EQUIPMENT TECHNIQUES OF MAGNETIC RESONANCE IMAGING

3.ST STAGE
LECTURE 3 - THEORY

GRADIENT COILS AND SPATIAL ENCODING (SELECT SLICE)

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Larmor Equation:

In a simple picture we can think of the spins as rotating at the Larmor or resonance frequency, which is also the frequency of the MR signal given by the equation:

$$\omega_0 = \gamma\, B_0$$

Where frequency is in Megahertz (MHz) and Magnetic field is in Tesla (T). The gyro-magnetic ratio (which has the symbol γ, pronounced 'gamma') and is a property of the nucleus in question. Its value more exactly is 42.56 MHz T^{-1} for hydrogen (water or fat protons).

- This equation is the only you need to know and to understand spatial localization.

Gradient Coils:

It is coils of wire that alter the Magnetic field strength in a linear fashion when a current is passed through them. Gradient coils are used to produce deliberate variations in the main Magnetic field (B_0).

- Spatial localization of the MR signal requires the use of three orthogonal linear Magnetic field gradients ('gradients' for short).

The gradients are applied repeatedly in a carefully controlled pulse sequence. They generate loud tapping, clicking or higher pitched beeping sounds during scanning, like a loudspeaker.

There are three sets of gradient coils, one each for the X, Y, and Z directions.

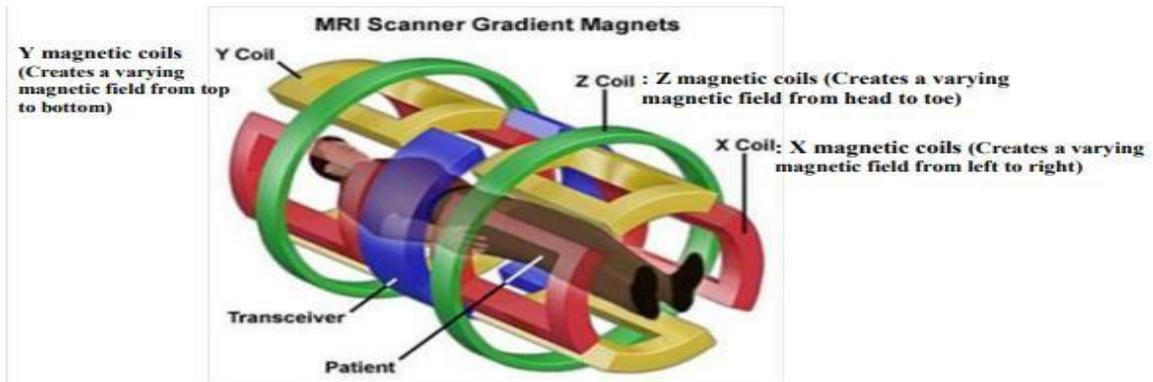Figure 1 shows the configuration of such coils.

Figure 1: Diagram of coils within the bore of the magnet that are used to create gradients.

Gradient coils are loops of wire on a cylindrical shell lying just inside the bore of an MR scanner, which enable us to create additional magnetic fields, which are, in a way, superimposed on the main Magnetic field B_0. When current is passed through these coils a secondary Magnetic field is created. This gradient field slightly distorts the main Magnetic field in a predictable pattern, causing the resonance frequency of protons to vary in as a function of position.

The gradient subsystem comprises the set of gradient coils, inside the bore of the magnet, and the gradient amplifiers which drive electrical current through them. In the standard cylindrical clinical systems,

the direction along the bore is termed the z-axis, the left–right direction is termed the x-axis and the top– bottom direction is termed the y-axis. The null point at the centre of the gradient coils, and also the centre of the magnet, is called the isocentre.
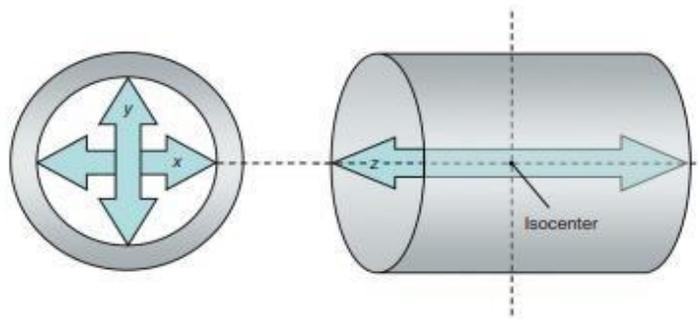

Figure 2: Gradient axes in a typical superconducting system.

