Let's Talk Physics

Introduction

It is difficult to decide where to start when you want to explain the physics of MRI. You could say, "start at the beginning", and you're right, that's where all good stories start. But with MRI physics it is a bit more difficult, because one first has to establish where the beginning is or, put in another way, how much do you want to know.
As the title already indicates this story is supposed to entice those people who are new to the business and who need to know the very basics of MRI physics. In one way this is easy to write up because I can leave out large chunks of physics. On the other hand it is very difficult because I have to assume you know nothing and yet I have to explain something complicated in an easy to understand manner. Believe me, it's not at all that easy.
But let that be my problem. After all, it's my job ☺.

Magnetization

Let us start the journey into MRI physics by looking around us. What do we see? Amongst a host of items which have nothing to do with MRI we see the earth. There are a few things we know about earth:

1. The earth is a giant ball that floats in space. Not randomly, but that's a different story.
2. The earth has a moon, which rotates around the earth.
3. The earth has an electrical charge. Whether it be positive or negative is not important.
4. The earth rotates (spins) around it's own axis. This is the interesting bit.
5. There's a heck of a lot of water on earth, around 70% worth, and most of it seems to be falling in my back garden while I'm writing this.

This giant, electrically charged and spinning ball is floating in space. Quite happily: nothing to worry about. From our physics lessons in school we may remember that a rotating electrical charge creates a magnetic field. And sure enough, the earth has a magnetic field, which we use to find our way from one place to another by means of a compass. The magnetic field strength of the earth is rather small: 30 μT at the poles and 70 μT at the equator. (Tesla is for magnetic fields what Ampere is for electric current).

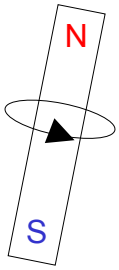


Figure 11

In short we can establish that the earth is a giant spinning bar magnet, with a north and a south pole (*Figure 11*). And don't forget it is wet, very wet.

You may wonder what all this has to do with MRI but I'll get to that in a moment.

Now let's have a look at ourselves, the Homo sapiens. What do we have in common with earth? Well on first sight not a lot, but when we take a bit from our body and we would put it under an electron microscope we can see things that look rather familiar. We see tiny little balls, which rotate around their own axes and also have an electrical charge and they have moons floating around it. What we are looking at are atoms. And atoms have everything to do with MRI, because we use them to generate our MR image.

Another thing we have in common with earth is water. Our body consists of 80% water.

From our chemistry lessons we know that there are many different elements, 110 to be precise. Because we exist mainly of water let's have a look at it. Water consists of 2 Hydrogen and 1 Oxygen atom. The hydrogen atom (the first element in the periodic table) has a nucleus, called proton, and 1 moon, called electron.
This proton is electrically charged and it rotates around its axis. There we have the analogy with the earth. Also the hydrogen proton can be looked at as if it were a tiny bar magnet with a north and a south pole.

Why do we take hydrogen as our MR imaging source?
There are two reasons. First off all we have a lot of them in our body. Actually it's the most abundant element we have. Secondly, in quantum physics there is a thing called "Gyro Magnetic Ratio". It is beyond the scope of this story what it represents; suffice to know that this ratio is different for each proton. It just so happens, that this gyro magnetic ratio for Hydrogen is the largest; 42.57 MHz/Tesla.

For who really wants to know, Hydrogen is not the only element we can use for MR imaging. In fact any element, which has an odd number of particles in the nucleus, can be used. Some elements, which can be used, are:

Table 1: MRI friendly elements

Isotope	Symbol	Spin Quantum number	Gyro Magnetic Ratio (MHz/T)
Hydrogen	^{1}H	1/2	42.6
Carbon	^{13}C	1/2	10.7
Oxygen	^{17}O	5/2	5.8
Fluorine	^{19}F	1/2	40.0
Sodium	^{23}Na	3/2	11.3
Magnesium	^{25}Mg	5/2	2.6
Phosphorus	^{31}P	1/2	17.2
Sulphur	^{33}S	3/2	3.3
Iron	^{57}Fe	1/2	1.4

If we look at a bunch of hydrogen protons (as in a molecule) we see, in fact, a lot of tiny bar magnets spinning around their own axes. As we may recall from classes, two north poles and two south poles of two magnets repel each other, while two poles of opposite sign attract each other.
In our body these tiny bar magnets are ordered in such a way that the magnetic forces equalize. Our bodies are, magnetically speaking, in balance.
Just as well, otherwise we would attract a lot of metal when we go about.

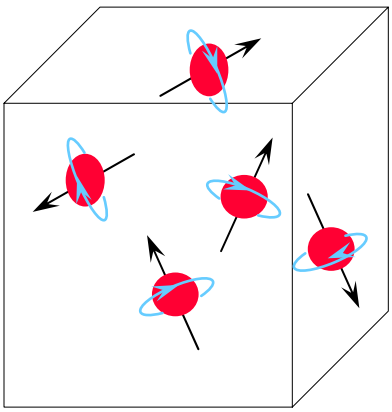


Figure 12

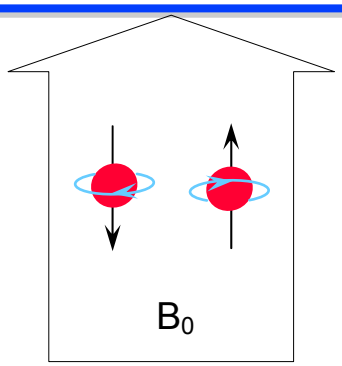
Now we have established some interesting facts of life, lets have a look at what happens when we go ahead and try to make an MRI examination.

As we have seen in the paragraph about the hardware, magnets used for MR imaging come in various field strengths. The magnetic field strength of a 1.5 Tesla magnet is ± 30.000 times stronger than the earth gravitational field! This indicates that we are working with

potentially dangerous equipment (more about that later).

When we put a person in a magnet some interesting things happen to the hydrogen protons:

1. They align with the magnetic field. This is done in two ways, parallel or anti-parallel.



(B_0 is the indication for the magnetic field of the MRI scanner)

Figure 13

2. They precess or "wobble" due to the magnetic momentum of the atom.

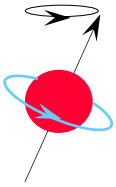


Figure 14

They precess at the Larmor frequency. This Larmor frequency is something, which needs a little further explanation. The Larmor frequency can be calculated from the following equation (don't worry, this will be the first of only two equations you will encounter in this story):

$$\omega_0 = \gamma B_0$$

Where: ω_0 = Precessional or Larmor frequency. (MHz)

γ = Gyro Magnetic Ratio. (MHz/T)

B_0 = Magnetic field strength. (T)

Here we see two things, which we discussed before, come together: the Gyro Magnetic Ratio and the Magnetic field strength.

So, there you have it. A cute little equation, if ever I saw one. But why is this so important? Well, we need the Larmor frequency to calculate the operating frequency of the MRI system.
If we have a MRI system of 1.5 Tesla then the Larmor or precessional frequency is: 42.57 x 1.5 = 63.855 MHz. The precessional frequencies of 1.0T, 0.5T, 0.35T and 0.2T systems would work out to be 42.57 MHz, 21.285 MHz, 14.8995 MHz and 8.514 MHz respectively. You can view these values on your own system by checking the Centre Frequency or similar phrase.

Now we know what happens to the individual protons when we put a victim in the scanner. Lets continue the story and see what happens further.

When the protons experience a strong magnetic field from the scanner we saw that they can align with the field in two ways: parallel and anti-parallel. You could call this also Low and High Energy State.
The distribution of the protons for both states is not the same. The protons are, just like many people, lazy. They prefer to be in low energy state. There are more protons aligned parallel or low energy state than there are anti-parallel or high energy state (*Figure 15*). However, it's not that big a difference. The excess amount of protons aligned parallel within a 0.5T field is only 3 per million (3 ppm = parts per million), in a 1.0T system there are 6 per million and in a 1.5T system there are 9 per million. So, the number of excess protons is proportional with B_0. That is also the reason why 1.5T systems make better images than systems with lower field strengths.
9 ppm excess protons don't seem very many, but in real life it adds up to quite a number. Have a look at the following calculation made by **Moriel NessAiver**, Ph.D. (He wrote an excellent book about MRI physics, which I highly recommend. See § recommended reading). He calculated how many excess protons there are in a single voxel (volume element) at 1.5T.

- Assume a voxel is 2 x 2 x 5 mm = 0.02 ml
- Avogadro's Number says that there are 6.02×10^{23} molecules per mole.
- 1 mole of water weighs 18 grams ($O^{16} + 2H^{1}$), has 2 moles of Hydrogen and fills 18 ml, so……
- 1 voxel of water has $2 \times 6.02 \times 10^{23} \times 0.02 / 18 = 1.338 \times 10^{21}$ total protons
- The total number of excess protons =

$$\frac{1.338 \times 10^{21} \times 9}{2 \times 10^{6}} = \mathbf{6.02 \times 10^{15}}$$ or **6 million billion**!!!

Don't do this at home! (In other words: don't remember this)

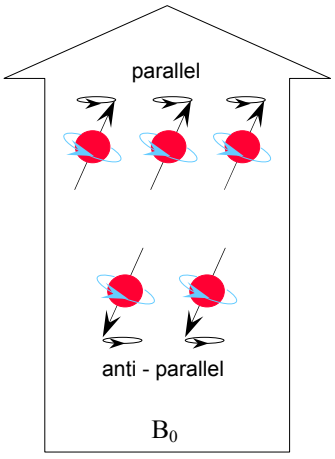


Figure 15

In the end we see that there is a **net magnetization** (the sum of all tiny magnetic fields of each proton) pointing in the same direction as the system's magnetic field.
It is with this net magnetization that we continue.

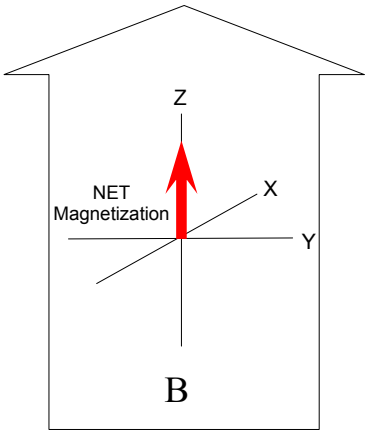


Figure 16

In order to see what happens with this net magnetization in our MRI experiment in an easy way, the scientific community came up with the brilliant idea to visualize it by means of vectors.
A vector (the red arrow in the *Figure 16*) has a direction and a force. To see what happens with the vector (net magnetization) we imagine a frame of rotation, which is nothing else than a set of axes called X, Y and Z.
The Z-axis is always pointing in the direction of the main magnetic field, while X and Y are pointing at right angles from Z. Here we see the (red) net magnetization vector pointing in the same direction as the Z-axis. The net magnetization is now called M_z or longitudinal magnetization.
It is now possible to make simplified drawings of the net magnetization in motion.

Now you are ready to dig a little deeper into the matter and we continue with our MRI experiment and see what happens when we start to play around with the net magnetization.

To obtain an image from a patient it is not enough to put him/her into the magnet. We have to do a little bit more than that. What we also have to do is discussed in the following pages.
The following steps can be divided into Excitation, Relaxation, Acquisition, Computing and Display.

Excitation

Before the system starts to acquire the data it will perform a quick measurement (also called pre-scan) to determine (amongst others) at which frequency the protons are spinning (the Larmor frequency). This centre frequency is important because this is the frequency the system uses for the next step.
Once the centre frequency is determined the system will start the acquisition.
This time we keep things real simple. No fancy pulse sequences. We come to that later.
For now we only send a radio frequency pulse into the patient and we look at what happens.

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 an 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. Resonance you know from the opera singer who sings a high note and the crystal glass shatters to pieces. MRI works with the same principle. Only protons that spin with the same frequency as the RF pulse will respond to that RF pulse. If we would send an RF pulse with a different frequency, let's say 59.347 MHz, nothing would happen.

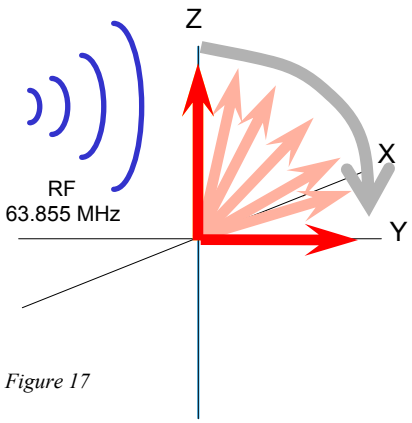


Figure 17

By sending an RF pulse at the centre frequency, with a 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 17*). (See how handy these vectors are. Without the vectors it would be quite impossible to draw this event).
We just "flipped" the net magnetization 90º. 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 1º to 180º. For now we only use an FA of 90º.
This process is called excitation. That's it, applause!!

Relaxation

Now it becomes interesting. We rotated the net magnetization 90º into the X-Y plane. We could also say that we lifted the protons into a higher energy state, same thing. This happened because the protons absorbed energy from the RF pulse. This is a situation that the protons do not like. You could compare this with walking on your hands, it is possible but you don't like it for a long time. You prefer to walk on your feet. Same thing for the protons, they prefer to align with the main magnetic field or, in other words, they would rather be in a low energy state.
Now something happens that is referred to as Relaxation. The relaxation process can be divided into two parts: T1 and T2 relaxation.

