Magnetic Resonance Imaging

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Lärandemål 1/3

1. Vad som mäts med MR
2. Varför väte går att avbilda med MR och varför just väte är ett lämpligt ämne att avbilda
3. Vad som händer när en atomkärna med spinn placeras i ett starkt magnetfält och varför
4. Följderna av en förändring i det externa magnetfältet B0
5. Ingående delar i en MR-utrustning samt deras funktion
6. RF-pulsens roll för generering av MR-signal

Lärandemål 2/3

7. MR-signalens uppkomst och utseende samt hur den detekteras
8. Hur kontrast skapas i MR-bilder
9. Hur parameterval i MR-utrustningen samt vävnadsegenskaper påverkar kontrasten i bilden
10. Hur MR-signalen lokaliseras i x, y – och z-led
11. Hur bildkvaliteten kan förbättras i en MR-bild
12. Vad som bestämmer temporal och spatiell upplösning i en MR-bild

Lärandemål 3/3

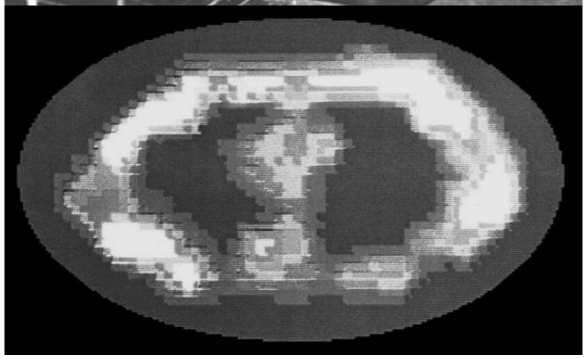
13. Hur spin-echosekvensen fungerar samt översiktligt känna till att det finns andra pulssekvenser
14. Säkerhetaspekter kopplat till MR-avbildning
15. Den kliniska nyttan av MR-avbildning samt redogöra för vilka kliniska situationer som MR är en lämplig avbildningsmetod
16. Vilka artefakter som kan uppstå vid MR-avbildning och hur de uppstår
17. MR-teknikens styrkor och svagheter jämförelse med andra tekniker?

Part I

Introduction

- Non-invasive
- No ionizing radiation
  - Low risk
  - Scan and rescan patients for follow-up
  - Scan healthy volunteers
- Excellent soft-tissue contrast
  - Not attenuation based (like x-ray)
- Versatile
  - Measure blood flow, diffusion, perfusion, cardiac motion, molecular content, metabolism
  - Most technological advances are now based on software

MRI then...



0.07T, Lund 1983

Philips 1.5T, 1985

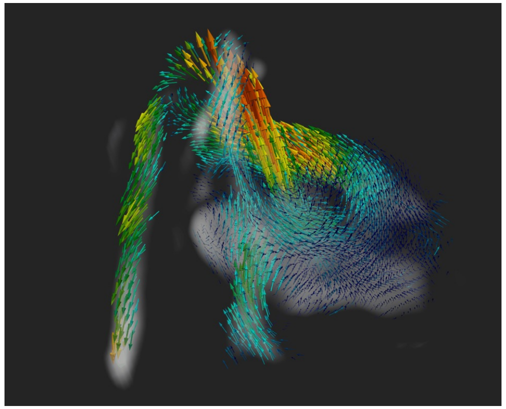
...and now!

Brain at 100 micron resolution

Siemens FREE.Max 0.55T
(University of Michigan 2021)

Siemens Terra.X 7T (2022)

In-utero cardiac 4D flow



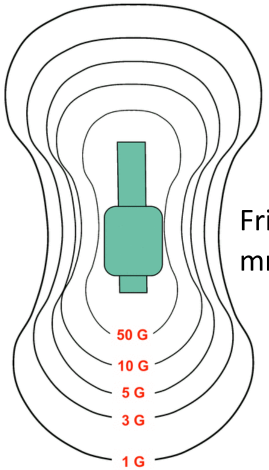
Hyperfine Swoop 64mT

Source	Approximate Magnetic Field Strength
Neuron depolarization (imaged by MEG)	0.5 pT (5×10^{-13} T)
Earth's magnetic field	0.5 G (50 μT)
Refrigerator magnet	50 G (5 mT)
Junkyard electromagnet	1 T
Clinical MRI scanners	0.5 - 3.0 T (typical)
Research MRI scanners (human)	7.0 T – 11.7 T
Laboratory NMR spectrometers	6 - 23 T
Largest pulsed field created in lab nondestructively	97 T
Largest pulsed field created in lab (destroying equipment but not the lab)	730 T

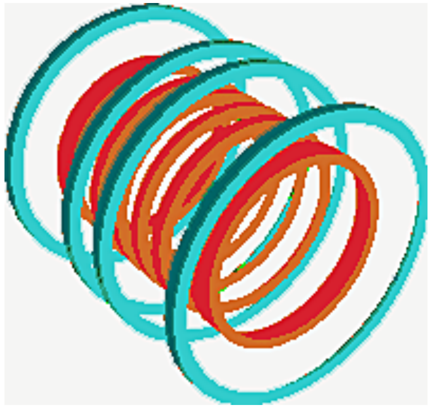
MRI Magnet design

- Niobium–titanium (Nb-Ti) multifilaments in copper matrix
- Super conducting at $T_c < 10$ K and $H_c < 15$ T
- 40 km wire, 500 A current for 1.5 T
- Helium bath
  - Capacity 1400 L, filled to 800 L
  - Boiling point 4.2 K
- No losses due to current, but due to cosmic radiation
  - Boiled off helium is recondensated using a cooling head for “zero boil-off”

Fringe fields

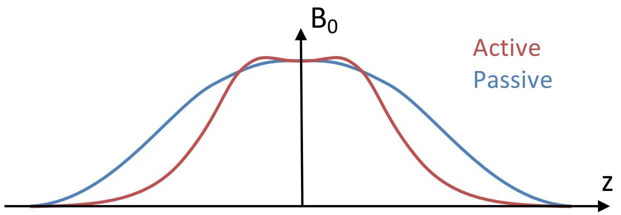


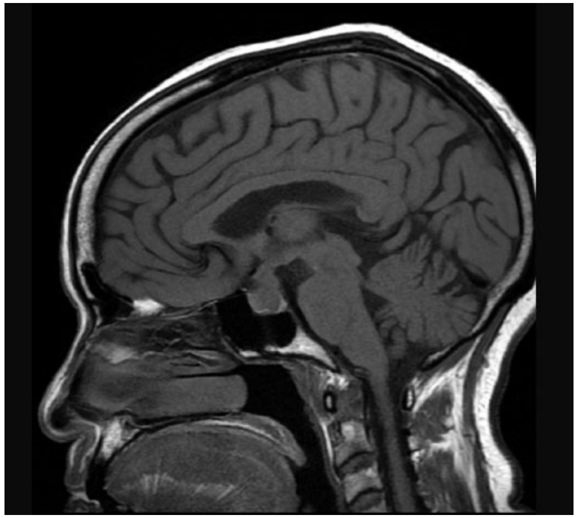
Fringe field (1 G = 0.1 mT)
mri-q.com

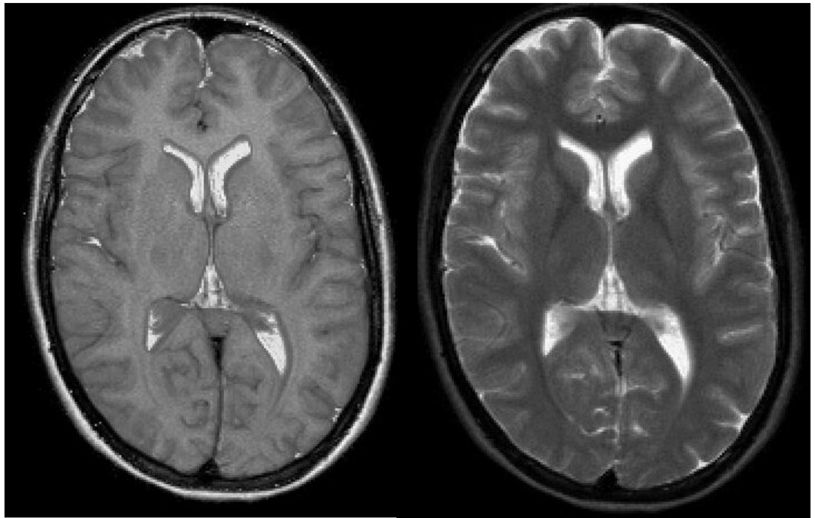


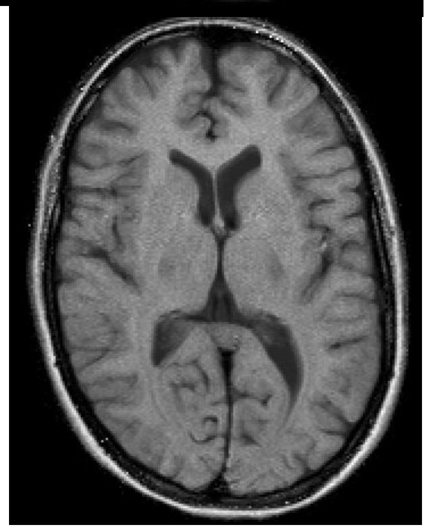
Active shield coils (blue) are in series with the main magnet windings (orange) but carry current in the opposite direction.
mri-q.com

- Actively shielded
  - Opposing field generated in series
  - Reduces fringe field
  - Higher spatial gradients (attractive force)

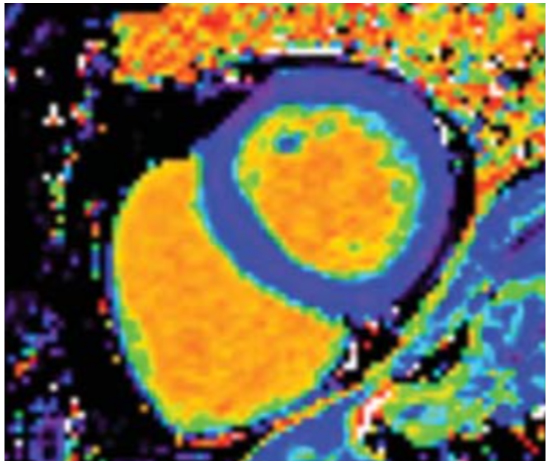




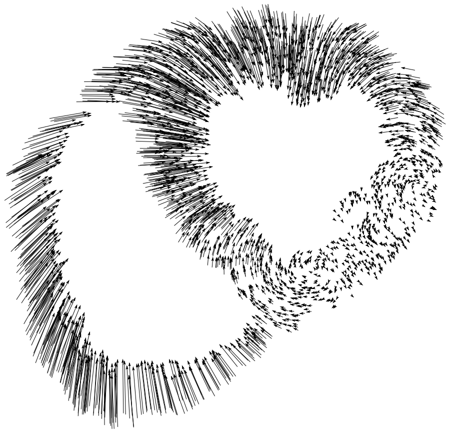




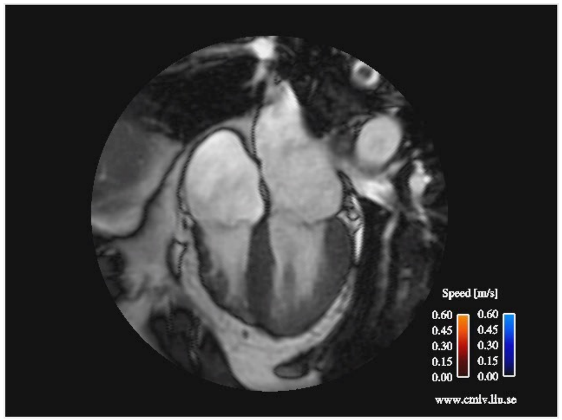
Diffusion tensor imaging



Myocardial tissue characterization



Myocardial motion



Blood flow

Nobel Prize winners

Rabi

- 1945: Isidor Isaac Rabi
  - For his resonance method for recording the magnetic properties of atomic nuclei
- 1952: Felix Bloch and Edward Purcell
  - for their development of new methods for nuclear magnetic precision measurements and discoveries in connection therewith
- 2003: Peter Mansfield and Paul C. Lauterbur
  - for their discoveries concerning magnetic resonance imaging

Bloch and Purcell

Mansfield and Lauterbur

Spin

1.2.5 The Pauli principle

The spin of particles has profound consequences. The *Pauli principle*[1] states:

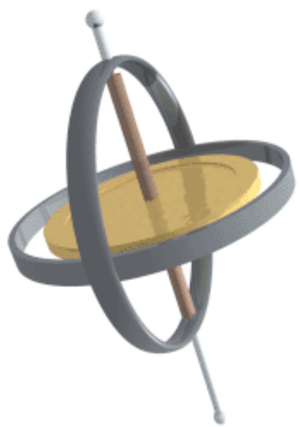
two fermions may not have identical quantum states.

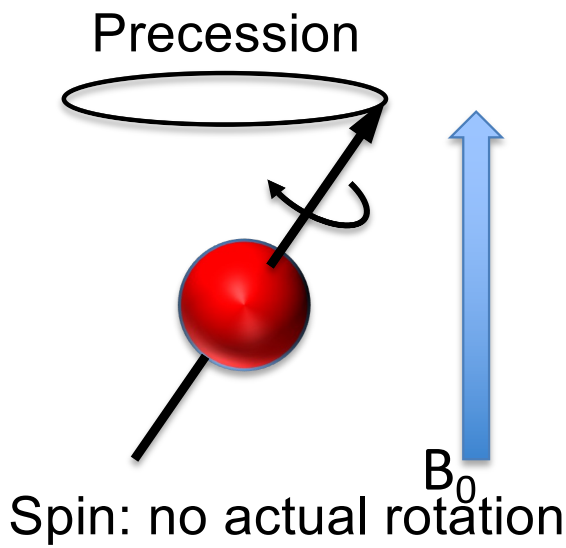
Since the electron is a fermion, this has major consequences for atomic and molecular structure. For example, the periodic system, the stability of the chemical bond, and the conductivity of metals, may all be explained by allowing electrons to fill up available quantum states, at each stage pairing up electrons with opposite spin before proceeding to the next level. This is called the *Aufbau principle* of matter, and is explained in standard textbooks on atomic and molecular structure (see *Further Reading*).

The everyday fact that one's body does not collapse spontaneously into a black hole depends on the spin-1/2 of the electron.

Nuclear magnetic moment and precession

- Atomic nuclei with odd mass number or odd charge number have non-zero spin
  - Magnetic moment, angular momentum
- The magnetic moment precesses in the presence of a magnetic field



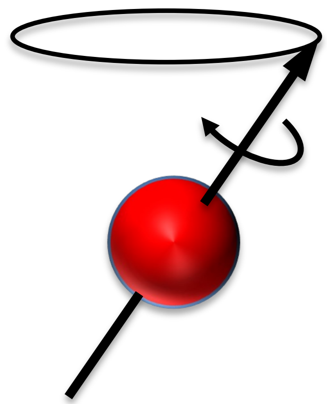


The name of the game

Larmor frequency

- Precession frequency proportional to spin magnetic moment and magnetic field

$$f_0 = \gamma B_0$$



γ	Gyromagnetic ratio
B_0	External magnetic field

^1H at B_0=1.5T:

$$\gamma = 42.58 \text{ MHz / T}$$

$$f_0 = 64 \text{ MHz}$$

Quantization misconception

- Not only two possible states
  - Superposition of eigenstates: $\psi(t) = a(t)\psi_a + b(t)\psi_b$
- Quantization only occurs when measuring a single spin
- MRI always consider many spins, and can keep the continuous, non-quantized representation.

JOURNAL OF APPLIED PHYSICS VOLUME 28, NUMBER 1 JANUARY, 1957

Geometrical Representation of the Schrödinger Equation for Solving Maser Problems

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(Received September 18, 1956)

beam, such that the wave function for any one individual

$$\psi(t)=a(t)\psi_a+b(t)\psi_b$$

du... b eig... single system corresponding to the energies $W+\hbar\omega_0/2$ and $W-\hbar\omega_0/2$ respectively. W is the mean energy of the two levels determined by velocities and internal interactions which remain unchanged. W will be taken as the zero of energy for each system. ω_0 is the resonant angular frequency associated with a transition between the two levels and is always taken positive.

It is usual to solve Schrödinger's equation with some perturbation V for the complex coefficients $a(t)$ and $b(t)$, and from them calculate the physical properties of the system. However, the mathematics is not always transparent and the complex coefficients do not give directly the values of real physical observables. Neither is it sufficient to know only the real magnitudes of a and b, i.e., the level populations and transition probabilities, when coherent processes are involved. We propose instead to take advantage of the fact that the phase of $\psi(t)$ has no influence so that only three real numbers are needed to completely specify $\psi(t)$. We constr... hich have ... 3-vector

$$\begin{aligned} r_1&\equiv ab^*+ba^*\\ r_2&\equiv i(ab^*-ba^*)\\ r_3&\equiv aa^*-bb^*. \end{aligned} \tag{2}$$

(*) al... ime depen... er's equation which gives

$$i\hbar da/dt=a[(\hbar\omega_0/2)+V_{aa}]+bV_{ab} \tag{3}$$

and similar equations for db/dt, da^*/dt, db^*/dt. The subscripts on V indicate the usual matrix elements. $V_{aa}=V_{bb}=0$ for most all cases of interest, and whenever these can be neglected compared to $\hbar\omega_0/2$, V need be neither small nor of short duration for the results to be exact. Using Eqs. (3) to find the differential equation f...

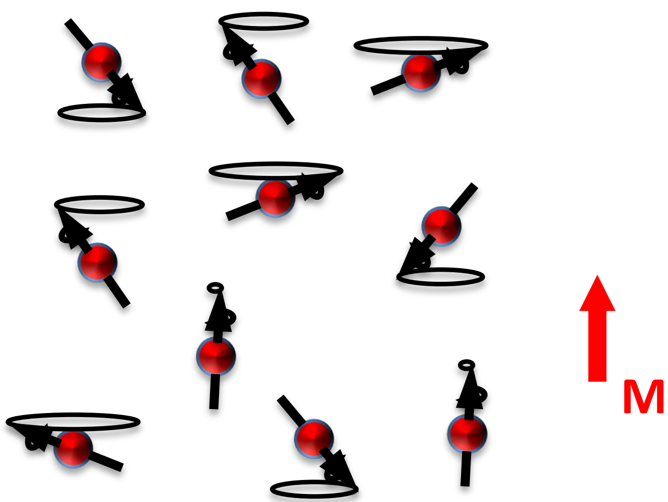
$$d\mathbf{r}/dt=\boldsymbol{\omega}\times\mathbf{r} \tag{4}$$

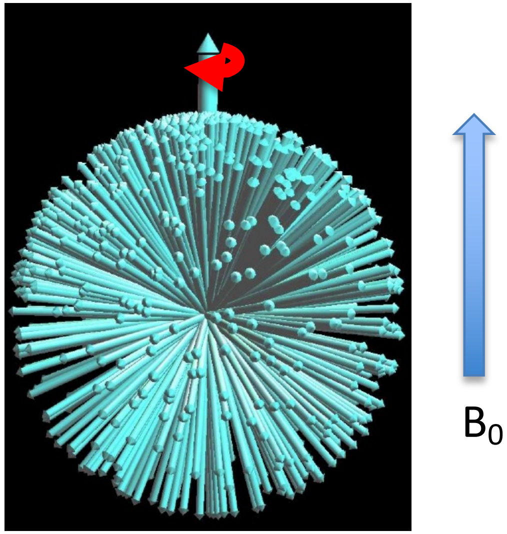
where $\boldsymbol{\omega}$ i... ed by the three real components:

$$\begin{aligned} \omega_1&\equiv(V_{ab}+V_{ba})/\hbar\\ \omega_2&\equiv i(V_{ab}-V_{ba})/\hbar\\ \omega_3&\equiv\omega_0. \end{aligned} \tag{5}$$

Net magnetization

- Many spins result in a net magnetization; M



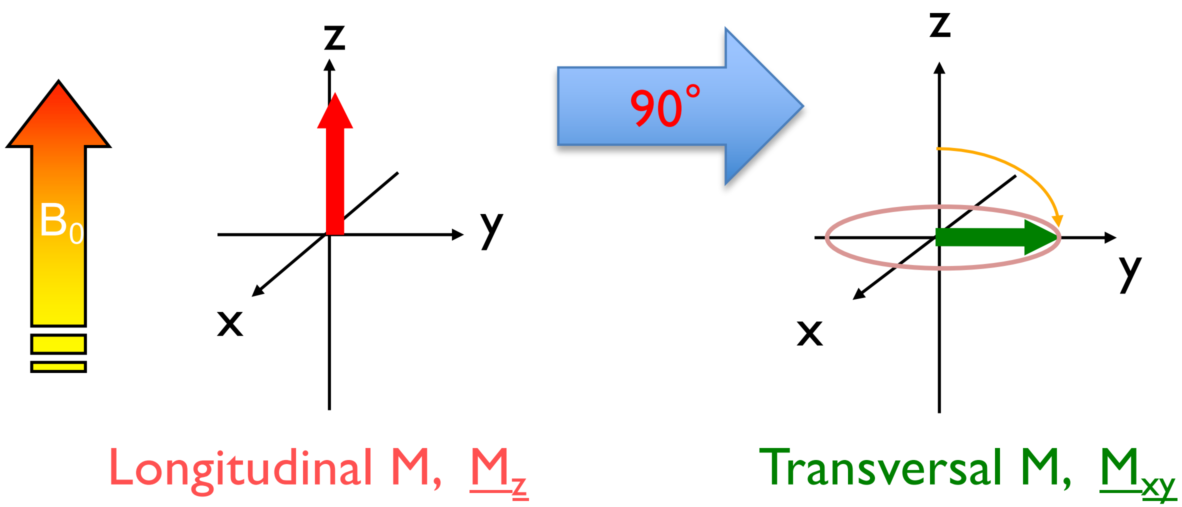


Hanson L, Concepts Mag Reson, 2008

Typical conditions (^1H at B_0 = 1.5 T, 310.15 K):
polarization ~ 0.00001 (100 ppm)

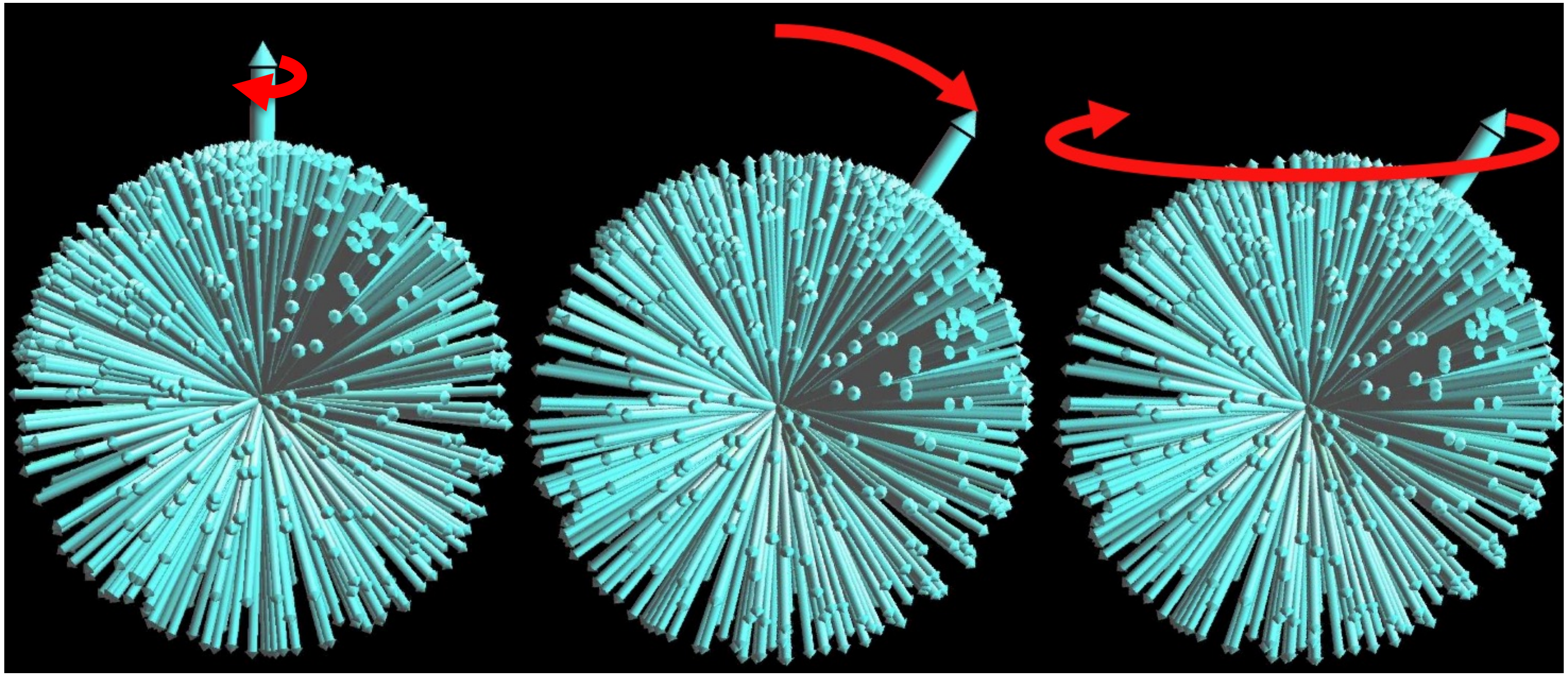
RF Excitation

- Time varying magnetic field, with Larmor frequency
  - 50-150 MHz, "radio frequency band"
  - Not propagating radio wave
- Turns magnetization away from the z axis



