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(Nuclear) Magnetic Resonance Imaging

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Operační program Výzkum, vývoj a vzdělávání

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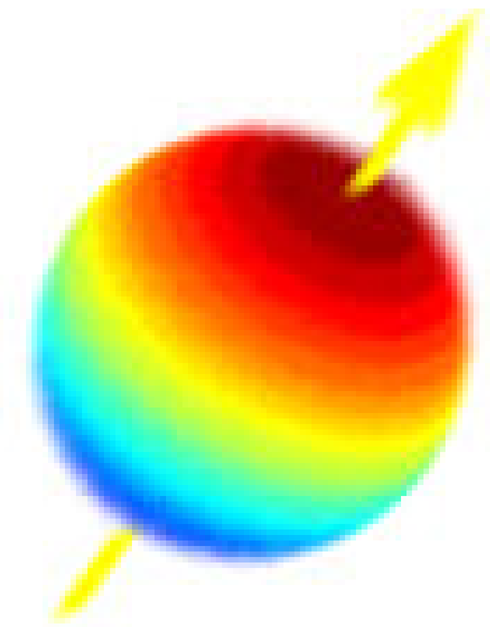
Magnetic resonance imaging

Spin

- **Proton and neutron** spins are known as nuclear spins.

- An unpaired component has a spin of ½ and two particles with opposite spin cancel one another.

- **In NMR it is the unpaired nuclear spins that produce a signal in a magnetic field.**



Common Nuclei with NMR properties

•Criteria:

Must have ODD number of protons or ODD number of neutrons.

Reason?

It is impossible to arrange these nuclei so that a zero net angular momentum is achieved. Thus, these nuclei will display a magnetic moment and angular momentum necessary for NMR.

Examples:

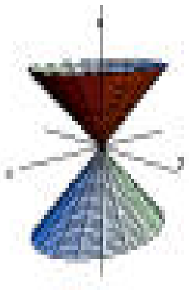
^{1}H, ^{13}C, ^{19}F, ^{23}N, and ^{31}P with gyromagnetic ratio of **42.58**, 10.71, 40.08, 11.27 and 17.25 **MHz/T**.

Since hydrogen protons are the most abundant in human body, we use ^{1}H MRI most of the time.

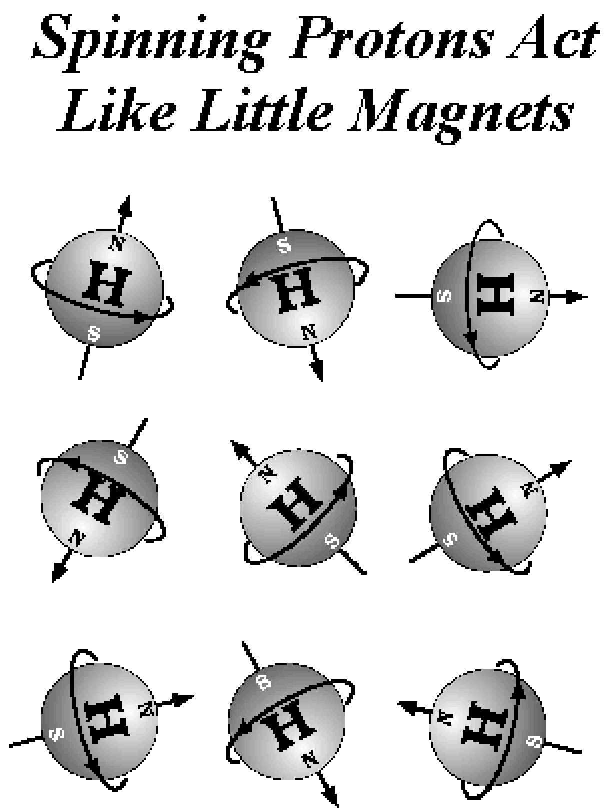
Hydrogen

- Human body is mainly composed of fat and water, which makes the human body composed of about 63% hydrogen.

- Why Are Protons Important to MRI?
  - positively charged
  - spin about a central axis
  - a moving (spinning) charge creates a magnetic field.
  - the straight arrow (vector) indicates the direction of the magnetic field.

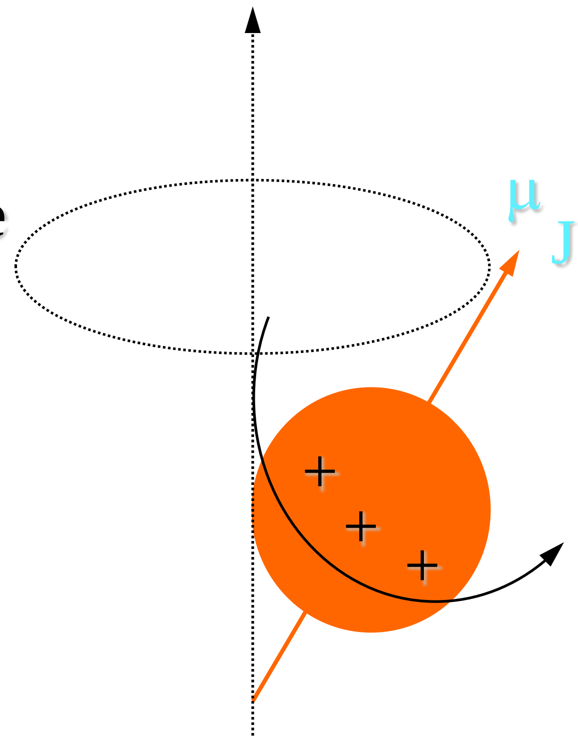


Protons of Hydrogen



Single Proton

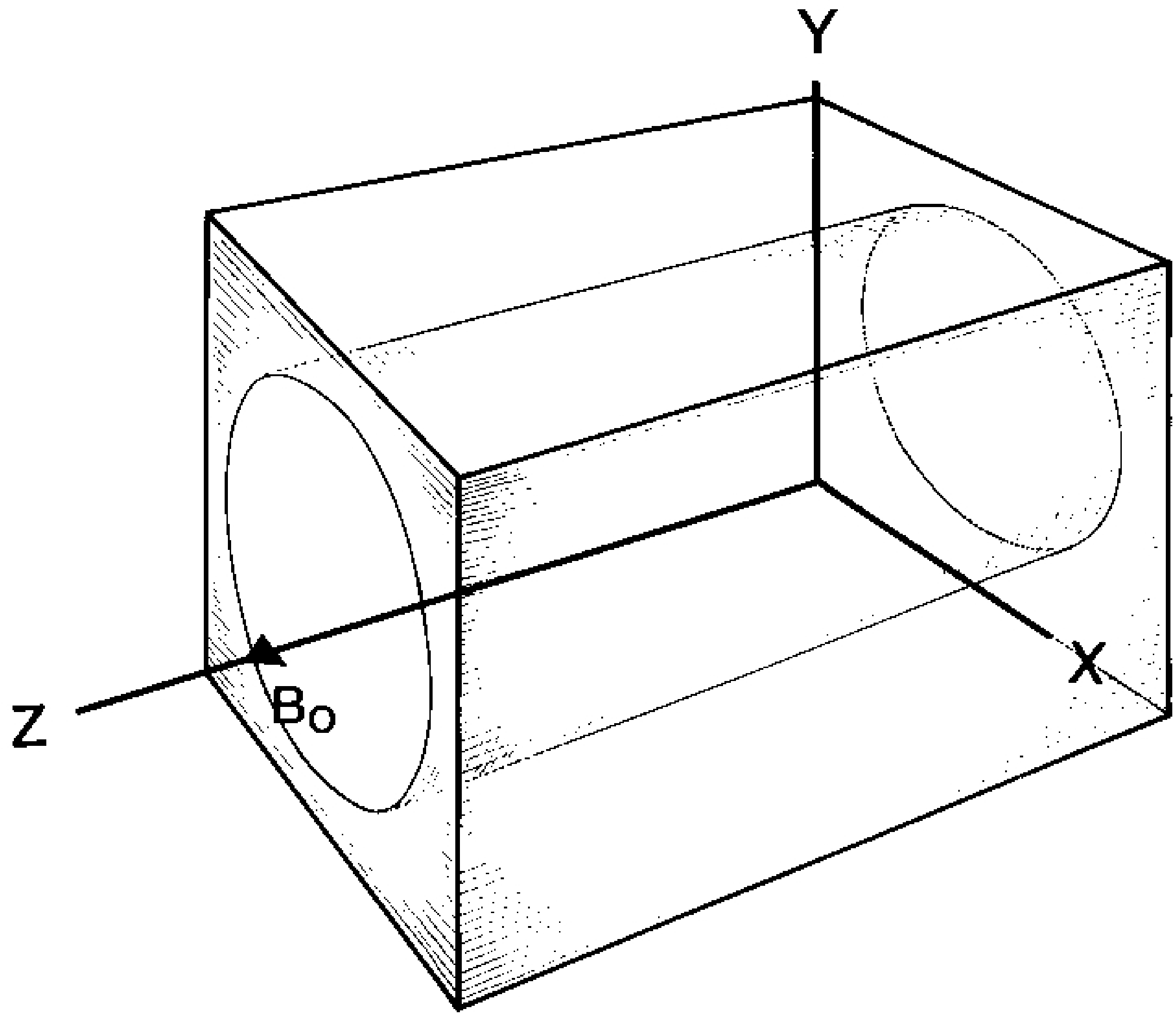
There is electric charge on the surface of the proton, thus creating a small current loop and generating magnetic moment μ.



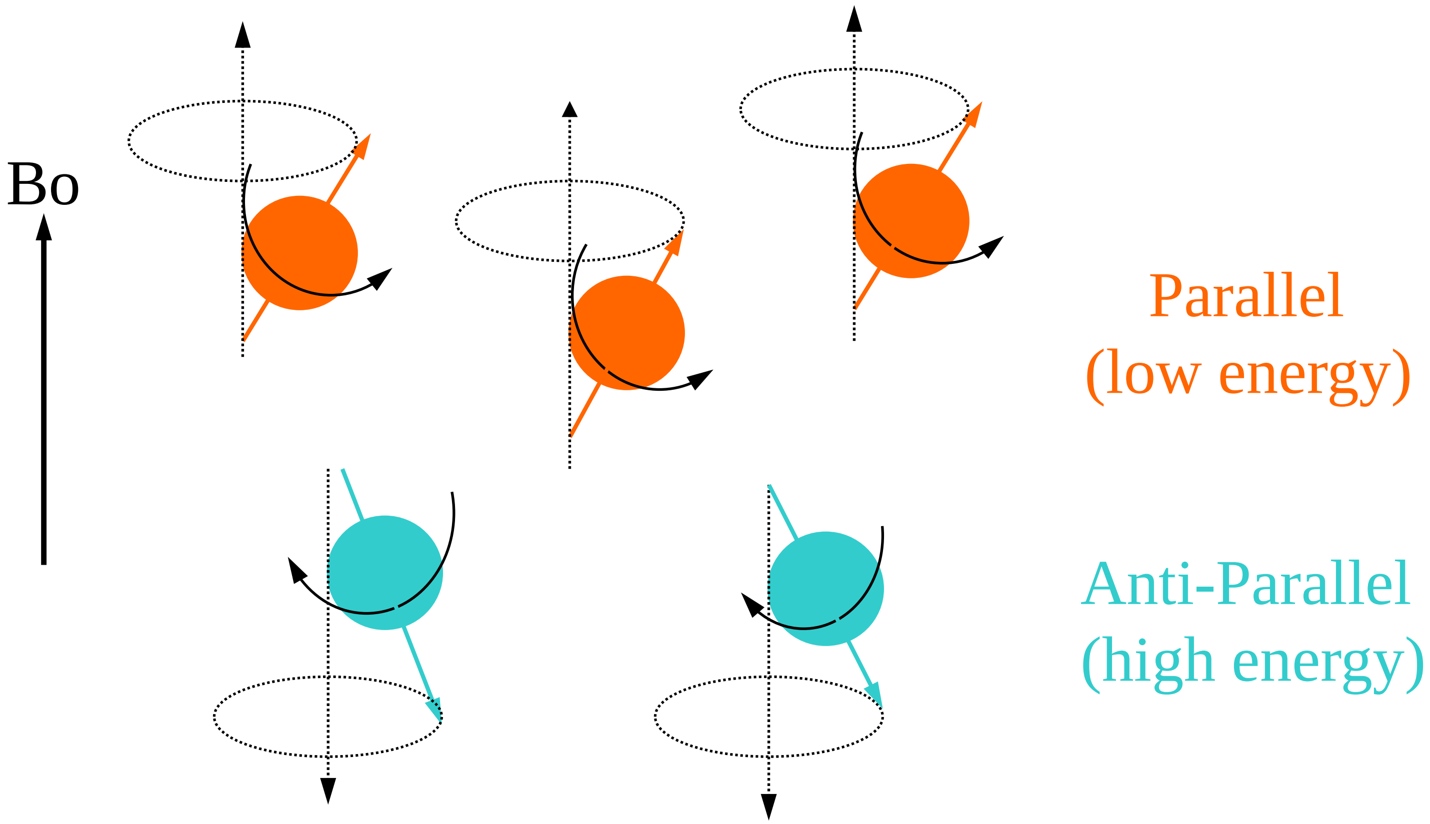
The proton also has mass which generates an angular momentum J when it is spinning.

Thus proton "magnet" differs from a magnetic bar in that it also possesses angular momentum caused by spinning.

Coordinates



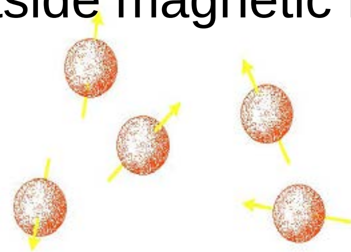
Protons in Magnetic Field $\mathbf{B_0}$



Spinning protons in a magnetic field will assume two states.
If the temperature is 0° K, all spins will occupy the lower energy state.