The coils heat up rapidly due to the large currents, and almost all commercial MR systems have a cooling system, tubes carrying chilled water through the gradient coils and amplifiers.
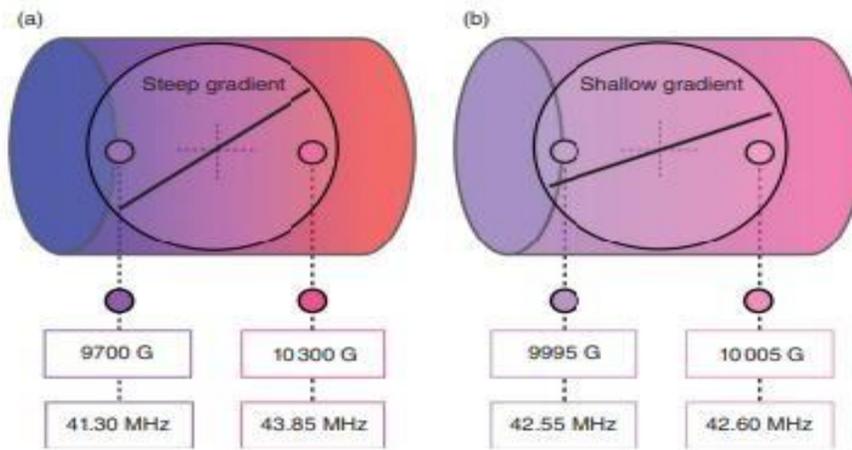

Figure 3: (a) Steep and (b) shallow gradient slopes.

In order to image one slice of the body we use gradient magnets. These magnets smoothly increase the strength of the external field across the length and width of the body. Because the field strength is different at different parts of the body, the frequency of precession will also be different. Applying an RF pulse at that frequency will only resonate the nuclei at that location.

- Whenever a magnetic field differs in magnitude or direction between two points in space, a magnetic gradient is said to exist.

Mathematically: The gradient (G) is defined as change in field (ΔB) divided by change in distance (Δs). As a simple example let us assume the magnetic fields at the two ends of a horizontal bore scanner are 1.000 T and 1.002 T respectively, and that these locations are separated by 1 meter.
The gradient is:

Gradient = (1.002 − 1.000)/1
= 2 mT/meter

1.000 T   1.002 T

1 meter

The requirement of the gradient coils are twofold.

- First. they are required to produce a linear variation in field along one direction,
- Secondly, to have high efficiency, low inductance and low resistance, in order to minimize the current requirements and heat deposition
- The design of the z-gradients is usually based on circular (Maxwell) coils, while the transverse (x- and y-) gradients typically have a saddle (Golay) coil configuration.

These physical effects of rapid current flows through the gradients give rise to three potential MR safety:

1. Acoustic noise.
2. Nerve/cardiac stimulation.
3. Heating around implants.

Acoustic noise: The switching of gradient currents is the source of acoustic noise, can reach levels which may damage hearing.

- The noise level depends on the gradient strength, the switching time, and the mechanical properties of the gradient set. The noise in the bore can easily be higher than 85 dB, a noise level which is high enough to cause temporary hearing damage. That's why you should always encourage patients to use hearing protection.

Gradient Strength and Slew Rate:

The gradient rise time, or how rapidly the field changes with time from zero to the peak amplitude, is usually expressed in microseconds (μs), with typical values from 1000 μs down to 200 μs. The gradient slew rate is calculated by dividing the peak gradient amplitude by the rise time.

Typical slew rates are in the range 20–200 T m^{-1} s^{-1}.