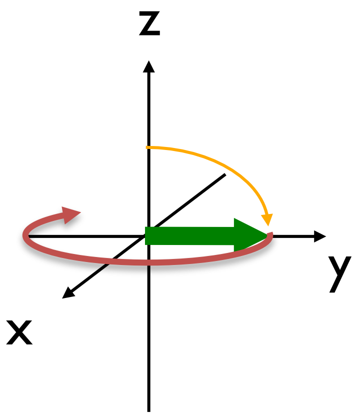
RF excitation and precession

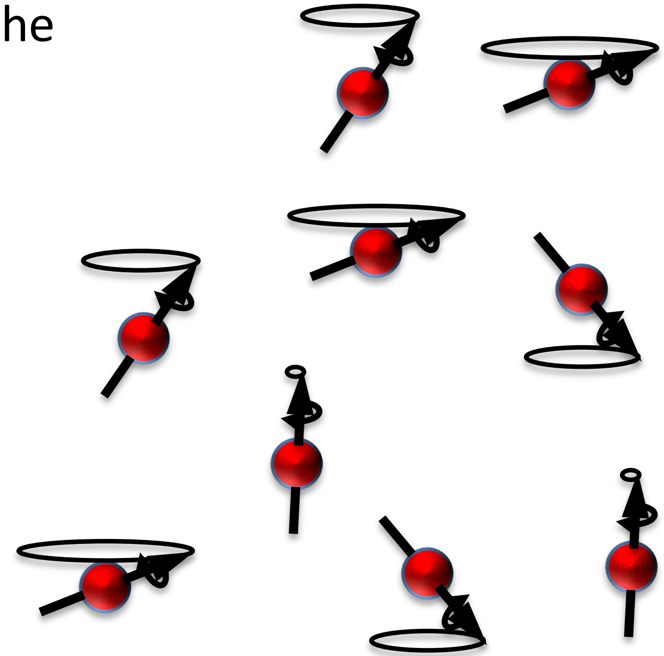
- RF excitation can be used to rotate magnetic moments
  - Net magnetization rotates and precesses



Net magnetization precession

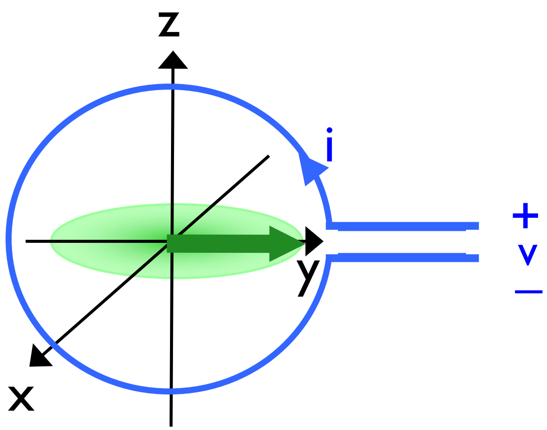
- After excitation, the net magnetization also precesses around the external magnetic field
    - Precession frequency is the Larmor frequency

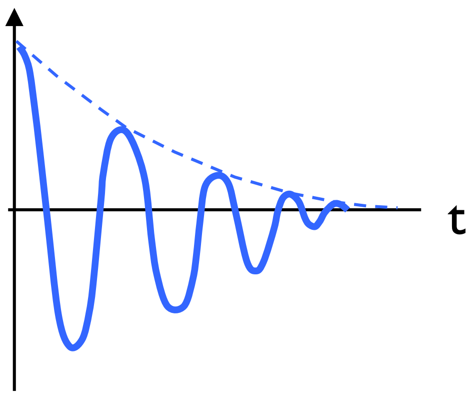




Signal reception

- Time varying current -> creates a magnetic field
- Time varying magnetic field -> induces a current
- Rotating M_{xy} induces a current in a conducting loop



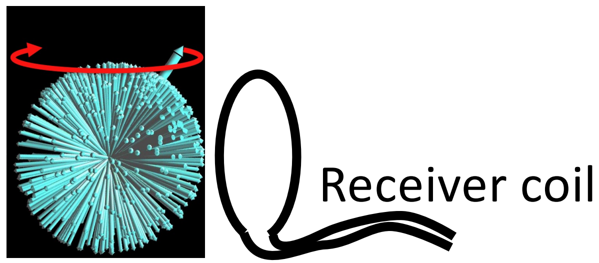


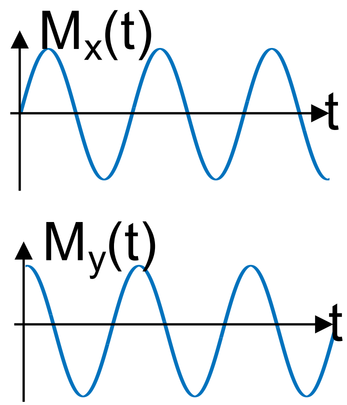
Signal reception

- Faraday's law of induction

$$V(t) = -\frac{\partial \Phi}{\partial t}$$

Φ - Magnetic flux through coil
$V(t)$ - Induced voltage in coil

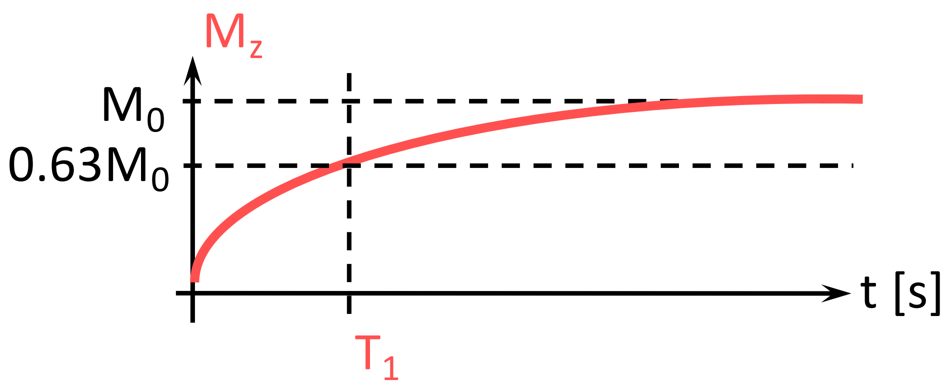




- Rotating M in xy plane induces a current in a conducting loop
- Signal is complex valued – proportional to $M_x + iM_y$

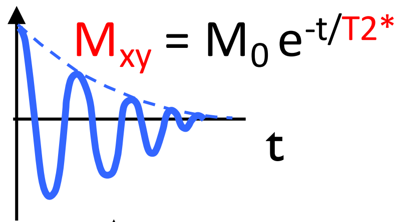
T_1 relaxation

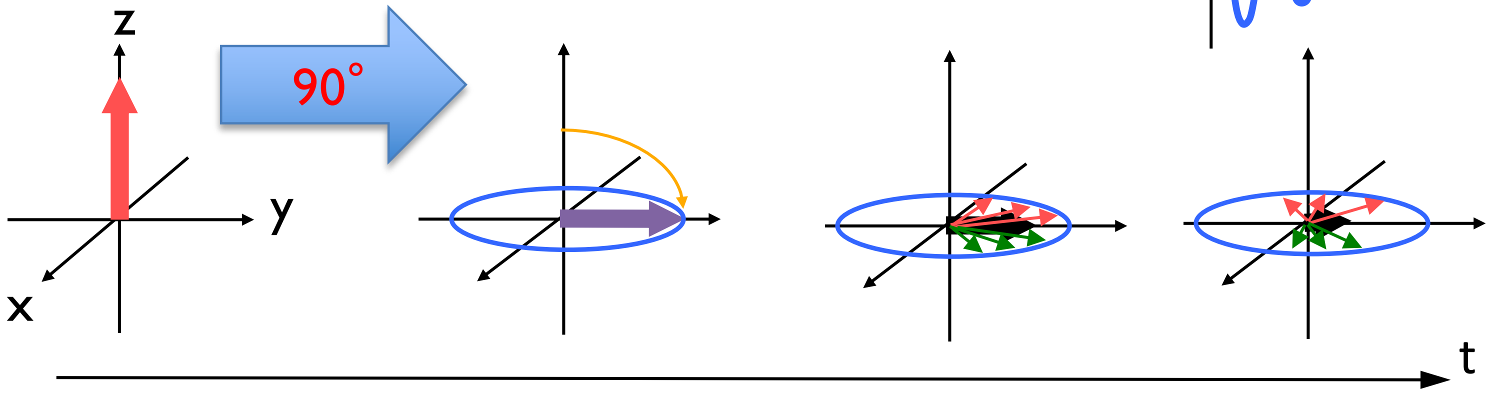
- After excitation, the system returns to equilibrium
- M_z returns exponentially
  - $M_z = M_0\,(1-e^{-t/T1})$
- Time constant: T_1
- Tissue dependent, typically in the order of 1 s for blood at 1.5 T



T_2^* relaxation

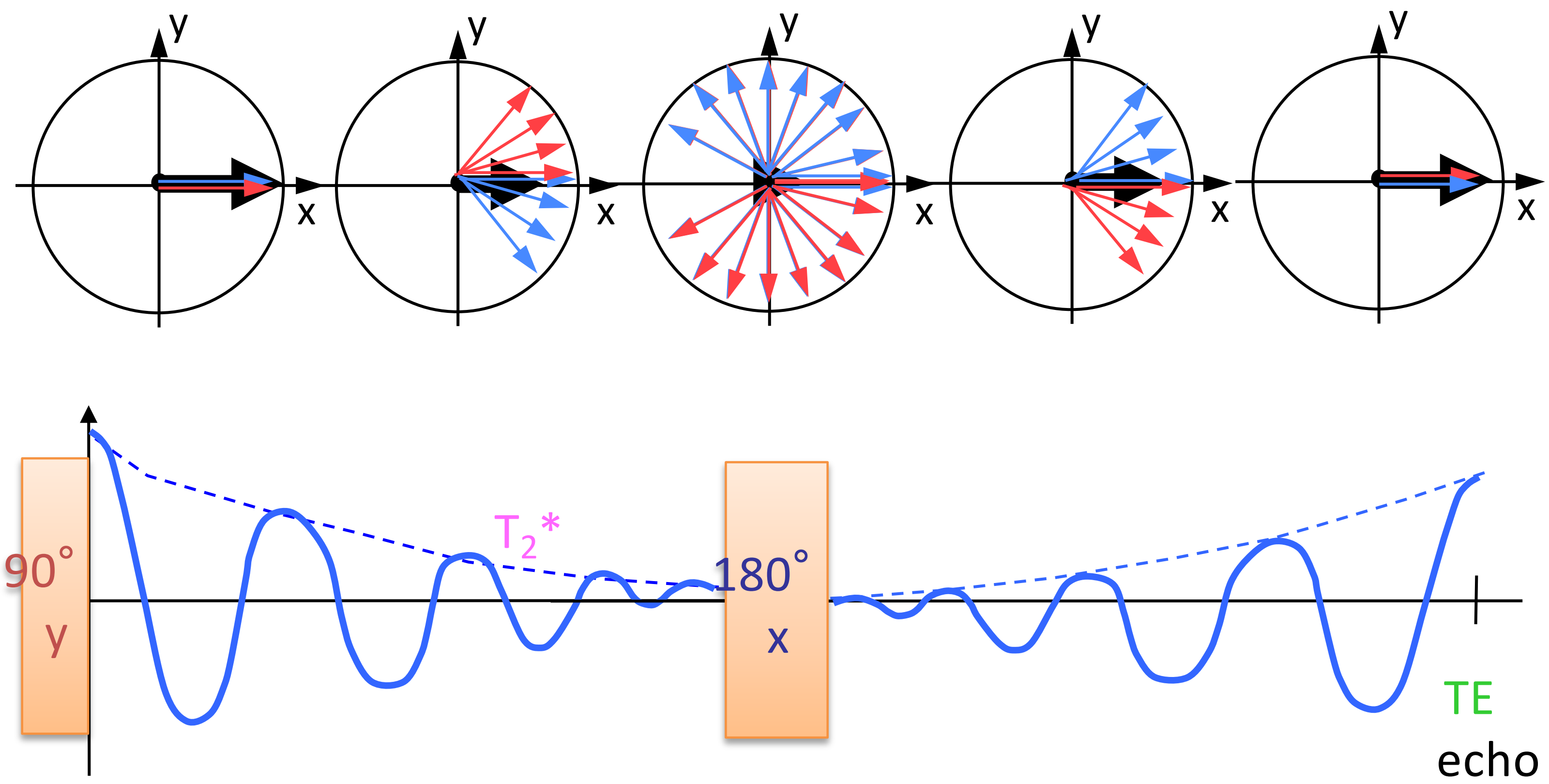
- Local variations of the magnetic field
  - Small magnetic field from surrounding electrons and nuclei
  - Non-homogeneous external magnetic field
- Varying magnetic field -> varying Larmor (precession) frequency
- Net magnetization “dephases”
- Exponential, time constant T_2^*, tissue dependent





The spin-echo pulse sequence

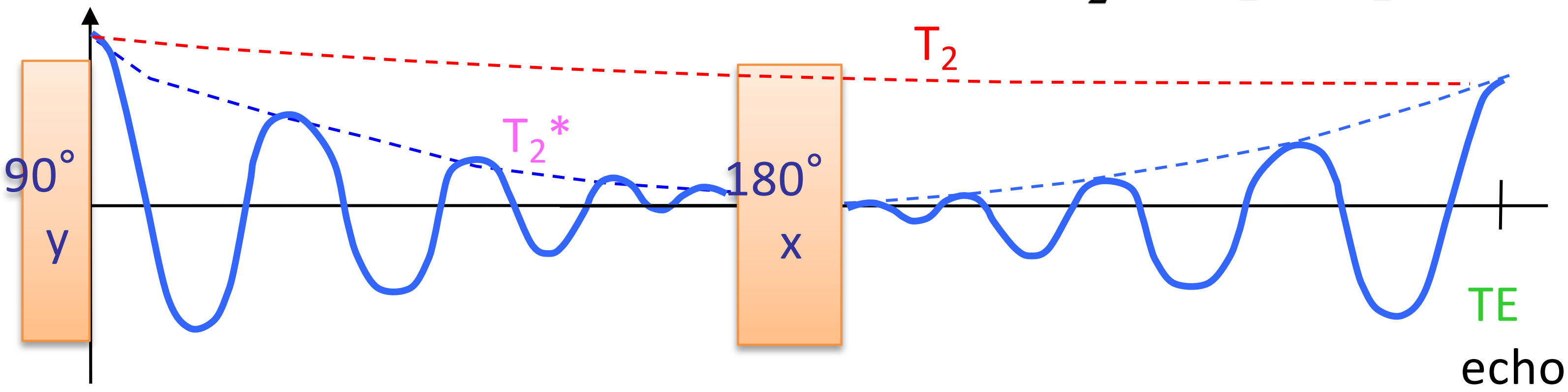
- Dephasing due to time-constant local variations can be refocused using an 180° pulse



T_2 and T_2^* relaxation

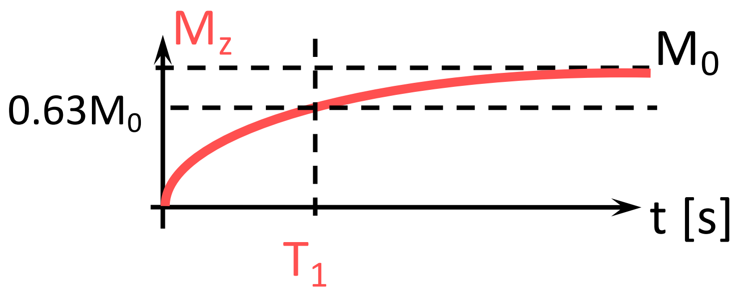
- Time-constant magnetic field variations can be refocused
  - Magnetic field inhomogeneities
- Time-varying variations cannot
  - Small magnetic field from surrounding electrons and nuclei
- Spin-echo pulse sequence not always applicable (no refocusing possible)
  - T_2^* is *apparent* T_2

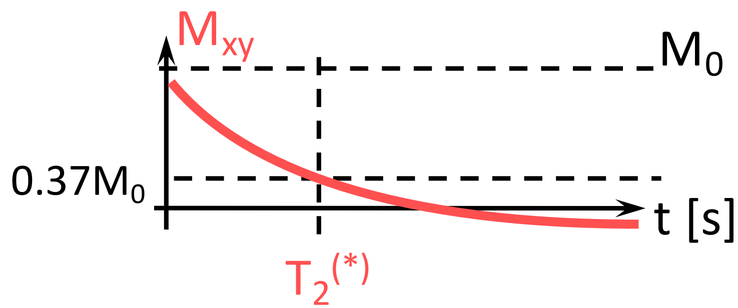
$$\frac{1}{T_2^*} = \frac{1}{T_2} + \frac{1}{T_2'}$$



Relaxation summary

- T_1: Longitudinal relaxation
    - Recovery of M_z to equilibrium
- T_2: Transverse relaxation
    - Dynamic magnetic field variations
    - Reduction of M_{xy} magnitude due to dephasing
    - Unavoidable
- T_2^*: Apparent transverse relaxation
    - Static *and* Dynamic magnetic field variations
    - If static variations are note refocused, it manifests as apparent T_2
- $T_2^* < T_2 < T_1$





T_1 and T_2 values at 1.5 T

Tissue type	T_1	T_2
CSF	2500 ms	1000 ms
White brain matter	780 ms	90 ms
Gray brain matter	920 ms	100 ms
Blood	1200 ms	360 ms
Myocardium	880 ms	75 ms
Fat	260 ms	110 ms

Spin-echo pulse sequence

TE: Echo time
TR: Repetition time

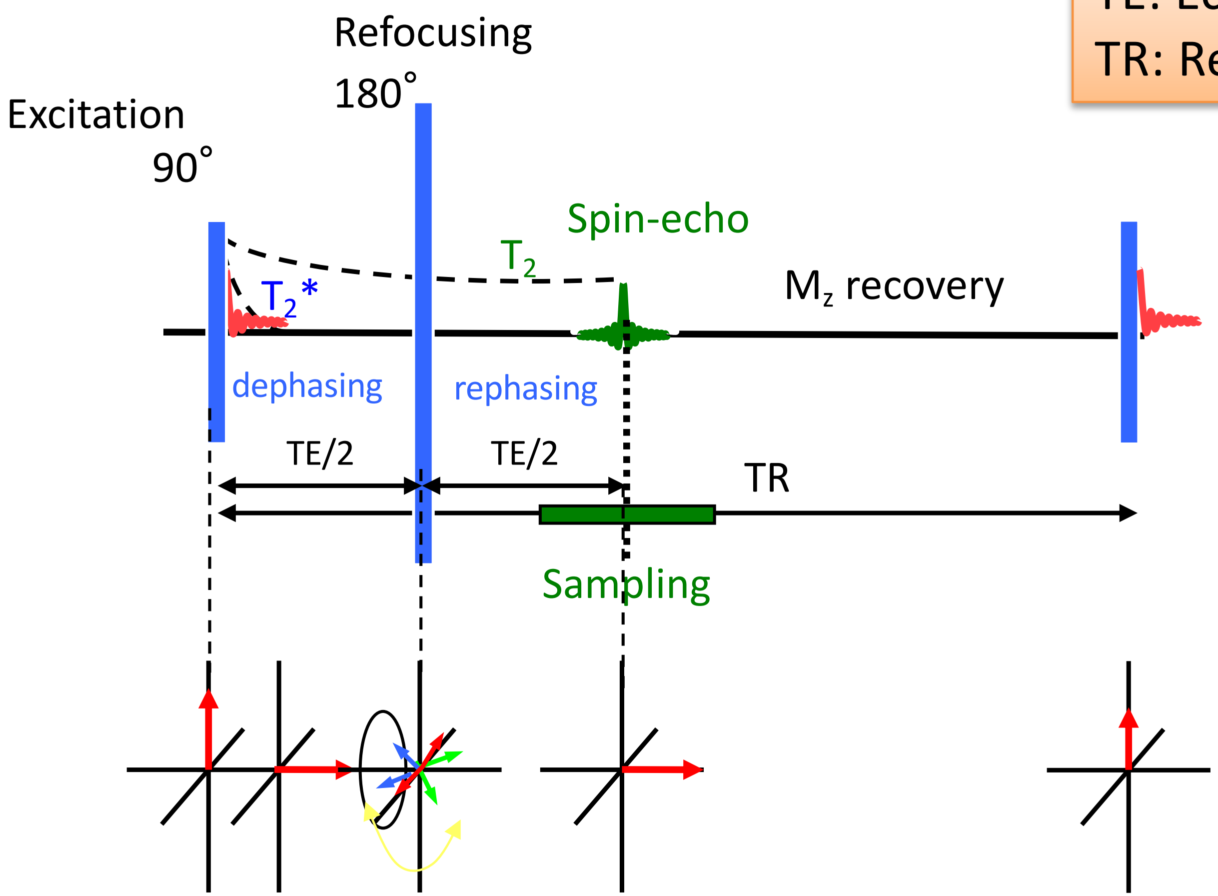
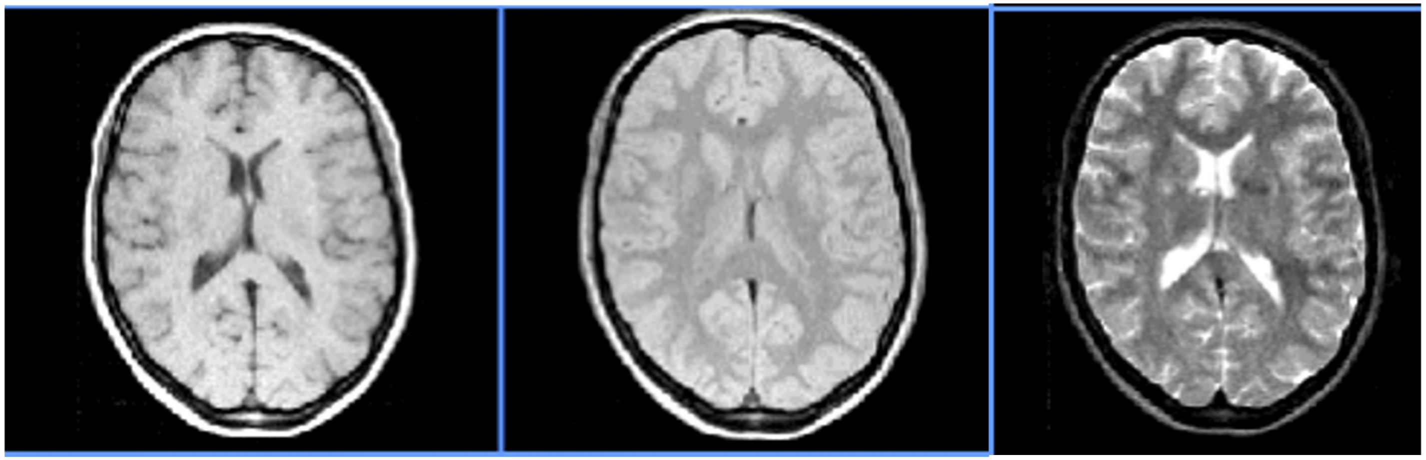
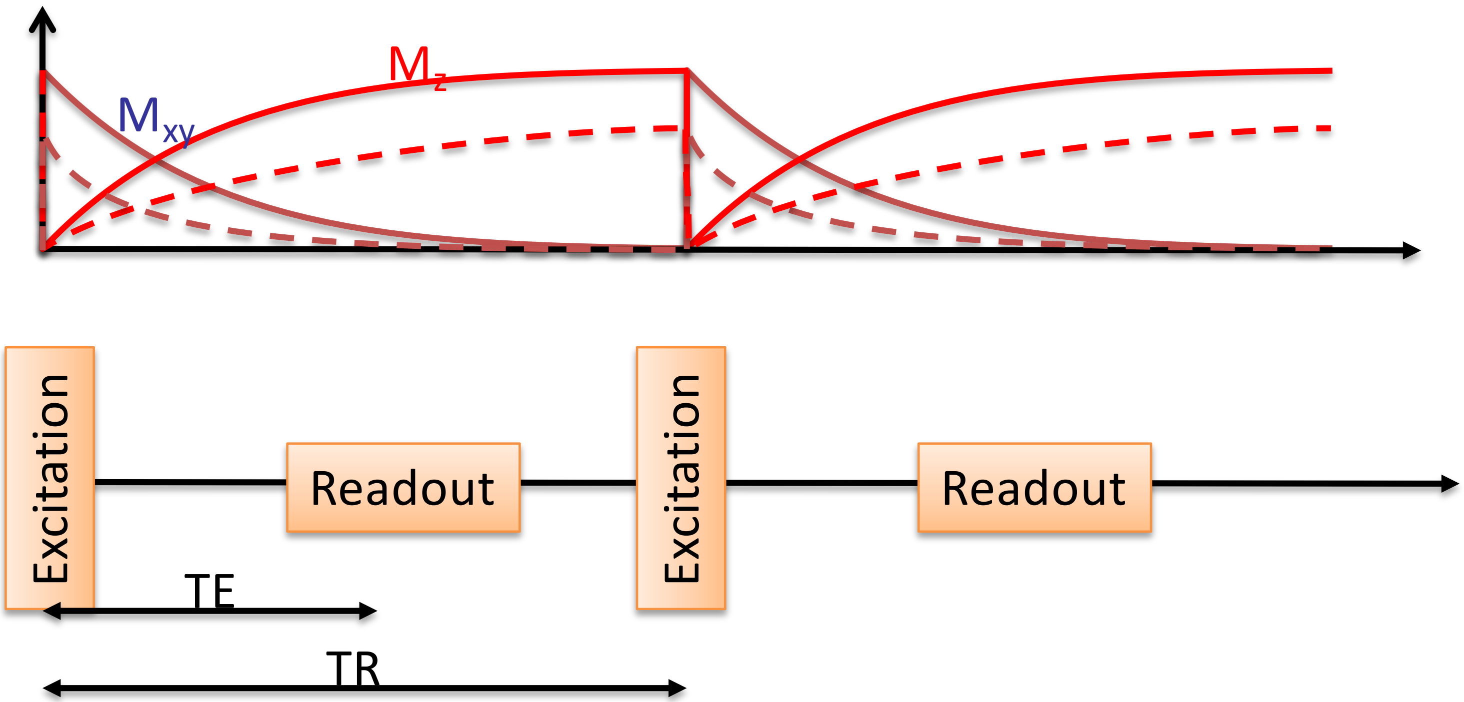