Protons align with Field $\mathbf{B_0}$

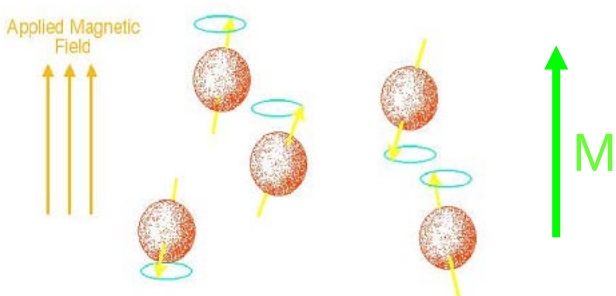
Outside magnetic field

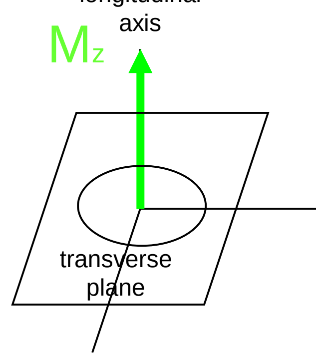


randomly oriented

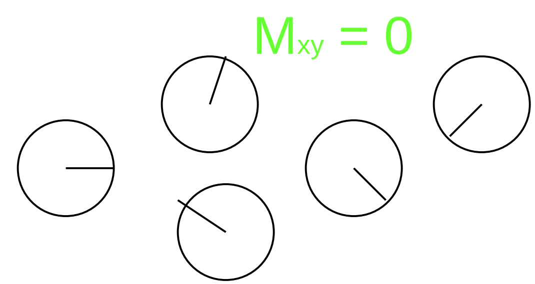
- spins tend to align parallel or anti-parallel to B_0
- net magnetization (M) along B_0
- spins precess with random phase
- no net magnetization in transverse plane

Inside magnetic field





Longitudinal magnetization



Transverse magnetization

Larmor Equation

- Frequency (rate) of precession is proportional to the strength of magnetic field

$$\omega = \gamma * B$$

Larmor Frequency II

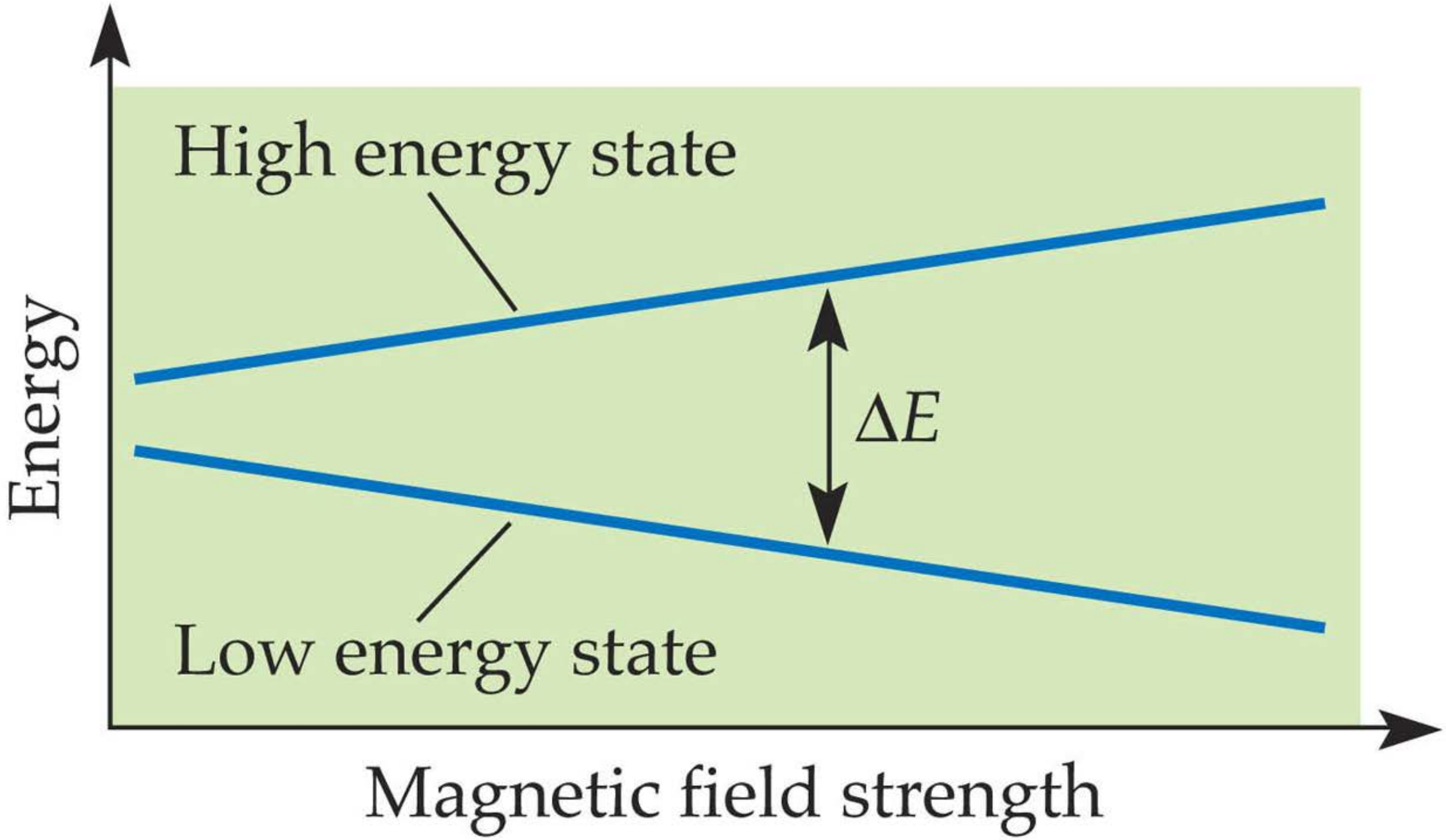
$$\omega = \gamma * B$$

ω= 63MHz If B = 1.5T

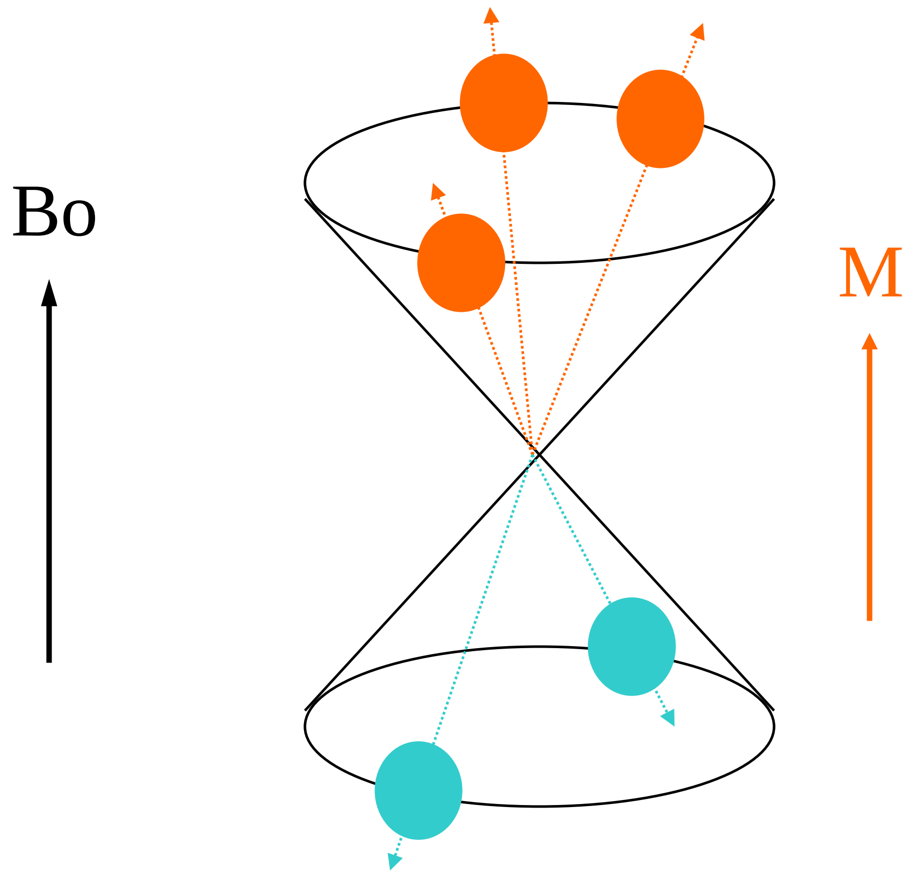
ω= 2 * 63MHz If B = 3.0T

= 126MHz

Zeeman Effect



Net Magnetization

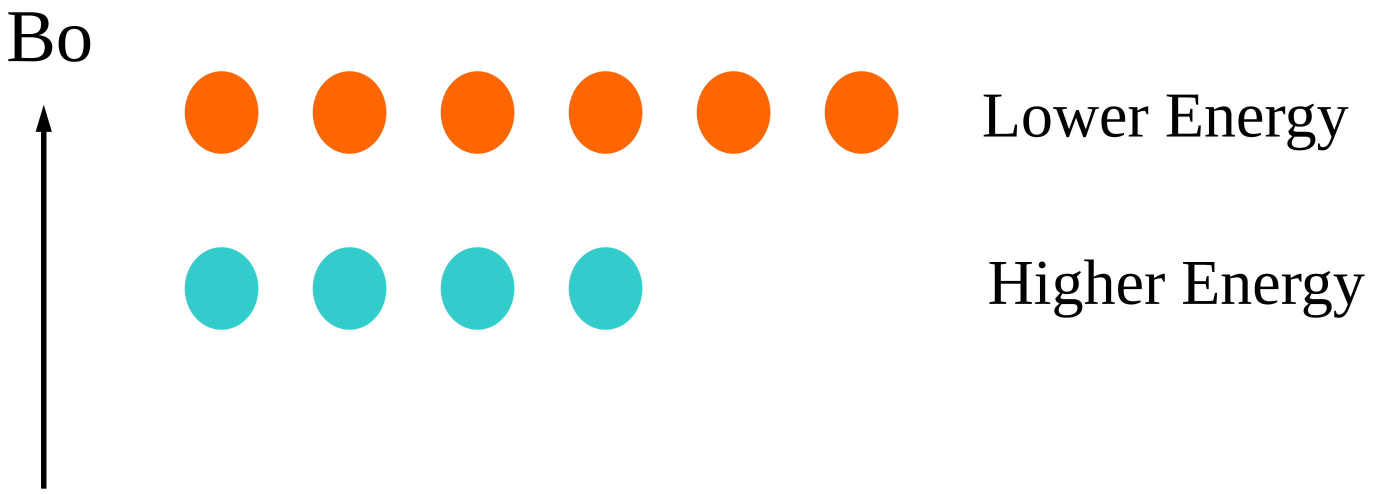


$$M = c\frac{B_o}{T}$$

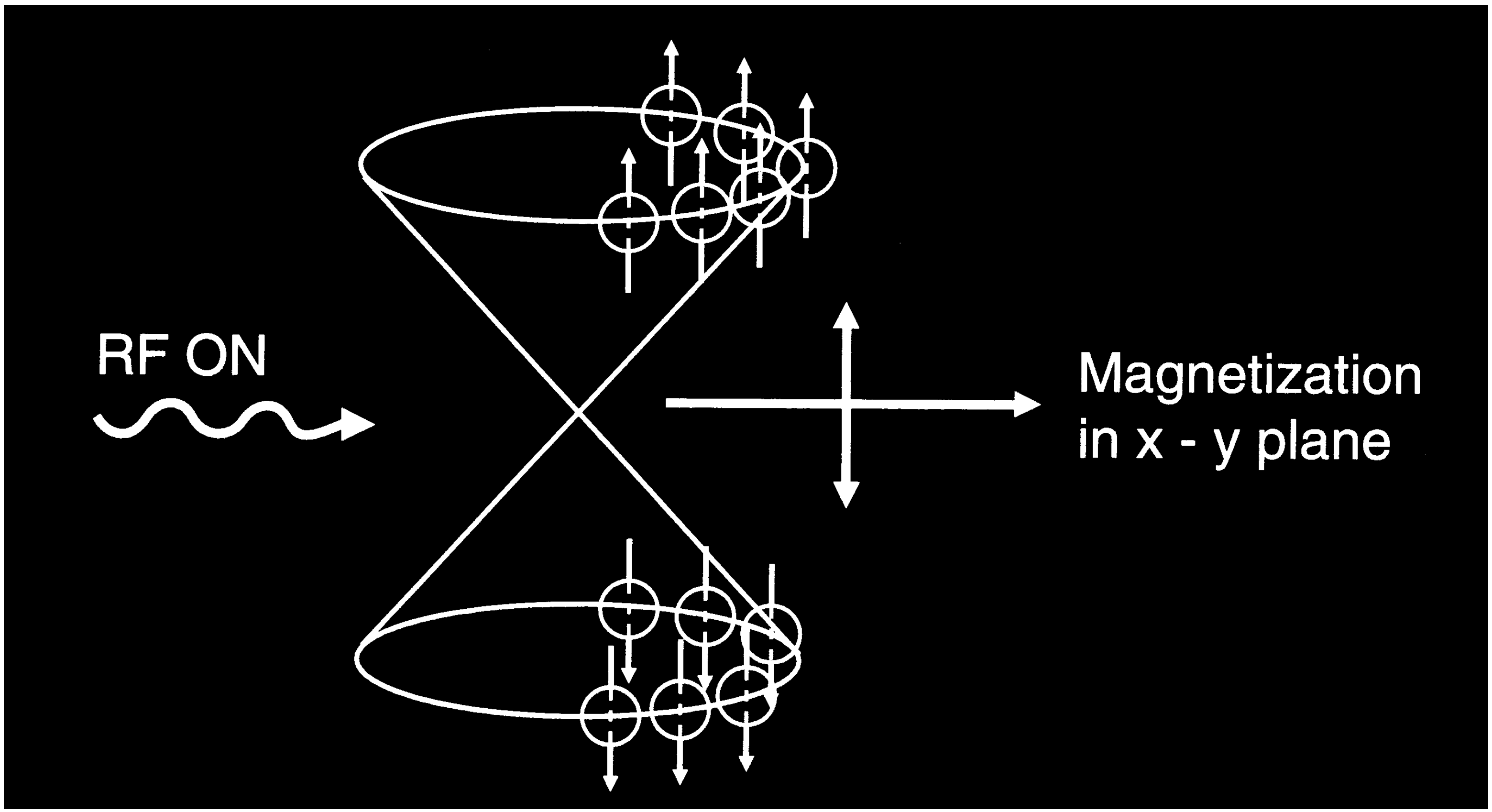
The Boltzman equation describes the population ratio of the two energy states:

$$N^-/N^+ = e^{-\Delta E/kT}$$

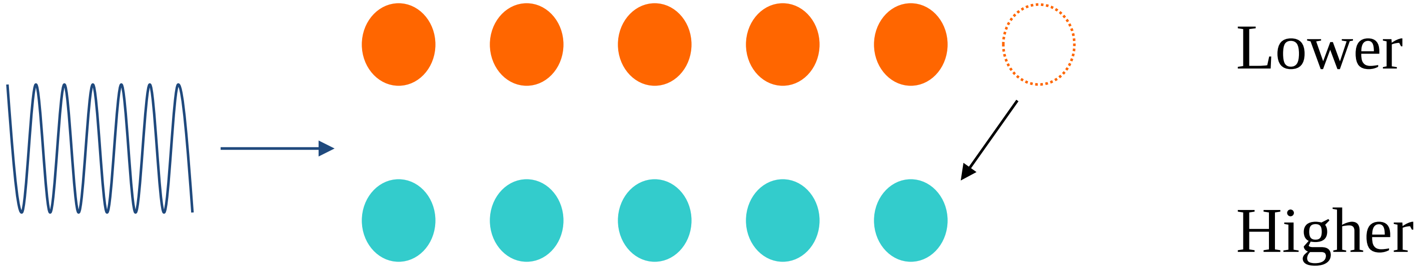
Spin System Before Irradiation



Spin Excitation
Tipping Protons into the Imaging Plane



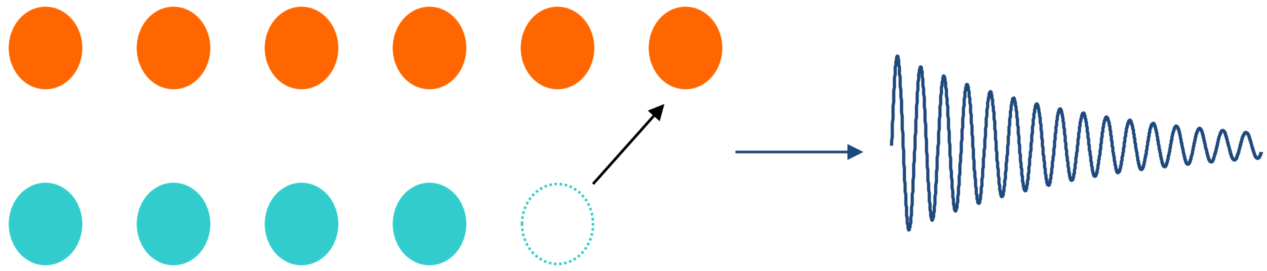
The Effect of Irradiation to the Spin System



The MRI Measurement
(Up to this point)

