EE225E/BIOE265 Spring 2016 Miki Lustig
Principles of MRI

Lab 2: Magnetic Resonance Imaging

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Preliminaries

- Never bring anything metal into the room with an MRI (i.e. keys, metallic jewelry, chairs)
- Do not enter if you have a pacemaker, metal implant, or injury with metal debris.
- Leave your wallet outside the room so that credit and access cards are not demagnetized.
- Do not unplug coils while system is running.
- In the event there is water near the scanner do not touch it. Let me know.
- The  icon indicates an action item.

Introduction

In this lab, we will get hands-on experience with an actual MRI scanner, a Siemens 3 Tesla system. Although clinical MRI systems generally have numerous coils for imaging many different parts of the body, we are borrowing this system from the Brain Imaging Center, and you will find a conspicuous absence of anything but brain imaging coils!

We will go over the basic workflow of MRI scans: patient positioning, landmarking, and scan prescriptions. Additionally, we will adjust the acquisition parameters to manipulate contrast, and demonstrate considerations of sampling for FOV, resolution, bandwidth and SNR.

positioning the patient

The MRI has a translatable table on which the patient lays down. Once the patient is positioned on the table, a laser marker is used to set the origin of the scan acquisition. After this is performed, the table moves so that the set landmark is at the isocenter of the B_0 magnet .

 Setup coil and landmark phantom.

setting up the exam

Before any scanning happens, we must enter relevant patient data including name, age, weight, medical IDs, etc... For medical applications, this is extremely important since this data gets written to the header of *every single image* taken during the exam. The images can then be transferred to a radiologist who is often off-site. By tagging every image with the patient data, we minimize the chance of clinical mix-ups. Patient weight also has safety implications since the specific absorption rate (SAR) - defined as the power

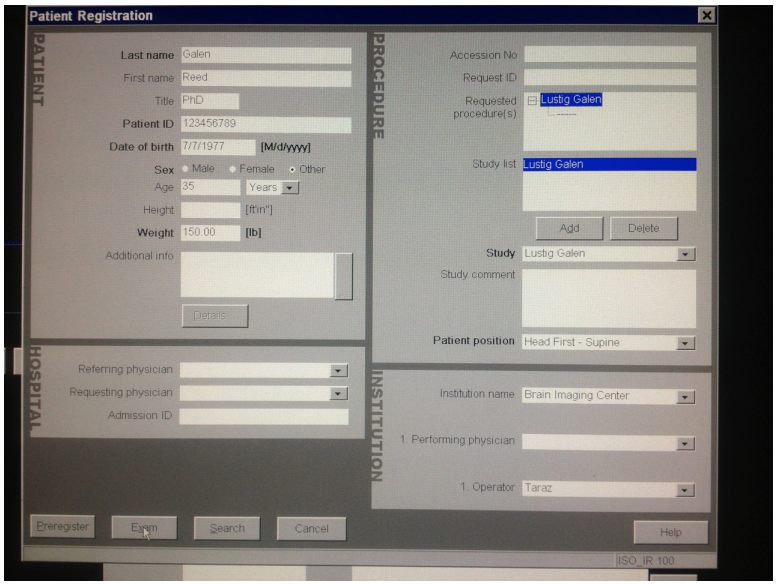


Figure 1: registering the patient

deposited per kilogram of tissue - increases with patient weight. See Figure 1 for the registration interface on the Siemens scanner.

Setup the exam for our phantom.

The Localizer Scan

Once the patient is in the bore of the magnet, we acquire a localizer scan. This is a quick reference scan, in which 1 or a few slices are acquired in the axial, sagittal, and coronal planes (see Figure 2). Most of the subsequent scans are "prescribed" off of these images. Figure 3 shows the localizer images, one per scan plane.