Image contrast

- By modifying pulse sequence timing (TE and TR), different image contrasts can be obtained
  - Tissue dependent T_1
  - Tissue dependent T_2

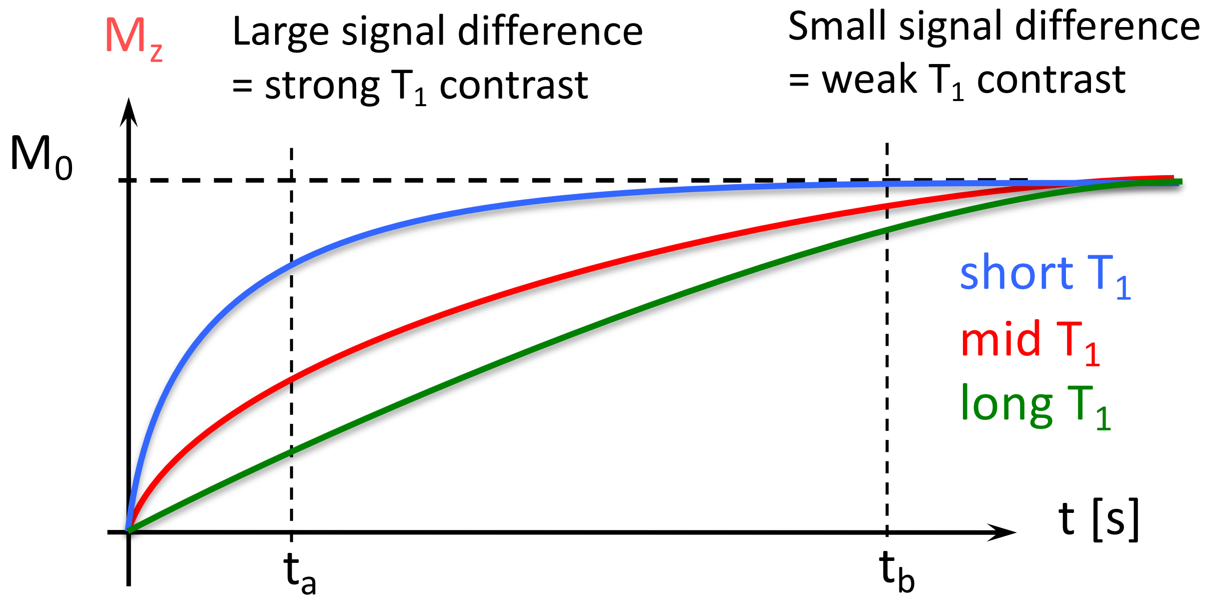




By adjusting TE and TR, signal magnitude will vary depending on T_1 and T_2

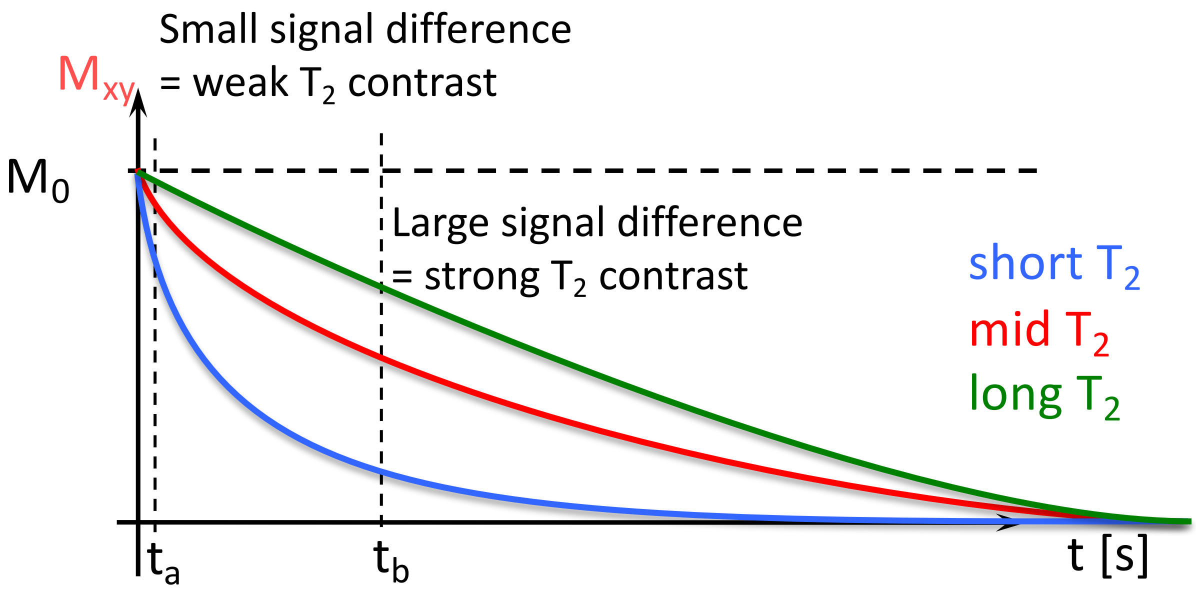
T_1 contrast

- Different tissues have different T_1 relaxation times
- T_1 contrast depend on sequence timing
  - Time after excitation (how much signal is *recovered*)



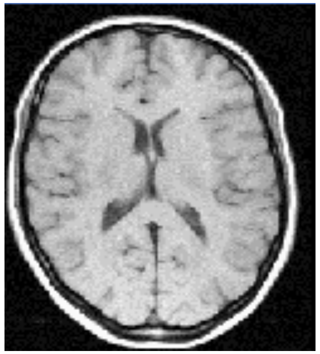
T_2 contrast

- Different tissues have different T_2 relaxation times
- T_2 contrast depend on sequence timing
  - Time after excitation (how much signal has *dephased*)

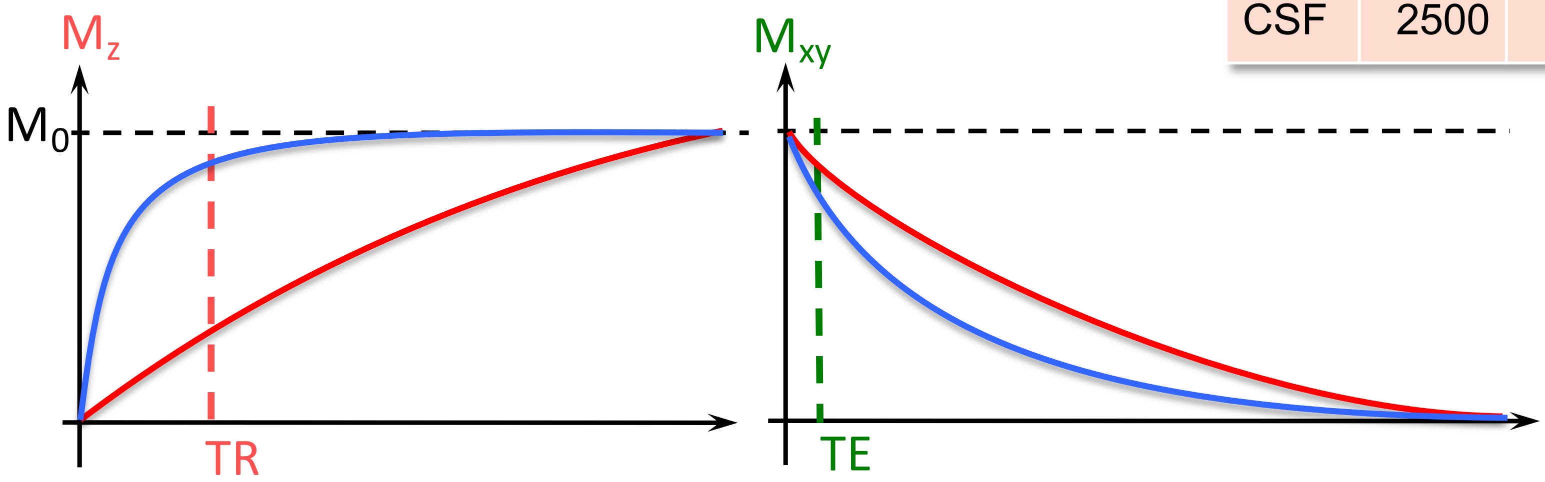


T_1 weighting

- Short TR: maximizes T_1 contrast
- Short TE: minimizes T_2 contrast
- TE cannot be made infinitely short

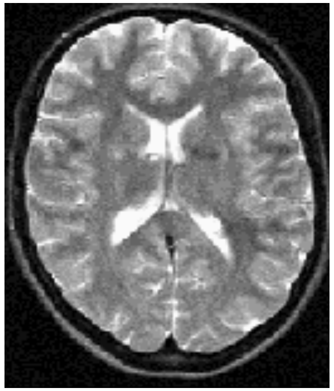


	T_1	T_2
WM	780	90
GM	920	100
CSF	2500	1000



T_2 weighting

- Long TR: minimizes T_1 contrast
- Long TE: maximizes T_2 contrast
- Too long TE and all signal is lost



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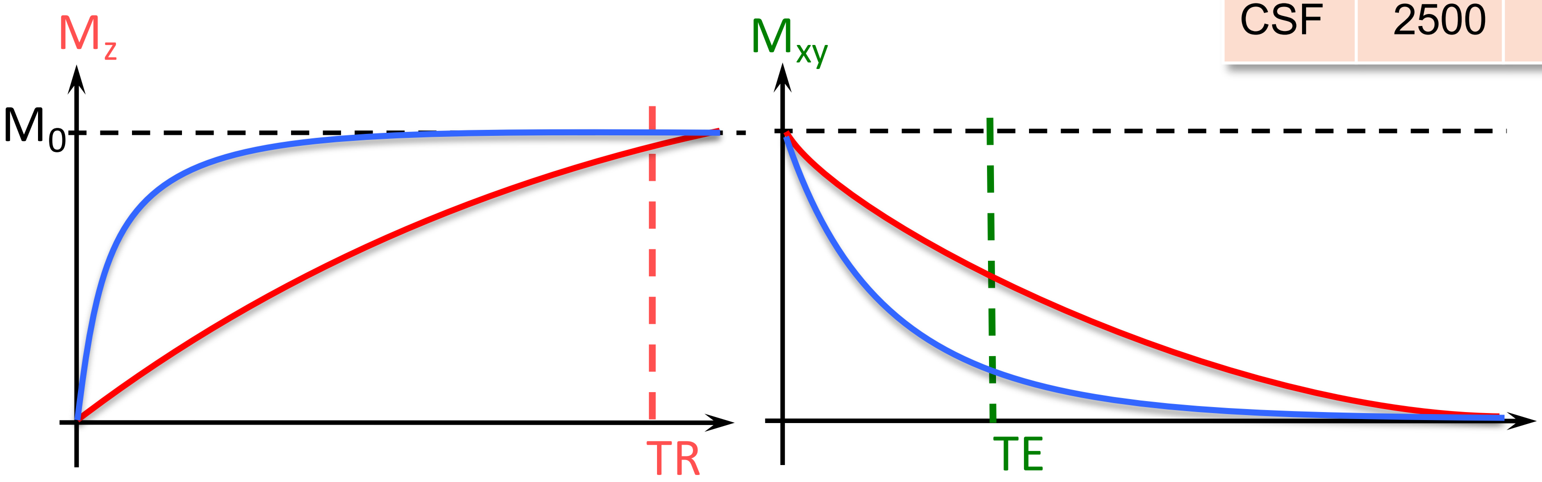
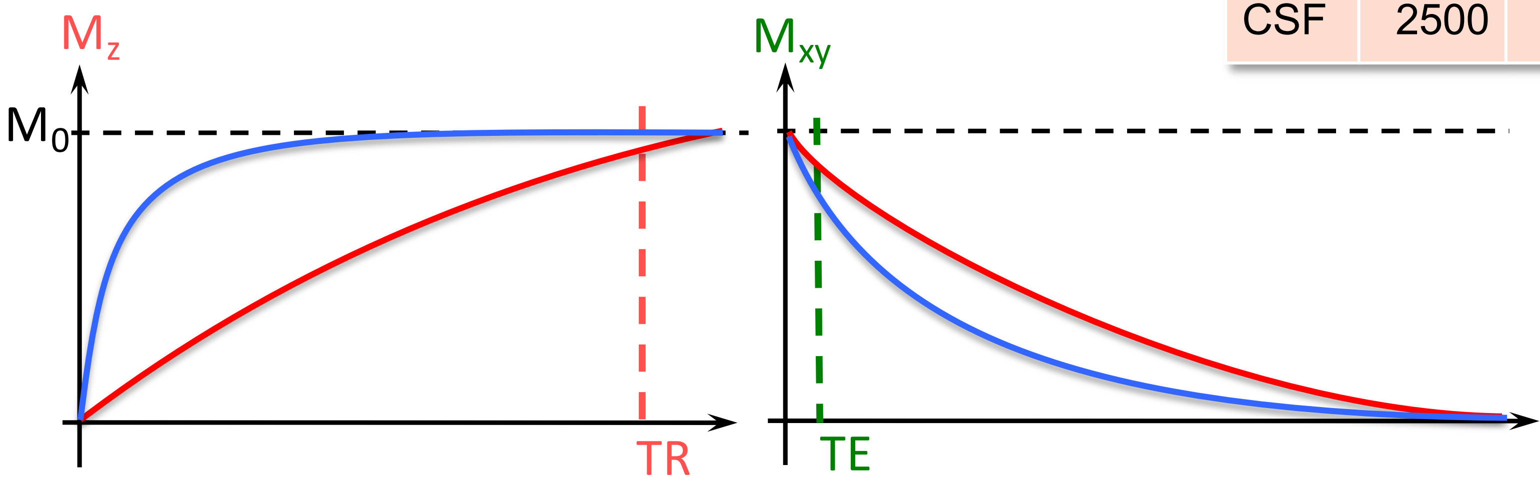


Proton Density (PD) weighting

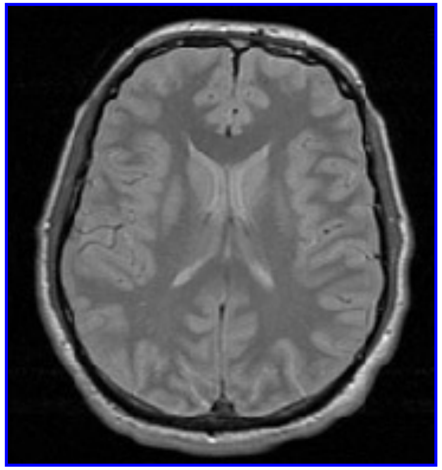
- Long TR: minimizes T_1 contrast
- Short TE: minimizes T_2 contrast
- Remaining contrast depends on proton density

	T_1	T_2
WM	780	90
GM	920	100
CSF	2500	1000



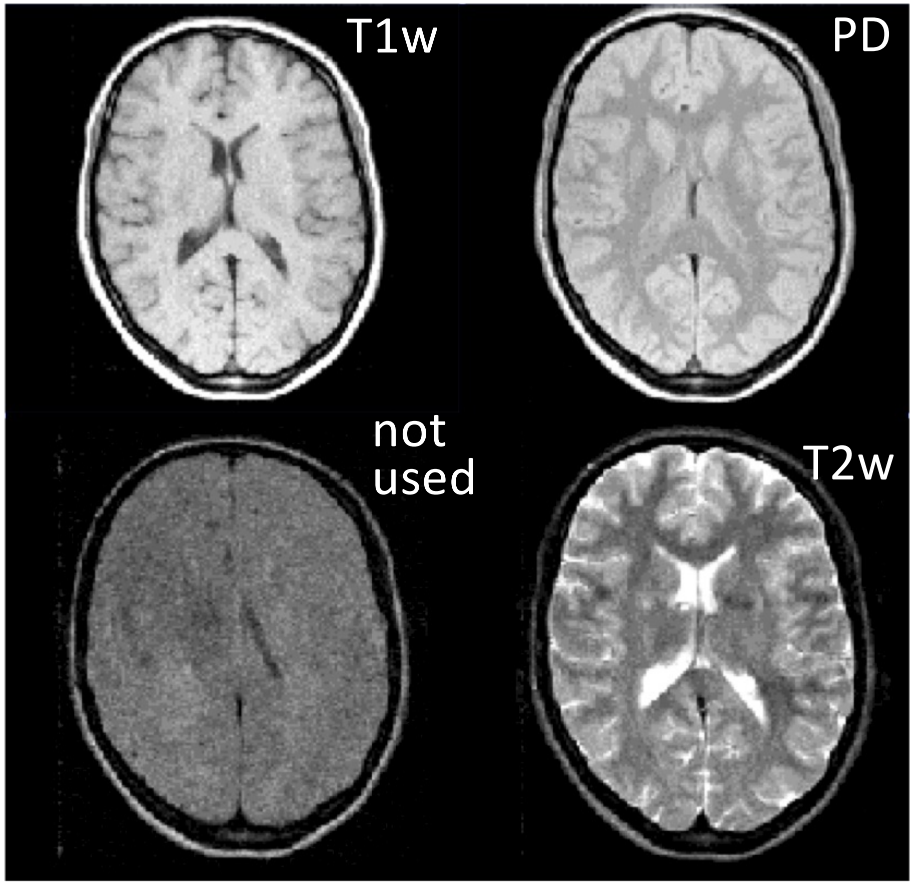
Proton Density (PD)

- The signal magnitude depends on the number of hydrogen nuclei (protons) in the tissue
  - -> Proton density weighting
- Different tissues have different proton densities
  - fluids: ~95 %
  - fat and water-based tissues 65-85 %
- Generally weak contrast images



Peter Jezzard

Contrast weightings



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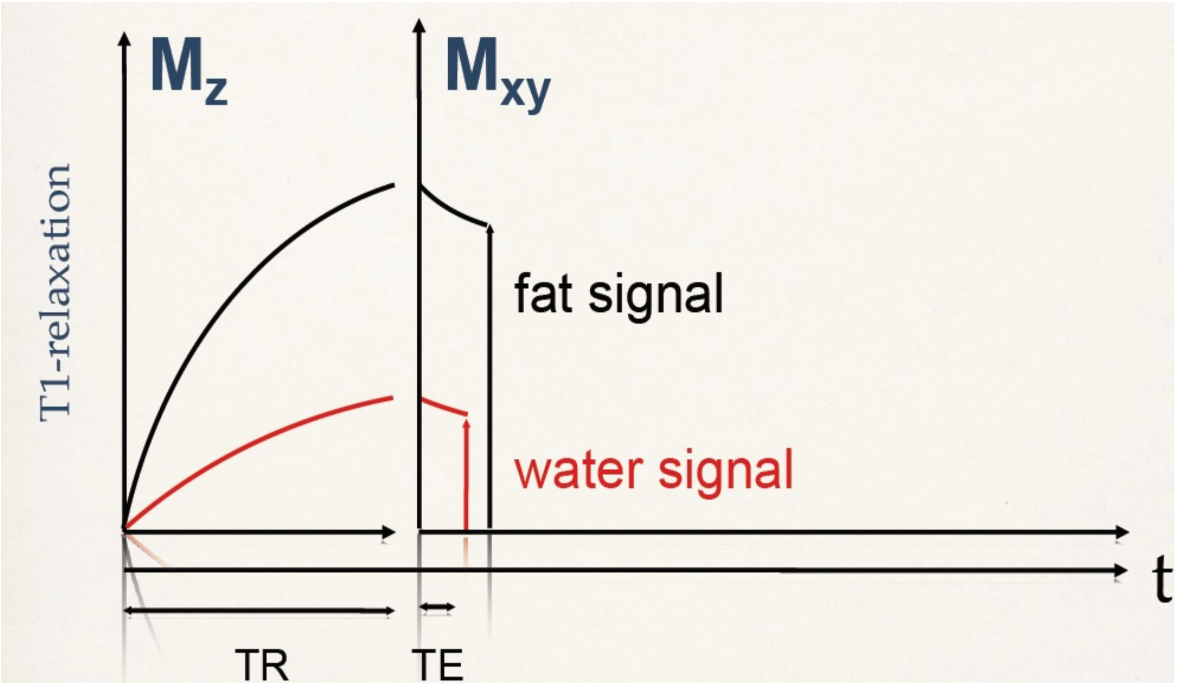
GM	920	100
CSF	2500	1000

