

Goals

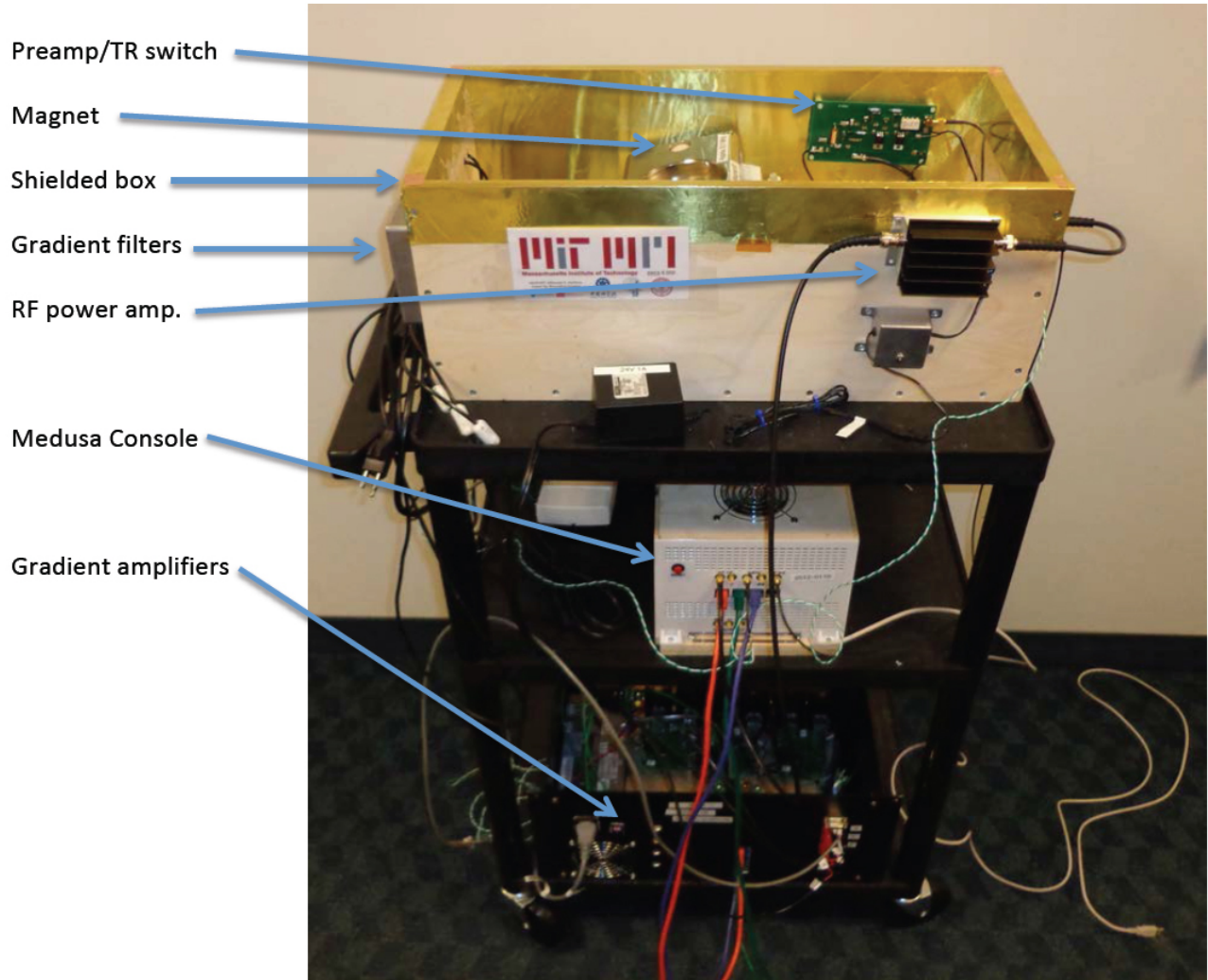
In this Tabletop MRI lab, you will get familiar with the portable MRI system and collect simple magnetic resonance signals and images from water-containing phantoms. After learning some basic procedures on the system, you will begin to use the spatial gradients to do one-dimensional projection imaging. Finally, you will use spatial gradients along multiple dimensions to encode multiple axes and create 2D and 3D images. Here we go!