T1 Relaxation

The protons want to go back to their original situation, called equilibrium. They do so by releasing the absorbed energy in the shape of (very little) warmth and RF waves.
In principle the reverse of excitation takes place. The net magnetization rotates back to align itself with the Z-axis.

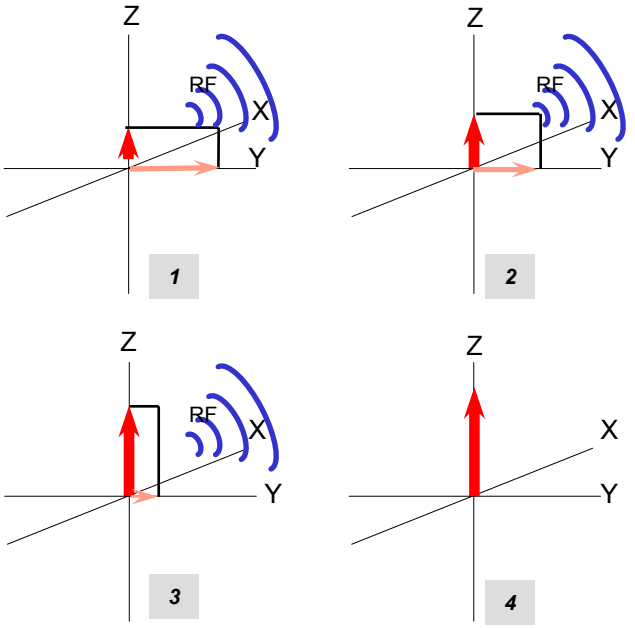


Figure 18

After the RF excitation pulse stops, the net magnetization will re-grow along the Z-axis, while emitting a radio-frequency waves (*Figure 18*).
T1 relaxation describes what happens in the Z direction.
So, after a little while, the situation is exactly as before we sent an RF pulse into the patient.
T1 relaxation is also known as Spin-Lattice relaxation, because the energy is released to the surrounding tissue (lattice).
So far, so good! This process is relatively easy to understand because one can, somehow, picture this in ones mind.

T1 Relaxation Curves

Longitudinal Relaxation T1

T1 relaxation happens to the protons in the volume that experienced the 90º-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 19*.

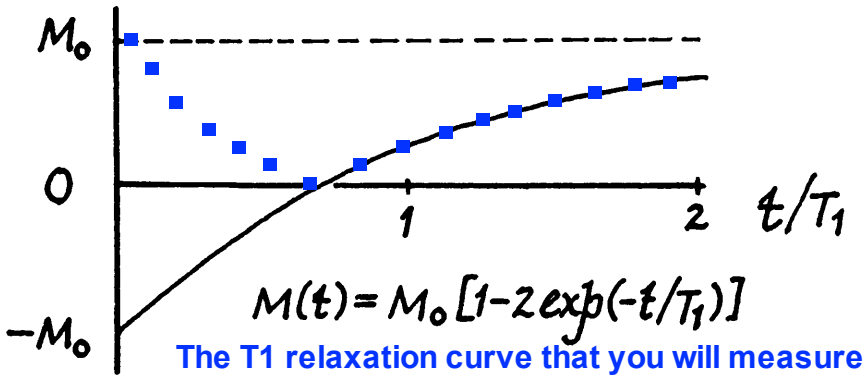
He does not show the magnetization along -z, so his figure actually shows T2 relaxation--it shows the induced non-equilbrium magnetization along y returning to equilibrium magnetization along +z. The shape of the T1 relaxation due to magnetization along +z and -z is shown on the next page.

Because the detector does not see negative signals, you will measure the dotted curve

and

$$M_o - M(t) = 2M_o \exp(-t/T_1) \ .$$

T1 relaxation



The T1 relaxation curve that you will measure is given by the absolute value of this equation

The practical difference is that any mis-estimation in the baseline for the signal cancels out for the second equation so that such errors do not lead to apparent non-exponentiality whereas they do for the first. The penalty you pay for that is that the value of the function being plotted at t=0 is unknown, whereas for the first equation it is fixed at 2.00000 (assuming that the experimental parameters are set perfectly) which is very helpful to know especially if S/N is poor because it is a good data point, possibly the only one.

A related question is that of how to space out the sampling points in time for the optimum accuracy in a given time. Since the quantity plotted (for either equation) falls off with time while the noise stays the same size, clearly the S/N decreases as the delay increases. In fact, you can reach a point where the data contributes more noise than signal. There is also a strong desire to not measure these long delay values since they take more time. On the other hand, these are the points that tell you whether the relaxation is exponential or not and this is a very important fact to establish.

This figure shows the return to equilibrium of the magnetization along +z, but it does not show the signal that you will measure--you will measure the difference in the magnetization along +z and -z.

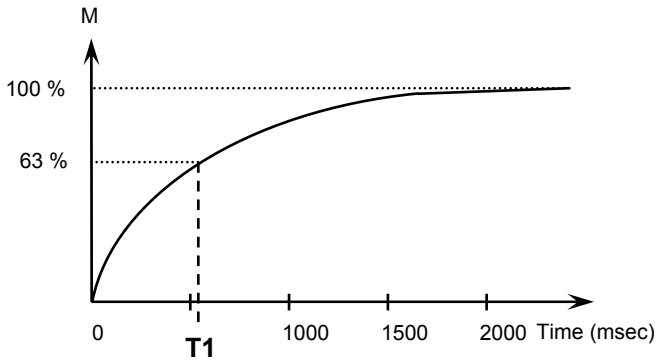


Figure 19

The T1 response you will measure is shown on the previous page.

The curve shows at time = 0 that there is no magnetization in the Z-direction right after the RF-pulse. But immediately the M_z starts to recover along the Z-axis. T1 relaxation is a time constant. T1 is defined as the time it takes for the longitudinal magnetization (M_z) to reach 63 % of the original magnetization.
A similar curve can be drawn for each tissue. That's what **Damadian** and **Lauterbur** discovered many moons ago. Each tissue will release energy (relax) at a different rate and that's why MRI has such good contrast resolution.

T2 Relaxation

As said before, the relaxation process is divided into two parts. The second part, T2 relaxation, is slightly more complicated. I have found that people have difficulties in understanding this, but we'll have a jolly good go at it. Do not despair!!
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. I cannot emphasize this enough.

Phase and Phase coherence

Ever heard of Phase? Imagine this: if you see a group of soldiers marching along the road they all put their left leg forward at the same time. The sergeant tells them to do so: Left, Right; Left, Right, Left . . . Left . . . Left, Right. You could say that the group is walking in phase or in synchronization.

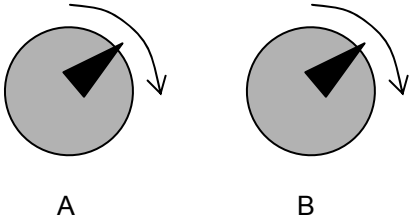


Figure 20

Another example: In *Figure 20* we see two wheels with an arrow. The wheels are rotating at the same speed. The arrows will therefore point in the same direction at any time. The wheels are said to be rotating In-Phase.

Let's go back one step and have a look at the net magnetization vector before we apply the 90º RF pulse. The net magnetization vector is the sum of all the small magnetic fields of the protons, which are aligned along the Z-axis.
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. The arrows of the two wheels from the previous example would point in different directions.
When we apply the 90º RF pulse something interesting happens. Apart from flipping the magnetization into the X-Y plane, the protons will also start spinning *in-phase*!!

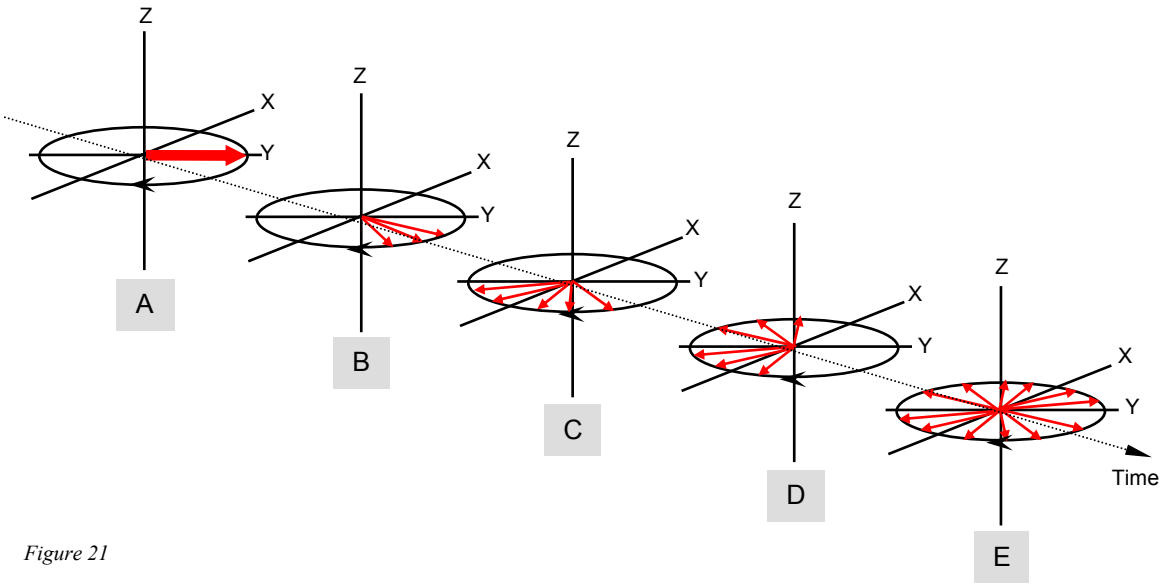


Figure 21

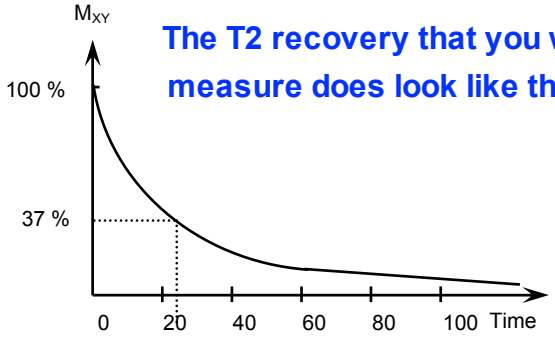
So, right after the 90º RF pulse the net magnetization vector (now called transverse magnetization) is rotating in the X-Y plane around the Z-axis (*Figure 21A*). The vectors all point in the same direction because they are In-Phase. However, they don't stay like this.
Image this: I'm sure as a child you played the game where you stood real close behind each other and then you tried to walk. This only works when both of you put your left foot forward at the same time. Then you are walking In-Phase. At some time one of you would stumble and your feet would get tangled up resulting in a mini chaos with both of you walking in a different direction: you got Out-Of-Phase or you were De-phasing. A similar situation happens with the vectors in MRI. Remember that each proton can be thought of as a tiny bar magnet with a north and a south pole. And two poles of the same sign repel each other. Because the magnetic fields of each vector are influenced by one another the situation will occur that one vector is slowed down while the other vector might speed up. The vectors will rotate at different speeds and therefore they are not able to point into the same direction anymore: they will start to de-phase. At first the amount of de-phasing will be small (*Figure 21B, 21C, 212D*), but quickly that will increase until there is no more phase coherence left: there is not one vector pointing in the same direction anymore. (*Figure 21E*) In the meanwhile the whole lot is still rotating around the Z-axis in the X-Y plane.

This process of getting from a total *in-phase* situation to a total *out-of-phase* situation is called T2 relaxation.

T2 Relaxation Curves

Transverse Relaxation T2

Just like T1 relaxation, T2 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.



The T2 recovery that you will measure does look like this.

Figure 22

Also here we can draw a curve. (*Figure 22*)
Right after the 90º 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 relaxation is also a time constant. T2 is defined as the time it

takes for the spins to de-phase to 37% of the original value.
The rate of de-phasing is different for each tissue. Fat tissue will de-phase quickly, while water will de-phase much slower.
One more remark about T2: it happens much faster than T1 relaxation. T2 relaxation happens in tens of milli-seconds, while T1 can take up to seconds. (Have a look at the relaxation times table in the § Appendix).

T2 relaxation is also called spin–spin relaxation because it describes interactions between protons in their immediate surroundings (molecules).

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 (Actually, T2* is much faster than T1)

When both relaxation processes are finished the net magnetization vector is aligned with the main magnetic field (B_0) again and the protons are spinning Out-Of-Phase; the situation before we transmitted the 90º RF-pulse.

You will measure the same value for T1 and T2.

Acquisition

See the plot and text on the next page.

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. An interesting, but ever so important, fact is the position of the receive coil.

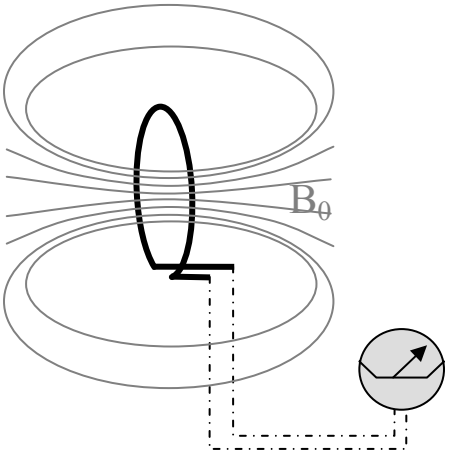


Figure 23

The receive coil must be positioned at right angles to the main magnetic field (B_0). 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. When a magnetic field goes through the loop, a current is induced (*Figure 23*). B_0 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_0 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 24*.

