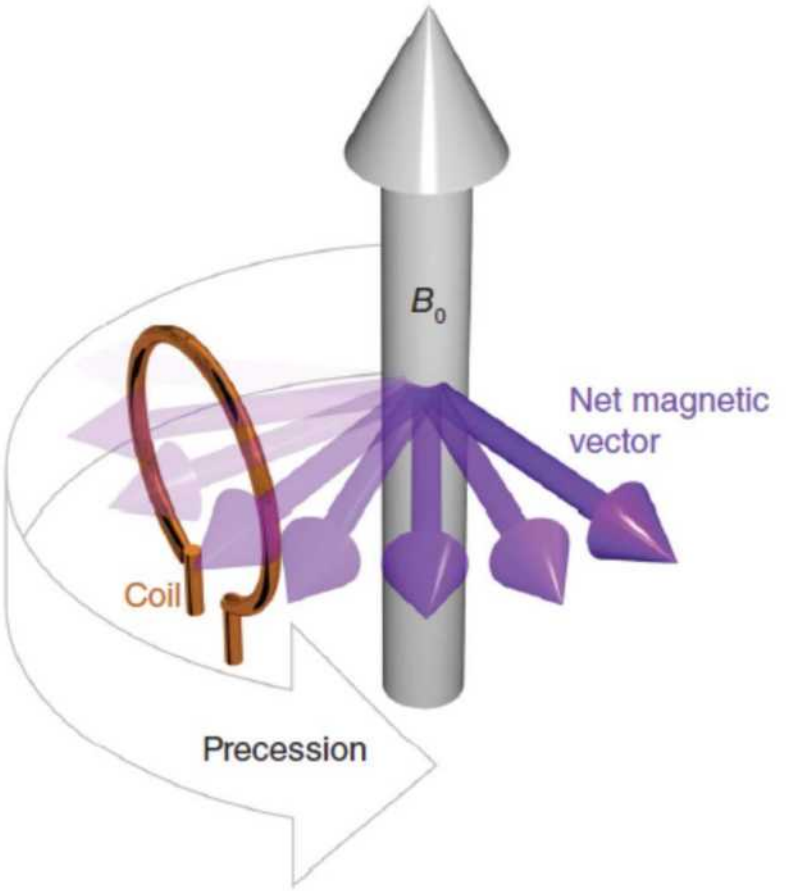
a changing magnetic field causes movement of charged particles, i.e. electrons.

This flow of electrons is a current, and if a receiver coil or any conductive loop is placed in a moving magnetic field, i.e. the magnetization precessing in the transverse plane, a voltage generated by this current is induced in the receiver coil.

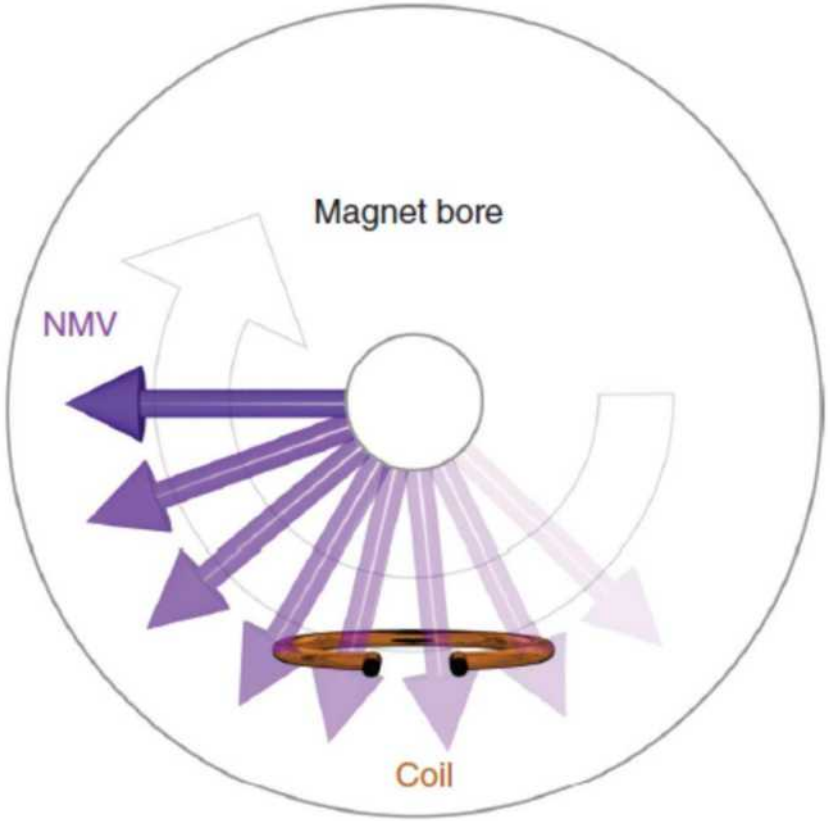
This voltage is called signal and is produced when coherent (in phase) magnetization cuts across the coil.

The frequency of signal depends on the frequency of rotation of the magnetic field – the magnitude of signal depends on the amount of coherent magnetization present in the transverse plane.

Net magnetic vector at 90° to B_0



End view, looking down on the transverse plane



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The free induction decay (FID) signal

Relaxation:

When the RF excitation pulse is switched off, the NMV is influenced only by B0, and it tries to realign with it.

To do so, the hydrogen nuclei lose energy given to them by the RF excitation pulse.

As relaxation occurs, the NMV returns to realign with B0 because some of the high-energy nuclei return to the low-energy population and therefore align their magnetic moments in the spin-up direction.

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The free induction decay (FID) signal

Finally, the magnetic moments of hydrogen lose coherency due to dephasing.

This occurs because of inhomogeneities in the B0 field and due to interactions between spins in the patient's tissue.

FID → The magnitude of transverse coherent magnetization decreases, so does the magnitude of the voltage induced in the receiver coil.

Spins freely precess influenced only by B0,

signal decays with time

magnetic moments of the spins induce a current in the receiver coil.

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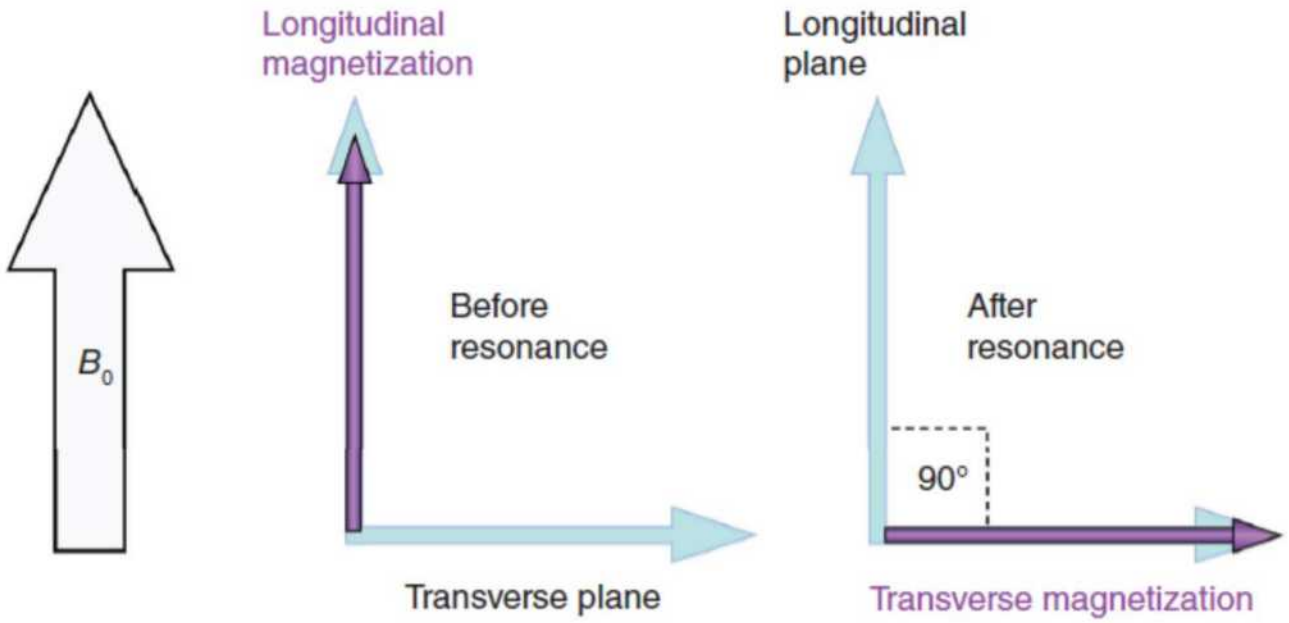
Longitudinal and transverse magnetization



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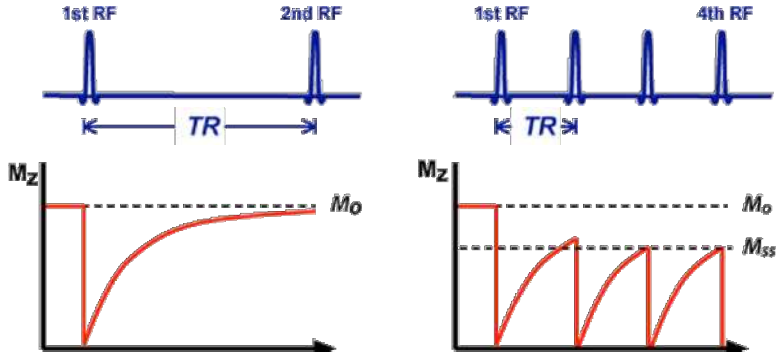
Pulse timing parameters

Repetition Time:

The TR is the time from the application of one RF excitation pulse to the application of the next RF excitation pulse for each slice and is measured in millisecond. The TR determines the amount of longitudinal relaxation that occurs between the end of one RF excitation pulse and application of the next. The TR thus determines the amount of T1 relaxation that has occurred when signal is read.



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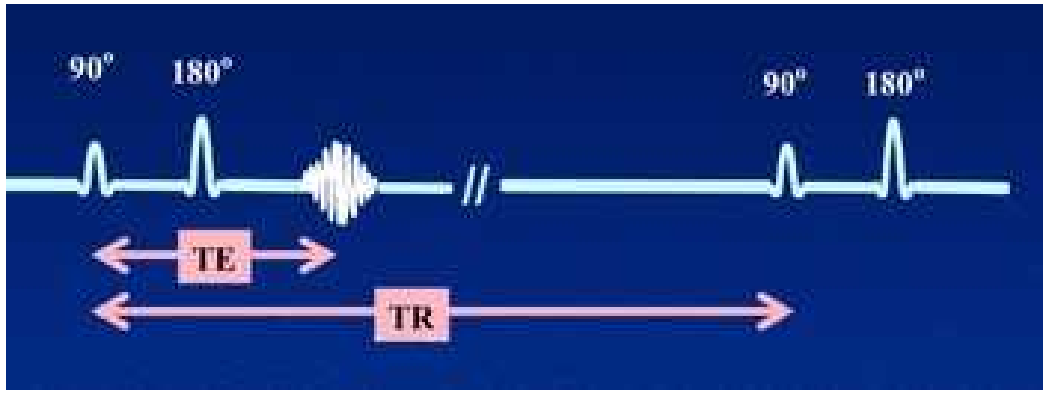
Pulse timing parameters

Time to Echo:

The TE is the time from the application of the RF excitation pulse to the peak of signal induced in the receiver coil and is also measured in millisecond. The TE determines how much decay of transverse magnetization occurs. TE thus controls the amount of T2 relaxation that has occurred when signal is read.



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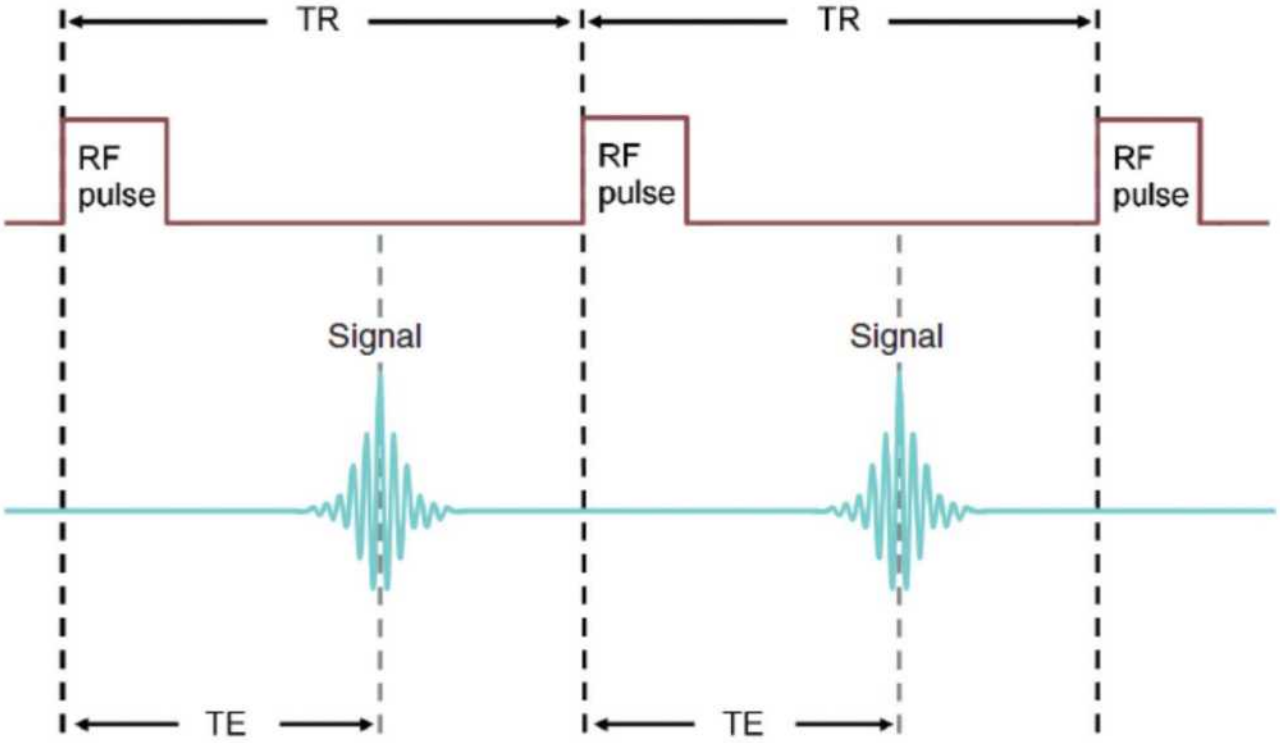
A basic pulse sequence



1.5T		3T	
SE		**SE**	
Short TE	Min–30 ms	Short TE	Min–15 ms
Long TE	70 ms+	Long TE	70 ms+
Short TR	600–800 ms	Short TR	600–900 ms
Long TR	2000 ms+	Long TR	2000 ms+
FSE		**FSE**	
Short TE	Min–20 ms	Short TE	Min–15 ms
Long TE	90 ms+	Long TE	90 ms+
Short TR	400–600 ms	Short TR	600–900 ms
Long TR	4000 ms+	Long TR	4000 ms+
Short TEL	2–6	Short TEL	2–6
Long ETL	16+	Long ETL	16+
IR T1		**IR T1**	
Short TE	Min–20 ms	Short TE	Min–20 ms
Long TR	3000 ms+	Long TR	300 ms+
TI	200–600 ms	TI	Short or null time of tissue
Short ETL	2–6	Short ETL	2–6
STIR		**STIR**	
Long TE	60 ms+	Long TE	60 ms+
Long TR	3000 ms+	Long TR	3000 ms+
Short TI	100–175 ms	Short TI	210 ms
Long ETL	16+	Long ETL	16+
FLAIR		**FLAIR**	
Long TE	80 ms+	Long TE	80 ms+
Long TR	9000 ms+	Long TR	9000 ms + (TR at least 4 × TI)
Long TI	1700–2500 ms (depending on TR)	Long TI	1700–2500 ms (depending on TR)
Long ETL	16+	Long ETL	16+
Coherent GRE		**Coherent GRE**	
Long TE	15 ms+	Long TE	15 ms+
Short TR	<50 ms	Short TR	<50 ms
Flip angle	20–50°	Flip angle	20–50°
Incoherent GRE		**Incoherent GRE**	
Short TE	Minimum	Short TE	Minimum
Short TR	<50 ms	Short TR	<50 ms
Flip angle	20–50°	Flip angle	20–50°
Balanced GRE		**Balanced GRE**	
TE	Minimum	TE	Minimum
TR	Minimum	TR	Minimum
Flip angle	>40°	Flip angle	>40°
SSFP		**SSFP**	
TE	10–15 ms	TE	10–15 ms
TR	<50 ms	TR	<50 ms
Flip angle	20–40°	Flip angle	20–40°

1.5T and 3T

Slice thickness 2D		**Slice thickness 3D**	
Thin	2–4 mm	Thin	<1 mm
Medium	5–6 mm	Thick	>3 mm
Thick	8 mm		
FOV		**Matrix**	
Small	<18 cm	Coarse	256 × 128 / 256 × 192
Medium	18–30 cm	Medium	256 × 256 / 512 × 256
Large	>30 cm	Fine	512 × 512
		Very fine	>1024 × 1024
NEX/NSA		**Slice number 3D**	
Short	1	Small	<32
Medium	2–3	Medium	64
Multiple	>4	Large	>128
PC-MRA 2D and 3D		**TOF-MRA 2D**	
TE	Minimum	TE	Minimum
TR	25–33 ms	TR	28–45 ms
Flip angle	30°	Flip angle	40–60°
VENC venous	20–40 cm/s	**TOF-MRA 3D**	
VENC arterial	60 cm/s	TE	Minimum
		TR	25–50 ms
		Flip angle	20–30°

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Image weighting and contrast

2

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Introduction

- All clinical diagnostic images must demonstrate contrast between normal anatomical features and between anatomy and pathology. If there is no contrast, it is impossible to identify anatomy or detect abnormalities within the body.

