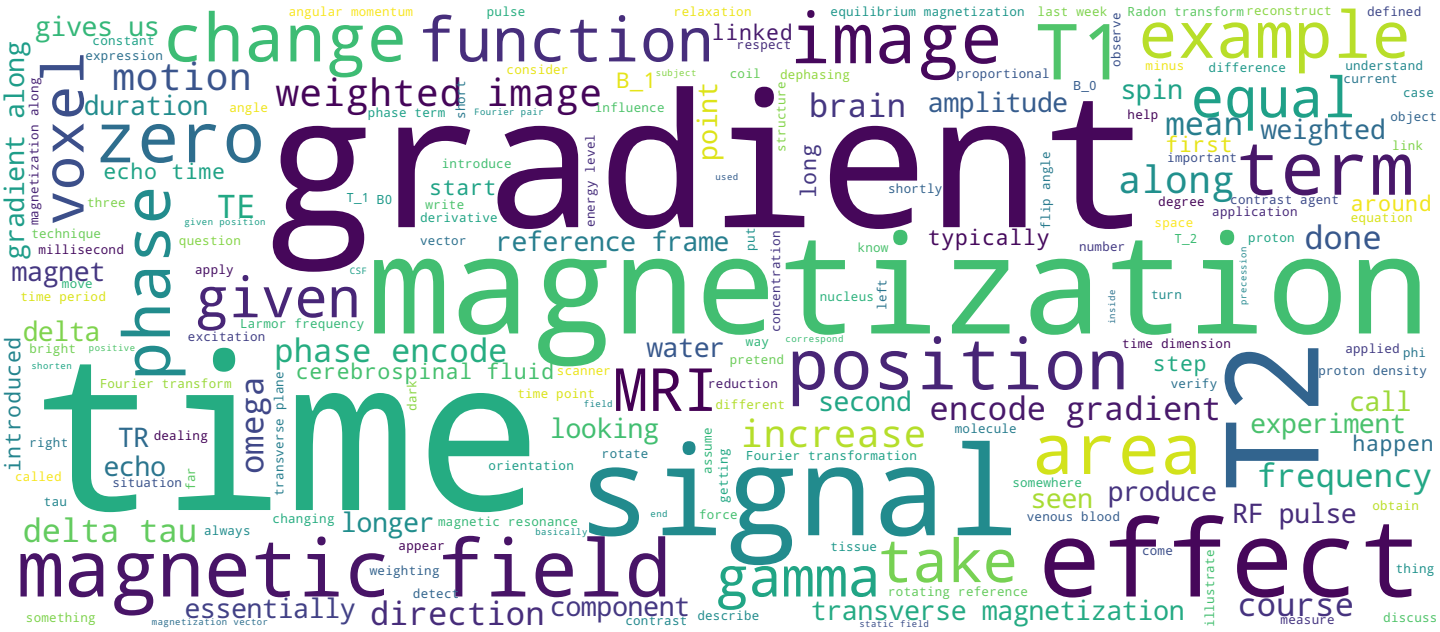
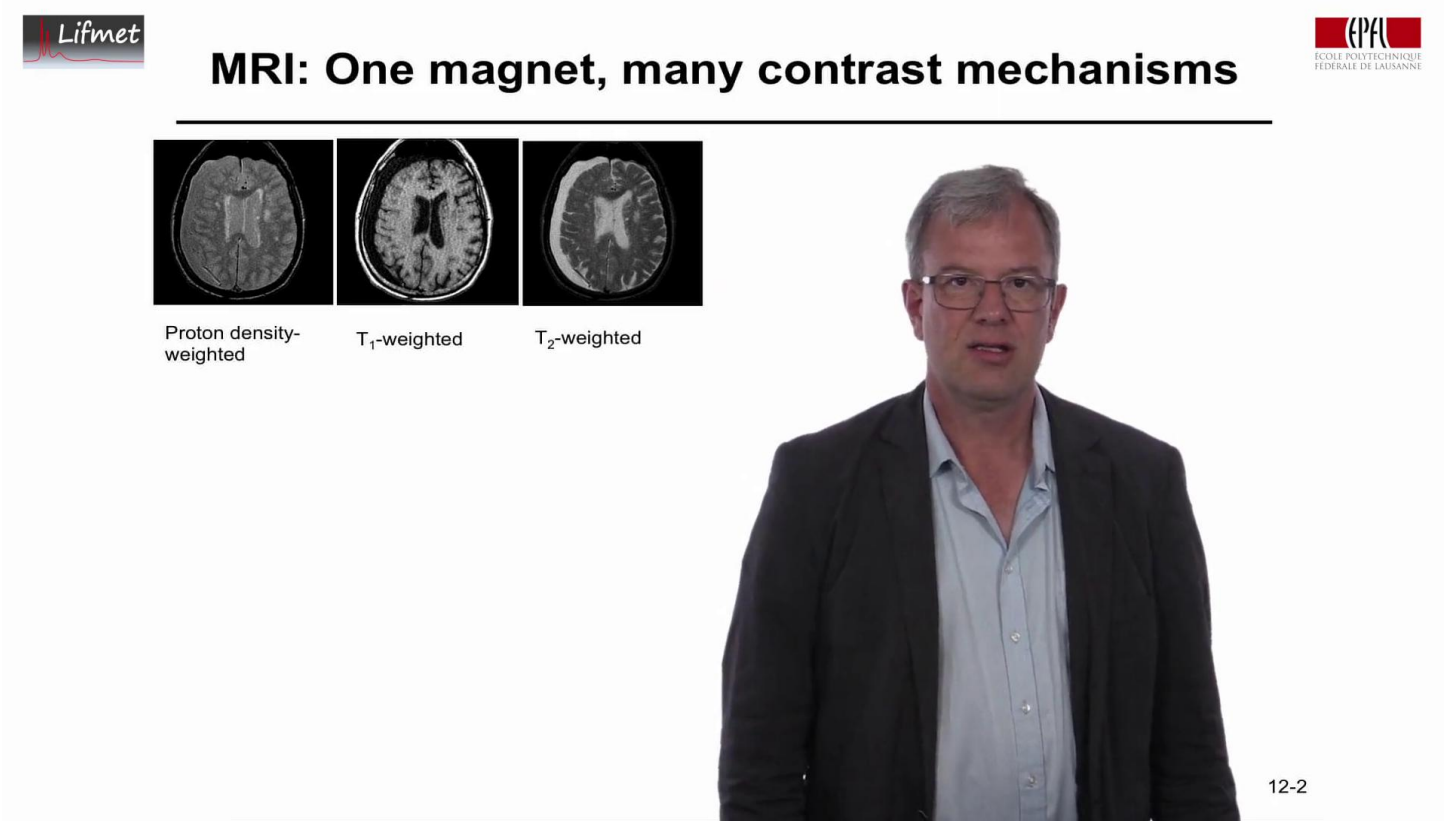
12.1 Gradient echo contrast (T2*)

Fundamentals of Biomedical Imaging

Prof. Rolf Gruetter





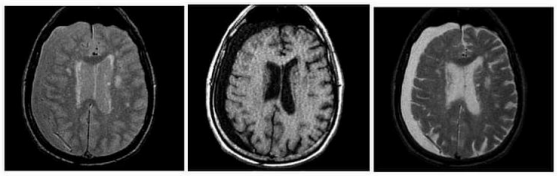
This week's lectures are dealing with the basic contrast mechanisms of MRI and so I want to first, however, start out with a very important message that is dear to me and this is: we are dealing here with one physical phenomenon and that is magnetic resonance. We're dealing with one instrument that is one magnet but what we are getting out of this modality, out of this biophysical mechanism is many different contrasts. So we have many different contrast mechanisms. So I want to first start out with some examples. This is a so called proton density weighted image. So we have a cut through the head here. This is from the brain and if it's proton density then essentially what has been imaged here is the equilibrium magnetizations, so that reflects the density of the protons in the MRI. Then here's an example of a *T1* weighted image. Again a similar cut from the subject. It's a transversal cut, *T1* weighted, so, and here is a *T2* weighted image. So these are some very basic contrasts, proton density *T1* weighted and *T2* weighted and you can see that just by changing some experimental parameters, one can substantially change the appearance of the image.

Notes

Summary

0m 04s

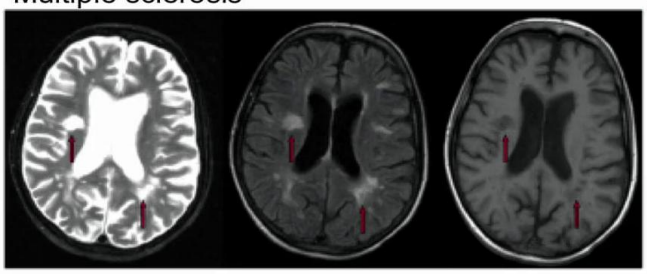
MRI: One magnet, many contrast mechanisms

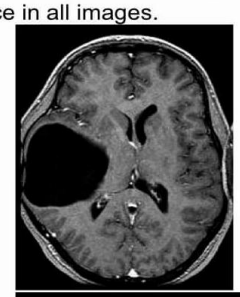


Proton density-weighted | T_1-weighted | T_2-weighted

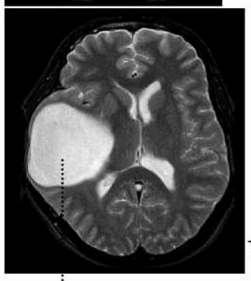
Examples of proton density, T_1, and T_2-weighted images, from the Whole Brain Atlas site at Harvard. Note fluid appearance in all images.

Multiple sclerosis





T_1-weighted



T_2-weighted

Large cyst
(just cerebrospinal fluid)

12-2

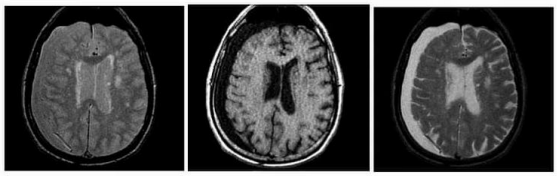
So here, what is interesting to note here is the cerebrospinal fluid, the ventricles, they are filled with cerebrospinal fluid. Here they are more intense than the others because they have a bit more water In the *T1* weighted image they're dark because they have a long *T1* and in *T2* weighted image they are bright because they have a long *T2*. Here's an example which is more pathological so this is clearly not a normal part of the brain. This is *T1* weighted and here it's *T2* weighted. Now what is this space here? It's dark on *T1* weighted like the cerebrospinal fluid. It's bright on a *T2* weighted like cerebrospinal fluid. So this is a cavity where tissue has been eliminated. in fact it is a large cyst. It's just cerebrospinal fluid. Probably a resected tumor or some complication early in life. Now here's another example, it's a bit more elaborate that I want to illustrate here and this is imaging multiple sclerosis, so we're looking at the plaques. So we're looking here at the *T2* weighted image on the left and we see that cerebrospinal fluid is bright, it's here in the ventricles, and it's also in the cell site.