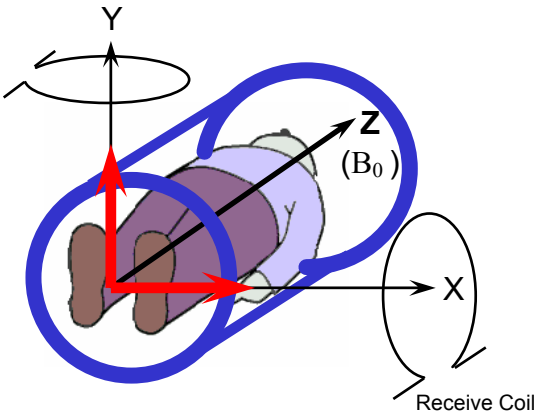


Figure 24

It is quite interesting to try this for yourself with your scanner. Just make a series of scans where you position the receive coil at different angles. Start with the coil at a right angle with B_0, and then turn it a bit such that B_0 is allowed to run through

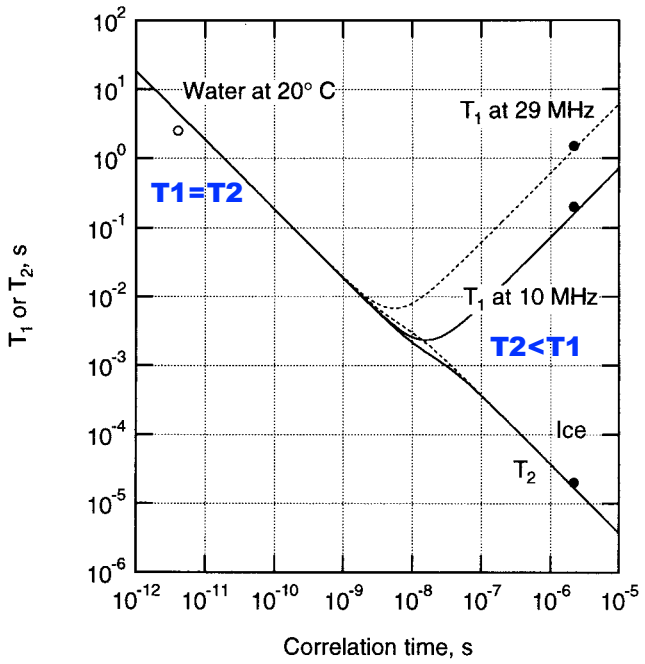


FIGURE 17.12. Plot of T_1 and T_2 vs correlation time of the fluctuating magnetic field at the nucleus. Experimental points are shown for water (open dot) and ice (solid dot).

$$\frac{1}{T_1}=\frac{C\tau_c}{1+\omega_0^2\tau_c^2}, \tag{17.33}$$

where C is the proportionality constant.

The correlation time in a solid is much longer than in a liquid. For example, in liquid water at 20 °C it is about 3.5×10^{-12} s; in ice it is about 2×10^{-6} s. Figure 17.12 shows the behavior of T_1 as a function of correlation time, plotted from Eq. (17.33) with $C=5.43\times10^{10}$ s^{-2}. For short correlation times T_1 does not depend on the Larmor frequency. At long correlation times T_1 is proportional to the Larmor frequency, as can be seen from Eq. (17.33). The minimum in T_1 occurs when $\omega_0=1/\tau_c$ in this model.

Table 17.2 shows some typical values of the relaxation times at 20 MHz. Neighboring paramagnetic atoms reduce the relaxation time by causing a fluctuating magnetic field. For example, adding 20 ppm of Fe^{3+} to water reduces T_1 to 20 ms.

Differences in relaxation time are easily detected in an image. Different tissues have different relaxation times. A contrast agent containing gadolinium is often used in magnetic resonance imaging. It is combined with many of the same pharmaceuticals used with ^{99m}Tc, and it reduces the relaxation time of nearby nuclei. The hemoglobin that carries oxygen in the blood exists in two forms: oxyhemoglobin and deoxyhemoglobin. The former is diamagnetic and the latter is paramagnetic, so the relaxation time in blood depends on the amount of oxygen in the hemoglobin. The technique that exploits this is called BOLD (blood oxygen level dependence).

TABLE 17.2. *Approximate relaxation times at 20 MHz.*

	T_1 (ms)	T_2 (ms)
Whole blood	900	200
Muscle	500	35
Fat	200	60
Water	3000	3000

It was pointed out in Sec. 17.4 that because of dephasing, T_2 is less than or equal to T_1. The same model for the fluctuating fields which led to Eq. (17.33) gives an expression for T_2:

$$\frac{1}{T_2}=\frac{C\tau_c}{2}+\frac{1}{2T_1}. \tag{17.34}$$

There is a slight frequency dependency to T_2 for values of the correlation time close to the reciprocal of the Larmor frequency.

Another effect that causes the magnetization to rapidly decrease is dephasing. Dephasing across the sample occurs because of inhomogeneities in the externally applied field. Suppose that the spread in Larmor frequency and the transverse relaxation time are related by $T_2\Delta\omega=K$. (Usually K is taken to be 2.) The spread in Larmor frequencies $\Delta\omega$ is due to a spread in magnetic field ΔB experienced by the nuclear spins in different atoms. The total variation in B is due to fluctuations caused by the magnetic field of neighbors and to variation in the applied magnetic field across the sample:

$$\Delta B_{\text{tot}}=\Delta B_{\text{internal}}+\Delta B_{\text{external}}.$$

Therefore

$$\Delta\omega_{\text{tot}}=\Delta\omega_{\text{internal}}+\Delta\omega_{\text{external}}.$$

The total spread is associated with the experimental relaxation time, $T_2^*=K/\Delta\omega_{\text{tot}}$. The "true" or "non-recoverable" relaxation time $T_2=K/\Delta\omega_{\text{internal}}$ is due to the fluctuations in the magnetic field intrinsic to the sample. Therefore

$$\frac{1}{T_2^*}=\frac{1}{T_2}+\frac{\gamma\Delta B_{\text{external}}}{K}. \tag{17.35}$$

T_2 is called the *nonrecoverable* relaxation time because various experimental techniques can be used to compensate for the external inhomogeneities, but not the atomic ones.

17.7. DETECTING THE SIGNAL

We have now seen that a sample of nuclear spins in a strong magnetic field has an induced magnetic moment, that it is possible to apply a sinusoidally varying magnetic field and nutate the magnetic moment to precess at any arbitrary

the Phase Encoding gradient (G_{PE}) was switched on to do the first phase encoding. Then (**4**) the Frequency Encoding or Read Out gradient (G_{RO}) was switched on during which (**5**) the signal, the Free Induction Decay (FID), was sampled.

This is a very simple and basic sequence. We also saw that the signal dies out very quickly. In the early days that was a problem. The hardware could not be switched quick enough to sample the entire signal. They could only sample the last part of the signal when most of the signal was gone. The resulting image showed it! It was signal starved. In order to improve the amount of signal the engineers came up with a brilliant solution.

Spin Echo Sequence

After the 90º-excitation pulse the net-magnetization is in the X-Y plane. It immediately starts to dephase due to T2 relaxation (spin-spin interactions). It is because of this dephasing that the signal drops like a stone. Ideally, we would like to keep the phase coherence because this gives the best signal. The brilliant solution the engineers came up with is this: a short time after the 90º RF-pulse a second RF-pulse is given. This time it is an 180º pulse. The 180º pulse causes the spins to *re*phase. When all the spins are rephased the signal is high again, and when we make sure we sample the signal at this instant we'll have a much better image. *Figure 50* shows it better.

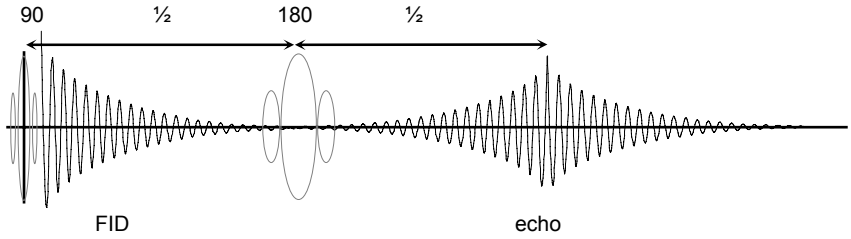


Figure 50

The signal we sample is called: an Echo, because it is "rebuilt" from the FID.
Notice that the 180º *rephasing* pulse is exactly in the middle of the 90º pulse and the echo.

There is a book (highly recommended, see reference) called: "MRI made easy… Well almost" by Schering, Germany, in which you can find a really nice analogy of rephasing:

Imagine a number of runners on a racetrack. When the whistle blows they all start to run (dephasing). Obviously they all run at the different speeds and after, let's say 30 seconds, the fastest one will be way ahead of the slower one. Then the whistle blows for the second time. The runners were instructed to turn around, *without losing speed*, at the second blow of the whistle (180º RF pulse). The fastest runner, now way behind the slower one, will catch up with the slower one (rephasing). After another 30 seconds they all arrive at the starting point *at the same time* (echo).

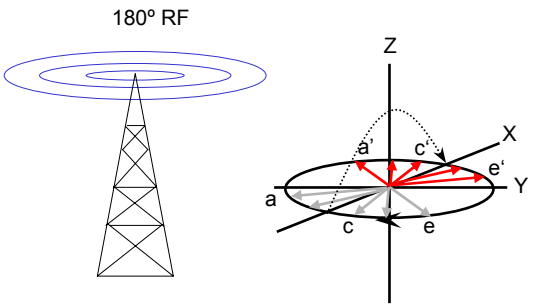


Figure 51

The effect of the 180º RF pulse is called: rephasing.
Figure 51 shows how this works. The spin system is mirrored around the Y-axis. Note that the rotation direction in the X-Y plane does not change.

So, what do we have? Let's have a look at *Figure 52*.

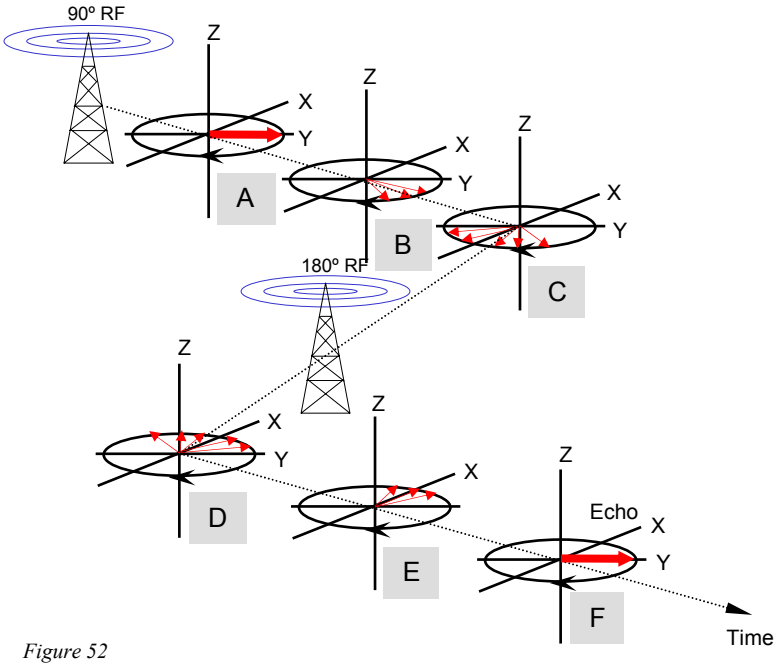


Figure 52

A. It starts with a 90º excitation pulse. The magnetization is flipped into the X-Y-plane.
B. Immediately the spins dephase…
C. The spins dephase a bit more… then a 180 rephasing pulse is given.
D. The spins are mirrored around the Y axis.
E. The spins rephase until…
F. The spins are in phase again creating an "echo".

This is what is known as a Spin-Echo sequence.
As with everything in MRI, the spin-echo sequence is a compromise:

Advantages:

- the signal is strong
- compensation for local field inhomogeneities: less artifacts.

Disadvantages:

- It takes time to do the rephasing step. This will increase the total scan time.
- It increases the amount of RF one has to put into the body (not that it's dangerous, but there are certain limits).

In spite of the increased scan time and the amount of RF the spin-echo sequence is widely used and has become the routine sequence in MRI.

Appendix

Tissue Relaxation Times

Following **table 4** shows T1 and T2 relaxation times for various tissues (normal and pathological) at different magnetic field strengths.