- One of the main advantages of MRI compared to other imaging modalities is its excellent soft tissue discrimination. The contrast characteristics of each image depend on many variables, and it is important that the mechanisms that affect image contrast in MRI are clearly understood.

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Image contrast

- The factors that affect image contrast in diagnostic imaging are usually divided into two categories:

- **Intrinsic contrast** parameters are those that cannot be changed because they are inherent to the body's tissues.
- **Extrinsic contrast** parameters are those that can be changed because they are under our control.

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Intrinsic contrast

❑ **Intrinsic contrast parameters are as follows:**

- **T1 recovery time**
- **T2 decay time**
- **Proton density (PD)**
- **Flow**
- **Apparent diffusion coefficient (ADC)**

All these are inherent to the body's tissues and cannot be changed. T1 recovery time, T2 decay time, proton density, and ADC are discussed in this chapter.

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Extrinsic contrast

- **Extrinsic contrast** parameters are as follows:

- TR
- TE
- Flip angle
- TI
- Turbo factor/echo train length
- b value.

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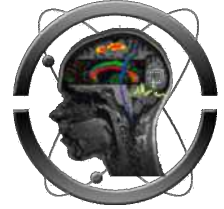
Relaxation

as soon as the B1 field is removed, hydrogen nuclei are only under the influence of B0. One of the principles of thermodynamics is that a system always seeks its lowest possible energy level.

This occurs in MRI when the RF excitation pulse is switched off ,hydrogen nuclei return to their low-energy state and their magnetic moments dephase. The process by which this occurs is called relaxation



Relaxation

- During relaxation, hydrogen nuclei give up absorbed RF energy, and the net magnetic vector (NMV) returns to B0.

- At the same time but independently, magnetic moments of hydrogen nuclei lose phase coherence.

- Relaxation therefore results in recovery of magnetization in the longitudinal plane and decay of coherent magnetization in the transverse plane.

- **The recovery of longitudinal magnetization is caused by a process termed T1 recovery.**

T1 recovery and T2 decay occur at two different rates.
T2 decay occurs 5–10 times faster than T1 recovery

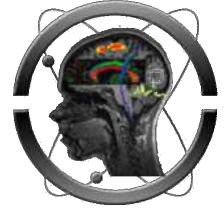
- **The decay of coherent transverse magnetization is caused by a process termed T2 decay.**

T1 recovery

T1 recovery is caused by hydrogen nuclei giving up their energy to the surrounding environment or molecular lattice. The term recovery refers to the recovery of longitudinal magnetization, and T1 relates to the fact that it is the primary relaxation process.

This type of relaxation is called spin–lattice energy transfer.

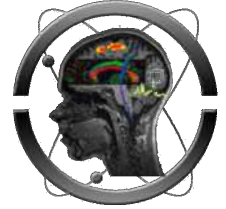
Finally, the magnetic moments of hydrogen nuclei to recover their longitudinal magnetization.



T1 recovery

- According to quantum theory, the number of high-energy spins decreases, and the number of low-energy spins increases as energy is lost by high-energy spins during the relaxation process.

- According to classical theory, the NMV gradually realigns itself in the longitudinal plane as the proportion of spin-up and spin-down hydrogen nuclei changes.



T1 recovery

- The rate of T1 recovery is an exponential process and occurs at different rates in different tissues.
- The longitudinal magnetization is related exponentially to recovery time. There is a time constant associated with this exponential relationship.
- This is called the T1 recovery time and is the time it takes for 63% of the longitudinal magnetization to recover in a tissue.
- The T1 recovery time of a tissue is an intrinsic contrast parameter that is inherent to the tissue.

The TR therefore determines how much T1 recovery occurs in a tissue

Table 2.1 Typical T1 recovery times of brain tissue at 1 T.

Tissue	T1 recovery time (ms)
Water	2500
Fat	200
CSF	2000
White matter	500

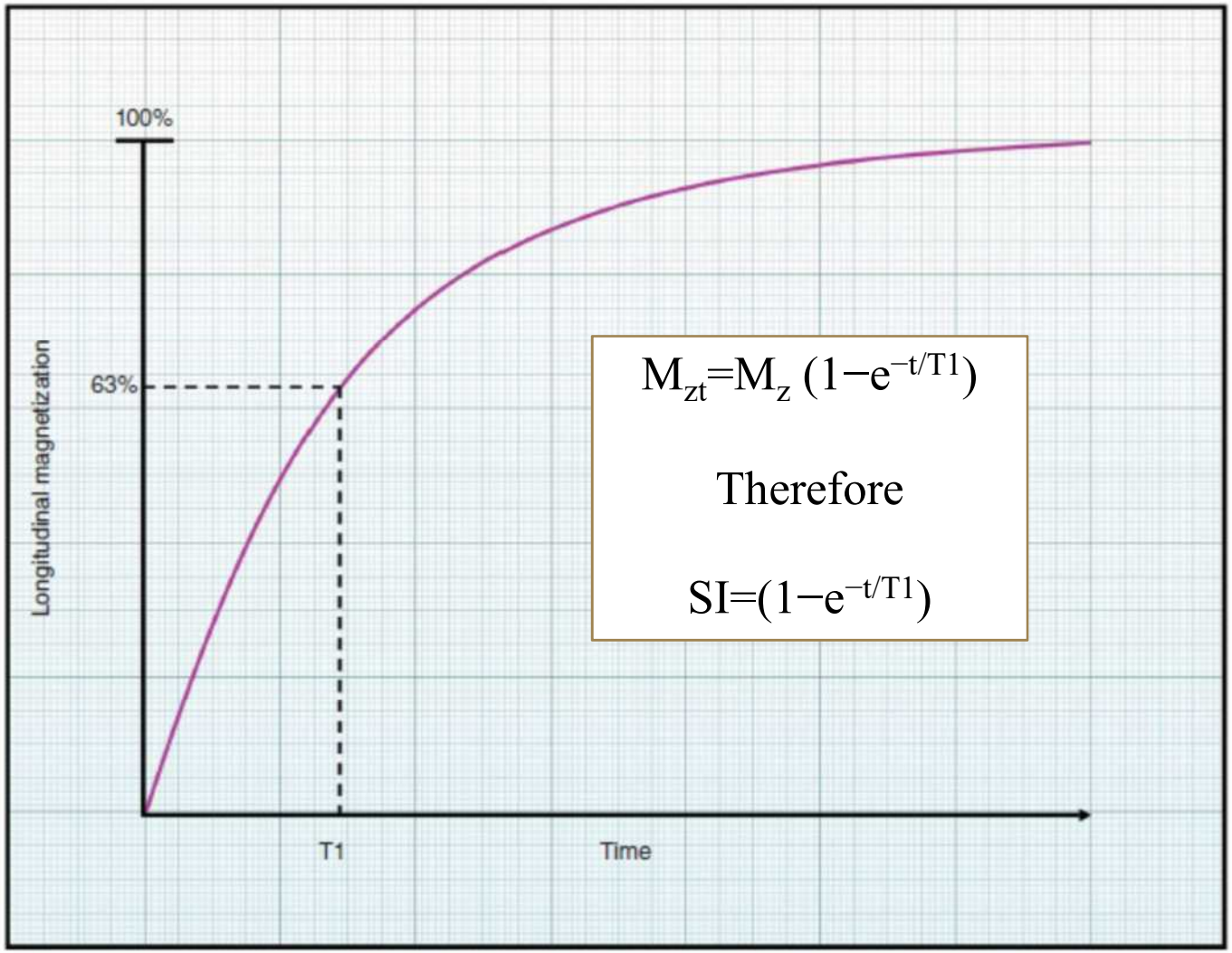


Figure 2.1 The T1 recovery curve.

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T2 decay

- ✓ **T2 decay is caused by the magnetic fields of neighboring hydrogen nuclei interacting with each other.**
- ✓ **The term decay refers to the loss of coherent transverse magnetization, and T2 relates to the fact that it is the secondary relaxation process.**
- ✓ **This type of relaxation is termed spin–spin relaxation and causes dephasing of magnetic moments of the spins.**

Spin–spin relaxation is caused by one spin transferring energy to another spin rather than into the lattice

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T2 decay

Imagine two spins close to each other,

one with its magnetic moment aligned in the same direction as B0 and the other in the opposite direction.

The spin whose magnetic moment is aligned in the same direction as B0 creates a slightly larger magnetic field than is experienced by the neighboring spin .As a result, the precessional frequency of the magnetic moment of this spin increases.

Conversely, the spin whose magnetic moment is aligned in the opposite direction to B0 causes a slightly lower magnetic field than is experienced by the other spin, and its precessional frequency decreases.

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T2 decay

❑ NOTE

It occurs because hydrogen nuclei are in the same environment and experiencing the same B0 field.

Magnetic moments of all the hydrogen nuclei (spin-up and spin-down) lose phase coherence in this way.

These small changes in frequency are sufficient to cause dephasing of magnetic moments of the spins.

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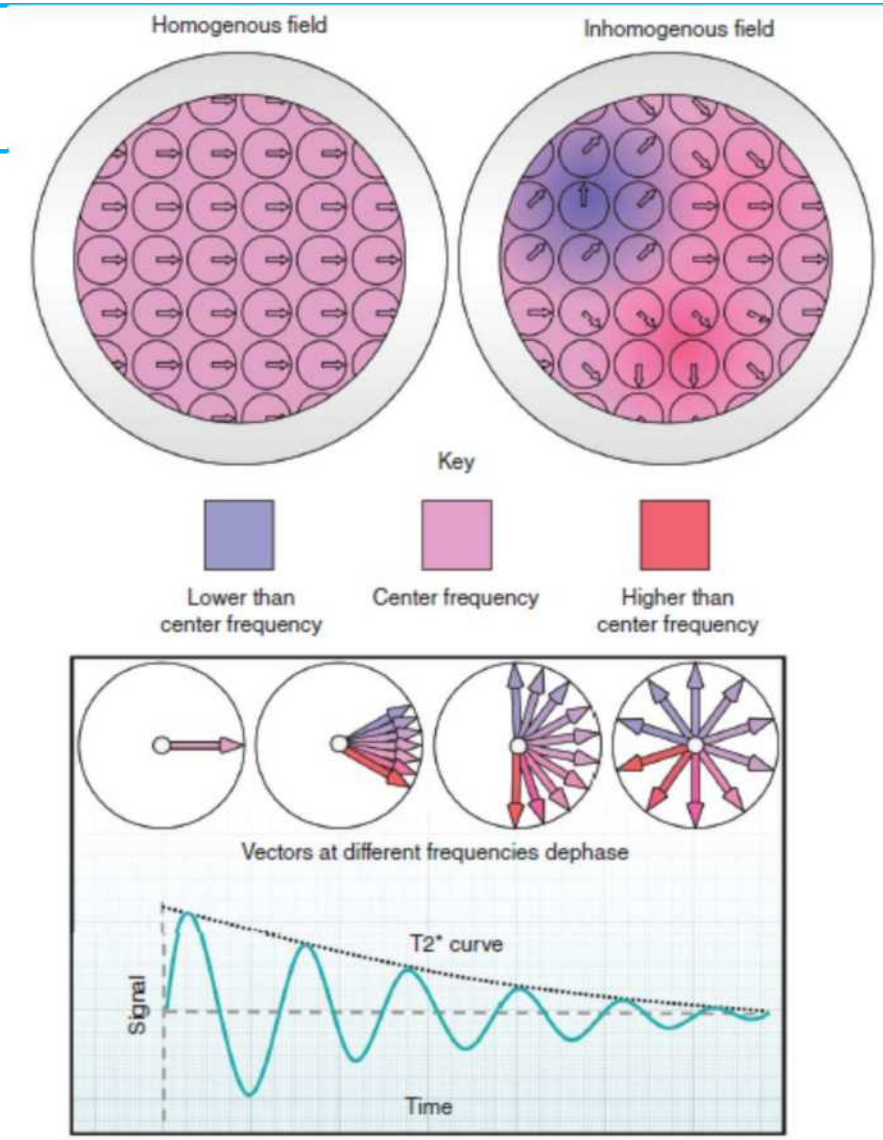
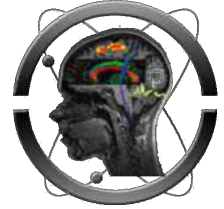
T2 decay

- **Spin–spin interaction** is inherent to the tissue, but dephasing is also caused by **inhomogeneities** in the B0 field.

- **Inhomogeneities** are areas within the magnetic field that do not exactly match the external magnetic field strength.

- Some areas have a magnetic field strength slightly less than the main magnetic field, while other areas have a magnetic field strength slightly higher than the main magnetic field.


Homogenous field
Inhomogenous field
Key
Lower than center frequency
Center frequency
Higher than center frequency
Vectors at different frequencies dephase
T2* curve
Signal
Time



T2 decay

According to the Larmor equation,

the Larmor frequency of an MR-active nucleus is proportional to the magnetic field strength it experiences. If a hydrogen nucleus lies in an area of inhomogeneity with higher field strength, the precessional frequency of its magnetic moment increases, i.e. it speeds up.

However, if a hydrogen nucleus lies in an area of inhomogeneity with lower field strength, the precessional frequency of its magnetic moment decreases, i.e. it slows down.

This relative acceleration and deceleration of magnetic moments due to magnetic field inhomogeneities, and differences in the precessional frequency in certain tissues, causes immediate dephasing of the magnetic moments of the spins and produces a free induction decay (FID).

