physics of diagnostic radiology

Lecture 10

First semester

2025-2024

1.10. In-Phase and Diphase

To explain first what do we mean by **Phase**? Through the following simple example can illustrate Phase.

- In Figure 1 below, (a) the wheels rotate in the same speed and in the same angle. The arrows will therefore point in the same direction at any time. Say the wheels rotate in the same phase (**in-Phase**). Another two wheels (b) with different angle therefore, we say out of phase (**de-phase**).

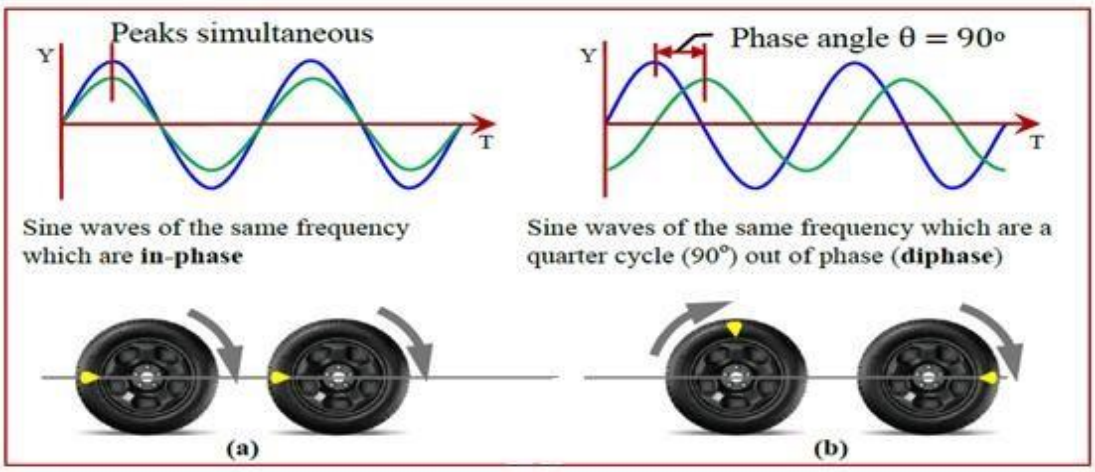

Figure 13: (a) two wheels are rotating **in-Phase** (b) two wheels are rotating **de-phase.**

1.11 RF Pulse

If the person placed in the MRI scanner, the first thing which happens is the precession (Precession is a change in the orientation of the rotational axis of spins around the direction of the external magnetic field (the z-axis)).

The spin precesses around a circle in the xy-plane and the net magnetization is the length, amplitude or magnitude of the magnetization vector M_{xy}: the quantity normally represented on a pixel-by-pixel basis in a MR image (thus amplitude or magnitude image) will be a complicated motion due to precession from the **two static fields**. However, if the second magnetic field which is temporarily applied is oscillating with the frequency of precession of the precessing spins a simple rotation of the net magnetization vector results. (Rotation of magnetization into the x-y-plane is a "90° pulse". The dephasing of the components of magnetization in the x-y-plane starts to occur straight away, as does the re-growth of magnetization in the z-direction as shown in next section).

1.11 Excitation

Once the Larmor frequency is determined the system will start the acquisition. The oscillating magnetic field at the Larmor frequency is switched on for a very small amount of time (a few milliseconds) to achieve such a rotation. This magnetic field is called an RF pulse; it is short (a burst or pulse) and the Larmor frequency for MRI is in the radio frequency range (tens of MHz). This process is sometimes called RF excitation of the spin system. Different amounts of rotation can be achieved by applying the oscillating magnetic field for different durations.

To understand it more deeply can through the following example:

Let us assume we work with a 1.5 Tesla system. The centre or operating frequency of the system is 63.855 MHz. In order to manipulate the net magnetization we will therefore have to send a Radio Frequency (RF) pulse with a frequency that matches the centre frequency of the system: 63.855 MHz. This is where the Resonance comes from in the name Magnetic Resonance Imaging. Only protons that spin with the same frequency as the RF pulse will respond to that RF pulse. If we send an RF pulse with 59.347 MHz, nothing would happen. Therefore, by sending an RF pulse at the **Larmor Frequency**, with certain strength (amplitude) and for a certain period of time it is possible to rotate the **net magnetization** into a plane perpendicular to the Z-axis, in this case the X-Y plane (Figure 14).

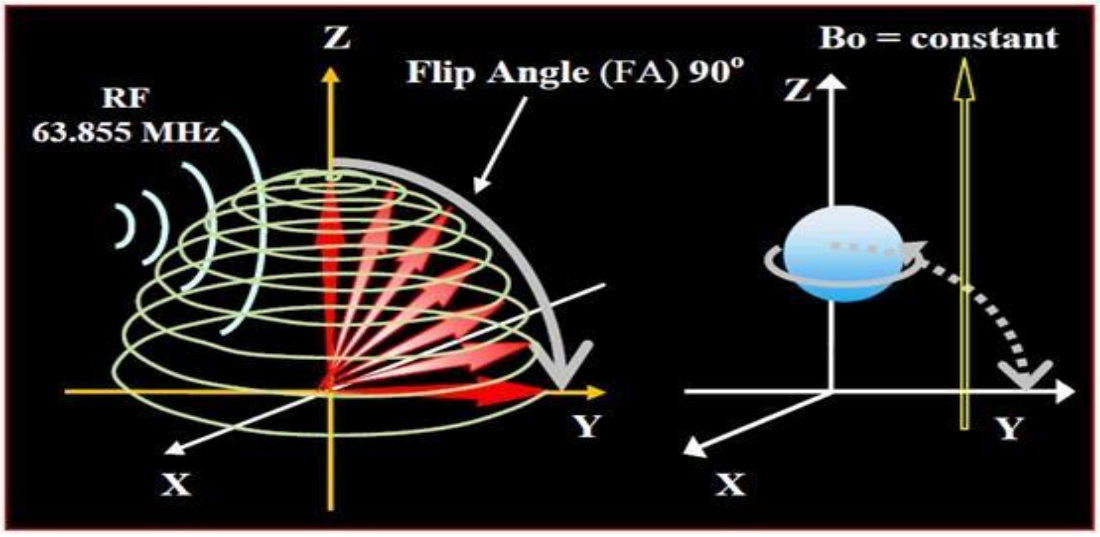

Figure 14: RF pulse at the Larmor Frequency, it is possible to rotate the net magnetization into a plane perpendicular to the Z-axis.

We just ―flipped‖ the net magnetization 90°. Later we will see that there is a parameter in our pulse sequence, called the **Flip Angle** (FA), which indicates the amount of degrees we rotate the net magnetization. It is possible to flip the net magnetization any degree in the range from 1° to 180°. For now we only use an FA of 90°. This process is called **excitation**.