Notes

Summary

1m 35s

MRI: One magnet, many contrast mechanisms



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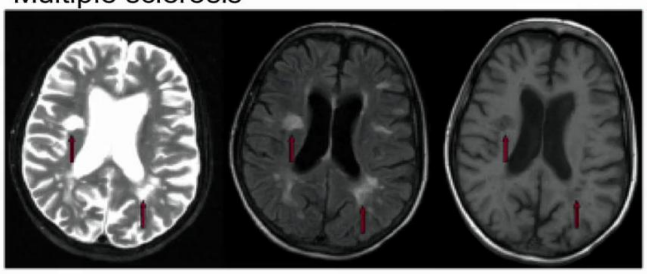
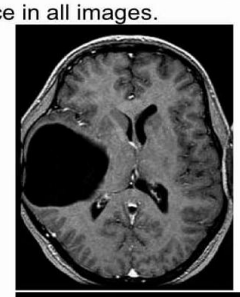
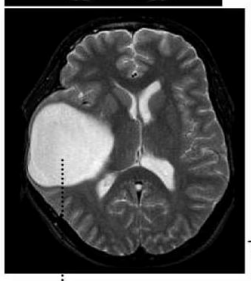


T_2-weighted [$TE=T_2(CSF)$]

FLAIR: T_2 and T_1 weighted (inversion recovery CSF-nulled) [$TI=\ln 2T_1(CSF)$]



T_1-weighted



T_2-weighted

Large cyst (just cerebrospinal fluid)

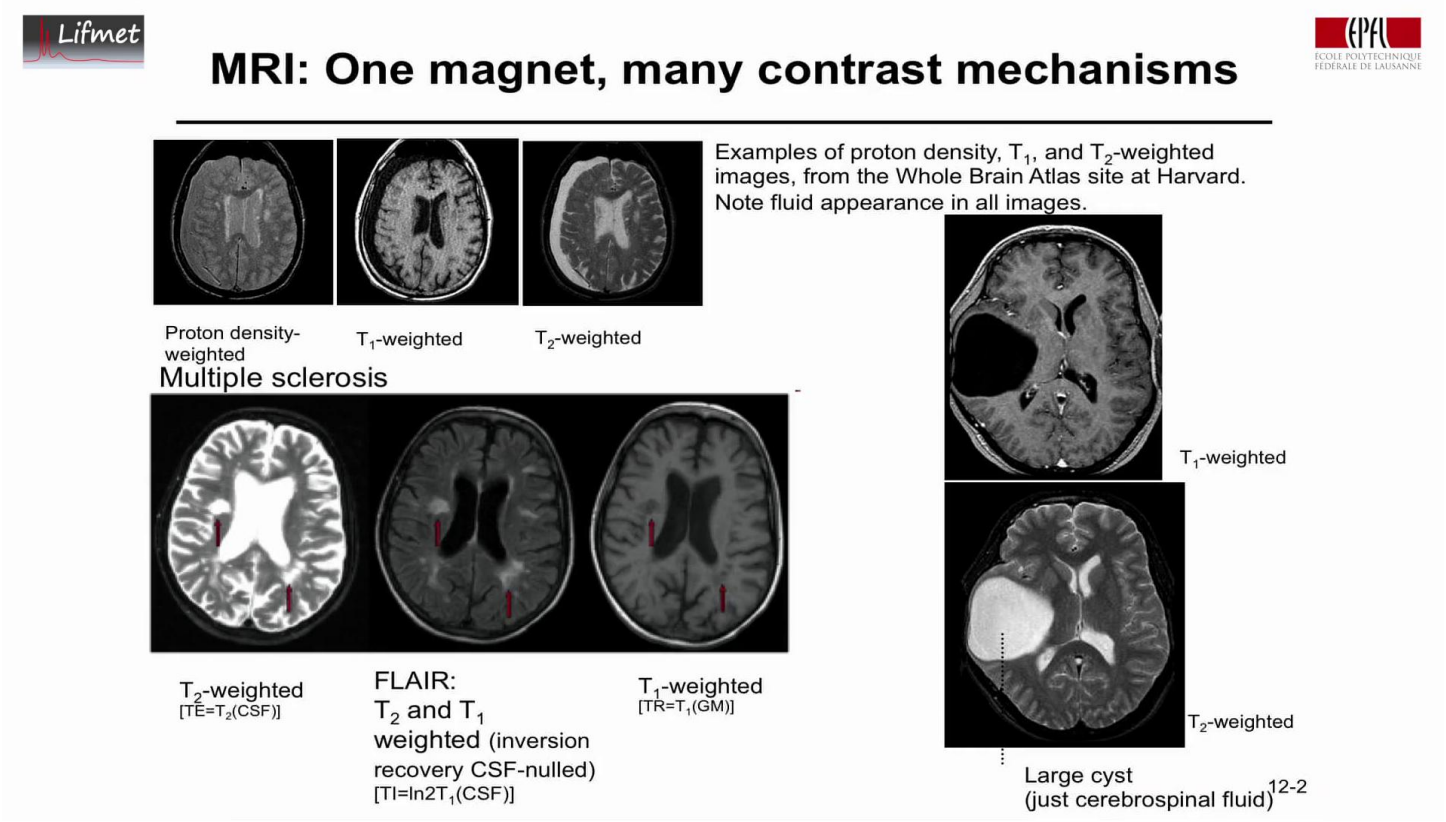
12-2

And if one looks very carefully among all these bright signals, one sees some signals here that are somewhere where they shouldn't be. But this is, with this-- all this bright signal is difficult to detect. So then the approach that is often used in multiple sclerosis is to use the so-called fluid attenuated inversion recovery approach. So or FLAIR. That's the acronym for that and this is a combination of *T1* and *T2* weighted. First what is done with an inversion pulse and an inversion time *TI* that is such that the signal from the CSF, the magnetization, the longitudinal magnetization from the CSF is zero. At that point when the longitudinal magnetization in the cerebrospinal fluid is zero, the signal is read out and then detected with a long echo time so it is *T2* weighted. What then happens is that these lesions here in the brain tissue the signal changes appreciably compared to the brain tissue and there's substantially more contrast to noise ratio. Here the signal to noise ratio is substantially better but here the contrast ratio that's distinguishes these demyelinating legions from the normal appearing tissues around them is much easier detectable.

Notes

Summary

2m 54s



And finally, we have the *T1* weighted image. On the T1 weighted image because demyelinating legions are destruction of tissue, so there's less tissue, less membranes, less proteins around, *T1* and *T2* are getting longer, *T1* is longer, they appear as dark and then again not as visible as they are here on the middle image. So we have here just the examples *T1*, *T2*, and proton density that give us already three basic contrasts in MRI.

Notes

Summary

4m 18s

T_2*-Weighted Images

Venography
Designed to make venous blood (rich in deoxy-hemoglobin) darker than normal tissue (=reduce magnetization in blood)

BOLD fMRI=functional MRI

Normoxia

12-3

Another contrast I want to discuss this week is *T2** weighted. We're actually going to discuss this contrast first. With *T2** weighting, what is often done is venography. And the goal here is to use experimental conditions that at which point the magnetization that's in the venous blood has decayed so venous blood, the signal from venous blood is darker than normal tissue. So it's a reduction in magnetization in blood. What is a venogram? What does it look like? Here is from the brain. So this is a *T2** weighted image and all these black stripes here, that are in here, they're in fact normal manifestation of venous blood that has a short *T2**. That is deoxygenated and we'll see shortly. We'll discuss this shortly. In research there has been functional MRI has been coming along. It's based on blood oxygen level dependent functional MRI or BOLD fMRI and what I want to highlight here is an example of the very first images, or some of the very first images that Seiji Ogawa at Bell Labs obtained at the time in the late 1980's from the rat's brain. So this is a rat brain under normal conditions. This is what such a *T2** weighted image looks like.

Notes

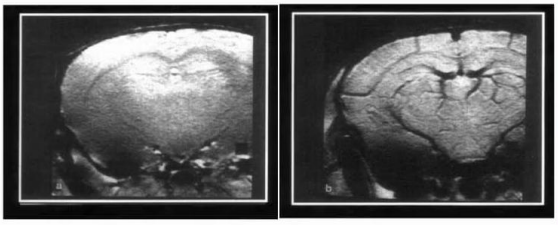
Summary

4m 49s

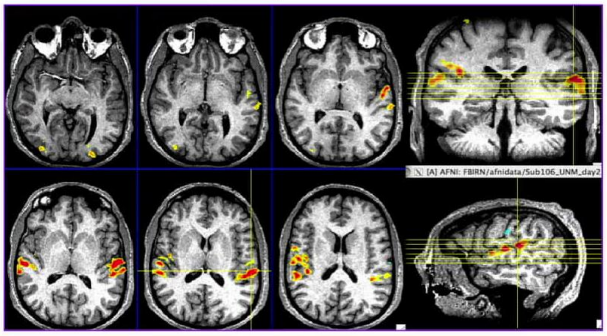
T_2*-Weighted Images

Venography
Designed to make venous blood (rich in deoxy-hemoglobin) darker than normal tissue (=reduce magnetization in blood)

BOLD fMRI=functional MRI



Normoxia Hypoxia



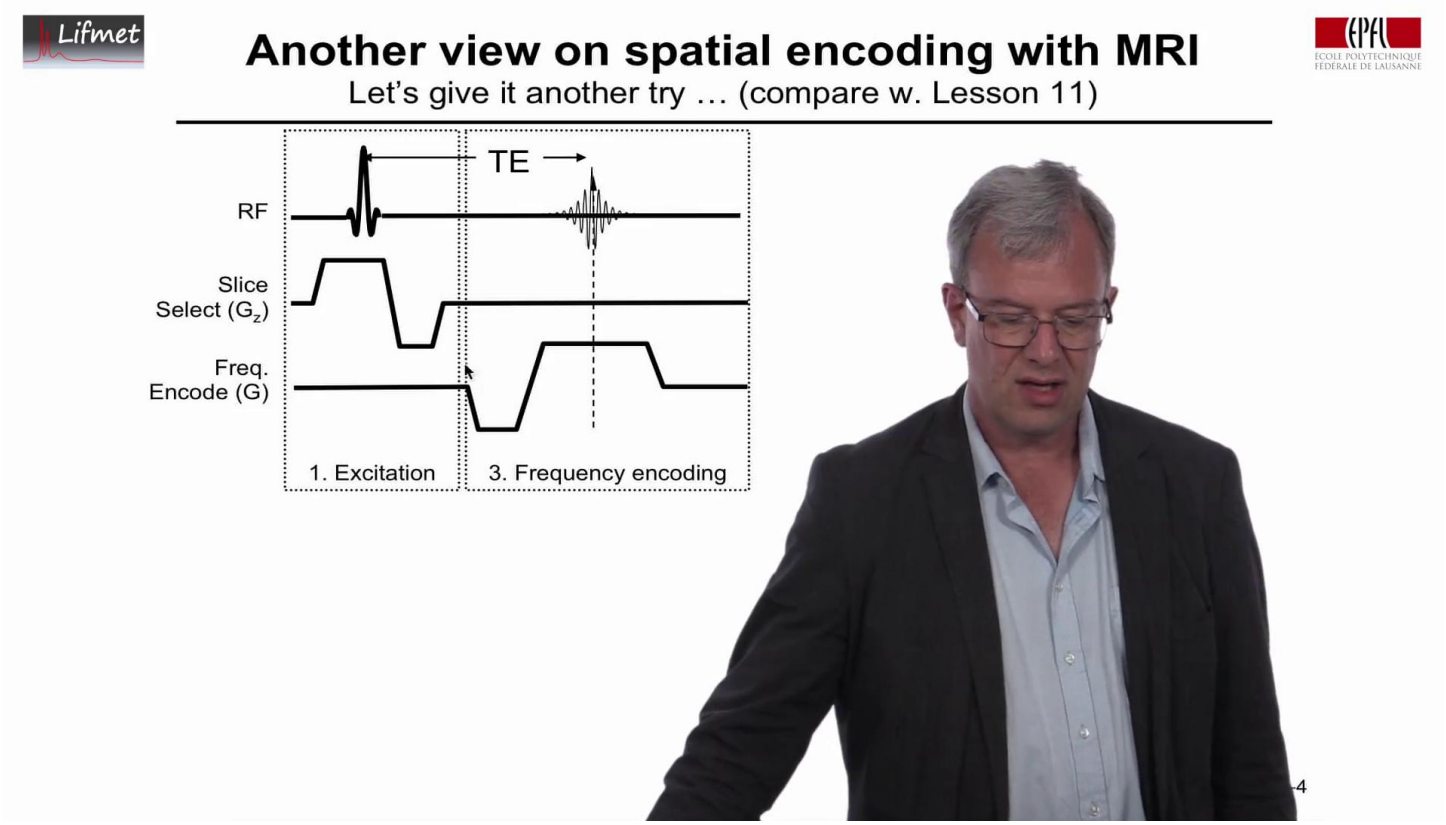
12-3

And one day he noticed that in one of the animals which was having trouble breathing, he obtained an image that looked like this. There were the venous structures that were dark appearing and general the sensitivity of the signal had decreased. And he had the enormous insight to realize that from neuroscience we knew that there is an increase in cerebral blood flow much more than oxygen extraction. So this would lead to a change in deoxygenation in the blood vessels and he proposed at the time that this is possible to exploit in normal brain activation studies to see the activity of the brain and hence in the early 90's the functional MRI was born. Nowadays the datasets used for fMRI for brain activation studies depict datasets like this, so we have the anatomical image. Here are the eyeballs, retrobulbar fat. this is axial cut, this is a coronal cut, these are sagittal cuts and overlaid on the anatomical images are the area of the brain that have been activated in response to an external stimulus. That stimulus can be finger tapping, it can be looking at an image, it can be listening to a sound, it can be a mental task, and so forth.