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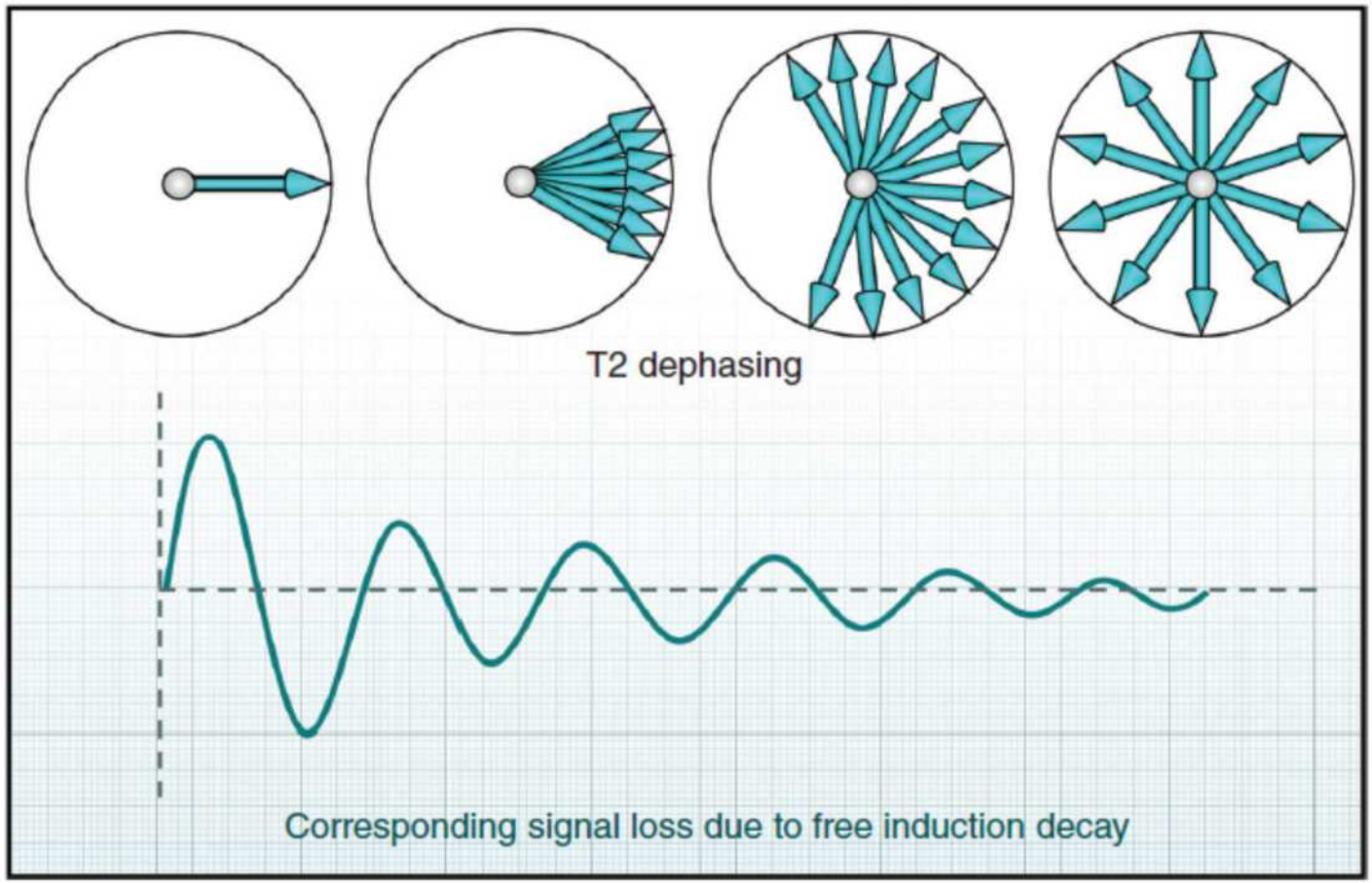
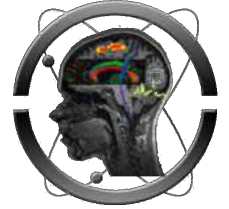
T2 decay





- **The rate of T2 decay is an exponential process and occurs at different rates in different tissues;**

Table 2.2 Typical T2 decay times of brain tissue at 1 T.

Tissue	T2 decay time (ms)
Water	2500
Fat	100
CSF	300
White matter	100

T2 decay

- **Coherent transverse magnetization** is related exponentially to decay time.
- This means that there is more coherent transverse magnetization at the beginning of the time-frame and, as time progresses, there is less coherent transverse magnetization until all the magnetic moments dephase.
- There is a time constant associated with this exponential relationship.

It is called the T2 decay time and is the time it takes for **63%** of the transverse magnetization to dephase (**37% is left in phase**) in a tissue.

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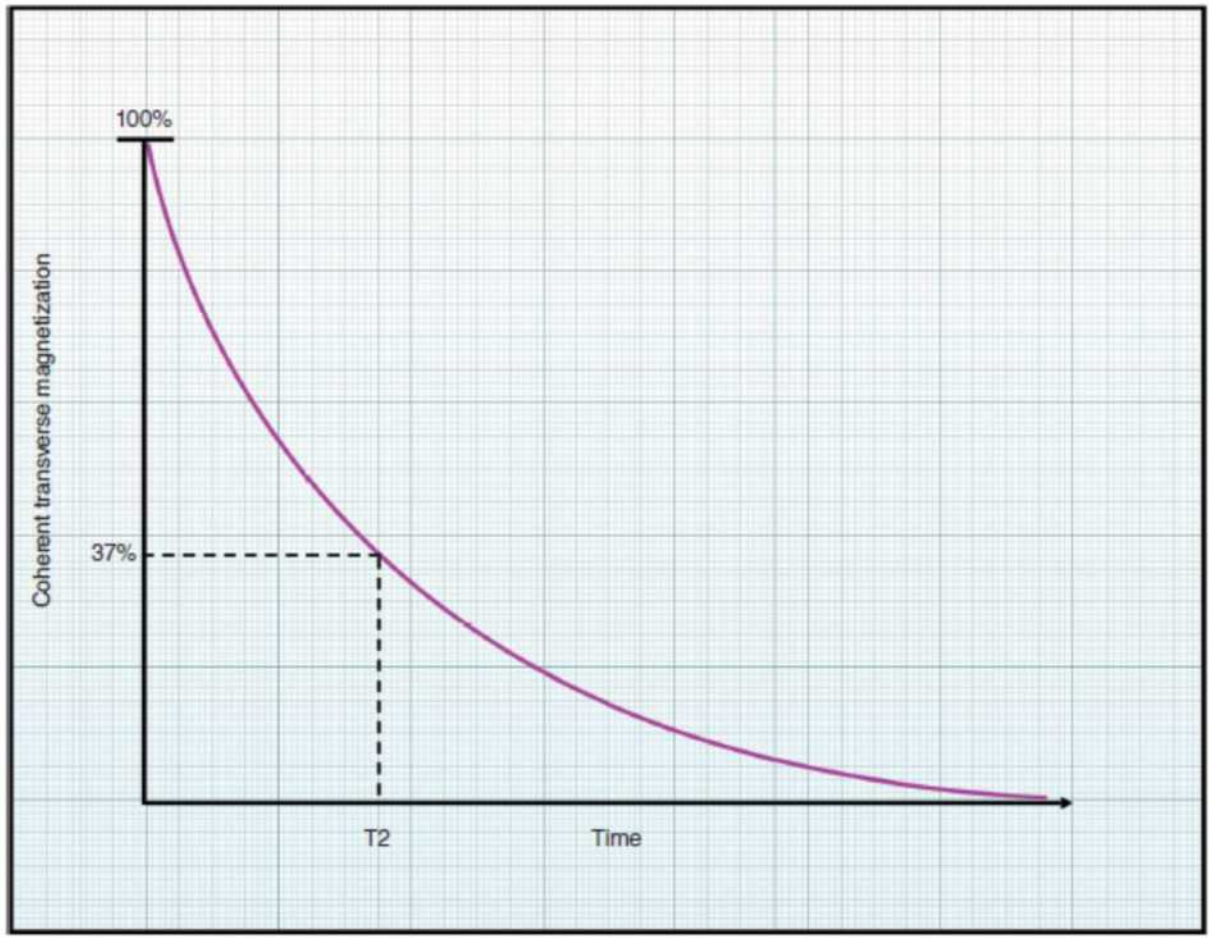
T2 decay



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T2 decay

- The T2 decay time of a tissue is an intrinsic contrast parameter that is inherent to the tissue.

- The time during which this occurs is the time between an RF excitation pulse and when signal is collected in the receiver coil.

The echo time (TE) therefore determines how much T2 decay occurs in a tissue when signal is collected

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T2 decay

- Dephasing caused by inhomogeneities in the B0 field produces its own decay curve. This is differentiated from T2 decay by using the term T2*.
- When the RF excitation pulse is switched off, magnetic moments of the hydrogen nuclei dephase very quickly (within about 10 ms), and this is caused by T2* decay.
- T2 decay, from inherent tissue dephasing, takes longer than T2* decay.
- The purpose of pulse sequences is to refocus or rephase magnetic moments of the hydrogen nuclei so that inherent tissue dephasing is measured at time TE and images of different contrast can be acquired.

$$1/T2^* = 1/T2 + 1/2\gamma\Delta\beta0$$

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T2 decay

Table 2.3 Things to remember – relaxation.

Relaxation is a general term that refers to a loss of energy. In MRI, this is energy that is delivered to the spins via the RF excitation pulse and then lost once it is switched off
Spin lattice energy transfer is a relaxation process where spins give up the energy absorbed through RF excitation to the surrounding molecular lattice of the tissue. It causes T1 recovery
T2 decay is an irreversible loss of phase coherence due to spin–spin interactions on an atomic and molecular level. It is one of the causes of T2 decay
Pulse sequences are mechanisms that permit refocusing of spins so that images can be acquired with different types of contrast

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Contrast mechanisms

- An MR image has contrast

if there are areas of

high signal (hyperintensity – white in the image)

and areas of

low signal (hypointensity – black in the image).

- Some areas have an

intermediate signal (shades of gray in between white and black).

The NMV is separated into individual vectors of the tissues such as fat, cerebrospinal fluid (CSF), and muscle.

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Contrast mechanisms

- ✓ A tissue has a high signal if it has a large transverse component of coherent magnetization at time TE.
- If there is a large component of coherent transverse magnetization, the amplitude of signal received by the coil is large, resulting in a hyperintense area on the image.
- ✓ A tissue returns a low signal if it has a small or no transverse component of coherent magnetization at time TE.
- If there is a small or no component of transverse coherent magnetization, the amplitude of signal received by the coil is small, resulting in a hypointense area on the image.

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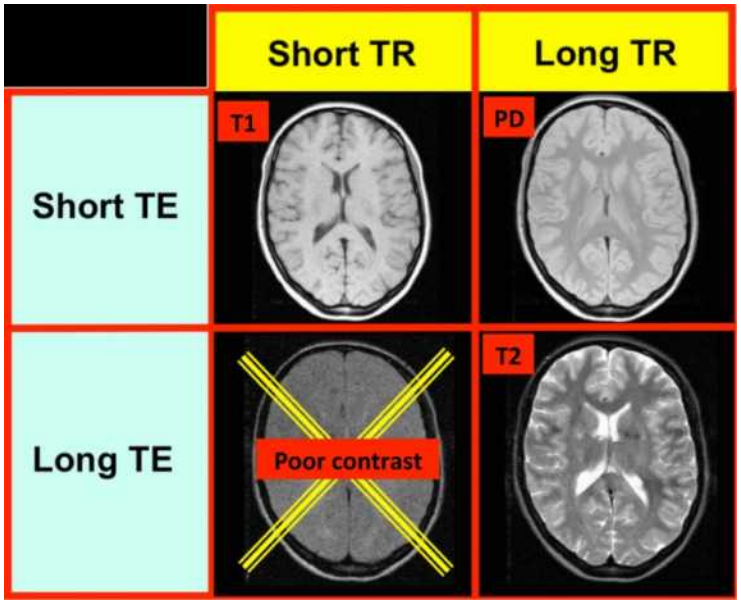

The proton density (PD) of a tissue is the number of mobile hydrogen protons per unit volume of that tissue. The higher the proton density of a tissue, the more signal available from that tissue.

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Contrast mechanisms

T1 and T2 relaxation depend on two factors:

- **If the molecular tumbling rate matches the Larmor frequency of hydrogen. If there is a good match between the rate of molecular tumbling and the Larmor frequency of magnetic moments of hydrogen, energy exchange between hydrogen nuclei and the molecular lattice is efficient. When there is a bad match, energy exchange is not as efficient.**

This is important in both T1 recovery and T2 decay processes

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Contrast mechanisms

- If the molecules are closely packed together. In tissues where molecules are closely spaced, there is more efficient interaction between the magnetic fields of neighboring hydrogen nuclei. The reverse is true when molecules are spaced apart.

This is especially important in T2 decay processes, which rely on the efficiency of interactions between the magnetic fields of neighboring hydrogen nuclei (spin–spin relaxation).

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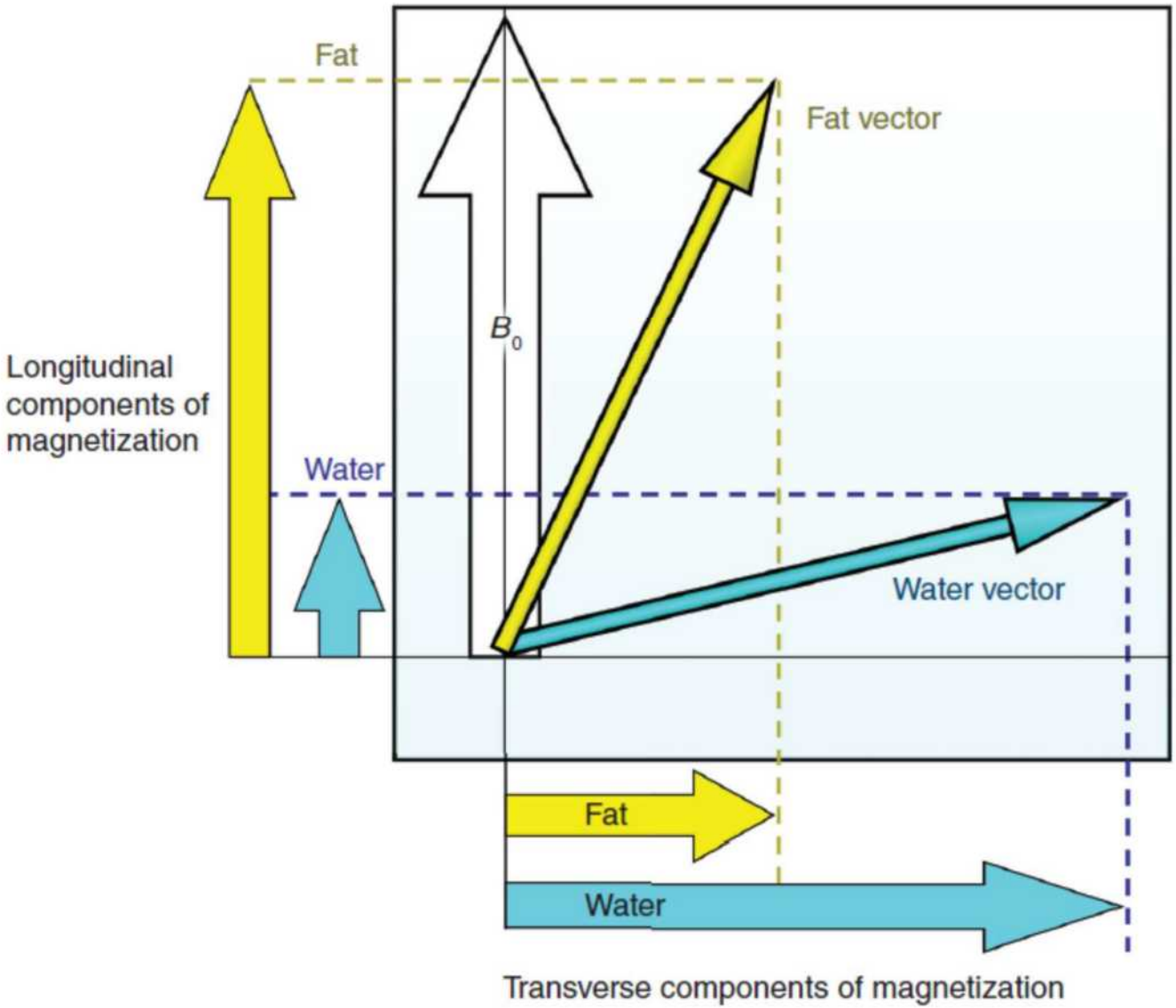
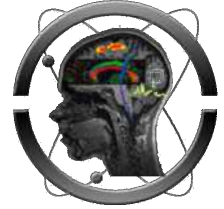
Relaxation in different tissues

As discussed earlier,

T1 recovery and T2 decay are exponential processes with time constants T1 recovery time and T2 decay time, which represent the time it takes for 63% of the total magnetization to recover in the longitudinal plane due to spin–lattice energy transfer (T1 recovery time), or lost in the transverse plane via spin–spin relaxation (T2 decay time).

Generally, the two extremes of contrast in MRI are fat and water.


