While saturation recovery yields proton (spin) density images, the images are T_1-weighted with partial saturation.

- A spin echo sequence has a 90° pulse, which is followed by one – or more – 180° pulses, to rephase the dephasing protons resulting in one – or more – spin echoes. This sequence can give proton density-weighted, T_1-weighted, or T_2-weighted images. This is determined by the imaging parameters which are chosen (TR, TE).
- In the inversion recovery sequence, a 180° pulse is followed by a 90° pulse, resulting in T_1-weighted images.
- Fast imaging sequences use flip angles that are smaller than 90°, and so-called gradient echoes. Image weighting is also determined by the specific type of sequence and the imaging parameters chosen.

About imaging time

As we have just seen, fast imaging sequences decrease **imaging time**.

Is there any other way to decrease this time? What does actually determine the **imaging time**?

For MR imaging with normal pulse sequences, this can be easily calculated; the **acquisition time (a.t.)** is:

a.t. = TR x N x N_{ex}

This looks a little complicated but it isn't really. Let us start at the back. N_{ex} is the number of excitations. What does that mean?

For certain reasons, it is necessary to use not only one signal measurement, but to repeat the measurement several times. As the MR signal coming out of the patient is very weak, it may be good to add up signals from several measurements, to take several "**averages**", to get a good quality image.

Actually, what you get is an image with a better signal-to-noise ratio.

Naturally, imaging time increases with every additional measurement.

Repeated measurements result in a better signal-to-noise ratio.

To illustrate this: Just imagine that you are sitting in a large audience, where people are making a lot of noise. Someone sitting next to you whispers something in your ear, but you cannot really understand him, because there is so much background noise. What you will probably do, is ask him to repeat what he said once or several times. You mentally add up the information which you receive each time. As this signal is always the same, it will increase by adding it up. The **background noise**, however, is not always the same.

Instead, it is random and fluctuates and does not add up the way the signal does. So altogether you will have a better **signal-to-noise ratio** (which you would also have if the person spoke louder). Back to our formula: What is "N"? As you know from other imaging methods (or your PC), pictures are made of **picture elements**, which all together make up the image matrix, e.g. a 1,024 x 768 matrix has 768 rows of 1,024 picture elements (**pixels**).

In our equation, N is the number of rows in a **matrix**, like rows in a letter.

The more rows you have, the more time it takes for the image.

Just think about this as if you were writing a letter: if you have paper with 5 rows on a page, you will finish a page faster than if you have 25 rows to write. However, you have more content, more detail on a page/picture, when you work with more rows.

And why does TR influence acquisition time?

If you choose a long time TR to repeat your pulse sequence, to perform additional signal measurements, imaging takes longer than with a short TR. However, there is a trick that can shorten imaging time a bit.

While we are waiting to repeat our imaging sequence in one slice, i.e. while we wait for TR to go by (slice A in figure 56), we might as well make measurements in one or more different slices (slices B, C and D in figure 56). The longer the TR, the more slices we can excite in the meantime.

So by just adding a little extra time, we will examine many slices instead of one, and imaging time per slice decreases substantially.

We perform so-called **multislice imaging**. Another way to possibly reduce TR, and thus imaging time, is the use of a **contrast medium**: as we have seen, Gadolinium-DTPA shortens T_1.

And when T_1 is shorter, the TR can also be shorter, without a loss in signal intensity of the tissue in question (see figure 49).

