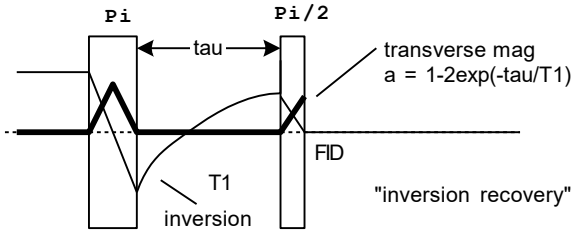
---------- **Chapter 11: Experiments on non-interacting spins (315-335) ------- 20 p ---------- ###**

11.1 Measuring T_1 by doing inversion recovery (315). This is a two pulse experiment as shown page 316 with a delay between that you vary. The initial π_x pulse creates an "inversion" in the population, then that inversion "recovers" some amount during the τ pulse, then you monitor the amount of recovery by hitting the sample with a $\pi/2_x$ pulse which then tilts whatever you have over into the -y axis, and then things spin from there, in the x-y plane, making M_x which you read out. I did the math, I see that the expression given for $a(\tau)$ is obviously correct. Suppose there are three resonances when you do this. Do your FT only in the t domain (not the τ domain) and you get a surface as shown page 318. For very long τ, you have recovered all the way to TE so you see "the usual peaks" as you go left to right. But at τ=0, $a(\tau)$ is negative as the expression for $a(\tau)$ obviously shows, so all three peaks start out negative on this surface, and by the same amount. Then you "graphically" measure the exponential parameter for each of the recovering peaks and that gives you a separate T_1 for each resonance. Got it.

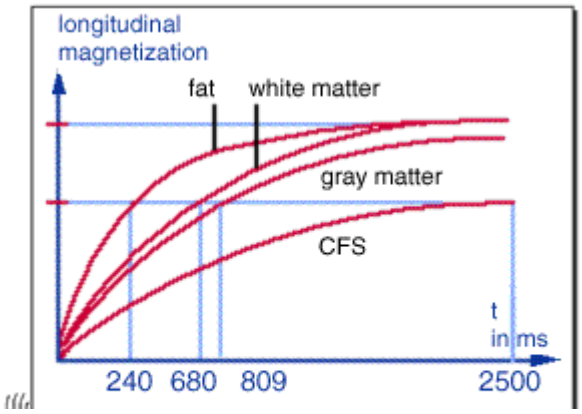
Perhaps the T_1 time itself might tell you something. I think this is the basis of the T_1-weighted MRI image, but of course T_1 is computed automatically somehow at each pixel in the slice. Page 319 shows another example of inversion recovery, in a case where there are 4 resonances.

Notice that you have to do the spectrum in this case to get a picture you can work with if there are multiple peaks with different T_1's. Otherwise you are stuck with a surface bottom page 317 which is hard to understand because you don't know where the resonances are located.

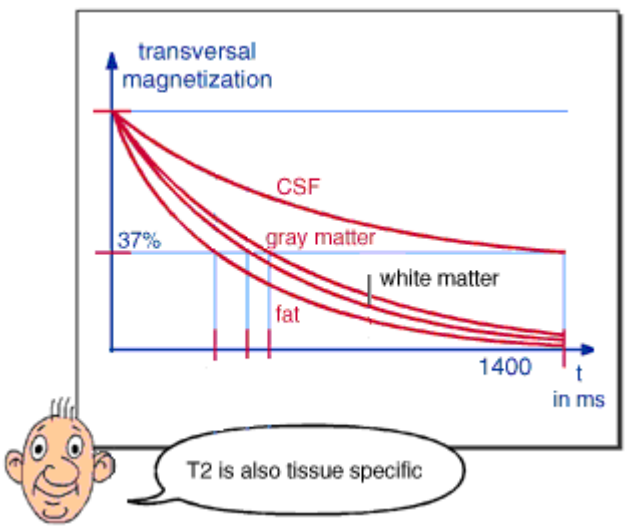
I know that "short τ inversion recovery" in MRI is called STIR, but Levitt does not mention this here, and I don't think anywhere, but I certainly will be able to read about it now that I have the tools. Here is what this experiment looks like:



The second pulse takes however much inversion is left and converts it to coherence hence signal.



Comment: Computer could automatically compute this data and report out your $(T_1)_\ell$ for each peak ℓ. For MRI, we are looking at the H peaks from water molecules, so probably only one peak to worry about. Then $a(0) = 1-2\exp(-\tau/T_1)$ is a measure of T_1 for each sample pixel, so you could then get $a(t=0,x,y) = 1-2\exp(-\tau/T_1(x,y))$. If



you choose τ to be on the order of brain T_1 times, Then the intensity a^2 of your image will in effect be a measure of $T_1(x,y)$ and this would be the famous "T_1-weighted image". Here is some web info on the right. Upper is T_1, lower is T_2.