1.12 Relaxation

If the net magnetization rotated 90 degrees in x-y plane and this means the same thing if we say that the protons raised to a **higher energy state**. This occurs because the protons absorbed energy from the **RF pulse**. This is called the perturbation that protons do not "like or want" continue in high energy situation "**excitation**" they tend to return to the normal or low energy situation "**equilibrium**". This can be compared with the abnormal situation in the case of walking on your hands, this is possible, but you do not want to continue this case for a long time and you inevitably you prefer the natural state is walking on your feet. A general principle of thermodynamics is that every system seeks its lowest energy level. The **relaxation** means the return of a perturbed system into the original situation "equilibrium" and each relaxation process can be characterized by a relaxation time. The relaxation process can be divided into two parts: **T1** and **T2 relaxation**.

1.12.1 T1 Relaxation

T1 is spin-lattice relaxation time which relates to the recovery of the magnetization along z direction **after RF pulse**. We can say that this as the time it takes tissue to recover from an RF pulse so you can give another pulse and still get signal. T1 is called the **spin-lattice relaxation time** because it refers to the time it takes for the **spins** to give the energy they obtained from the RF pulse back to the surrounding tissue (**lattice**) in order to go back to their equilibrium state. T1 relaxation describes what happens in the Z direction. After the magnetization has been flipped 90° into the x–y plane, the RF pulse is **turned off**. Therefore, after the RF pulse is turned off, two things will occur:

1. The spins will go back to the lowest energy state.
2. The spins will get **out of phase** with each other.

The Protons are returning to its original situation "equilibrium" by the releasing the absorbed energy in the form of "very little" RF waves. That means, in principle the net magnetization rotates back to align itself with the Z-axis. After the stops of the RF excitation pulse, the net magnetization will re-grow along the Z-axis, while emitting radio-frequency waves (Figure 15).

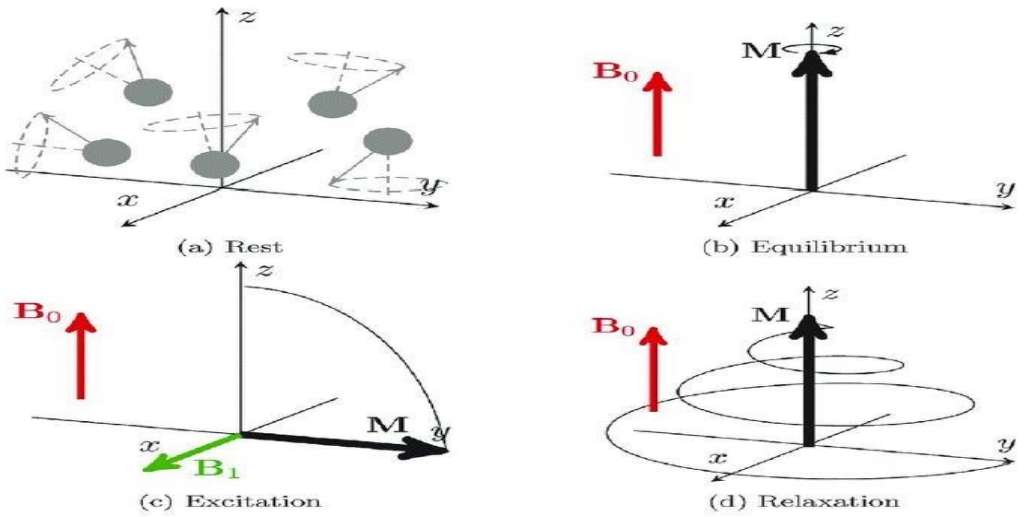

Figure 15: The net magnetization will re-grow along the Z-axis after the stops of the RF excitation pulse.

T1 Relaxation Curves

T1 relaxation happens to the protons in the volume that experienced the 90°-excitation pulse. However, not all the protons are bound in their molecules in the same way. This is different for each tissue. One ^{1}H atom may be bound very tight, such as in fat tissue, while the other has a much looser bond, such as in water. Tightly bound protons will release their energy much quicker to their surroundings than protons, which are bound loosely. The rate at which they release their energy is therefore different. The rate of T1 relaxation can be depicted as shown in Figure 16.

The curve shows at time = 0 that there is no magnetization in the Z-direction right after the F-pulse. But immediately the M_Z starts to recover along the Z-axis. T1 is defined as the time it takes for the longitudinal magnetization (M_Z) to reach 63 % of the original magnetization. In other words, magnetization (M_Z) be at the beginning and before sending a pulse on the Z-axis with the maximum value (100%), after sending a pulse, M_Z go down to zero on the Z-axis and a full appear in the xy plane. This means that the protons are in an excited state.

Immediately after cutting-off the pulse, begin declining in the xy plane and at the same time grow on the Z-axis to reach 63% according to the following equation:

$$M_Z = M_O(1-e^{-t/T1})$$

For example; After $t = T_1$

$$M_z = M_o(1 - e^{-T_1/T_1})$$
$$= M_o(1 - e^{-1}) \quad \text{where} \quad e \cong 2.72$$
$$= 0.63M_o \quad \text{or} \quad 63\%\ M_o$$

After $t = 5T_1$

$$M_z = M_o(1 - e^{-5T_1/T_1})$$
$$= M_o(1 - e^{-5})$$
$$= 0.99\ M_o \quad \text{or} \quad 99\%\ M_o$$

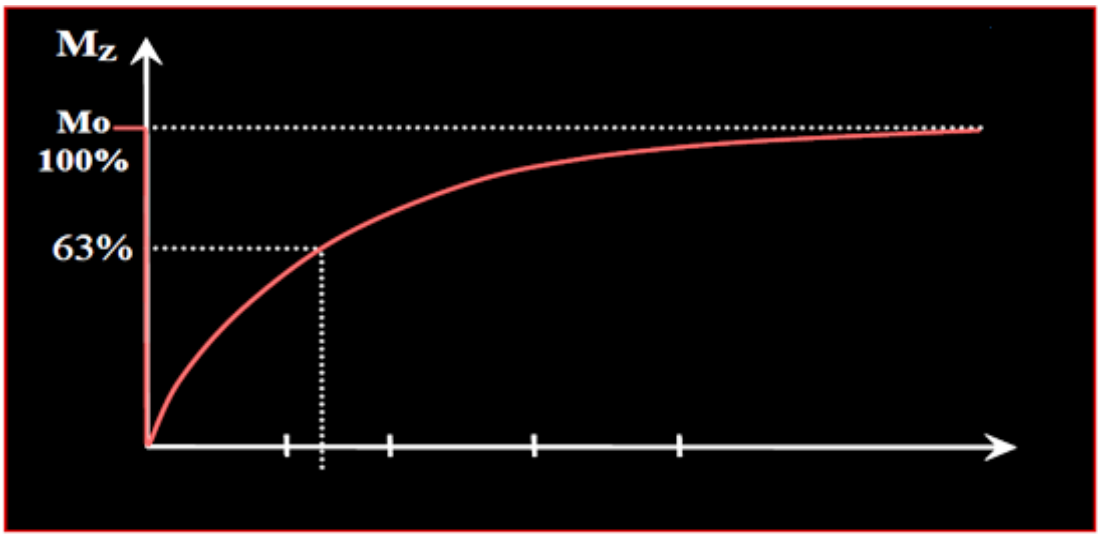

Figure16: The rate of T1 relaxation.