1. Introduction to the equipment

1.1 About...

The system you are about to use came together across 3 continents and was the work of at least 20 people. It was coordinated by and assembled at the Massachusetts General Hospital (MGH), Martinos Center MR Physics group, with key components coming from other MR groups around the world. The rare-earth magnet was designed and constructed by the group of Prof. Wenhui Yang at the Institute of Electrical Engineering of the Chinese Academy of Sciences in Beijing. The scanner's Medusa console has been a >5 year development project by Greig Scott and Pascal Stang at the Stanford EE Department (<http://mrsrl.stanford.edu/~medusa/hardware/>). The gradient coil current contours were calculated by Maxim Zeitsev and Feng Jia of Freiburg University using target field design software they developed, and implemented into circuit board by Cris LaPierre of MGH. The gradient amplifier was designed by Thomas Witzel at MGH. The sequence software and GUIs were written by Jason Stockmann, Bo Zhu and Clarissa Zimmerman. The systems were constructed and tested by Clarissa Zimmerman, Jason Stockman, Lawrence Wald, Cris LaPierre and Bo Zhu at MGH. This handout was adapted from literature developed by Clarissa Zimmerman and Audrey Fan for the MIT subject 6.S02.

1.2 System orientation

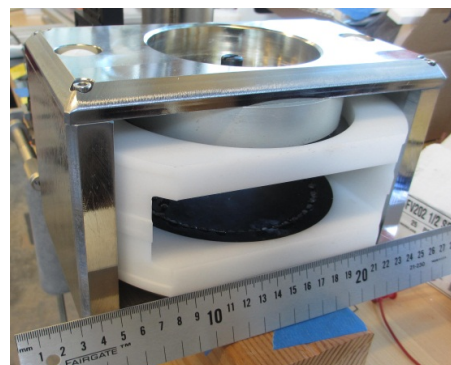


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1.3 System components

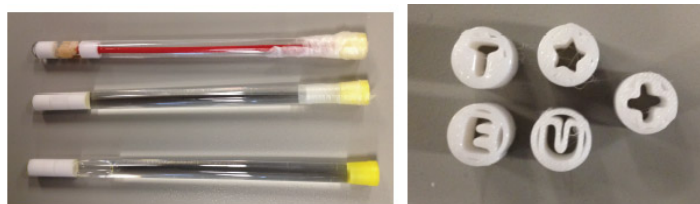
1.3.1 Magnet

The B_0 field is created by a small 0.19 Tesla permanent magnet. The two rare-earth magnet disks are held apart with an iron yoke, which also provides a flux-return path for the magnetic field, containing the magnetic field to the gap between the pole pieces (and inside the iron).



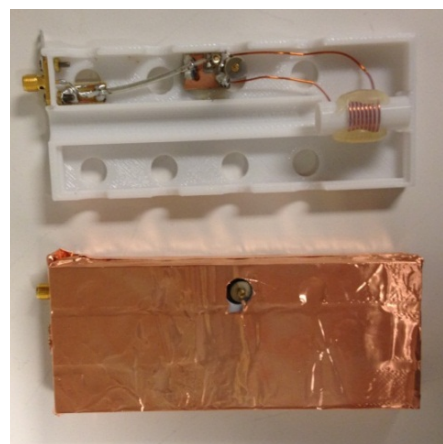
1.3.2 Phantoms

Phantoms are artificial imaging samples with known dimensions and features. For this MRI system, the imaging phantoms are contained in 1cm diameter glass tubes.



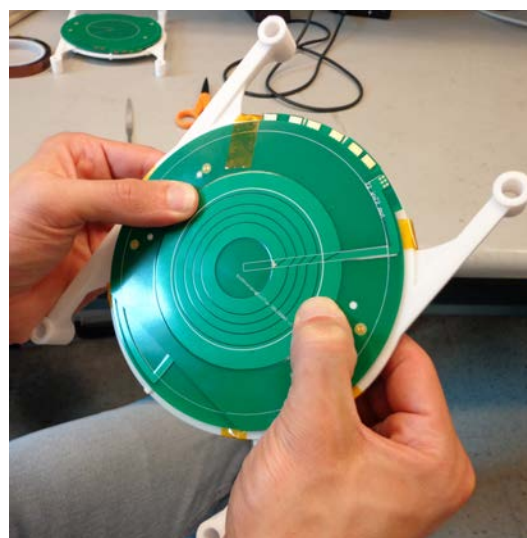
1.3.3 RF coil

The RF coil for this system is used for transmitting the excitation pulse to get the magnetization precessing and to detect the MRI signal through the Faraday detection principle. The coil is part of a resonant circuit that is tuned to the Larmor frequency of the B_0 magnet. The coil is a solenoid into which the NMR tubes fit snugly; it is contained in a plastic case which is wrapped in copper foil to shield against external RF noise sources.