Notes

Summary

6m 24s

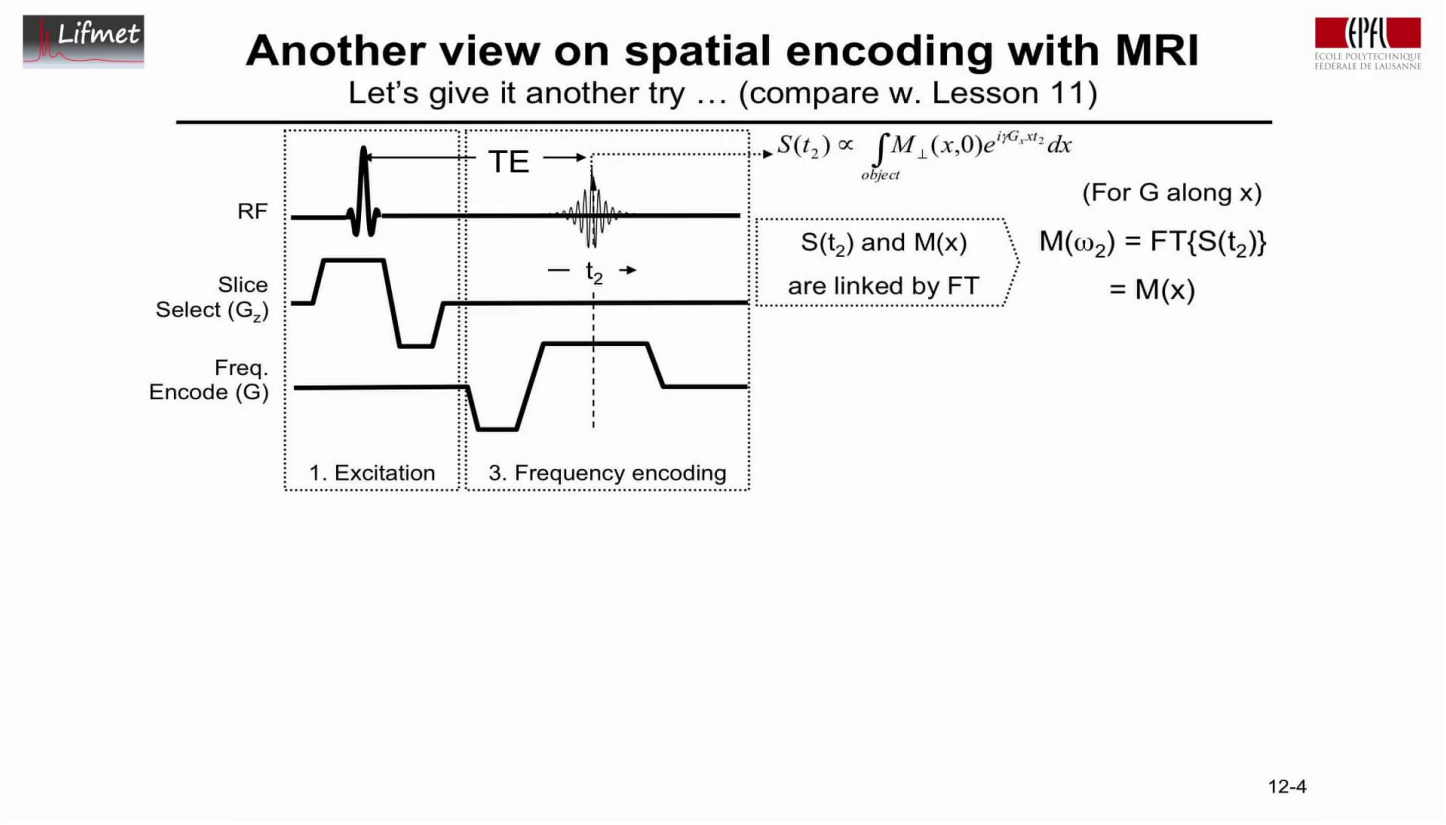


Now I want to give here another view on spatial coding with MRI. Last week we have introduced a two dimensional spatial encoding with MRI and I'm sure you have found it not the easiest topic to understand. So I want to give it here another try and give you another argument to rationalize how we reconstruct images with MRI. So let's start with the basic experiment. We need to have an *RF pulse* to flip the magnetization into transfer plane so we have something that precesses that we can detect. This is done typically with a slice select gradient so we're selecting a slice like this in the brain and then we have the frequency and encode gradient that is switched. The first part of the experiment is excitation and then we have frequency and coding, so two phases of the experiment. We observe here the echo time at this time point. The echo, we observe the echo at this time point and this is defined by the echo time. We have seen that we need this negative gradient refocusing gradient to have maximal sensitivity to minimize the effects of dephasing under slice select gradient. Here is the frequency encoding dephase and rephase and when these two areas are the same we have echo.

Notes

Summary

7m 49s



So now I'll introduce a time and I'll call it *T2*. You'll see shortly why I call it *T2* and not just simply, *T2*. So as we record the signal with this scanner, we'll call the time during which this happens, *T2*. And now we're going to look at the signal as a function of *T2* starting at this time point. So this is time zero here. This is equal to the transverse magnetization as a function of position *x*. We have the gradient here, we'll assume along *x*. So gradient will assume the frequency codes along *x*. So magnetization starting here, they're all aligned, they're collinear. And then we have the different precessional frequency based on the deposition along *x*. So it's gamma *G x* times x times *T2* is the face as it evolves. And of course the entire [scene] we see is the summation of all the signals along *x*. So now the link between the signal as a function of *T2* the time domain signal and the magnetization as a function of position, the link between the two is given by the Fourier transformation. And we take a standard Fourier pair, time and frequency. So the Fourier transform of a signal and time gives you a function as a function of frequency.

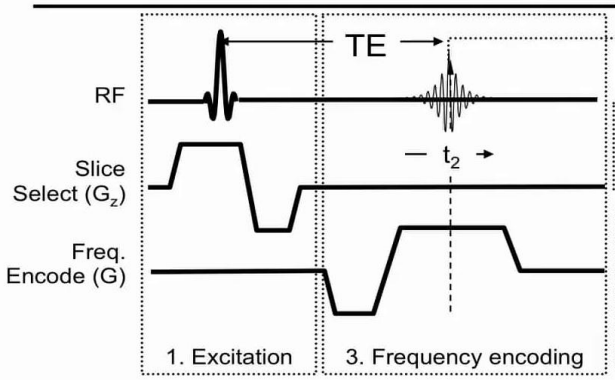
Notes

Summary

9m 10s

Another view on spatial encoding with MRI

Let's give it another try … (compare w. Lesson 11)



$$S(t_2) \propto \int_{object} M_\perp(x,0) e^{i\gamma G_x x t_2} dx$$

(For G along x)

$S(t_2)$ and M(x) are linked by FT

$M(\omega_2) = FT\{S(t_2)\}$

$= M(x)$

M(x) is Radon transform measured along the direction of the Gradient **G**

$$M(x) = \int M(x,y) dy$$

12-4

That's a standard Fourier pair, time frequency. So if we have *T2* as our time variable, then *omega 2* is our -- we'll call *omega 2* the corresponding Fourier pair variable. Now *omega 2* is the *gamma Gx* times *x*, that is linked to the position because the gradient is fixed. The *gamma* is fixed so this is *omega 2* is linearly related to the position. So the magnetization is a function of this frequency is equal to the magnetization as a function of position. Okay so actually if we look at a two-dimensional object at a given position in *x*, all the magnetizations that have the same coordinate along *x*, for this gradient along *y*, they'll have the same precessional frequency so we cannot distinguish them. They'll just appear as a stronger signal and this is reflected that we have to sum along *y*, all the magnetizations along *y*, for a given position *x*. That's our magnetization along *x*. And so it turns out that after Fourier transform of our echo here of this signal we're getting *M* of *x* and *M* of *x* is the Radon transform of the magnetization along the direction *x*. Now what we can do is here, this is for a particular direction *x* but I have not defined what *x* is so I can choose any coordinate system and have *x* different.

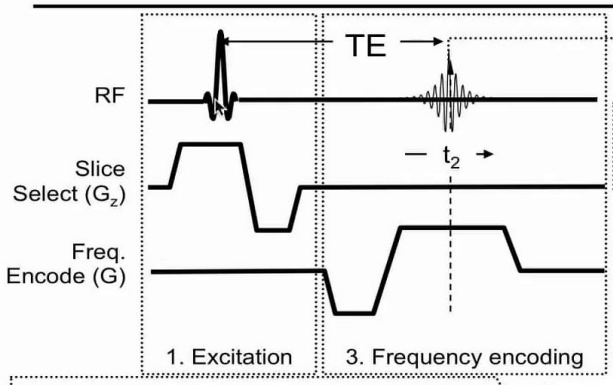
Notes

Summary

10m 29s

Another view on spatial encoding with MRI

Let's give it another try … (compare w. Lesson 11)



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$\mathbf{G}=(G_x,G_y)=\cos\phi G, \sin\phi G$

Measure Radon Transforms along ϕ (as in CT) every TR s (T_1 relaxation) (→ sinogram, projection reconstruction, see also central slice theorem)

12-4

And so we can actually change the orientation of this projection, the magnetization of *x*, by now changing the orientation of the gradient, because the direction of this Radon transform is entirely defined by the gradient orientation. So if I now choose a gradient, and remember we have introduced gradient as vector as a linear combination of-- as with two components basically with a constant amplitude but I will rotate it by an angle *phi*. What this amounts to is, that we are recording the projection, the echo, along an orientation that is rotated with respect to *x* by the angle *phi*. You can verify that, if *phi* is zero, then we have a gradient along *x*. If *phi* is 90 degrees, then we have a gradient along *y*. And of course we can be changed between the two gradually by changing *phi* gradually. And so this we can repeat, we can repeat different projections. We'll collect after Fourier transformation the Radon transforms of our magnetization. We'll do this in many steps just like with CT. We'll have to take into account that we have to every time to measure new Radon transform. We'll have to do a new excitation.