A similar curve can be drawn for each tissue as shown in figure 17 which illustrates four tissues found in the head. Each tissue will release energy (relax) at a different rate and that's why MRI has such good contrast resolution.

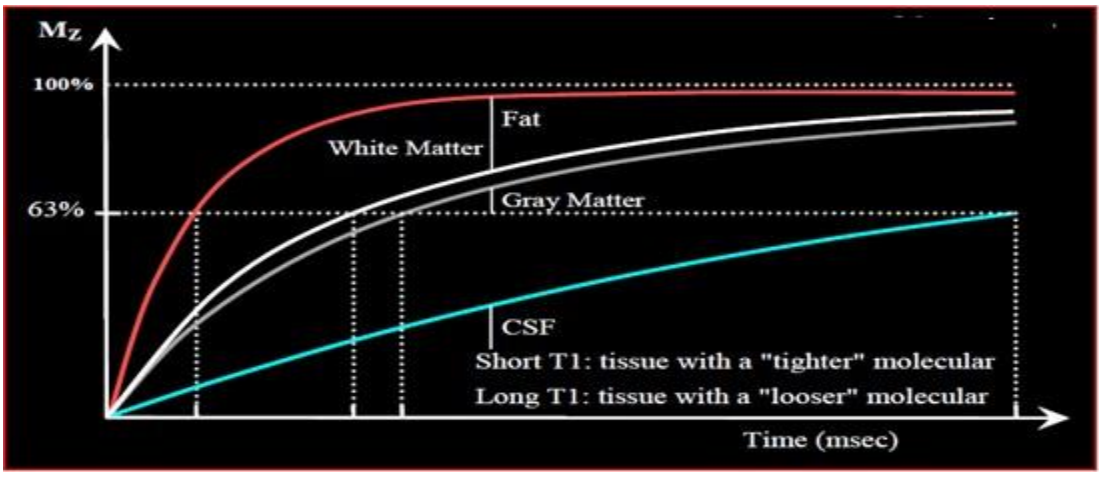

Figure 17: Example T1 curves for four tissues found in the head.

1.13.2 T_2 Relaxation

First of all, it is very important to realize that T1 and T2 relaxation are **two independent processes**. **The one has nothing to do with the other**. The only thing they have in common is that both processes happen simultaneously. T1 relaxation describes what happens in the Z direction, while T2 relaxation describes what happens in the X-Y plane. That's why they have nothing to do with one another. Each individual proton is spinning around its own axis. Although they may be rotating with the same speed, they are not spinning **in-phase** or, in other words, there is no **phase coherence**. When we apply the 90° RF pulse something interesting happens. Apart from flipping the magnetization into the X-Y plane, the protons will also start spinning **in-phase**. So, right after the 90° RF pulse the net magnetization vector (now called **transverse magnetization**) is rotating in the X-Y plane around the Z-axis at the Larmor frequency (Figure 18 A).

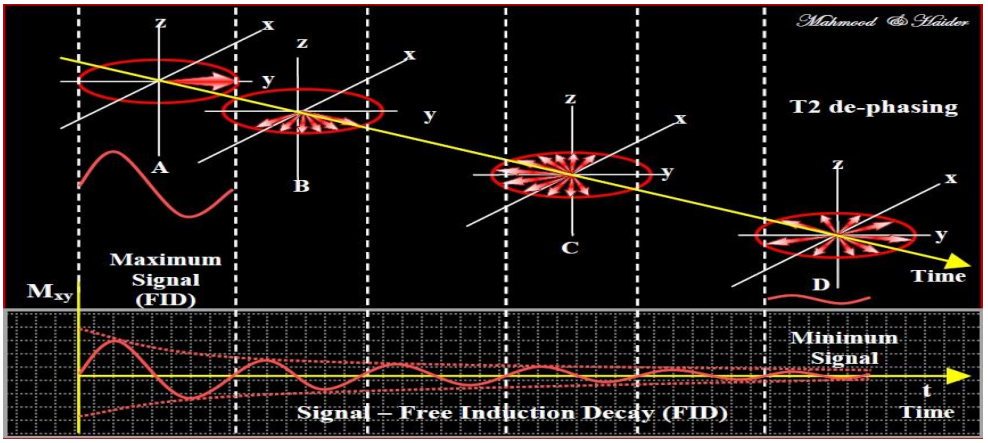

Figure 18: De-phasing and free induction decay (FID).

The transverse magnetization induces an MR signal in the radiofrequency coil immediately after its formation, it has a maximum magnitude, and all of the protons are in phase. Therefore the vectors all point in the same direction because they are **in-Phase**. However, they don't stay like this. The transverse magnetization starts decreasing in magnitude immediately as protons start going out of phase. This process of **de-phasing** and reduction in the amount of transverse magnetization is called transverse relaxation. At first the amount of de-phasing will be small (Figure 18 B, C), but quickly that will increase until

there is no more phase coherence left: there is not one vector pointing in the same direction anymore.

In the meanwhile the whole lot is still rotating around the Z-axis in the X-Y plane (Figure 18 D). A characteristic time representing the decay of the signal 37%, is called the T2 relaxation time. This process of getting from a total **in-phase** situation to a total **out-of-phase** situation is called **T2 relaxation**.

T2 Relaxation Curves

T_1 relaxation, T_2 relaxation does not happen at once. Again, it depends on how the Hydrogen proton is bound in its molecule and that again is different for each tissue. Right after the 90° RF-pulse all the magnetization is —flipped‖ into the xy-plane. The net magnetization changes name and is now called M_{xy}. At time = 0 all spins are in-phase, but immediately start to de-phase. T2 is defined as the time it takes for the spins to de-phase to 37% of the original value. Immediately after cutting-off the pulse, begin declining in the xy plane, according to the following equation:

$$M_{xy} = M_O e^{-t/T2}$$

The rate of **de-phasing** is different for each tissue. Fat tissue will de-phase quickly, while water will de-phase much slower (Figure 19)