- In the presence of the static magnetic field
  - Protons align with the field
  - Protons precess about the magnetic field
- Briefly turn on RF pulse
  - Provides energy to tip the protons at least partially into the imaging plane
- ***What happens to the protons next?***

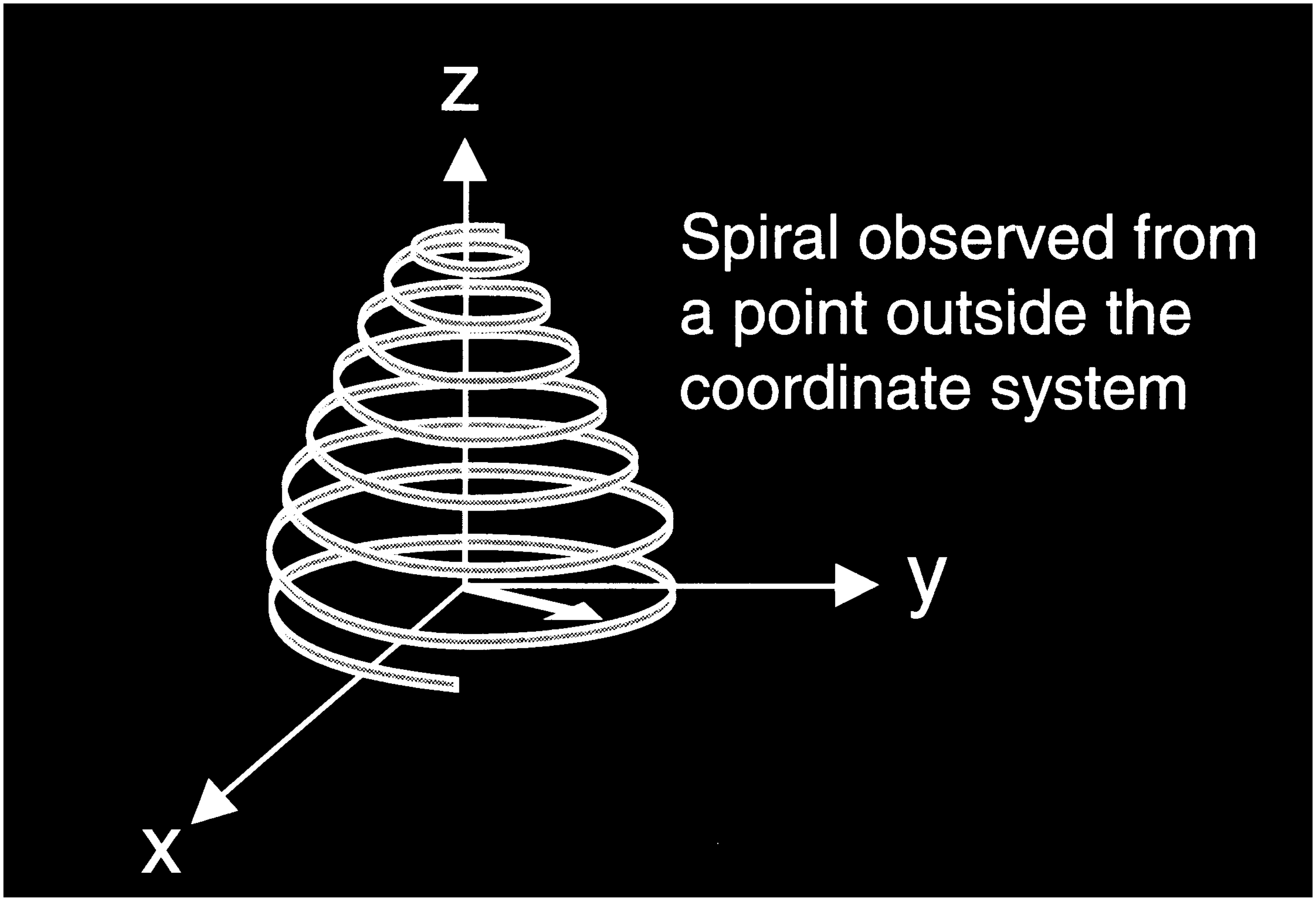
Spin System After Irradiation



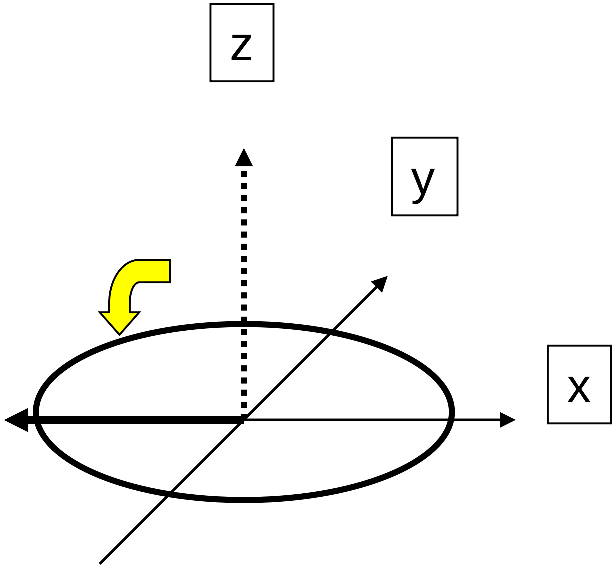
Types of Relaxation

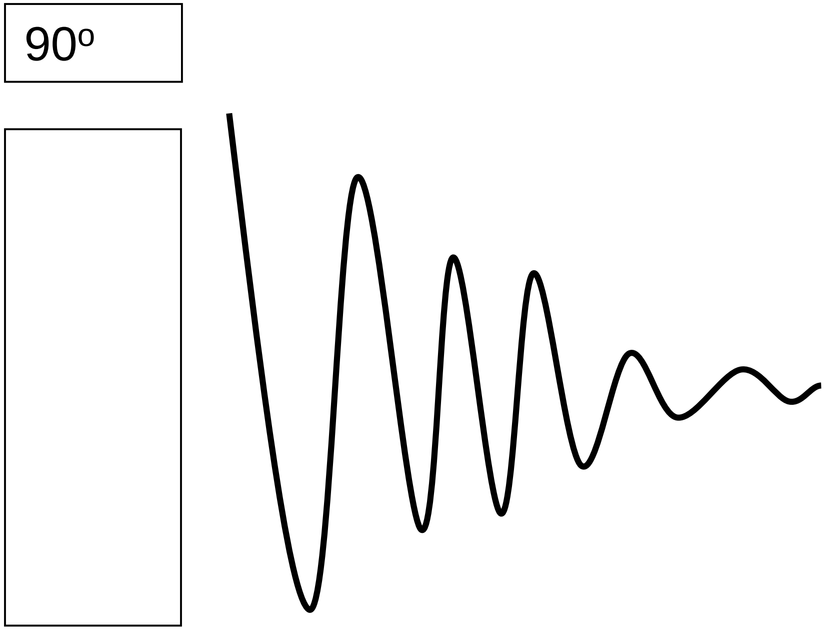
- *Longitudinal* – precessing protons are pulled back into alignment with main magnetic field of the scanner (B_o) reducing size of the magnetic moment vector in the x-y plane
- *Transverse* – precessing protons become out of phase leading to a drop in the net magnetic moment vector (M_o)
- Transverse relaxation occurs much faster than Longitudinal relaxation
- *Tissue contrast is determined by differences in these two types of relaxation*

Longitudinal Relaxation in 3D



Longitudinal Relaxation in 2D

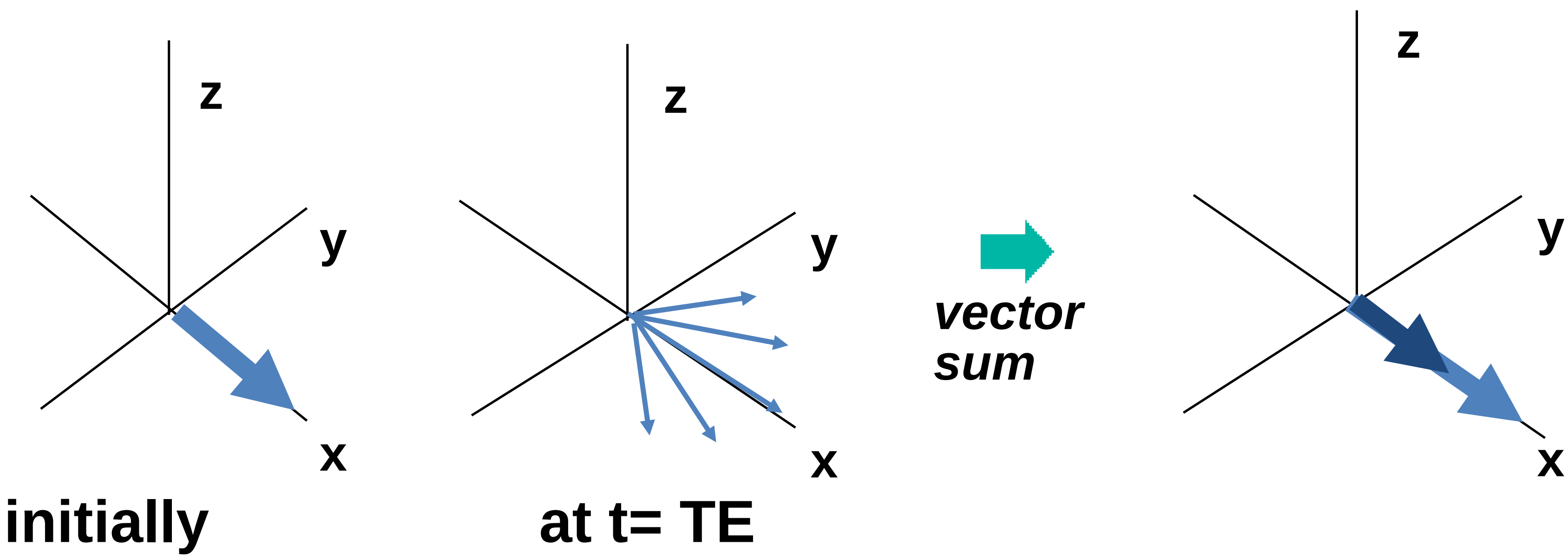




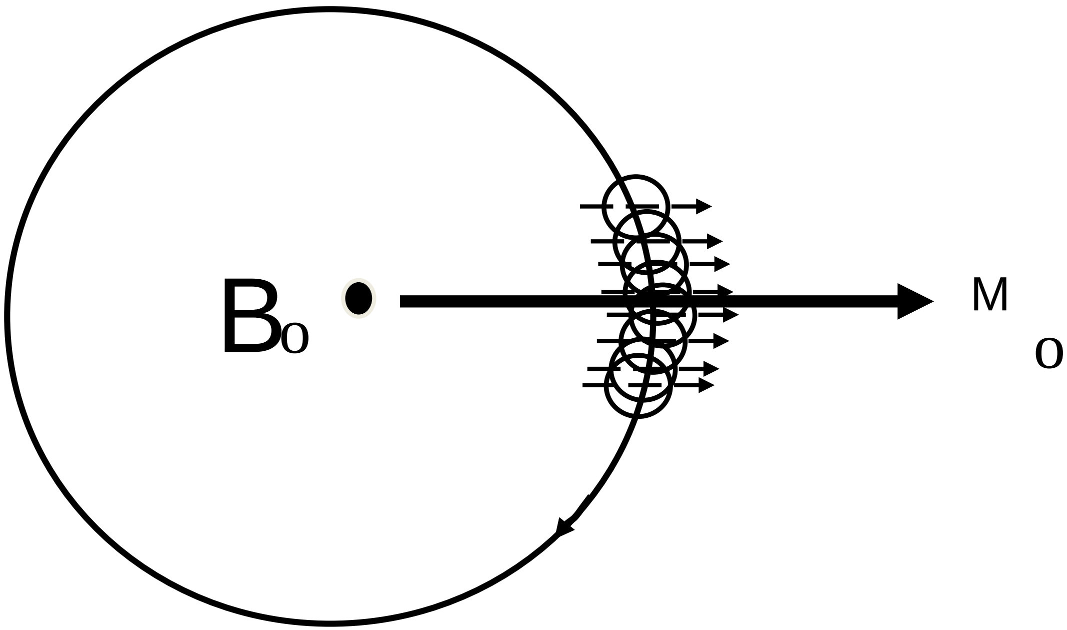
Free Induction Decay

Transverse Relaxation

Wait time <u>TE</u> after excitation before measuring M when the shorter T2 spins have dephased.