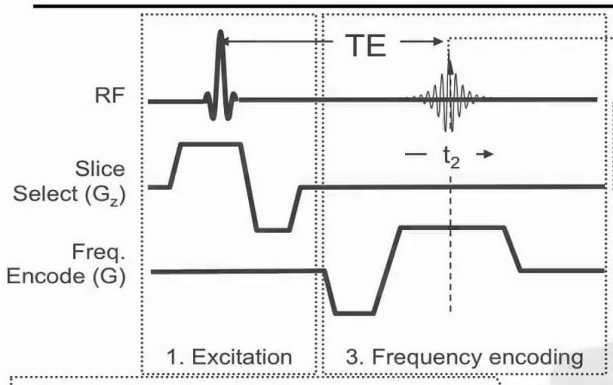
Notes

Summary

11m 59s

Another view on spatial encoding with MRI

Let's give it another try … (compare w. Lesson 11)



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Measure Radon Transforms along ϕ (as in CT) every TR s (T_1 relaxation) (→ sinogram, projection reconstruction, see also central slice theorem)

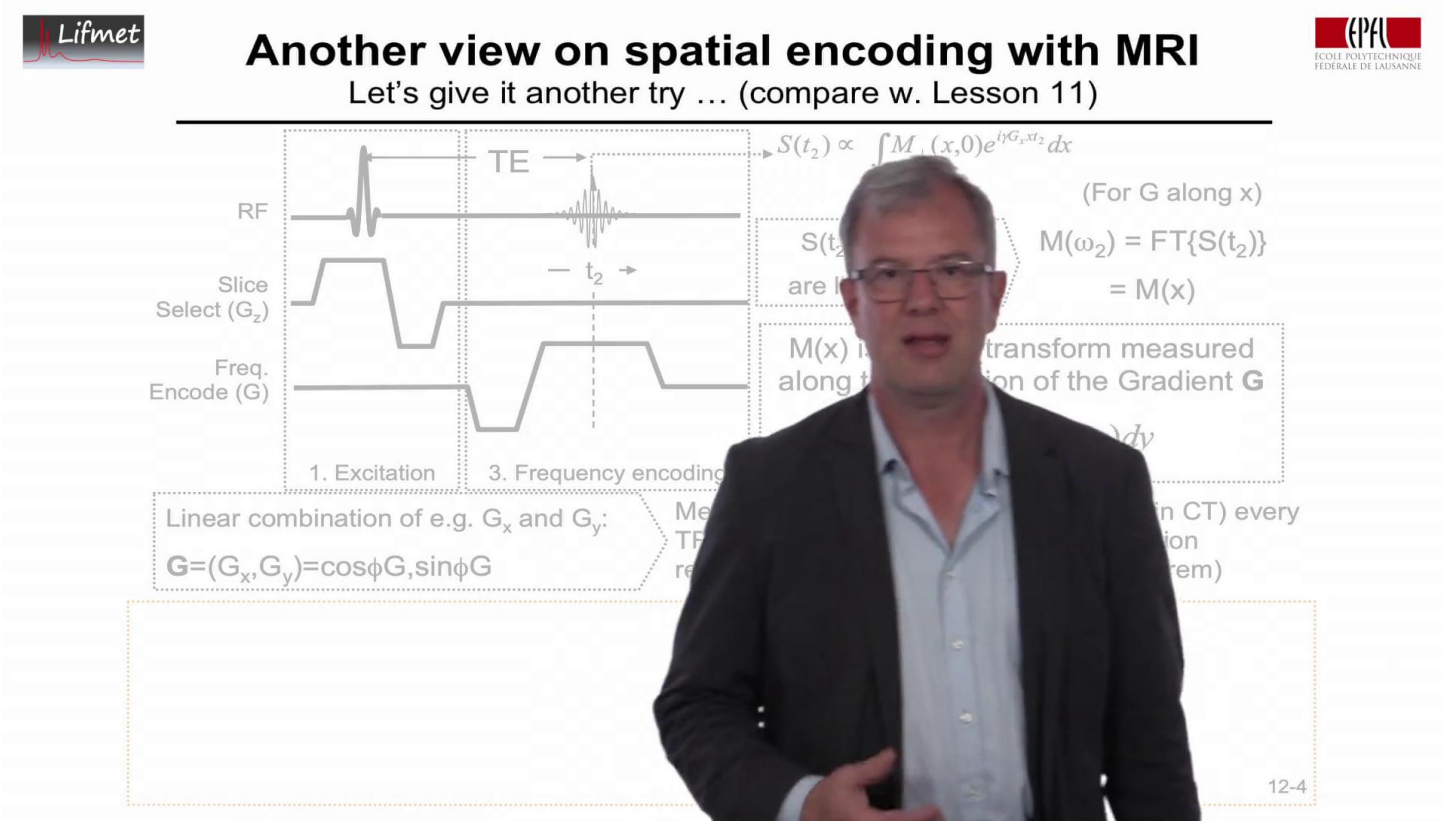
12-4

So this will be done every *TR* seconds based on the repetition time. This leads to a sinogram. In principle we can reconstruct any image with projection reconstruction or we can make a link to Fourier transformation with a central slice theorem.

Notes

Summary

13m 15s



So if this helps you to understand why MRI scans take some time we are going to have to collect the different projections, the different angles, we can do that just as well. It has been done in the X-ray techniques and then reconstruct the image using those techniques is definitely a possibility. So this could convince you that it indeed, A, it is possible to reconstruct an image, and B, why collecting the data of an image does take considerable time because if we want to have a 256 by 256 matrix that means we have to collect 256 angles *phi* and for this we have to wait 256 *TR* periods. *TR* is linked to *T1* so it is typically on the order of the *T1* relaxation time and therefore that explains why it takes typically a couple of minutes to obtain an MRI image. Okay so if this helps you, but now I'm going to try to look at the phase encode gradient because the MRI reconstruction and data requisition is rarely done using the Radon transform approach. There are some applications but they are the vast majority. The major application, the major way of obtaining the image is done by two-dimensional encoding using the phase encode gradient.

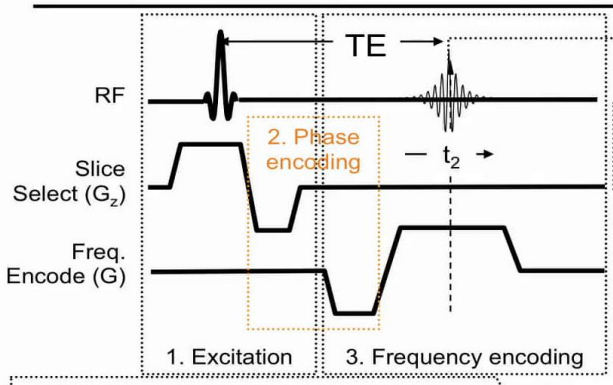
Notes

Summary

13m 34s

Another view on spatial encoding with MRI

Let's give it another try … (compare w. Lesson 11)



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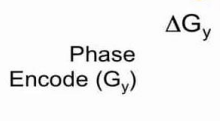
Phase encoding is just frequency encoding in a 2nd time dimension



$$M_\perp(\tau)\big|_{n^{th}\ step} = M_\perp(0) e^{i\gamma n \Delta G_y \tau}$$

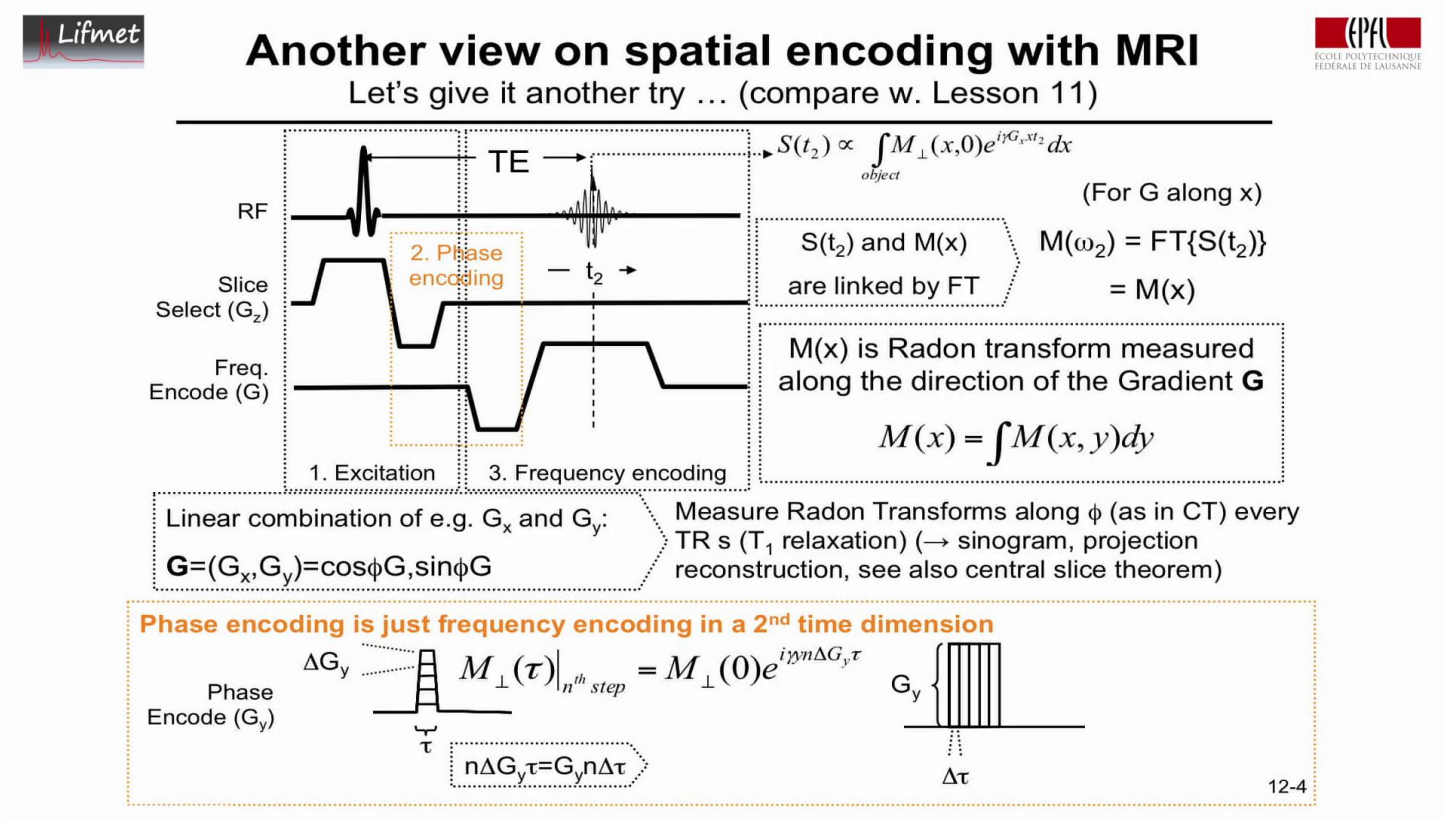
12-4

Now what I want to make here the argument is phase encoding is essentially just frequency encoding in a second time dimension. So we'll just put in here a phase encode gradient. We're looking here at this time period. It's the third-dimension and now we'll look at here at the *y* gradient we've had last week we have talked about phase encode gradient having a certain amplitude, and a certain duration *tau.* And if we have now an *Nth* step then the amplitude of the gradient is *N* times *delta gamma y,* that's the gradient strength times *y* times *gamma* times the duration that gives us the phase twist induced by this gradient. So for the *Nth* phase encode state the transverse magnetization after this phase encode gradient so right here is given by the magnetization before so that's the transverse magnetization here times this phase term here. And what we have done is we've argued that if we now step through these amplitudes with a decrement or an increment of *delta Gy,* we do this several times. That this essentially allows us to encode the second spatial dimension.

