6.S02 MRI Lab: Pre-Lab 1

--L.L. Wald, April 2013

1. Introduction to Magnetic Resonance Imaging

In a 2001 survey (Health Affairs 2001 20(5), p.30), Fuchs and Sox asked U.S. physicians to rank recent medical technologies by their importance to medical practice. The list covered a wide range; life-saving pharmaceuticals such as ACE inhibitors, Statins, and Tamoxifen, less-than-life-saving but popular drugs such as Viagra, surgical procedures such as balloon angioplasty, gastrointestinal endoscopy and coronary artery bypass and medical devices such as replacement hips/knees. You guessed it; the top ranked innovation was 3D and cross-sectional imaging by MRI and CT. Viagra was #28. Although occasionally disparaged in the healthcare debate as over-used and expensive (both true), MRI has changed the way medicine is practiced. Just like Dr. House cant get thru an episode without an MRI, a surgeon would never cut before having a detailed "look inside", removing the phrase "exploratory surgery" from the medical lexicon in less than a generation. Thus, the ability to "see inside" and plan treatment improves patient's outcomes (and lowers cost) simply by allowing better use of existing procedures and drugs.

MRI is particularly prized for the multiple ways it can distinguish soft tissues based on subtle differences in water content and environment. Unlike x-ray based methods such as Computed Tomography (CT or CAT scan), MR uses no ionizing radiation and can detect multiple physical properties of the tissue (rather than just X-ray absorption). It employs magnetic fields and radio waves using principles covered in freshman physics to detect small changes in the magnetic properties of the water molecule due to its environment.

Although the background of those advancing MRI technology varies from physicists to physicians, the principle developers of the technology at the commercial R&D level are electrical engineers. To the engineer, MRI technology offers a rich playground of electromagnetic design, Radio Frequency engineering, high power analog circuitry, and signal and image processing. In this laboratory, we provide 10 stations with "table-top" MRI scanners capable of scanning small objects (1cm diameter or less) that are controllable through MATLAB programming and will acquire and manipulate the signals ultimately producing an image. Goals of the laboratory are to understand the basic phenomena and image encoding methods employed in MRI, and the engineering principles behind them.

The proton in an external magnetic field

The motions of the nuclear magnets in a strong external B field are at the core of MRI. When placed in an external magnetic field (the big magnet) the tiny bar-magnet of the proton processes like a top, with the frequency of procession proportional to the strength of the external magnetic field. We measure the frequency of the procession (and thus the local field) with high accuracy using a simple radio receiver. Add some clever tricks to encode the position of the water by making the magnetic field vary linearly in space, include some basic principles of signal processing and you soon have an image of water content in the body. Exploiting minute alterations of the phenomena based on other physical properties allow us to map everything from water flow and/or diffusion, to temperature, to mechanical stiffness of the tissue. It is this rich variety of variations on a theme that makes MR research an active area today.

<u>Static nuclear magnetism</u> The nucleus of the Hydrogen atom is a single proton, and given the two Hydrogen atoms in every water molecule, it is the most abundant nuclei in the body. The proton is stable (not radioactive) has a non-zero radius and carries a positive charge. It also contains intrinsic angular momentum, $\vec{l}$, called spin angular momentum meaning we can roughly think of it as a spinning charged ball. Like all moving charges, the small currents on the spinning proton also generate a magnetic field in the pattern that a spinning, charged ball would. Many proton radius' away, the pattern is identical to that around a small current loop or small bar magnet, i.e. a magnetic dipole pattern (see figure). Since the magnetic dipole is an oriented entity, it can be described as a vector. Since $\vec{m}$ is essentially created by the angular momentum $\vec{l}$ and the charge, the two are co-linear and proportional to each other; $\vec{m} = \gamma\vec{l}$, where γ is called the "gyromagnetic ratio". Note that given $\gamma = |\vec{m}|/|\vec{l}|$, It should have been termed the "magneto-gyric ratio". One can show that the gyromagnetic ratio for a spinning uniformly charged ball is determined only by its charge, q. and mass, m; $\gamma = \frac{q}{2m}$.