Table 4: Tissue relaxation times

	T1 (ms) 0.25 T	T1 (ms) 0.5 T	T1 (ms) 1 T	T1 (ms) 1.5 T	SD (%)	T2 (ms)	SD (%)
BRAIN							
Gray matter	530	657	813	921	17	101	13
White matter	422	537	683	787	17	92	22
Tumours	667	802	963	1073	36	121	63
Meningioma	586	714	871	979	18	103	31
Glioma	845	887	931	959	35	111	33
Edema	667	806	973	1090	23	113	73
BONE							
Normal marrow	607	648	695	732	78	106	60
Osteosarcoma	740	811	888	973	28	85	30
BREAST							
Fibrotic tissue	409	547	732	868	18	49	16
Adipose tissue	190	214	241	259	28	84	36
Tumours	483	634	832	976	28	80	35
Carcinoma	451	595	785	923	25	94	48
Adenocarcinoma	490	686	959	1167	10	81	12
Fibroadenoma	508	715	989	1195	29	60	11
KIDNEY							
Normal tissue	417	496	589	652	27	58	24
Tumours	733	796	864	907	37	83	42
LIVER							
Normal tissue	250	325	423	493	22	43	14
Tumours	713	782	857	905	26	84	31
Hepatoma	621	769	951	1077	16	84	26
Chirrosis	328	367	410	438	21	45	
LUNG							
Normal tissue	488	599	735	829	19	79	29
Tumours	407	535	703	826	51	68	45
MUSCLE							
Normal tissue	409	547	732	868	18	47	13
Tumours	597	752	946	1083	32	87	40
Carcinoma	608	750	925	1046	16	82	73
Fibrosarcoma	831	896	967	1011	15	65	14
Rhabdomyosarcoma	664	827	1031	1173	27		
Edema	652	897	1235	1488	26	67	26
PANCREAS							
Normal tissue	302	371	455	513	25		
Tumours	718	942	1235	1448	15		
SPLEEN							
Normal tissue	431	543	683	782	19	62	27
Tumours						69	1

Final Words

You must have a sore head by now and I don't blame you ☺.

There is one more thing, though, I would like to stress: **A synonym for MRI is COMPROMISE**.
It can't be said enough. In fact I'm going to say it again: A synonym for MRI is COMPROMISE.
I think this is the most important thing to remember from this story.
MRI is a balancing act for Signal to Noise. Whenever you change a parameter, you'll have to change another to counteract for the signal change, if you want to keep the SNR the same.
Many systems offer some sort of "SNR Indicator". When you change a parameter it shows how much signal you will gain or lose. It's a very nice tool, because sometimes it is not at all clear how much impact changing one parameter can have. If your system does not have a tool like that, then be very careful when changing parameters.

I hope that reading this story has helped you to understand the complex matter of MRI physics.
It only scratched the surface and you're by no means ready yet to build your own MR scanner, but at least you know what the abbreviation MRI means ☺.
I know it is not easy to get the total picture after reading the story once. Don't blame yourself. Read the story again, and again, if need be. Then open up a book from the reference list and try it again. One day there will be a moment that the penny drops and you will be rewarded with an immense feeling of happiness and you realize that MRI is even better than sliced bread.

These pages are the result of many years of teaching MR physics to radiographers and technicians who just started with MRI. They are by no means complete, but I hope it will encourage you to explore the exciting world of MRI a bit more. Take it in good time, though. Do not move on until you understand the basics. It is very easy to get lost in the myriad of subjects.

This is the right time to express my thanks to all those people who encouraged and supported me to put it all on paper; Aida Ferreira, one of my first students, who asked me to do it; my colleague Johan Roubroeks for test reading and, above all, Deirdre, my wife, for editing the story and who put up with the long hours.
Thanks all.

Have fun, ☺

Evert Blink

FREE NUCLEAR INDUCTION

By E. L. Hahn

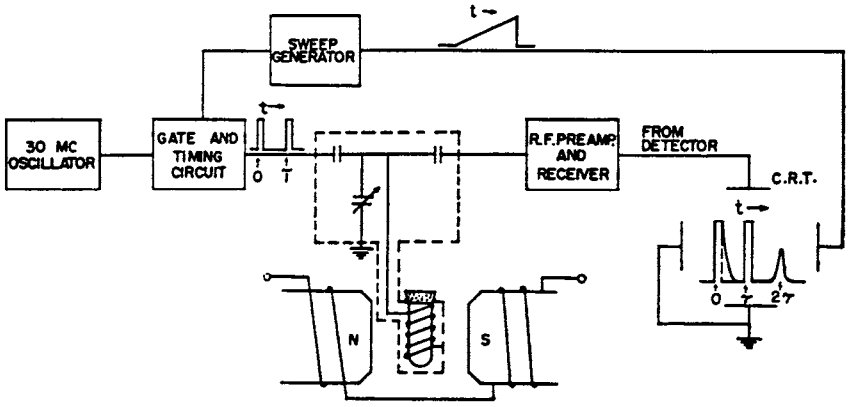


Fig. 1. Arrangement for obtaining spin echoes.

Unusually clear measurements of the time precession of nuclear magnetic moments and other properties of nuclei have become possible with the development of the free induction "spin echo" technique in the still new science of nuclear magnetic resonance.

THE STUDY of nuclear magnetic resonance or nuclear induction, a recent field of research for which F. Bloch [1] and E. M. Purcell [2] have been awarded the Nobel Prize, has been carried out by a variety of techniques. The usual approach has been to observe the nuclear resonance of an ensemble of nuclear moments in a large static magnetic field as a function of a slow change in this field. Meanwhile, a small radio-frequency field is applied continuously to the nuclear sample in a direction perpendicular to the large field. An alternative method to this steady state or "slow passage" technique is one by which the radio-frequency energy is applied to the sample in the form of short, intense pulses, and nuclear signals are observed after the pulses are removed. The effects which result can be compared to the free vibration or "ringing" of a bell, a term often applied to the free harmonic oscillations of a shocked inductive-capacitive (LC) circuit. The circuit is first supplied with electrical energy from some source, and the supply of energy is suddenly removed. The LC circuit then remains for a time in the "excited state", and the energy is gradually dissipated into heat, mostly in the circuit resistance. Similarly the atom or nucleus in the excited state can store energy for a time before it is completely dissipated, and in the case discussed here, the free oscillation or precession of an ensemble of nuclear spins in a large static field provides the ringing process.

It appears that the topic of free nuclear precession or "spin echoes", as it will be called here, can be classified under one of the following two approaches to the study of excited states: (a) A study can be made of the absorption or emission of radiation by a system. The system gains or loses radiation (or particles), and the experiment involves a measurement of the energy and intensity of radiation. (b) The behavior of some systems can be studied, not by observing directly whether they gain or lose radiation, but by detecting their mode of motion under the influence of applied *static* electric and magnetic fields. While observing such motion it is possible to infer whether or not the system has gained or lost radiation in the past.

It is well known that the latter approach is involved in the Rabi molecular beam technique [3] in which molecules and atoms are deflected in space. Also, this approach applies to the free nuclear precession or induction effect. The viewpoint of the experiment has been particularly emphasized by Bloch,[1] and is very much like the scheme for detecting the ringing of a tuned LC circuit. A pickup loop can be coupled to the magnetic flux about the inductance and a voltage of induction is measured. In the actual experiment the magnetic induction is provided by the precession of an ensemble of nuclear moments in a static magnetic field after the moments are "shocked" into a coherent state of precession by one or more pulses of radio-frequency magnetic field. One might classify this measurement under (a) above, and deduce that here the ringing of the system is measured by detecting the stored magnetic energy which the system dissipates. Certainly some radio-frequency (rf) energy is consumed by the loop which couples to the ringing LC circuit, but this can be made

E. L. Hahn, a physicist at the Watson Laboratory and a member of the physics staff at Columbia University, received his PhD at the University of Illinois in 1949. He was a National Research Council Fellow and instructor at Stanford University in 1951–52.

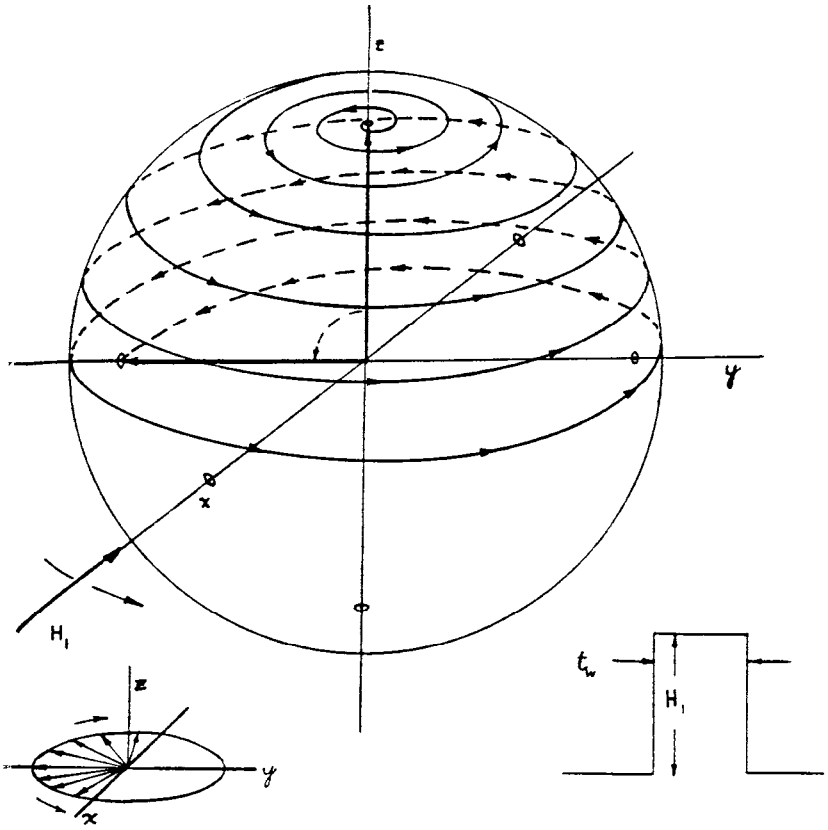


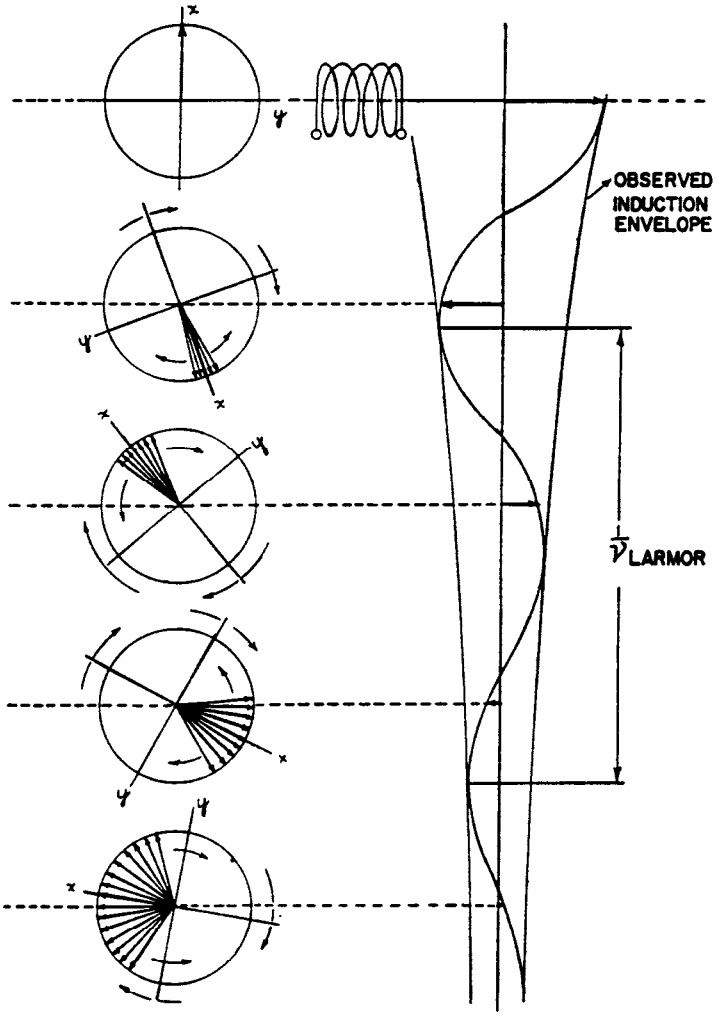
Fig. 2a. As seen in the laboratory frame of reference the classical magnetic moment precesses toward the xy plane in a spiral motion due the torque effect of the rf field H_1 at nuclear resonance. In the frame of reference rotating with H_1 the magnetic moment appears to precess in a plane perpendicular to H_1.

Fig. 2b. Following the orientation of the total magnetic moment into the xy plane by a 90° pulse the spectrum of moment vectors provides a free nuclear induction signal. The precessional motion relative to the moving and fixed frames of reference is indicated.

negligibly small by using a loop of high-circuit impedance. In the free precession experiment, a sample of nuclei is placed in a pickup loop comprising the inductance of an LC circuit tuned to the Larmor precession frequency. A low voltage of induction (about a millivolt at most due to protons) is produced across the inductive coil by the nuclear spins, and the current which flows in the coil produces a weak rf magnetic field throughout the volume of the sample. This field does react upon the nuclear spins, but causes only a negligibly small rate of induced emission. The ringing oscillator, in this case the nuclear moments, is again very weakly loaded by the pickup coil.

If any clear distinction can be made between the terms "nuclear magnetic resonance absorption" and "nuclear induction", it can be made in the free precession experiment. Nuclear resonance absorption belongs under heading (a) above. Radio-frequency energy in the form of pulses, absorbed by the nuclear moments, prepares the moments for the ringing process which is observed only by nuclear induction under heading (b).