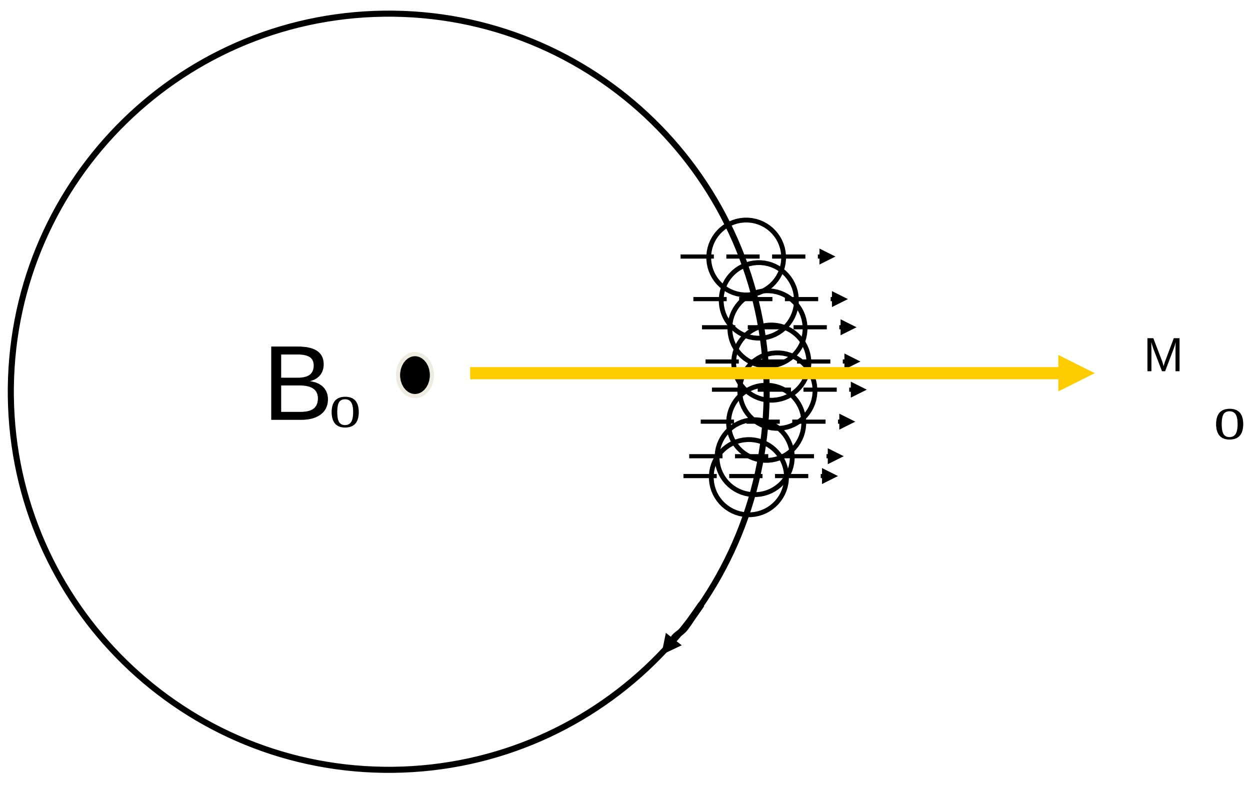


Magnetic Moment Measurable After RF Pulse

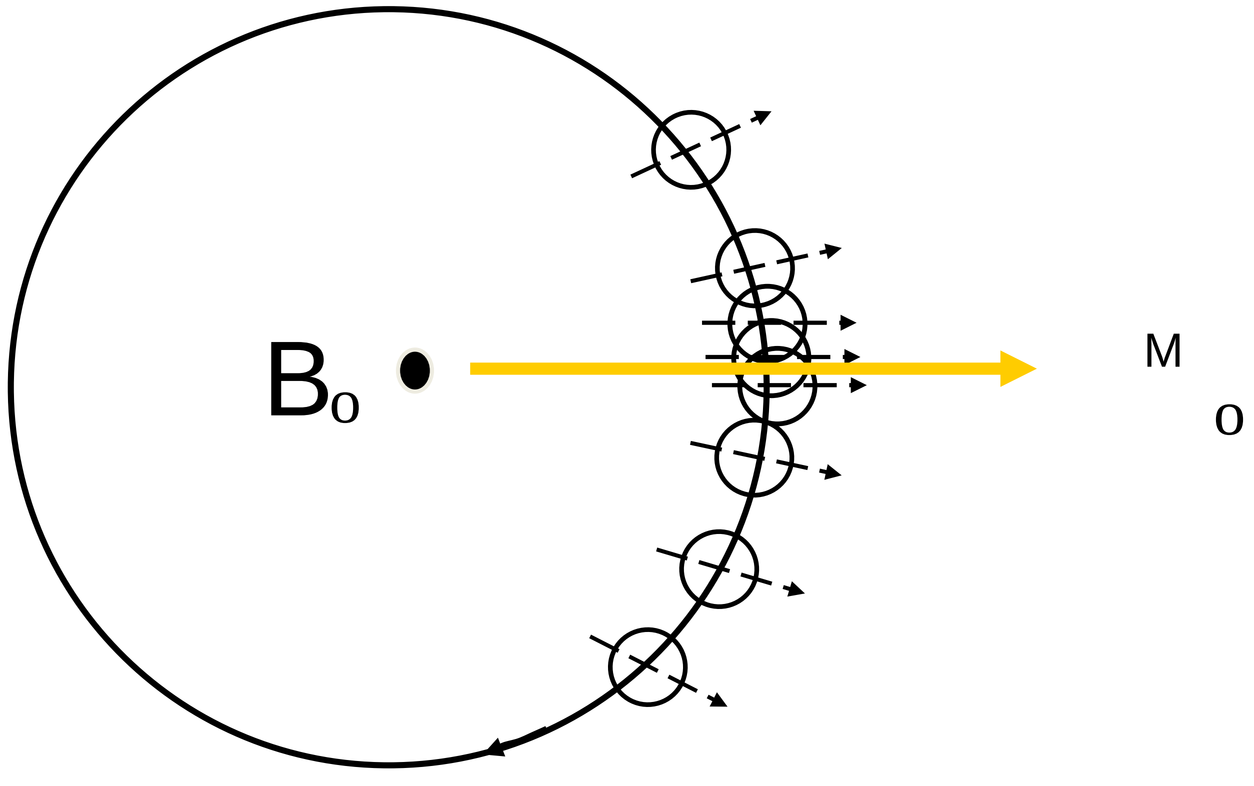


Following an RF pulse the protons precess in the x-y plane

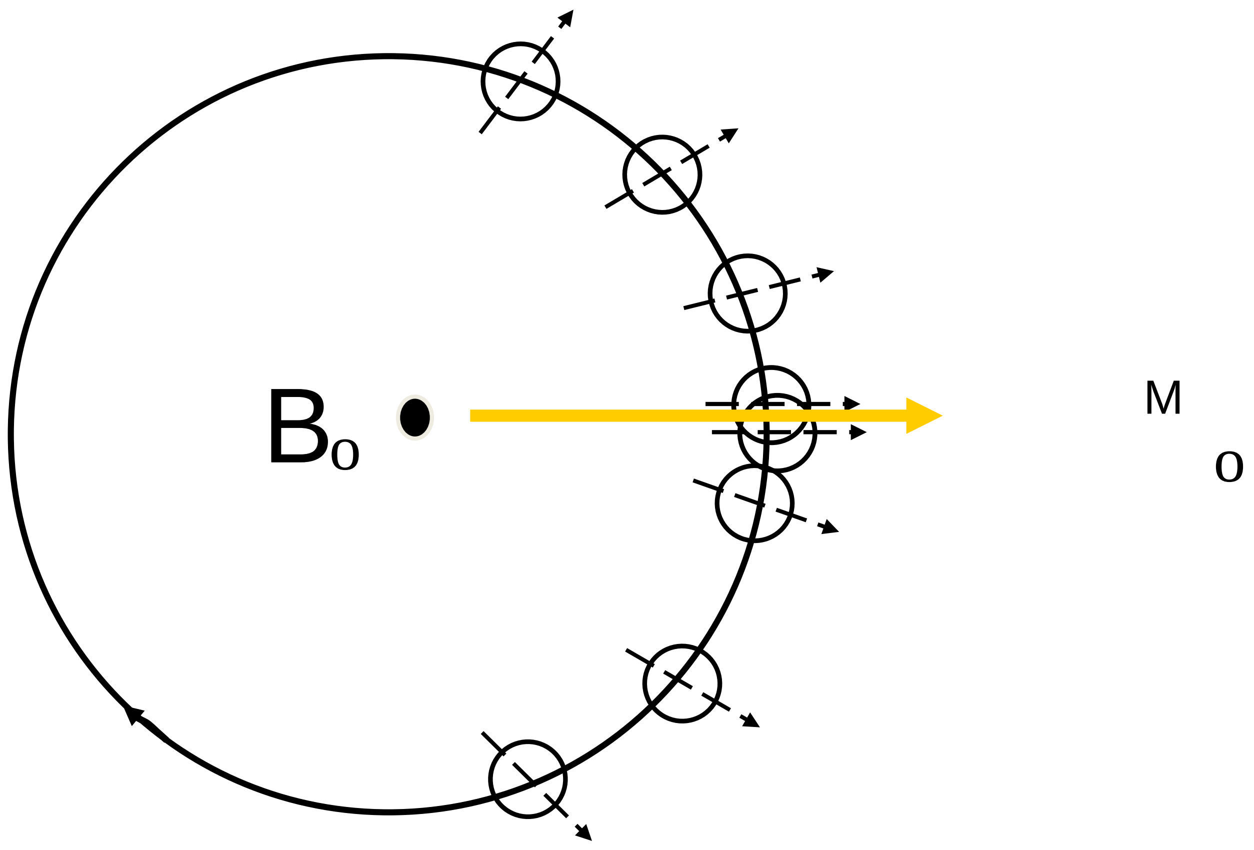
Transverse Relaxation



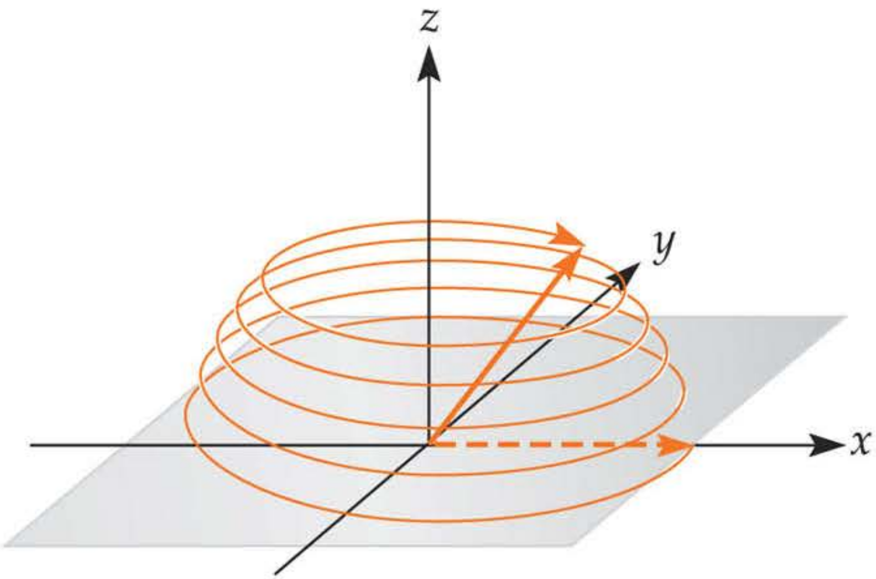
Transverse Relaxation



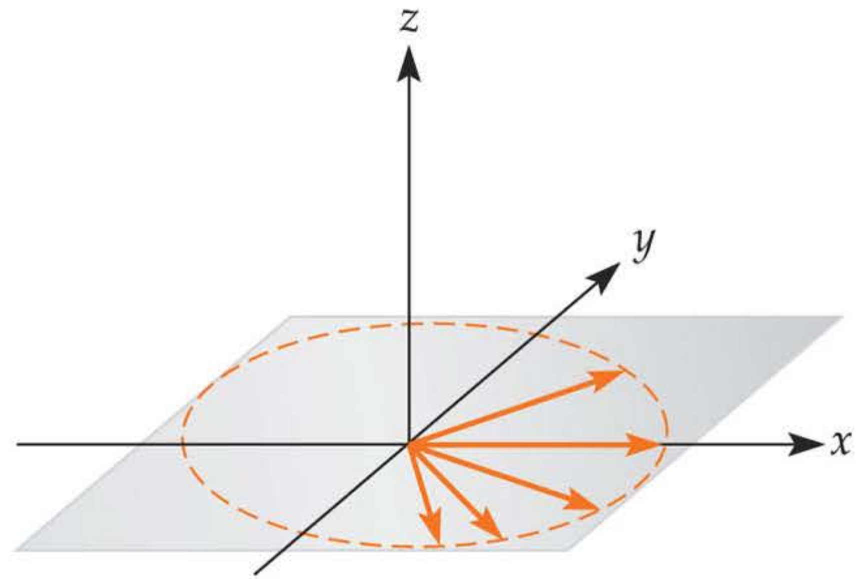
Transverse Relaxation



T_1 and T_2 relaxation

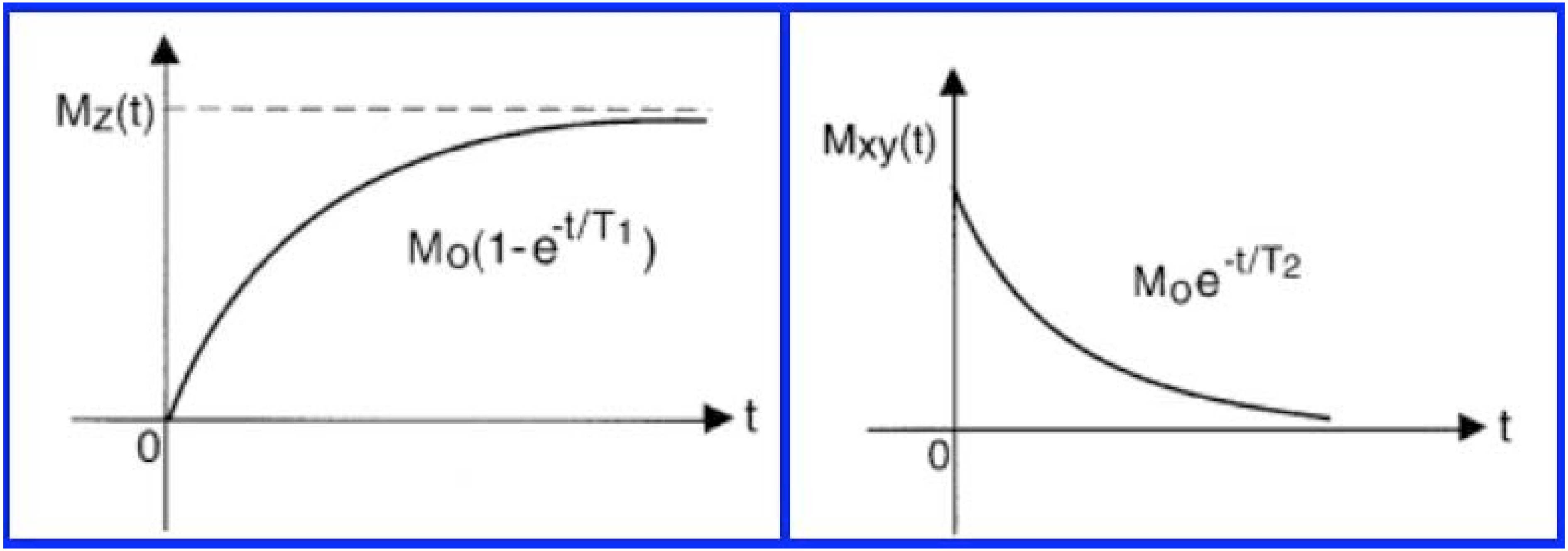


Longitudinal relaxation

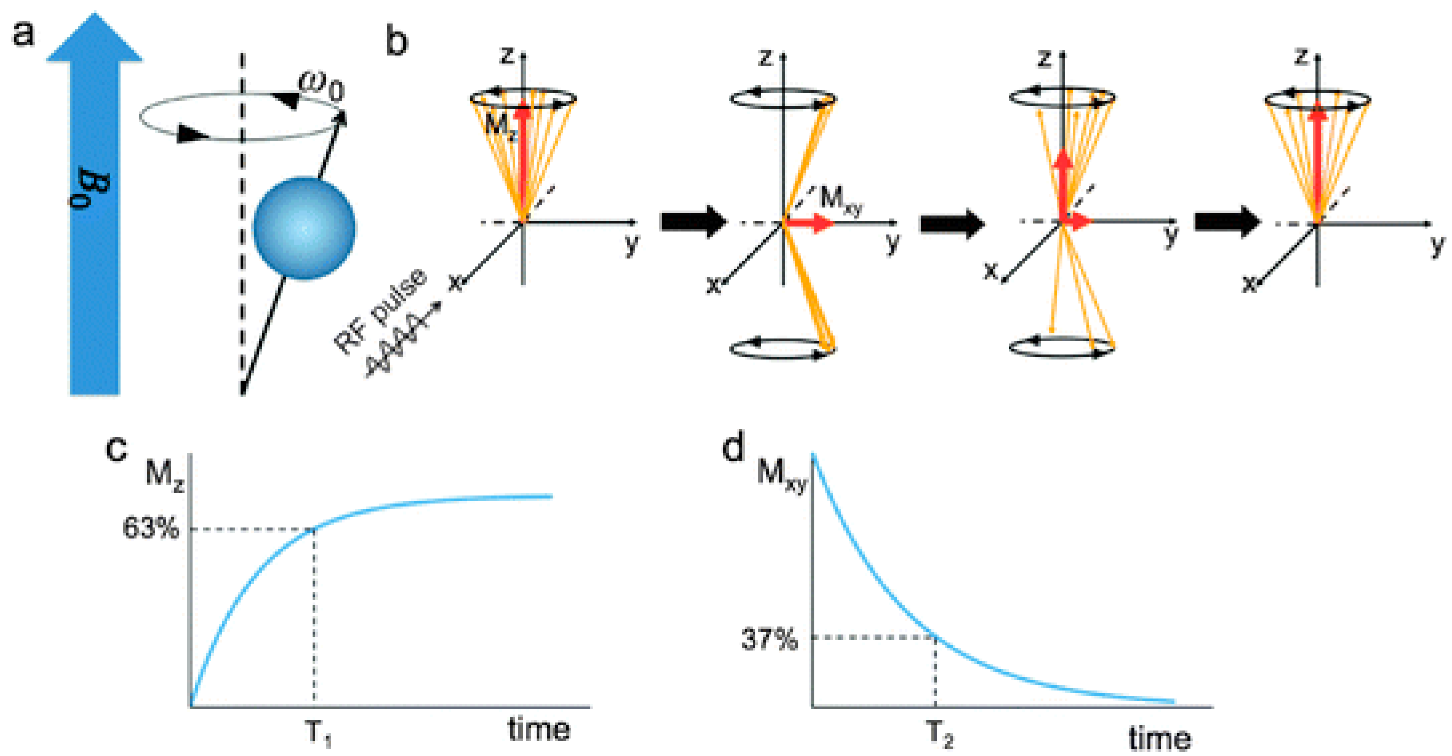


Transverse relaxation

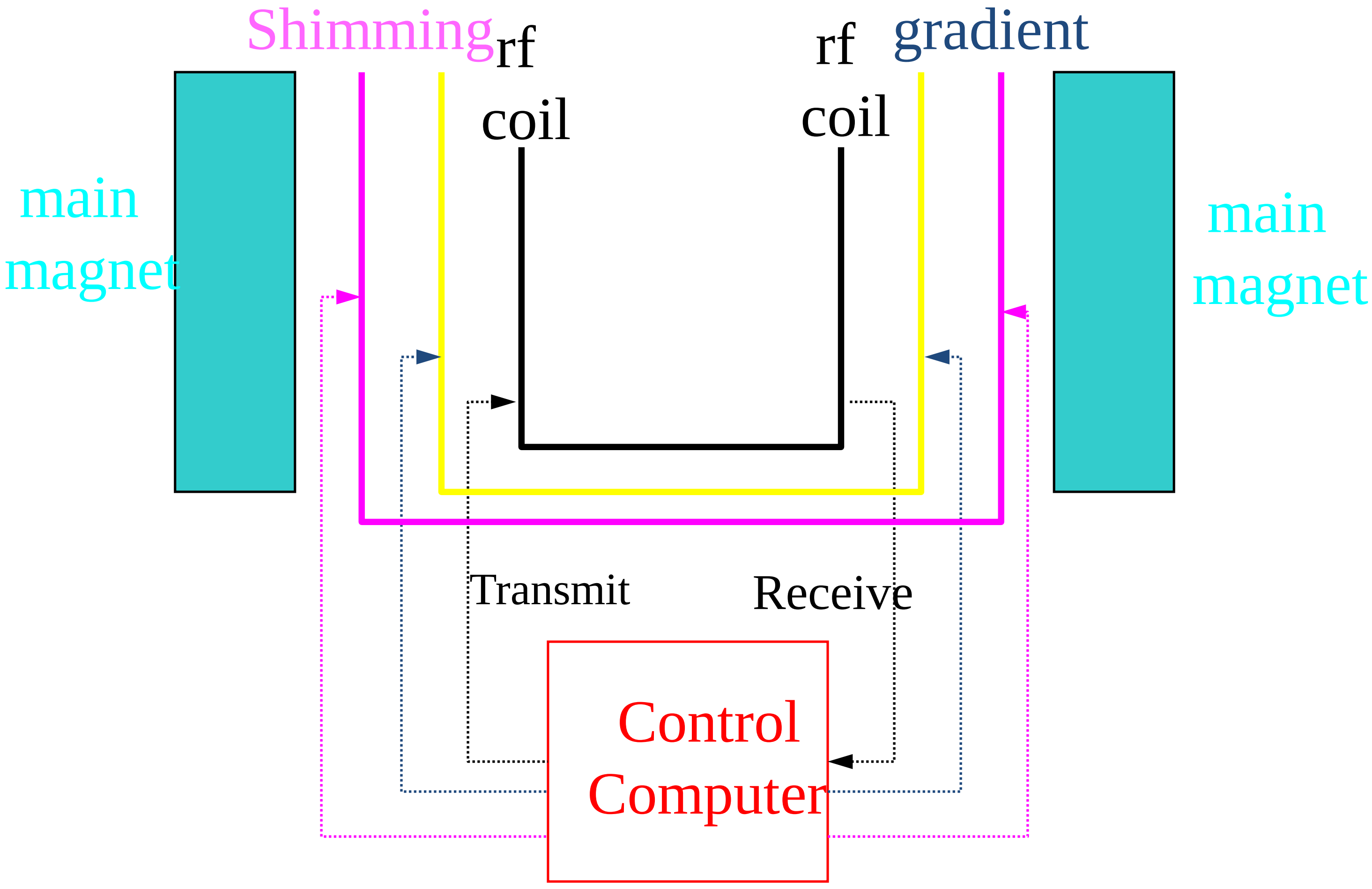
Magnetization



MRI Principle



Components of MRI



Magnetic resonance

Spatial Localization

Slice selection

In a homogeneous magnetic field (B_0), an magnetic field (B_1) oscillating at or very near the resonant frequency, ω, will excite nuclei in the bore of the imaging magnet (Larmor equation):

$$\omega = \gamma B_o$$

Spatial Localization

Slice selection (continued)

We can selectively excite nuclei in one slice of tissue by incorporating a third magnetic field: the "gradient" magnetic field. A gradient magnetic field is a small magnetic field superimposed on the static magnetic field. The gradient magnetic field produces a linear change in the total magnetic field.

Here, "gradient" means "change in field strength as a function of location in the MRI bore".

Spatial Localization

Slice selection (continued)

Since the gradient field changes in strength as a function of position, we use the term "gradient amplitude" to describe the field:

Gradient amplitude = Δ (field strength) / Δ (distance)

Example units: gauss / cm

Spatial Localization

Slice selection (continued)

To recap, we use these magnetic fields in MRI:

- B_0 – large, homogeneous field of superconducting magnet.
- B_1 – temporally-oscillating, RF magnetic field that excites nuclei at resonance.
- B_g – spatially-varying, small field responsible for the linear variation in the total field.

Spatial Localization

Slice selection (continued)

The linear change of the gradient field can be along the Z axis (inferior to superior), the X axis (left to right), the Y axis (anterior to posterior), or any combination (an "oblique" scanning prescription).

Switching the gradient magnetic fields on/off produces the MRI acoustic noise.

Spatial Localization

Slice selection example:

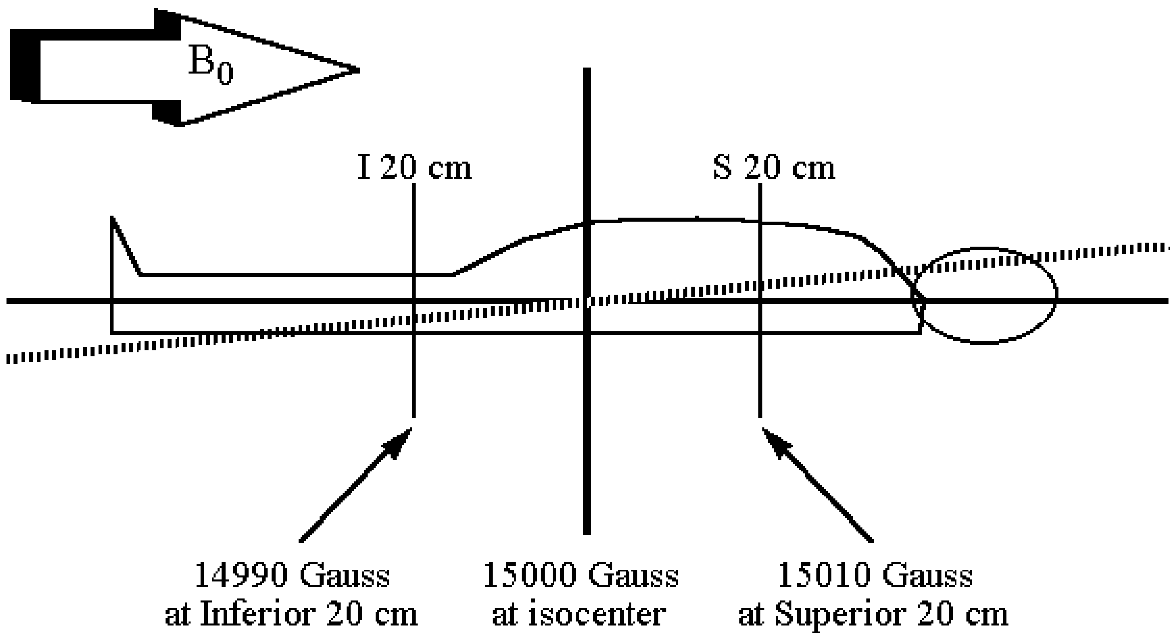
Q: How does the gradient field affect the resonant frequency?