Slew Rate = Peak Gradient Strength / Rise Time

$$\text{Slew Rate} = \frac{G_{max}}{Rise\ Time}$$
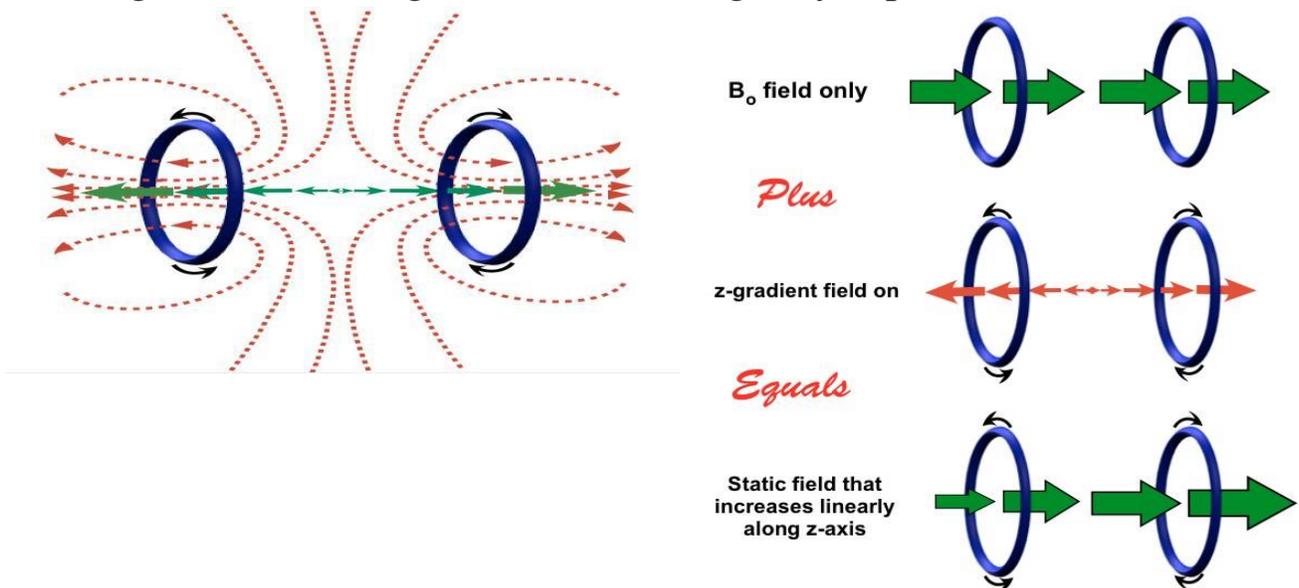
Peak Gradient Strength
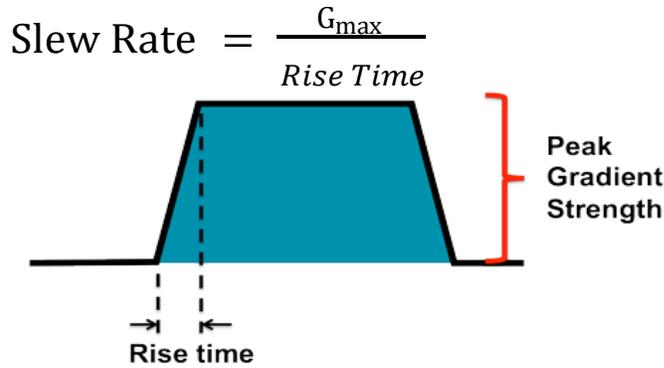
Rise time

(H.W) A gradient magnetic field has a maximum amplitude of 40 mT/m, with a rise time of 250 μs. What is the slew rate?

High slew rates are necessary for acquiring high-quality EPI images or for very short TE and TR, e.g. for cardiac imaging, and to achieve short echo spacing in TSE. Systems with very high slew rates may require software limits to prevent peripheral nerve stimulation

Z- Gradients:

The Z gradient coils are usually a pair of circular coils, each of which is wound on the cylinder at opposite ends of the imaging volume (Figure11-6). These are often called Maxwell coils, If a direct current with opposite polarity is passed through the two coils, a small change in the magnetic field along the Z-axis of the gantry is produced.