A qualitative description of the experiment involves many of the basic features of nuclear resonance techniques. A large constant magnetic field H_0 must be available in which polarization and precession of the nuclear spins take place. In practice this field is never perfectly homogeneous for all of the nuclei throughout the volume of the sample. Instead the field varies throughout space in a manner which is determined by the inhomogeneity of the magnet or also by local magnetic fields in the lattice of the sample. In liquids and gases, however, except for certain special but small molecular effects, any local magnetic field in the lattice



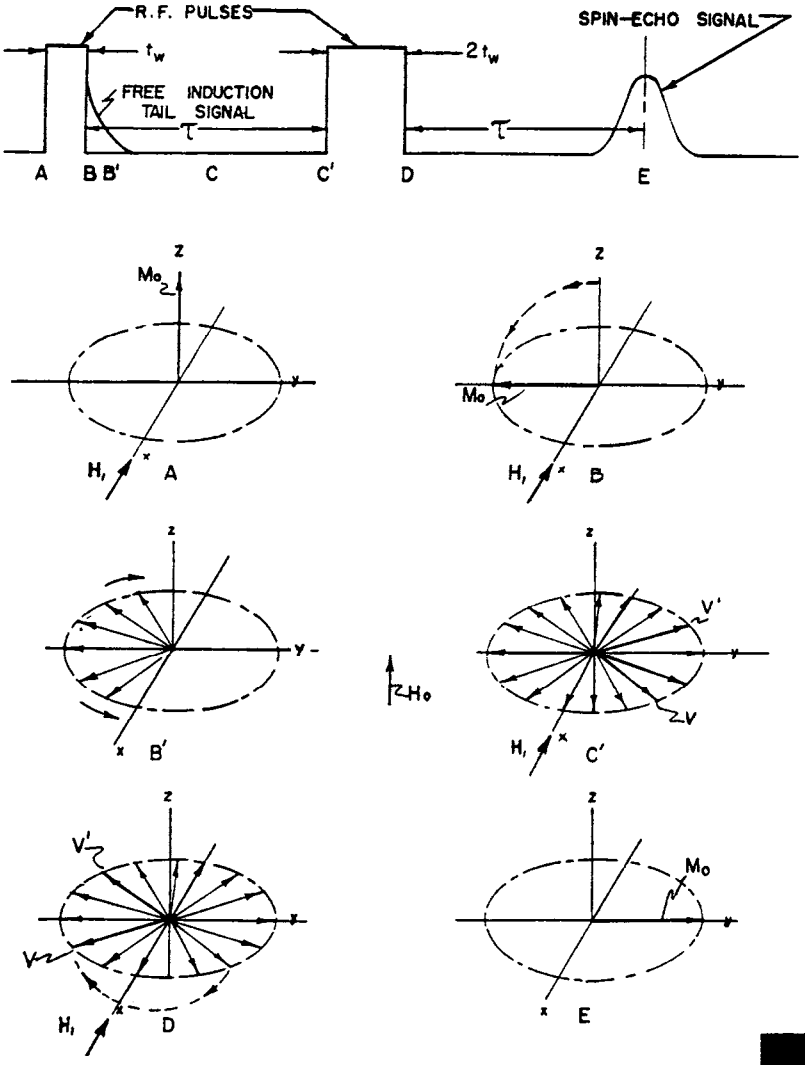
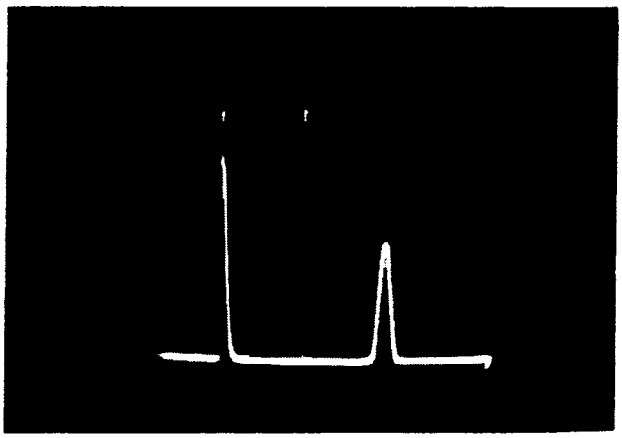


Fig. 3. Vector schematic of the spin echo formation.

Fig. 4. Oscillographic display of the spin echo due to protons in glycerine. The rf pulse widths are narrow compared to the duration of the signals.



which a given nuclear moment sees as a result of its neighbors will average completely to zero over a Larmor period. This happens because the tumbling and translational frequencies common to liquid molecules at normal temperatures, for example, are extremely large compared to the Larmor frequency of precession (greater by a factor of 10^4 to 10^5). During one Larmor period of precession the field caused by a neighboring dipole does not remain at a given value long enough to influence the rate of precession determined by the externally applied field H_0. It becomes possible, therefore, to ascribe to each volume element of the liquid an isochromatic or classical magnetic moment, which is due to the preponderance of nuclear moments pointing in the direction of the external field. This field can be assigned as homogeneous over the volume element. The entire liquid sample provides a distribution of magnetic moments according to how much volume of spin sample is assigned to each value of H_0 as it varies in space. Such a spectrum may be described by some symmetric distribution, where the maximum number of nuclear moments may be subject to a field of 7000 gauss, for example, and fewer moments see values of H_0 smaller or greater than this average value.

At thermal equilibrium the net magnetic moment M_0 which is aligned with the field can be compared to the "sleeping" mechanical top that spins with its axis along the direction of the gravitational field. If the top is perturbed by applying torque perpendicular to its spin axis, it will precess in a certain direction at a given frequency for any angle θ which exists between the spin axis and the direction of the gravitational field. The nuclear magnetic top has a torque exerted upon it by the magnetic field H_0 in place of gravity, and when it is tipped away from the H_0 or z axis direction by rotating rf magnetic fields in the xy plane, it consequently precesses about the z axis at the Larmor frequency given by $\omega_0 = \dot{\gamma} H_0$ after the xy perturbing field is removed. The constant γ is the gyromagnetic ratio defined by $\gamma = \mu/I\hbar$, where μ is the magnetic moment and $I\hbar$ is the nuclear spin angular momentum. The time that the induction signal due to a classically precessing $\bar{M}_0$ vector can persist is also the time for which constituent nuclear spins precess in phase before damp-

ing effects due to the lattice become appreciable. This coherence time is given by T_2, often referred to as the "transverse relaxation time". Another relaxation time of importance, which determines in part the value of T_2, is the longitudinal or thermal relaxation time T_1, the time in which a precessing spin remains in the magnetically excited state regardless of its phase. In liquids both of these relaxation times may vary from fractions of milliseconds to several seconds.

After a few preliminary definitions, we shall impose special experimental conditions for purposes of clearly explaining the echo nuclear induction effect. An inductive driving coil which surrounds the nuclear sample is tuned to the Larmor frequency $\omega = \omega_0'$ where ω_0' is the average angular Larmor frequency of a sample of nuclear moments. rf pulses applied to the tuned LC circuit each last for t_w seconds during which time approximately $\omega_0' t_w$ Larmor oscillations take place. The effective rf magnetic field is referred to as the $\bar{H}_1$ Gauss vector which precesses in the direction that $\bar{M}_0$ precesses, and is the vector which remains essentially perpendicular to the plane defined by $\bar{M}_0$ and $\bar{H}_0$ (see Fig. 2). In a coordinate system which rotates with frequency $\omega_0'/2\pi$ about the z axis, the $\bar{M}_0$ vector will then appear to precess about $\bar{H}_1$ through angle $\theta = \gamma H_1 t_w$ from the time $\bar{H}_1$ is turned on. This $\bar{H}_1$ field is one of two circularly polarized field components which sum to provide the alternating magnetic field $2H_1$ along the axis of the inductive coil. The other $\bar{H}_1$ field component, rotating in the opposite direction, can be ignored because its torque acting upon $\bar{M}_0$ is alternately positive and negative, and averages to zero for all practical purposes. We shall assume that $\bar{H}_1$ is turned on infinitely fast, and that it shall be removed after t_w seconds infinitely fast. During the time t_w, all isochromatic moments, initially aligned along $\bar{H}_0$, will be turned away from the z axis toward the xy plane. The driving radiofrequency pulses may be transmitted by the same coil used for receiving signals, or two separate coils perpendicular to each other may be used, one for transmitting and one for receiving. If $\bar{H}_1$ rotates in the xy plane at a frequency ω, all those isochromatic moments which are tuned to Larmor frequency ω_0, different from ω, will, strictly speaking, not precess exactly in a plane perpendicular to the direction of $\bar{H}_1$. However, $\bar{H}_1$ is chosen to be sufficiently intense so that most of the isochromatic moments are rotated toward the xy plane in a time short compared to the time in which they would get out of phase with $\bar{H}_1$ (i.e. they would scarcely deviate from the plane defined by $\bar{M}_0$ and $\bar{H}_0$, perpendicular to $\bar{H}_1$). Therefore all the isochromatic moments are substantially in phase at the time t_w when they have reached the xy plane. At this time t_w the field $\bar{H}_1$ is suddenly removed. A nuclear induction signal in the receiver or pickup coil will persist after the field $\bar{H}_1$ is removed, but will finally die out because each isochromatic moment is now free to precess at its natural frequency. Since these frequencies differ for each isochromatic moment, the moments will, after a time, get out of phase and their inductive effects will interfere or cancel among themselves.[4] The time required for this loss of a net observed induced output signal is usually determined by the inhomogeneity of the magnet. After such a time, the isochromatic moment vectors are uniformly distributed about the z axis, but the original magnitudes of these isochromatic moments are preserved if it is assumed that there are no relaxation effects. Although

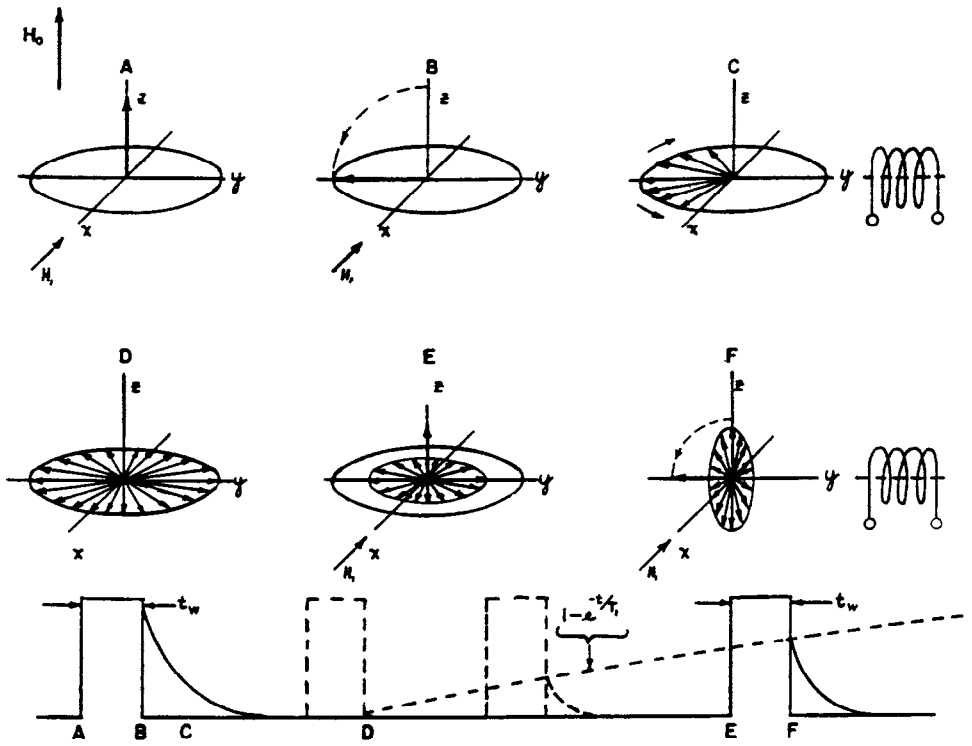


Fig. 5. Scheme for measurement of T_1. The sequence from *A* through *D* is the same as in Fig. 3. At the time a pulse is applied at time *E* a fraction of the total nuclear magnetism has returned to thermal equilibrium along the *z* axis. Following this pulse at *F* a free induction signal results which is proportional to this magnetism.

these isochromatic moment vectors do not provide a resultant moment, they do have definite phase relations among themselves, and each vector occupies a position which has been determined by its past history. If at a time τ after cessation of the first pulse, a second pulse is transmitted to the driving coil, this past history is manifested in what is called the "spin echo".

The echo effect [5] can be explained from a very simple analogy. Let a team of runners with different but constant running speeds start off at a time $t = 0$ as they would do at a track meet (see the cover of this issue). At some time T these runners will be distributed around the race track in apparently random positions. The referee fires his gun at a time $t = \tau > T$, and by previous arrangement the racers quickly turn about-face and run in the opposite direction with their original speeds. Obviously, at a time $t = 2\tau$, the runners will return together precisely at the starting line. This will happen once and only once, just as it will be shown in the case of two rf pulses and the echo. From this analogy, one can see that if even more than two pulses are applied to the ensemble, a pattern of echoes or constructive interference events will occur which will uniquely be related to the pulses which were applied in the past. For example, if the referee again fired his gun for a third time after the racers came together at the starting point and fanned out again around the track, and the runners again repeated the about-face procedure, they would again come back to the starting line.