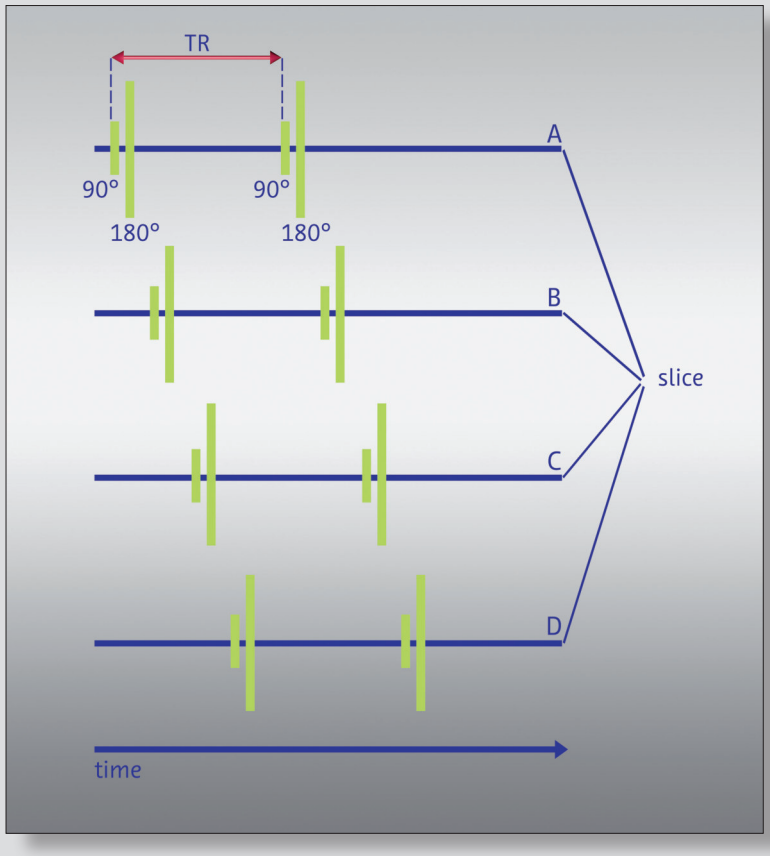


Fig. 56: Multislice imaging: while we wait for the time TR to pass by for another signal measurement in slice A, we perform signal measurements in additional slices. So during time TR, we actually recorded signals for more than one slice.

Let us review important factors that influence signal intensity in MR.

These are:

- proton density (page 45)
- T_1 (page 24)
- T_2 (page 28)
- flow (page 67)
- the pulse sequence (page 74-80)
- TR (page 45)
- TE (page 54)
- TI (page 77)
- flip angle (page 80)
- use of contrast medium (page 71)

If you are not sure about any of these, go back to the corresponding page.

If you feel familiar with these facts, continue with the next section, and learn about some important things in MR imaging that we have not talked about yet.

How can we select a slice which we want to examine?

When we put a patient into an MR scanner, he or she is in a rather homogeneous magnetic field.

So all the protons in the whole body have the same **Larmor frequency**, and will be excited/disturbed by the same RF pulse. To examine a specific slice only, a second magnetic field is superimposed on the external field, which has different strengths in varying locations. The magnetic field is therefore stronger or weaker in some places than in others (figure 57).

This additional field is called a **gradient field**, and is produced by the so-called **gradient coils**. This gradient field modifies the strength of the original magnetic field.

In figure 57, magnetic field strength increases for different cross sections from the feet towards the head.

Consequently, the protons in the different slices experience different magnetic fields, and thus have different precession frequencies.

So the RF pulses which disturb the protons in the different slices, must have different frequencies as well – otherwise there would be no resonance.

As gradient fields can be superimposed in any direction, it is possible to define not only transversal slices, but all kinds of different imaging planes without moving the patient. The gradient field that enables us to examine a specific slice, is also called **slice selecting gradient**.