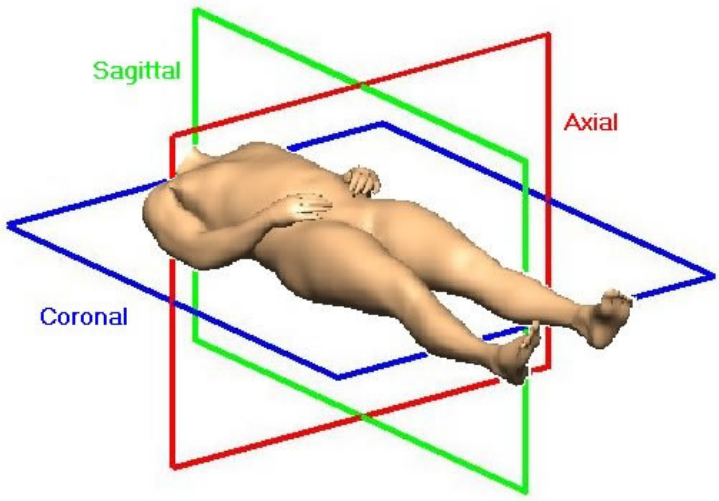


Figure 2: the definition of the scan planes

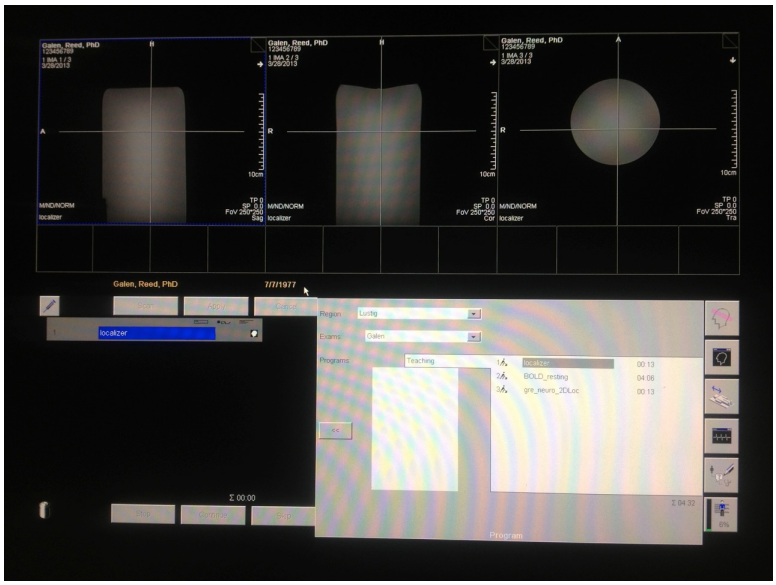


Figure 3: The 3-plane localizer images

Prescribing Sequences

To prescribe an image means to tell the system where to center the slices, FOV offsets, etc... so the image shows up in the correct place. This is performed by dragging and dropping a box representing the scan FOV, slice thickness, etc... The box can usually be rotated for oblique acquisitions. Once the prescription is set and the sequence is saved, the host computed computes the hardware parameters (gradient pulse amplitudes, durations, RF frequencies, etc...) so the acquisition matches the desired prescription. Pseudocode showing a very bare-bones program for a 2DFT sequence is shown below. In addition to looping through slices and phase encodes, all gradient amplitudes (for each phase encode) are calculated and stored into memory as soon as you hit "apply."

%the slice loop;
for $i = 1$ **to** N_{slices} **do**
  %set transmitter center frequency: shifting in slice dir;
  $\omega_{RF} = \gamma G_{ss} z_{slice}$;
  %set receiver freq: shifting in FE dir;
  $\omega_{rec} = BW_{rec} \times \frac{x}{FOV_x}$;
  %the phase encoding loop;
  for $j = 1$ **to** N_y **do**
    %set phase encoding grad amplitude;
    set_pe_amp(j);
    %set receiver phase: add linear phase for shifting in PE dir;
    $\phi_{rec} = 2\pi j \times \frac{y}{FOV_y} + 2\pi y \times \frac{N_y}{2FOV_y}$;
    %do a frequency encode;
    acquiredata(i,j);
  end
end

Algorithm 1: pseudocode for a 2DFT sequence

Experiments

A. 2D RF-Spoiled Gradient Echo Sequence

The first MRI subject we will image is a seeded watermelon. We will use a standard 2DFT spoiled gradient echo (Figure 4) for imaging. This sequence uses a spoiler gradient pulse at the end of the TR in order to dephase any residual transverse magnetization after each data acquisition. In addition, the phase of the RF is incremented from TR to TR. Therefore in each TR signal should arise from the $M_z \rightarrow M_{xy}$ excitation of the RF pulse and is T_1 weighted. Since there is no refocusing pulse, the signal decays with T_2^* after excitation.