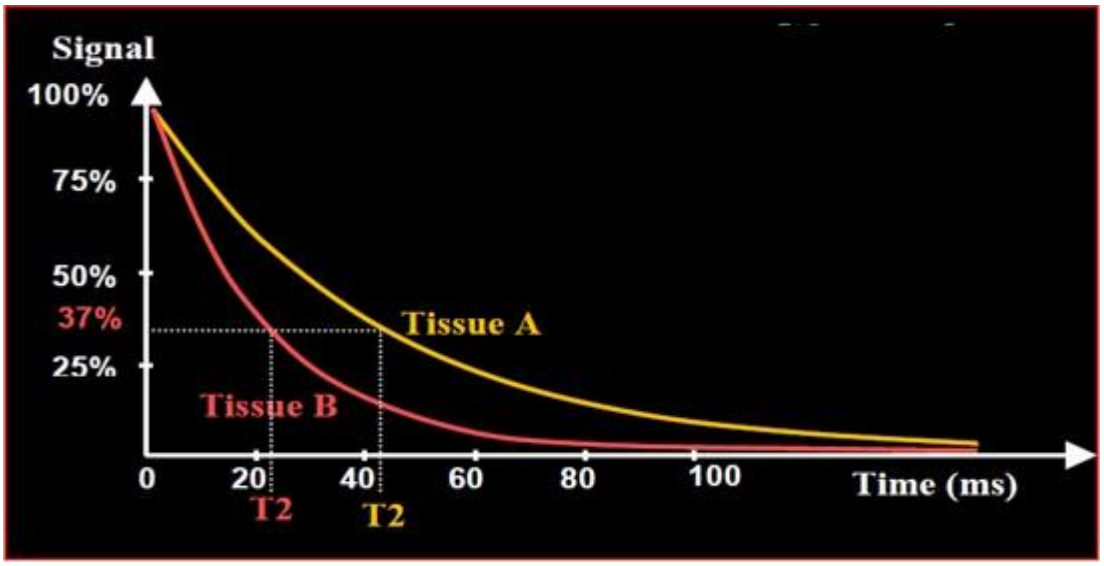

Figure 19: Transverse magnetization decay (T2)

One more remark about T2: it happens much faster than T1 relaxation. T2 relaxation happens in tens of milliseconds, while T1 can take up to seconds. T2

relaxation is also called spin–spin relaxation because it describes interactions between protons in their immediate surroundings (molecules).

Table 2: Approximate spin density (SD) and relaxation times (Tl, T2) for various tissues.

Tissue	SD	Tl (ms)	T2 (ms)
Water	100	2700	2700
Skeletal muscle	79	720	55
Cardiac muscle	80	725	60
Liver	71	290	50
Fat	-	360	30
Bone	<12	<100	<10
Spleen Kidney	79 81	570 505	50
Gray matter	84	4053/2	105
White matter	70	345	65

1.14 T2* Relaxation

All relaxation mechanisms mentioned so far are heavily influenced by temperature and molecular environment. Transverse relaxation is the result of random interactions at the atomic and molecular levels. Transverse relaxation is primarily related to the intrinsic field caused by adjacent protons (spins) and hence is called spin-spin relaxation. Transverse relaxation causes irreversible de-phasing of the transverse magnetization.

By contrast, the so-called T2* relaxation is a result of dephasing processes due to an inhomogeneous magnet field which can be minimized by manual justification ("shimming"). Since T2* is usually much smaller than T2, the signal decay of an FID is almost completely caused by T2* effects. In general, T1>T2>T2*.

Remember this:

- T1 and T2 relaxation are two independent processes, which happen simultaneously.
- T1 happens along the Z-axis; T2 happens in the X-Y plane.
- T2 is much quicker than T1

The differences in magnetic susceptibility among various tissues or materials, chemical shift, and gradients applied for spatial encoding. This de-phasing can be eliminated by using a 180° pulse, as in a spin-echo sequence. Hence, in a spin-echo sequence, only the —true‖ T2 relaxation is seen as shown in figure 20.

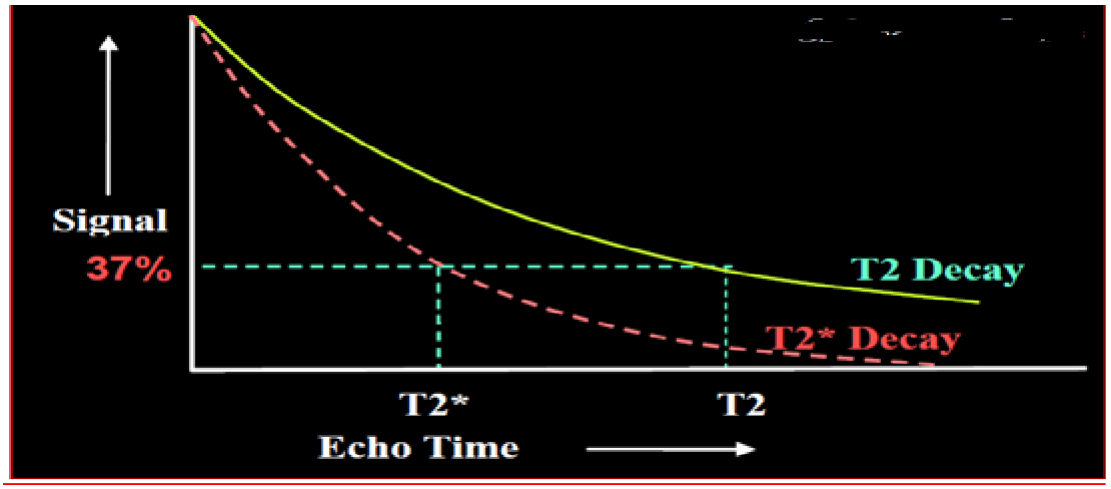

Figure 20: graph shows T2 andT2* relaxation curves. T2* is shorter than T2.

Ch. 5

2. 1 Acquisition

During the relaxation processes the spins shed their excess energy, which they acquired from the 90º RF pulse, in the shape of radio frequency waves. In order to produce an image we need to pick up these waves before they disappear into space. This can be done with a **Receive coil**. The receive coil can be the same as the **Transmit coil** or a different one.

The receive coil must be positioned at right angles to the main magnetic field. Failing to do so will result in an image without signal. This is why: if we open up a coil we see it is basically nothing but a loop of copper wire (see Figure 1).

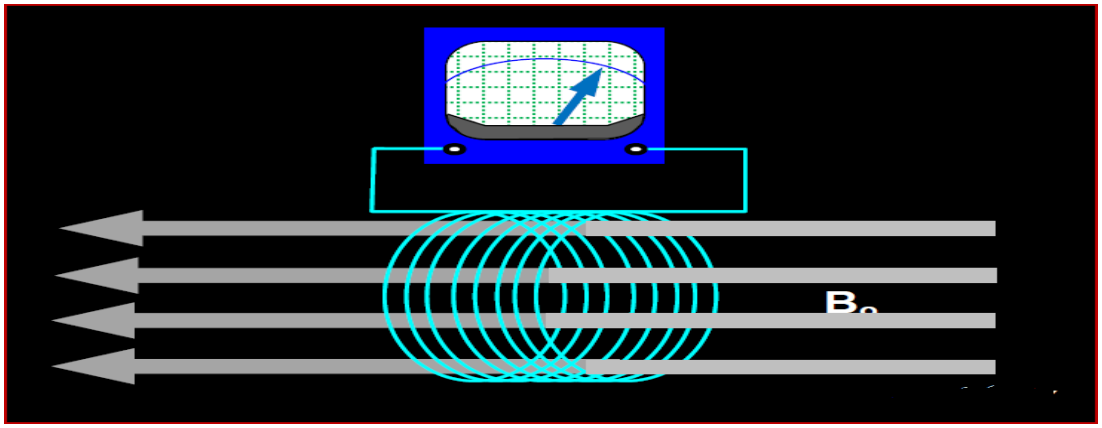

Figure 1: A magnetic field goes through the loop, a current is induced.