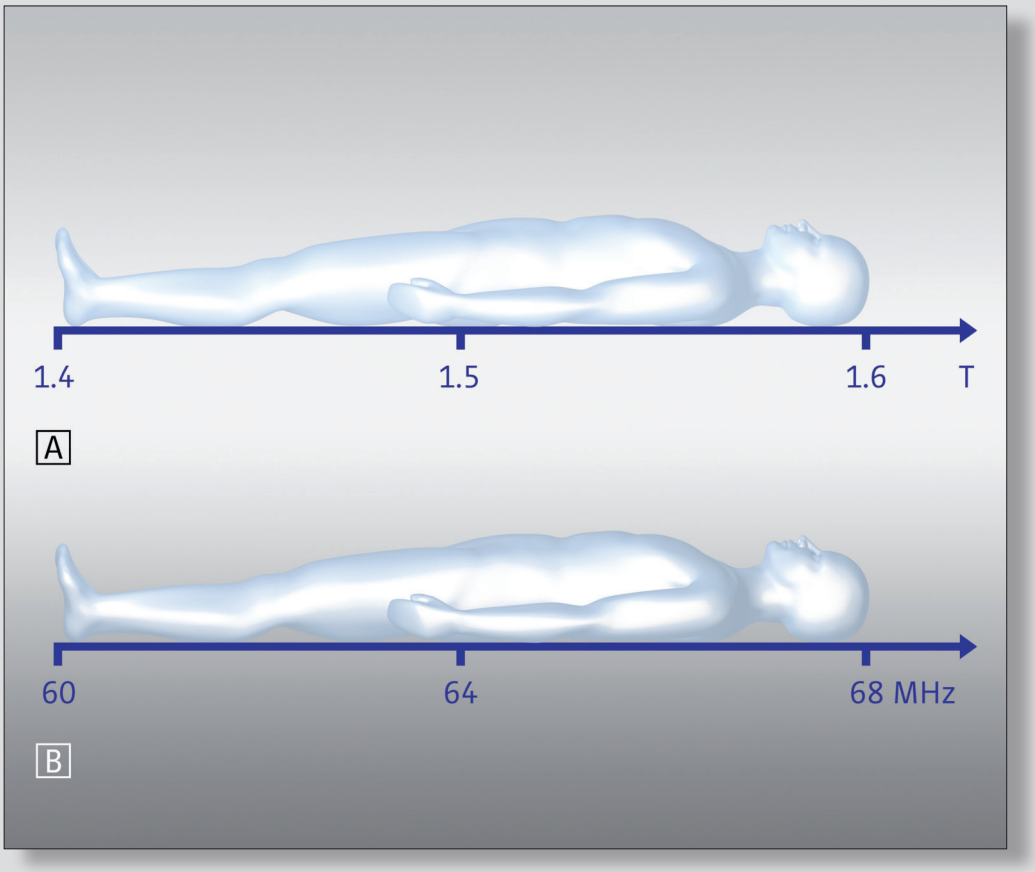


Fig. 57: Magnetic gradient fields are superimposed on the field of the MR magnet, so that different cross sections of the body experience magnetic fields of differing strength. In the illustration, the resulting magnetic field strength is increasing from 1.4 Tesla at the feet to 1.6 Tesla at the head. As magnetic field strength and precessing/ resonant frequency are directly correlated (Larmor equation), the resonant frequency at the feet is about 60 MHz, while it is about 68 MHz at the top of the head in our example. By selecting a certain RF pulse frequency, we determine the location of the slice which we examine.

How can we determine or select a certain slice thickness?

We can select a different **slice thickness** in two ways (figure 58):

One solution is to send in not only one specific frequency (which is not done in practice), but an RF pulse that has a range of frequencies, which is often referred to as **bandwidth**; the wider the range of frequencies, the wider the bandwidth, the thicker the slice in which protons will be excited. This is illustrated in figure 58.