A: The resonant frequency will be different at different locations.

Consider a gradient magnetic field of 0.5 Gauss / cm, using a Z gradient superimposed on a 1.5 T static magnetic field (1.5T = 15,000 Gauss).

Spatial Localization

Here's a picture of the total magnetic field as a function of position:



Spatial Localization

- Recall the Larmor equation:

$$\omega = \gamma B_o$$

For hydrogen:

γ = 42.58 MHz/Tesla = 42.58 x 10^6 Hz/Tesla

Calculating the center frequency at 1.5 Tesla:

$\omega = \gamma B_o$

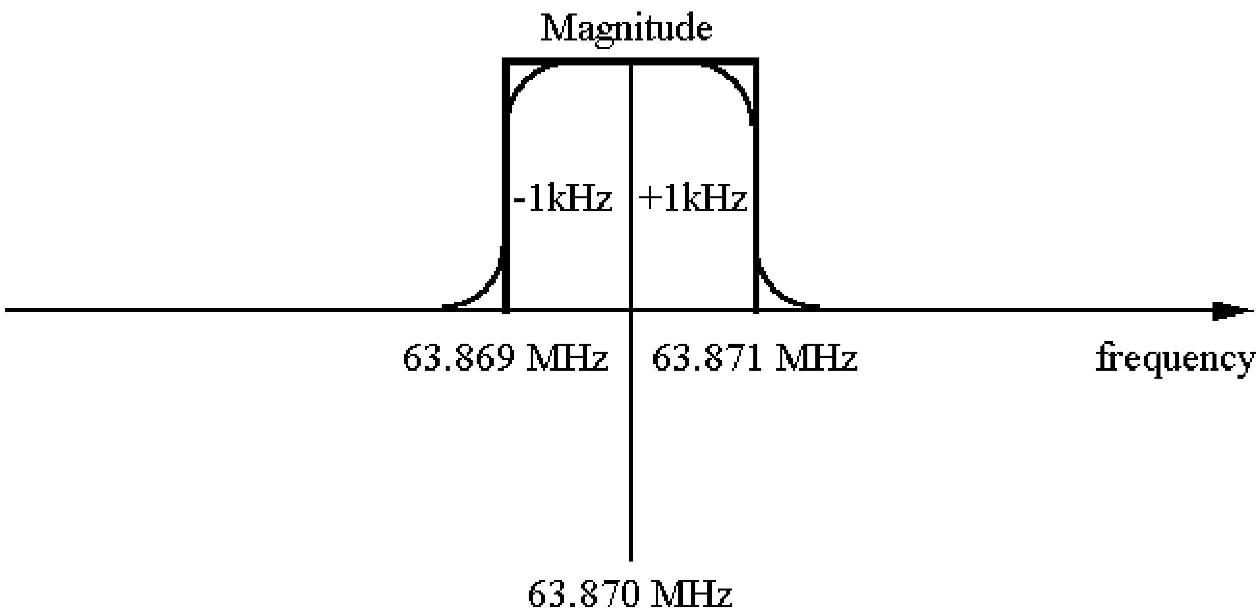
$\omega = (42.58 \text{ MHz / Tesla})(1.5 \text{ Tesla})$

$\omega = 63.87 \text{ MHz}$

RF Bandwidth

- The RF frequency of the oscillating B_1 magnetic field has an associated bandwidth. Rather than oscillating at a single frequency of 63.870 MHz, a range or "bandwidth" of frequencies is present. A typical bandwidth for the oscillating B_1 magnetic field is $\pm$ 1 kHz, thus RF frequencies from 63.869 MHz to 63.871 MHz are present. The bandwidth of RF frequencies present in the oscillating B_1 magnetic field is inversely proportional to the duration of the RF pulse.

RF Bandwidth (continued)



±1 kHz RF bandwidth

Spatial Localization

What are the frequencies at Inferior 20cm and Superior 20cm?

At I 20cm, B_{tot} = 1.499 T:

$\omega_I = \gamma B_{tot}$

$\omega_I = (42.58 \text{ MHz / Tesla})(1.499 \text{ Tesla})$

$\omega_I = 63.827 \text{ MHz}$

At S 20cm, B_{tot} = 1.501 T:

$\omega_S = \gamma B_{tot}$

$\omega_S = (42.58 \text{ MHz / Tesla})(1.501 \text{ Tesla})$

$\omega_S = 63.913 \text{ MHz}$

Difference in frequencies: .086 MHz

RF Bandwidth (continued)

- Where will excited nuclei be located, assuming an 63.87 MHz ($\pm$ 1 kHz) RF bandwidth and a 0.5 Gauss / cm gradient field superimposed on a 1.5 Tesla static magnetic field?

At the inferior-most position of excitation, ω is:

(63.87 – 0.001) MHz = 63.869 MHz

= (42.58 MHz/Tesla)(B inferior)

B inferior = 63.869 MHz/(42.58 MHz/T)

= 1.4999765 Tesla = 14999.765 Gauss

RF Bandwidth (continued)

So, change of field from center to the inferior extent of excitation = (15000 – 14999.765) Gauss

= -0.235 Gauss

The gradient imposes a field change of

0.5 Gauss/cm, so a change of –0.235 Gauss occurs at the following distance from the center:

$$\frac{-\ .235\ \text{Gauss}}{.5\ \text{Gauss}/\text{cm}} \Rightarrow -\ .47\ \text{cm}$$

RF Bandwidth (continued)