B_o is a very strong magnetic field; much stronger than the RF signal we are about to receive. That means if we position the coil such that B_o goes through the coil an enormous current is induced, and the tiny current induced by the RF

wave is overwhelmed. We will only see a lot of speckles (called: noise) in our image. Therefore, we have to make sure that the receive coil is positioned in such a way that B_0 can't go through the coil. The only way to achieve this is to position the receive coil at right angles to B_0 as shown in Figure 2.

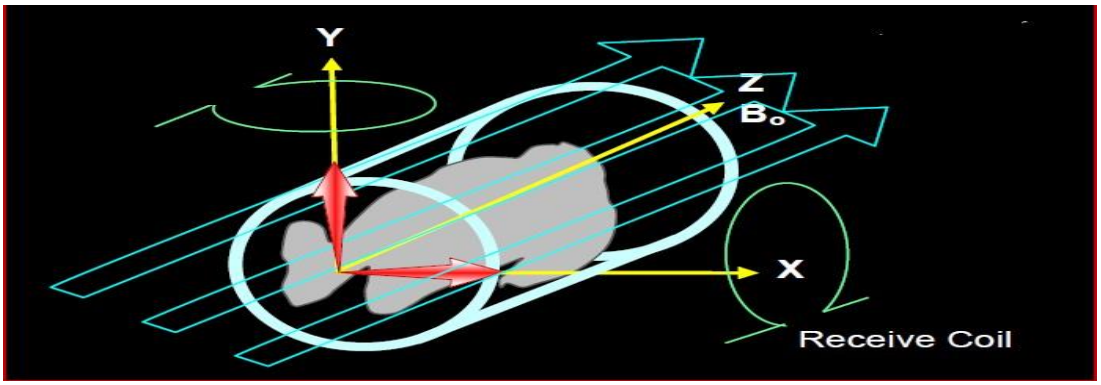

Figure 2: The receive coil at right angles to B_0.

Designing a coil for a bore type magnet where B_0 runs through the length of the body is exceptionally difficult. If you open up a Head coil you'll see probably two copper wires, which are saddle shaped and positioned at right angles to one another. In order to receive enough signal there are two coils, because saddle shaped coils are relatively inefficient.

2.2 Computing and Display:

In general, system of MRI consists of five major components: **magnet**, **gradient systems**, **RF coil system**, **receiver**, and **computer system**. Figure 3 shows the entire process graphically. The received signal is then fed into a computer.

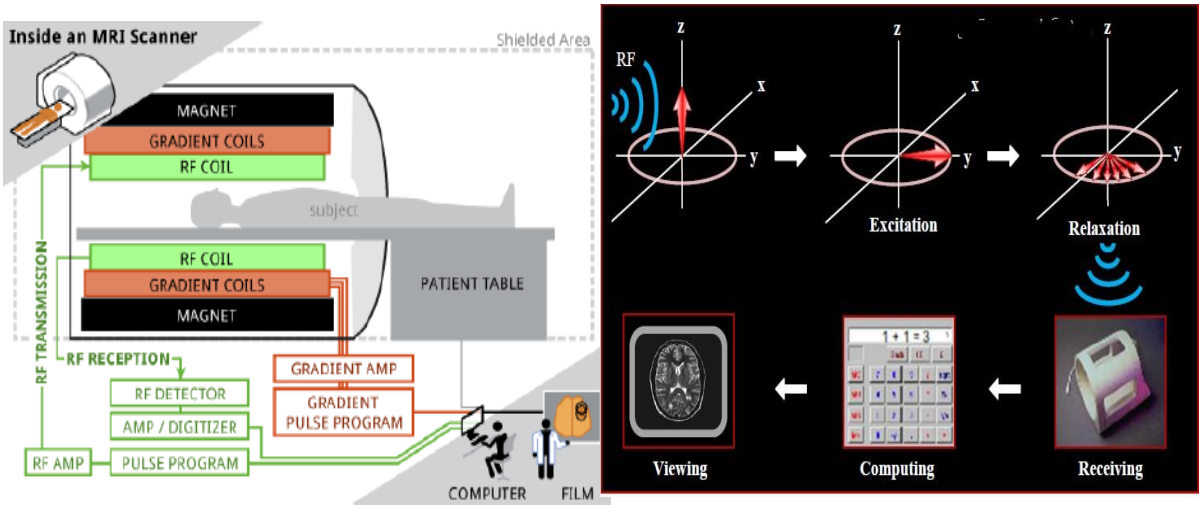

Figure 3: General Major Components of MRI system.

2.3 Gradient Coils:

As we explained previously to produce an image, you must stimulate the hydrogen nuclei in the body, and then determine the location of those nuclei within the body. These tasks are accomplished using the **gradient coil** (The MRI gradient coil is an essential component that modulates the primary magnetic field to provide spatial encoding during imaging). If we assume a completely homogeneous magnetic field (this ideal situation does not exist), then all the protons in the body will **spin** at the **Larmor frequency**. This also means that all protons when you return to equilibrium give the same signal. In this case we will not know whether the signal coming from the head or foot.

The solution to our problem can be found in the characteristics of the RF-wave, which are: **Phase, Frequency** and **Amplitude**

First, we will divide the body up to the volume elements, also known as: **voxels**.

- The protons within that voxel will emit RF wave with known **phase** and **frequency.**
- **Amplitude** of the signal depends on the amount of protons in the voxel.

The answer to our problem is: **Gradient Coils**

The **gradient coils** are resistant type electromagnets, which enable us to create additional magnetic fields, which are, in a way, superimposed on the main magnetic field B_0. The gradient coils are used to spatially encode the positions of the MRI spins by varying the magnetic field linearly across the imaging volume such that the **Larmor frequency** varies as a function of position.

To achieve adequate image quality and frame rates, the gradient coils in the MRI imaging system must rapidly change the strong static magnetic field by approximately 5% in the area of interest. High-voltage (operating at a few kilovolts) and high-current (100s of amps) power electronics are required to drive these gradient coils. Notwithstanding the large power requirements, low noise and stability are key performance metrics since any ripple in the coil current causes noise in the subsequent RF pickup. That noise directly affects the integrity of the images. To differentiate tissue types, the MRI systems analyze the magnitude of the received signals. Excited nuclei continue to emit a signal until the energy absorbed during the excitation phase has been released. The

time constant of these exponentially decaying signals ranges from tens of milliseconds to over a second; the recovery time is a function of field strength and the type of tissue. It is the variations in this time constant that allow different tissue types to be identified.