Fat
Fat vector
B_0
Longitudinal components of magnetization
Water
Water vector
Fat
Water
Transverse components of magnetization



Fat and water

- **Fat** molecules contain atoms of hydrogen arranged with carbon and oxygen. They consist of **large molecules** called **lipids** that are closely packed together and whose molecular motion or tumbling rate is relatively slow.

- **Water** molecules contain **two hydrogen** atoms arranged with **one oxygen** atom (H2O). Its molecules are spaced apart, and their molecular tumbling rate is relatively fast.

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NOTE

The oxygen in water tends to steal the electrons away from around the hydrogen nucleus. This renders it more available to the effects of the main magnetic field.

In fat, the carbon does not take the electrons from around the hydrogen nucleus. They remain in an electron cloud, protecting the nucleus from the effects of the main field.

Therefore, hydrogen in fat recovers more rapidly along the longitudinal axis than water and loses transverse magnetization faster than water. Subsequently, fat and water appear differently in MR images

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T1 recovery in fat

- T1 recovery occurs due to hydrogen nuclei giving up their energy to the surrounding molecular lattice.
- Fat has a low inherent energy and easily absorbs energy into its lattice from hydrogen nuclei.
- The slow molecular tumbling in fat allows the T1 recovery process to be relatively rapid because the molecular tumbling rate matches the Larmor frequency.
- Consequently, there is efficient energy exchange from hydrogen nuclei to the surrounding molecular lattice.

This means that magnetic moments of fat hydrogen nuclei quickly relax and regain their longitudinal magnetization. The NMV of fat realigns rapidly with B0, so the T1 recovery time of fat is short

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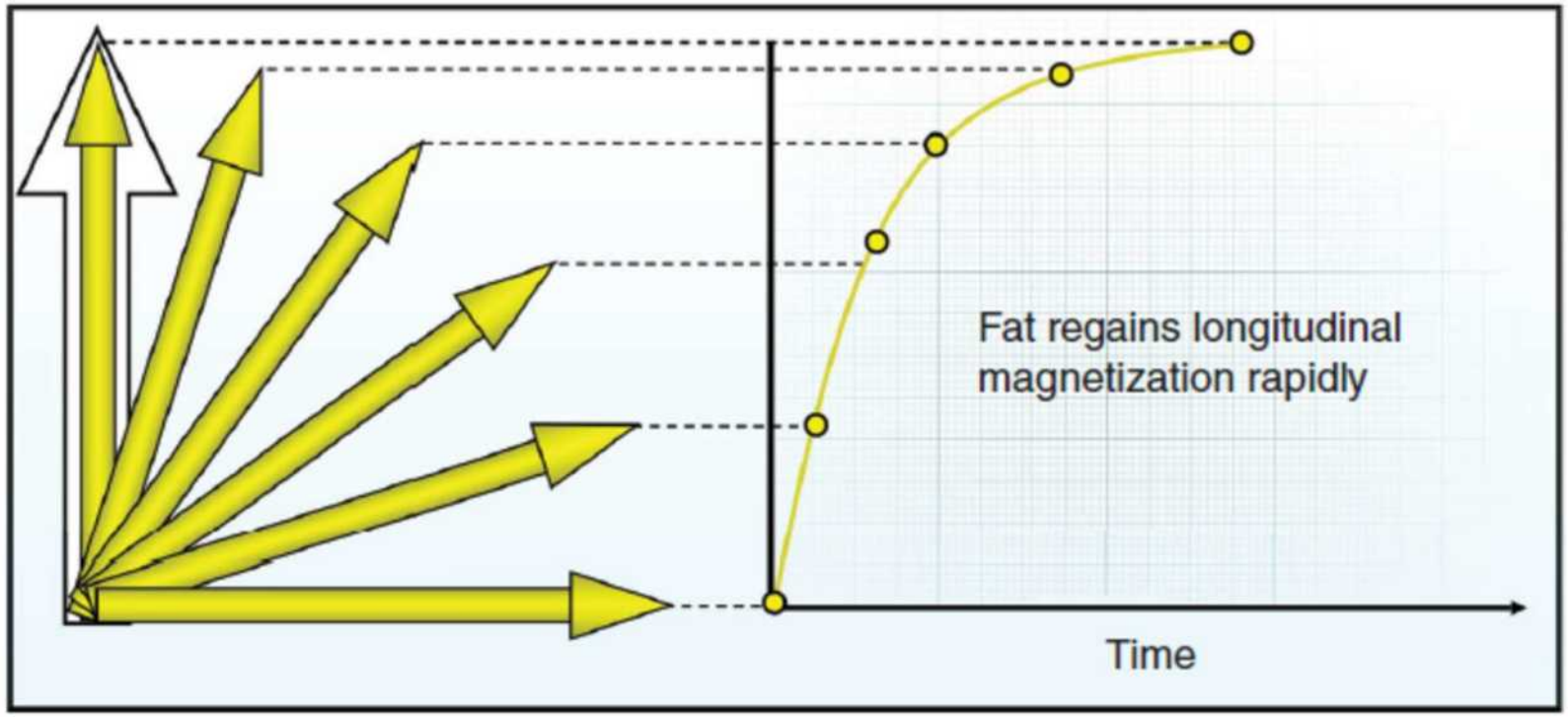
T1 recovery in fat



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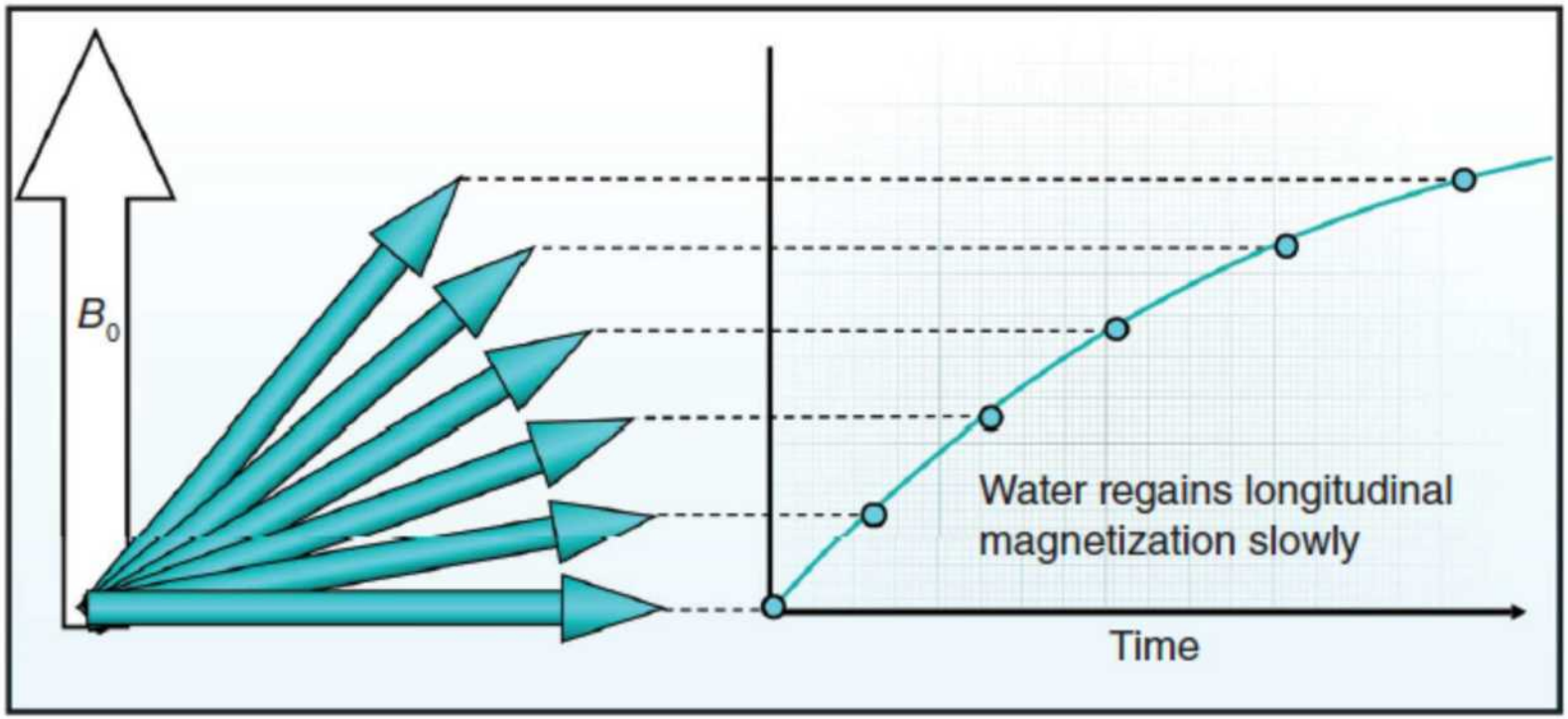
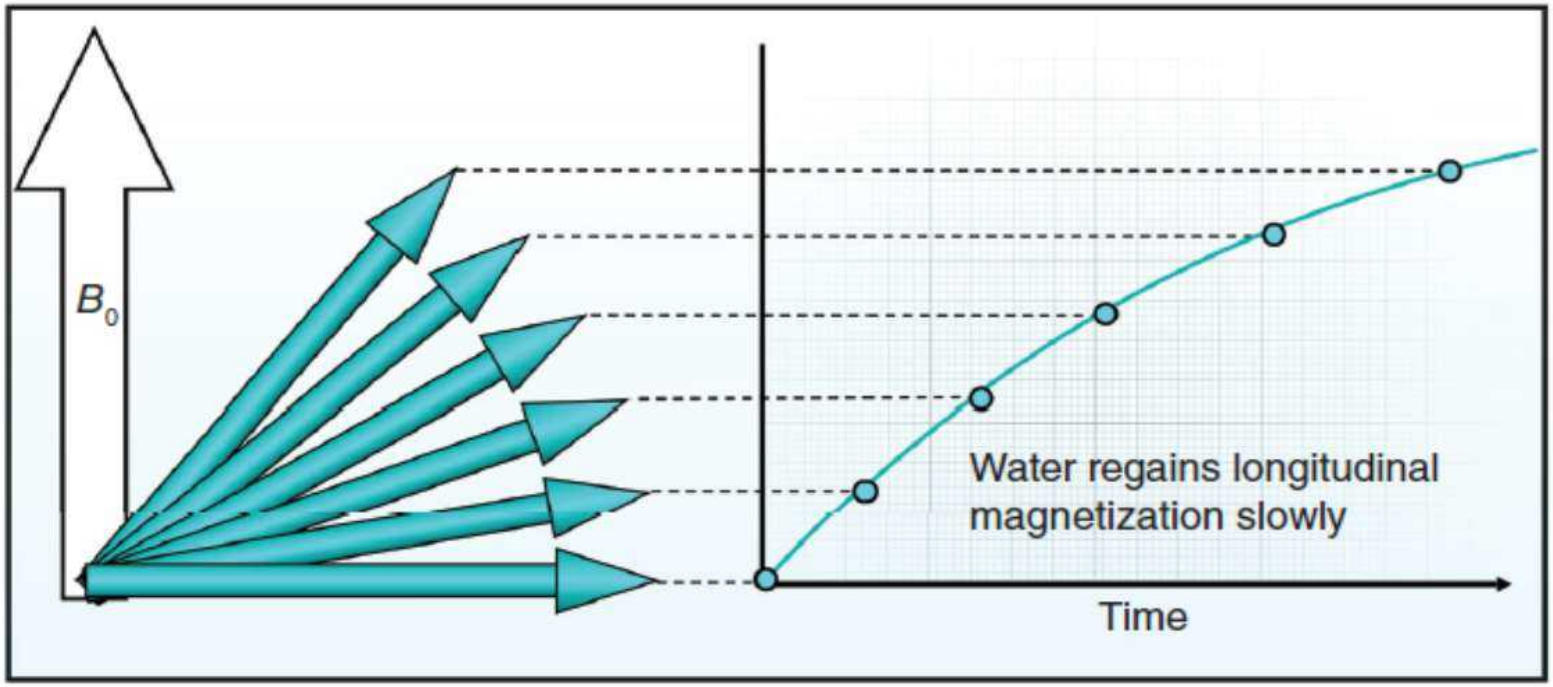
T1 recovery in water




B_0
Water regains longitudinal magnetization slowly
Time

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T2 decay in fat

- T2 decay occurs because the magnetic fields of hydrogen nuclei interact with each other.
- This process is efficient in hydrogen in fat, as the molecules are packed closely together, and therefore spin–spin interactions are more likely to occur.
- It also occurs because magnetic moments of hydrogen nuclei in fat precess at a similar frequency to molecular tumbling.
- As a result, magnetic moments dephase quickly, and there is a rapid loss of coherent transverse magnetization.

The T2 decay time of fat is therefore short

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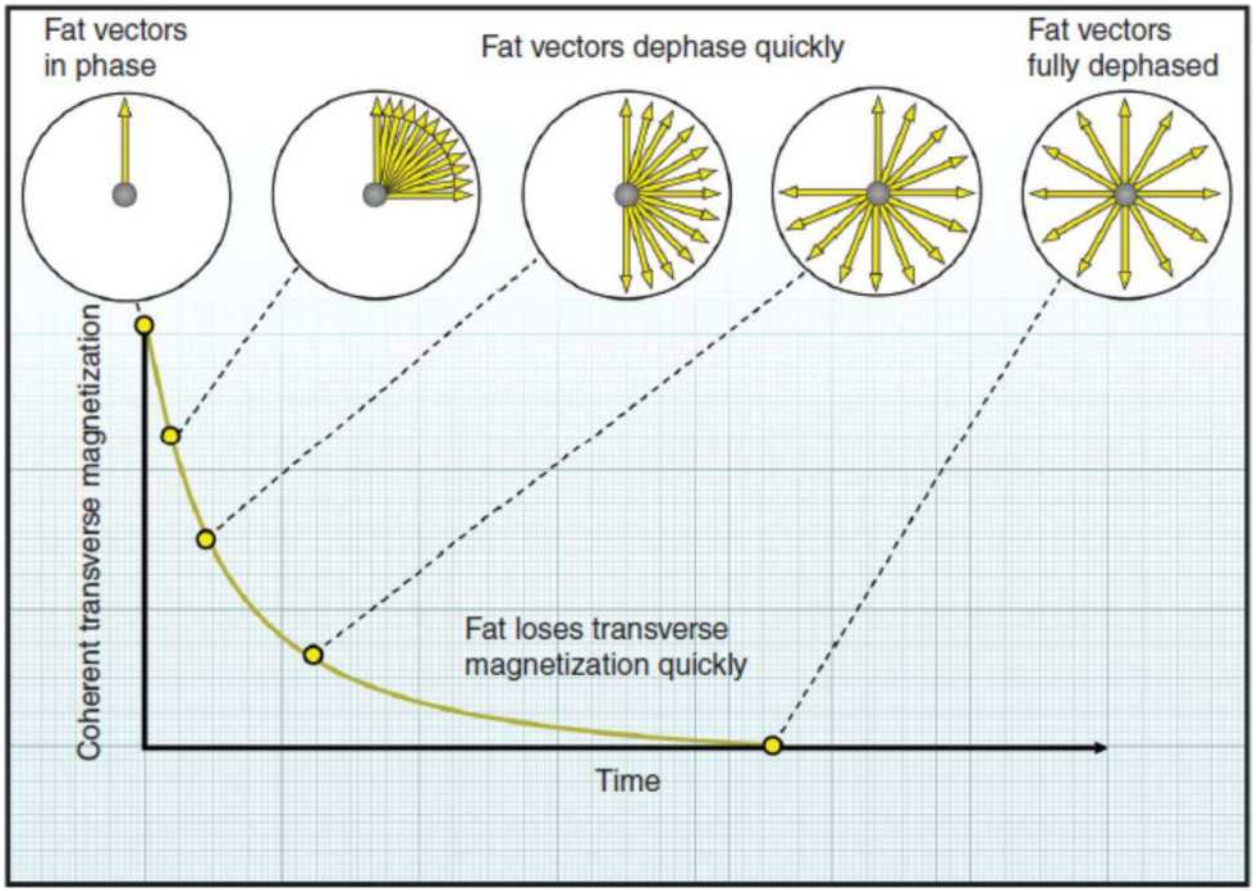
T2 decay in fat