Contrast weighting quiz 1/3

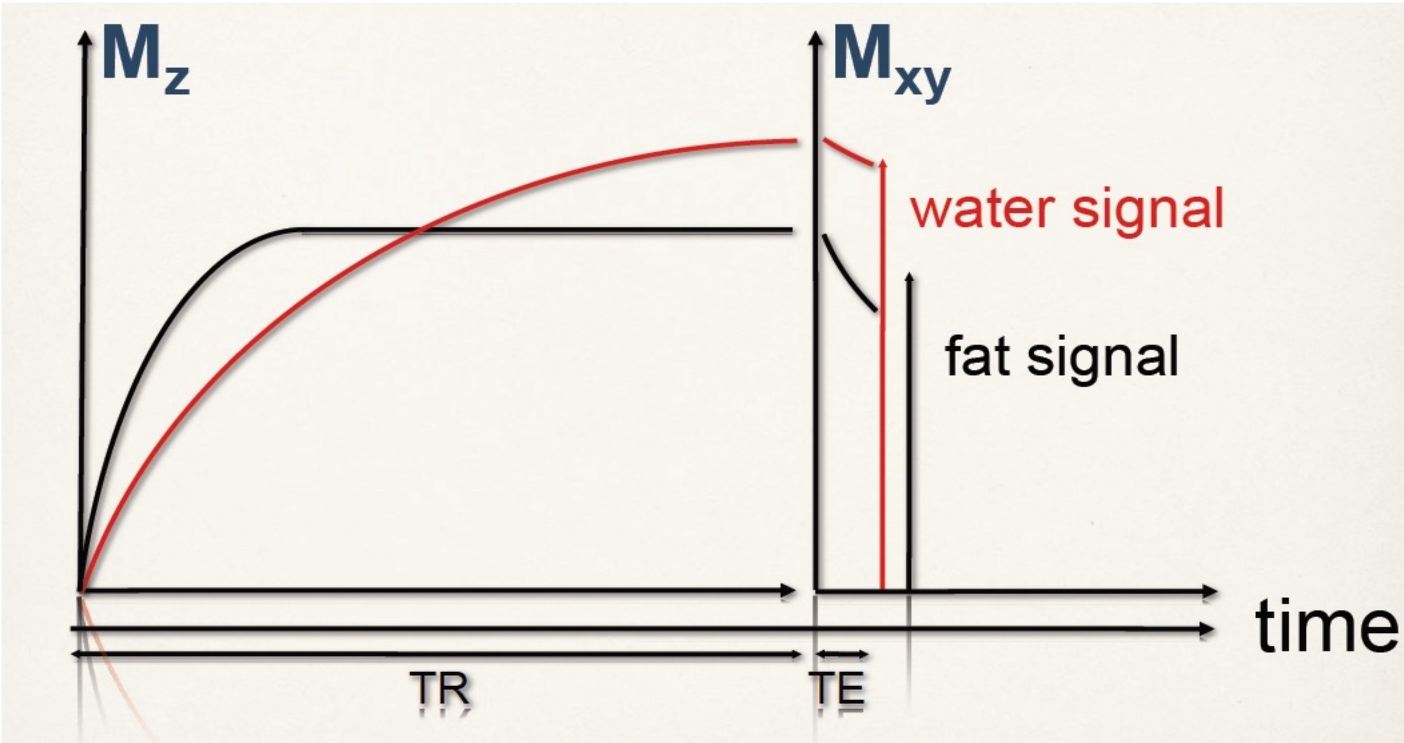
- **TE = <15 ms (short), TR = 400-600 ms (short)**

- **Water dark, fat bright - T1-weighted**



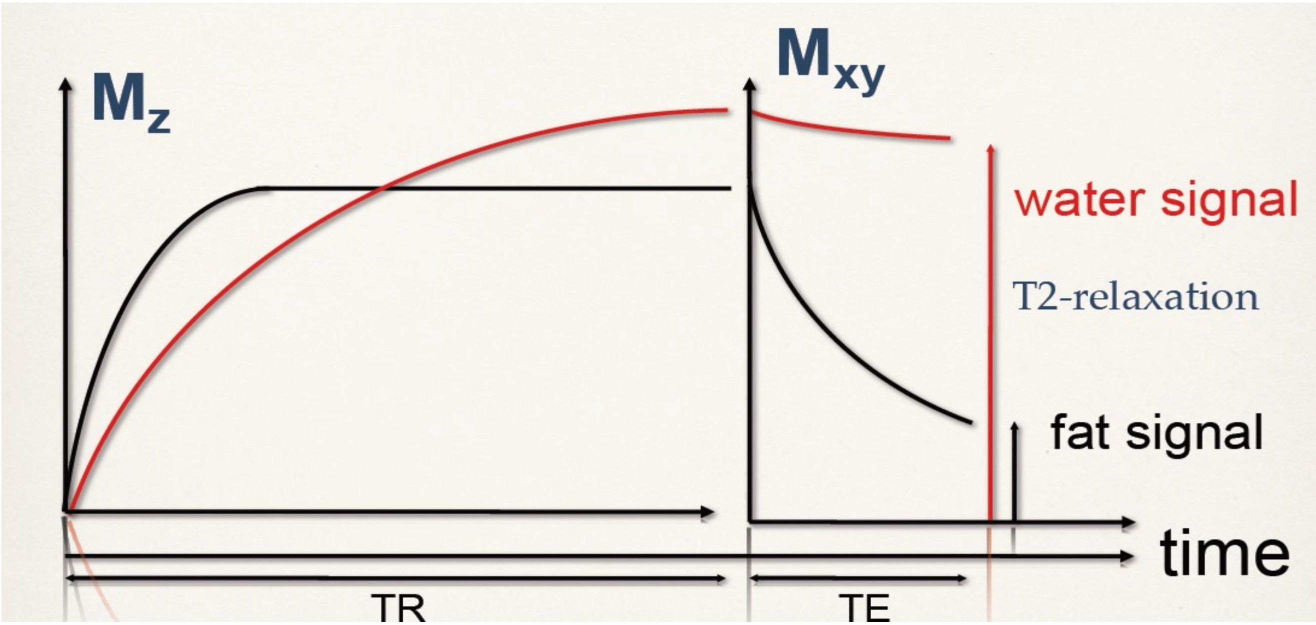
Contrast weighting quiz 2/3

- TE = <15 ms (short), TR > 4000 ms (long)
- Proton density weighted



Contrast weighting quiz 3/3

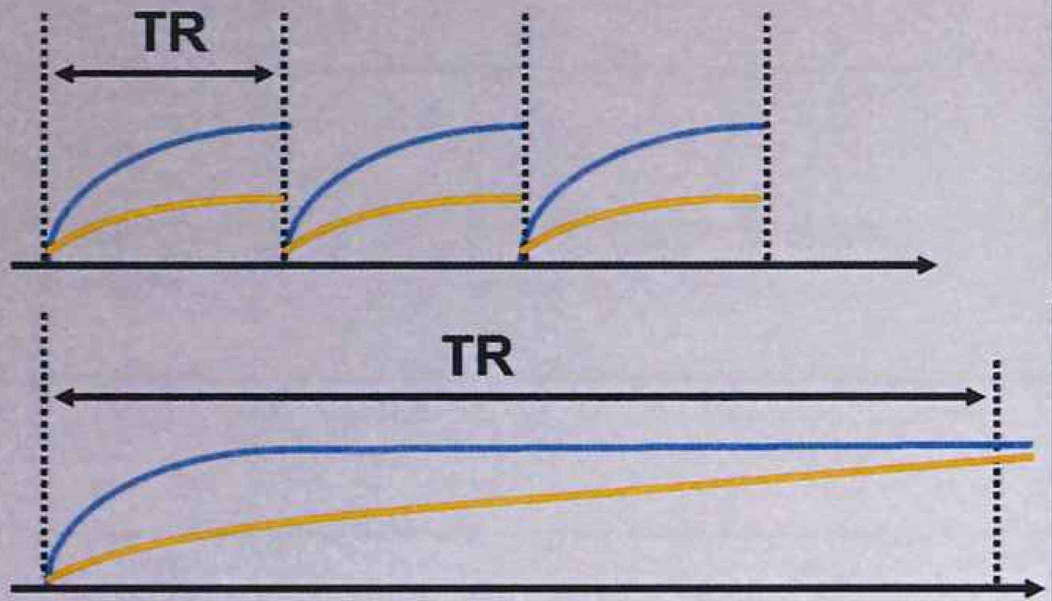
- TE = 80-90 ms (long), TR > 4000 ms (long)
- Water bright, fat dark – T_2 weighted



Olika TR

SIEMENS

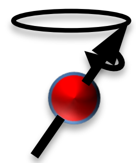
Effekten av lång och kort TR



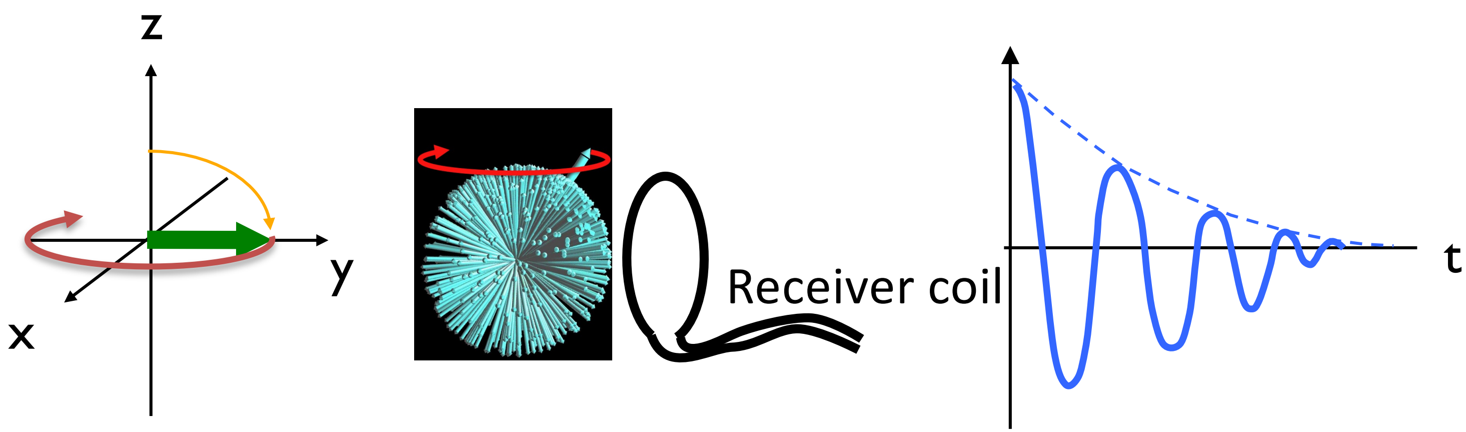
Kort TR trycker ner magnetiseringen från vävnader med lång T1 så att vävnader med kort T1 får starkare signal relativt sätt

Lång TR gör att magnetiseringen från alla vävnader hinner återhämta sig

Summary, nuclear magnetic resonance

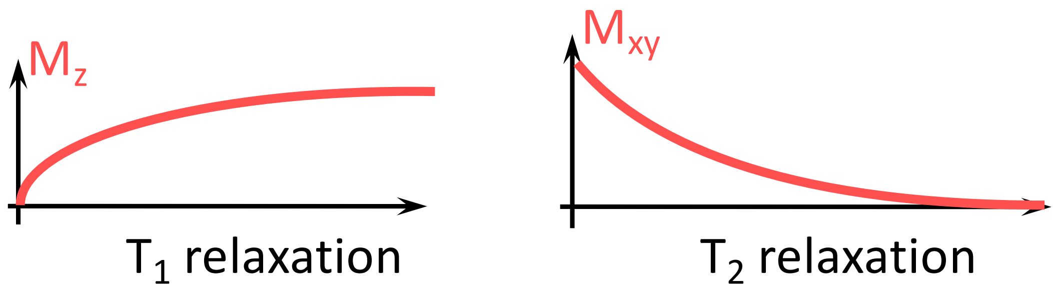


- Spins precess with a Larmor frequency proportional to external field
- Net magnetization vector can be flipped into the transverse plane M_{xy}
- The precessing net magnetization vector in M_{xy} induces current in a loop

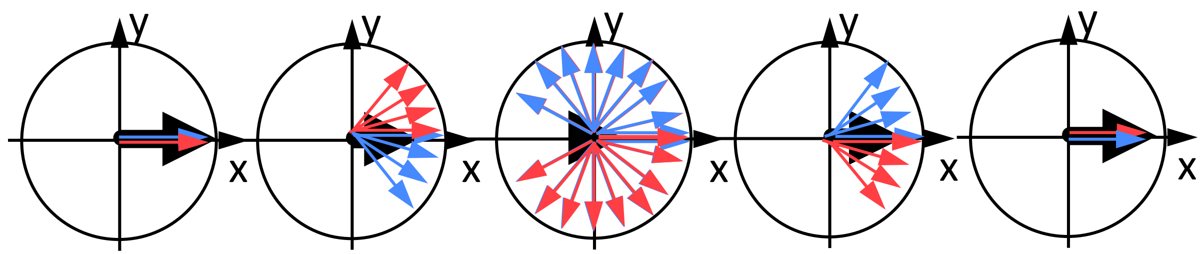


Summary, relaxation

- The longitudinal magnetization M_z returns to equilibrium; T_1 relaxation
- The transverse component dephases due to inhomogeneities; T2 relaxation

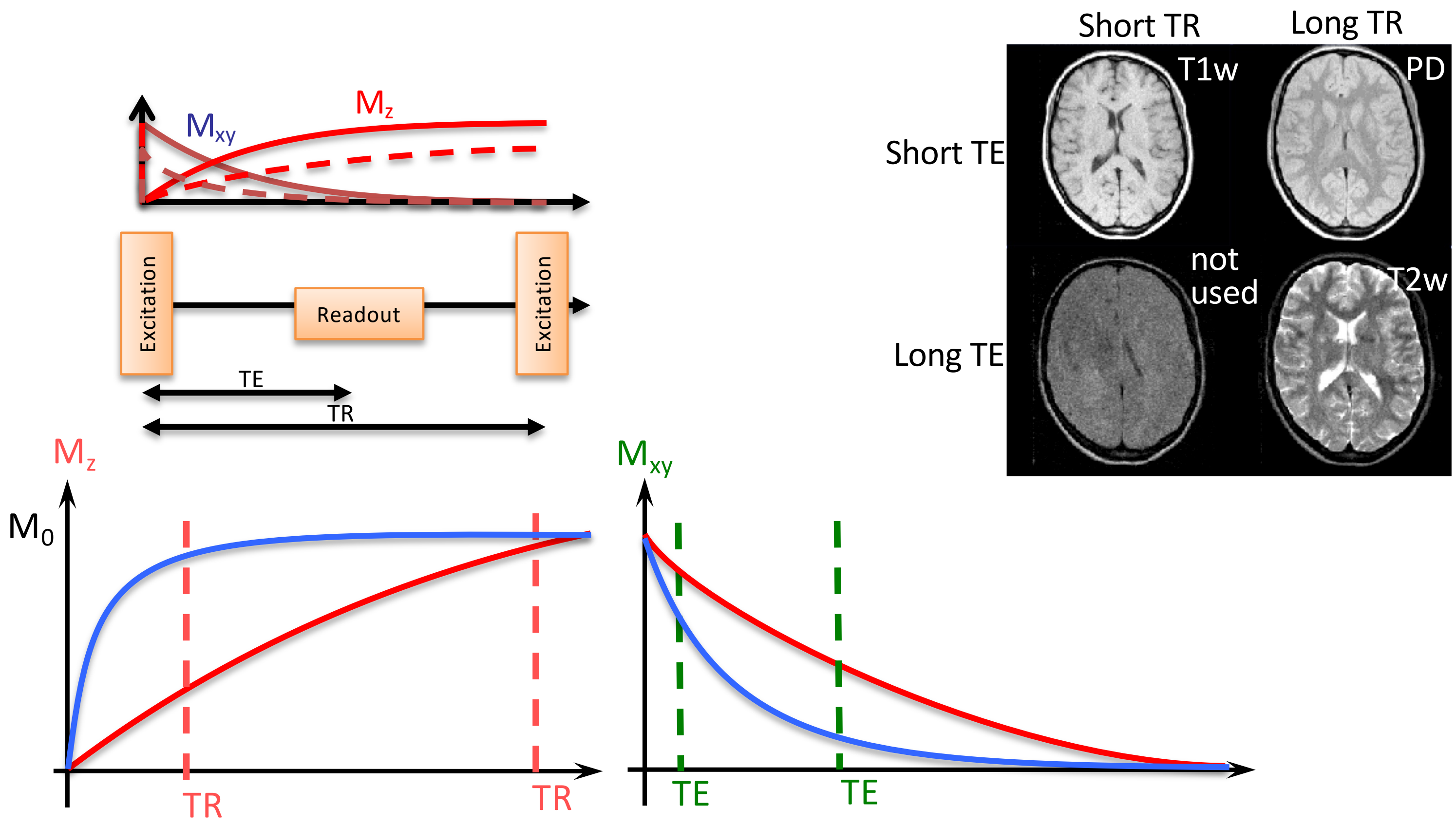


- Using spin-echo, a 180 degree refocusing pulse can recover static inhomogeneities



Summary, image contrast

- By choosing TE and TR appropriately, a desired image contrast can be obtained



Part II

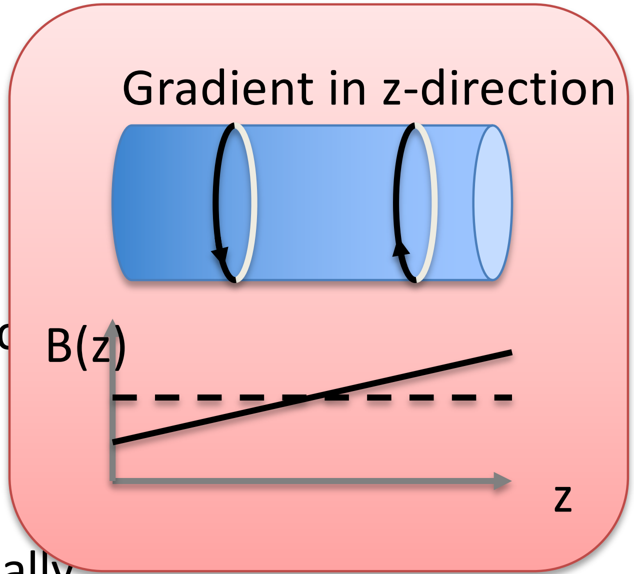
Recap

- Static magnetic field (B_0)
  - Net magnetization precesses with Larmor frequency
- Varying magnetic field with Larmor frequency (B_1, “RF pulse”)
  - Rotates the magnetization of all spins on resonance
  - Excites signal
  - Refocuses static magnetic field variations
- Sequence timing (TE and TR) controls image contrast depending on T_1 and T_2 relaxation

- Spatial localization??

Magnetic field gradients

- Using coils, an additional magnetic field can be superimposed over the B_0 field
  - Field strength varies
  - Field direction *always* in z-direction
- The current can be varied in the gradient coils
  - One coil set in each direction x, y and z
  - Combined will create a gradient in arbitrary direc



- Larmor frequency is proportional to magnetic field
- Using gradients makes Larmor frequency vary spatially

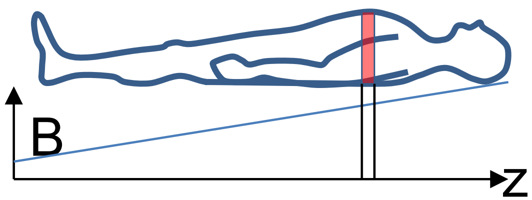
Spatial localization of signal

Use magnetic field gradients to modify the local field strength

- Larmor frequency changes linearly with space

Combine RF pulse with gradient on

- Excites one slice, leaving rest of body unaffected

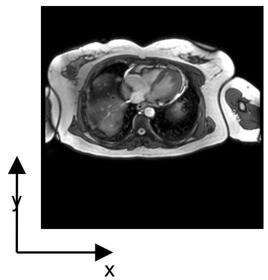


$$f = \gamma(B_0 + G_z z)$$

Read signal with in-plane encoding

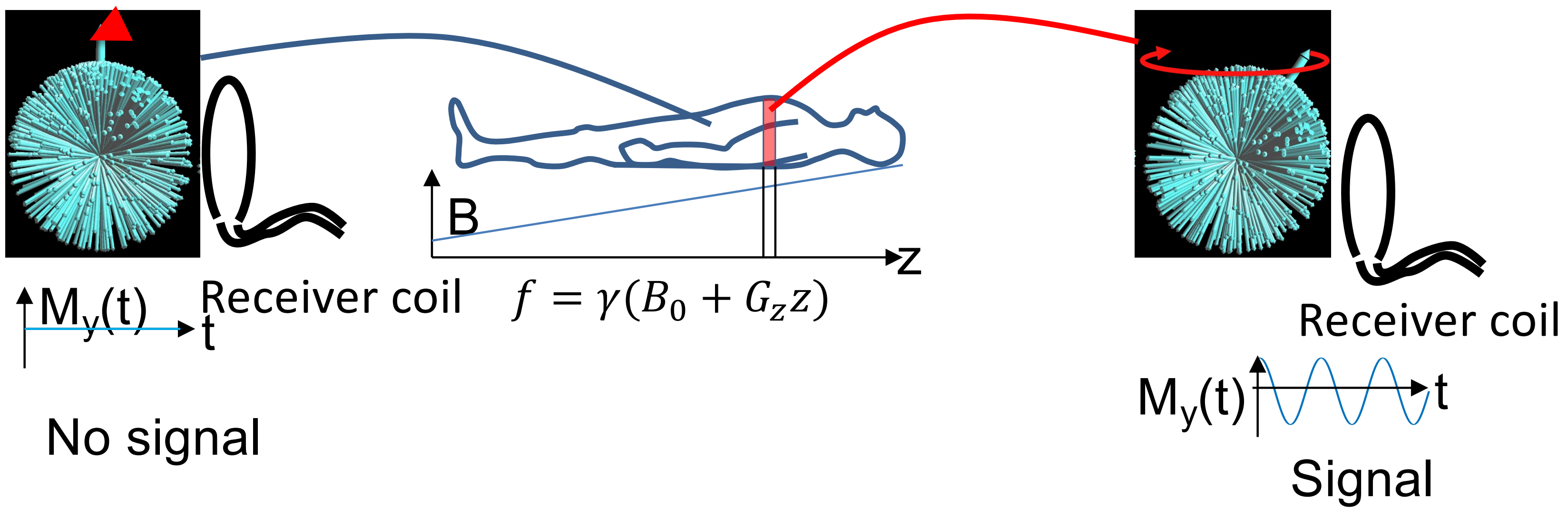
- Signals can be separated by frequency
- *k*-space encoding

$$f = \gamma(B_0 + G_x x)$$

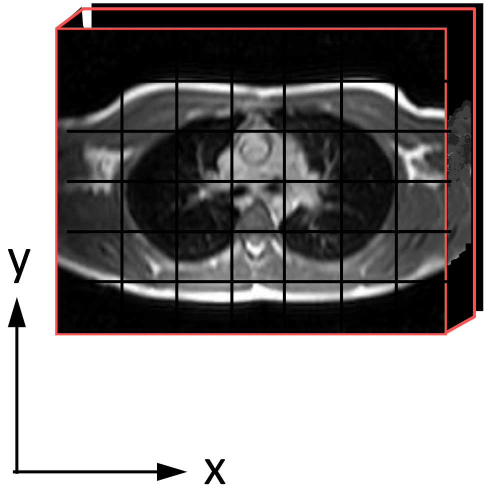


Slice selection

- With a gradient, the Larmor frequency varies linearly with space
- Only the spins in resonance will be excited and thus provide signal
- Use a non-zero bandwidth to excite a slice



2D in-plane encoding

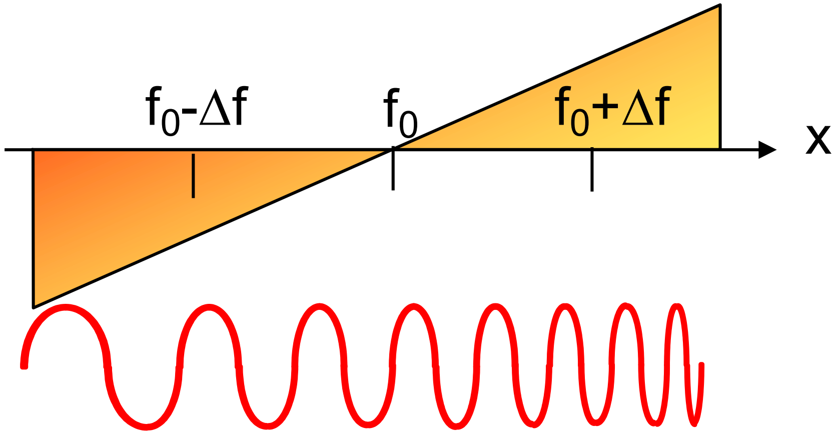


- The signal received comes from the whole excited volume
- We use gradients in two steps to encode the x and y directions
- Frequency encoding (convention: x)
- Phase encoding (convention: y)