FOV, Sampling and Aliasing

Phase encoding sampled discretely at Δk_y intervals, we can model our sampled image $\hat{F}(k_x, k_y)$ as

$$\hat{F}(k_x, k_y) = F(k_x, k_y) \text{Ш}\left(\frac{k_y}{\Delta k_y}\right) \tag{1}$$

After inverse FT reconstruction, the image is

$$\begin{aligned}
\hat{f}(x,y) &= f(x,y) * \Delta k_y \text{Ш}\left(\Delta k_y y\right) \\
&= f(x,y) * \sum_{n=-\infty}^{\infty} \delta\left(y - \frac{n}{\Delta k_y}\right) \\
&= \sum_{n=-\infty}^{\infty} f\left(x, y - \frac{n}{\Delta k_y}\right)
\end{aligned}$$

Even though the k_x is also sampled discretely, we don't get aliasing in this dimension since there is an anti-aliasing filter, a band-pass frequency filter which gets rid of frequencies greater than those

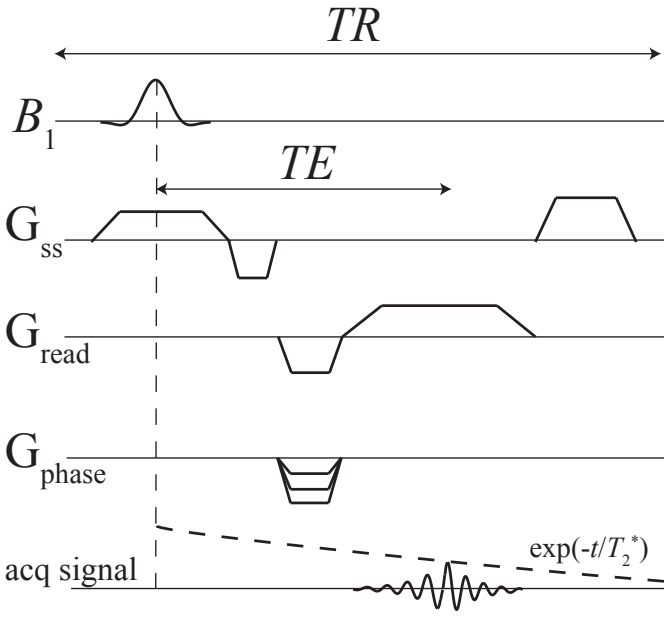


Figure 4: The gradient echo sequence

corresponding to FOV_x. Recall that the FOV relates to the kspace sample size Δk by

$$FOV = \frac{1}{\Delta k} \tag{2}$$

Therefore, we can rewrite the reconstructed image as

$$\hat{f}(x,y) = \sum_{n=-\infty}^{\infty} f\left(x, y - n \times FOV_y\right) \tag{3}$$

This is represents replication in the y direction spaced at FOV_y.

Prescribe a 3-plane localizer sequence and scan. This should produce axial, sagital and coronal images. This will let you localize the object and prescribe future scans. You should get a result similar to Figure 3.

Then select the RF spoiled GRE sequence. On the prescription screen, choose a box in the axial plane, and make the FOV as tight as possible while still covering the entire object (Figure 5). Adjust the resolution so we get as close to a 1 mm pixel as possible. Acquire this image with a 5 mm thick slice. Change the name of the prescription to **"Original"** you are going to use it later. Note the TE, minimum possible TE, Bandwidth per pixel. The quality of the resulting image should be nice!

TE =	$minTE$ =	$BWPP$ =

We will look now at the effect of prescribing a smaller FOV than the object.

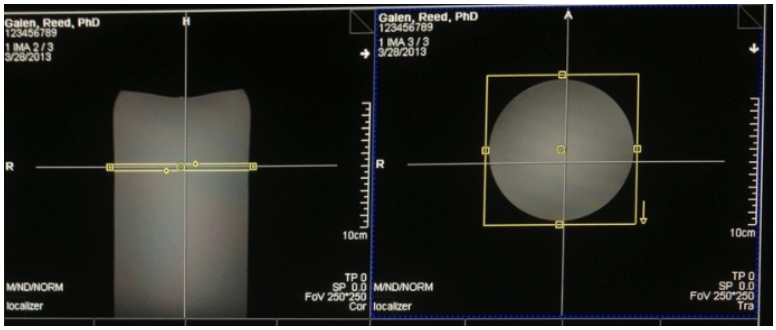


Figure 5: *FOV* just covering the object

Reduce the *FOV* so that it is smaller than the object in both dimensions (Figure 6). By reducing the FOV and keeping the number of encodes the same, we are increasing the resolution. You may encounter a situation where the interface will not allow you to reduce the size of the box. In that case, change the value of the FOV manually in the prescription menu. Change the name of the prescription to **"Small FOV"**.