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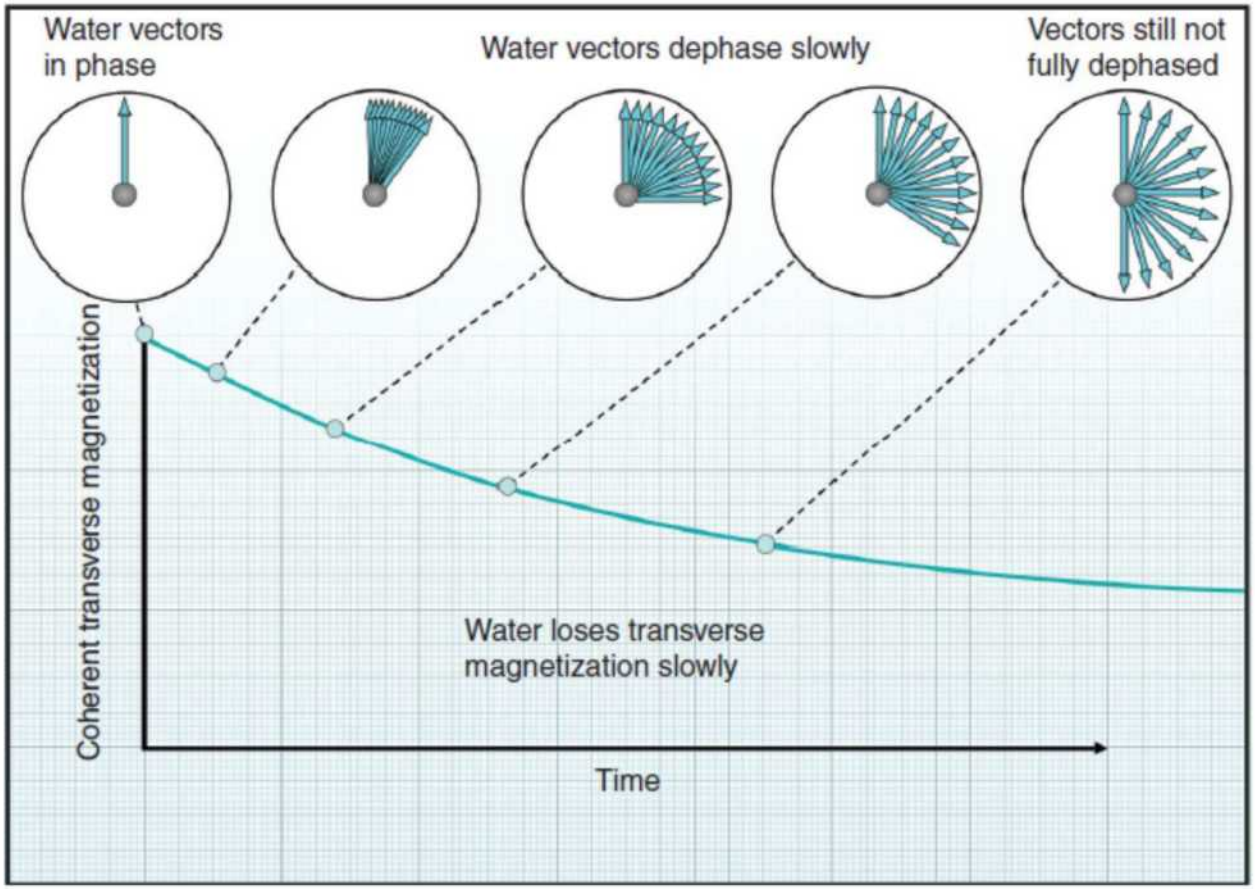
T2 decay in water



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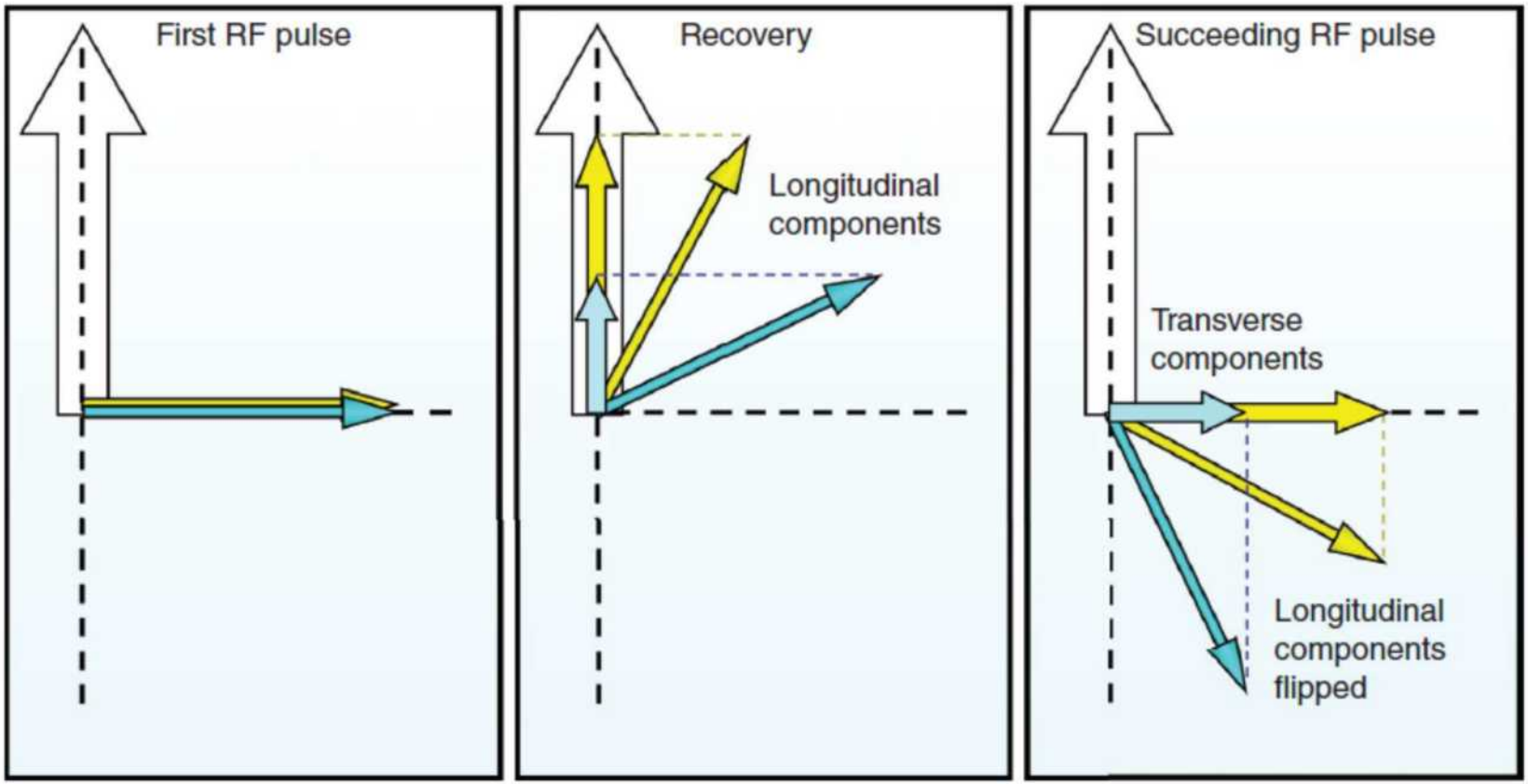
T1 contrast



Learning tip:
Saturation

Whenever the NMV is pushed beyond 90°, it is said to be **partially saturated**. When the NMV is pushed to a full 180°, it is said to be **fully saturated**. If partial saturation of the fat and water vectors occurs, TI contrast is maximized.

Look at Figure 2.11. Before application of the first RF excitation pulse, the fat and water vectors are aligned with B_0. When the first 90° RF excitation pulse is applied, the fat and water vectors are flipped into the transverse plane. The RF excitation pulse is then removed, and the vectors begin to relax and return to B_0. Fat has a shorter T1 recovery time than water and so returns to B_0 faster than water. If the TR is shorter than the T1 recovery times of the tissues, the next RF excitation pulse flips the vectors beyond 90° and into partial saturation because their recovery was incomplete. The fat and water vectors are saturated to different degrees because they were at different points of recovery before the 90° RF excitation pulse was applied. The transverse component of magnetization for each vector is therefore different. The transverse component of fat is greater than that of water because its longitudinal component recovers to a greater degree before the next RF pulse is applied, and so more longitudinal magnetization is available to be flipped into the transverse plane. The fat vector therefore generates a higher signal than does water. Fat is hyperintense, and water is relatively hypointense.

Now look at Figure 2.12. If the TR is longer than the T1 recovery times of the tissues, both fat and water fully recover before the next RF excitation pulse is applied. Both vectors are flipped directly into the transverse plane and are not saturated. The magnitude of the transverse component of magnetization for fat and water depends only on their individual proton densities rather than the rate of recovery of their longitudinal components. The flip angle has a significant impact on saturation effects. This is discussed in more detail in Chapter 4.

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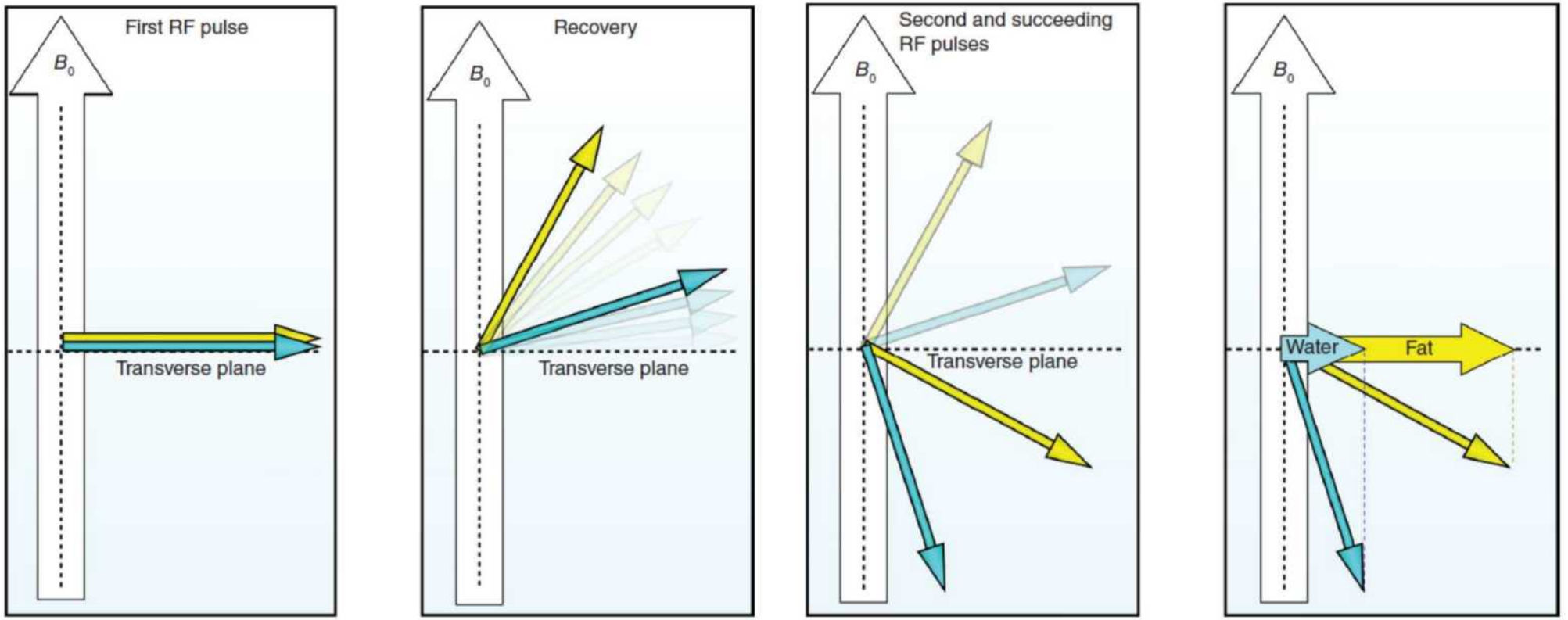
Saturation

Short TR



the first RF excitation pulse, the fat and water vectors are aligned with B0

the next RF excitation pulse flips the vectors beyond 90° and into partial saturation because their recovery was incomplete

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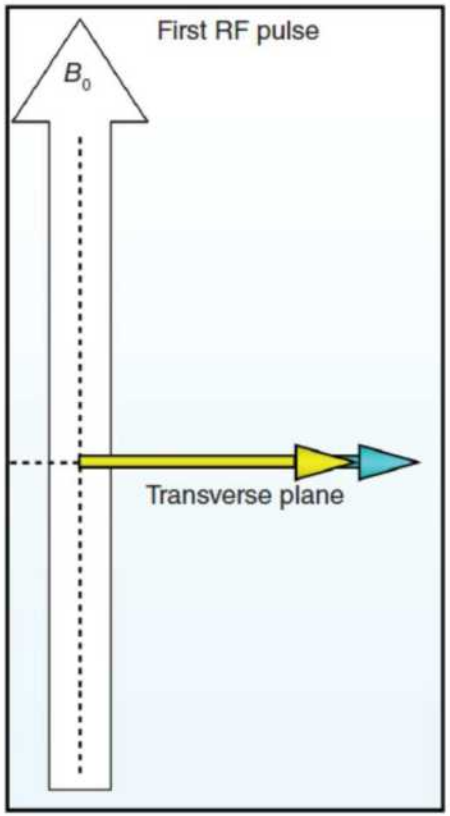
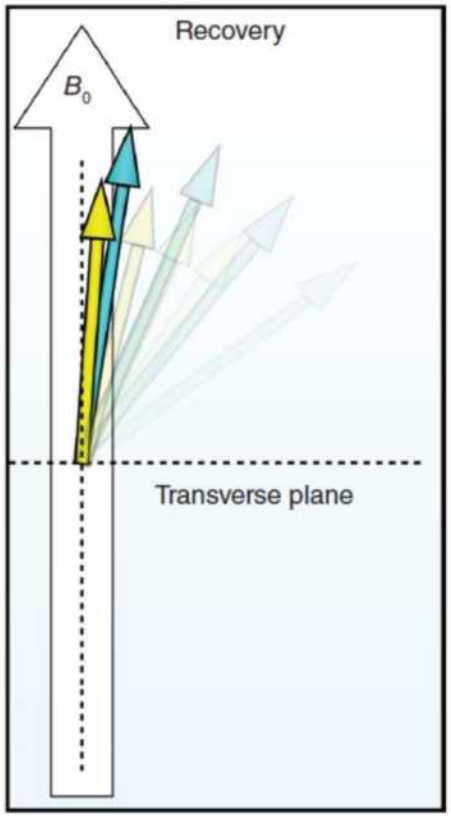
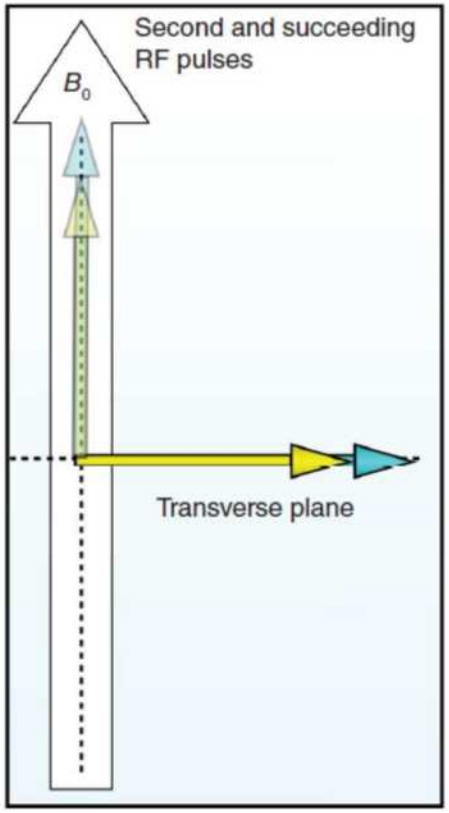
Saturation