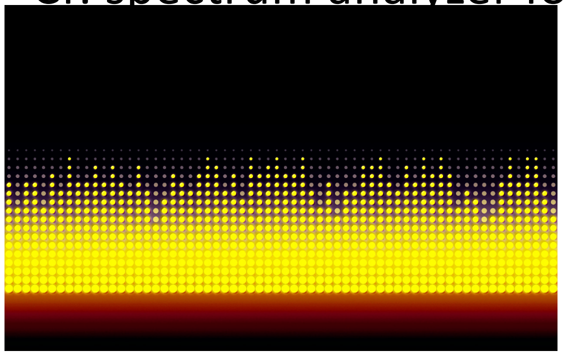
Frequency encoding direction (x)

- A gradient is turned on during signal reception
- Signals from different positions have different frequencies

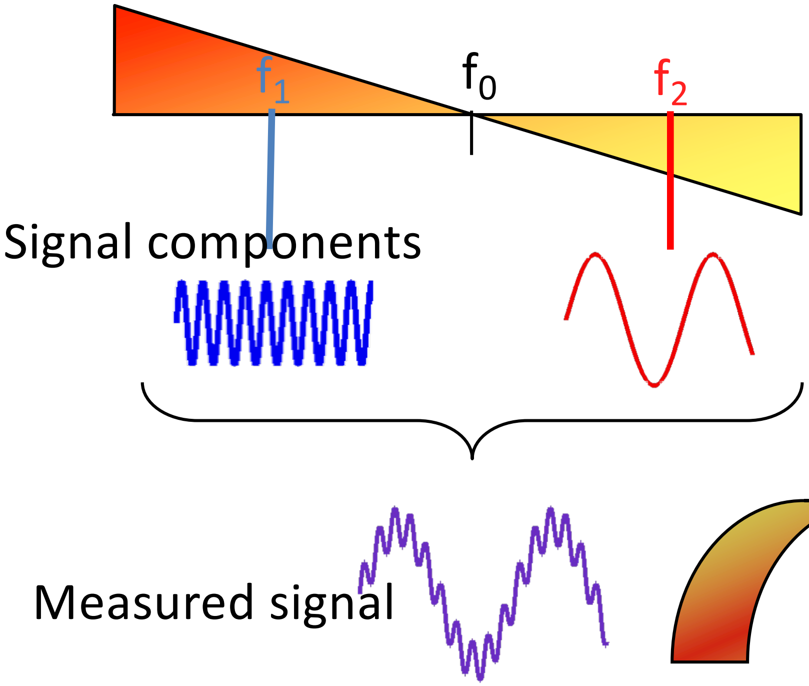
$$f = \gamma(B_0 + G_x x)$$



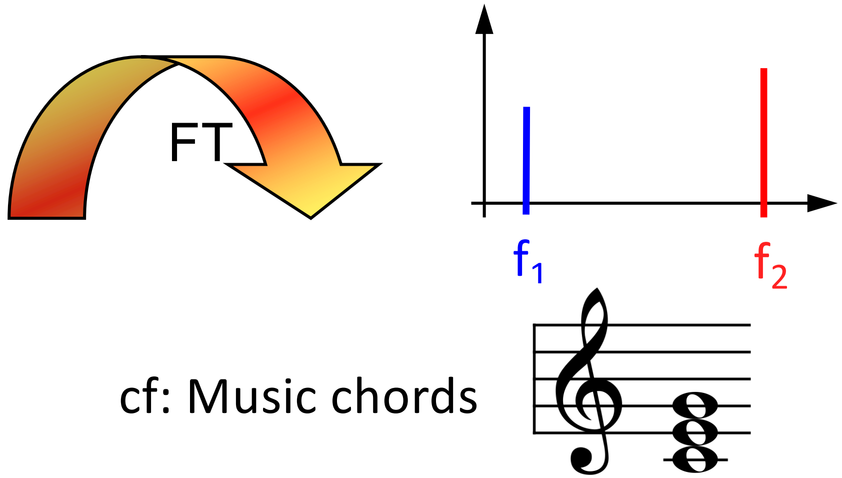
Cf: spectrum analyzer for music



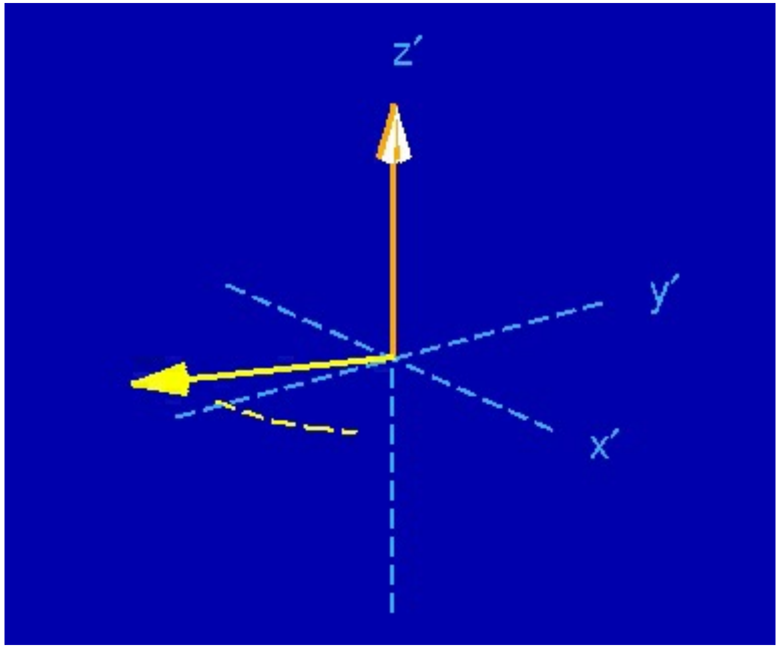
The Fourier transform



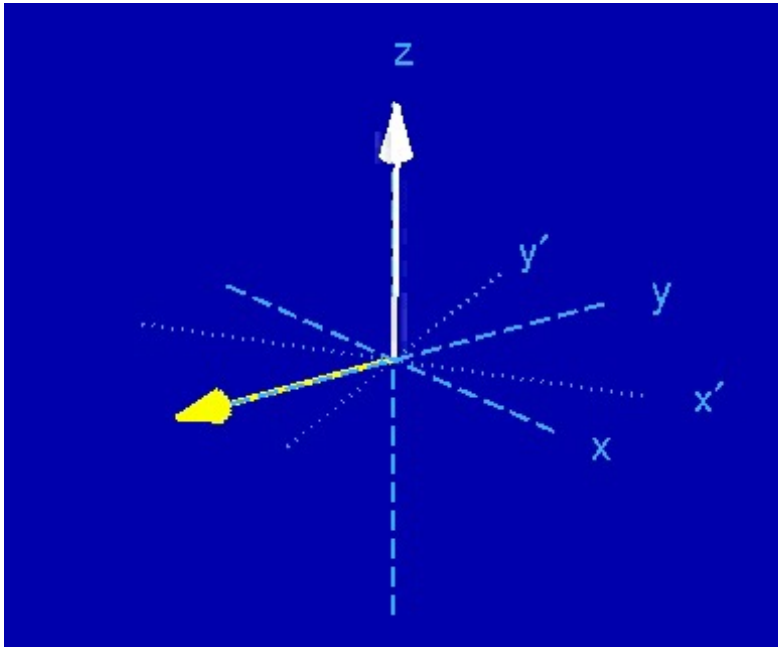
The Fourier transform recovers the frequencies and separates into f_1 and f_2



Rotating frame of reference



Static frame of reference



Rotating frame of reference

Movies from http://mrsrl.stanford.edu/~brian/intromr/ (Brian Hargreaves)

Frequency encoding

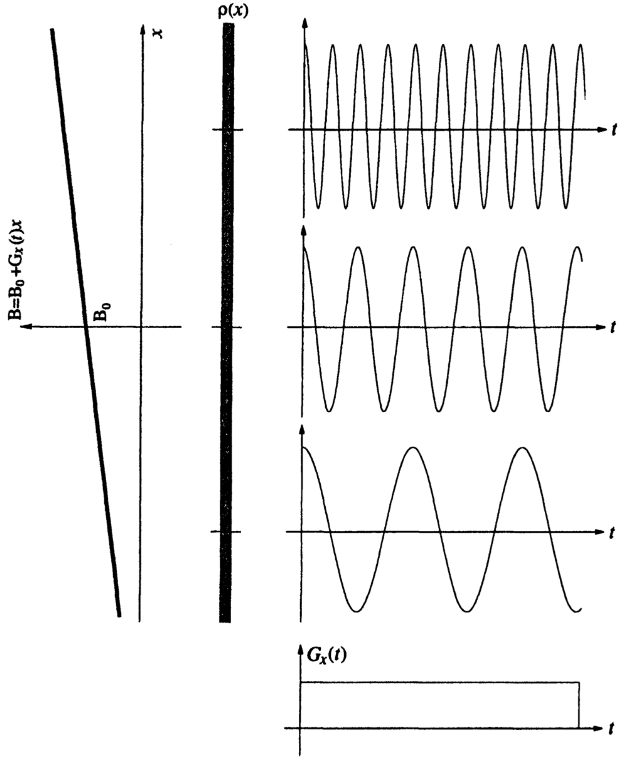


Figure 5.7 Localized signals from a hypothetical one-dimensional object in the presence of a frequency-encoding gradient.

Phase encoding

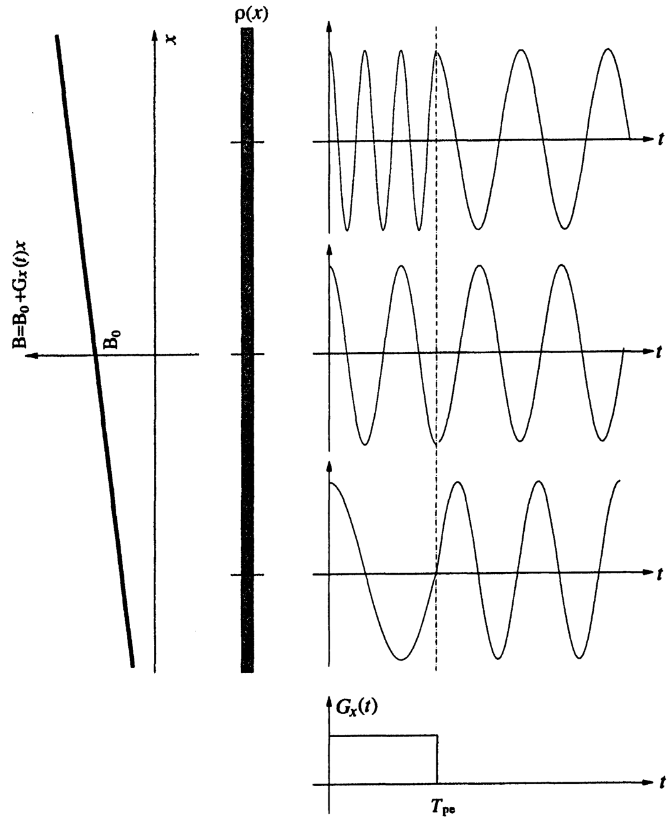


Figure 5.9 Phase-encoded signals from a one-dimensional object. Note that phase encoding is achieved by pre-frequency encoding the signals for a short period of time T_{pe}.

k-space

Signal from entire object with phase evolution due to local field deviation
$$S(t) \propto \int_{object} M_{xy}(\vec{r},0)e^{-i\Delta\omega(\vec{r})t}d\vec{r}$$

Local field in the presence of of a gradient (x-direction)
$$\Delta\omega(\vec{r})t = x\gamma\int_0^t G_x(\tau)d\tau$$

Insert and substitute
$$S(t) \propto \int_{object} M_{xy}(\vec{r},0)e^{-ix\underbrace{\gamma\int_0^t G_x(\tau)d\tau}_{2\pi k_x}}d\vec{r}$$

k-space coordinate
$$2\pi k(t) = \gamma\int_0^t G(\tau)d\tau$$

k-space encoding and Fourier transform

$$S(t) \propto \int_{object} M_{xy}(\vec{r},0)e^{-ix\underline{\gamma \int_0^t G_x(\tau)d\tau}}d\vec{r}$$

$$2\pi k_x$$

$$FT[g(x)] = G(k) = \int_{-\infty}^{+\infty} g(x)e^{-2\pi ikx}\,dx$$

Frequency encoding

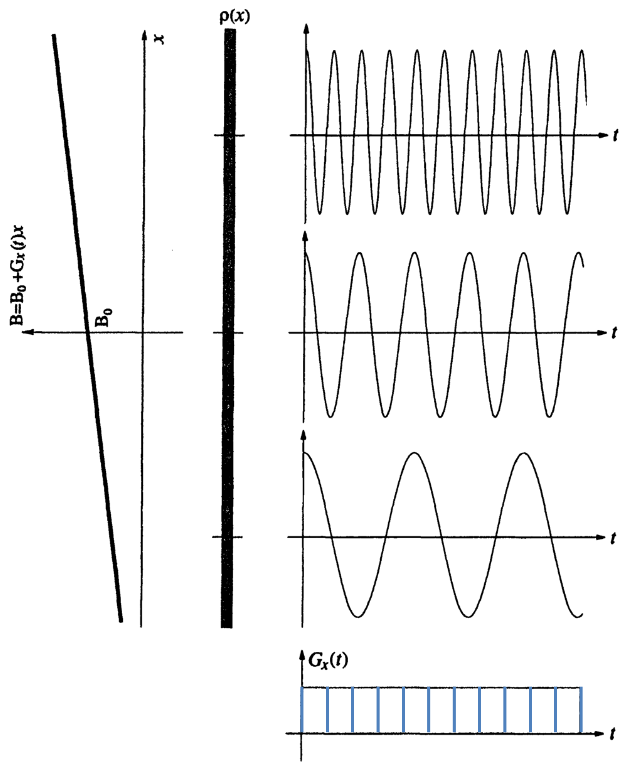


Figure 5.7 Localized signals from a hypothetical one-dimensional object in the presence of a frequency-encoding gradient.

Phase encoding

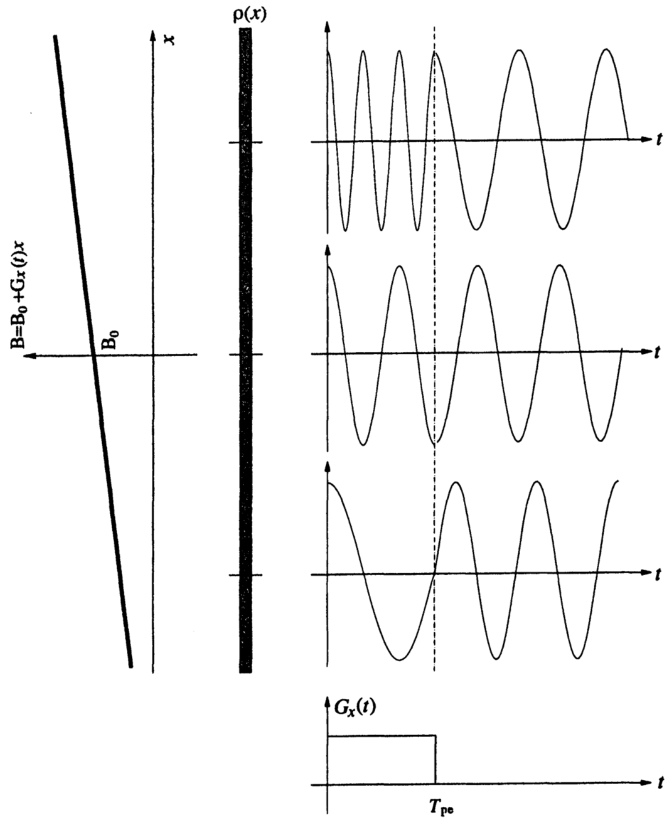
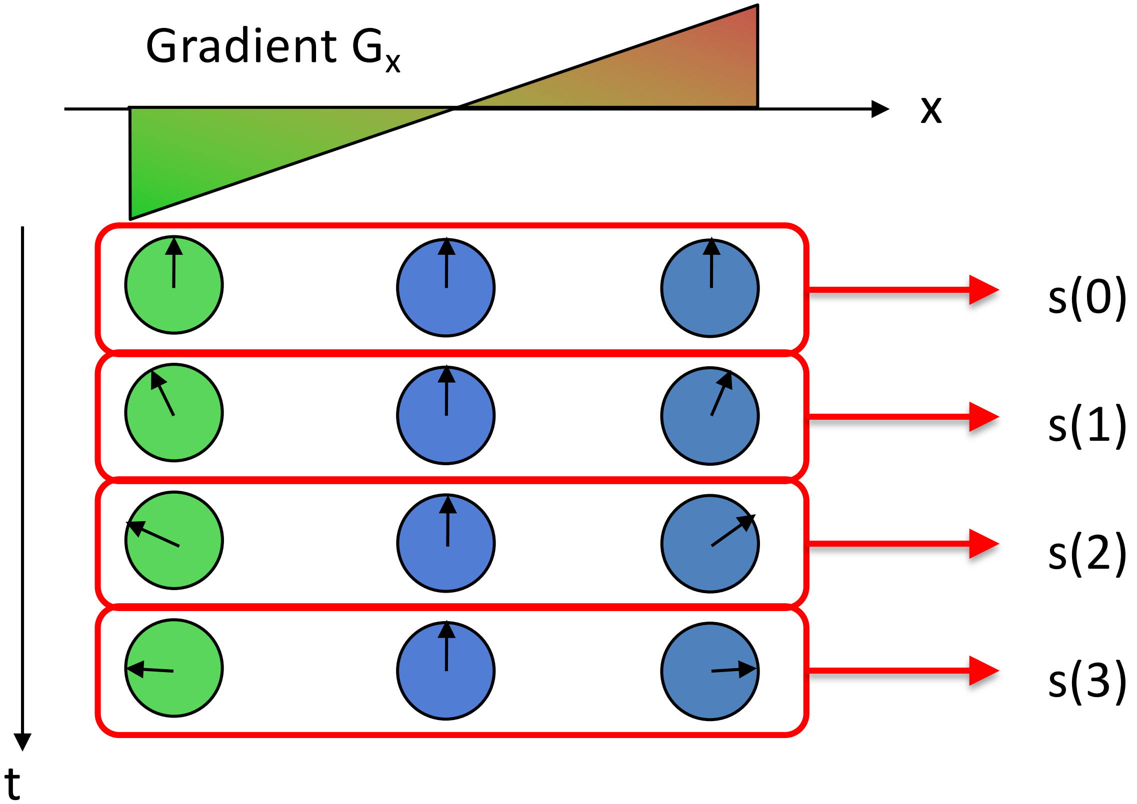
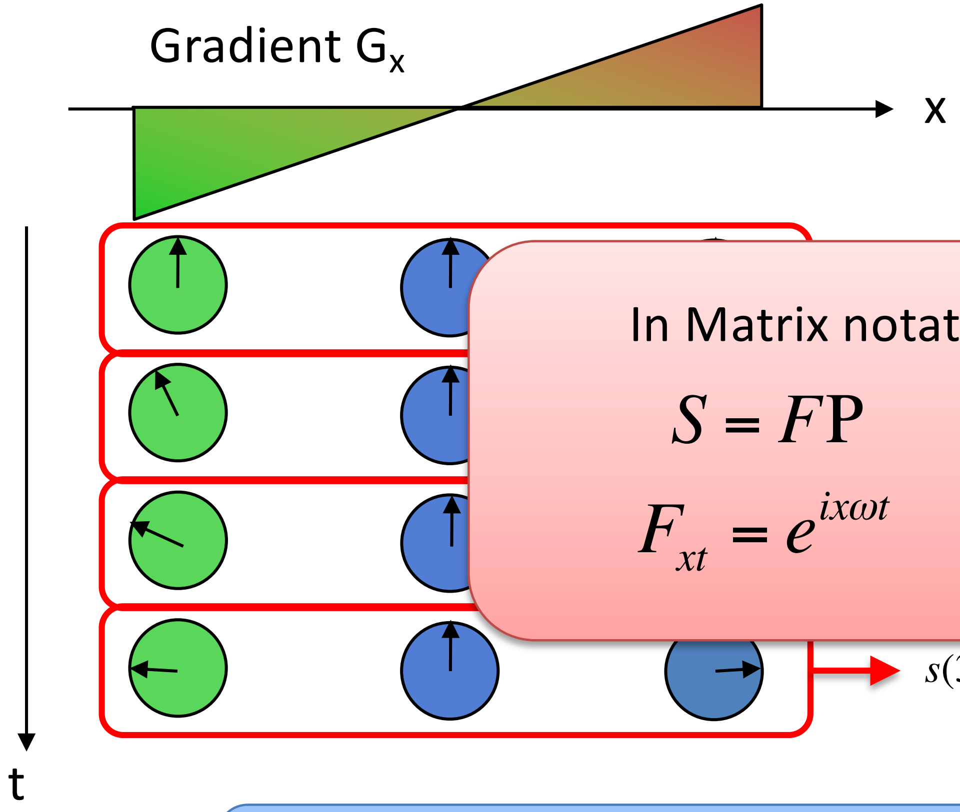


Figure 5.9 Phase-encoded signals from a one-dimensional object. Note that phase encoding is achieved by pre-frequency encoding the signals for a short period of time T_{pe}.