1.3.4 Gradient coils

The gradient coils generate the linear gradient fields that are used for imaging. They produce a magnetic field in the z direction, with amplitude that changes linearly as a function position, with a slope of 14 Gauss/cm when driven with 1 Ampere of current. With 2-A current, the slope is 28 G/cm so we refer to the sensitivity of the gradients as 14 G/(cm A). The gradient coils for this system are contained on a printed circuit board that is inserted into the magnet bore leaving a space just over 1cm for the sample and RF detector/excitation coil.



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1.3.5 Gradient amplifier

The gradient amplifier is used to supply the current to the gradient coils. Since it's the fields we care about, and the fields are proportional to current, this amplifier can be viewed as a voltage to current transducer; it takes a voltage waveform from the console and creates a current proportional to that voltage in the gradient coil. It is similar to a common audio power amplifier except that it must also be able to output DC currents. It uses a power op-amp followed by a current sensor. The output of the current sensor is compared to the input voltage to ensure that the current itself is proportional to the input voltage. A current sensor is created by measuring the voltage across a small resistor in series with the output.

1.3.6 Transmit/Receive Switch

The RF coil is used for 2 things: transmitting the RF pulse and receiving the MRI signal. This means that it should sometimes be connected to the transmit amplifier and sometimes be connected to the receive amplifier. This TR switch uses passive components to effectively switch the coil connection between the two amplifiers.

1.3.7 Console

The console interfaces with the computer via MATLAB®. It produces the RF transmit pulses and gradient waveforms based on vectors created in a MATLAB sequence. It also acquires the received MRI signal at a time specified in the sequence. The console samples the received signal and demodulates it to baseband. Some specifics are at:

<http://mrsrl.stanford.edu/~medusa/hardware/>

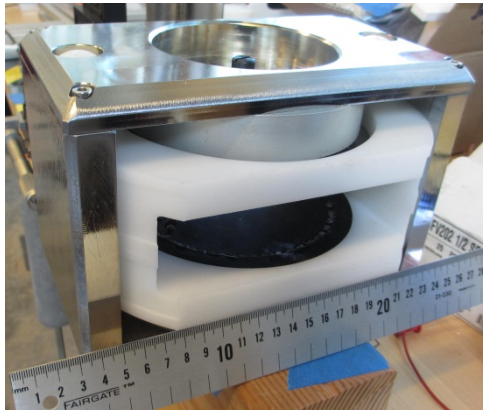
1.3.8 MATLAB GUI

The MATLAB GUI is what you will use to edit and run sequences. You may run an FID sequence or a spin echo sequence. You can control the amplitude of the RF pulses and gradient waveform. You will also control the repetition time (TR), echo time (TE), and the read out time.

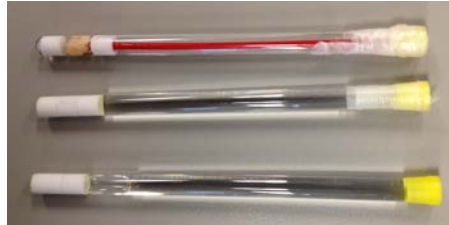


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1.4 Summary: MRI system components