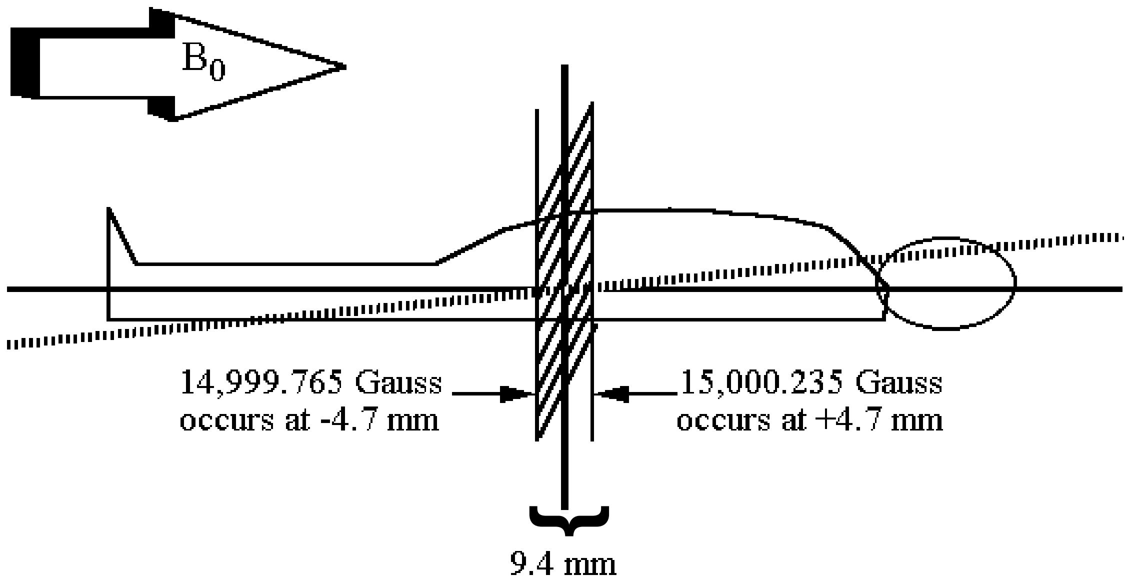
- Same is true of the superior position:

$$63.871\ \text{MHz} = (42.58\ \text{MHz / Tesla})(\text{B superior})$$

B superior = 1.5000235 Tesla = 15000.235 Gauss

$$\frac{.235\ \text{Gauss}}{.5\ \text{Gauss / cm}} \Rightarrow .47\ \text{cm}$$

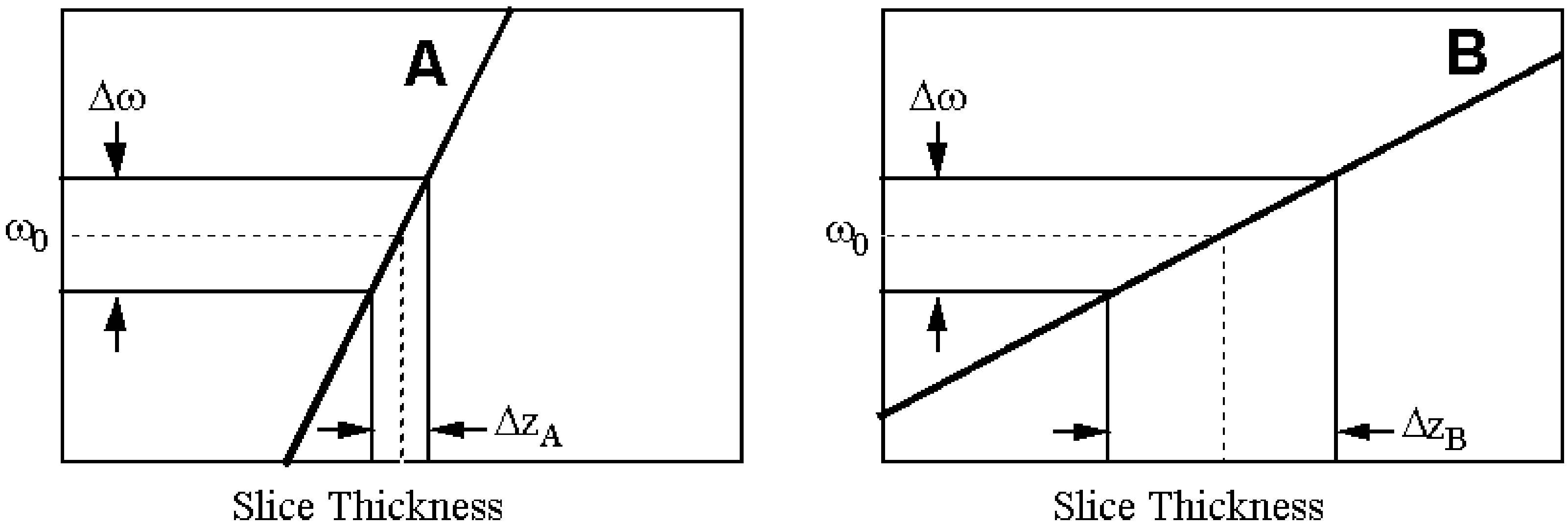
RF Bandwidth (continued)



RF Bandwidth (continued)

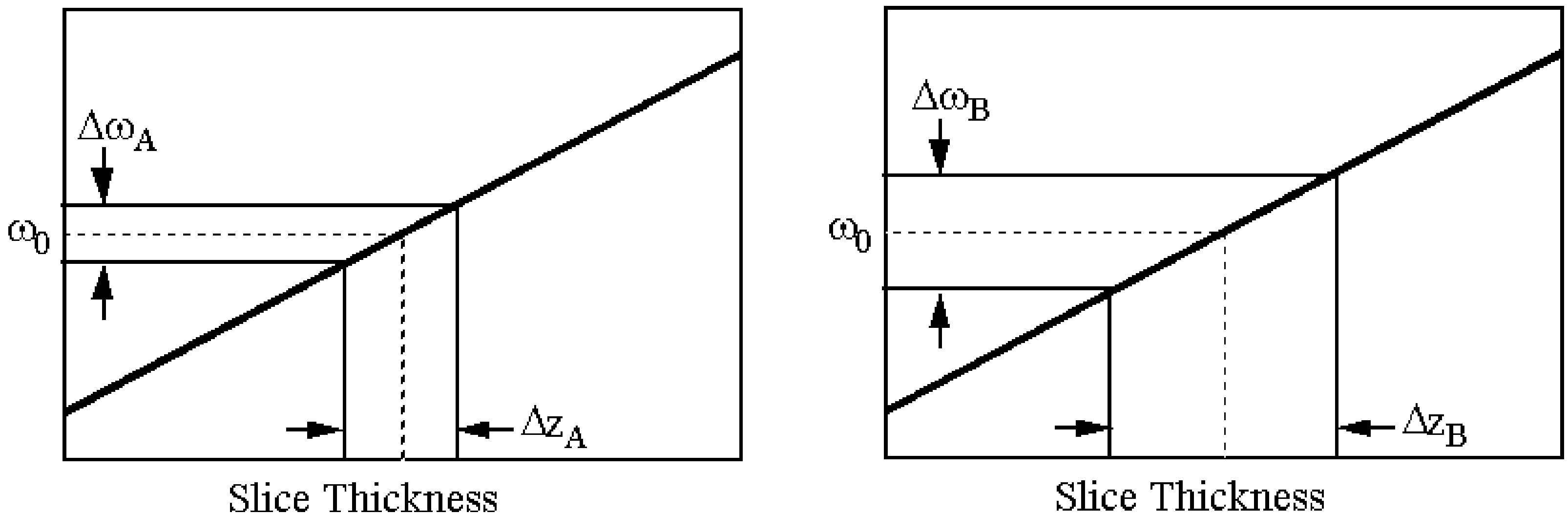
- A $\pm$ 1 kHz RF pulse at 63.870 MHz will excite a 9.4 mm thick slice in the presence of a 0.5 Gauss/cm gradient at 1.5 Tesla.
- The maximum gradient on the GE LX Horizon Echospeed+ is 3.3 Gauss/cm. The maximum gradient on the Siemens Vision is 2.5 Gauss/cm.
- Slice thickness can be changed with RF bandwidth or gradient magnetic field or both.

RF Bandwidth (continued)



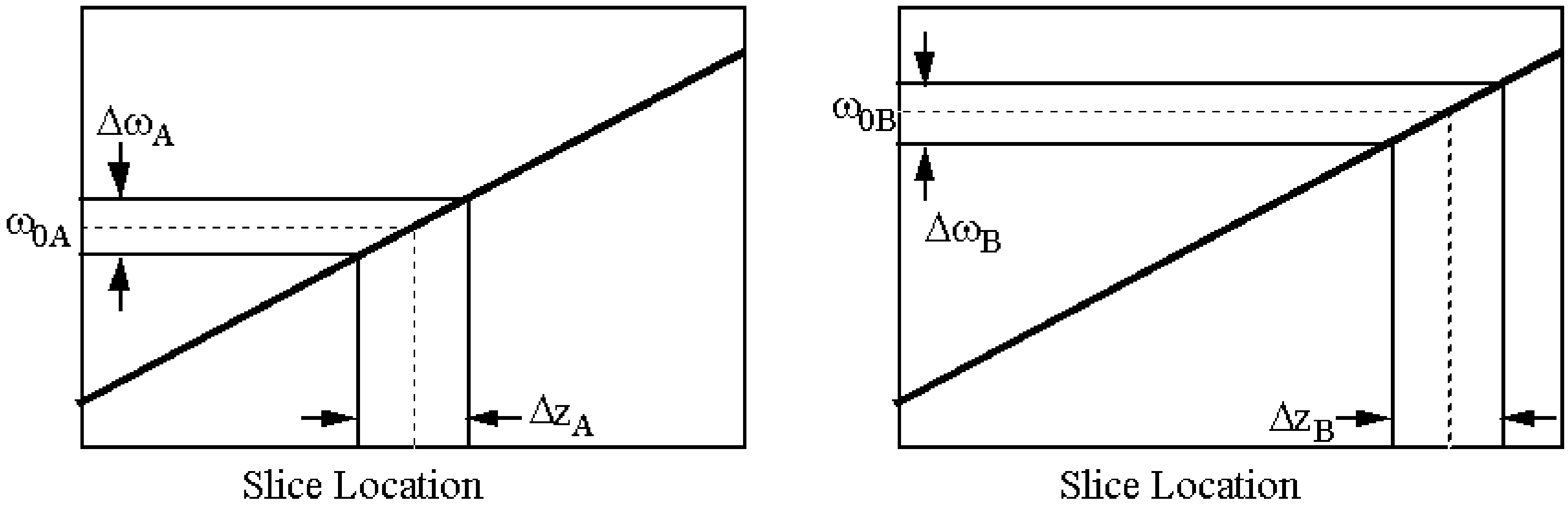
At a specific RF bandwidth ($\Delta\ \omega$), a high magnetic field gradient (line A) results in slice thickness ($\Delta\ z_A$). By reducing the magnetic field gradient (line B), the selected slice thickness increases in width ($\Delta\ z_B$).

RF Bandwidth (continued)



An increase in the RF bandwidth applied at a constant magnetic field gradient results in a thicker slice ($\Delta z_B > \Delta z_A$).

Slice Location (RF frequency)



The location of the selected slice can be moved by changing the center RF frequency ($\omega_{0B} > \omega_{0A}$).

Two-Dimensional Fourier Transform MRI (2DFT)

- Planar imaging - a plane or slice of spins has been selectively excited as shown previously. After the Z gradient and RF pulse have been turned off, (and after a brief rephasing with the Z gradient) all spins in the slice are precessing in phase at the same frequency.

- A **2 Dimensional Fourier Transform (2DFT)** technique can now be used to image the plane.

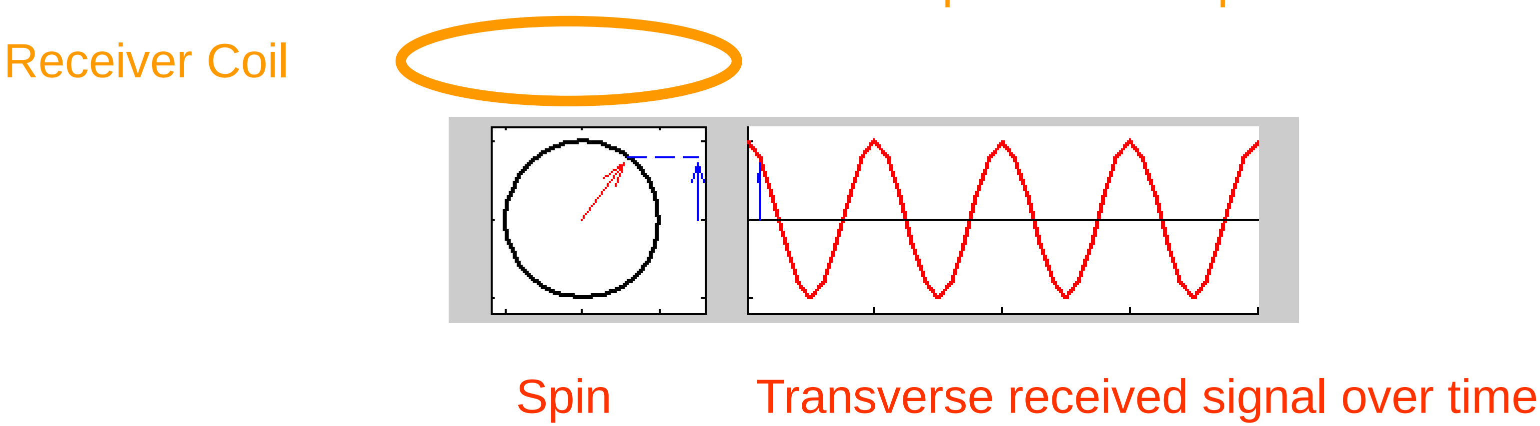
- After imaging a plane, the RF frequency can be changed to image other planes in order to build up an image volume.

Two-Dimensional Fourier Transform MRI (2DFT)

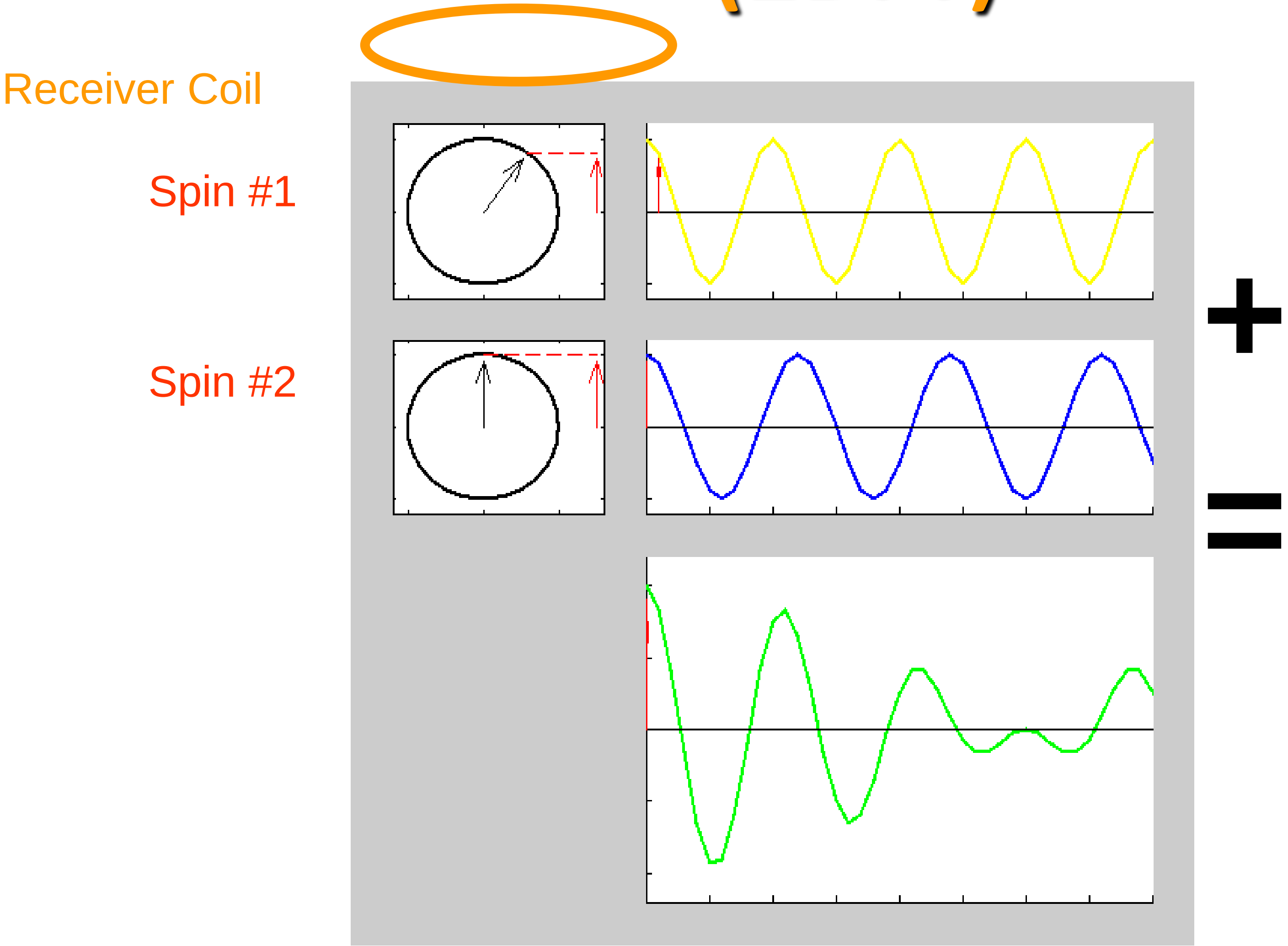
Q: Once spins in a slice are excited, how does the scanner "observe" the data?