Notes

Summary

15m 02s

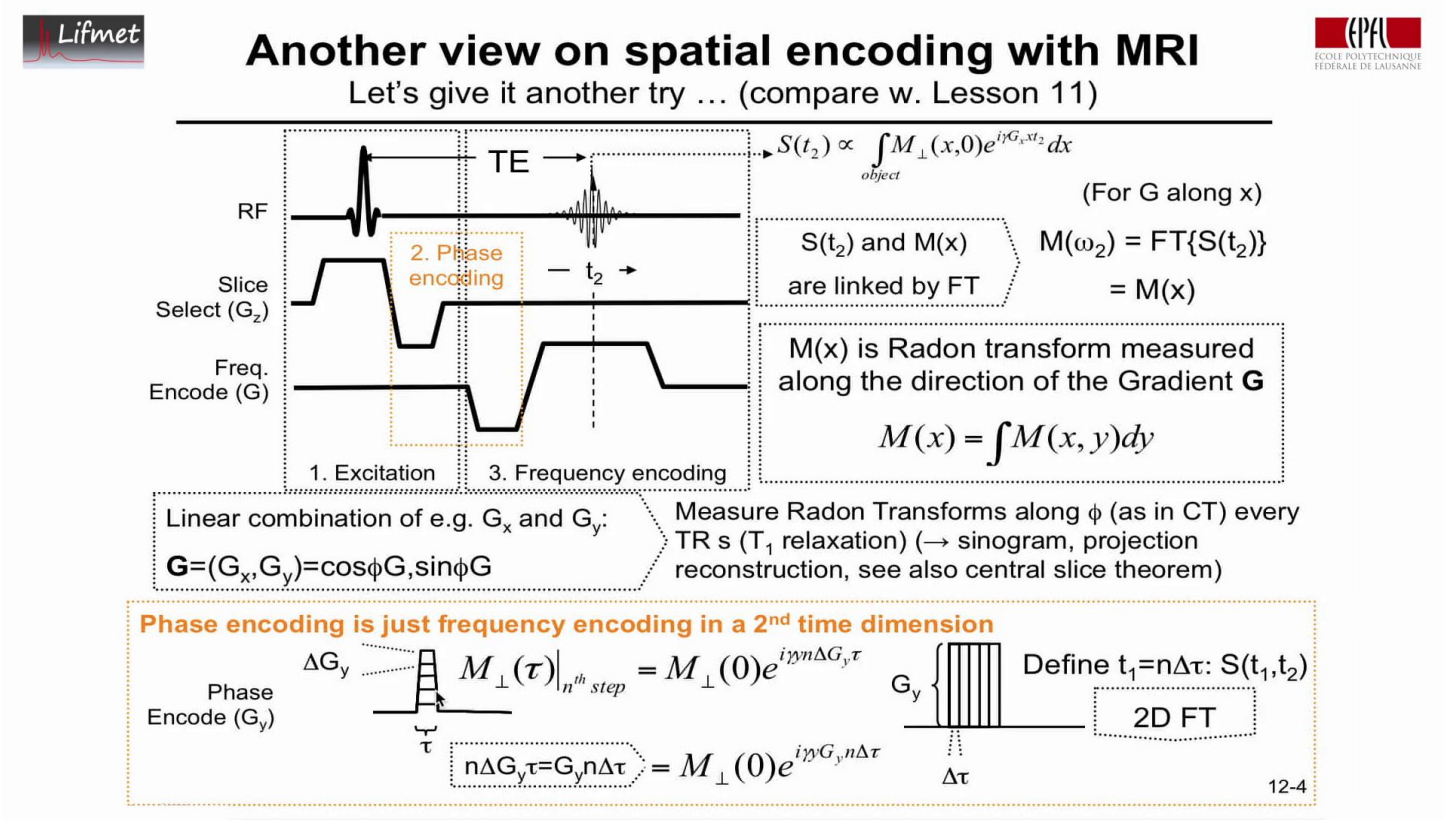


The argument there was that I made it was graphical to say, well in the end the computer cannot distinguish whether we have collected one-dimension immediately and the other one step by step or if we do it in reverse and therefore these two things are equivalent. Now what I want to argue here is that we can actually look-- if you look here at the gradient times time term, the *n delta Gy tau*, I can rewrite this and I'll say, well *n* times *delta Gy tau,* I'll express this as a constant gradient times *n* times *delta tau.* So now I have a constant gradient along *y* and I have this term *n delta tau.* What does this mean? So for the *Nth* gradient step we'll start out with the first gradient step we'll have an amplitude, *Gy*, this is the gradient along *y,* and we'll have a duration of *delta tau.* If we do now the second step, then we have a duration of two *delta tau,* and three *delta tau,* four *delta tau,* and so on. Five *delta tau* here. So it's up to *n delta tau.* And now we'll rewrite the magnetization the effect of this gradient again, gradient times time gives you the phase effect.

Notes

Summary

16m 17s

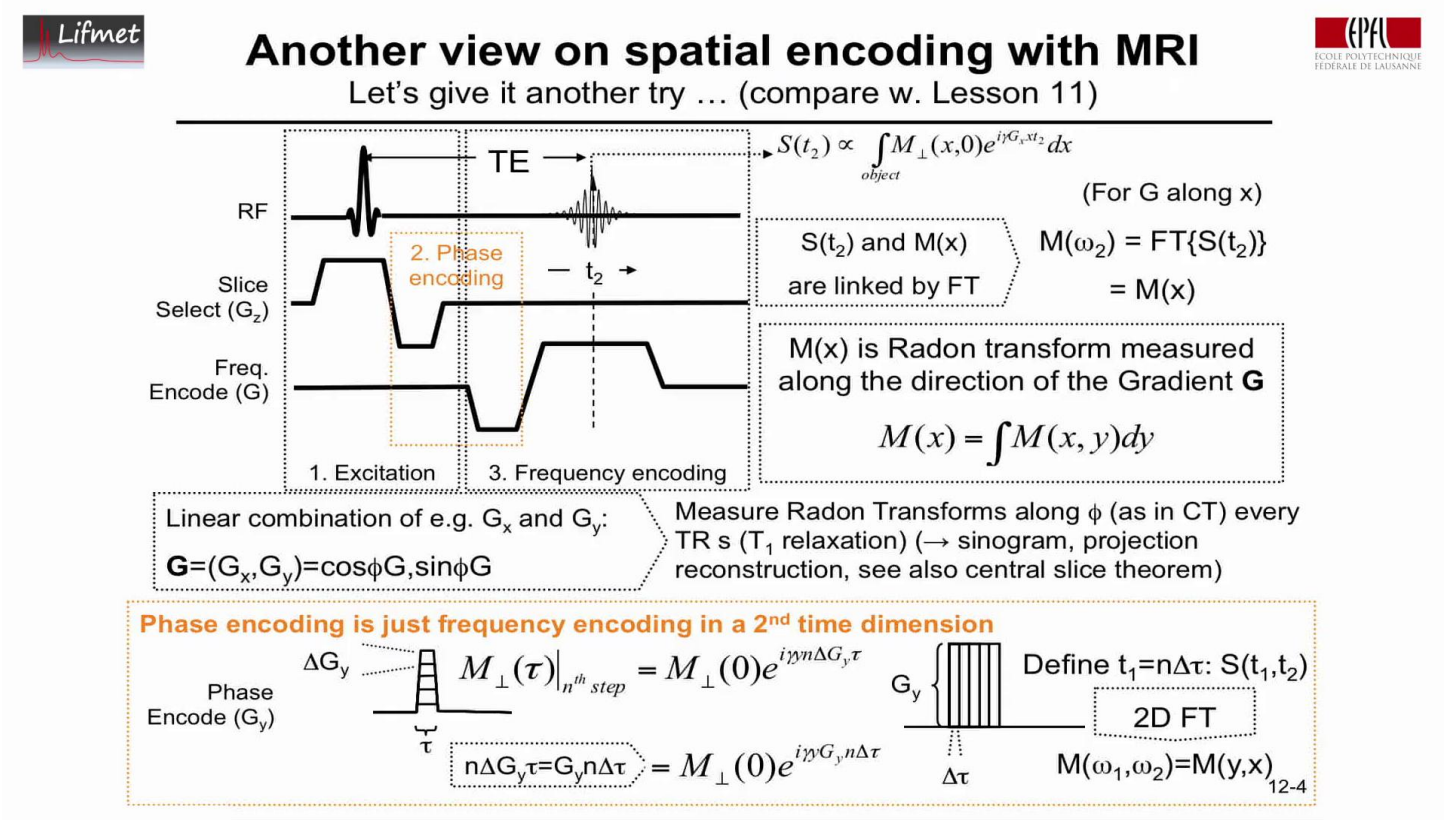


So gradient times time *n delta tau* times *y* times *gamma* is the phase term. We're just using this identity here so we're getting the magnetization now expressed in this form, the evolution of magnetization on to the gradient. And now I can say well *n delta tau* that's just nothing but a time dimension which I call *T1* that has been discretely sampled at intervals *delta tau.* So in the end effect what we are now getting is a signal that depends on two time dimensions. *T2* is the time as we recorded, *T1* has been defined as the time step by step as we increase the gradient duration for this example but we have seen that this is equivalent to increasing the gradient amplitude step by step. So if we consider the signal is now a two dimensional time signal, there are of course not two dimensional times as such but it's a second virtual time dimension, we can do the frequency, the Fourier transformation, two dimensional Fourier transformation, we got two time variables, and this gives us a-- from the signal of *T1* and *T2,* gives us a magnetization as a function of *omega 1* and *omega 2,* and therefore a magnetization as a function of *y* and *x,* because *omega 1* is linked to the position in *y* here and *omega 2* is linked to the position in *x* that's here.

Notes

Summary

17m 32s

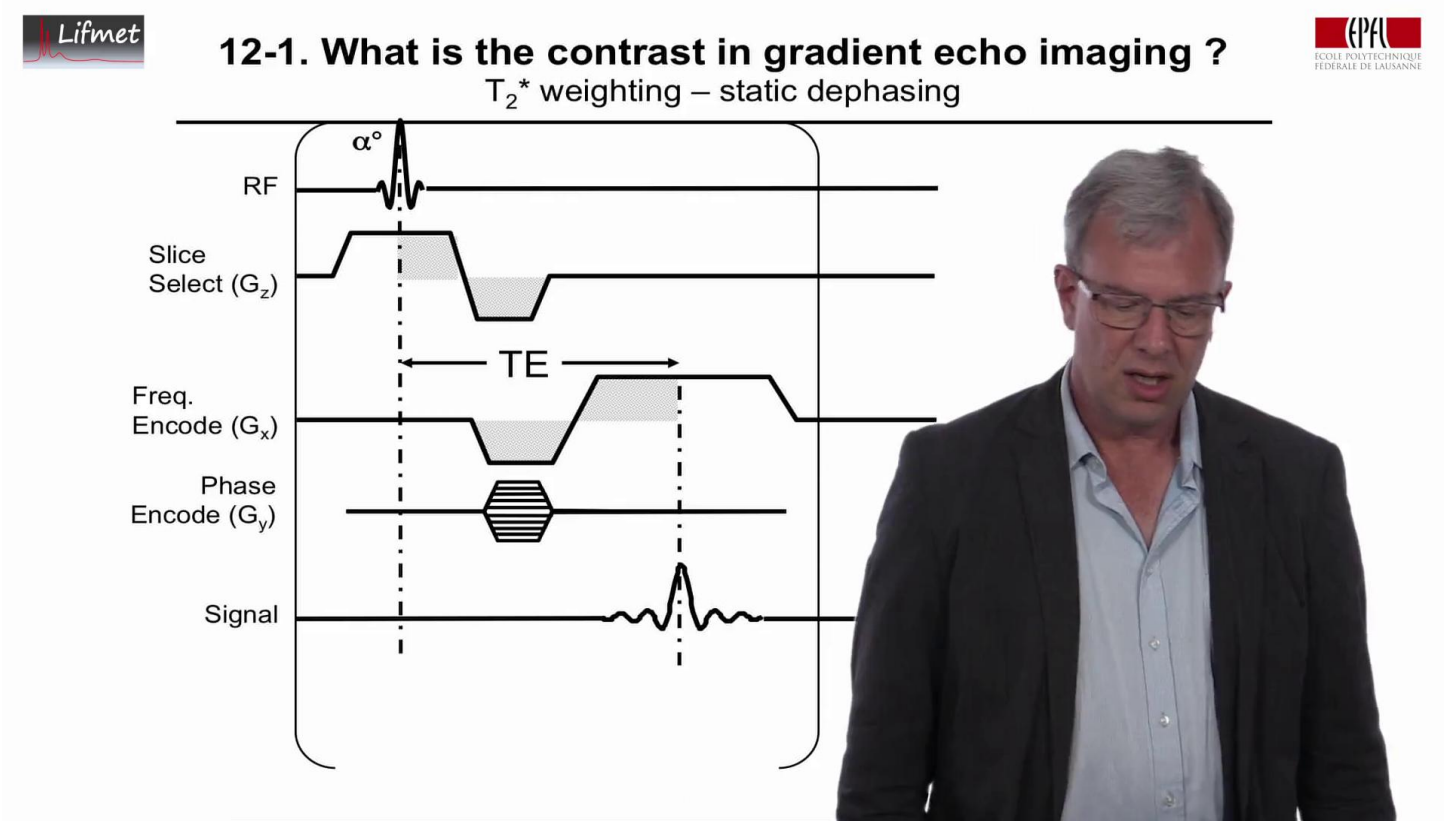


So we have a two dimensional time signal we can view it as that way, it's not standard way done in the MRI but if it helps you this is another way to look at it. Two dimensional Fourier transform gives us two frequencies. A two dimensional object and these frequencies through the Larmor equation are linked to position with the gradient condition. Okay so much to rehash the spatial encoding with MRI, the phase encoding and I hope this helps you with this relatively difficult topic and understanding how we generate 2D images.