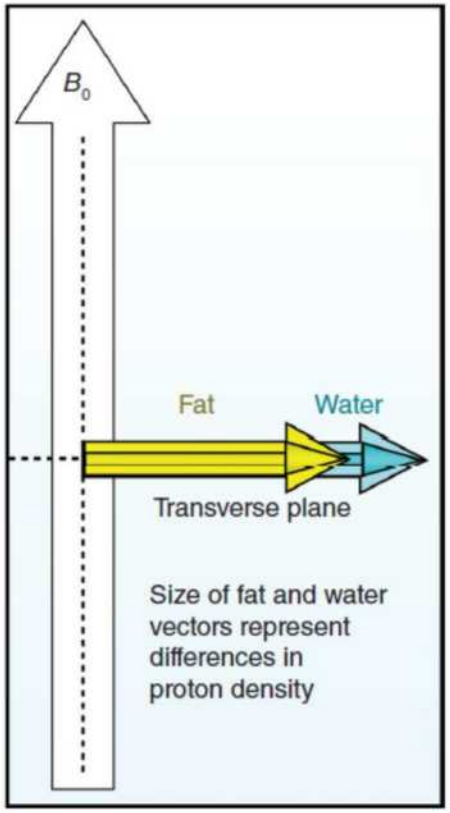
Long TR

. The flip angle has a significant impact on saturation effects;









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T2 contrast

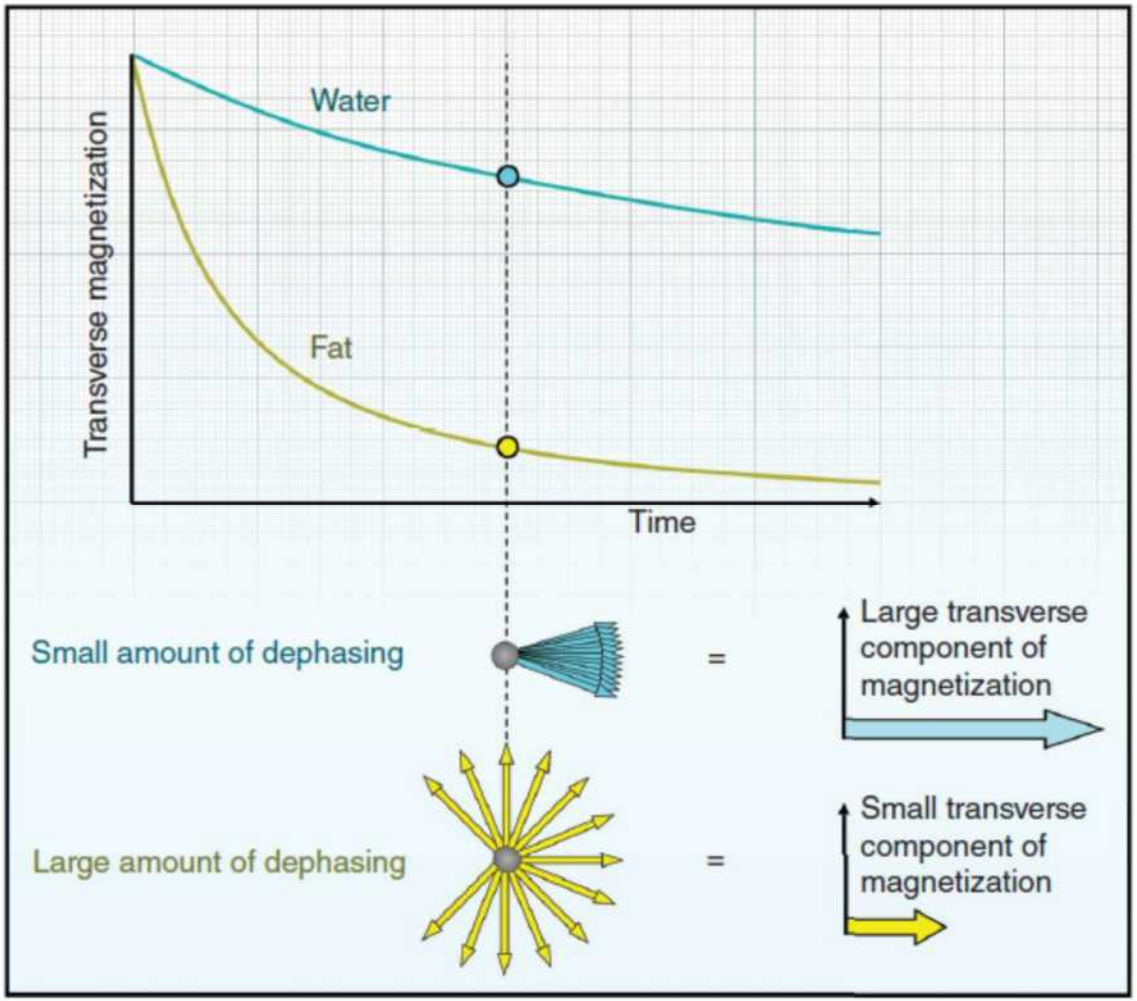

Table 2.5 Things to remember – T2 decay.

Fat has a short T2 decay time
Water has a long T2 decay time
T2 decay is caused by spin–spin relaxation. The efficiency of this process depends on how closely the molecules are packed together
T2 decay times are dependent upon magnetic field strength. As field strength increases, tissues take longer to dephase
T2 contrast is controlled by the TE. For good T2 contrast, the TE must be long

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Proton density contrast

- **Proton density (PD) contrast refers to differences in signal intensity between tissues that are a consequence of their relative number of mobile hydrogen protons per unit volume.**
- **To produce contrast due to differences in the proton densities between the tissues, the transverse component of magnetization must reflect these differences.**
- **Tissues with a high proton density have a large transverse component of magnetization (and therefore a high signal) and are hyperintense.**
- **Tissues with a low proton density have a small transverse component of magnetization (and therefore a low signal) and are relatively hypointense.**

Proton density contrast is always present and depends on the patient and the area under examination. The signal intensity of a tissue therefore depends on its intrinsic contrast properties

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Proton density contrast

Equation 2.4		
$SI = PD\ e^{-TE/T2}(1 - e^{-TR/T1})$	SI is the signal intensity in a tissue PD is proton density TE is the echo time (ms) T2 is the T2 relaxation time of the tissue (ms) TR is the repetition time (ms) T1 is the T1 relaxation time in the tissue (ms)	This equation shows why the signal intensity from a tissue depends on intrinsic and extrinsic contrast parameters. Equation (4.1) shows how this equation is modified in gradient-echo pulse sequences

Learning tip:
Achieving the steady-state

It is clear from the previous learning tip that if the TR is shorter than the T1 relaxation times of the tissues, the first few RF excitation pulses result in different amounts of transverse magnetization. This is because they have recovered different amounts of longitudinal magnetization before the RF excitation pulse was applied. It takes a few TR periods for things to settle down into what is called the **steady-state**. When the steady-state is achieved, vectors recover to the same point and achieve the same amount of longitudinal magnetization during the TR period and they are always flipped to the same point by the 90° RF excitation pulse. They therefore create the same amount of transverse magnetization every TR.

Once the steady-state is achieved, signals (or echoes as they are usually called) are detected by the receiver coil. Before then, signals are not detected because they are different every TR. This is because the amount of transverse magnetization that is created is different. The first few RF excitation pulses are known as **preparatory** or **dummy pulses** because the signals they produce are ignored. Once the longitudinal and transverse vectors of magnetization have settled down and are steady, then these signals are detected and used to create the image. The time it takes to achieve the steady-state depends on B_0, proton density, flip angle, T1 relaxation time, and the duration of the RF excitation pulse [5]. The steady-state is discussed further in Chapter 4 in relation to gradient-echo sequences.

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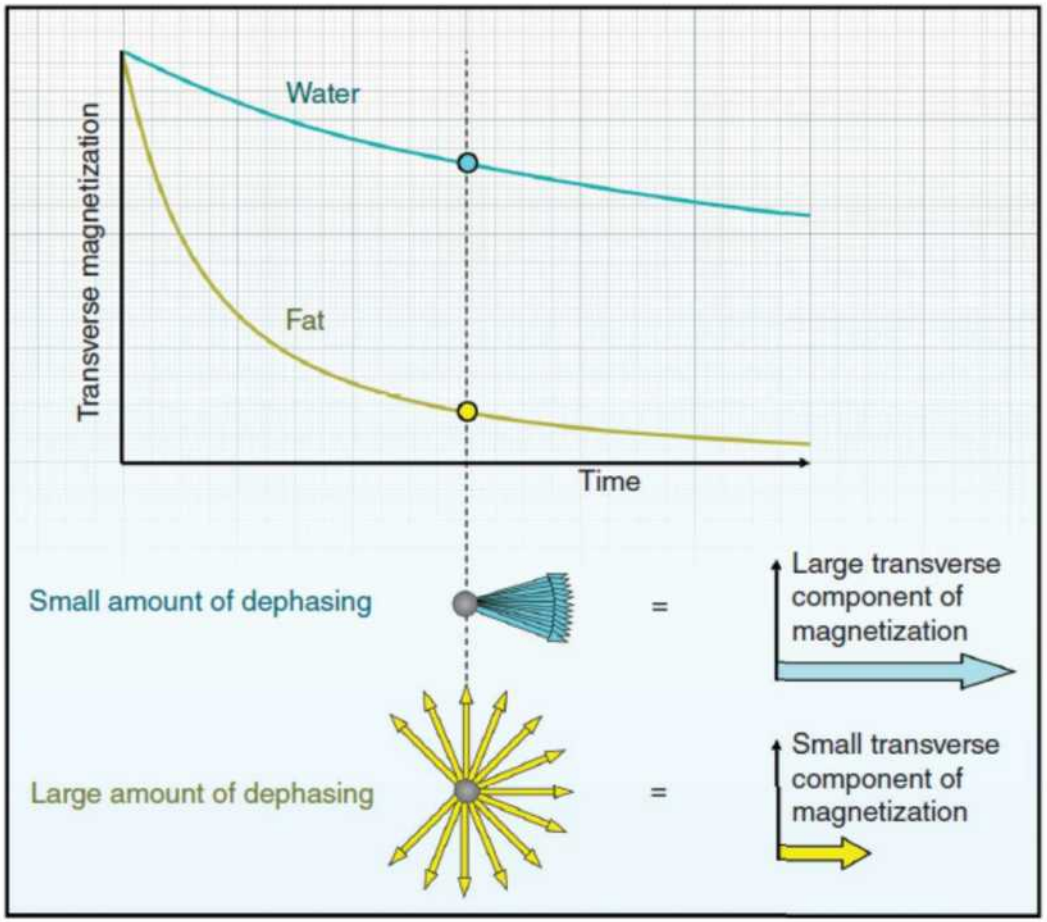
T2 contrast



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T2 decay times are dependent upon magnetic field strength. As field strength increases, tissues take longer to dephase
T2 contrast is controlled by the TE. For good T2 contrast, the TE must be long

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Weighting

- All the intrinsic contrast parameters listed at the beginning of this chapter simultaneously affect image contrast, and therefore, it is possible to obtain images of mixed appearance.

- This means that when looking at an image it is very difficult to determine the relative contribution of each intrinsic contrast parameter to the contrast observed.

Weighting

- To minimize this, extrinsic contrast parameters are selected to weight image contrast toward one of the intrinsic contrast parameters and away from the others.

- This is achieved by applying our knowledge of how extrinsic contrast parameters control the relative contribution from each intrinsic contrast parameter. To demonstrate T1, T2, or proton density weighting, specific values of TR and TE are selected.

- The appropriate selection of these parameters weights an image so that one contrast mechanism dominates the other two.

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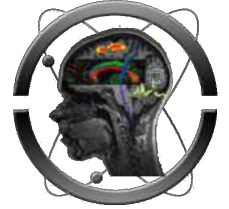
T1 weighting

A T1-weighted image is one where contrast depends predominantly on the differences in the T1 recovery times between fat and water (and all the tissues with intermediate T1 recovery times).

The TR controls how far each vector recovers before the slice is excited by the next RF excitation pulse. To achieve T1 weighting, the TR must be short enough so that neither the vector in fat nor the vector in water has sufficient time to fully return to B0.



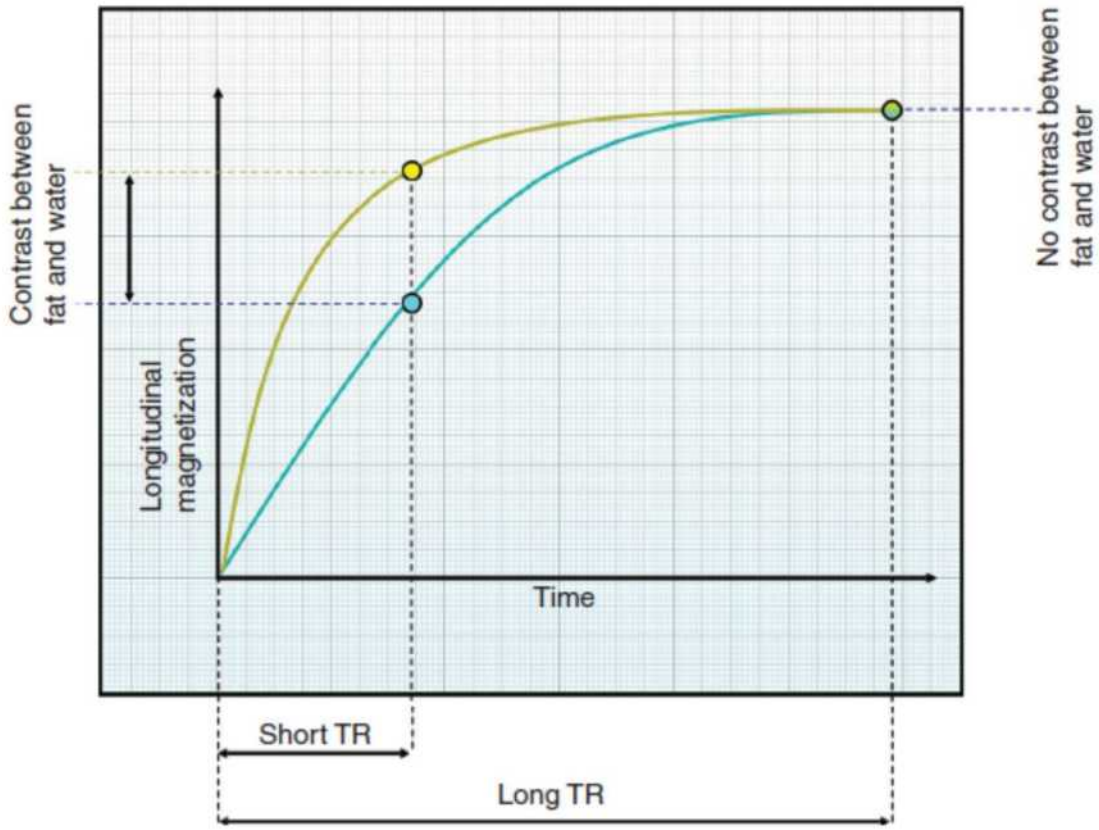
T1 weighting

- If the TR is too long, both the vectors in fat and water return to B0 and fully recover their longitudinal magnetization.

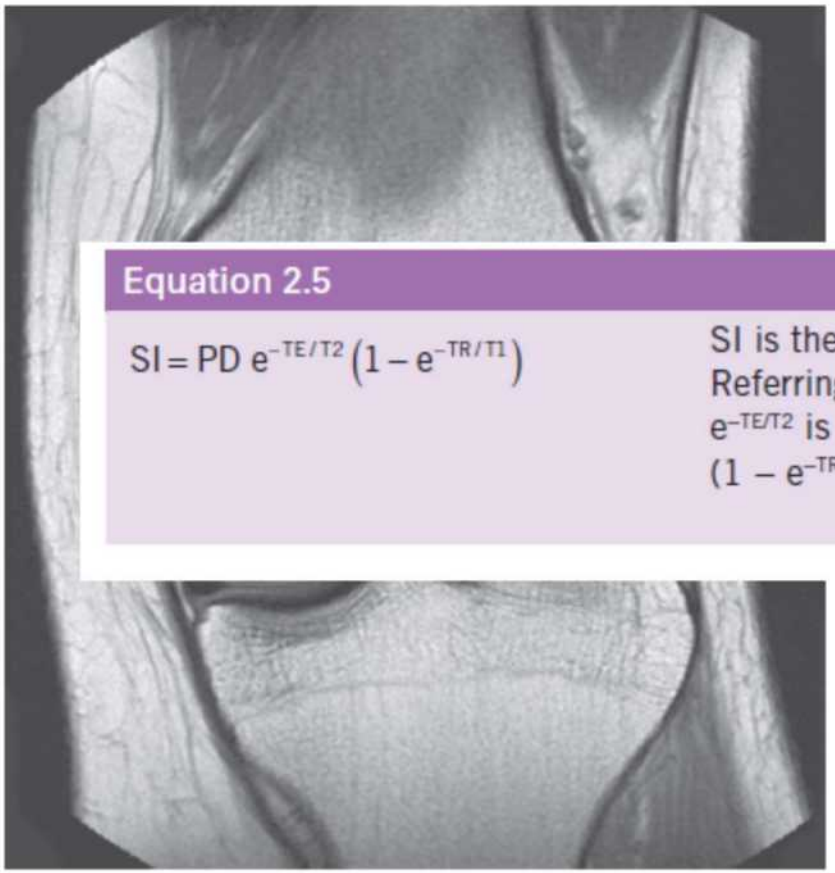
- When this occurs, T1 recovery is complete in both tissues, and the differences in their T1 recovery times are not demonstrated.