A: The receive coil in the scanner detects the TOTAL transverse magnetization signal in a particular direction, resulting from the SUM of ALL excited spins.

Example #1: one spin case



Two-Dimensional Fourier Transform MRI (2DFT)

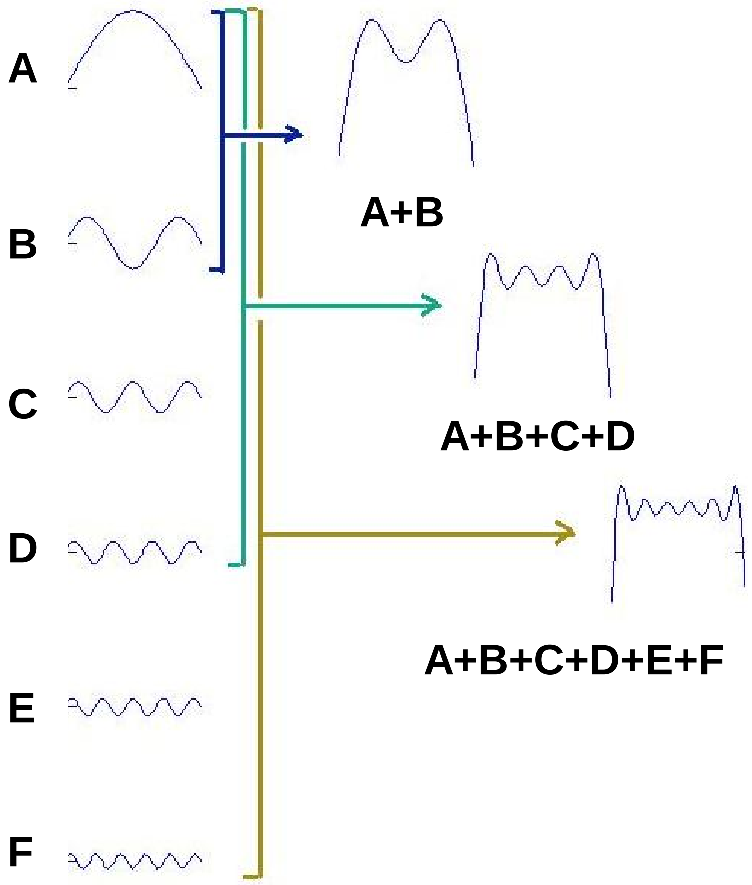


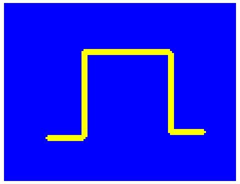
Example #2: two spin case, with different frequencies

Summed signal can be complicated…but, this is useful.

Fourier Transform Basics

In 1-D, we can create a wave with a complicated shape by adding periodic waves of different frequency together.

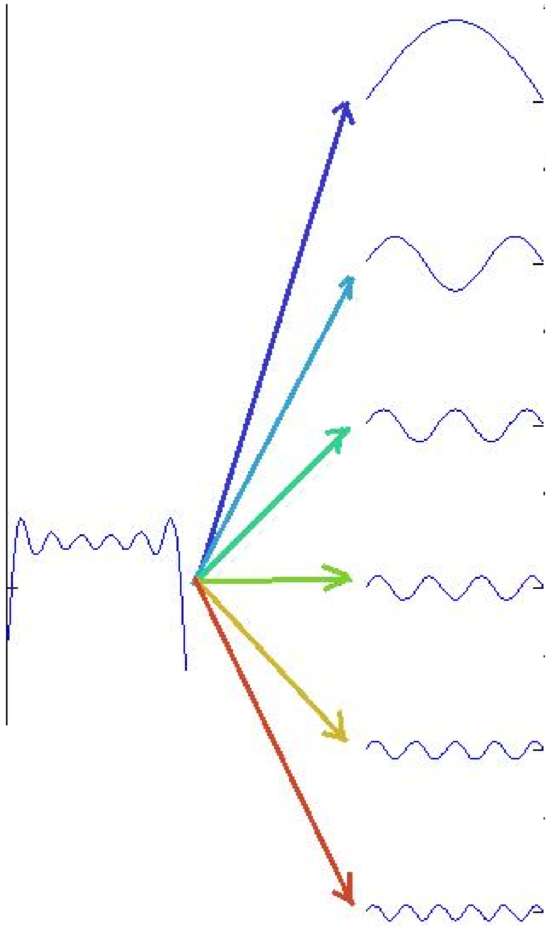




In this example, we could keep going to create a square wave, if we wanted.

Fourier Transform Basics

This process works in reverse as well: we can decompose a complicated wave into a combination of simple component waves.

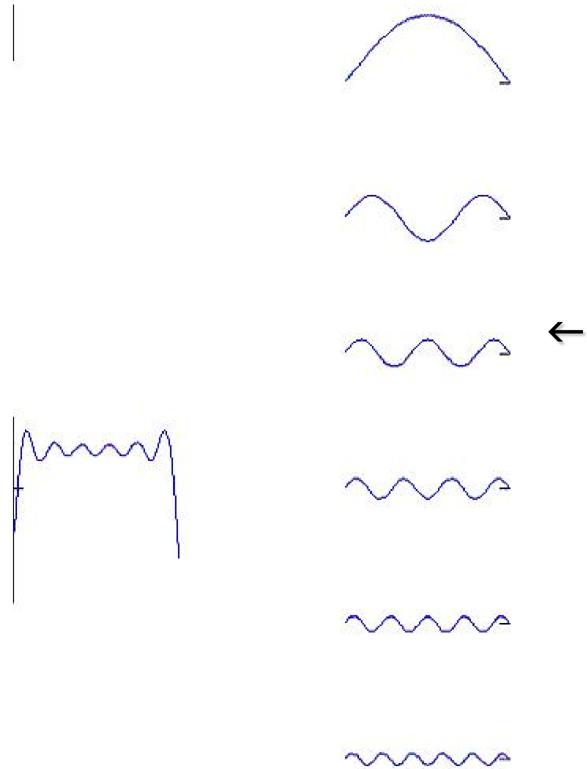


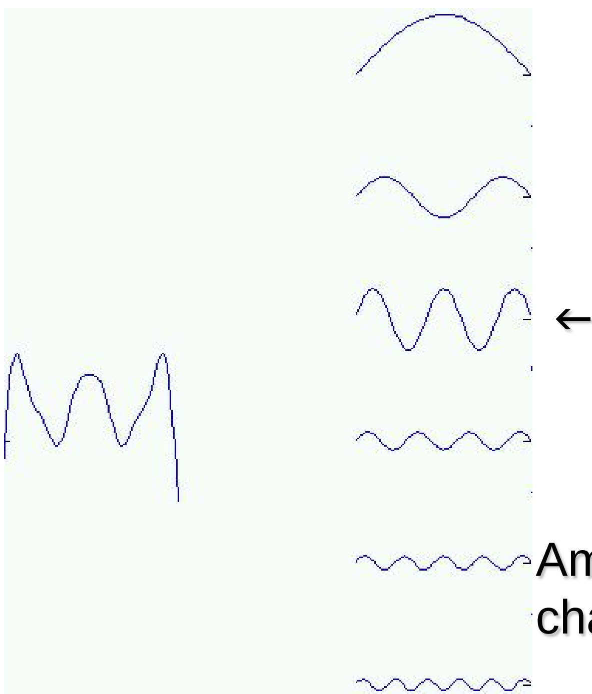
The mathematical process for doing this is known as a “Fourier Decomposition”.

Fourier Transform Basics

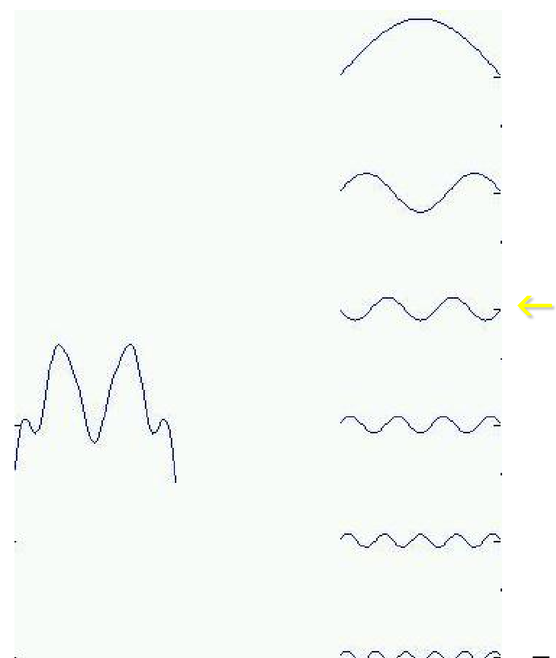
For each different frequency component, we need to know the amplitude…

…and the phase, to construct a unique wave.





Amplitude change



Phase change

Fourier Transform Basics

So, if spins in an excited slice were prepared such that they precess with a different frequency and phase at each position, the resulting signal could only be constructed with a unique set of magnetization amplitudes from each position.

Thus, we could apply a Fourier transform to our total signal to determine the transverse magnetization at every position.

So, let's see how we do this…

Phase and Frequency Encoding

Consider an MRI image composed of 9 voxels (3 x 3 matrix)

ω ϕ	ω ϕ	ω ϕ
ω ϕ	ω ϕ	ω ϕ
ω ϕ	ω ϕ	ω ϕ

ω = precessional frequency

ϕ = phase

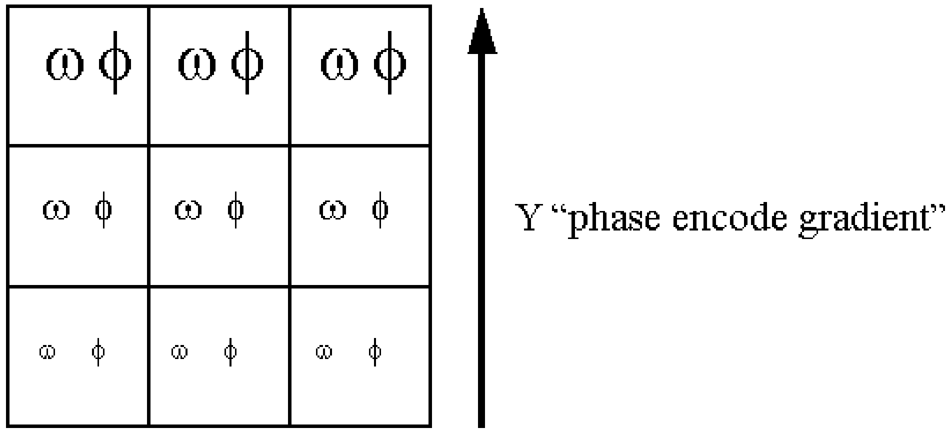
All voxels have the same precessional frequency and are all “in phase” after the slice select gradient and RF pulse.

Phase and Frequency Encoding
(continued)

1. Apply a Y gradient or "phase encode gradient"

2. Nuclei in different rows experience different magnetic fields. Nuclei in the highest magnetic field (top row), precess fastest and advance the farthest (most cycles) in a given time.

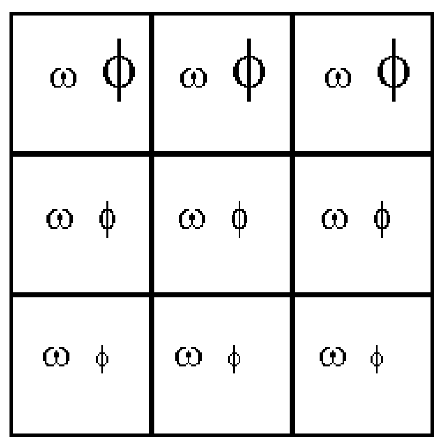
Phase and Frequency Encoding
(continued)



When the Y “phase encode gradient” is on, spins on the top row have relatively higher precessional frequency and advanced phase. Spins on the bottom row have reduced precessional frequency and retarded phase.

Phase and Frequency Encoding
(continued)

3. Turn off the Y "phase encode gradient"
4. All nuclei resume precessing at the same frequency
5. All nuclei retain their characteristic Y coordinate dependent phase angles

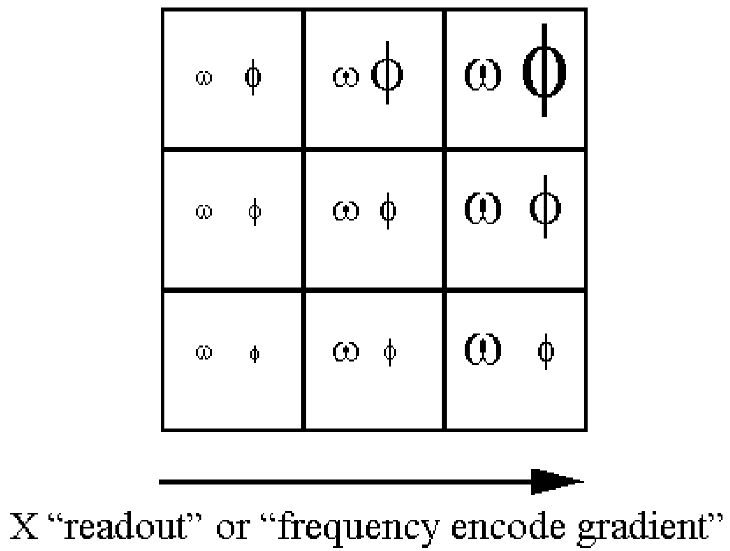


Phase and Frequency Encoding
(continued)

6. A “read out” gradient is applied along the X axis, creating a distribution of precessional frequencies along the X axis.