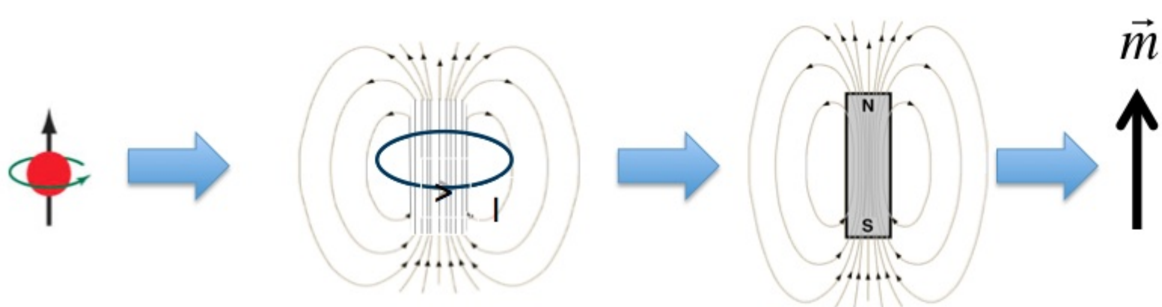


Figure 1. From Left to Right; the proton as a spinning "ball" of charge with its angular momentum vector. The magnetic field pattern produce by a small current loop or a small bar magnet. The vector representation of a small magnetic dipole.

The magnetic moment, angular momentum and gyromagnetic ratio are intrinsic and constant properties of the proton. Note also that $\vec{m}$ has units of current · area. Current has SI units of [C/s] thus $\vec{m}$ has units of [C m^2/s]. Angular momentum has units of [kg m^2/s] giving γ units of [C/kg]. Note; the magnetic field unit of 1 Tesla is defined as the field in which a 1 Coulomb charge moving at 1m/s experiences a force of 1 Newton suggests that Tesla can be expressed in units of [Ns/C m]. Using [N] = [kg m/s^2] from F=ma shows that [C/kg] is equivalent to [T^{-1}s^{-1}] or more commonly as [Hz]/[T]. For the proton, $\boldsymbol{\gamma = 42.577\frac{MHz}{T}}$. This will be the only fundamental nuclear constant that we will need to know.

When a magnetic dipole is placed in an external magnetic field, the north end of the dipole tries to align with the field vectors (think compass needle in the Earth's field). The energy of the system can be expressed as: $U = -\vec{m} \cdot \vec{B}$ and is thus lowest in the aligned state. A cubic millimeter of the body contains ~6x10^{18} water protons. The net magnetization in this macroscopic ensemble is $\vec{M} = \sum \vec{m}$. We will derive a measureable signal from this net magnetization, but its important to note that even with strong aligning fields, most of the $\vec{m}$ end up cancelling one another; i.e. they are nearly randomly oriented due to the thermal motion and small energy of alignment. Nonetheless, M does add up to a small macroscopic magnetization that we will observe (and image the distribution of); $|\vec{M}| \cong 10^{-4} \sum |\vec{m}|$. Similarly, the ensemble of proton angular momentum vectors, $\vec{l}$, add up to a macroscopic quantity $\vec{L} = \sum \vec{l}$. We use the capital letter for the macroscopic quantity and the small letter for the microscopic. We will always work with macroscopic ensembles of billions of spins that on average behave under the laws of classical physics, even though the individual proton spins behave quantum mechanically. This is a general principle of quantum physics; quantum laws must reduce to classical laws when a macroscopic ensemble is considered. Finally, note that this is an unusual form of magnetism: it arises from the magnetic properties of the nucleus. Magnetism arising from the electron (and electron orbitals) such as Ferromagnetism, Diamagnetism, and Paramagnetism are much more common (and much stronger).

Nuclear magnetism in motion The weakness of nuclear magnetism makes it difficult to detect unless it is moving. A moving magnet can be detected through the Faraday (generator) effect since a moving magnet in a coil of wire generates a voltage across the coil. We will observe the nuclear magnetism in this way, but first must figure out how it moves (its equations of motion) and get it started. Even to have a net (non-zero) magnetic moment, $\vec{M}$, we must apply an external field to create a net alignment of the individual moments, $\vec{m}$, so that is step 1. This external field exerts an alignment torque which will rotate the magnetization into the low energy state (aligned with $\vec{B}$) just as the earth's field torques a compass needle into alignment with the earth's field. Note this one-time motion is too slow for us to detect directly via the Faraday effect; we will see that there is a periodic motion we can establish which lasts for thousands of cycles.

The force on a dipole is $\vec{F} = -\vec{\nabla} U = \vec{\nabla}(\vec{m} \cdot \vec{B})$. Note that there is no net force on the dipole if B is uniform in space. There is, however, a torque on the dipole. Figure 2 shows the forces on a small square current dipole. Note the forces cancel out, but still create a net torque on each current element which tries to align the dipole moment (which points normal to the plane) with the applied field.