- T1-weighted images are used to show anatomy and pathology after administration of a contrast agent.

- TR controls the amount of T1 contrast
- For T1 weighting, the TR must be short and the TE must also be short



T1 weighting



Equation 2.5

$SI = PD\, e^{-TE/T2}\left(1 - e^{-TR/T1}\right)$

SI is the signal intensity in a tissue
Referring to Equation (2.4):
$e^{-TE/T2}$ is the T2 component
$(1 - e^{-TR/T1})$ is the T1 component

This equation shows that if the TE is infinitely short then $e^{-TE/T2} = 1$. Therefore, T2 contrast is minimized and signal intensity depends mainly on PD and T1 contrast

Table 2.6 T1 contrast examples.

High signal	Fat Hemangioma Intraosseous lipoma Radiation change Degeneration Fatty deposition Methemoglobin Cysts with proteinaceous fluid Paramagnetic contrast agents Slow-flowing blood
Low signal	Cortical bone Avascular necrosis Infarction Infection Tumors Sclerosis Cysts Calcification
No signal	Air Fast-flowing blood Tendons Cortical bone Scar tissue Calcification

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T2 weighting

- A T2-weighted image is one where contrast predominantly depends on the differences in the T2 decay times between fat and water (and all the tissues with intermediate T2 decay times).
- The TE controls the amount of T2 decay that occurs before signal is received. To achieve T2 weighting, the TE must be long enough to give the vectors in both fat and water time to dephase. If the TE is too short, neither the vector in fat nor the vector in water has had time to dephase, and, therefore, the differences in their T2 decay times are not demonstrated.
- T2-weighted images are used to image pathology because most pathology has a high water content and is therefore relatively hyperintense on T2-weighted images.

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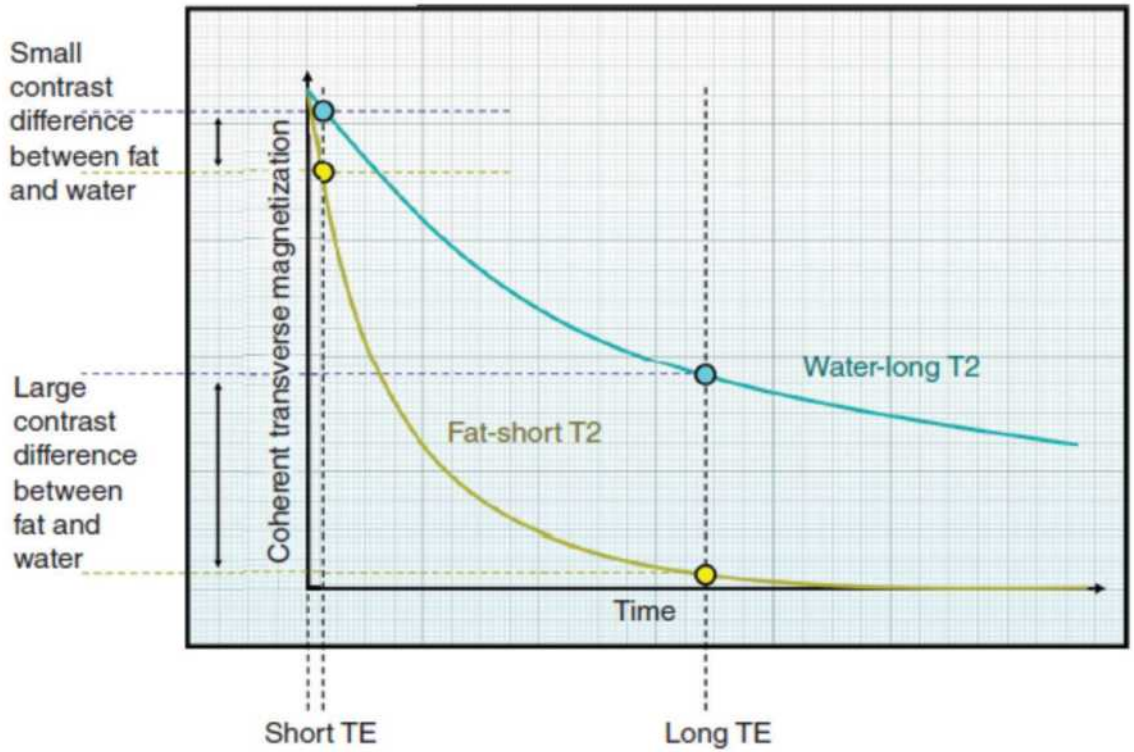
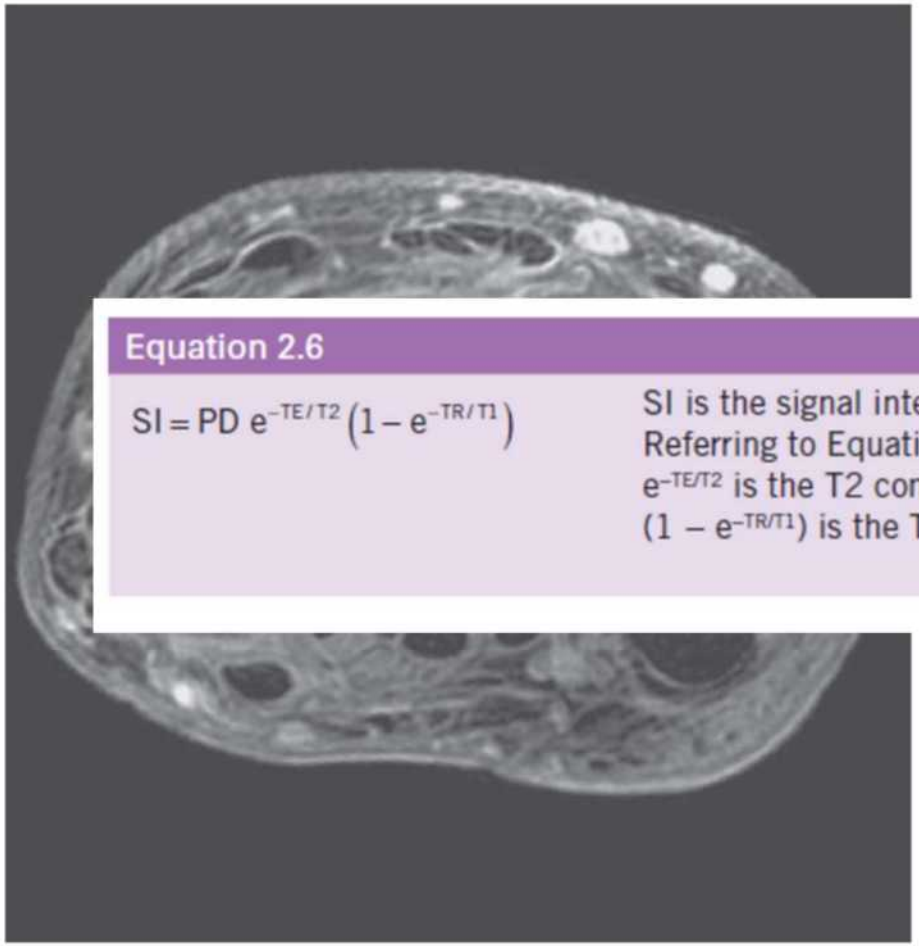
TE controls the amount of T2 contrast

For T2 weighting, the TE must be long and the TR must also be long





Equation 2.6

$SI = PD\, e^{-TE/T2}\left(1 - e^{-TR/T1}\right)$

SI is the signal intensity in a tissue
Referring to Equation (2.4):
$e^{-TE/T2}$ is the T2 component
$(1 - e^{-TR/T1})$ is the T1 component

This equation shows that if the TR is infinitely long, then $(1 - e^{-TR/T1}) = 1$. Therefore, T1 contrast is minimized, and signal intensity mainly depends on PD and T2 contrast

Table 2.8 T2 contrast examples.

High signal	Water Synovial fluid Hemangioma Infection Inflammation Edema Some tumors Hemorrhage Slow-flowing blood Cysts
Low signal	Cortical bone Bone islands Deoxyhemoglobin Hemosiderin Calcification T2 paramagnetic agents
No signal	Air Fast-flowing blood Tendons Cortical bone Scar tissue Calcification

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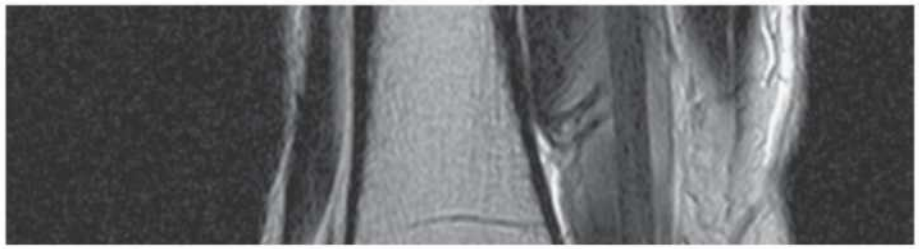
Proton density weighting

- A PD-weighted image is one where differences in the number of mobile hydrogen nuclei per unit volume of tissue are the main determining factor in forming image contrast Table.
- PD weighting is always present to some extent. To achieve PD weighting, the effects of T1 and T2 contrast are diminished so that proton density contrast dominates.
- A long TR allows the vectors in both fat and water to fully recover their longitudinal magnetization and so diminishes T1 contrast. A short TE does not give the vectors in fat or water time to dephase and so diminishes T2 contrast. PD-weighted images are used to image anatomy and pathology.



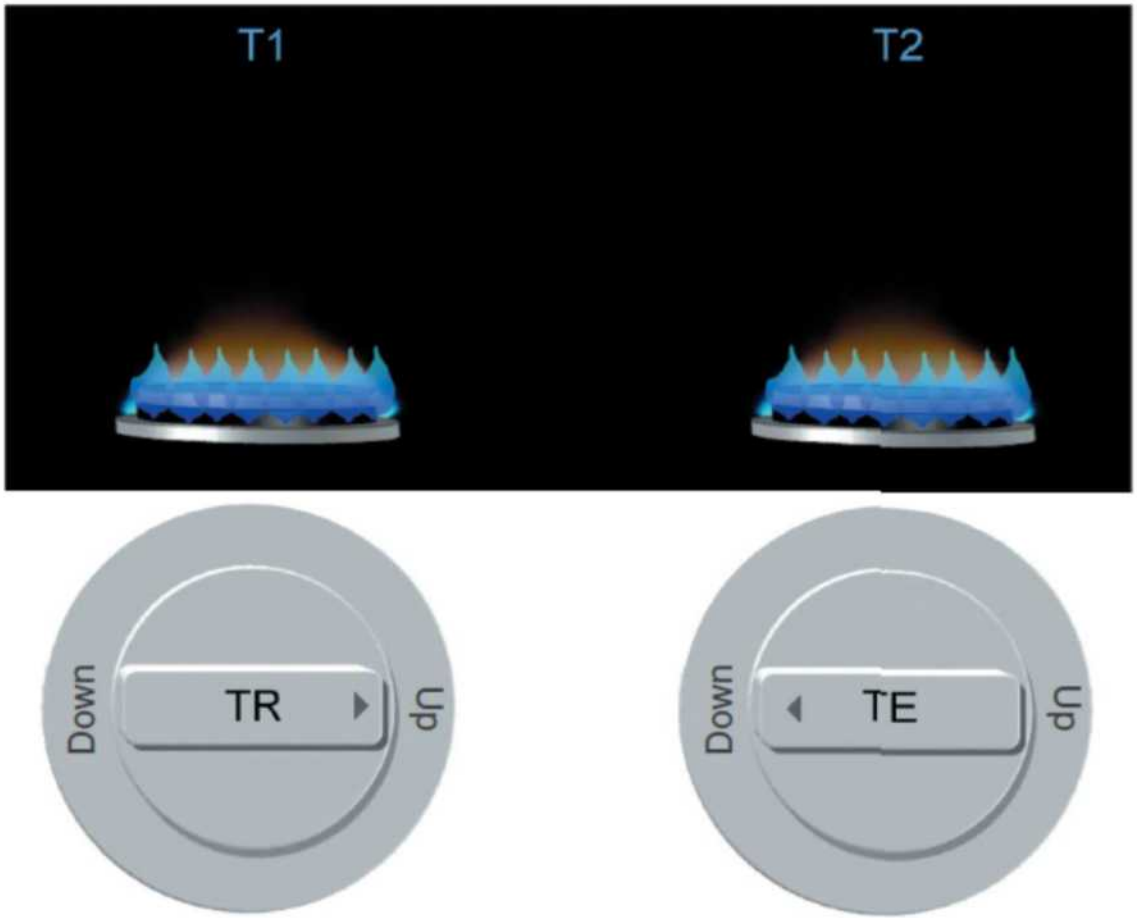
Equation 2.7

$$SI = PD\, e^{-TE/T2}\left(1 - e^{-TR/T1}\right)$$

SI is the signal intensity in a tissue
Referring to Equation (2.4):
$e^{-TE/T2}$ is the T2 component
$(1 - e^{-TR/T1})$ is the T1 component

This equation shows that if the TR is infinitely long, then $(1 - e^{-TR/T1}) = 1$ and if the TE is infinitely short then $e^{-TE/T2} = 1$. Therefore, T1 and T2 contrast are minimized and signal intensity mainly depends on PD

High signal	CSF Synovial fluid Slow-flowing blood Infection Inflammation Edema Cysts Fat
Low or no signal	Air Fast-flowing blood Tendons Cortical bone Scar tissue Calcification


T1
T2
Down
TR
Up
Down
TE
Up

T1 recovery	The recovery of longitudinal magnetization due to spin–lattice relaxation after the RF excitation pulse is switched off
T1 recovery time	The time it takes for 63% of the longitudinal magnetization to recover in a tissue
T1 contrast	An image where fat is hyperintense and water is relatively hypointense because the TR is short enough to not allow full recovery of the vectors
T1 weighting	An image whose contrast is predominantly due to the differences in the T1 recovery times of the tissues
T2 decay	The decay of coherent transverse magnetization due to spin–spin relaxation after the RF excitation pulse is switched off
T2 decay time	The time it takes for 63% of the coherent transverse magnetization to decay in a tissue
T2 contrast	An image where fat is hypointense and water is relatively hyperintense because the TE is long enough to allow full dephasing
T2 weighting	An image whose contrast is predominantly due to the differences in the T2 decay times of the tissues

Other contrast mechanisms

- We have explored the main image contrast mechanisms. These are the types of contrast you are likely to encounter on a day-to-day basis. However, there are other techniques that are used to generate very specific image contrast:

- Diffusion-weighted imaging (DWI)
- Functional MRI (fMRI)
- Magnetization transfer contrast (MTC)
- Susceptibility-weighted imaging (SWI)
- Contrast agents