For the purpose of simplifying the description of the actual echo formation, as shown in Fig. 3, the second pulse is made either of the same intensity and twice as long as the first pulse or twice as intense with the same duration as the first pulse. From the time $\bar{H}_1$ appears again in the xy plane, each isochromatic moment vector precesses in a cone whose axis is the direction of $\bar{H}_1$. At the instant the second pulse is removed, all vectors will have been rotated from whatever xy plane quadrants they happened to have been in (at the onset of the second pulse) on one side of $\bar{H}_1$ to a mirror image position on the opposite side of $\bar{H}_1$. The second pulse, so to speak, has flipped the "pancake" of isotropically distributed moments by 180°. When this flip has occurred it can be seen that if we refer to the central average isochromatic moment, each isochromatic moment which lay ahead of this average moment by a given angle before the second pulse now lies behind it by the same angle. Furthermore, each isochromatic moment which lay behind the average isochromatic moment by a given angle will lie ahead by the same angle. Now if these isochromatic moments continue to precess as before, those behind the reference vector or average isochromatic moment will be catching up and those ahead will be falling back. Hence, at time τ beyond the second pulse, all the moment vectors will be back in phase and the echo of the first pulse will occur at time 2τ where τ is the time between the first pulse and the second or reversing pulse. This can be seen by tracing the history of a pair of vectors from the figure. At the onset of the first pulse, the moments M_0 lie on the z axis at thermal equilibrium at A. Following a rotation of 90°, completed at B by the first pulse, the isochromatic moments spread out as shown in B'. During this time, an induction signal forms as a "tail" following the first pulse. At C and C' the tail is absent because the isochromatic moments are evenly distributed in the xy plane. Follow, for example, the precessional motion of isochromatic moment vectors V and V' shown at C'. Here they happen to be oriented in positions indicated at the onset of the second pulse, which now rotates them and the whole array in the xy plane by 180°, as shown at D. After the pulse is removed at D, the vectors V and V' will proceed to precess through angles in the xy plane again in a time τ as they did after the first pulse. They obviously must coincide at the time of the echo at E. This argument holds for every pair of vectors in the ensemble for the special case given here. The spin echo has a shape which grows and dies out symmetrically in the time it takes for the isochromatic moments to get in phase and then out of phase.

It should be noted at this point that although it has been assumed for the sake of simplicity that the first pulse rotates all the vectors by 90°, and the second pulse rotates them by 180°, these rotations happen to be the ones which give the maximum available echo. Useful results may also be obtained by use of other arbitrary combinations of t_w and H_1 giving different angles of rotation. For example, the second pulse may be of equal length and equal intensity as the first pulse, in which case the array is rotated 90° rather than 180° as described above.

Useful information about the local magnetic fields due to chemical environment about the precessing nucleus can be obtained in certain cases from measurements of shapes and amplitudes of free induction echo signals. Steady state resonance techniques of course can provide the same information, which in some cases is more direct, particularly when a number of closely spaced transitions between stationary states of nuclear Zeeman levels are indicated directly by resonance lines.[6] The equivalent of such small differences in energies of closely spaced Zeeman levels is manifested in the echo method by interference beats [7] between the various components of precessing magnetic moments which precess at different Larmor frequencies in the same sample of spins subjected to a given external H_0. If the maximum echo signal amplitude is measured for increasing values of τ, where for each setting of τ the ensemble is initially at thermal equilibrium, a plot of the echo signal amplitude as a function of τ displays predominantly a monotonic decay. In many cases the decay is exponential and serves as a direct measure of the nuclear spin relaxation parameter T_2, but the decay may also be determined in part by other factors. There are other special ways of studying echo signal plots which do not require that the ensemble be at equilibrium for just a pair of pulses, where the nature of the information obtained is essentially the same.

The decay of the echo may be understood in terms of the race track analogy if it is assumed now that the runners become fatigued after the start of the race. For this reason they may change their speeds erratically or even drop out of the race completely. Consequently, following the second gun shot (the second pulse) some of the racers may return together at the starting line but not all of them. In the nuclear spin system a similar situation prevails: either (1) the nuclear spins return to thermal equilibrium or (2) they lose phase memory of Larmor precession. The effect (1) occurs when the magnetic energy of precession contained by the moment is transferred to a molecule completely in the form of kinetic energy. The time in which this effect occurs is called T_1, the spin-lattice thermal relaxation time. The effect (2) arises when magnetic energy is transferred from spin to spin. One must also add to this the effect due to fluctuating local magnetic fields caused by neighboring moments and paramagnetic substances. The over-all time which relates to processes in the lattice which shorten the phase memory of precession has been denoted by T_2, which includes the effect of T_1 as well.

Another influence which destroys the phase memory of precession is that due to the self diffusion of molecules which contain resonant nuclei. Since there is an established gradient of the magnetic field over the volume of the sample, a molecule whose nuclear moment has been flipped initially in a field H_0, may, in the course of time 2τ, drift by Brownian motion into a randomly differing field H_0. Therefore, as τ is increased, a lesser number of moments participate in the generation of in-phase nuclear radio-frequency signals. The theory of the diffusion effect can be incorporated into the nuclear equations, and a useful expression is obtained by which the self-diffusion coefficient of molecules can be measured from the plotted envelope curve and known parameters.

It is possible to measure the thermal relaxation time T_1, independently of all other effects, by measuring signals proportional to the excess population of moments in the z direction at any time, as shown in Fig. 5. This can be done by measuring the amplitude at some arbitrary point on the free induction tail following the second radio-frequency pulse. This is compared to the amplitude at the corresponding point on the free induction tail on the induced signal following the first pulse, which is the amplitude proportional to the maximum available moment M_0. It can be shown that if $\gamma H_1 t_w = \pi/2$ for both pulses, then the tail signal following the second pulse is proportional to the number of gyromagnetic moments which have been thermally relaxed during the time τ. There are alternate methods of measuring T_1, for instance, from the observation of echoes obtained from an application of more than two radio-frequency pulses.

Recently a very interesting phenomenon of a ringing system in the case of an ensemble of molecular electric dipole moments has been demonstrated by R. H. Dicke at Princeton. The principle is very much like the case for pulsed nuclear induction except that a coherent electric field due to molecular rotation induces a signal in a microwave cavity following a strong microwave pulse at resonance. One aim of Dicke's method is to obtain higher resolution in spite of Doppler broadening. Similarly the pulsed echo method permits high resolution where long relaxation times (the equivalent of narrow natural line widths) can be measured in spite of artificial line broadening due to an inhomogeneous magnetic field.

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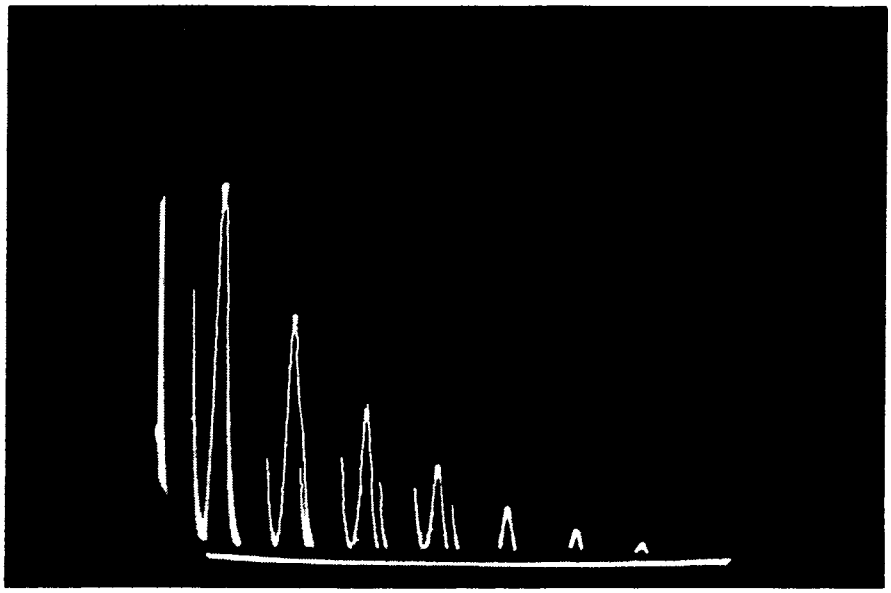


Fig. 6. Multiple exposures of single proton echoes. The first rf pulse occurs at the beginning of the trace and the second pulse is spaced from the origin at equal intervals for each exposure with the sample at thermal equilibrium. The echo envelope provides a measure of the phase coherence parameter T_2.

Appendix

Tissue Relaxation Times

Following **table 4** shows T1 and T2 relaxation times for various tissues (normal and pathological) at different magnetic field strengths.

Table 4: Tissue relaxation times

	T1 (ms) 0.25 T	T1 (ms) 0.5 T	T1 (ms) 1 T	T1 (ms) 1.5 T	SD (%)	T2 (ms)	SD (%)
BRAIN							
Gray matter	530	657	813	921	17	101	13
White matter	422	537	683	787	17	92	22
Tumours	667	802	963	1073	36	121	63
Meningioma	586	714	871	979	18	103	31
Glioma	845	887	931	959	35	111	33
Edema	667	806	973	1090	23	113	73
BONE							
Normal marrow	607	648	695	732	78	106	60
Osteosarcoma	740	811	888	973	28	85	30
BREAST							
Fibrotic tissue	409	547	732	868	18	49	16
Adipose tissue	190	214	241	259	28	84	36
Tumours	483	634	832	976	28	80	35
Carcinoma	451	595	785	923	25	94	48
Adenocarcinoma	490	686	959	1167	10	81	12
Fibroadenoma	508	715	989	1195	29	60	11
KIDNEY							
Normal tissue	417	496	589	652	27	58	24
Tumours	733	796	864	907	37	83	42
LIVER							
Normal tissue	250	325	423	493	22	43	14
Tumours	713	782	857	905	26	84	31
Hepatoma	621	769	951	1077	16	84	26
Chirrosis	328	367	410	438	21	45	
LUNG							
Normal tissue	488	599	735	829	19	79	29
Tumours	407	535	703	826	51	68	45
MUSCLE							
Normal tissue	409	547	732	868	18	47	13
Tumours	597	752	946	1083	32	87	40
Carcinoma	608	750	925	1046	16	82	73
Fibrosarcoma	831	896	967	1011	15	65	14
Rhabdomyosarcoma	664	827	1031	1173	27		
Edema	652	897	1235	1488	26	67	26
PANCREAS							
Normal tissue	302	371	455	513	25		
Tumours	718	942	1235	1448	15		
SPLEEN							
Normal tissue	431	543	683	782	19	62	27
Tumours						69	1

Final Words

You must have a sore head by now and I don't blame you ☺.

There is one more thing, though, I would like to stress: **A synonym for MRI is COMPROMISE**.
It can't be said enough. In fact I'm going to say it again: A synonym for MRI is COMPROMISE.
I think this is the most important thing to remember from this story.
MRI is a balancing act for Signal to Noise. Whenever you change a parameter, you'll have to change another to counteract for the signal change, if you want to keep the SNR the same.
Many systems offer some sort of "SNR Indicator". When you change a parameter it shows how much signal you will gain or lose. It's a very nice tool, because sometimes it is not at all clear how much impact changing one parameter can have. If your system does not have a tool like that, then be very careful when changing parameters.

I hope that reading this story has helped you to understand the complex matter of MRI physics.
It only scratched the surface and you're by no means ready yet to build your own MR scanner, but at least you know what the abbreviation MRI means ☺.
I know it is not easy to get the total picture after reading the story once. Don't blame yourself. Read the story again, and again, if need be. Then open up a book from the reference list and try it again. One day there will be a moment that the penny drops and you will be rewarded with an immense feeling of happiness and you realize that MRI is even better than sliced bread.

These pages are the result of many years of teaching MR physics to radiographers and technicians who just started with MRI. They are by no means complete, but I hope it will encourage you to explore the exciting world of MRI a bit more. Take it in good time, though. Do not move on until you understand the basics. It is very easy to get lost in the myriad of subjects.

This is the right time to express my thanks to all those people who encouraged and supported me to put it all on paper; Aida Ferreira, one of my first students, who asked me to do it; my colleague Johan Roubroeks for test reading and, above all, Deirdre, my wife, for editing the story and who put up with the long hours.
Thanks all.

Have fun, ☺

Evert Blink

amplitude of the magnetization as it precesses around the axes not the least of which is the fact that knowing exactly where in the rotating frame the magnetization is at a given time (rather than just an amplitude away from the z' axis, for example) represents more information. A phase sensitive detector, described in Chapter V, is used to gain information about projections of the rotating magnetization along any axis in the x'-y' rotating plane. With phase sensitive detection, the return sweep in the fully relaxed case will yield an inverted signal so that there will be a special delay time $t_{\frac{1}{2}}$ between the two sweeps for which the magnetization is a null. This will occur when $t_{\frac{1}{2}}=T_1 \ell n 2$, where T_1 is the relaxation time of the magnetization towards the positive z axis.

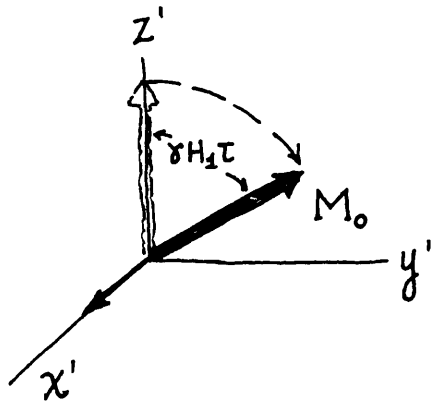
I.C. PULSE NMR

The first NMR experiments were the cw (continuous wave) experiments by Bloch, Purcell, and their respective colleagues. The first pulse experiments followed shortly thereafter and Hahn published his now famous spin echo paper in 1950. For the following decade and a half, pulse NMR was used mostly for determinations of relaxation times T_1 and T_2. In the meantime Lowe and Norberg proved that an NMR frequency spectrum is related to an FID by Fourier transformation and Cooley and Tukey rediscovered an efficient Fourier transform algorithm. Ernst and Anderson recognized the potential for great efficiency in using Fourier transform pulse NMR for multiline high resolution spectra and performed the first successful demonstration. More recently, line narrow-

ing experiments with solid samples have become possible due to the pioneering efforts of Hahn, Mansfield, Waugh, and their colleagues, and the overwhelming majority of these utilize pulse techniques. In the next two sections, we introduce you to the pulse NMR experiments without getting into the hardware or even the Fourier transformation process.