2.4. Signal Coding

Signal coding (it is the process of representing an information signal in a way that realizes a desired communications objective such as analog-to-digital conversion, low bit rate transmission, or message encryption) To explain this subject clearly and easy assimilation, suppose some of the assumptions: considering an axial image of the brain using a 1.5 Tesla magnet. Also we work with a homogeneous magnetic field, which covers the whole body from head to toe. Just as important as the strength of the main magnet is its precision. The straightness of the magnetic lines within the centre of the magnet needs to be near-perfect. This is known as homogeneity. Inhomogeneities in the field strength within the scan region should be less than three parts per (3 ppm). When we put a patient in the magnet, all the protons, from head to toe, align with B_0. They spin at the Larmor frequency of 63.6 MHz. (Figure 4).

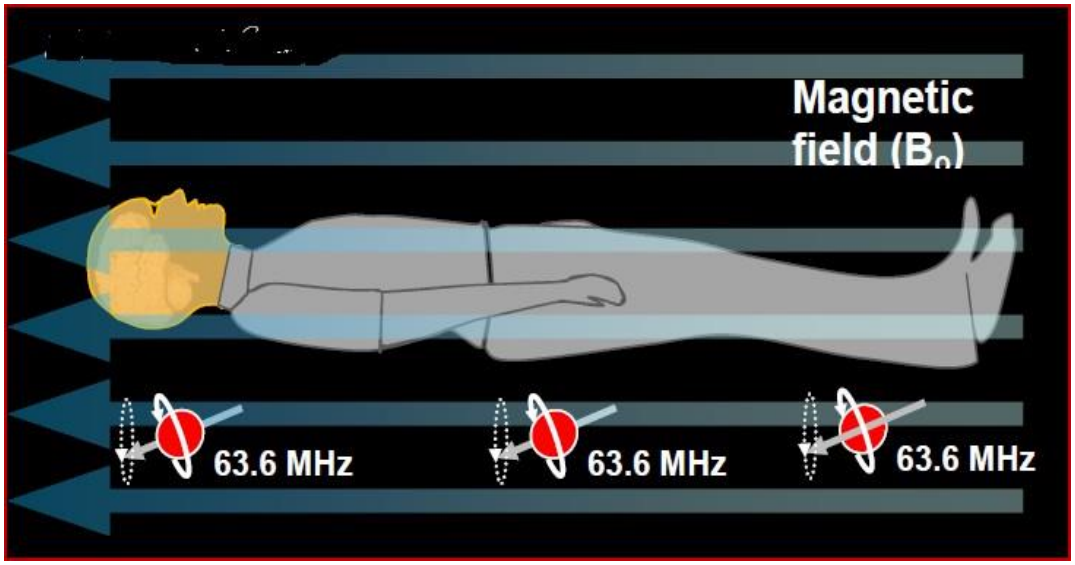

Figure 4: Patient in the magnet, all the protons align with B_0 and they spin at the Larmor frequency.

If we use a 90° excitation RF-pulse to flip the magnetization into the x-y plane, then all the protons would react and return a signal. We would have no clue where the signal comes from: from head or toe.

2.5 Slice Encoding Gradient

The magnetic field gradient (e.g. Z-gradient) is temporarily applied (Z-gradient is switched on) at the same time as the RF pulse (see figure 5).

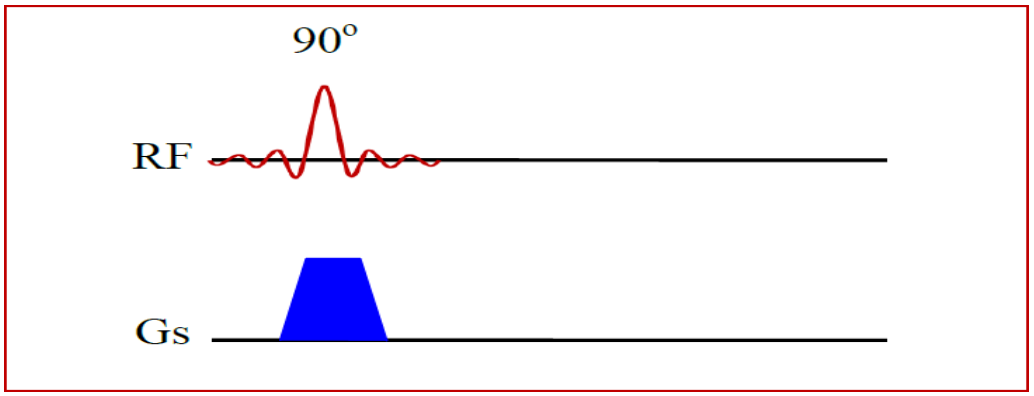

Figure 5: Slice selection pulse sequence, a gradient magnetic field is applied at the same time as the RF pulse.

This will generate an additional magnetic field in the Z-direction, which is superimposed on B_o. The indication +Gz in Figure 13 means there is a slightly stronger B_o field in the head as there is in the isocentre of the magnet. A stronger B_o field means a higher Larmor frequency. Along the entire the slope of the gradient there is a different B_o field and consequently the protons spin at slightly different frequencies. Therefore, the protons in the head will spin slightly faster than the ones in the isocentre. The reverse goes for the protons in the feet. Figure 14 shows that the protons in the feet now spin at 63.5 MHz, the ones in the isocentre of the magnet still at 63.6 MHz and the ones in the head with 63.7 MHz.

This means we can "pick out" the section which we want to excite by choosing the right frequency range of RF excitation pulse. The section which contains Larmor frequencies which match the frequencies of the oscillating magnetic field will respond. An MRI signal will be generated only from that section of the patient. This is called **Slice-Encoding** or **Slice-Selection**. Usually the slice selection gradient is applied in the z-axis—the head-foot direction in the scanner. But because a gradient magnetic field may be applied in any orientation, slices may be acquired at literally any angle or orientation in the patient. This is one of strength MRI. Now, if we apply an RF-pulse with a frequency of 63.7 MHz only the protons in a thin slice in the head will react because they are the only ones

which spin with the same frequency. In this example Gz is the slice-encoding gradient. If we would stop here, this means that we receive the returned signal comes from the single slice in the head. That is we have identified the site by using the Z-gradient (Gz), as shown in Figure 6.