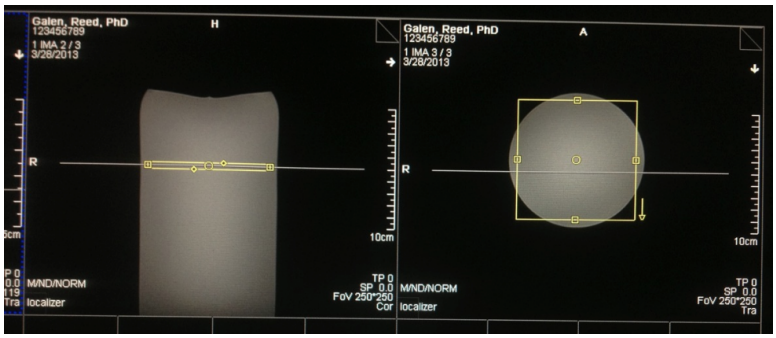


Figure 6: *FOV* reduced below the object size

What has to change in order to reduce the FOV?

Scan the object.

Describe what happened. In which direction do you see aliasing?

Now, sap the phase encode direction and rescan the image.

What artifacts do you see?

B. **SNR tradeoffs**

We can estimate how SNR changes with sequence parameters if we ignore T_1, T_2. and T_2^* decay. Recall that

$$SNR \sim \sqrt{t_{AD,total}} \tag{4}$$

The parameter $t_{AD,total}$ is the total time the receiver is acquiring data. This parameter can be broken down as

$$t_{AD,total} = t_{AD} \times N_{PE} \times N_{avg} \tag{5}$$

where t_{AD} is the A/D time in 1 TR, N_{PE} is the total number of phase encodes, and N_{avg} is the number of signal averages. Note that we are definitely ignoring transverse decay effects, since continually increasing the A/D time in a single TR will not give any SNR benefit after the signal has decayed. The SNR scales linearly with pixel volume V

$$SNR \sim V = \delta x \times \delta y \times \delta z \tag{6}$$

since increasing V will linearly increase the number of spins contributing to the signal. Therefore, the SNR scaling with pixel volume and scan time is

$$SNR \sim V \times \sqrt{t_{AD,total}} \tag{7}$$

The A/D time in a single TR (t_{AD}) is frequently represented in terms of the receiver bandwidth (BW), which is related to the single-sample time t_s as

$$BW = \frac{1}{t_s} \tag{8}$$

The system we use quotes this in terms of the "bandwidth per pixel" ($BWPP$) which is defined as

$$BWPP = \frac{BW}{N} \tag{9}$$

where N is the number of frequency encodes. Therefore, the single-TR A/D time (t_{AD}) is just

$$t_{AD} = N \times t_s = \frac{N}{BW} = \frac{1}{BWPP} \tag{10}$$

Copy and paste the original gradient echo sequence with the full FOV (Figure 5).

Based on the $BWPP$, resolution and scan prescription, what are the following:

1. the duration of the A/D window =

2. the lower limit of TE =

3. number of phase encodes =

Copy sequence and change the name of the prescription to **"SNR = 100"**. Scan the image again. This will serve as our reference SNR.

Change the prescription to obtain double the spatial resolution (better resolution) by keeping the same FOV, and doubling the number of frequency and phase encodes. Choose Magnitude and Phase acquisition.

Calculate the expected relative SNR compared to the reference.

Change the name of the prescription to **"SNR = xxx"**. Scan the object.

Do you see the increase in resolution and loss of SNR?

Based on the resulting SNR, how many averages are required to get back to a similar SNR as the reference?

Prescribe the sequence with the number of averages. Comment on the image quality.

How long was the scan time compared to the reference?

Another way to increase the SNR is to reduce the bandwidth of the acquisition. This will result in more A/D time during a TR, but the overall scan time will not increase.

Copy the prescription from **"SNR = xxx"**. Reduce the bandwidth per-pixel parameter such that the resulting SNR should be the same as the reference. The scanner may complain that in order to do so, another parameter needs to be adjusted. Why?

Change name to **"reduced BWPP: SNR = xx '** and acquire magnitude and phase. Scan the object. Comment on the image quality. Did the contrast change? Does the SNR resembles the reference? If not, why? What happened to the phase?

What happened to the phase?

C. T_2 **versus** T_2^*

In this experiment we will acquire T_2 and T_2^* "weighted" images of the watermelon. In the gradient echo sequence (Figure 4), the encoded signal at the echo will be

$$S_{gre} \sim e^{t/T_2^*}. \tag{11}$$

This assumes that T_2^* is the limiting decay term (this is not always the case, since the proton NMR signal in many environments - especially in solids - has extremely short T_2 values). In order to weight the image heavily with T_2^* contrast, we will use a longer echo time. We will compare this image with a regular spin echo sequence (Figure 7) with identical TE, TR, and resolution. Since the spin echo sequence will be proportional to the actual T_2 decay

$$S_{SE} \sim e^{t/T_2}, \tag{12}$$

we can attribute image intensity differences to the T_2 / T_2^* weighting.