B_o field only

Plus

z-gradient field on

Equals

Static field that increases linearly along z-axis

Maxwell coils produce an incremental field that is zero at magnet isocenter but increases linearly outward in both the +z and -z directions. When this is added to the constant (Bo) field, the result is a gradually increasing gradient along the z-axis.
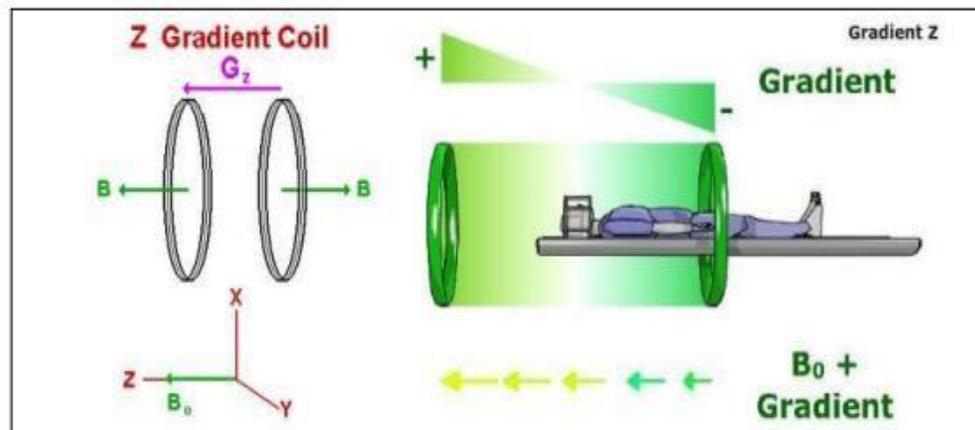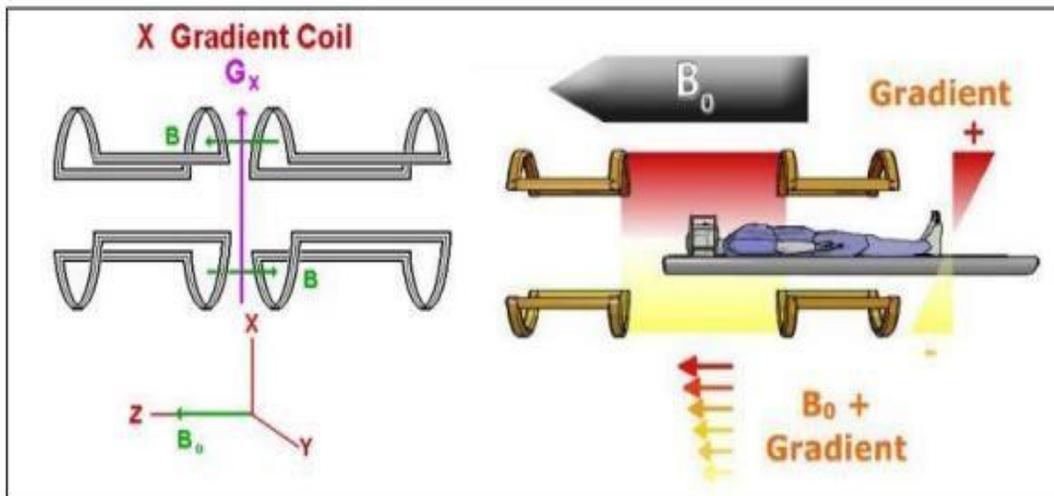

Variations in the field produced by gradients in direction z
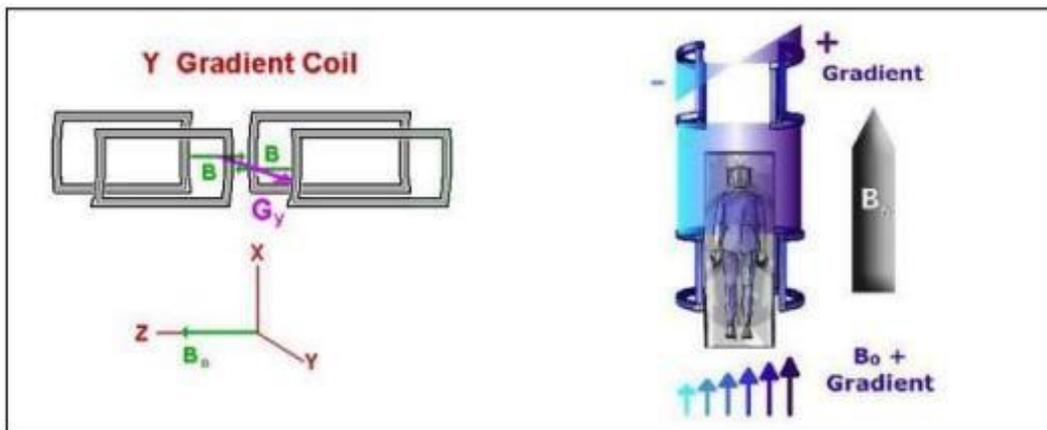
X- and *Y-*Gradients:

The X gradient magnetic field is induced by a pair of coils—actually four saddle-shaped coils in sets of two—positioned on either side of the cylinder. By convention, these coils are positioned so that the gradient magnetic field is across the patient laterally. The axis is therefore the horizontal axis across the patient from side to side and can also be used for slice selection (sagittal images).

the Y-axis is the vertical axis through the patient in the anteriorposterior direction. It can also serve to perform slice selection (coronal images).

**Variations in the field produced by gradients in direction x**

Variations in the field produced by gradients in direction y

The X and Y gradients in the B_0 field are created by a pair of coils (Golay Coils). The X axis coils create a gradient in B_0 in the X direction due to the current through the coils. The Y axis coils provides a similar gradient in B_0 along the Y axis. The design for transverse gradients used in cylindrical MR magnets is based on a "double-saddle" coil configuration originally described in 1958 by Marcel Golay The simplest form of this coil set consists of 4 inner and 4 outer arcs on the surface of a cylinder connected by 8 straight wires running parallel to the z-axis.