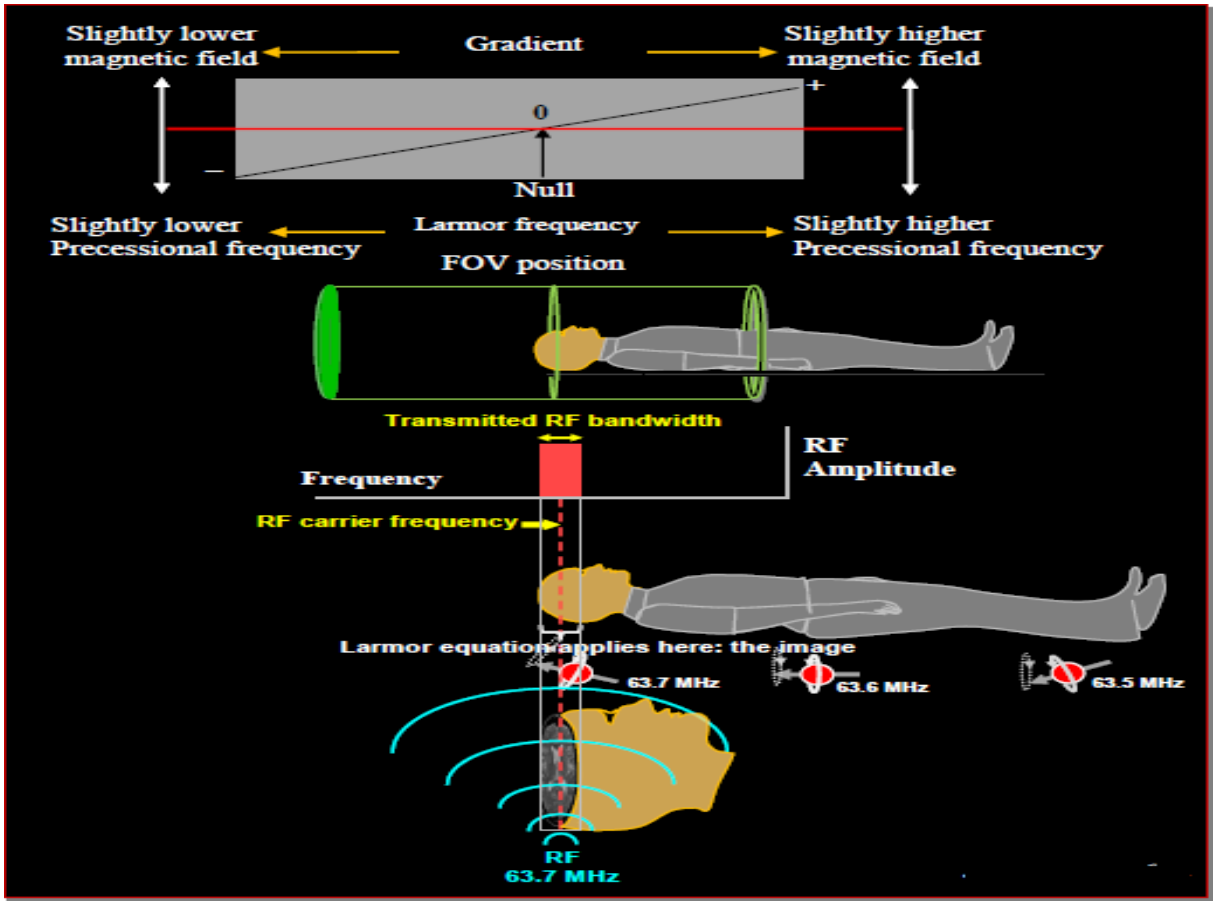

Figure 6: Generate an additional magnetic field in the Z-direction.

2.5.1 Slice Location

If there's time between imaging different slices, we can simply move the patient-table so that the section of interest within the patient is at the isocentre. This is preferable because placing the section of interest in the part of the main magnetic field which is most homogenous will give us better images.

- The carrier frequency of the RF excitation pulse may be changed as shown in figure 7.

The carrier frequency of the transmitted RF pulse determines which spins along the patient will resonate (because they have a matching Larmor frequency). If multiple slices are to be acquired in quick sequence, the carrier frequency can be set to determine the location of the imaging slice in the patient.

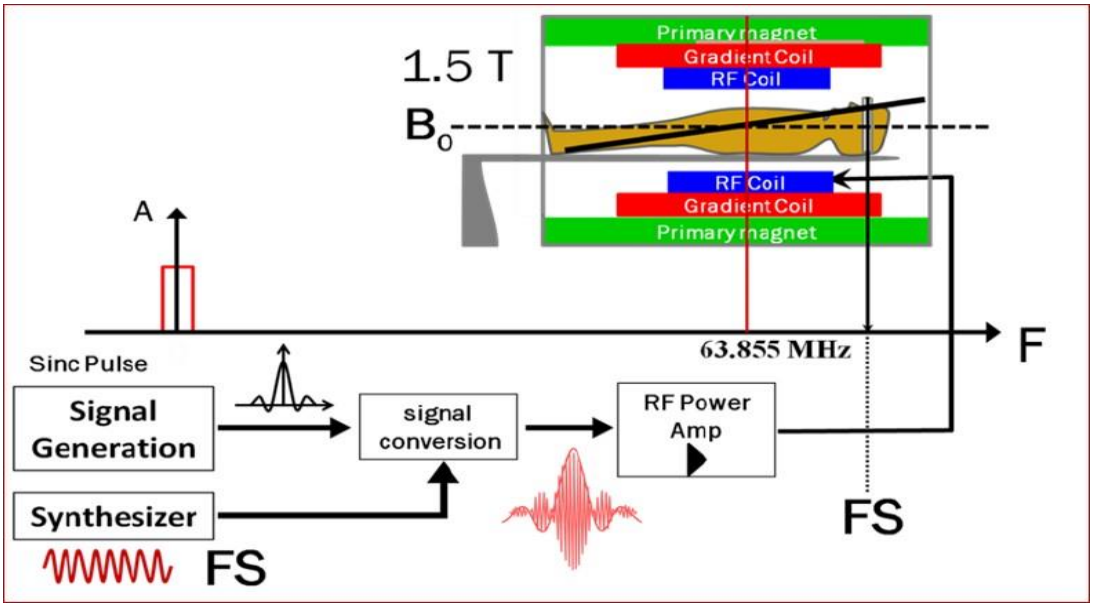

Figure 7: Process of choice the slice location

2.5.2. Slice Thickness

Slice thickness helps get better resolution and finer detailed images. The slice thickness is governed by the following equation:

$$\text{Thickness} = \frac{\text{BW}_{\text{trans}}}{\gamma_0 G_s}$$

Where BW_{trans} is the transmitted RF bandwidth (the range of frequencies it covers), γ_0 is the gyro-magnetic ratio and G_s is the magnitude of the slice selection magnetic field gradient.

In Figure 8A show that varying the steepness of the gradient, while keeping the RF-pulse bandwidth the same. Alternatively, Figure 8 B the steepness of the gradient is kept the same, while the bandwidth of the RF-pulse is varied. Can also change the slice thickness, the slice thickness may be reduced by either increasing the gradient of the magnetic field (dashed line in figure 8 A) or by decreasing the RF pulse width, (or transmit bandwidth, figures 8 B). A thinner slice produces better anatomical detail, the partial volume effect being less, but it takes longer to excite.

In practice, the slice thickness is determined by a combination of both gradient steepness and RF-pulse bandwidth.