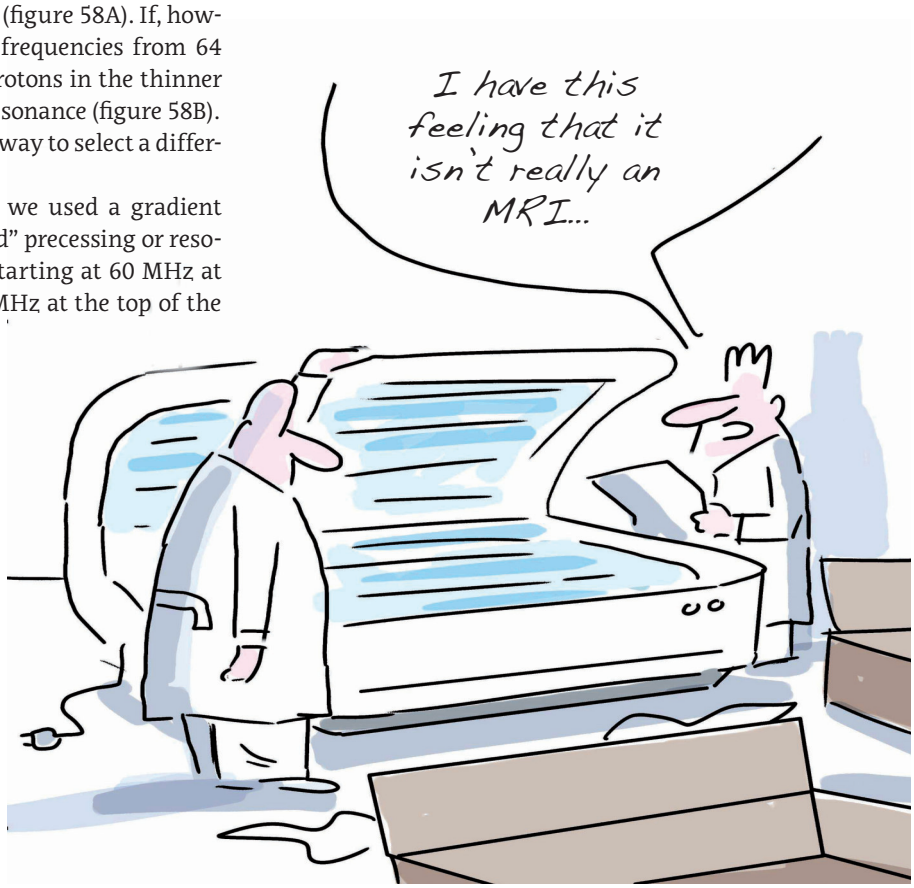
If we use an **RF pulse** with frequencies from 64 to 65 MHz, we will get the thickness of slice 1 (figure 58A). If, however, we only use frequencies from 64 to 64.5 MHz, the protons in the thinner slice 2, will show resonance (figure 58B).

There is another way to select a different slice thickness:

In our example, we used a gradient field that "produced" precessing or resonant frequencies starting at 60 MHz at the feet, up to 68 MHz at the top of the head. If we, however, have a steeper gradient field, i.e. one that has more difference in field strength over a specific distance, the precession frequencies will also vary to a larger degree, let us say from 56 MHz to 72 MHz.

If we now use an RF pulse of the same bandwidth as in A, containing frequencies between 64 and 65 MHz, the slice thickness in our example C with the steeper gradient field is, however, smaller than in our example A with the more shallow gradient field.

So using the same range of radio frequencies, the same bandwidth as it is called, slice thickness can be modified by the slope of the gradient field.