Apply gradient:

A magnetic field gradient is applied in the Z-axis superimposed on the background magnetic field. Going back to the Larmor equation the frequency of precession depends on the magnetic field. This means that nuclei will have different frequencies throughout the z-axis
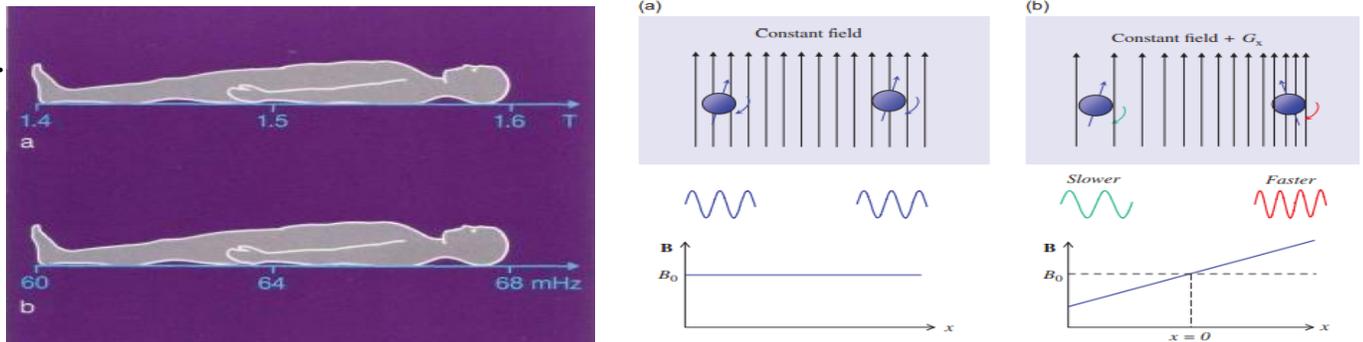
**Spatial Encoding (Select Slice)**

Selecting the slice:

The first step of spatial encoding consists in selecting the slice plane.

- When a gradient coil is switched on, the magnetic field strength, and therefore the precessional frequency of nuclei located along its axis, is altered in a linear fashion. Therefore a specifi c point along the axis of the gradient has a specific precessional frequency, and nuclei situated within a slice have a particular precessional frequency

To do this, a magnetic field gradient, the Slice Selection Gradient (Gs), is applied perpendicular to the desired slice plane. This is added to Bo, and the protons present a resonance frequency variation proportionate to Gs (Larmor equation).

- The slice select gradient (Gs) is a magnetic field gradient applied to select the slice position in the direction of this gradient. The field strength is measured in Tesla per meter (T/m)
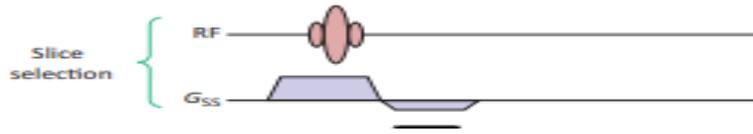
Top line, an RF pulse is applied simultaneously with a slice-selective gradient GSS (line 2). The RF pulse stimulates the MR interactions in tissue which lead to the MR signal.

An RF wave is simultaneously applied, with the same frequency as that of the protons in the desired slice plane. This causes a shift in the magnetization of only the protons on this plane. As none of the hydrogen nuclei located outside the slice plane are excited, they will not emit a signal. The RF wave associated with the slice selection gradient and the adapted resonance frequency, is called the selective pulse.

Selecting the slice plane:

The slice select gradient (Gs) is a magnetic field gradient applied to select the slice position in the direction of this gradient.
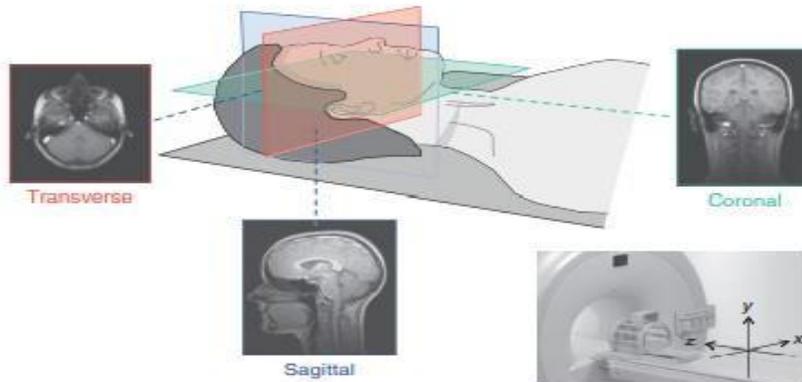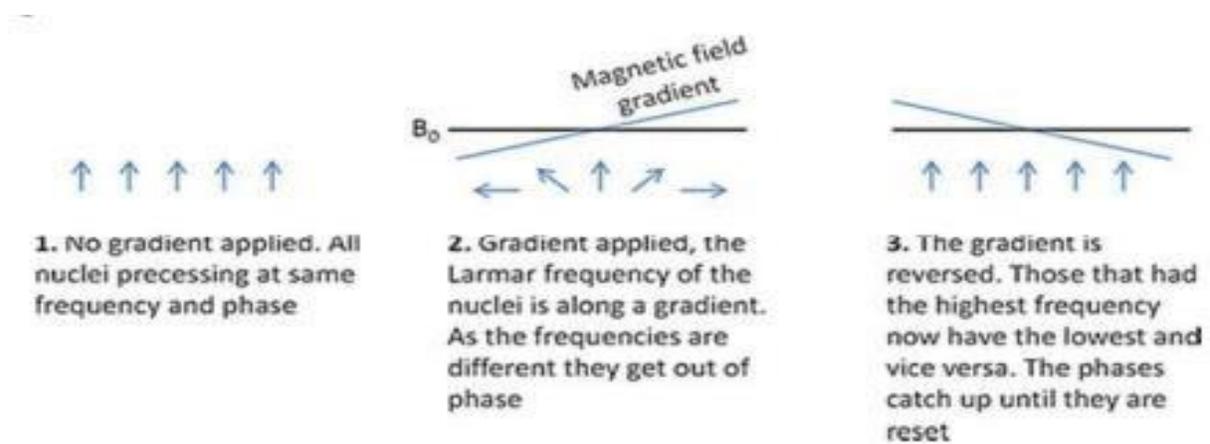

The magnitude of B_0 is altered in a linear fashion by the gradient coils, so that the magnetic field strength and therefore the precessional frequency experienced by nuclei situated along the axis of the gradient can be predicted. This is called spatial encoding.

- Selective excitation of an image slice by applying a shaped RF pulse and a field gradient at the same time.

Dephasing and Rephasing:

If we apply a gradient G for a given time, what will happen to the MR signal? There will be variations in frequency of the MR signal, either faster or slower, depending upon position.

The spins which are precessing faster, because of the action of a gradient, appear to move apart or dephase, and those which are precessing slower dephase in the opposite direction.

1. No gradient applied. All nuclei precessing at same frequency and phase

2. Gradient applied, the Larmar frequency of the nuclei is along a gradient. As the frequencies are different they get out of phase

3. The gradient is reversed. Those that had the highest frequency now have the lowest and vice versa. The phases catch up until they are reset

Before selecting the next slice we need to reset the nuclei. This is done by temporarily reversing the gradient to reverse the precessional frequencies. The nuclei then rephase.

Slice thickness:

In practice, the system automatically applies the appropriate gradient slope and transmit bandwidth according to the required slice thickness.
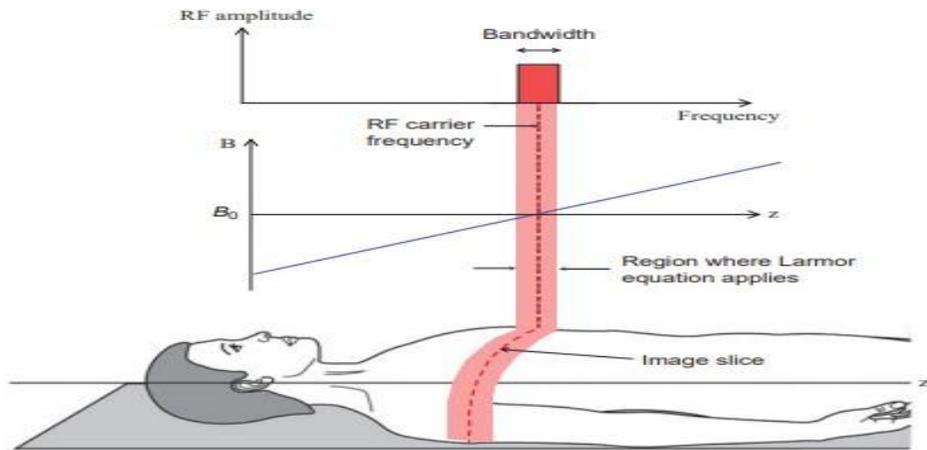
The thickness of the slice can be controlled by changing either the shape of the designer RF pulse (changing its bandwidth) or the strength of the gradient.