Notes

Summary

19m 00s

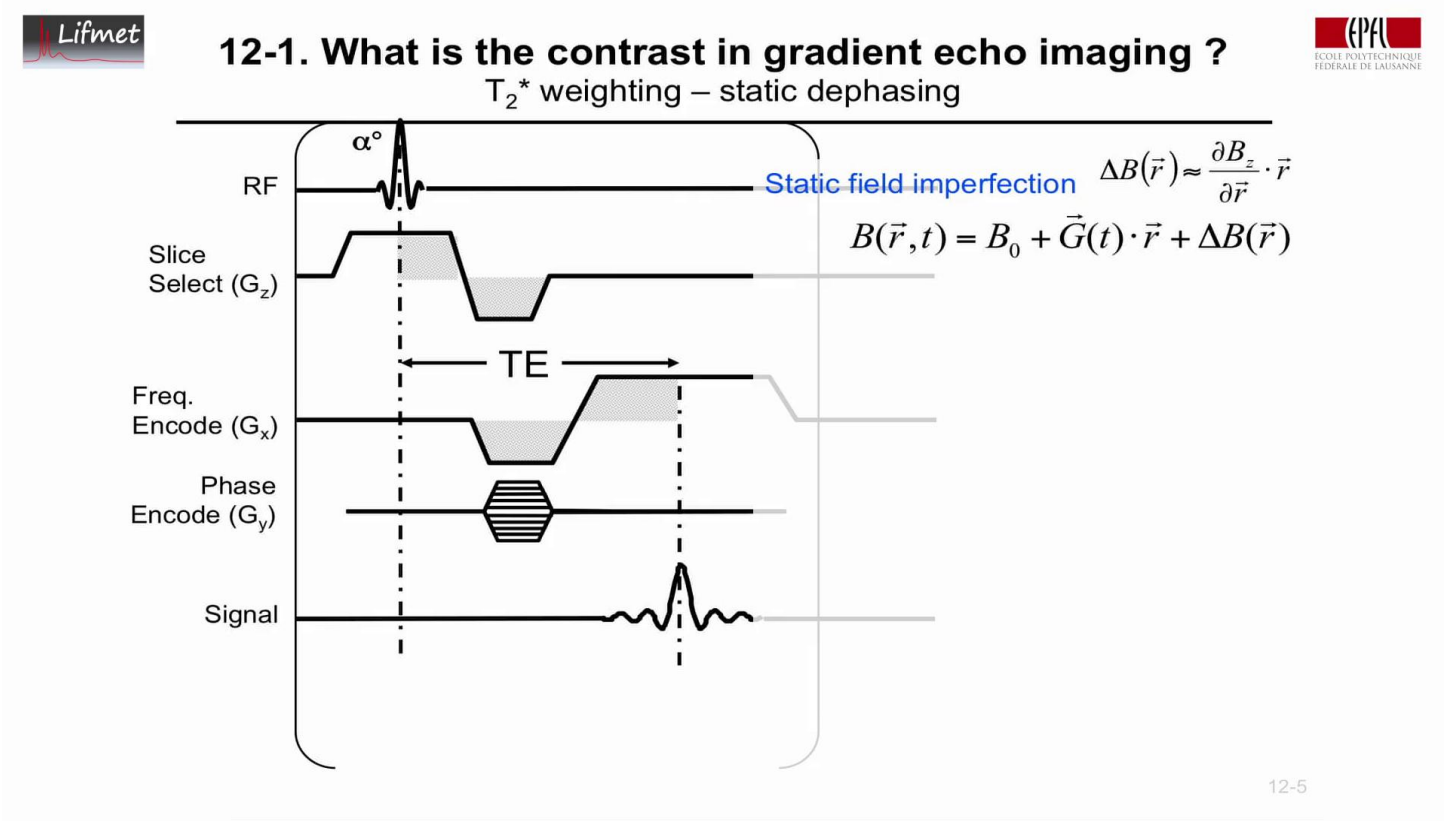


Okay so last week we introduced a gradient echo sequence, this is the basic experiment with which one generates images. We've introduced how the spatial information is encoded. So now the question that I want to go after is: What is the contrast in gradient echo imaging? And here we will introduce *T2** weighting. So we have the *RF pulse* which is at an *alpha* degree. We need this pulse to generate transverse magnetization. It is done with slice selection. So we're selecting a slice here and the areas of these two gradients are equal but opposite. This is refocusing so we have maximum signal. Then we have the frequency encode gradient with dephasing and rephasing. And this area here then defines the rephasing. And we have the phase encode gradient. This gives us the signal that is ultimately recorded. We'll set this time here to zero and here the time is defined by the frequency encode gradient. This is the echo time *TE*. So this is our basic imaging sequence, this is repeated *n* times to generate the image. Now what we want to look at is one effect that produces *T2** weighting and that is to look at the effect of static field imperfection, which we haven't dealt with yet.

Notes

Summary

19m 40s

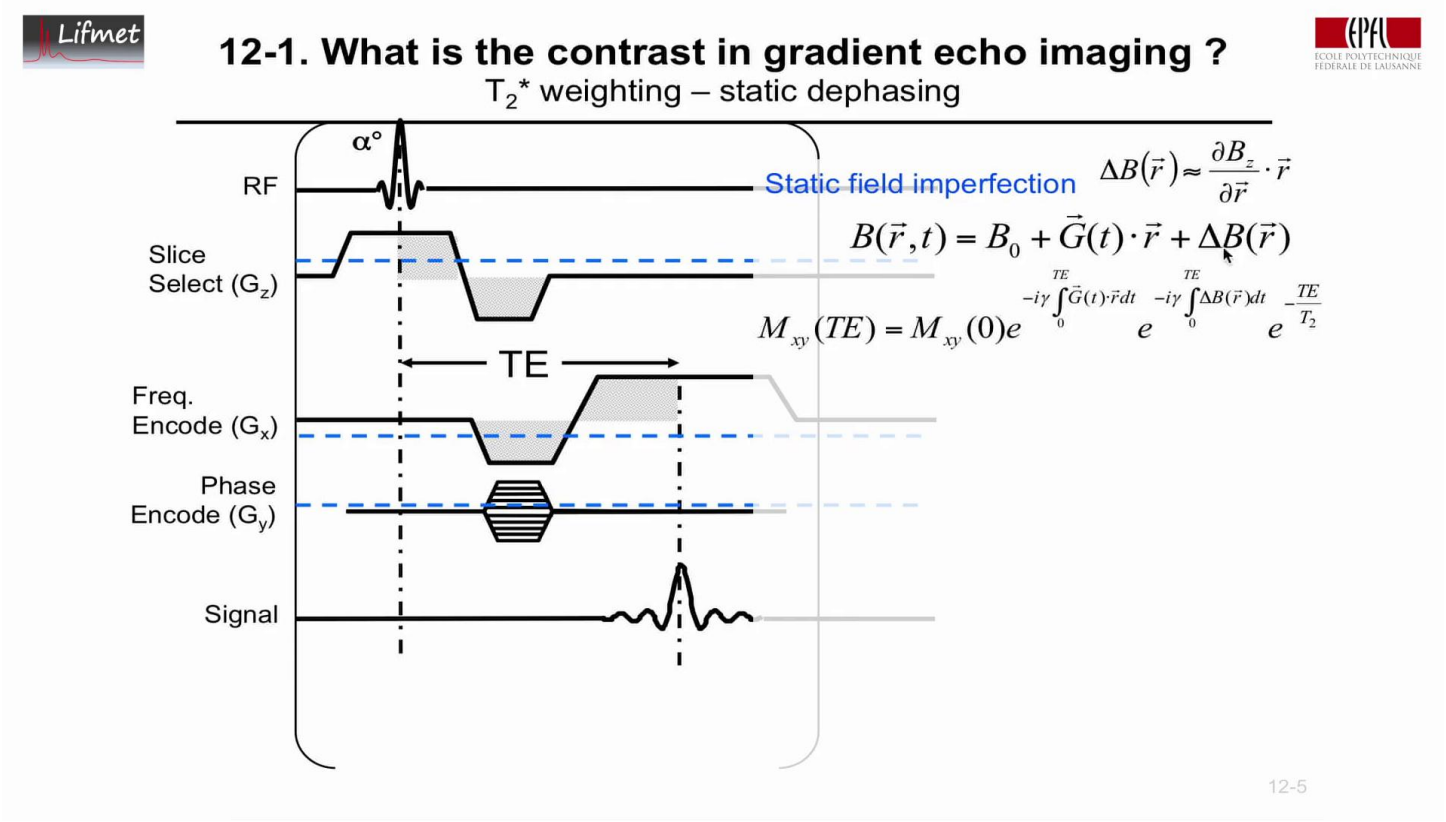


So far I have introduced you to the relaxation times and so far we have dealt with the *B0*, aside from the gradient that we introduce in the scanner, the *B0* is perfectly homogeneous in space and this is unfortunately not true. In very general terms we can write this static magnetic field to be around which the magnetization precesses has to be zero. We have the gradient term, now this can be time variant as we can see here, times the position plus a term *delta B of r.* This *delta B of r* is nothing but the first order approximation of the Taylor series expansion of the magnetic field, if you will. So it is the derivative of the *z* component of the magnetic field, with respect to position times the position scalar product. This is a vector, this is a scalar product. So this produces a changing magnetic field and that is spatially dependent. It is static, it's not time variant and we can assume that we have an image voxel and in that image voxel we don't have a perfect *B0.* So we have a distribution of magnetic fields in that voxel of our image. So what happens now, this basically means that if we have a *dBz* along that, then basically what we've drawn here the gradient as zero, it's actually not truly at zero.