B₀ Magnet



Imaging Phantoms



RF Coil and Enclosure



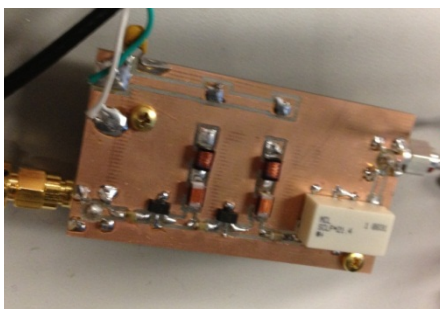
Gradient Coil PCB



Gradient Amplifiers



Console



Receive Amplifier

Transmit/Receive Switch



Transmit Amplifier

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2. Acquire MR signals

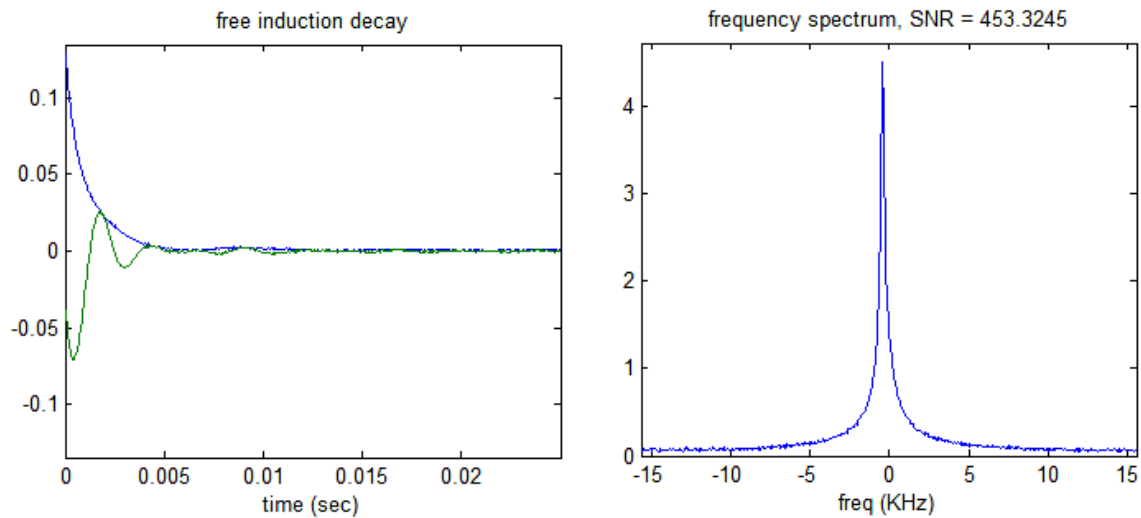
2.1 Making sure your scanner is connected properly

- Plug in the power cord of your portable MRI system, turn it on, and make sure the Medusa console displays a flashing green light.
- Plug in the USB cable from the portable MRI system onto your desktop station to connect the MRI hardware to your computer.
- The staff has installed the driver for the Medusa system associated with your portable MRI. To launch the imaging software interface, click on the desktop icon: MRI_GUI_v4p2
- Your current directory should be “*dir*” = C:\Users\student\Desktop\MRI_GUIs_v4p2\. We need to add the appropriate filepaths in MATLAB. On the left-hand panel of MATLAB, navigate to “*dir*”, right-click on “*dir*”, and then “Add to path → Selected folders and subfolders”. Also add the “*medusa_driver*” folder and subfolders in the same way.

2.2 Free Induction decay (FID)

- Insert the two-cylinder water tube (phantom #3) all the way into the RF coil in the magnet. Ask the lab instructor for help with positioning, if necessary.
- For this lab, you will use the “FID GUI”. Please click on this button in the imaging software window.
- There is a piece of paper with the recorded Larmor frequency for your particular magnet. If you cannot find this value, check on the following pdf document under your magnet unit number: *6.S02_2014_MRI_Lab_Frequency_and_Shim_Settings.pdf*. This Larmor frequency should have a value close to 8.15 MHz; type this into the GUI for under “**Frequency (MHz)**”.
- Click “Run Scan”.

Check yourself: Spend a few minutes to acquire the first free induction decay (FID) if necessary, and if you are still having trouble observing an FID similar to the one below, raise your hand and ask for help from the staff.



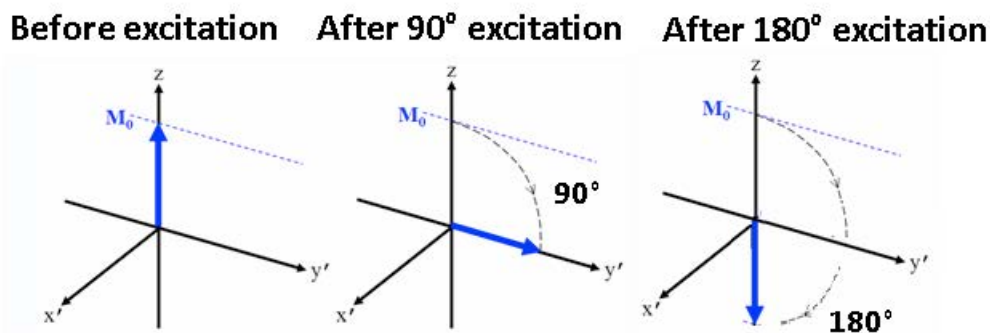
Courtesy of [Professor Lawrence Wald](#). Used with permission.