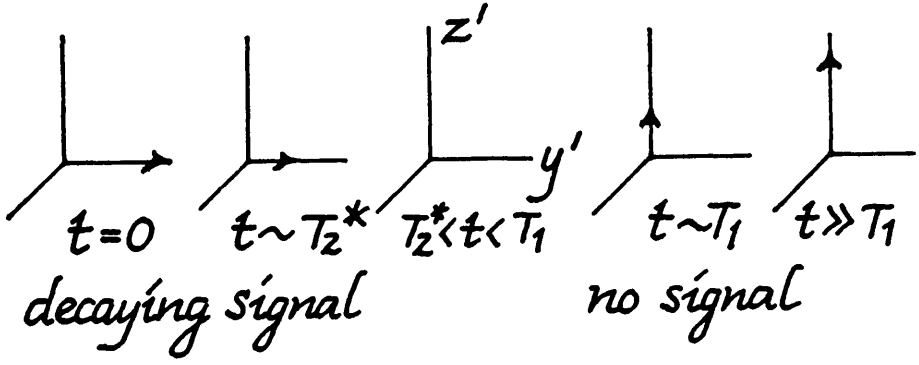
I.C.1. INTRODUCTION TO PULSE NMR

In pulse nuclear magnetic resonance (NMR) experiments, an intense rf pulse in the laboratory frame, with an amplitude H_1 larger than that used in cw experiments, is applied to the sample. At resonance, $\overline{H}_e=\overline{H}_1$, say along the rotating x' axis, and the magnetization will rotate about $\overline{H}_1$ in the rotating frame y'-z' plane with an angular velocity $\gamma\overline{H}_1$. If such an $\overline{H}_1$ field is turned on at resonance for a time τ, the magnetization vector will be rotated by an angle $\gamma H_1\tau$. An H_1 field of the correct amplitude and duration to produce a rotation of 90° is called a 90° (or a $\pi/2$) pulse. If the H_1 field is applied for twice as long, this results in a 180° (or a π) pulse. Similarly one could obtain a 270° pulse, a 360° pulse, etc.



Consider the 90° pulse. Immediately following it, the magnetization lies along the rotating y' axis. This magneti-

zation will decay in time as the system comes back to equilibrium due to two mechanisms. The magnetization dephases in the x'-y' plane because the field inhomogeneity, for example, will cause some of the nuclei to precess faster than others but it also tends to regrow along the z'=z direction because thermal equilibrium is being re-established. After the 90° pulse, the magnetization will induce a decaying sinusoidal voltage in a pickup coil in the x-y plane. For a single Lorentzian line the decay of the magnetization in the x-y plane will be exponential with the time constant T_2^* and the output of the pickup coil will be a sinusoidal wave whose amplitude is decaying at the rate $\exp(-t/T_2^*)$. If one waits long enough, the magnetization will re-establish itself along the z direction in equilibrium with the applied field with a time constant called the spin-lattice (or longitudinal) relaxation time T_1, i.e., it will recover as $M_z=M_o[1-\exp(-t/T_1)]$. The decay rate for the magnetization in the x-y plane is usually larger than or equal to the decay rate for the recovery of the magnetization along the z direction.



Because the pickup coil is sensitive only to the component of magnetization in the x-y plane, the magnetization being re-established along the z' axis due to the T_1 process is not

detectable until it is rotated away from the z' axis. Therefore a second pulse is always necessary to measure the spin-lattice relaxation time T_1. The basic idea is to perturb the magnetization from its equilibrium state with one pulse and examine its recovery along the z' axis after a variable delay with another pulse, for example, a 90° pulse. Several methods for measuring T_1 are described in III.D.

The effect of a 90° pulse can be duplicated in cw NMR although with more difficulty than in pulse NMR. When a resonance line is swept rapidly in the conventional cw high resolution experiment, a ringing pattern follows the line if phase detection is used. Such a ringing pattern occurs in the following way: As one sweeps into resonance, the magnetization is tipped away from the applied magnetic field as in adiabatic fast passage, as discussed in I.B. However, if one sweeps through resonance too rapidly, that is, non-adiabatically, the magnetization gets left behind by the effective field H_e. (The best chance of this happening is near resonance because the precession frequency γH_e about H_e is the smallest and the rate of change of the H_e direction is the greatest there.) After the passage, the field has been increased way beyond the resonance value for that line and the magnetization will freely precess. If a phase sensitive detector is used, the signal from the pickup coil will get in and out of phase with the reference signal in the detector giving rise to a decaying ringing pattern. The length of time during which the ringing pattern persists is the time during which there is a net x-y component of the magnetization. Therefore, under conditions in which the H_1 field does not exert a torque on the magnetization, the ringing pattern will be observed to decay in time as $\exp(-t/T_2^*)$. The T_2^* time constant here is identical to the T_2^* observed following a 90° pulse in a pulse experiment.

The magnetization decay rate constant in the x-y plane, $1/T_2^*$, consists of a term related to the recovery rate constant for the magnetization along z, $1/T_1$, as well as other mechanisms which give rise to a dispersal of the magnetization only in the x-y plane. Two examples of the latter are $1/T_2'$ which results from dipolar processes (to be discussed later) and $\gamma\Delta H_0$ which is due to the applied magnetic field inhomogeneity ΔH_0. In this case the relaxation rate is approximately

$$1/T_2^* \cong 1/(2T_1) + 1/T_2' + \gamma\Delta H_0$$

although usually one of these mechanisms will dominate. In many high resolution spectra, T_2^* is determined by the spin lattice relaxation time T_1 which, in turn, is affected by motional parameters of the molecule such as characteristic times for reorientation and diffusion. In a typical solid, on the other hand, the dipolar contribution T_2' is much more likely to determine T_2^* because the effects of these fields from the neighboring spins do not cancel as they do in liquids wherein the molecular tumbling averages out the dipolar interaction. Strictly speaking, the line will not even be Lorentzian in most solids but T_2' and T_2^* still can characterize some decay or dephasing time. For very narrow lines such as for natural abundance ^{13}C nuclei without proton and fluorine neighbors, the magnetic field inhomogeneity can be the main contribution to the line width and the relaxation time T_2^*. This is usually the least interesting and informative contribution to the T_2^* process and there are ways such as the CPMG method discussed in the following section to get around this problem.

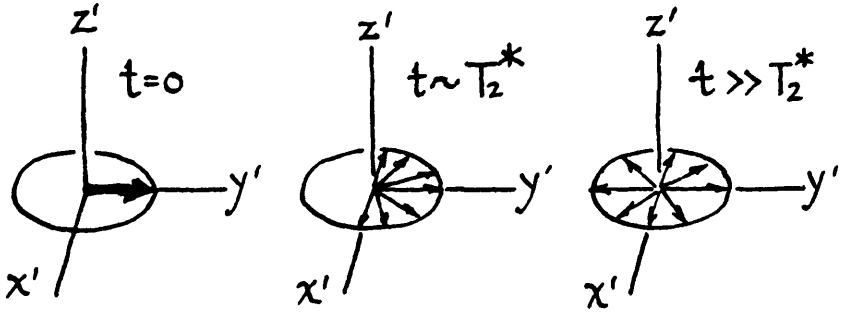
I.C.2. PULSE NMR EXPERIMENTS

At this point, we study in more detail the behavior of the magnetization following an rf pulse. Consider a sample placed between the poles of an electromagnet. In any real magnet, different parts of the sample experience different fields owing to magnet imperfections. In NMR spectroscopy it is important to minimize this inhomogeneity in the applied static field for maximum resolution, although the required homogeneity varies widely depending on the experiment.

Suppose a 90° pulse is applied to a sample in an inhomogeneous magnetic field. If the pulse is very intense, the entire magnetization will be rotated through the appropriate angle in a very short time. Shortly thereafter, however, the magnetization amplitude will decrease due to the field inhomogeneity in the following way. The total magnetization vector is the sum of smaller magnetization vectors each arising from a small volume experiencing a homogeneous field. After a 90° pulse, each of these components of the magnetization will precess with its own characteristic Larmor frequency. Therefore, the magnetization from those portions of the sample with slightly larger magnetic fields will precess faster than those which are in a smaller field. As a result, the different contributions of the magnetization will get out of phase with each other. The contribution to the magnetization which arises from one such small segment of the sample, experiencing a homogeneous applied field, is called a spin isochromat. The term isochromat is used because it implies a constant frequency (i.e., color) and the implication is that all the nuclear moments in that segment of the sample will precess with the same Larmor frequency.

In the following figure, the results of applying a 90°

pulse in an inhomogeneous field is seen in the rotating reference frame. In the rotating frame the magnetization immediately following the pulse starts out along the rotating y' axis. Shortly thereafter, some of the components of the magnetization, i.e., some of the spin isochromats, start getting ahead of the average and some start getting behind.



The result is that the net magnetization along the y' axis decays to 0 as the spin isochromats fan out in x'-y' plane. If there is a pickup coil in the x-y plane, the voltage induced in the coil will decay as the spin isochromats fan out. This signal following a pulse is the free induction decay (FID). [See the last figure in Section I.A.] As a quantitative example, consider protons in a field inhomogeneity ΔH of 1 gauss across the sample. [That's quite a poor magnet -- or a huge sample.] Since the angular velocity is given by γH, its spread in this sample is $\gamma\Delta H=(4.3\ \text{kHz})\times 2\pi$. This means that in a time of the order of 1/(4.3 kHz) the fastest isochromats would have crossed the slowest isochromats.

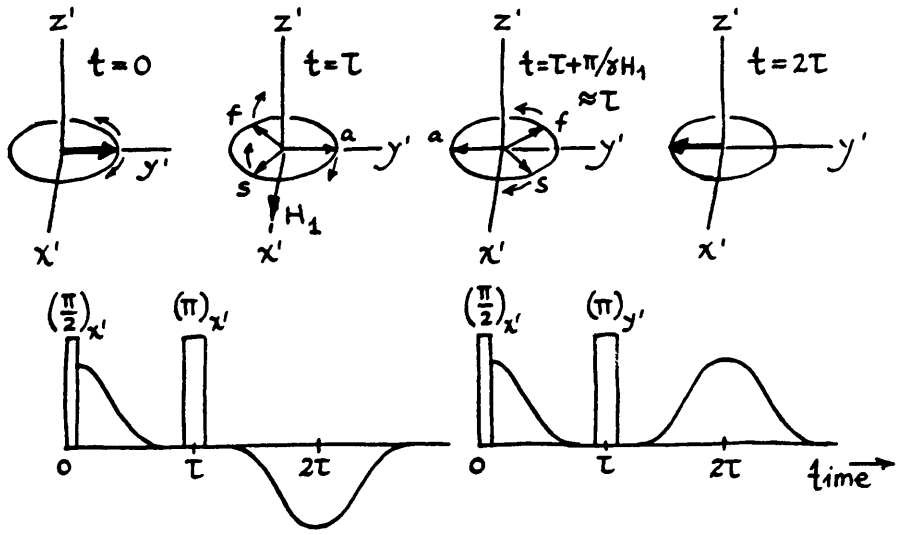
As mentioned in the previous section, the time constant which describes the decay of the magnetization in the x-y plane is T_2^*. We define an intrinsic relaxation time which is characteristic of the magnetization decay in one of the spin

isochromats without any field inhomogeneity effects and this is the spin-spin or the transverse relaxation time T_2. Thus, T_2, T_2^*, and the spread in Larmor speed $\gamma\Delta H_0$ due to field inhomogeneity ΔH_0 are related by $1/T_2^*=1/T_2+\gamma\Delta H_0$. If the magnetic field homogeneity is improved, T_2^* will increase until it becomes equal to T_2. When T_2^* is dominated by magnetic field inhomogeneity, it will give us little information about the sample. However, the T_2 processes which cause the magnetization decay within a spin isochromat, will be due to fundamental molecular processes and are likely objects of the experiment. The time constant T_2^* is related to the line width in the frequency spectrum as discussed in III.B.

While all this is going on, spin-lattice relaxation is also taking place. The spin-spin relaxation does not involve any exchange of energy with the world outside the spin system, whereas the spin-lattice relaxation depends on an outside agent (which is called the lattice) accepting energy from the spin system so that the latter can relax towards the thermal equilibrium state given by the Boltzmann populations discussed near the beginning of this chapter. As we will discuss in III.A., the lattice can be of many possible forms such as molecular rotation, diffusion, or lattice vibration. We only note here that the spin-lattice relaxation process can be quite weak in the absence of molecular motion or paramagnetic ions as in many rigid solids but it can be very strong in many solutions and in some solids which exhibit molecular motion. The spin-spin relaxation process contains a contribution from the spin-lattice process so that $T_1>>T_2^*$ in most solids while $T_1\sim T_2^*$ in most liquids.