Copy and paste the **"Original"** gradient echo sequence with the full FOV (Figure 5). Increase TE to 40 ms and the TR to 500 ms. Change the name to **"GRE TE40/TR500"**. Acquire magnitude

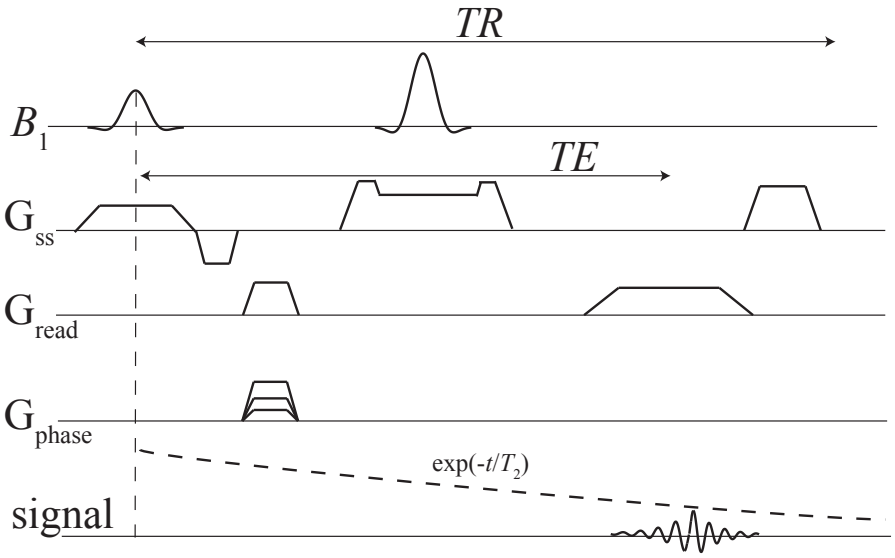
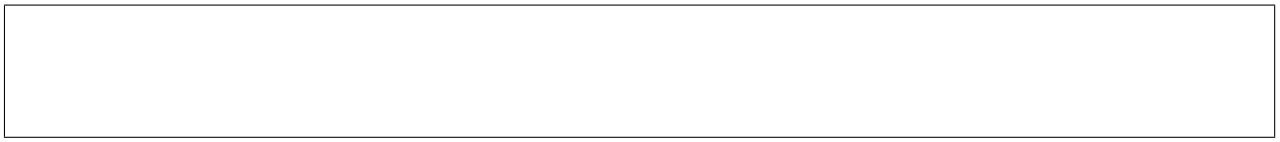


Figure 7: The spin echo sequence

and phase. Also, change the flip-angle to 90 degrees. Acquire this scan, and note the regions of signal dropout. What causes this spatially-nonuniform signal loss?

Prescribe a spin echo sequence with the same TE/TR resolution and BWPP. Name it **"SE TE40/TR500"**. Again, acquire magnitude and phase. Scan the object with the new sequence. The difference should be striking! How do explain the difference in contrast and SNR?



D. **MRI of Phantoms: T_1, T_2, and Chemical Shift Response**

In the next experiment, we image several different phantoms. These are containers of liquid with varying relaxation and chemical shift properties.

Prescribe a localizer and scan. Prescribe a 2DFT gradient echo sequence with the original parameters from **"Original"**. Scan the object. Can you see a chemical shift?

Prescribe a new scan. Change the BWPP to 40Hz. Name it **" GRE BWPP40"**. How much chemical shift should you expect?

The scanner might complain about the echo time. Increase if necessary. Scan the object. Do you see the chemical shift? Why do you see changes in signal intensity?

E. Echo Planar Imaging

The EPI sequence diagram is given in figure 8. It consists of an excitation RF pulse followed by a rapidly oscillating readout gradient train and phase encoding blips. In this manner, some or all of k-space can be acquired in a single shot (excitation). EPI is extremely quick, but it suffers from some high sensitivity to off-resonance. Recall that the off-resonance image-domain shift (in pixels) is given by