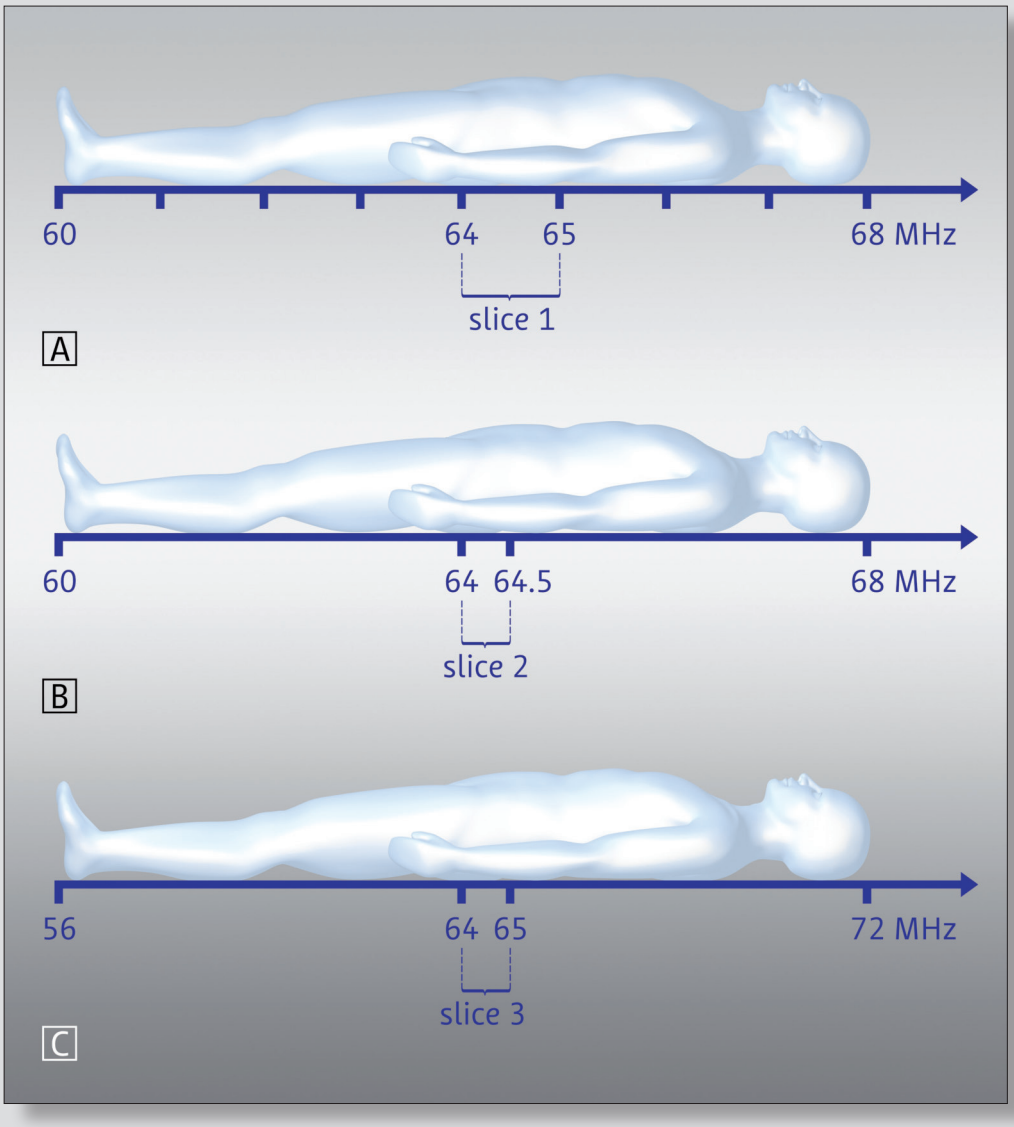


Fig. 58: There are two ways to determine slice thickness. The first is to use an RF pulse that has not only one specific frequency, but a certain range of frequencies, a so-called bandwidth. If, for example, we send in an RF pulse, which contains frequencies between 64 and 65 MHz, protons in slice 1 will be influenced by the RF pulse.
When the RF pulse only contains frequencies between 64 MHz and 64.5 MHz, thus has a smaller bandwidth, slice 2, which is half as thick as slice 1, will be imaged.
When there is more difference in magnetic field strength between the level of the feet and the head, i.e. the magnetic gradient is steeper, the resulting slice will be thinner, even though the RF pulse bandwidth is the same. This is illustrated in (C), where the magnetic field strength varies more between the feet and the head than in (A); the corresponding resonant frequencies are 56 to 72 MHz in (C) vs. 60 to 68 MHz in (A). Using the same RF pulse containing frequencies from 64 to 65 MHz results in imaging of a thinner slice 3 in (C) than in (A).