Notes

Summary

21m 05s

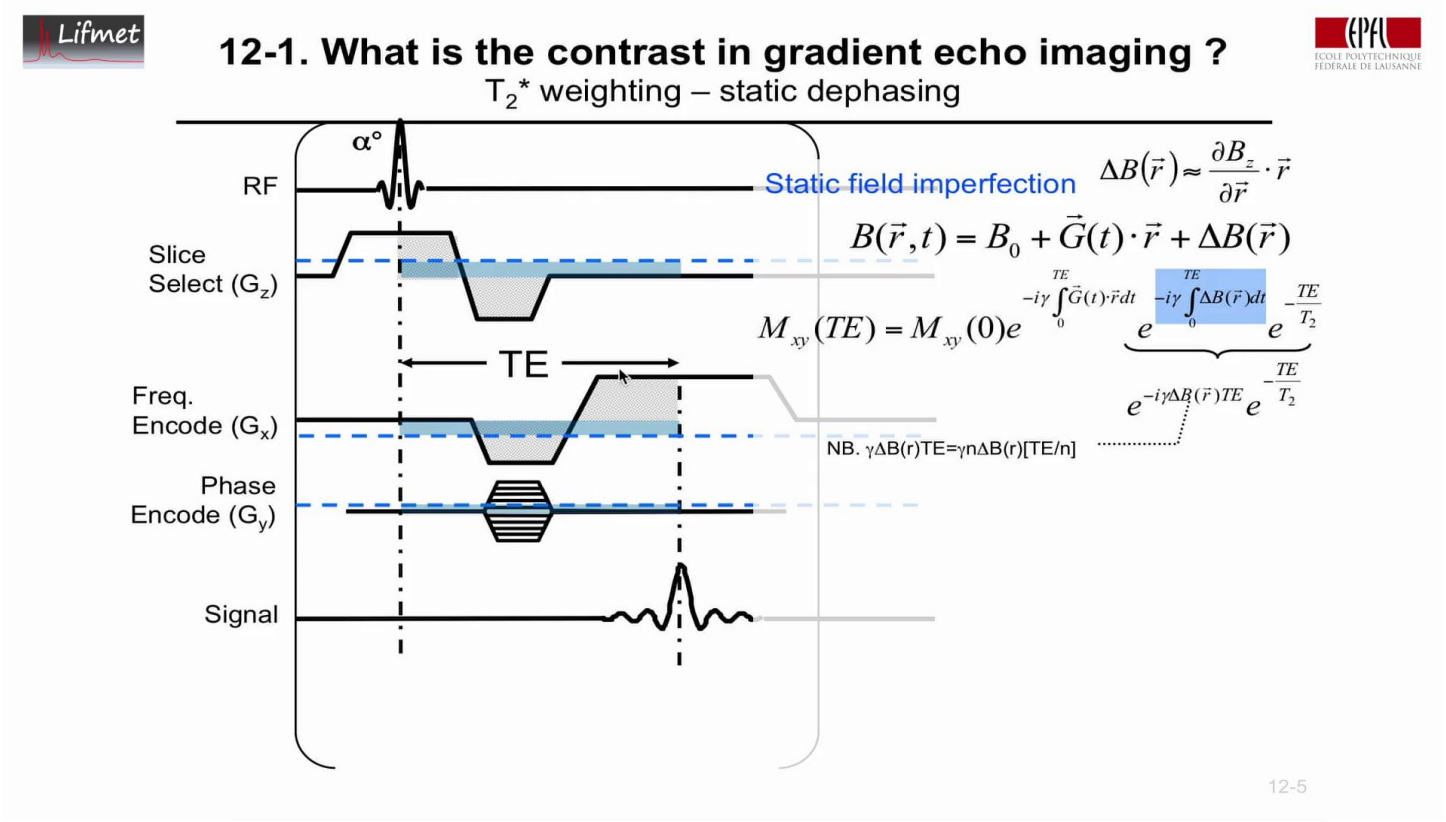


There is a little offset here that is caused by this term *dBz* times *z*. Same for *x* so this is *dBz* derived by *x* times *x* and the same for *y*, the *y* direction. We're here just thinking about a particular voxel. So now our magnetization at time *TE*, the signal here our echo at this time, is given by the magnetization just after the pulse, we'll assume we have perfect refocusing of the slice selection, then we have *e to the minus i gamma* from zero to *TE*. The effect of the gradient actually can include the slice here as well. This will be nullified but we have the effect of the gradient and we have seen that the phase term is proportional to the time integral of the gradient of *T* times *r*. And then we have the term here from the static field, *e to the minus i gamma*, again integrated from zero to *TE delta B of r dT*. That's our deviation that's producing also a phase. So we have-- we are not perfectly at the Larmor frequency now for at a given position *r* and then we have the *T2* effect here.

Notes

Summary

22m 32s

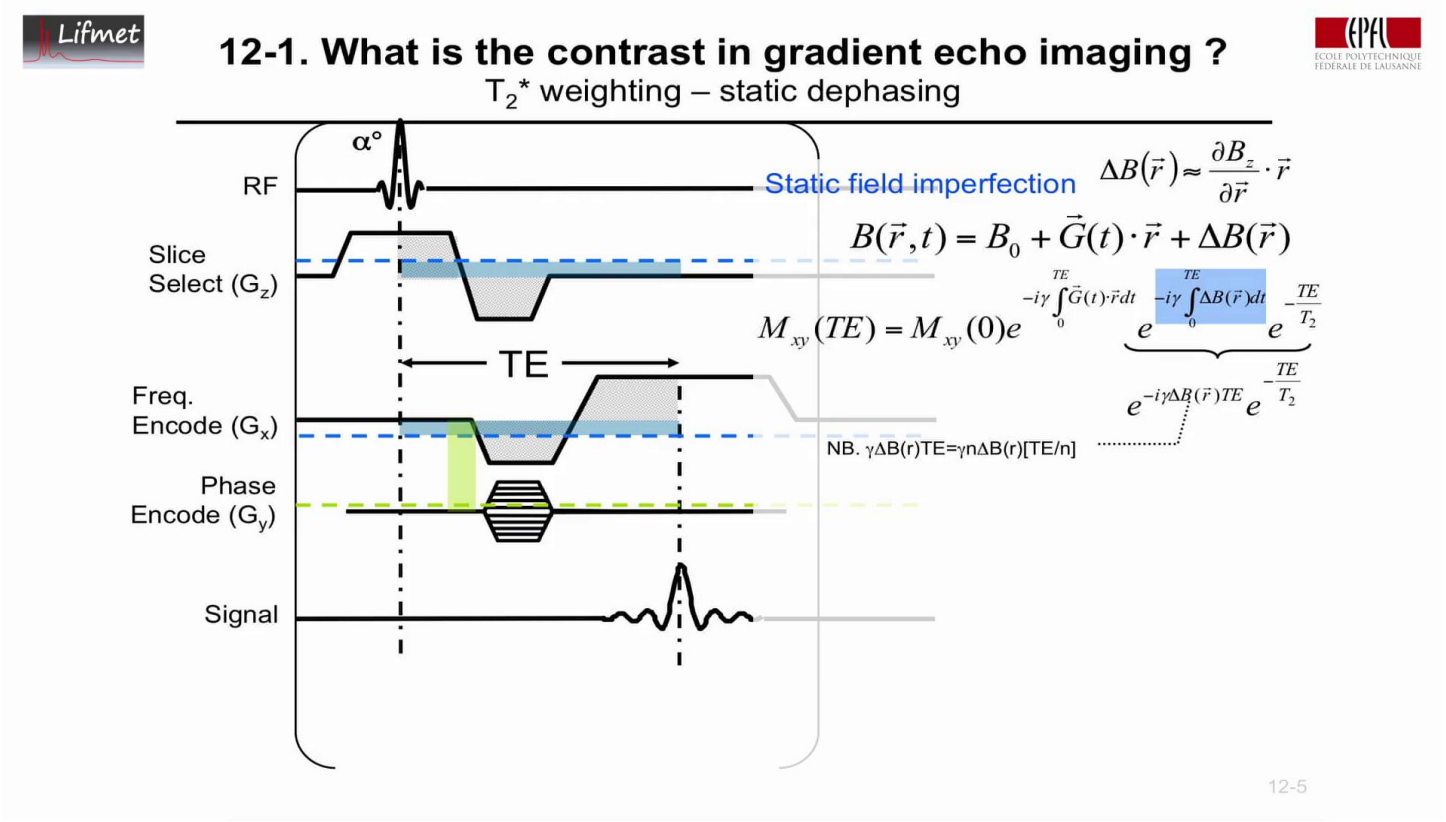


So what this means now is that if you look at this term this is an area, this is the area under the gradient, the effect of the phase, this is also an area on the gradient because we can write this as a gradient just like here but now it's a static gradient that doesn't vary in time, it's constant. And so essentially, we have now the phase effect of the static imperfection and in addition to the effect of the gradient. And now we're going to be concerned with the effect of this gradient. We can write this now. Empirically what one typically writes that this is equals given to, actually this is the integral directly evaluated so *delta B* does not change with time so *delta B* times *TE*, that's the phase effect. And so now what we are going to do to understand what is the effect of this area here, the blue area in the three directions. We're going to use the feature that I introduced last week and that is that a gradient applied simultaneously or sequentially before the echo has the same effect on the echo phase.

Notes

Summary

23m 56s

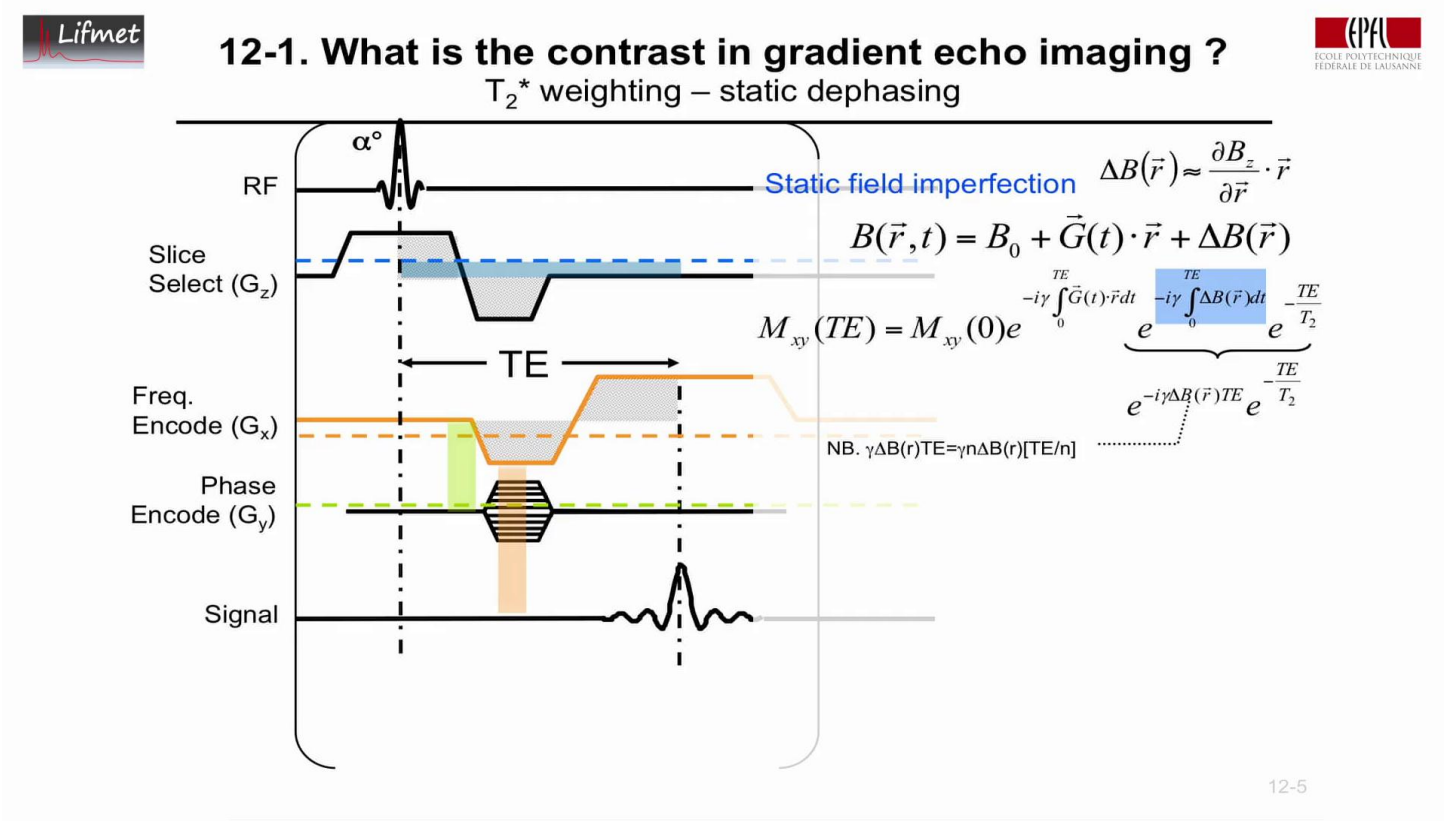


So what we're going to use now here is that the *gamma delta B* times *TE*, I can write this as: *gamma* times *n* times *delta B* times *TE* over *n*. You see I multiplied by *n* and divided by *n*. What does this mean here is now I can pretend to like I'm implying an *n* times bigger static field homogeneity for an *n* times smaller time period. The effect on the phase will be the same. You can verify this term is in here. The *n* cancels out so mathematically it has no influence on the image phase but we can easily visualize this. So let's look at the phase encode gradient which now here appears in green. I will do this now, I will shorten the duration of this inhomogeneity. I'll pretend that this is only applied for a short period of time and I'll increase the amplitude by the same amount so it is now *n* times bigger. The area stays the same. And then I will just say, well I can again with the properties I introduced last week, we can say that this gradient is applied for a short time just before the phase encode gradient. So what does this mean?