7. The signal in the RF coil is now sampled in the presence of the X gradient.

Phase and Frequency Encoding (continued)

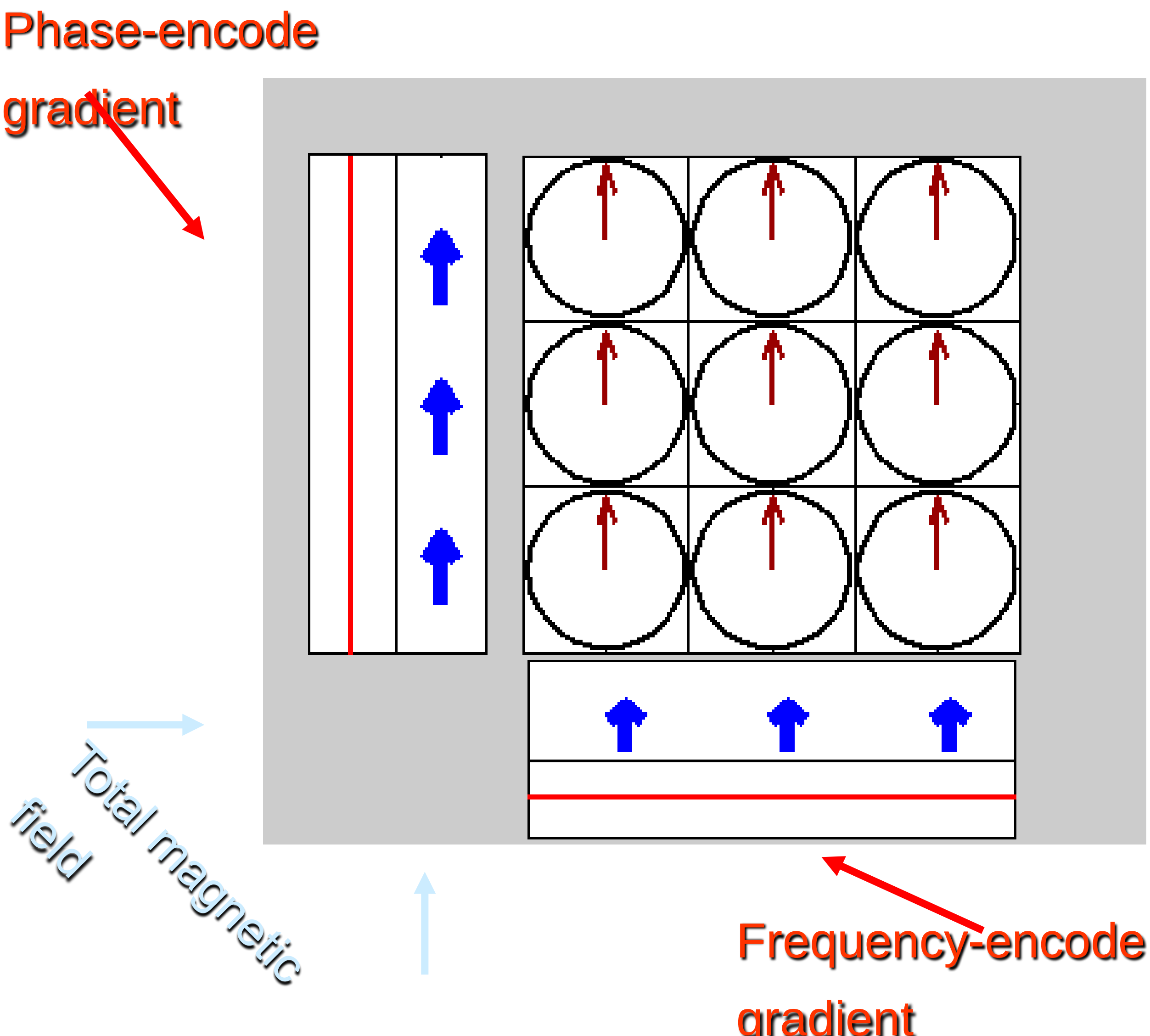


While the frequency encoding gradient is on, each voxel contributes a unique combination of phase and frequency. The signal induced in the RF coil is measured while the frequency encoding gradient is on.

Phase and Frequency Encoding
(continued)

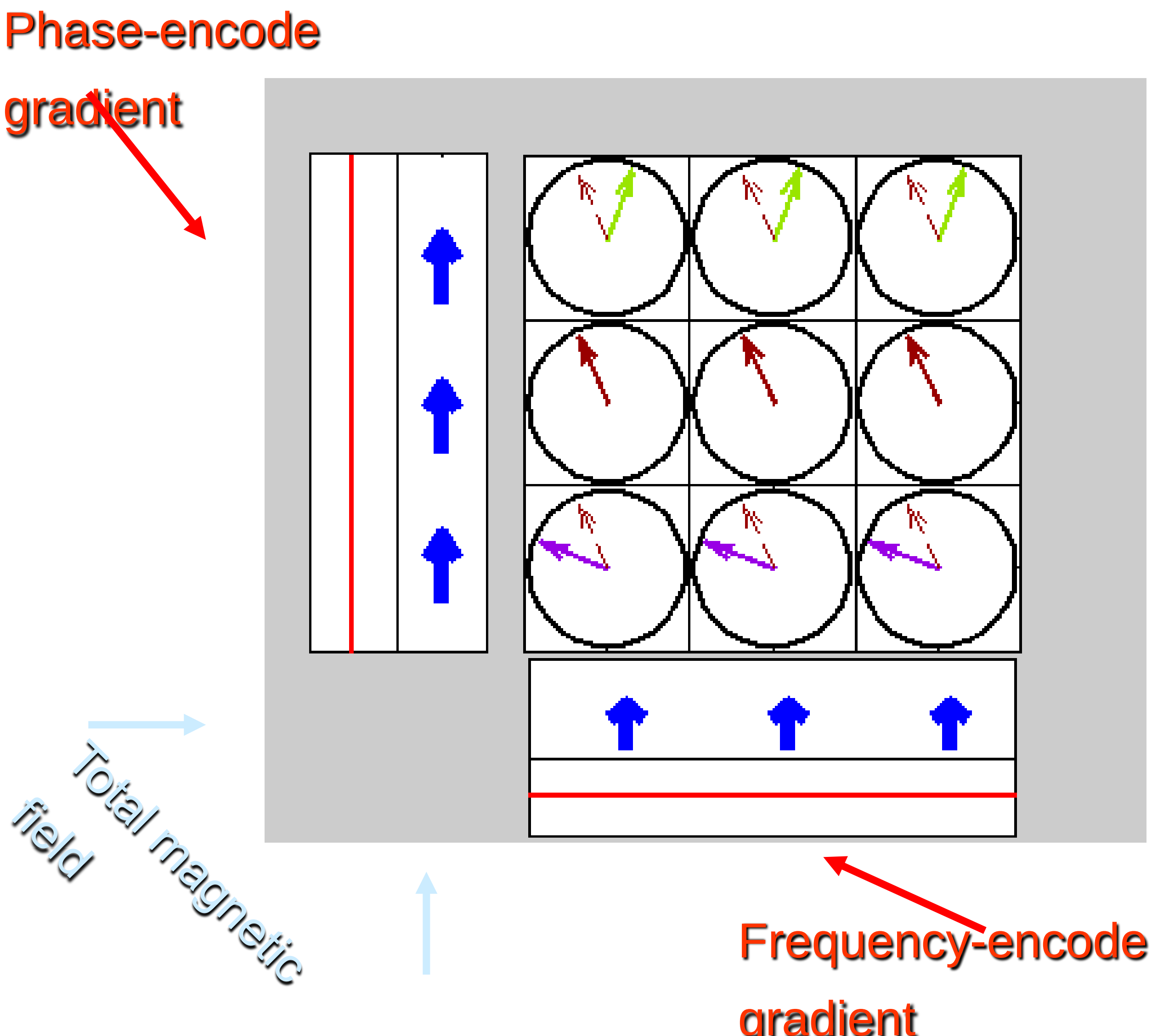
Let’s watch a movie of this process…

Phase and Frequency Encoding
(continued)



Phase-encoding of these rows occurs by turning a gradient on for a short period of time…

Phase and Frequency Encoding
(continued)



…then, frequency-encoding of the columns occurs by turning a gradient on for a different axis and leaving it on during the readout.

Phase and Frequency Encoding (continued)

8. The cycle is repeated with a different setting of the Y phase encoding gradient. For a 256 x 256 matrix, at least 256 samples of the induced signal are measured in the presence of an X frequency encoding gradient. The cycle is repeated with 256 values of the Y phase encoding gradient.

9. After the samples for all rows are taken for every phase-encode cycle, 2D Fourier Transformation is then carried out along the phase-encoded columns and the frequency-encoded rows to produce intensity values for all voxels.

Phase and Frequency Encoding
(continued)

A 2DFT can be accomplished around any plane, by choosing the appropriate gradients for slice selection, phase encoding and frequency encoding.

k - space

The Fourier transformation acts on the observed "raw data" to form an image. A conventional MRI image consists of a matrix of 256 rows and 256 columns of voxels (an "image matrix").

The "raw data" before the transformation ALSO consists of values in a 256 x 256 matrix.

k - space

This “raw data” matrix is affectionately known as “k-space”. Two-dimensional Fourier Transformation (2DFT) of the “k-space” produces an image.

Each value in the resulting image matrix corresponds to a grey scale intensity indicative of the MR characteristics of the nuclei in the voxels. Rows and columns in the image are said to be “frequency encoded” or “phase encoded”.

k - space

For an MRI image having a matrix of 256 rows and 256 columns of voxels, acquisition of the data requires that the spin population be excited 256 times, using a different magnitude for the phase encoding gradient for each excitation.

k - space
(continued)

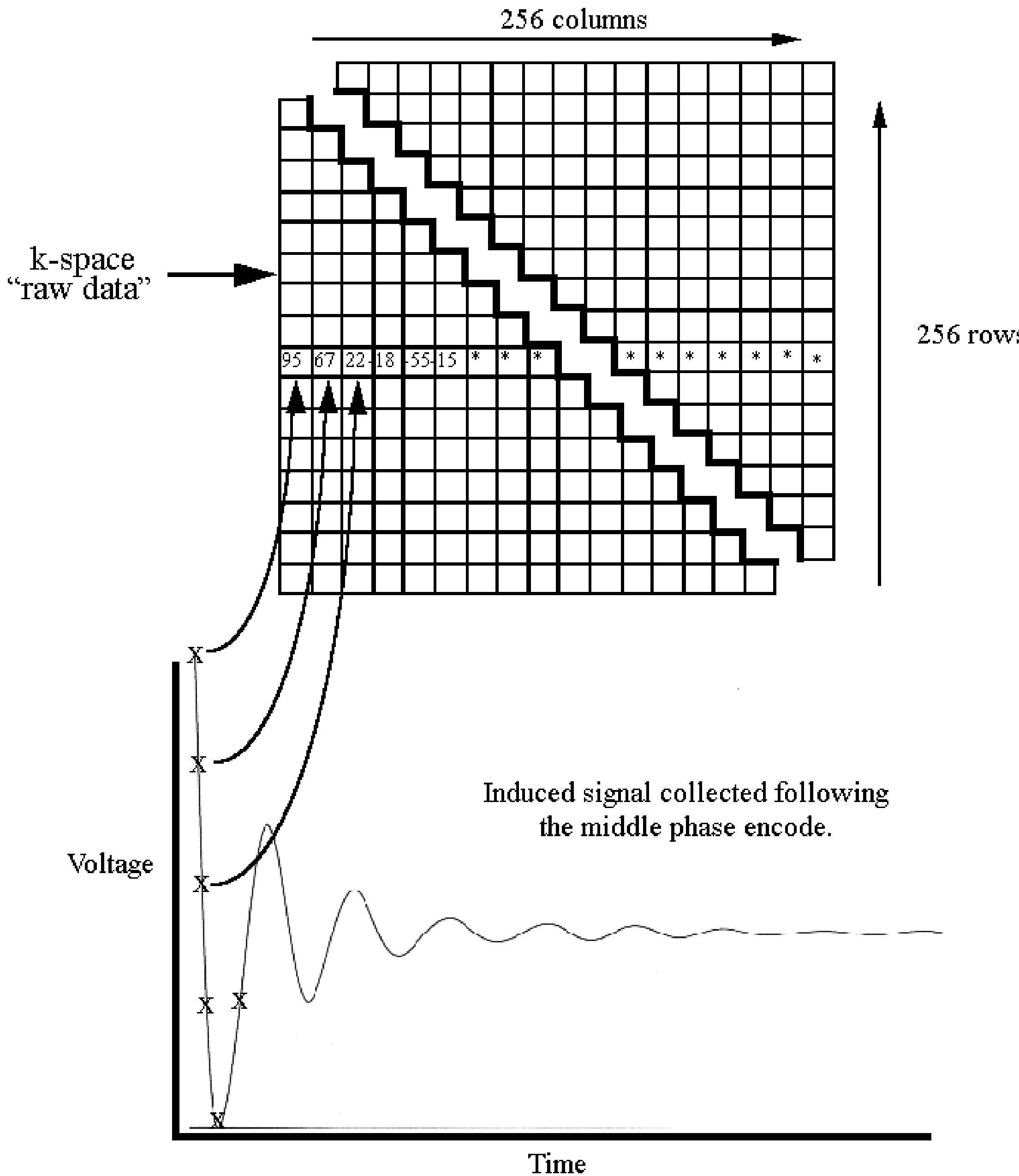
The top row of k-space would be measured in the presence of a strong positive phase encode gradient.

A middle row of k-space would be measured with the phase encode gradient turned off.

The bottom row of k-space would conventionally be measured in the presence of a strong negative phase encode gradient.

k - space
(continued)

While the frequency encoding gradient is on, the voltage in the RF coil is measured at least 256 times. The 256 values measured during the first RF pulse are assigned to the first row of the 256 x 256 "raw data" matrix. The 256 values measured for each subsequent RF pulse are assigned to the corresponding row of the matrix.



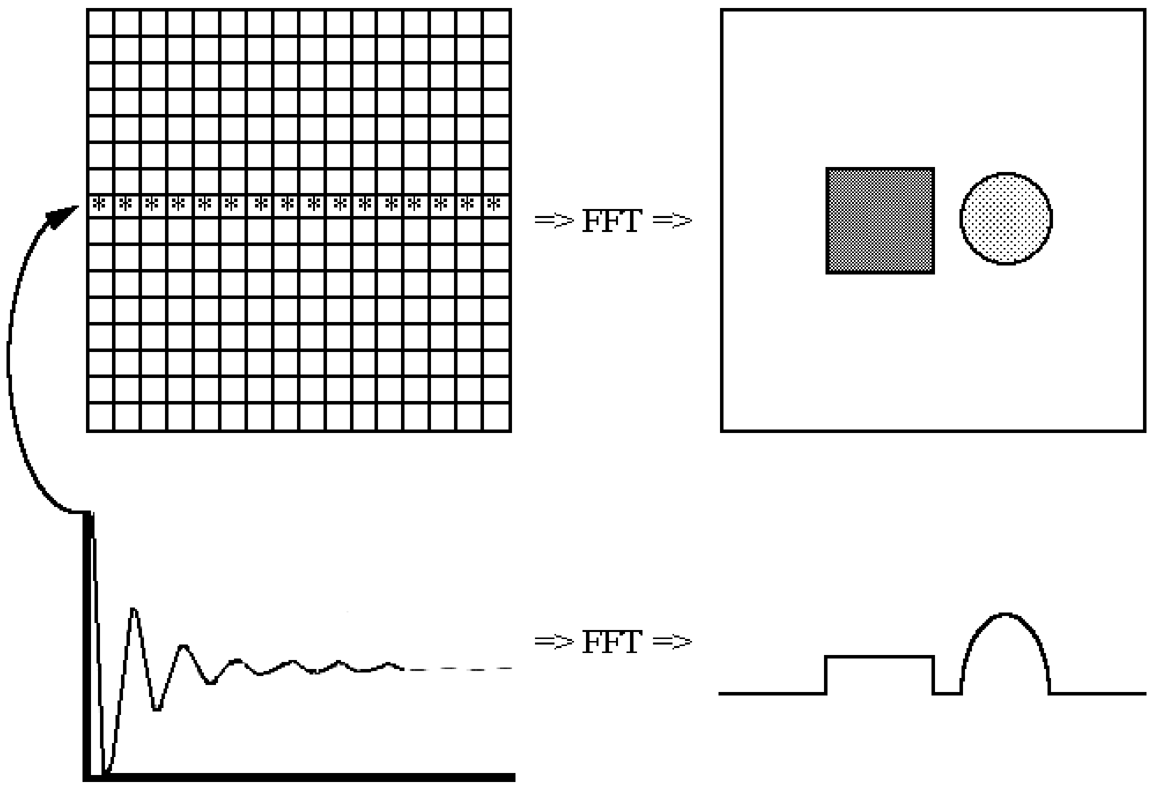
256 columns
k-space
"raw data"
256 rows
95
67
22
-18
-55
-15
Induced signal collected following
the middle phase encode.
Voltage
Time

k - space
(continued)

There are many techniques of “filling” or “traversing” k-space, each of which may convey different imaging advantages. These techniques will reviewed in subsequent sections.

k - space

(continued)



The central row of k-space is measured with the phase encode gradient turned off. An FFT of the data in the central row produces a projection or profile of the object.

Použité zdroje

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