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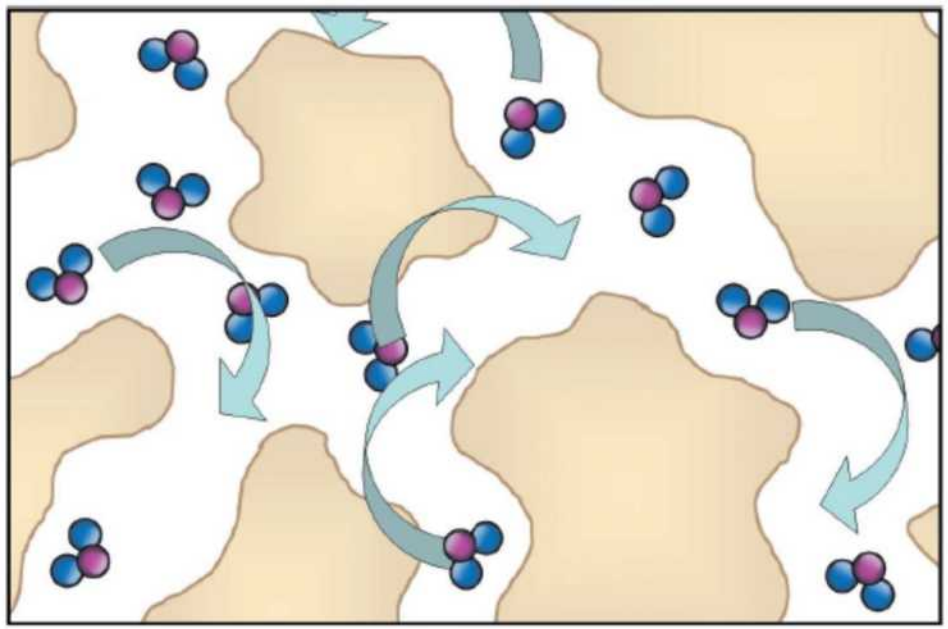
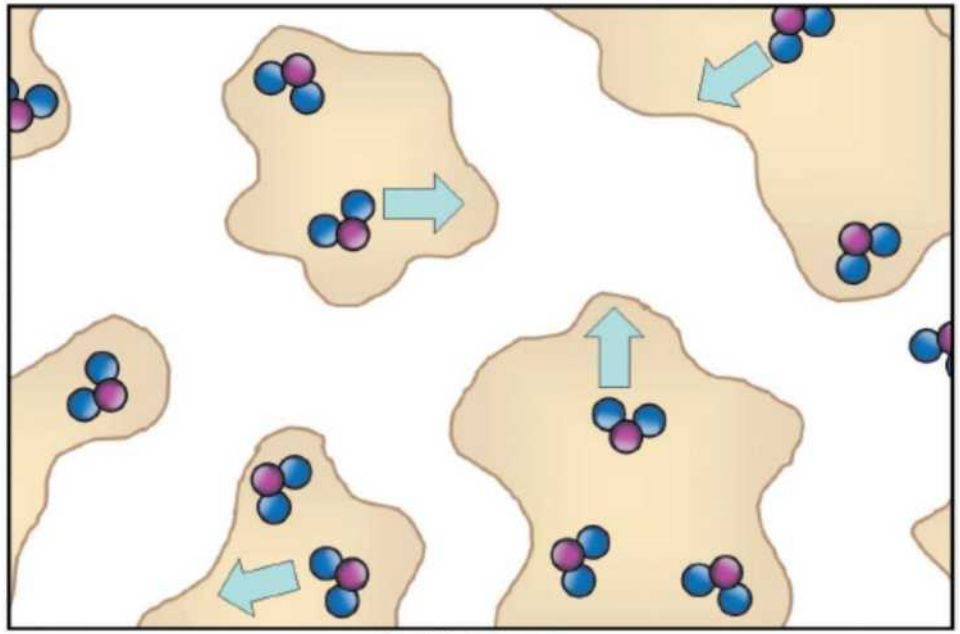
Diffusion-weighted imaging (DWI)



Freely diffusing water



Restricted water

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Diffusion-weighted imaging (DWI)

- **Diffusion** is a term used to describe **the movement of molecules in the extracellular space** due to random thermal motion.

- This **motion** is restricted by boundaries such as ligaments, membranes, and macromolecules.

- Sometimes **restrictions** in diffusion are directional, depending on the structure of the tissues, and diffusion is also restricted in pathology.

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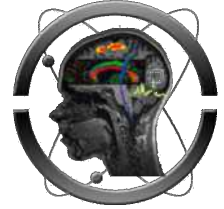
apparent diffusion coefficient

- The net displacement of molecules diffusing across an area of tissue per second is called the apparent diffusion coefficient (ADC), and this is one of the intrinsic contrast parameters.
- It is therefore a parameter that affects image contrast, but it is intrinsic to the tissue and not therefore under our control.
- In areas of restricted diffusion, the ADC is low because the extracellular space is small. Examples of this type of tissue are ligaments, and many types of pathology.
- In areas of free diffusion, the ADC is high because the extracellular space is large. Examples of this type of tissue are normal gray matter and normal liver tissue.



Diffusion Weighted Images (DWI)

- Diffusion Weighted Images (DWI) denotes those whose contrast is determined by the ADC. This is achieved by using gradients.
- In this technique, differences in the ADC are revealed by applying two gradients.
- The first gradient dephases magnetic moments of hydrogen nuclei, and the second gradient attempts to rephase them. In tissues where the ADC is low, the molecules (and therefore the hydrogen nuclei that make them up) are essentially stationary because their diffusion is limited. Magnetic moments of these spins acquire no net phase change after the gradients are applied.

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Diffusion Weighted Images (DWI)

- This is because they do not move between each gradient application. The first gradient dephases the magnetic moments of the hydrogen nuclei, but then the second gradient rephases them.

- As a result, a high signal is obtained from tissues with a low ADC, as the magnetic moments of spins within them are coherent and produce a large component of transverse magnetization.

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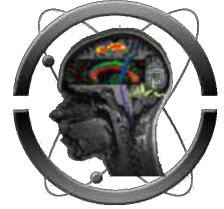
Diffusion Weighted Images (DWI)

- Magnetic moments of moving hydrogen nuclei, however, acquire phase change, and this results in a signal loss.
- This is because the molecules (and therefore the hydrogen nuclei that make them up) diffuse and therefore move between the application of each gradient.
- The first gradient dephases the magnetic moments of the hydrogen nuclei, but then the second gradient cannot rephase them because they move in the meantime.
- In diffusion imaging, normal tissue that exhibits a high ADC has a lower signal intensity than abnormal tissue that has a low ADC, as the molecules within it are free to move.



- ✓ **Diffusion becomes restricted when pathology is present, and so the signal intensity is higher. Signal change depends on the ADC of the tissue and the strength, duration, and interval of the gradients (collectively known as the b factor/value expressed in units of s/mm2).**
- ✓ **In DWI, an extrinsic contrast parameter (b factor) controls how much a tissue's intrinsic contrast parameter (ADC) contributes toward image weighting. As the b factor increases, so does diffusion weighting because the contribution from differences between the ADC of different tissues to image weighting also increases.**

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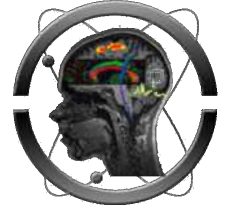
Diffusion-weighted imaging (DWI)

Table 2.13 Typical ADC values in the brain.

	ADC ($\times 10^{-3}$ mm^2/s)	Relative signal when $b = 1000$
Cerebral spinal fluid	2.94	0.05
Gray matter	0.76	0.47
White matter	0.45	0.63



Diffusion-weighted imaging (DWI)

Equation 2.8

$$b = \gamma^2 \times G^2 \times \delta^2 \times (\Delta - \delta/3)$$

b is the *b* value or *b* factor (s/mm²)
γ is the gyromagnetic ratio (MHz/T)
G is the gradient amplitude (mT/m)
δ is the gradient duration (ms)
Δ is the time between two gradient pulses (ms)

The *b* value or *b* factor is a function of the amplitude, duration, and interval of the gradients in the Stejskal–Tanner scheme

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Functional MRI

- **Functional MRI (fMRI)** is a rapid MRI technique that acquires images of the brain during activity or stimulus and then at rest. Image contrast depends on a **physiological process called blood oxygenation level dependent (BOLD).**

- **BOLD** exploits differences in the magnetic susceptibility of oxyhemoglobin and deoxyhemoglobin because of increased cerebral blood flow and little or no increase in local oxygen consumption that occurs during stimulation.

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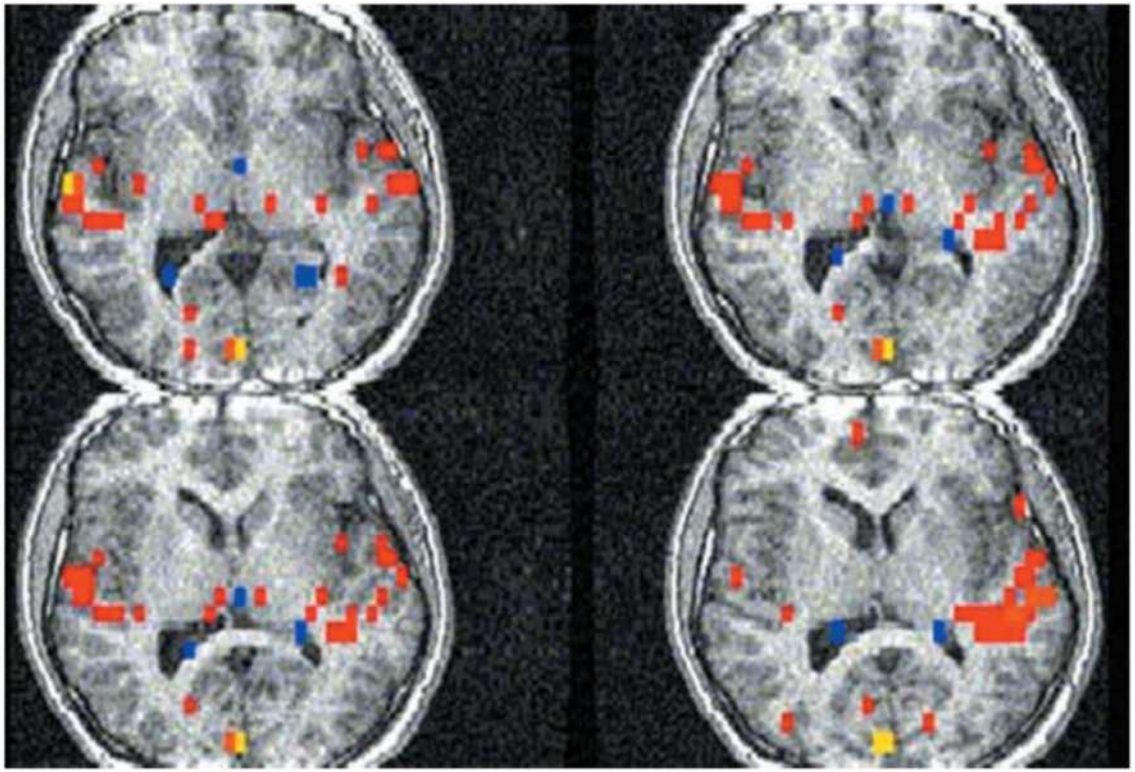
Functional MRI

- Because deoxyhemoglobin is paramagnetic, vessels containing a significant amount of this molecule create local field inhomogeneities causing dephasing and therefore signal loss.
- During activity, blood flow to the cortex increases, causing a reduction of deoxyhemoglobin, which results in a decrease in dephasing and a corresponding increase in signal intensity.
- These effects are very short-lived and therefore require extremely rapid sequences.



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Magnetization transfer contrast

- **Magnetisation Transfer Contrast (MTC)** is another mechanism that generates images with a certain contrast. It involves the fast exchange of energy between **bound and free nuclei**.

- **Bound nuclei are those that are restricted and have a very short T2 decay time**.

- **Their T2 decays times are so short they cannot normally be imaged.**

- **However, bound nuclei reduce the signal intensity of the free nuclei.**

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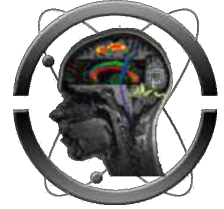

Magnetization transfer contrast

Free nuclei are observable because they have longer T2 decay times.

The magnetic moments of the bound nuclei have a much broader precessional frequency range than the magnetic moments of free nuclei and are therefore excited by an RF excitation pulse that is several kilohertz away from the frequency of the free nuclei.

The energy absorbed by the bound pool of nuclei causes saturation, and magnetization is transferred to the free pool of nuclei.

This causes a reduction in their signal intensity.

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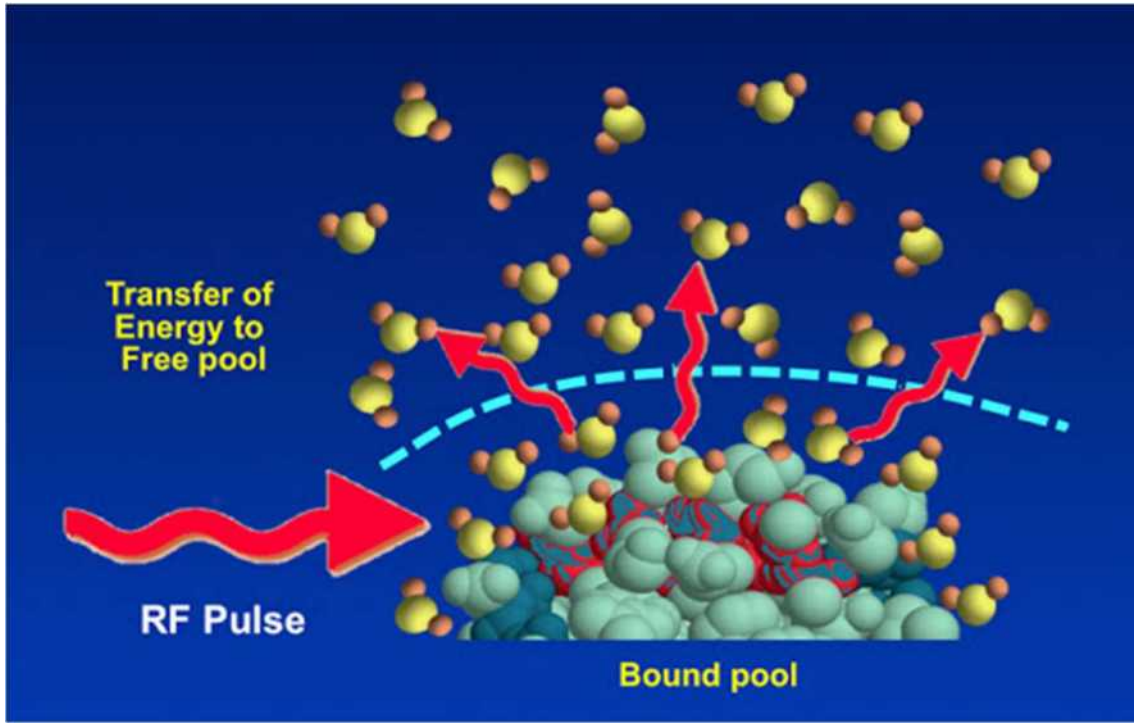
Magnetization transfer



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Susceptibility weighting (SWI)

- SWI uses the magnetic susceptibility differences between tissues to generate image contrast.

- Gradient-echo sequences, in conjunction with a long TE, are commonly used, as they enhance the differences in magnetic susceptibility between tissues

DWI is a technique that sensitizes a spin-echo-type sequence to diffusion motion by using strong gradients

The ADC is an intrinsic contrast parameter and signifies the net displacement of molecules in the extracellular space per second

The *b* value is an extrinsic contrast parameter that controls how much the intrinsic ADC influences image contrast – hence the term diffusion-weighted imaging

Functional imaging techniques are used to image the function or physiology of a system rather than its anatomy

fMRI relies on a process called BOLD to produce a signal in areas of the brain where there is increased activity after performing a function (such as finger tapping)

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Contrast agents

- **Contrast agents** are usually characterized by whether they affect T1 or T2 relaxation times.
- Those that shorten T1 recovery times are called **T1 agents**, and those that shorten T2 decay times are called **T2 agents**.
- **The degree of shortening** depends on **the concentration of the agent**.
- Most agents used in MRI are **T1 agents**.
- The most commonly used agent is **Gadolinium**.
  - Gadolinium (Gd) is a rare-earth metal and, in its natural form, is highly toxic. It cannot be excreted by the body and would cause long-term side effects as it binds to membranes. It is made safe by binding or chelating the gadolinium to other molecules such as diethylene triaminepentaacetic acid (DTPA) (a ligand), which is safely excreted.