The slice thickness can be determined by:

(1) the steepness of the slope of the gradient field (Gs) and
(2) the bandwidth of the 90o RF-pulse.
The slice thickness is governed by the following equation:

$$thk = \frac{BW_{trans}}{\gamma_0 . G_s}$$

Where;
thk is the slice thickness.
BW_{trans} is the transmitted RF bandwidth (the range of frequencies it covers).
γ_0 is the gyromagnetic ratio.
G_s is the magnitude of the slice selection magnetic field gradient.
(H.W) For example:

Using a gradient magnetic field strength of ($10mTm^{-1}$), the transmitted RF pulse bandwidth would be about 4.3 kHz (using γ_0 = 42.58 MHz T $^{-1}$).

If we increase our slice thickness, we will be including many more types of tissues in to our slice. This may cause an image blurring artifact called partial voluming artifact.

Image Resolution:

In order to get optimal image resolution, must be very thin slices with a high signal to noise ratio (SNR). But whenever were thinner slices the noise was more, the SNR decreases and spatial resolution increase.
Spatial resolution is the ability to distinguish between different tissues neighboring each other.

Effects the poorer SNR of thin slices can be addressed for to some extent by increasing the number of acquisitions (acquisitions: the process of measuring and storing image data.) or by a longer TR. Yet this is accomplished only at the expense of the overall image acquisition time (the period of time required to collect the image data. This time does not include the time necessary to reconstruct the image. ADC - analog-to-digital converter) and reduces the cost efficiency of the MR imaging system.

For example, if we were to have tissue associated with cerebral spinal fluid (CSF) and brain tissue within one slice, areas where these structures overlap will be blurred because each will produce different signal intensity.
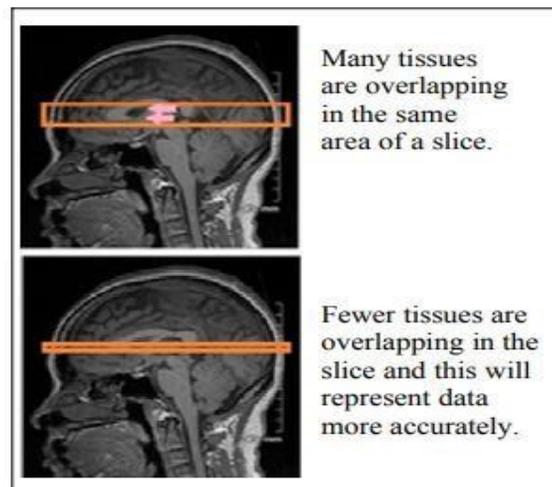

- By decreasing our slice thickness, we will be creating a more detail and better representation of the anatomy that we are sampling.
- Thinner slices will decrease the size of our voxel size. Smaller voxels will increase our resolution.

Therefore, as mentioned previously, a typical slice thickness is 2-10 mm

Bandwidth:

The RF excitation pulse contains the same range of frequencies to match the difference in precessional frequency.

Large bandwidth = large range of frequencies = larger slice
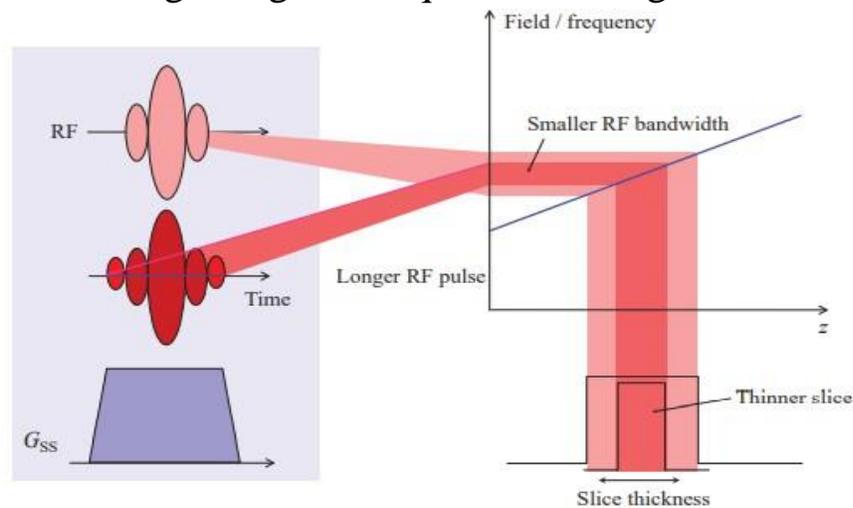
We can use a narrower RF pulse bandwidth. According to Fourier theory, this means using a longer duration RF pulse. can have a thinner slice but it will take longer to excite.