Signal in time domain



Signal in time domain



$$s(t) = \sum_x \rho(x) e^{i\psi(t,x)}$$

$$\psi(t,x) = x\omega t$$

$$-1)e^{i\cdot -1\cdot\omega\cdot 0} + \rho(0)e^{i\cdot 0\cdot\omega\cdot 0} + \rho(1)e^{i\cdot 1\cdot\omega\cdot 0}$$

$$-1)e^{i\cdot -1\cdot\omega\cdot 1} + \rho(0)e^{i\cdot 0\cdot\omega\cdot 1} + \rho(1)e^{i\cdot 1\cdot\omega\cdot 1}$$

$$-1)e^{i\cdot -1\cdot\omega\cdot 2} + \rho(0)e^{i\cdot 0\cdot\omega\cdot 2} + \rho(1)e^{i\cdot 1\cdot\omega\cdot 2}$$

$$s(3) = \rho(-1)e^{i\cdot -1\cdot\omega\cdot 3} + \rho(0)e^{i\cdot 0\cdot\omega\cdot 3} + \rho(1)e^{i\cdot 1\cdot\omega\cdot 3}$$

In Matrix notation

$$S = F\mathrm{P}$$

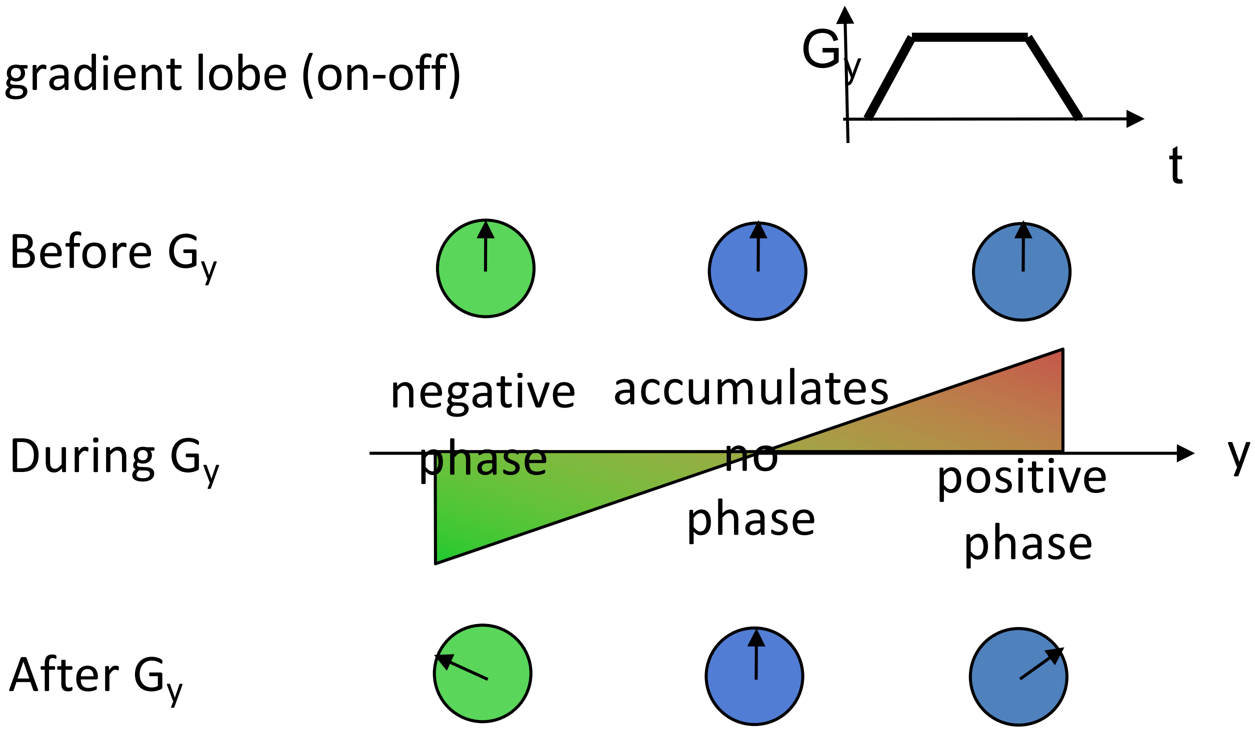
$$F_{xt} = e^{ix\omega t}$$

Reconstruction

$$\tilde{\mathrm{P}} = F^{-1}S$$

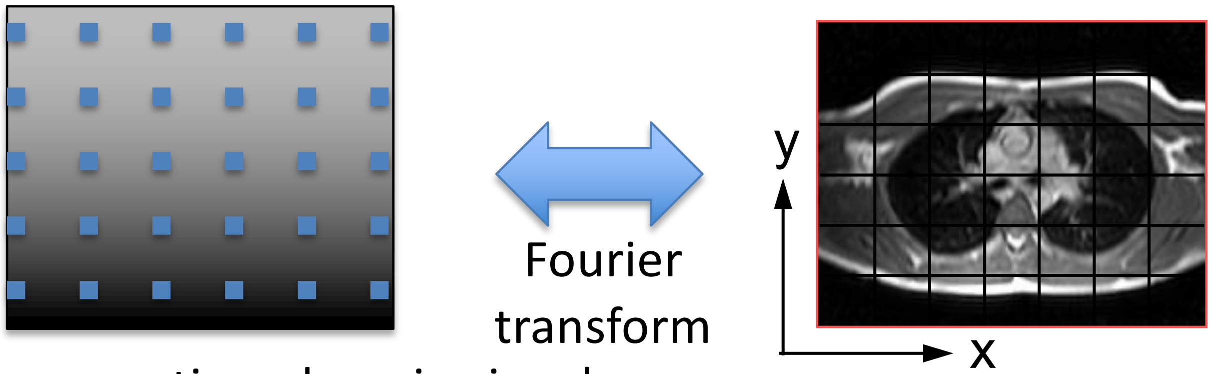
Phase encoding in the y direction

- Apply a short gradient lobe (on-off)



In-plane encoding in two directions

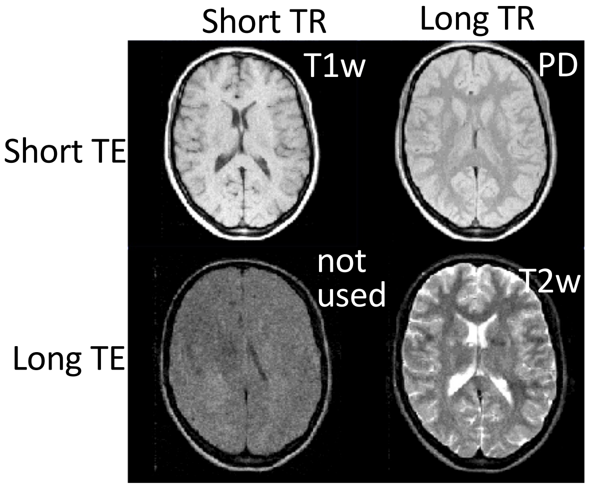
- Need all combinations to reconstruct an N x N pixel image
- x direction in one go (frequency encoding)
  - acquires N samples per readout
- y direction sequentially (phase encoding)
  - repeated N times, with varying strength of the G_y gradient pulse

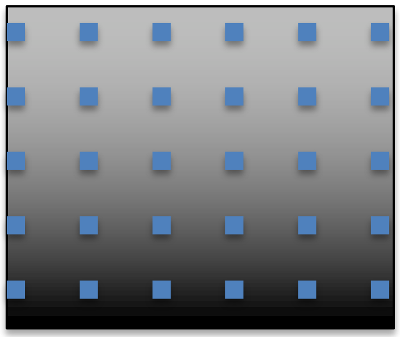


k-space – time domain signal

Scan time

- Need to collect N phase encodings
- Each TR, we can acquire N time domain samples (frequency encoding)
- For an image of N x N, the scan time becomes N x TR
- TR is chosen for the appropriate contrast
  - 256 x 256 with TR of 100 ms => 25.6 s scan time for one slice





k-space – time domain signal

Summary

- Slice direction is encoded during excitation
  - z gradient on -> linear frequency in z-direction
  - RF pulse with suitable bandwidth excites only spins within a band of Larmor frequencies
- In-plane encoding during signal reception
  - Frequency encoding (x) direction, gradient on during readout
    - Different positions have different frequency
  - Phase encoding (y) direction, gradient lobe of varying strength before each readout
    - Different positions have different phase
  - Reconstructed using Fourier transform
- MR measures the complex-valued Fourier transform of the image

1. Vad står förkortningen MR för?
☐ Magnetröntgen
☐ Mjukdelsröntgen
☐ Magnetresonans

2. Från vad kommer MR-signalen?
☐ Från vatten
☐ Från alla atomkärnor
☐ Från väteatomkärnan

3. Vilken av följande komponenter behövs för att ta bilder med en magnetkamera?
☐ En supraledande magnet
☐ Gradientspolar
☐ Ytspolar

4. Den kraftigaste kontrastmekanismen i MR bygger på skillnader i?
☐ T_1 och T_2
☐ Protondensitet
☐ Resonansfrekvens

5. Frekvensen på Larmor-precessionen beror på?
☐ Frekvensen på RF-sändaren
☐ T_1-relaxationen
☐ Magnetfältstyrkan och typ av atomkärna

6. Vilket påstående är sant?
☐ T_1-relaxationen är alltid snabbare än T_2-relaxationen
☐ T_2^* är alltid kortare än T_2
☐ Efter excitering avklingar signalen på grund av T_1-relaxationen

7. Gradienter används för?
☐ Frekvenskodning och faskodning
☐ Snittselektion

8. Vilket påstående om MR-signalen är sant?
☐ Bara magnetisering i samma riktning som huvudmagnetfältet kan detekteras
☐ MR-signalen är proportionell mot nettomagnetiseringen (M_0)
☐ Bara magnetisering i transversalplanet (xy-planet) kan detekteras

9. Vilket påstående är sant?
☐ Signalen innehåller bidrag från hela den valda skivan
☐ Intensiteten i en bildpixel är densamma som motsvarande pixel i *k*-space
☐ I varje faskodningssteg kodas en rad pixlar av MR-bilden

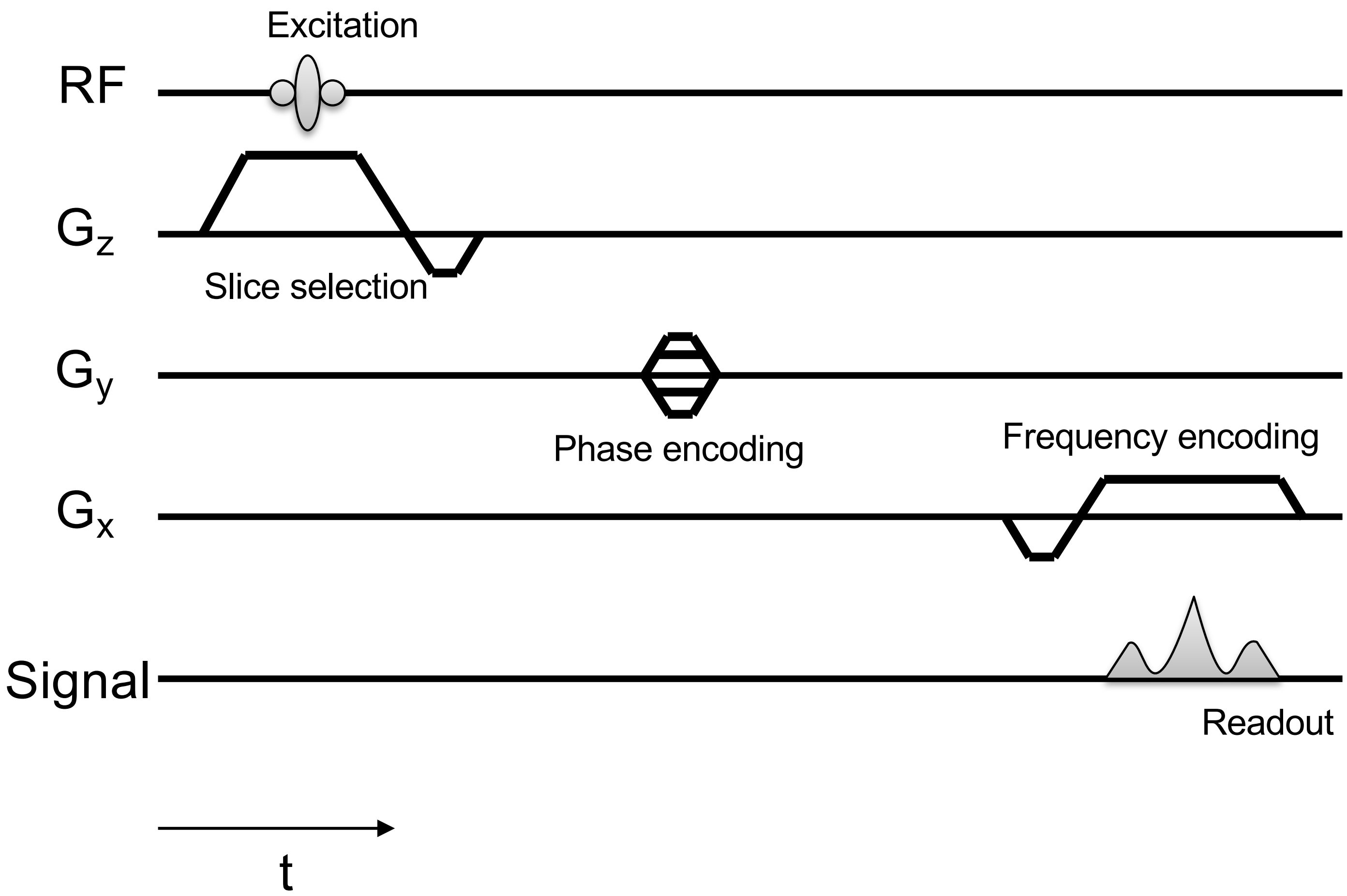
10. En vävnad med lång T_1 och lång T_2 är ljus
☐ På en T_1-viktad bild
☐ På en T_2-viktad bild

Part III

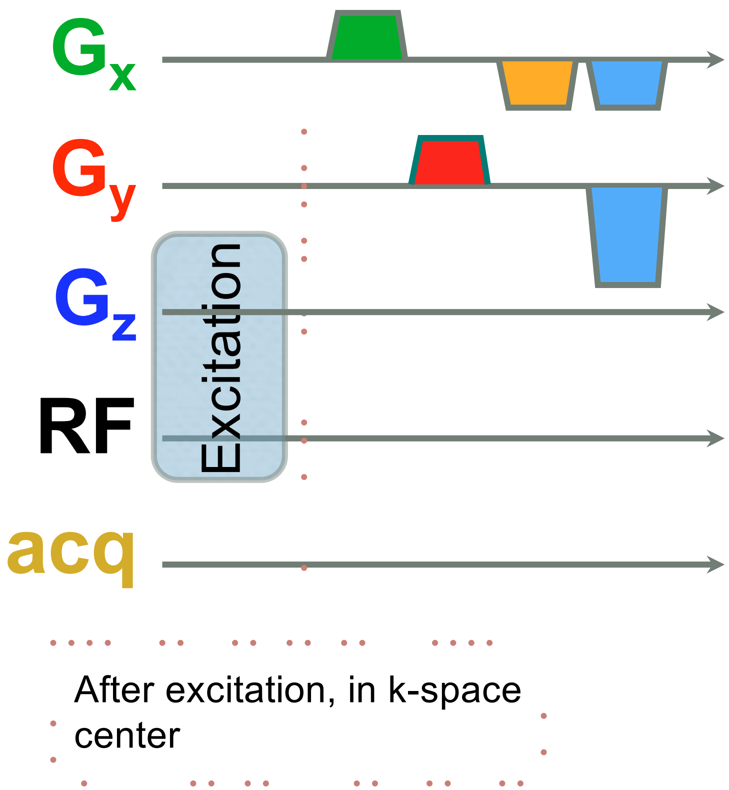
MRI is versatile

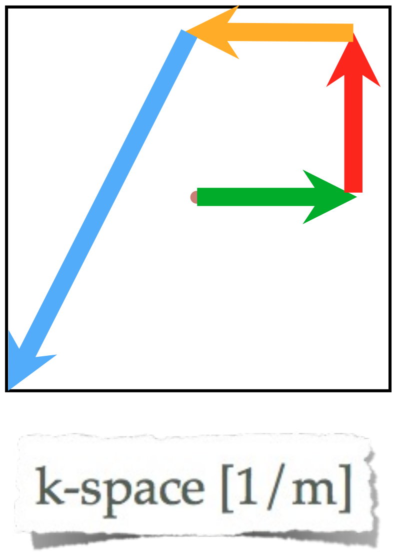
- Means of manipulation
  - RF pulses
    - Only on-resonance spins get excited
    - Arbitrary flip angle
    - Prepulses
  - Gradients
    - Slice selection: change resonance frequency spatially
    - Spatial encoding: position gets encoded in phase
  - Time
    - Relaxation, transverse (T_2) and longitudinal (T_1)
  - Motion
    - Phase shifts and dephasing during gradients