$$\Delta x = \Delta f \times BWPP, \tag{13}$$

where $BWPP$ is defined as

$$BWPP = \frac{1}{t_s N} \tag{14}$$

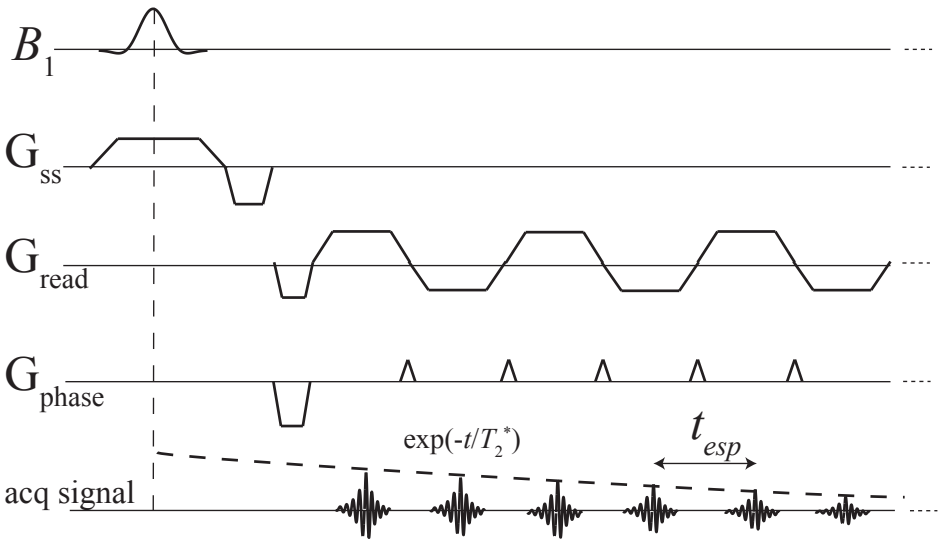


Figure 8: The EPI sequence

t_s is the time to acquire a single sample, and N is the total number of pixels in the direction under consideration. In the frequency encoding direction, we are often using $t_s \sim 10$ to 100μs , so the total bandwidth is ~ 10 kHz, and the bandwidth per pixel is ~ 100 Hz. In EPI, however, we must replace t_s with t_{esp} - the spacing between echoes- in the slow (phase encoded) direction (see Figure 8). This value is typically a few ms, so the $BWPP$ is reduced by at least an order of magnitude and is on the order of 10 Hz per pixel. This means that the fat signal ($\Delta f = 440$ Hz at 3T) is shifted 50 pixels, or significant proportion of the FOV!

First, run the localizer with the new phantoms loaded in the scanner. Prescribe an EPI sequence in the axial plane and turn off the lipid suppression. What is the $BWPP$ in the x and y directions? How many pixels will fat be shifted in each dimension?

Change the name to **"EPI"** and acquire an EPI scan. Comment on the result. Comment on the signal from the other bottles.

EPI is very sensitive to many system imperfections. The scanner attempts to correct these by performing a calibration scan. However, the correction is never perfect. The results is the so called N/2 Ghost. Can you identify the ghosts? Why are their ghosts? And what causes their displacement?

F. Fat Suppression

The large spatial shift of fat creates intolerable artifacts when imaging human subjects, so we must always use some sort of lipid suppression with EPI. There are typically two major types of lipid suppression techniques. In the "fat saturation" method, we apply a selective 90 degree pulse at 440 Hz off-resonance. A dephasing gradient pulse is played immediately after this pulse to get rid of any M_{xy}, and then the slice selective excitation pulse is played immediately afterwards. Since there was a 90 degree pulse played on lipids milliseconds prior, there is no M_z left to contribute to the signal. The other lipid suppression technique uses the slice select pulse as also as a frequency-selective pulse. It selectively excited water while leaving lipids in the passband. See Figure 9 for a diagram of these suppression strategies.

First, we analyze the robustness of the water-selective excitation strategy (Figure 9b). Recall the the