Ingeneous experiments have been designed to remove the effect of the applied field inhomogeneity. They are called spin-echo techniques and have application in the realm of

Fourier transform NMR as well as pulse NMR (Hahn, 1950). Suppose that a 90° pulse is applied along x' at time 0 to a spin system for which magnetic field inhomogeneity is the major contribution to $1/T_2^*$. Shortly thereafter, the spin isochromats will have dephased in the x'-y' plane. As a result, there is no net magnetization component in the x'-y' plane, although the individual spin isochromats have not yet dephased. Suppose at a time τ later, a 180° pulse is applied along the x' axis. Any magnetization along the z direction would simply



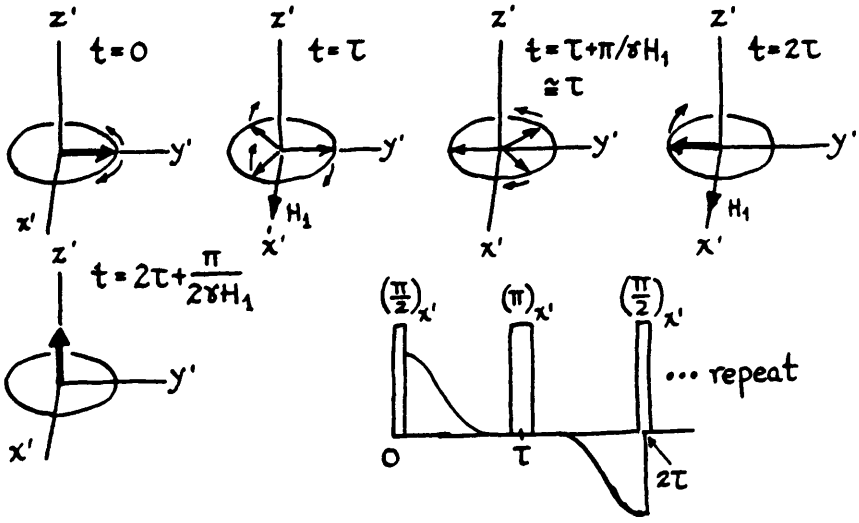
be inverted to the -z direction and be of no consequence. Of the magnetization remaining in the x'-y' plane each one of the spin isochromats would be rotated 180° about the x' axis. As a consequence, those spin isochromats which had gotten ahead of the average spin isochromats by a certain angle are now behind the average of the pack of spin isochromats by the same amount. Those spin isochromats which were going slower

than average and had gotten behind the rotating y' axis are now ahead of the rotating y' axis by the equivalent amount. Therefore, following the 180° pulse, the spin isochromats begin to rephase to form a net magnetization as the rapid isochromats catch up with the slow ones. The result is that the magnetization becomes refocussed along the -y' axis at time 2τ and it will cause an inverted spin echo, as shown in the sketch at left. The spin echo consists of two FID's back-to-back. It is also possible to get spin echoes by applying the 180° pulse along the y' axis in the rotating frame. The refocussing will then take place along the y' axis so the echo will have the same sign as the FID.

At this point we pause in our discussion of pulse experiments in order to give an application of spin echoes. One of the problems in accumulating FID's is that enough time must elapse between the FID's in order for the magnetization to build up along the z'(=z) axis so that the signal is reasonably large. If $T_1 \cong T_2^*$ as in many liquids this is not a serious problem since the recovery along the z axis will already have taken place by the time the FID has been recorded. If, on the other hand, T_1 is long compared with T_2^*, time intervals of the order of T_1 are still needed between the FID's even though the FID has disappeared long before.

One way to increase the efficiency of data taking in such a case is a sequence called DEFT for driven equilibrium Fourier transform (Becker, et al. 1969). The idea of this experiment is to force the system back to equilibrium more rapidly after each FID so that it can be pulsed more frequently. In a DEFT sequence, a 90° pulse is applied to the system, and the FID is recorded. A 180° pulse is applied at a time τ when the FID has disappeared due to the dephasing of the spin isochromats to form an echo at time 2τ. At the

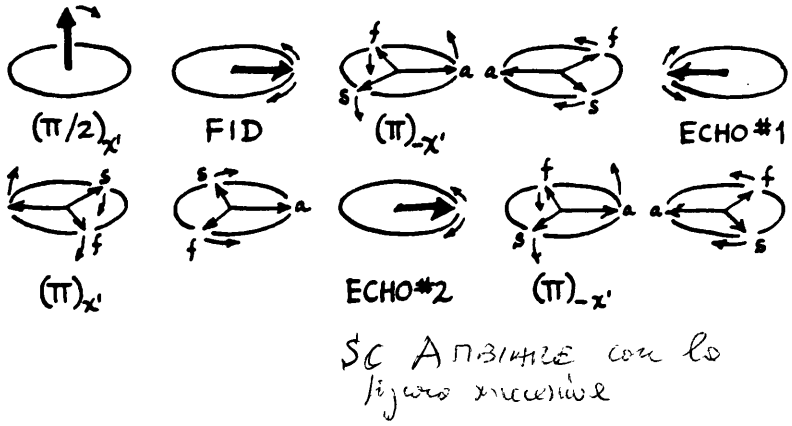
peak of the spin echo, when all the magnetization in the x-y plane that can be refocussed will be along the -y' axis, a 90° pulse is applied to rotate that magnetization back to the z



direction. In other words, the system has been forced back close to thermal equilibrium and we are prepared to repeat the experiment by applying a 90° pulse and recording the FID again.

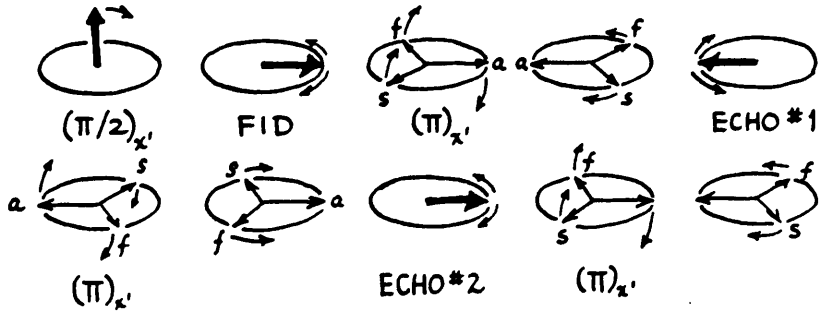
We shall now return to spin echoes by expanding the discussion to include echo trains. For simplicity, let us assume that we have only one line in a high resolution spectrum. On exact resonance in an inhomogeneous field, the 90° pulse yields an FID with a time constant T_2^*. At a time τ later, a 180° pulse is applied and the echo maximum occurs at time 2τ, since the time required for rephasing the spin iso-

chromats equals the time it took for dephasing. Because the spin echo arises from magnetization that has regrouped along the -y' axis, the echo will be inverted compared to the FID. Another 180° pulse at time 3τ will result in another echo at time 4τ and this will be right side up. One can continue to apply 180° pulses with a spacing of 2τ with echoes occurring between each of the pulses. This spin echo train is called a Carr-Purcell (CP) spin echo train (Carr and Purcell, 1954).



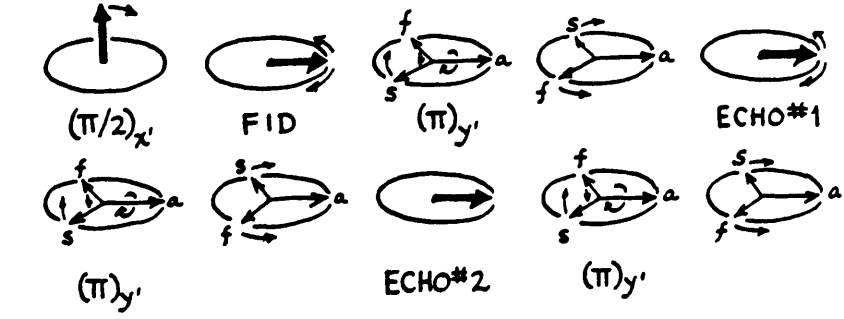
The echo amplitude maxima should decay with the time constant T_2, the intrinsic spin-spin relaxation time, which is the time it takes for the magnetization to decay in the x-y plane in the absence of any external field inhomogeneity.

In practice, Carr-Purcell echo trains usually result in measured T_2's that are too short because of cumulative errors of each pulses not being exactly 180° pulses and of H_1 inhomogeneity which spreads out the magnetization in a plane containing H_0 and H_1. One way to compensate for these errors is to alternate the phase of each 180° pulse by 180° phase shifts as shown below. The first 90° pulse occurs with the rotating H_1 field along the rotating x' axis. The first 180° pulse, how-



ever, would have its rotating H_1 field along the -x' axis. The second 180° pulse would have the rotating H_1 field along the rotating x' axis and so on. In this way, any pulse length errors are cancelled on alternate echoes.

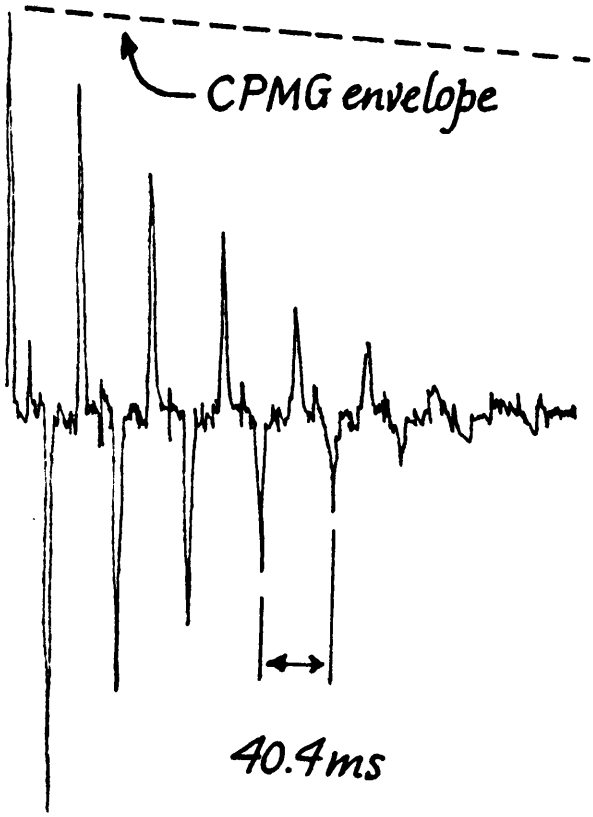
A slightly simpler and more common spin echo sequence for measuring long T_2's is the Meiboom-Gill modification of the



Carr-Purcell sequence (CPMG) in which all the 180° pulses in the train are phase shifted 90° with respect to the inital 90°

pulse, i.e., if the 90° pulse is along x', the 180° pulses are along ±y' (Meiboom and Gill, 1958). Now all the echoes form along H_1 for the 180° pulses in the rotating frame regardless of the exact tip angle, and this is shown at left (bottom).

CPMG echoes also differ from CP echoes in their having the same sign since the echoes are always formed along the same direction in the rotating frame. This is a slight inconvenience compared to the bipolar CP echoes which have no baseline ambiguity. An example of a CP signal with the corresponding CPMG envelope indicated is shown below.



This example is taken from ^{13}C T_2 measurements in liquid CO at 77K at 8.9 MHz where T_2 was 5.8 seconds as determined by

the CPMG data.

Suppose we had decided to measure this T_2 without using the CPMG method. From the relation

$$1/T_2^* = 1/T_2 + \gamma\Delta H_0$$

we know that $\gamma\Delta H_0 << 1/T_2$ in order for the measured T_2^* to accurately reflect T_2. Because $T_2 \sim 6$ s, let us say that $\gamma\Delta H_0$ must be less than 10^{-2} radians/sec. For ^{13}C, $\gamma \sim 2\pi$ kHz/G so that ΔH_0 must be less than $(10^{-2}/2\pi x 10^3)$ G or about 1.5 μG! In a field of about 10^4 G, that is an inhomogeneity of 1.5×10^{-10} which is well beyond the capability of the usual NMR magnets.

It is obvious that the CPMG sequence is particularly useful for determining the spin-spin interaction time T_2. Long T_2's requiring CPMG measurements usually contain information about very slow motions of the molecules containing the nuclei. See Sections III.E.1. for details.

The spin echo train has an additional use in NMR. Remember that a spin echo consists of two FID's back to back. Suppose we co-add the echoes. Out of a spin echo train, one could get a fairly large number of FID's which could be added together to produce an enhanced S/N. This could be done simply by triggering the digitizer at each echo. With more sophistication, S/N can be further improved by taking the left-hand half of each echo, reversing it left to right, and adding it to the right-hand half. Using spin echo trains to improve the S/N in Fourier transform NMR is called SEFT (Spin Echo Fourier Transform) (Allerhand and Cochran, 1970). It is clear that SEFT will be advantageous under the same conditions that DEFT is advantageous, namely when T_2^* is much smaller than T_1. Neither DEFT nor SEFT have found

widespread application in NMR but are discussed here because they are techniques which add to the understanding of pulse NMR.

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I.D. INTRODUCTION TO FOURIER TRANSFORM NMR

In the following three sections, we will get our feet wet with simple concepts of Fourier transform NMR and the time and frequency domains which are related by the transform. Detailed treatments will be given in Chapter II.