Pulse sequences

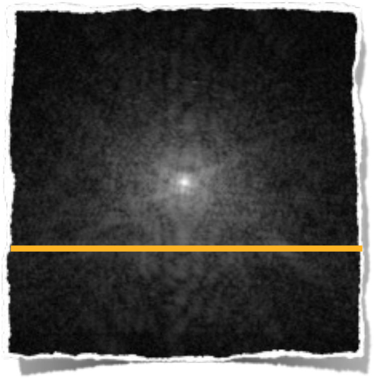


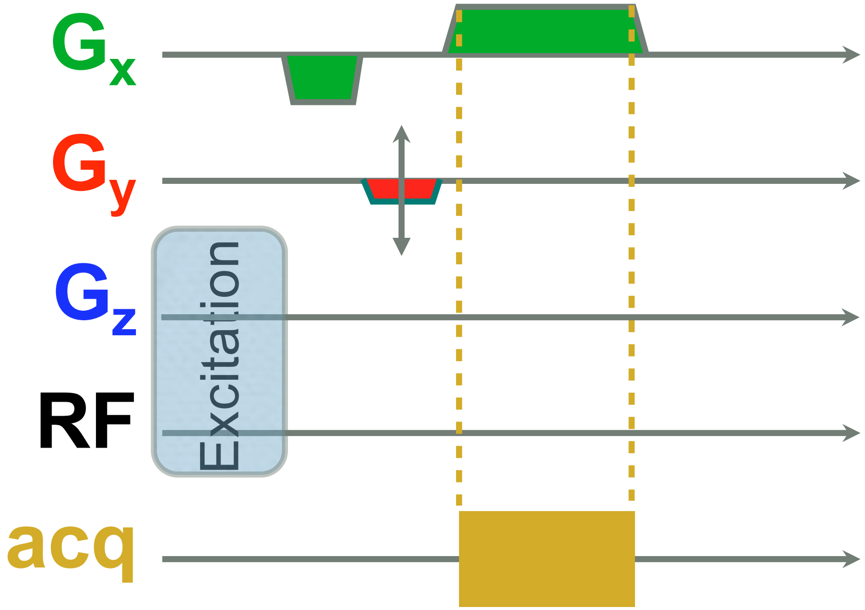
Gradient area & k-space

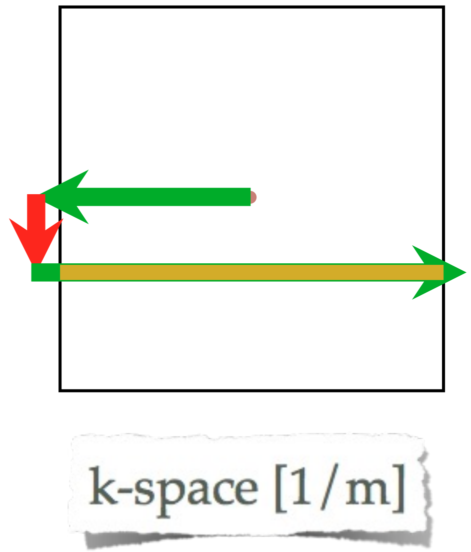




Gradient echo







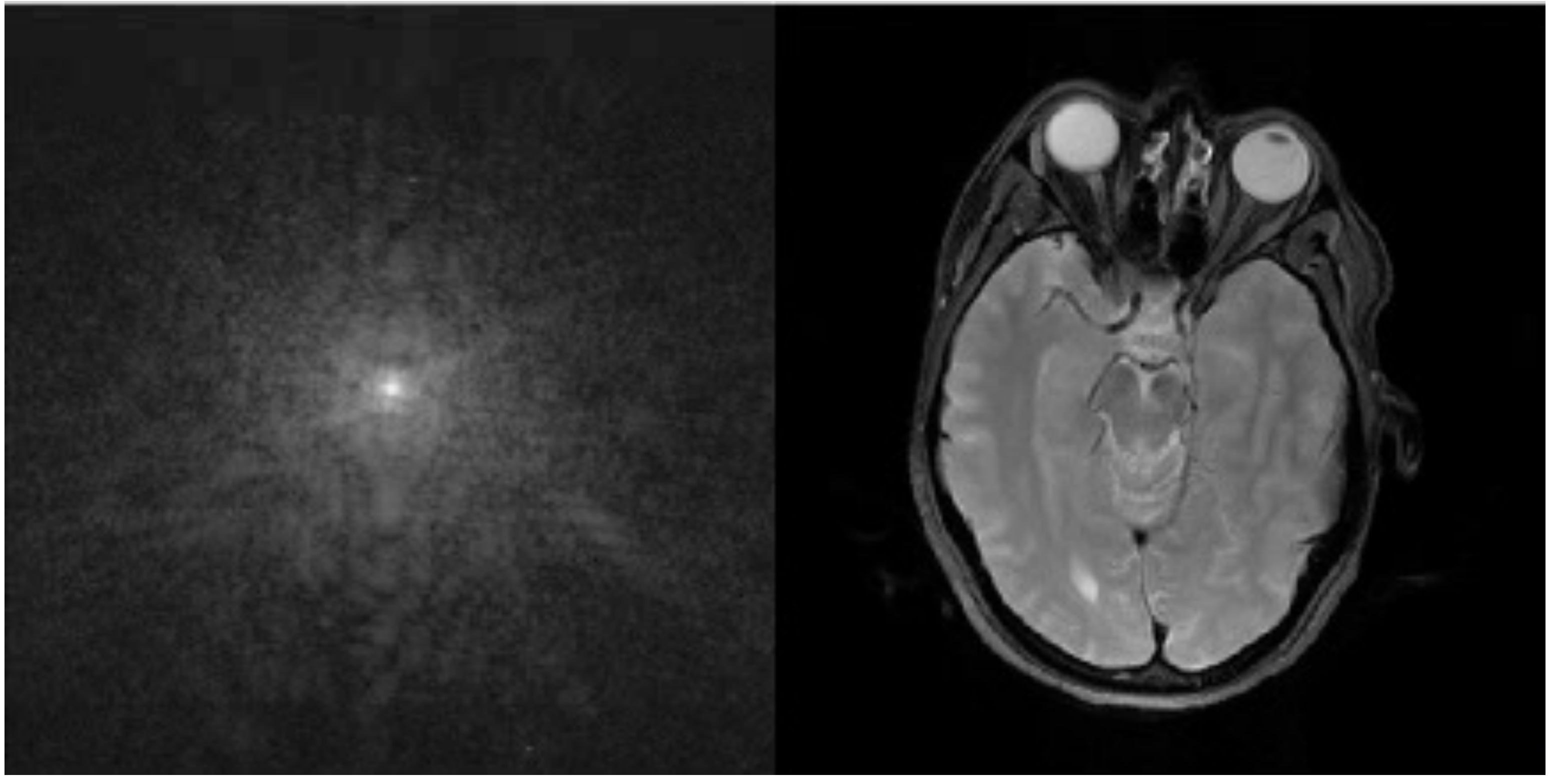
Linear left-right k-space filling

Linear top-down k-space filling

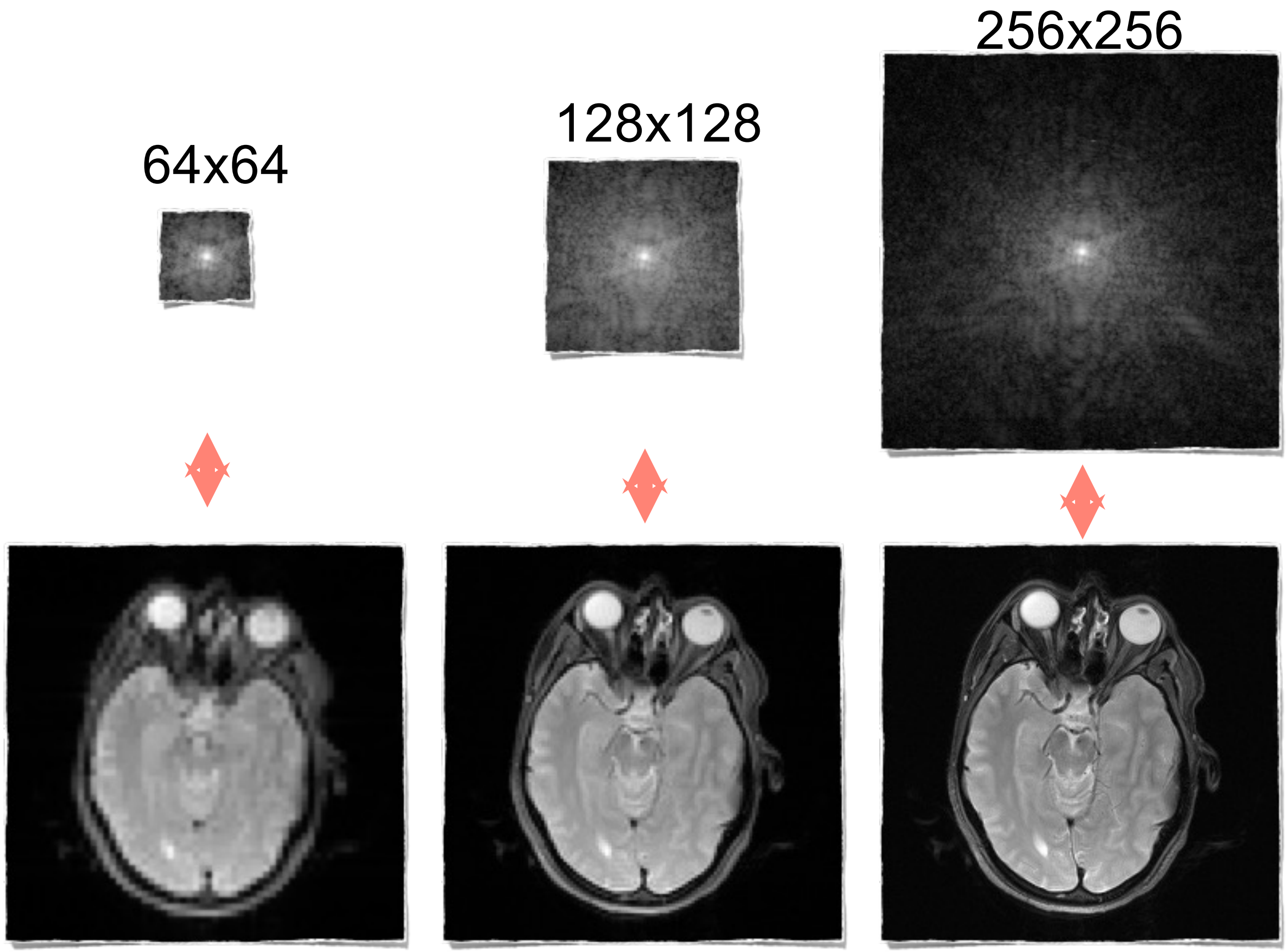
Center-out vertical k-space filling

From 2x2 to 256x256

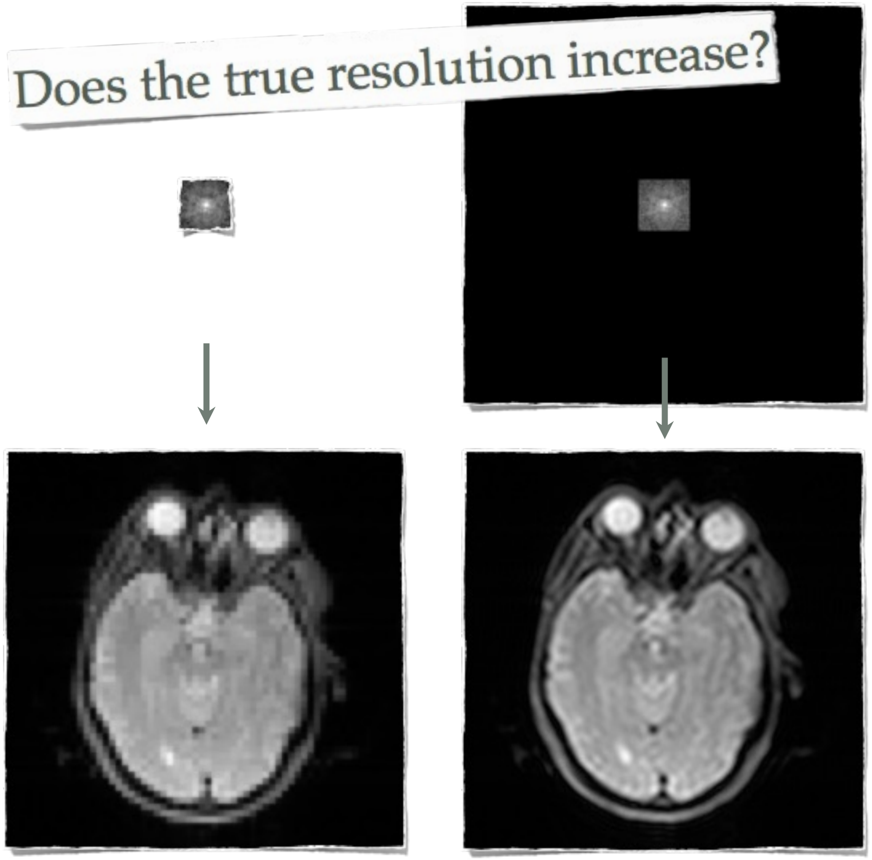
Removal: center-out



k-space size

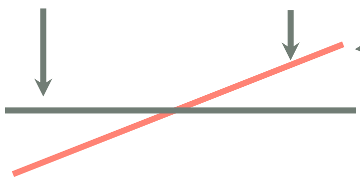


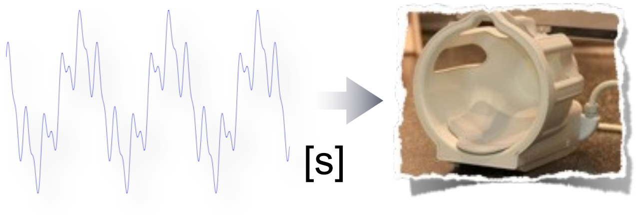
Zero-filling 64x64 -> 512x512

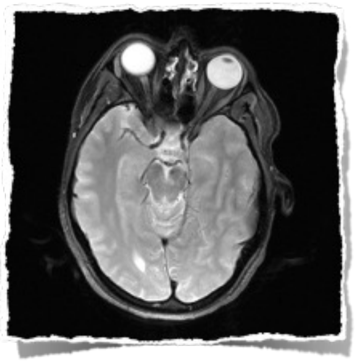


Signal, k-space, imag

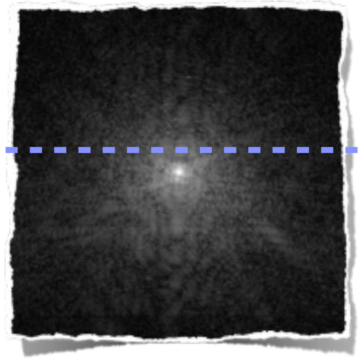
With a gradient G_x
we map frequencies [1/s] to distances [m]





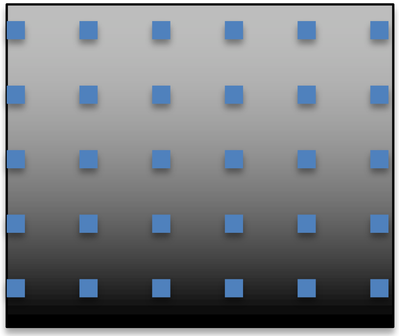


Fourier transform
(units are inverted!)



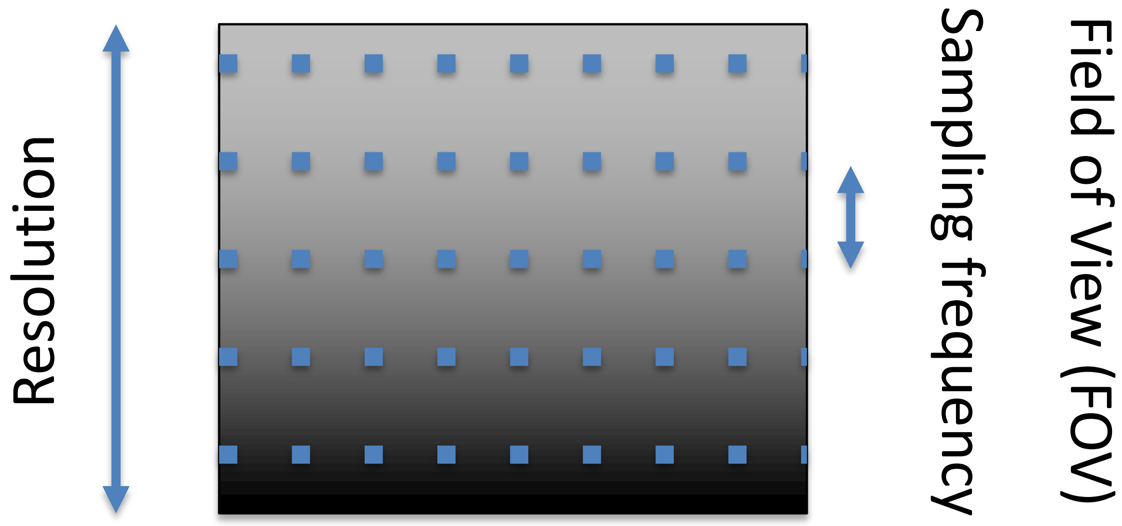
Scan time

- 256 x 256 with TR of 100 ms => 25.6 s scan time for one slice
- Shorten scan time
  - Reduce TR
    - Changes contrast
  - Reduce number of lines in k-space
    - Reduces resolution
    - or reduces Field Of View (FOV)



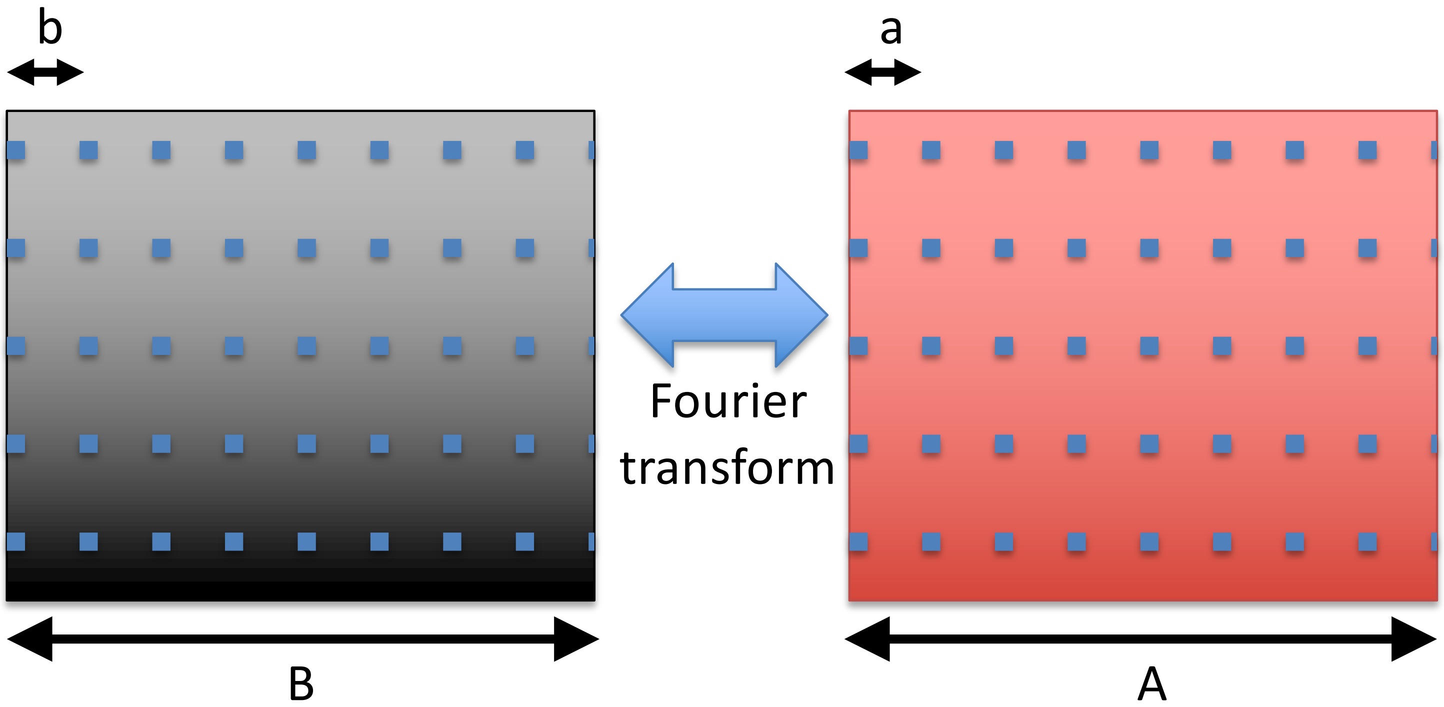
k-space – time domain signal

k-space matrix properties



k-space – time domain signal

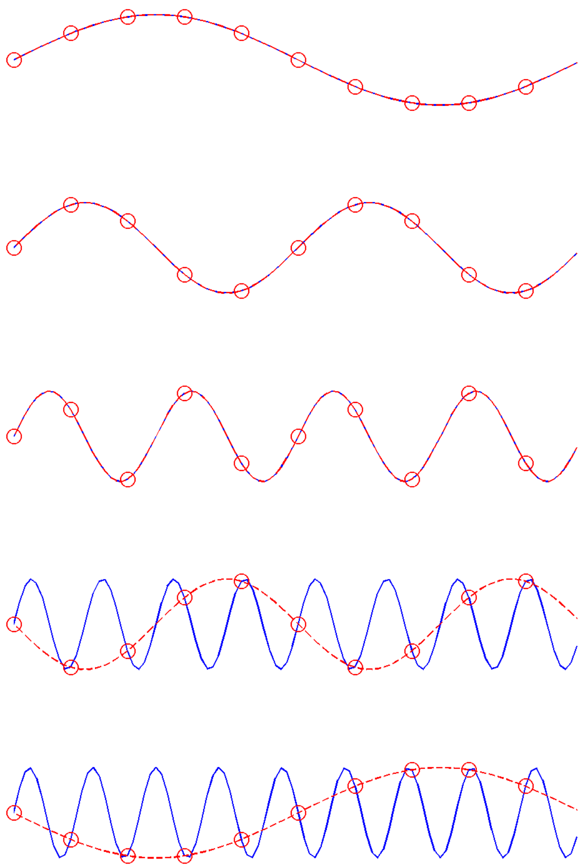
k-space relationships



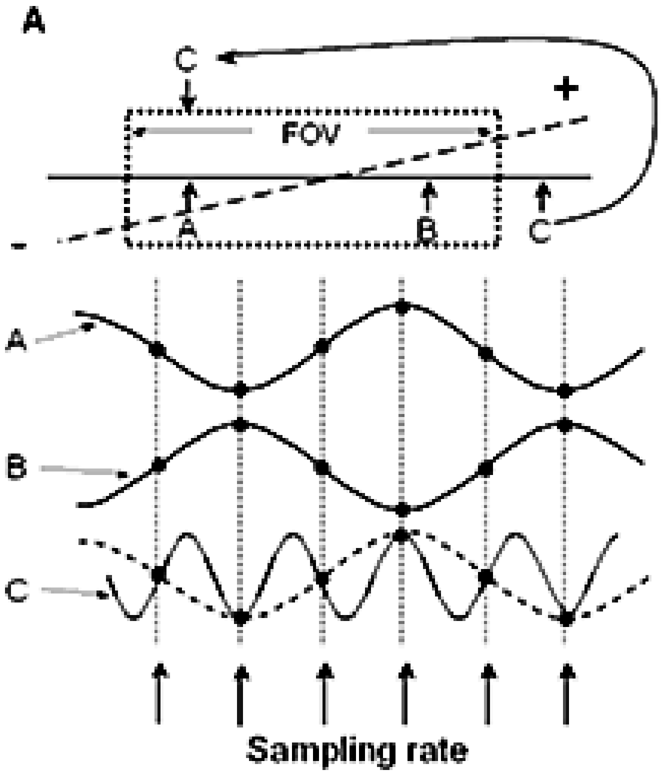
a ~ 1 / B
b ~ 1 / A

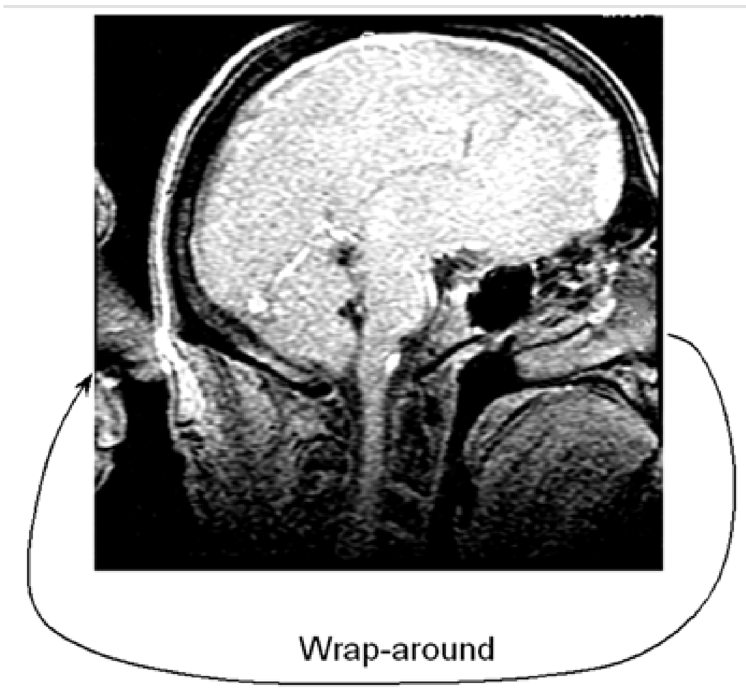
Fold-over artifacts

- Frequency determination relies on sufficient sampling frequency
- Can only resolve up to Nyquist frequency
- Nyquist frequency is half of sampling frequency
  - Corresponds to two samples per period
- If Nyquist sampling criterion is not fulfilled "aliasing" occurs



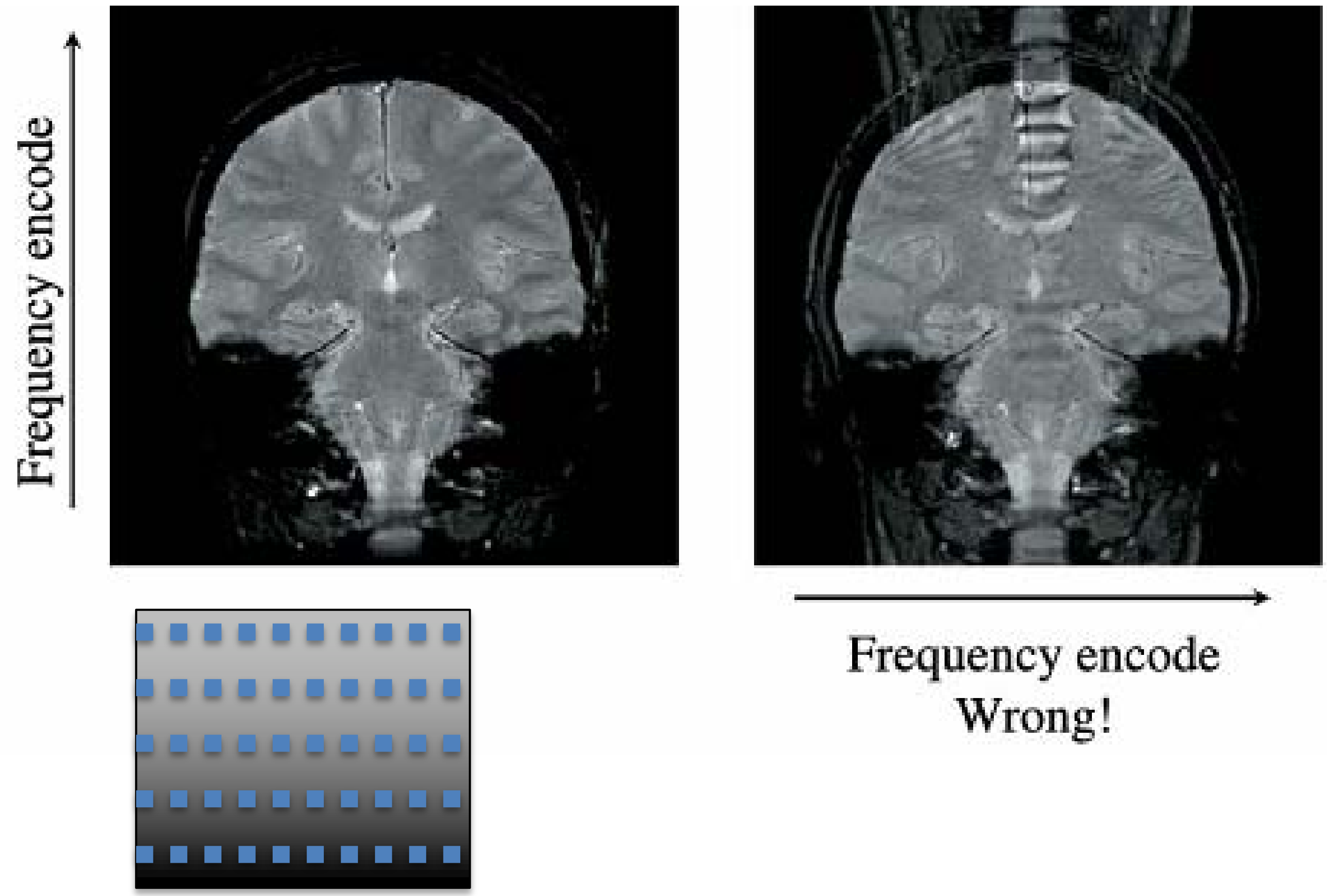
Fold-over artifacts





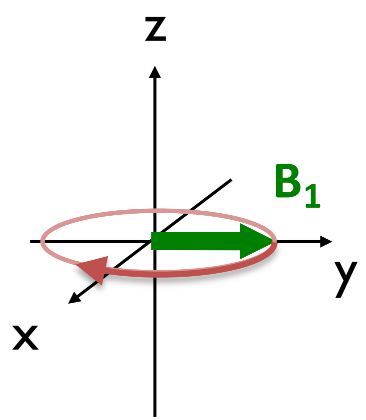
Fold-over artifacts

Importance of readout direction

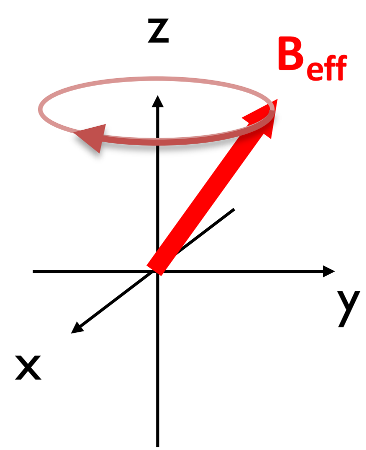


RF pulses

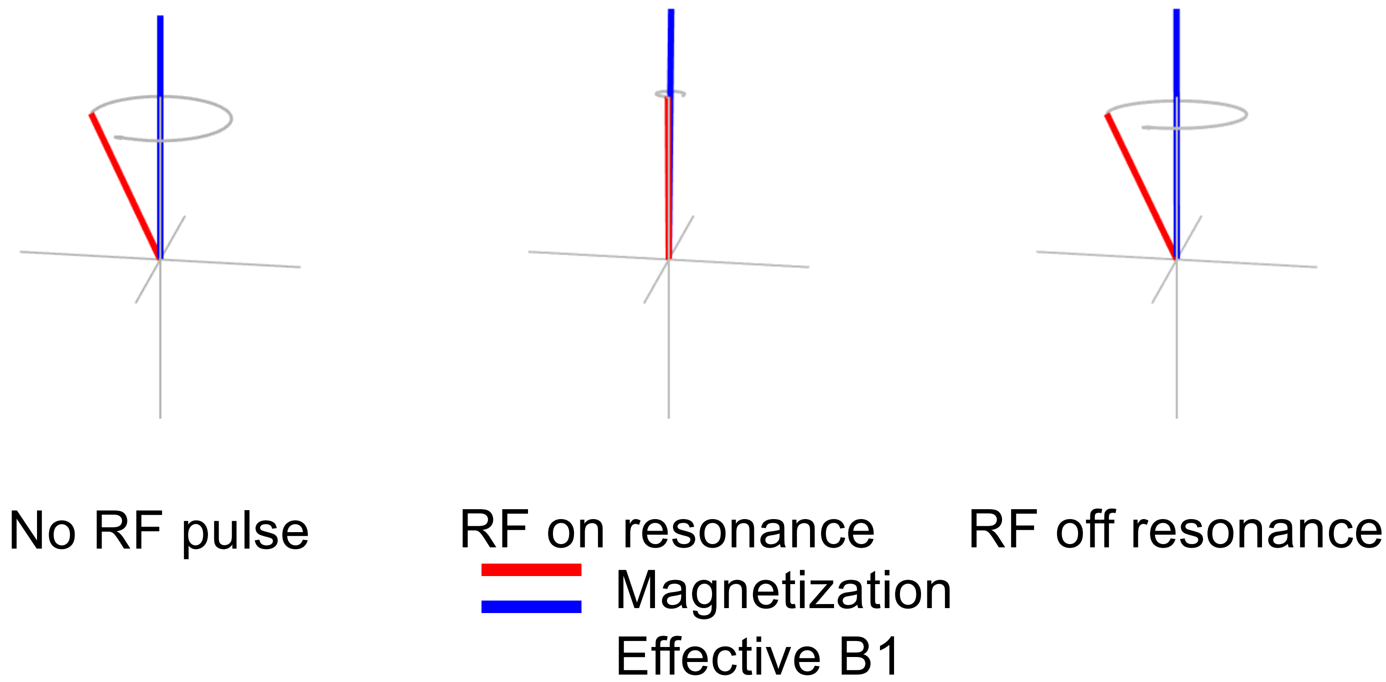
- Rotating magnetic field
  - Not radio waves
  - H field, ideally no E field