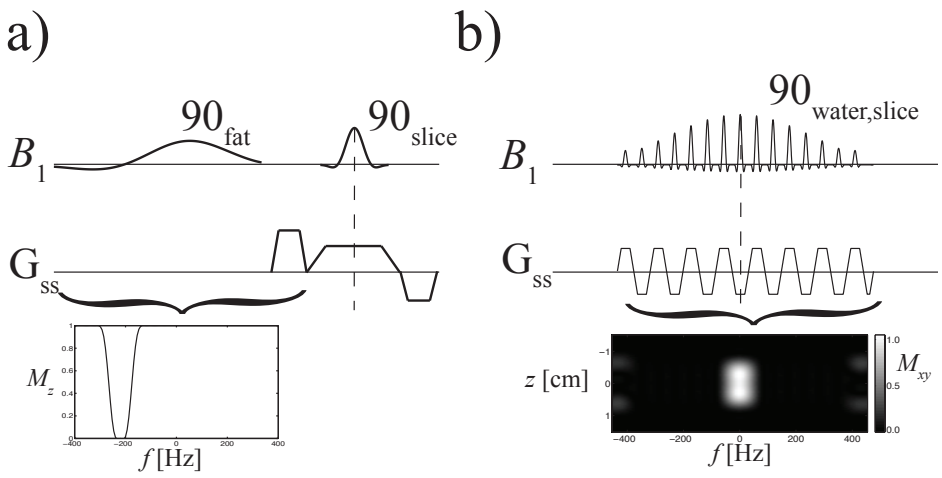


Figure 9: EPI fat suppression techniques. Fat saturation (a) selectively eliminates the M_z at the fat frequency. A dephasing gradient is applied after the pulse so the excited magnetization from the first pulse does not contribute to the signal. Water-only excitation (b) uses a spectral spatial pulse

Fourier approximation for the pulse excitation profile:

$$\begin{aligned} M_{xy}(t) &= ie^{-i\omega t}\int_0^t B_{1,xy}(\tau)e^{-i\omega\tau}M_z\,d\tau \\ &\approx iM_0e^{-i\omega t}\mathcal{F}\left\{B_{1,xy}(t)\right\} \end{aligned}$$

The approximation is valid when θ - the pulse flip angle - is small, since $M_z = M_0\cos\theta \approx M_0$. Lets let the RF pulse be a perfectly selective pulse (we can neglect its finite duration for now):

$$B_1 = |B_1|\ \mathrm{sinc}(t/T) \tag{15}$$

so

$$M_{xy} = iM_0e^{-i\omega t}\,|B_1|\sqcap(T\omega) \tag{16}$$

The above expression indicates that transmitter magnitude imperfections (variations of $|B_1|$) have no effect whatsoever in the stop band of the pulse. In other words, in the pulse stopband ($|\omega| > 1/T$), the excited magnetization M_{xy} is zero regardless of changes to the pulse amplitude $|B_1|$. So if the lipid signal is in the stopband of a water-selective excitation (Figure 9b), then transmitter variations will not effect the performance of the lipid suppression.

Now take the selective saturation of the lipid resonance (Figure 9a). In this case, we place the lipid signal in the *passband* of a 90 degree pulse prior to excitation in order to eliminate the M_z of lipid. Assuming the saturation pulse is exactly placed on the lipid resonance, the flip angle imparted to the fat signal is just

$$\theta_{fat} = \gamma\int_0^T B_1(t)\,dt \tag{17}$$

The residual M_z will be proportional to $\cos\theta$, and the fat signal leaking into the imaging slice will be

$$\begin{aligned} S_{fat} &\sim M_{z,fat}\sin\theta_{slice} \\ &\sim \cos\left(\theta_{fat}\right)\sin\theta_{slice} \end{aligned}$$

If we assume that we have an imperfect saturation pulse, then

$$\theta_{fat} = 90^\circ + \delta\theta_{fat} \tag{18}$$

where $\delta\theta_{fat}$ is the amplitude of the imperfection of the saturation pulse. Since

$$\cos\left(90^\circ + \delta\theta_{fat}\right) = -\sin\left(\delta\theta_{fat}\right) \approx -\delta\theta_{fat} \tag{19}$$

then

$$S_{fat} \sim -\delta\theta_{fat} \sin\theta_{slice} \tag{20}$$

In other words, the signal magnitude grows linearly with pulse amplitude imperfection $\delta\theta_{fat}$. Therefore, in circumstances where we have an spatially-inhomogeneous transmitter, the spectral-selective excitation almost always has superior performance to "pre-saturation" in keeping the lipid signal low.

Prescribe an EPI sequence using water excite, spectral spatial pulse. Name it **"EPI WE"**. Scan the image. Comment on the result.

Prescribe an EPI sequence using fat saturation pulse. Name it **"EPI FS"**. Scan the image. Comment on the result.

Which of the methods performed better overall? Depending on the day, you might get a different answer!

G. T_1 estimation

The steady state response of the spoiled gradient echo sequence (Figure 4) is given in Nishimura (pg 151):

$$S = \frac{(1 - E_1)\sin\theta}{1 - E_1\cos\theta} M_0 \tag{21}$$

with

$$E_1 = e^{-TR/T_1} \tag{22}$$

The limiting case of $\theta = 90^\circ$ is called the "saturation recovery" sequence, and the response simplifies to

$$S = \left(1 - e^{-TR/T_1}\right) M_0 \tag{23}$$

If we take a short repetition time ($TR \ll T_1$), then

$$1 - e^{-TR/T_1} \approx 1 - \left(1 - \frac{TR}{T_1}\right) \tag{24}$$

so

$$S = \frac{TR \times M_0}{T_1} \tag{25}$$

Sometimes it is more natural to talk in terms of $R_1 = 1/T_1$, the relaxation rate. Then, from equation 25, the signal is linear in TR with proportionality R_1.