- On the left is an example of a time domain signal, and the right is the corresponding frequency domain signal. The signal to noise ratio may vary depending in particular on the water content of the sample you are using.
- The blue curve on the free induction decay (FID) is the magnitude of the signal and the green trace is the real component (M_x component) of the signal.
- The frequency domain is displayed with the center frequency (0 Hz) corresponding to the frequency you typed into the FID_GUI in the "Frequency (MHz)" box (for instance, 8.15 MHz).

Known bug: If the signal inexplicitly and suddenly disappears, press the "Stop Scan" button FID_GUI and press the red reset button on the Medusa console. Sometimes when changing the frequency, the Medusa console gets confused and even a reset doesn't bring it back. In this case, "power cycling" can help.

2.3 Flip angle calibration

- In the lecture and readings, you should have learned about the **flip angle**, which is the angle of the net magnetization vector relative to the main field (B_0) direction after RF excitation.



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- On the excitation side, we can adjust the power of the radiofrequency (RF) pulse. This adjusts the amplitude of the RF transmit (B_1) pulse. Because achieved flip angle is proportional to the area under the RF pulse curve, changing the transmit power will adjust the flip angle. In this part, we will find the RF transmit powers that corresponds to 90° and 180° flip angles.
- Stop any ongoing runs, check the “flip angle calibration” box in the FID_GUI, and then start again. This automatically steps through the power of the RF excitation pulse and plots the amplitude of the FID as it goes. It steps from power of 0 Watts to ~ 1 Watts.
- Use the cursor to click on the left-most maximum in the graph at the bottom right of the GUI. The x-value you see represents the power required for a 90° pulse.
- Record the 90° power setting: _____ Watts. Enter this power setting on the FID_GUI.

2.4 Magnet shimming

- The goal of shimming is to improve the homogeneity (uniformity) of the main field, which will narrow the linewidth of the frequency domain signal. We do this by adding small DC currents to the gradient coils. This produces a linear field variation along x, y, and z that can cancel the first-order variations in B_0 that exist in the magnet.
- Our goal is thus to maximize the homogeneity of the magnetic field over the sample. Since frequency is proportional to B, this means as **narrow** a linewidth as possible.
- The area of the frequency domain spectrum is determined by the amount of magnetization (number of protons magnetized) in the sample. This area is fixed because we are not changing the amount of water in the phantom. Because the area is fixed, the peak height of the frequency spectrum is inversely proportional to its line width. Thus to get a **narrow** line, we will **maximize** the peak height!
- Set the number of repetitions to 100 and begin scanning. The frequency spectrum at the top right will begin updating with each scan, enabling you to monitor the peak height as you shim. Note that there is a check box for autoscaling the frequency spectrum; this feature can be

helpful as long as you pay attention to the vertical scale on the y-axis, but you may also wish to disable autoscaling by unchecking the box.

- In the GUI, change the current offsets (mA) of the x, y, z gradient coils to apply linear shim fields along each direction. As you do this, try to further increase the magnitude of the frequency peak (i.e. make the line as narrow as possible). Save your best shim settings using the “Save shim settings” button.

Record the currents (mA): X shim: _____ Y shim: _____ Z shim: _____

- Re-center the frequency if necessary. Record the new frequency: _____ MHz

3. One-dimensional imaging

3.1 Acquire spin echoes

- Stop any ongoing scans in the FID_GUI and close this window. Now open the “Spin echo GUI” through the software interface.
- Enter your optimal f_0 , transmit amplitudes, and shim settings (use the “load shims” button) into the GUI. Make sure the acquisition type is set to “Spin Echo”. Click “Run Scan”. See if you can spot the following on the graph that appears at the upper left:
 - a. At what time point is the 180° RF pulse applied?
 - b. At what time point does the spin echo occur?