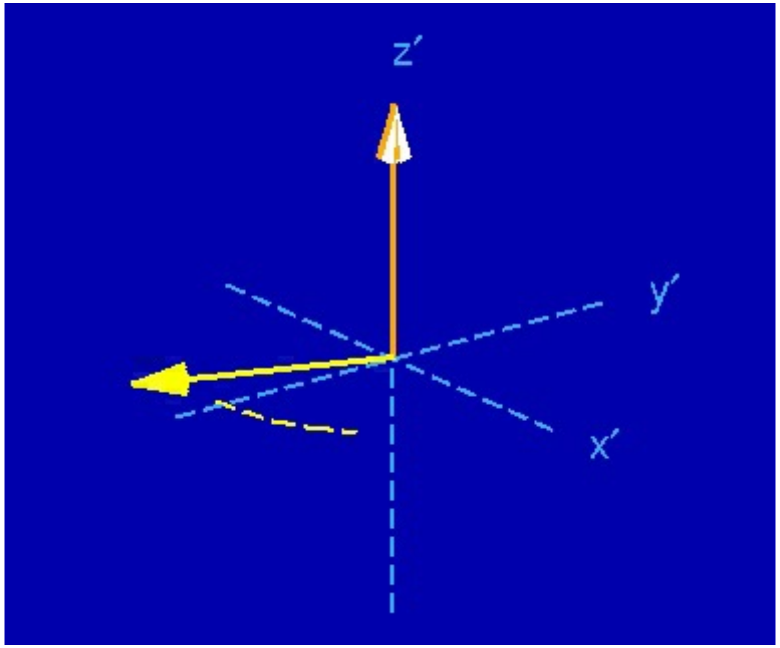
- $B_{eff} = B_0 + B_1$
  - B_0 ~ 1.5-3 T, 0 Hz
  - B_1 ~ 30 μT, 64-128 MHz



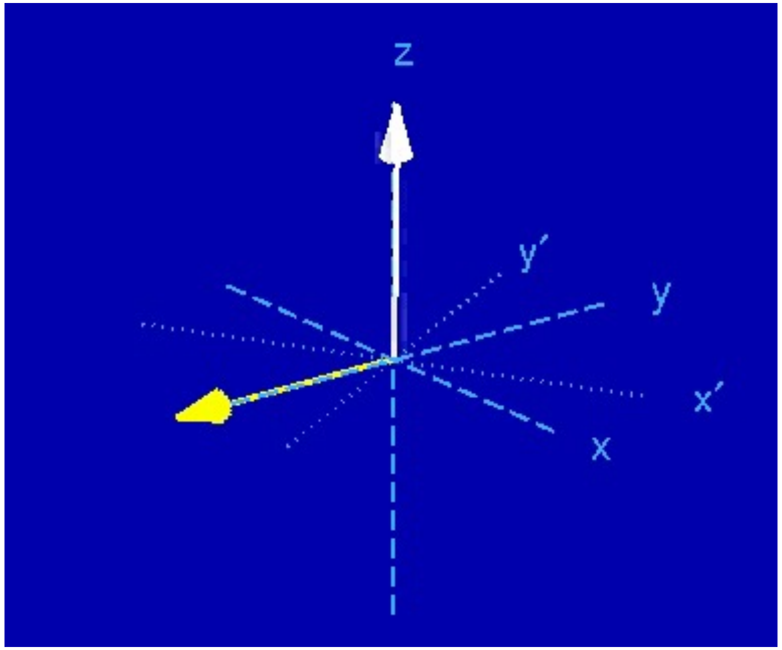
RF pulses - forced precession



Rotating frame of reference



Static frame of reference



Rotating frame of reference

Movies from http://mrsrl.stanford.edu/~brian/intromr/ (Brian Hargreaves)

Simulation

- http://www.drcmr.dk/bloch
- Bloch equations (after Felix Bloch); differential equations in matrix form that describe precession, relaxation and other manipulations

$$\frac{dM_x(t)}{dt} = \gamma(\mathbf{M}(t) \times \mathbf{B}(t))_x - \frac{M_x(t)}{T_2}$$

$$\frac{dM_y(t)}{dt} = \gamma(\mathbf{M}(t) \times \mathbf{B}(t))_y - \frac{M_y(t)}{T_2}$$

$$\frac{dM_z(t)}{dt} = \gamma(\mathbf{M}(t) \times \mathbf{B}(t))_z - \frac{M_z(t) - M_0}{T_1}$$

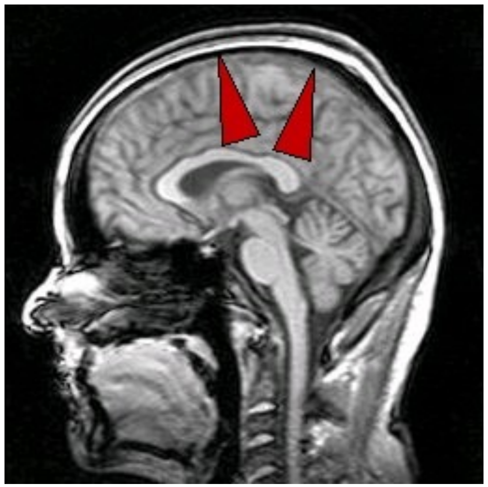
Chemical shift

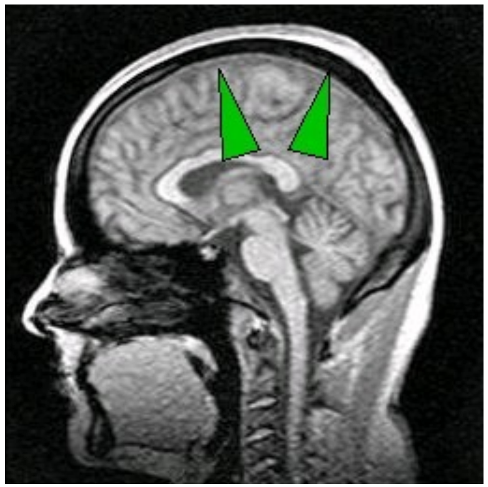
- Electrons shield the magnetic field
- Electron distribution depends on molecular structure
- Local magnetic field will be different for different molecules

- Chemical shift
- Can be used to discriminate different molecules
  - Despite that the signal is from the same nuclei
- Most apparent with water and fat
  - ~150 Hz/T difference

Water-fat shift

- Water-based tissues and fat-based tissues have different chemical shift
- The position encoding relies on frequency differences caused by the gradient
- Additional frequency differences due to chemical shift result in position decoding errors -> fat signal is shifted in the image



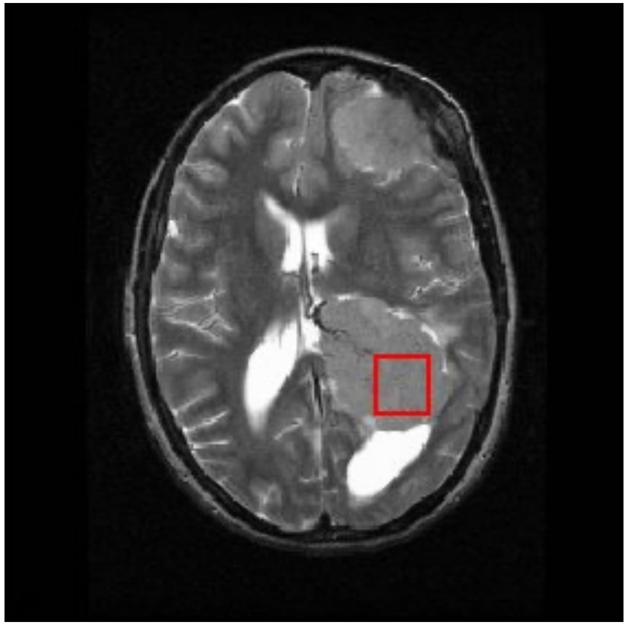


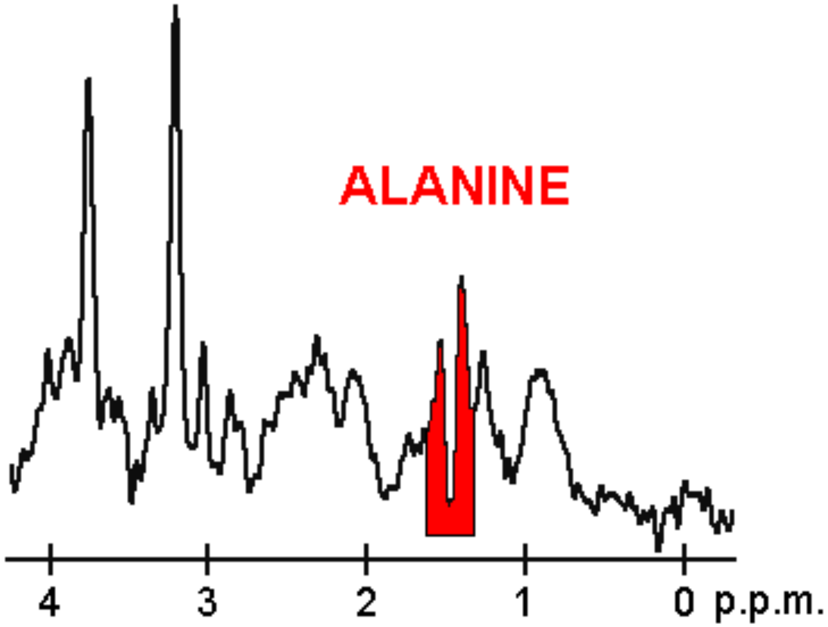
mr-tip.com

MR Spectroscopy

- In spatial encoding, a gradient was applied during readout
  - Reconstruct different positions depending on frequency
- MR spectroscopy
  - Readout without applied gradient
  - Different molecules give signal with different Larmor frequency
  - Similar reconstruction, but with a frequency dimension instead of a spatial dimension
- Spatial localization of MR spectroscopy requires extra effort
  - Single-voxel reasonably simple
  - MR spectroscopic imaging, requires more acquisition time

MR Spectroscopy



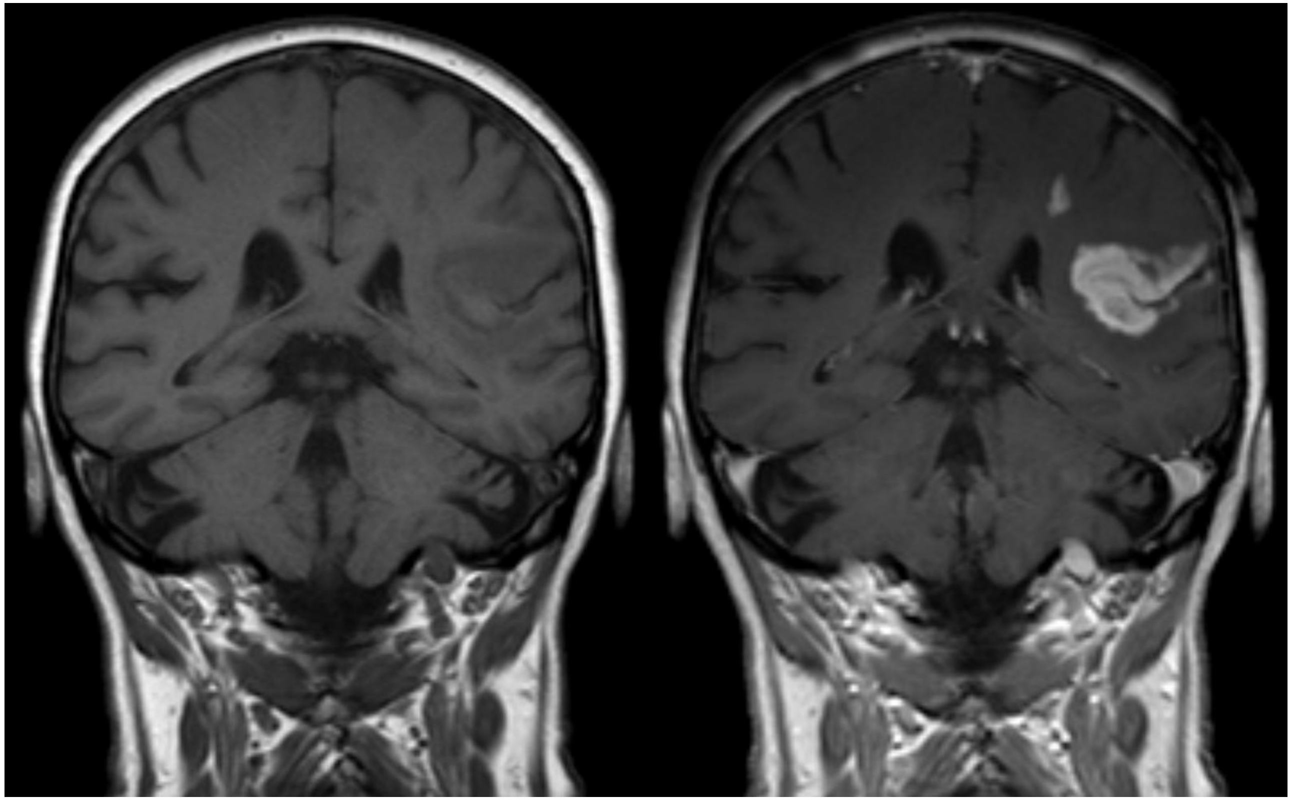


The University of Hull Centre for Magnetic Resonance Investigations (http://www.hull.ac.uk/mri).

Contrast agents

- Most MRI contrast agents don't give signal themselves
- Affects T_1 and T_2
- Lower dose than X-ray / CT contrast agents
- Gadolinium-based
  - Paramagnetic
  - Rare-earth metal
  - Toxic, but chelated in practice
  - Reduces T_1
- Iron oxide
  - Reduces T_2

Contrast agents



Defect of the blood-brain barrier after stroke shown in MRI. T1-weighted images, left image without right image with contrast medium administration.

The insides of an MRI system

Superconducting magnet,
cooling system (liquid N, He)

Gradient coils
sealed in epoxy

Integrated
RF coil

Electronics, cooling systems,
and more

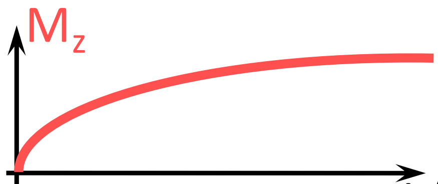
Summary

- Some nuclei have spin
  - Split energy level in presence of magnetic field
  - Energy difference results in a Larmor frequency precession
  - Larmor frequency *proportional to magnetic field* (^{1}H, 1.5 T => 64 MHz)
  - Population difference very small (10^{-5})

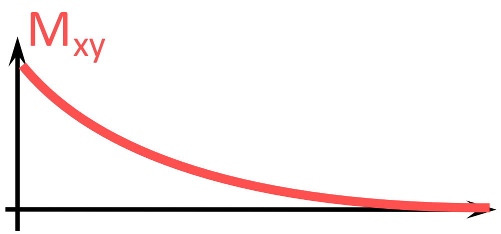
- Net magnetization vector M
  - A group of spins have a net magnetization vector
  - Can be measured when flipped to the transversal plane (M_{xy})
    - Complex signal with magnitude and phase
  - Relaxes back to thermal equilibrium (M_0 along B_0 direction)

Summary, relaxation

- T_1: Longitudinal relaxation, the net magnetization vector returns to thermal equilibrium along B_0

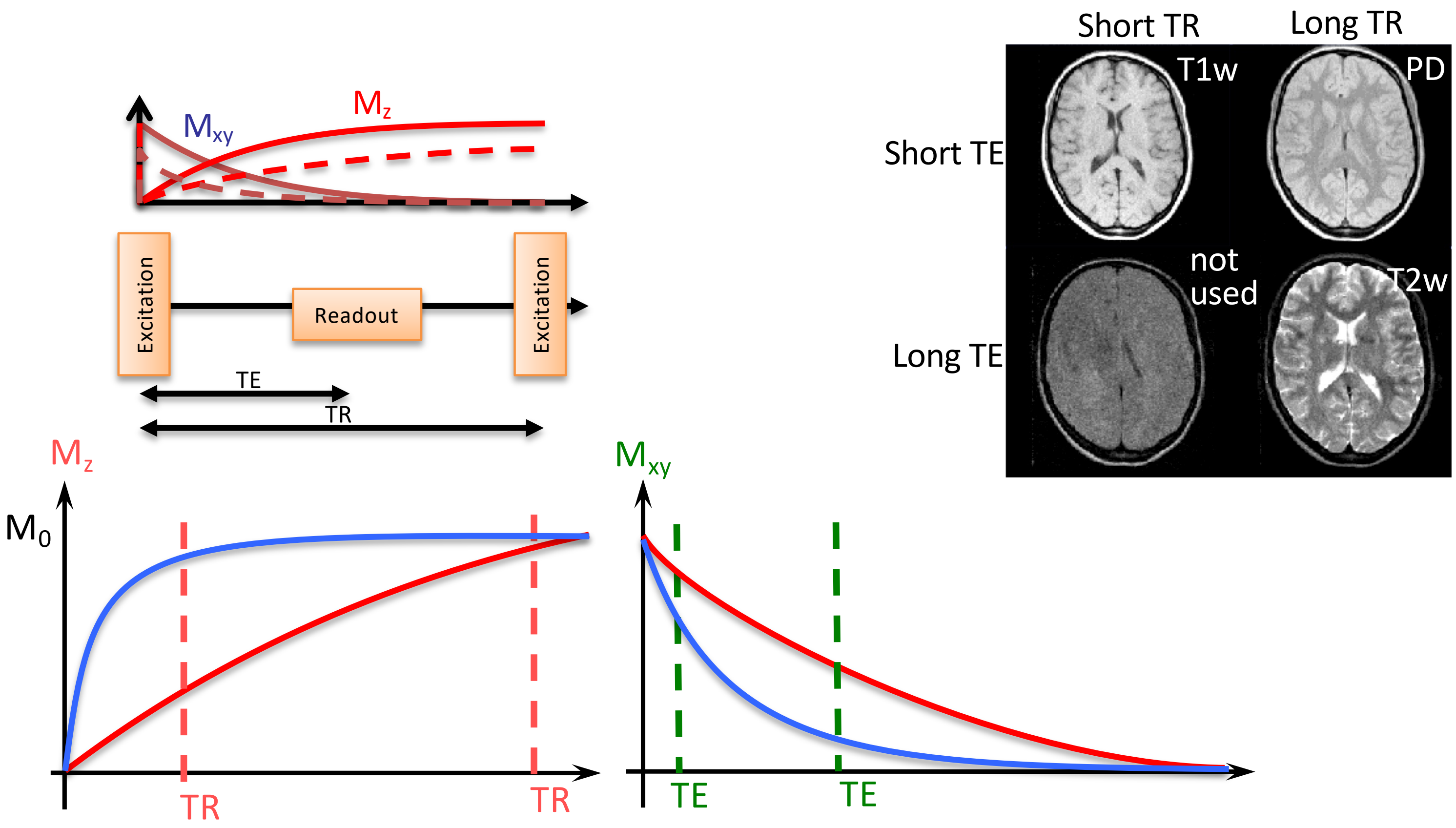


- T_2 and T_2^*: due to varying Larmor frequency, the transverse component (Mxy) of the net magnetization dephases and signal vanishes
  - T_2 is irreversible because they are random in time
  - T_2^* is the apparent T_2 when not using 180 degree refocusing pulses
  - The temporally static variations are refocused using a 180 degree pulse
- $T_2^* < T_2 < T_1$

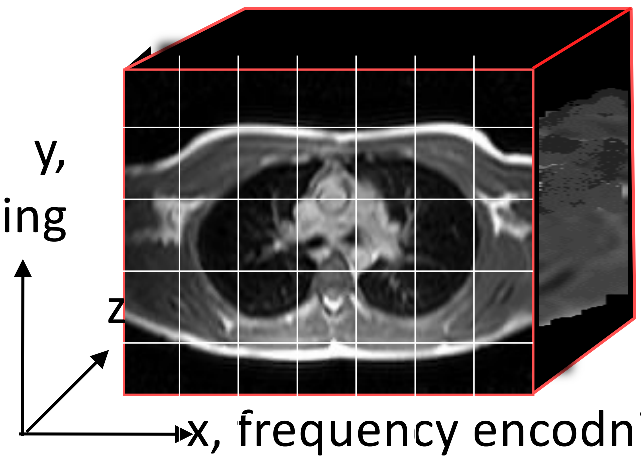


Summary, image contrast

- By choosing TE and TR appropriately, a desired image contrast can be obtained



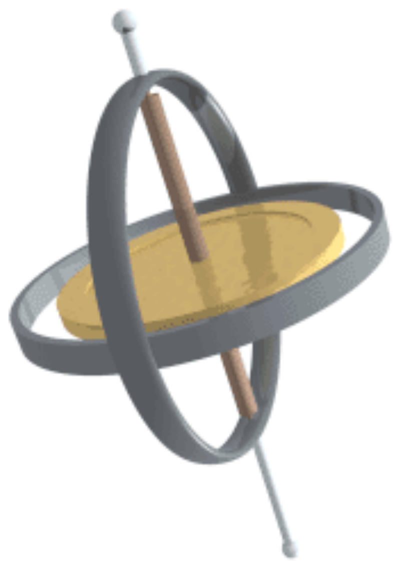
Summary, image localization



- Slice is selected during excitation
  - Gradient during RF pulse
  - Only spins within the bandwidth are excited
- Frequency encoding (x)
  - Gradient during signal reception (one TR)
  - Signal with frequency depending on position is emitted
  - Fourier transform (“spectrum analyzer”)
- Phase encoding (y)
  - Gradient lobe with varying strength before readout
  - Needs to be repeated for all combinations (N x TR)
  - Fourier transform for reconstruction

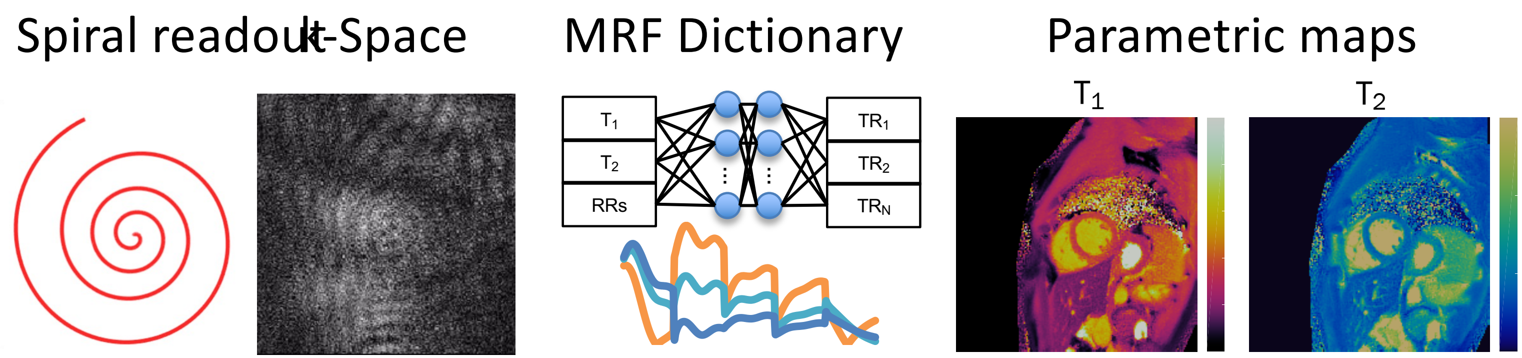
Acknowledgements

- Karin Markenroth Bloch, Lund (Slides)
- Andreas Sigfridsson, KI (Slides)



The next step

- Course in Magnetic Resonance Imaging
  - HL2011, autumn, 6 HP
- Master projects
  - MR Fingerprinting, neural network reconstructions alexander.fyrdahl@ki.se



Book chapter:
Fyrdahl A et. al. Magnetic Resonance Fingerprinting: The Role of Artificial Intelliger