Prescribe a saturation recovery sequence. Name it **" IR TR = 5ms"**. Scan using the sequence with a short 5ms TR. Notice the artifacts. What are these a result of? How can we avoid these artifacts?

Use several values of TR: 5, 10, 20, 50, 100 (name them appropriately). Identify the fast and slow T_1 phantoms. Estimate roughly the T_1 of one of the phantoms using equation 25. Based on the T_1 measurements, what do you expect the relative T_2 of each phantom to be?

H. T_2-weighted Imaging with the Fast Spin Echo Sequence

A diagram of the fast spin echo (FSE, also sometimes called turbo spin echo or TSE) sequence is given in Figure 10. The sequence consists of an excitation 90° followed by multiple refocusing pulses. These pulses can - and usually do- deviate from 180°, and they are almost always played using the CPMG phase schedule, meaning the refocusing pulses have a 90° advance with respect to the excitation. All pulses are played with slice-selective gradients, and there are also crushers surrounding the 180s which are usually bridged with the slice select gradients. These are for suppression the $M_z \rightarrow M_{xy}$ component of the pulse (excite magnetization from non-ideal 180° behavior).

Modeling the signal response when $\theta < 180°$ is complicated (see the extended phase graph algorithm). Generally, there is some mix of T_1 and T_2 decay, and the contrast can be manipulated by changing θ of the refocusing pulses. Even though we are suppressing the $M_z \rightarrow M_{xy}$ echo pathway by the crusher gradients, we could still have a $M_{xy} \rightarrow M_z \rightarrow M_{xy}$ "stimulated" echo. In this way, some T_1 weighting is introduced.

For simplicity, we will only treat the $\theta = 180°$ case, so the signal response is a pure T_2 decay curve. In this case, the phase encoded direction is windowed. i.e.

$$F(k_x, k_y) \rightarrow F(k_x, k_y) e^{-n_{ky} t_{esp}/T_2}$$

where t_{esp} is the echo spacing and n_{ky} is the phase encoding sample index. Therefore, we attenuate along the phase-direction.

Prescribe a fast spin echo sequence and name it **"FSE"**. Acquire a fast spin echo image in the axial scan plane.

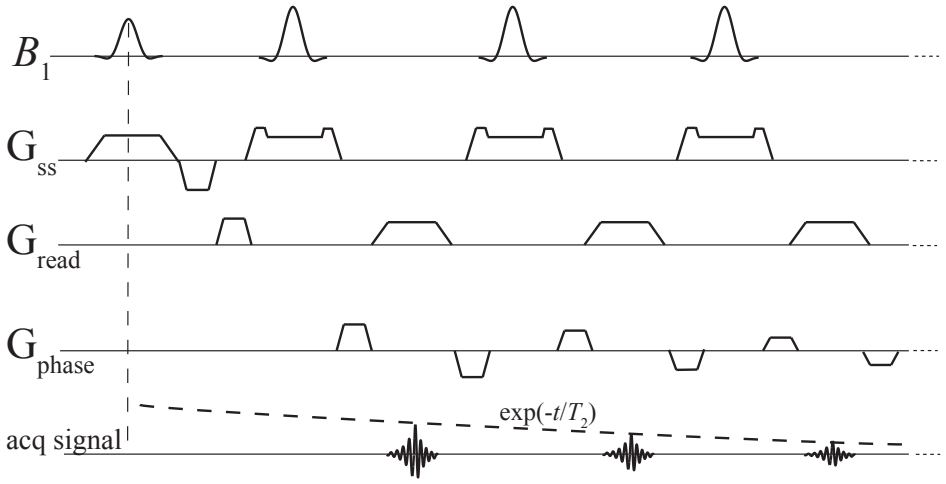


Figure 10: The fast spin echo sequence

What image artifact would this introduce for short T_2? Which phantoms exhibit this artifact the most? Do these correspond to the phantoms with short or long T_1 values?

Getting Scanned (Optional)

When getting scanned, there are 3 issues to be cognizant of. The first is noise: the scanner can be extremely loud when on the inside. For this reason, you are required to wear ear protection. The second is claustrophobia. When getting your brain scanned, your head will be in the center of the bore cylinder. Some people find this sensation uncomfortable. If you are susceptible to claustrophobia, talk to us before going in the scanner. Lastly, stay still! Your images will look best if you stay still as a mummy while the scanner is making noise.