Notes

Summary

25m 06s

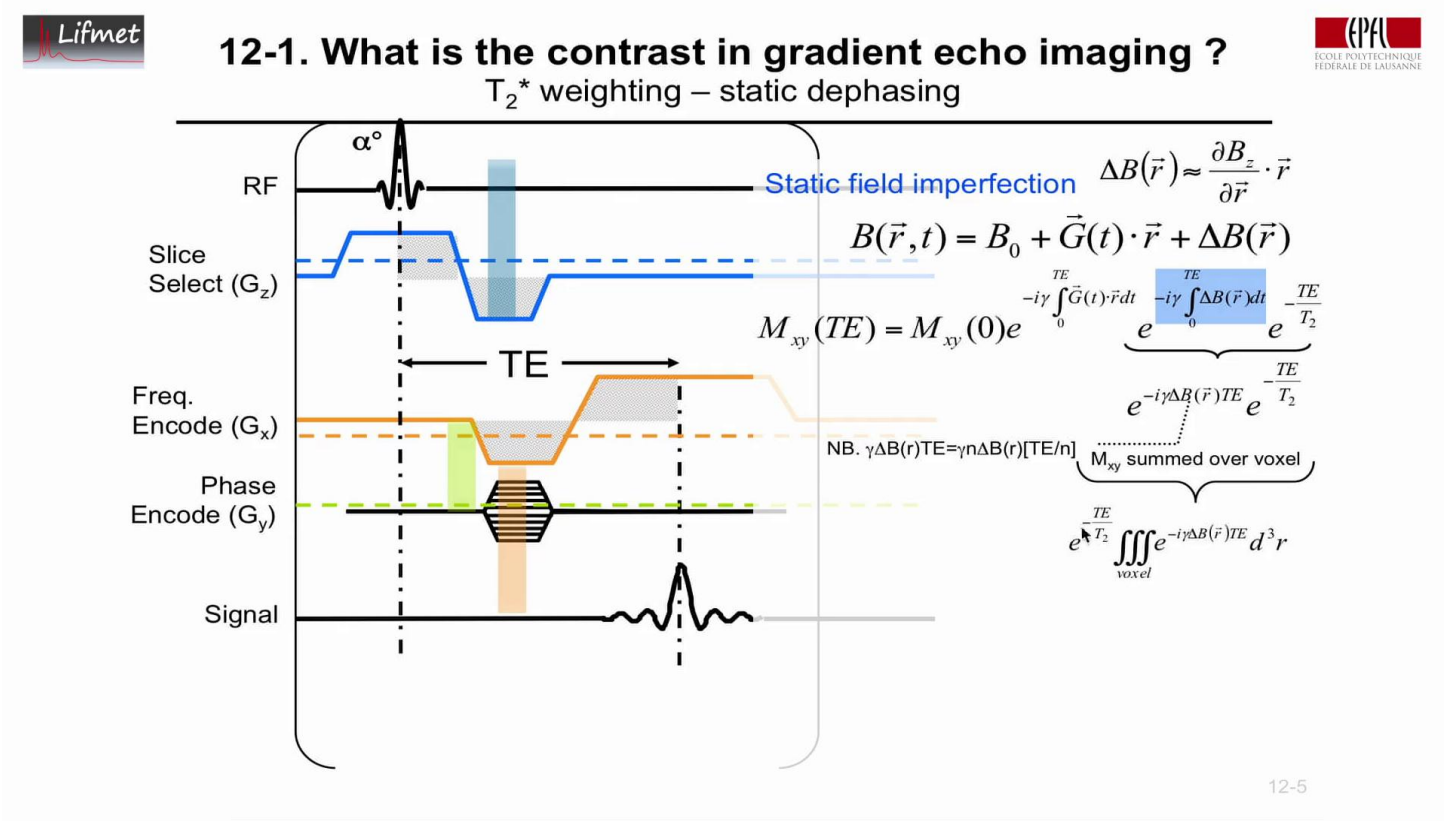


So if we have a voxel where there is a gradient along *y* in that voxel, it has an effect on the magnetization phase as if we had for a short time turned on an additional gradient that corresponds to the *dB0* derived by d*y*, the gradient *y* static gradient times-- as if we have turned on a little gradient here beforehand. We can do the same and they'll change the color here for the readout. So we have the area in orange here will short net increase the amplitude and now we'll move it somewhere else. So what do we see here now for a perfectly balanced imaging sequence we see that the area here is the same so the area the echo will be here. Now the effect of this inhomogeneity is to have a changing the area here. This is of course a huge exaggeration here these effects. But the effect here of the inhomogeneity is to introduce additional dephasing and this will change a little bit the position of the echo. So not much to worry about. However we have the *z* gradient, we'll do the same thing here, so we'll shorten the duration, we'll just mimic again the effect of the long time period.

Notes

Summary

26m 20s

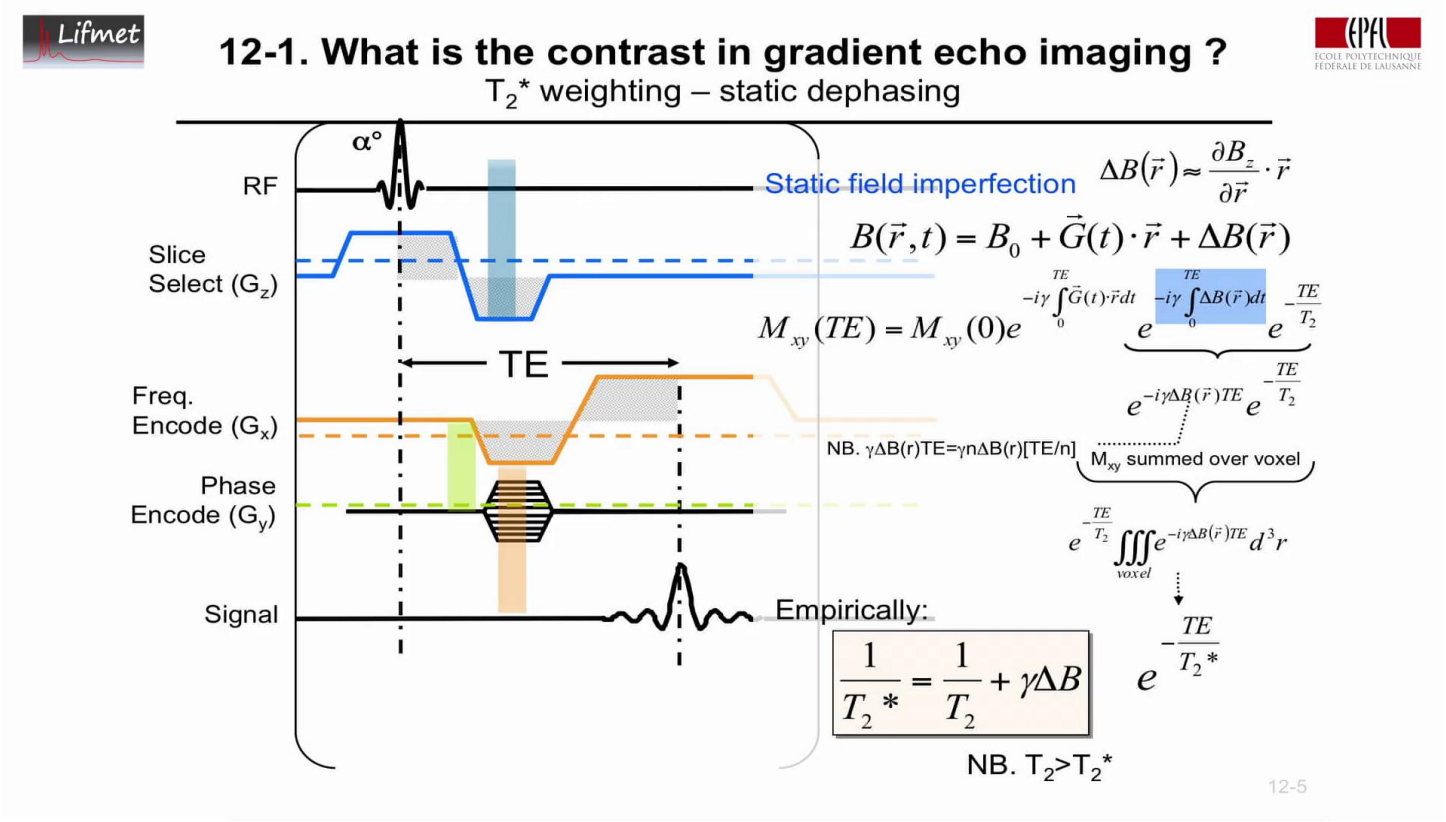


We'll pretend that the effect of the static gradient here is being described by something that is much shorter but has *n* times larger amplitudes. So I'll do this, I'll shorten it, increase the amplitude and I'll put it somewhere else. What is happening here is now we've had said we need to have the area of this gradient equal to the area of this gradient. Now the static inhomogeneity is changing the area of this gradient because it's in this direction and so now we no longer have perfect refocusing of magnetization along *z*. So we no longer have maximum signal. This gradient reduces our amplitude, this gradient and the redirection shifts the echo time, usually it's not perceptible. And the gray in phase encode direction, this additional gradient here will introduce a phase encode an additional dephasing can be substantial. So we have as a result, we have a reduction in signal amplitude. And this we can verify, we'll take this expression here that we've had and we'll now sum the magnetization over one voxel. So the *e to the minus TE* over *T2* is constant, We'll take it out. It's a three dimensional integral over the voxel times this phase term over the echo time.

Notes

Summary

27m 43s



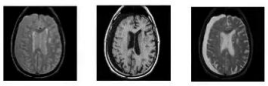
That's the effect of the phase of this term here. We'll assume that all this gradients have cancelled because we're exactly measuring here. So now this term here produces a dephasing integrated over the voxel so we have all sorts of different precessional frequencies in our voxel. This will create a dephasing that increases with *TE* and therefore a reduction in signal. Now what one does here is very empirical. We'll just say, well the signal actually is still decaying exponentially but now with a time constant *T2**, we'll call this time constant *T2**. So empirically there's a relationship between *T2 star and T2*, so one over *T2 star* is equal to one over *T2 plus gamma delta B. Delta B* is always considered to be a positive term, gamma is positive so this is-- the right side is bigger. If the right side is bigger than *T2, then* T2* *is smaller than* T2. Or *T2* is longer than *T2**. So *delta B* is here really an empirical parameter, a flux parameter that just describes the degree of inhomogeneities but it is a way to quantify the inhomogeneities that are present in the voxel.

Notes

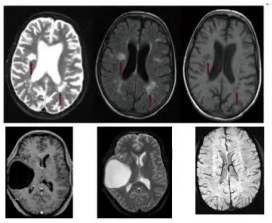
Summary

29m 03s

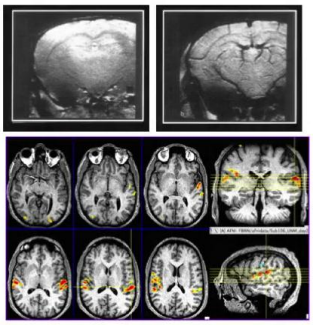
Credits



Walter Block and Franck R. Korosec, PhD, University of Wisconsin
http://slideplayer.com/slide/5780124/



"Your role in the care of patients with MS", Albert David Woods, December 15, 2008
https://www.reviewofoptometry.com/article/your-role-in-the-care-of-patients-with-ms



"Neuroimaging with MRI : some of the things we can see", Robet W. Cox, 2007
https://afni.nimh.nih.gov/pub/dist/edu/2007_12_MRIanat/MRIanat.ppt

So to summarize the effect of static inhomogeneities, imperfections in the static magnetic field creates a faster decay of the signal is faster than pure *T2* and also an opportunity for additional contrast, which we'll see in the next section.

Notes

Summary

30m 15s