The total magnetic field at a position Z_{SS} (SS = slice selection) during application of G_Z is given by: $B_0+Z_{SS}\cdot G_Z$ and the spatially selective excitation energy or frequency, respectively, can be easily calculated using the Larmor equation.

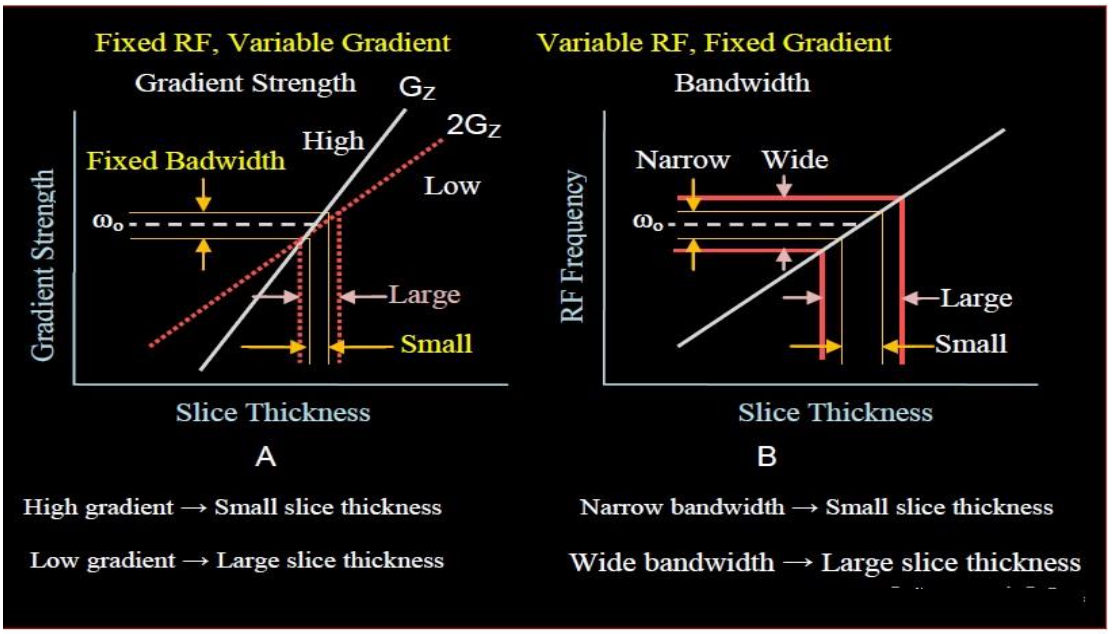

Figure 8: Slice thickness is dependent on RF bandwidth and gradient strength. (A) For a fixed gradient strength, the RF bandwidth determines the slice thickness, (B) for a fixed RF bandwidth, gradient strength determines the slice thickness.

A typical slice thickness is 2-10 mm. The RF pulse inevitably contains a certain amount of electromagnetic energy of frequencies slightly higher or lower than the intended bandwidth, thus mildly exciting tissues either side of the desired slice. To prevent this affecting the image slice, a gap (say 10% of the slice thickness) may be left between slices, although this is not necessary when the slices are interleaved.

For example, for a 10 mm slice thickness using a gradient magnetic field strength of (10 mmT / m), the transmitted RF pulse bandwidth would be about 4.3 kHz (using γ_0 = 42.58 MHz / T).

In order to get optimal image resolution, must be very thin slices with a high SNR. But whenever were thinner slices the noise was more, the SNR decreases and spatial resolution increases. Spatial Resolution is the ability to distinguish one structure from another. Conversely, increase of the slice thickness leads to increase signal to noise ratio and reduces spatial resolution. Because the thicker slices result other problems such as an increase in partial volume effects.

Effects the poorer SNR of thin slices can be addressed for to some extent by increasing the number of acquisitions 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.

2.5.3. Receiver Bandwidth

The receiver bandwidth is the range of frequencies collected by an MR system during frequency encoding. The bandwidth is either set automatically or can be changed by the operator. A wide receiver bandwidth enables faster data acquisition and minimizes chemical shift artifacts but also reduces SNR as more noise is included. Halving the bandwidth improves SNR by about 30%. With a narrow bandwidth, on the other hand, there will be more chemical shift and motion artifacts and the number of slices that can be acquired for a given TR is limited.

Is this enough? Certainly not, and we will know immediately why.

Figure 9 shows the axial slice, which has just been created by the Gz gradient. If we take a closer look at proton 1 and 2 in this slice we see that they both spin with the same frequency and have the same phase.

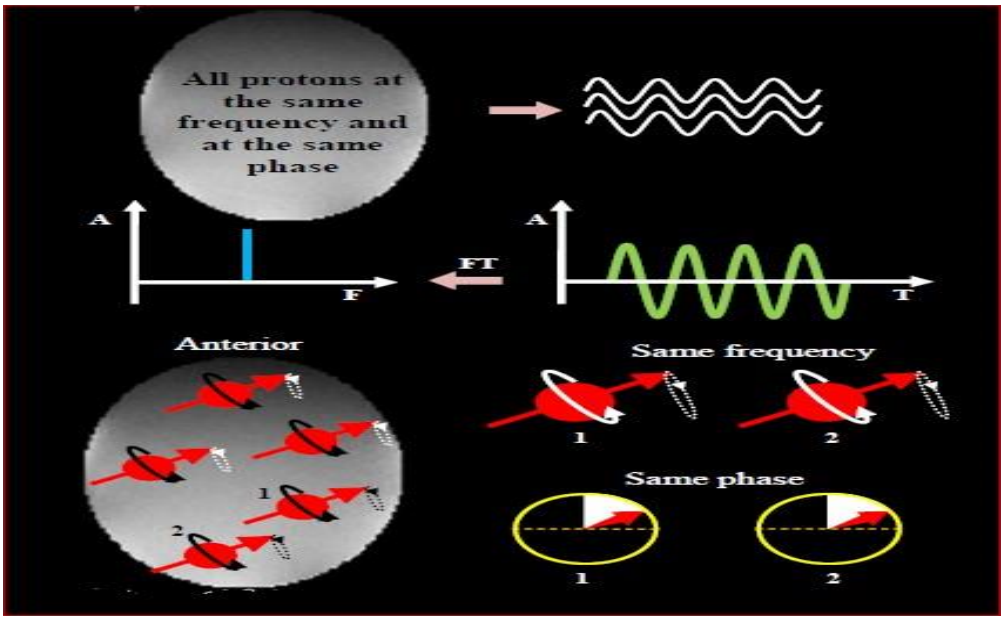

Figure 9: The axial slice, which has just been created by the Gz gradient.

Within the slice there are still an awful lot of protons and we still don't know from where the signal is coming from within the slice. Whether, it comes from anterior, posterior, left or right. Further encoding is therefore required in order to allow us to pinpoint the exact origin of the signals.