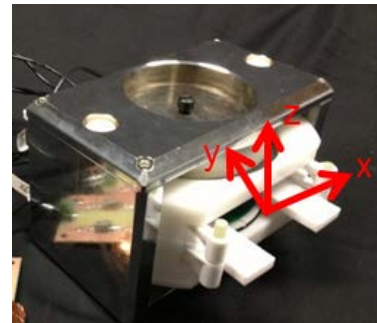
3.2 Acquire projection image

- You learned that magnetic field gradients enable the mapping from space to frequency that enables magnetic resonance to be used for imaging. Turn on gradient currents (and therefore add a gradient field) by choosing “projection along X, Y, or Z” as the “acquisition type”. The frequency domain signal is now a **projection** of the phantom onto the gradient field.

Check yourself: Look at the signals before and after turning on the gradient field. When you turn the gradient on, the frequency domain signal should look wider. This is because the gradient causes some spins to precess off-resonance, so the field distribution (and therefore the frequency distribution) becomes much wider when the x-gradient is on.

What happens to the time-domain signal when you turn the gradient on? Does this make sense?

- Stop any ongoing scans, turn on the X gradient using “projection along X”, set the repetition number to 100 and restart scanning.
- Open the MRI box and gradually rotate the phantom until you see the two projections collapsed into a single hump in the frequency display on the GUI. Define this as the 0 degree position.
- Stop the scanning. Without moving the phantom, turn on the Z gradient with “projection along Z,” and then restart. Now, the two projections should be maximally separated. Why do you think this is?



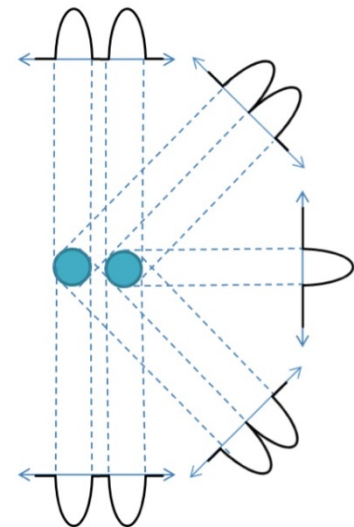
Gradient field directions in magnet

4. Acquiring and reconstructing 2D images

We will acquire a 2D image of the two-cylinder water phantom, using a readout sampling (dwell) time of $32 \mu\text{s}$ and a phase encode blip duration of 2 ms. We will obtain a 2.56 cm field of view (FOV) for our image and 128×128 points (matrix size) in our 2D image.

- From the software interface, open “IMAGING_GUI_1”.
- Enter your optimal f_0 , transmit amplitudes, and shim settings (use the “load shims” button) into the GUI.
- Enter the 128×128 matrix size if necessary, as well as the gradient values 28.7 and 29.4 mT/m for X and Z, respectively. Then run the scan and wait for it to build up.
- The output data are accumulated in a 128×128 matrix that holds the acquired complex valued k-space (spatial frequency) data. After the scan is done, save this data with an appropriate filename.
- To see the image, we will go out into the main MATLAB GUI. From the MATLAB command line, type the following lines:


```
load <filename>; % <filename> is the name you chose in the previous step
image = ifftshift( ifft2( ifftshift( data ) ) );
figure;
imagesc( abs( image ) );
```
- Can you see the tubes you imaged?



Projection images of two tubes of water seen from different directions.

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5. Three dimensional imaging

We will use the MIT Phantom, which has three layers of plastic structures printed with the letters as seen in the picture. We will image it using the 3D imaging GUI. In 3D imaging, the k-space matrix is now 3D. This sequence fills in one of the k_x , k_z planes per shot (TR), and then steps through the phase encodes in the k_y axis.



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- Open the 3D imaging GUI by typing “IMAGING_GUI_3” into the command window.
- Insert the MIT phantom. (You may need to check that f_0 is still correct, and the phantom is centered in the y-gradient.)
- Set the number of “slices” to 16 (matrix dimension in the y direction). This will give us 16 slices along the imaging FOV. If the FOV is 3 cm, that should cover the full length along the tube seen by the receive coil (detector), and give slices with ~ 1.875 mm thickness. Hopefully, you will see the individual letters in the MIT phantom more clearly, since they will no longer be collapsed.
- Try some different FOV’s and some averaging to improve the SNR of your 3D image. Use your newly acquired imaging expertise to try to get the best possible image.

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