11.2 Spin Echoes: Measurement of T_2 (318)

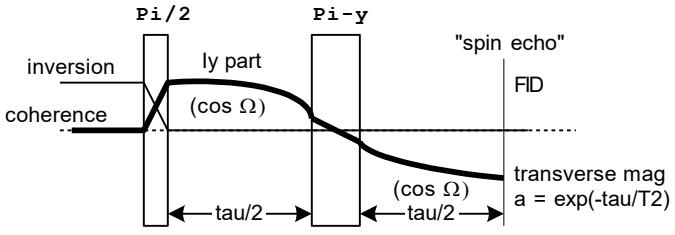
11.2.1 Broadening two ways (319). The inhomo type is due to variations in local environments of spins at the macroscopic scale, ie, variations in the sample material, or variations in the B field over this material. The homo type involves the microscopic mechanism which causes dephasing, and this is what T_2 is all about. But generally you get both types going at once, so you cannot just measure the width of a peak and thereby know T_2.

11.2.2 Inhomo broadening in the time domain (319). Let's jump to page 324 which shows the main idea. Different macroscopic chunks of your sample precess at different rates, so you dephase by this effect as shown. The little arrows from the different regions "fan out". As this happens, it helps make your FID decay more rapidly than it would just from the normal microscopic T_2 effect. This is shown in the upper picture on page 320. This is why you cannot just measure the width to get T_2.

11.2.3 The spin echo method. (320). This is very cool. The picture shows the pulse sequence with equal internal delays. First, if you do the usual math, you find that the second free propagation sort of reverses what the first one did. You end up with $a(\tau)$ as shown top page 323. Just as in the inversion recovery case, we get a simple expression and Ω^0 does not appear, although it appeared at intermediate points in the pulse sequence. Notice that you start getting Ω^0 dependence when you knock things over with your $\pi/2$ pulse. In the inversion recovery case, that happened just before the FID, but here you do it right off the bat! So, by running this experiment varying τ, you again could "plot" $a(\tau)$ and deduce the value of T_2. In this case, you don't have to do a spectrum if there is only one T_2 source spin set. If there were several sources, I guess you would to the FT in τ and look at the surface and read off the rate at which the peaks decay. Malcolm does not say this in this section, maybe later.

There is s subtle point here that is not shown so far. The claim is that this method removes the inhomo broadening in a very clever way. Next section.

Here is the experiment:



11.2.4 Refocussing (323). The figures at the bottom were first confusing to me. This is showing the FID signal from our spin echo experiment starting at the t=0 FID time. Mysteriously the signal grows and only then starts fading as it dephases. The explanation is well done on page 324 and 325 with great pictures. During the first pulse, you get that inhomo dephasing and the little arrows fan out, with the light ones rotating faster than the slow ones. But after you do your π_y flip, this same speed differential is used to de-

fan the vectors! As these vectors "de fan", the signal builds up, reaching a maximum after the end of the second pulse period (which is equal to the first). Then the signal fades away again. If it takes a long time for this pulse to build up from nothing, it looks like an isolated "echo" out there.

Now, if molecules MOVE during your experiment, the cancellation of the second part will not be perfect and the de-fanning will be less effective and the echo peak won't we as high. You can therefore use this method to measure flow rates !!!!

11.3 NMR Imaging (326). An excellent section. First we look at a 1D rod shown page 326 and we apply a gradient B field as shown page 327. This field points in the z direction, but varies with x across our rod sample. Obviously the Larmor is now a function of x. The Fourier spectrum is now spread out as shown on top page 329, because spins see a different field B(x). The spectrum is thus a measure of the spin density d(x).

Obviously, you need more to do a 2D picture, but here we get a simple method that shows the basic idea. Page 330 shows our trial 2D object and under that we have a pulse sequence. The key point is that during the first half t1 you activate a x-gradient, and during the second half you activate in this case a z-gradient. You then do lots of pulses varying both t1 and t2 to get data for a 2D FT. The time signal looks like Eq 11.5 where each Larmor "peak" is a function of its distance, x for the first, z for the second. So on the 2D Fourier spectrum picture, the point (x,y)'s density shows up as shown top page 333. And all points do this, so you end up with a complete 2D density image d(x,z). This would apply to a 2D object (with variable 2D density, though that feature is not emphasized here) located in the x-z plane. [Maybe the MRI noise I hear is these gradient magnets turning on and off very fast.]

Suppose you were to add a third pulse segment with a y gradient. Then each 3D point in the object would show up as a point in your 3D Fourier Transform. But they don't do it this way I know, but in theory they could. Probably not easy to make large uniform gradient fields. At some point I have to read how the MRI machines really work! We have just seen a "spin density" example.

So this is a climactic point so far in this book. I give the author A+ for quality work! And I think I will put this subject on hold just for a little while (which may end up being next year). This is a very good resting place, and I am about half way through the book.