Gradient Strength:

If we use the same range of radio frequencies, the same band width as it is called, slice thickness can be modified by the slope of the gradient field.

A **stronger gradient** will result in **a thinner slice**
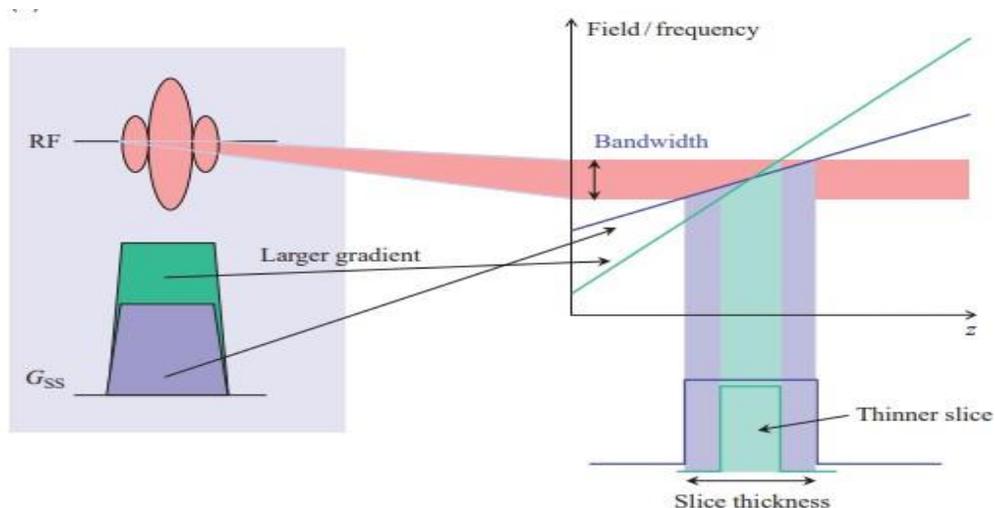
Inter-slice gap:

Due to imperfections in the RF pulses we usually have to introduce a slice gap to separate the slices. This is measured as the distance betweenthe slice edges. we generally try to keep the slice gap to a minimum, since tissues in the gap are not imaged at all.
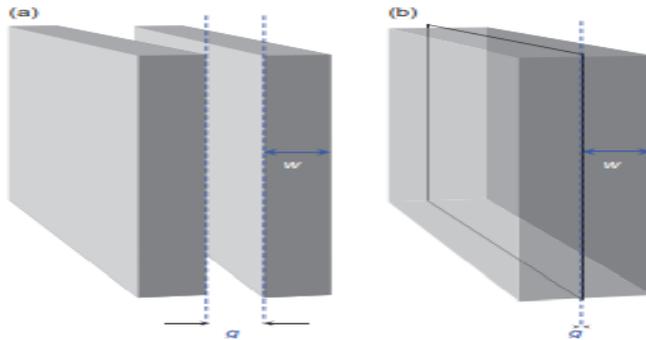
The RF pulse for one slice also excites protons in adjacent slices. Such interference is known as cross-talk. That is, when radio frequency pulse for one slice stimulates protons in the adjacent slices. This will lead to a reduction SNR.

1. The Larmor equation is:

A. $\omega_0 = B_0/\gamma$
B. $\omega_0 = \gamma B_0$
C. $\omega_0 = \gamma/B_0$
D. $\omega_0 = B_0^2\gamma$
E. $\omega_0 = \gamma^2 B_0$

2. The center of the magnet is called:

A. Isofrequency
B. Isocenter
C. Nodal point
D. Resonance center
E. Gradient origin

3. . Slew rate is calculated by:

A. Rise time / Gradient strength
B. Gradient strength $\times$ rise time
C. Gradient strength / rise time
D. Rise time $\times$ magnetic field
E. RF bandwidth / gradient

4. Dephasing occurs when:

A. Spins precess at the same frequency
B. Spins align perfectly
C. Spins precess at different frequencies
D. RF pulse stops
E. TE becomes zero

5. Partial volume artifact occurs when:

A. Slice is too thin
B. Slice is too thick
C. TE is short
D. TR is long
E. Field strength is low