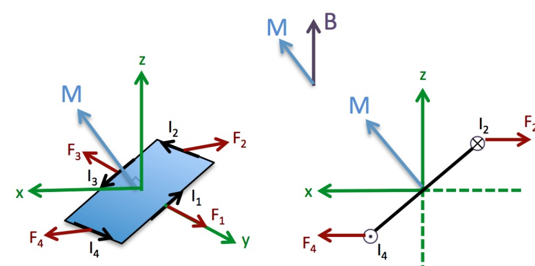


Fig. 2. A square current dipole in a magnetic field along z. There is no net force, but the torque (r x F) tries to align the dipole vector M, normal to the current loop) to the z axis.

$$\vec{\tau} = \vec{r} \times \vec{F}$$

The total torque calculated about the origin is of magnitude $2rF_2$ where r is the distance from the origin to I_2. The force, F_2 is of magnitude $I_2B(2r)$, where 2r is the length of the side. Then the torque ($\vec{\tau} = \vec{r}\times\vec{F}$) is of magnitude $4r^2IB\sin(\alpha)$ where α is the angle between M and B. Using M = I · (area) suggests $|\tau| = |M|B\sin\alpha$ Note that torque produce rotations, and have a funny direction assigned to them because we describe the angular rotation of a wheel as a vector pointing along the axis of rotation (and its sign given by the sense of rotation and the right hand rule). Thus we recognize the torque as:

$$\vec{\tau} = \vec{M}\times\vec{B}. \qquad \text{Eq. 1}$$

<u>Equations of motion:</u> Just like a force results in a change in linear momentum; $\vec{F} = d\vec{p}/dt$, a torque produces a change in angular momentum;

$$\vec{\tau} = \frac{d\vec{L}}{dt} = \gamma^{-1} d\vec{M}/dt. \qquad \text{Eq. 2}$$

The angular momentum also points along the axis of rotation. A torque with the same vector direction as the wheel's angular momentum causes the wheel to speed up (and if opposite, to slow down). Combining Eq. 1 and 2 gives the equation of motion for the magnetization:

$$d\vec{M}/dt = \gamma\vec{M}\times\vec{B} \qquad \text{Eq. 3}$$

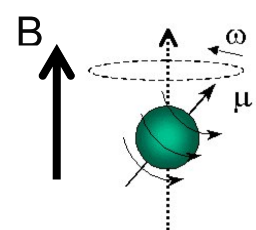
Note that since the cross product is always a vector perpendicular to M, thus the magnitude of M does not change. Instead it is simple to solve this differential equation by breaking it down into an equation for each component using $\vec{B} = B\hat{z}$;

$$\frac{dM_x(t)}{dt} = -\gamma M_y(t)B$$

$$\frac{dM_y(t)}{dt} = M_x(t)B$$

$$\frac{dM_z(t)}{dt} = 0$$

These are called the Bloch Equations without relaxation. The general solution is easy to guess:



$$M_x = -M_x(\text{t=0})\cos(\gamma Bt)$$

$$M_y = M_y(\text{t=0})\sin(\gamma Bt)$$

$$M_z = M_z(\text{t=0})$$

Here, M_x(t=0) is the initial value of the x component. We often call the total vector sum of the 3 components at t=0, M_0. This is the total starting magnetization. These equations of motion describe a spinning top whose axis of rotation precesses with a frequency ω around the B direction (taken to be z here). Note that M_z(t) is constant and will not contribute to our signal since our signal will be generated via Faraday induction (the generator effect) which requires a moving magnet. Therefore, we often keep track of the x,y component of M separately, since this is the magnet in our generator and its amplitude will be proportional to the signal voltage generated in the coil and its phase will be the phase of the AC voltage detected. This M_{xy} is a vector rotating in a clockwise sense (if γ is positive) in the xy plane and can thus be conveniently expressed in complex notation, where the real part is the x component and the imaginary is the y component.

$$\vec{M}_{xy} = M_{xy}(t = 0)e^{-j\omega t}$$

Here, $\omega = \gamma B$. We call the main static magnetic field of the magnet, the B_0 field, and always define its direction as the $\hat{z}$ direction. In this case, $\omega_0 = \gamma B_0$ is often called the "Larmor Frequency". And it's a little confusing in that earlier I wrote γ = 42.577 MHz/T which refers to a frequency in cycles per second. Here ω is an angular frequency (in Radians/s) so I will update gamma to:

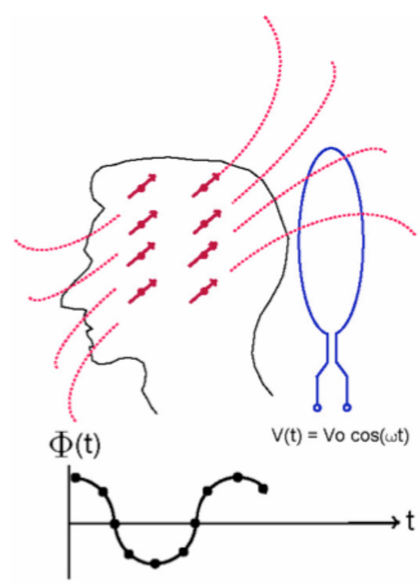
γ = 2π 42.577 radians/(s T).

In our tabe-top scanner, B_0 = 0.19T and $\omega_0 = 2\pi\ 8.1\times10^6 \frac{radians}{s\,T}$, a frequency of 8.1 MHz.

You can think of the analogy with a top like this. The top has the force of gravity pulling it down. This force tries to change the angular momentum, but the torque (r x F) is orthogonal to the angular momentum vector (which is parallel to r) therefore it can't change its magnitude but does change its direction, torqueing the vector around in circles around the z axis. In the spinning proton, the angular momentum vector plays an identical role, but instead of gravity torqueing it down, we have the external magnetic field torqueing it up (toward the z axis).

Detection Now we know how the net magnetization moves, how do we detect it. We detect it thru the Faraday Effect (Generator Effect). Faraday's Law tells us that a changing magnetic flux over a loop of wire induces an electro-motive force (EMF) across that loop which can be used to power electrical circuits (in fact this effect is at the heart of every power-plant and is used to power cities).

$$V(t) = -\frac{d\Phi}{dt}$$



Where Φ is the integral of the magnet's field lines over the surface of the wire loop. The time derivative guarantees that only moving magnets generate electricity and also that it will be AC electricity. In our case: $\Phi(t) = A\,\vec{M} \cdot \hat{n} = M_{xy}(t = 0)e^{-j\omega t}$

Where A is the area of the loop and $\hat{n}$ is the surface normal to the loop and we have assumed the coil is oriented so that the surface normal is in the xy plane. Then the precessing magnetization creates a flux that is oscillating at the frequency, ω. The detected voltage is:

$$V(t) = j\omega A\ M_{xy}(t = 0)e^{-j\omega t}$$

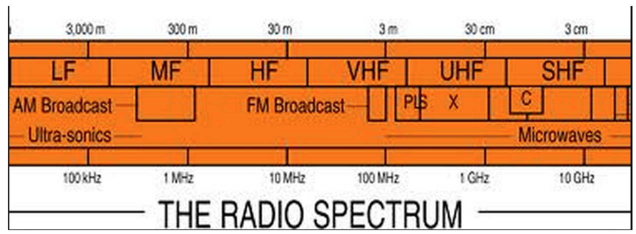
Thus we see that we get a voltage signal proportional to the magnetization. It's an AC voltage oscillating at the Larmor Frequency. For our desk-top MRIs, with a magnetic field of ~0.19T, this $\upsilon = \frac{\omega}{2\pi} \approx 8.1MHz$.

Will the signal last forever? Nothing lasts forever. We know that the low energy state requires the protons to align along the z axis (along the applied B field). In this case there is no component in the xy plane, only a stationary z component and thus no signal voltage. Thermodynamics tells us the system will eventually rest in this low energy state by giving up thermal energy to its surroundings. But it doesn't tell us whether it will get there in 1 us or 10^{10} years. This depends on how tightly coupled the magnetization is to "the world"; how strongly it exchanges energy with its surroundings. It turns out the magnetization returns to the z axis with a time-constant of about 1s. We call this return to equilibrium time constant T_1. Thus we expect our signal to decay at least this fast. There is a second method to loose the magnetization. Recall that the net magnetization M is made up of ~10^{20} individual protons. What if they do not experience the same field, B, but each are in slightly different fields? How would this happen? The simplest way is if the main magnet is not uniform over the sample. If the magnet is non-uniform to 10ppm over the sample in our 0.19T magnet, then the spread in frequencies will be 80 Hz and the fastest spins will pick up 2π radians of phase relative to the slowest ones in τ = 1/80 s = 13ms. Thus the signal will be decay significantly in about 13 ms. This is about what happens in our magnets; the signal decays approximately exponentially with a time constant of about 10ms. We call this time constant the T_2^* decay rate. We call the oscillating signal voltage with its decaying exponential envelop the "Free Induction Decay"; FID.

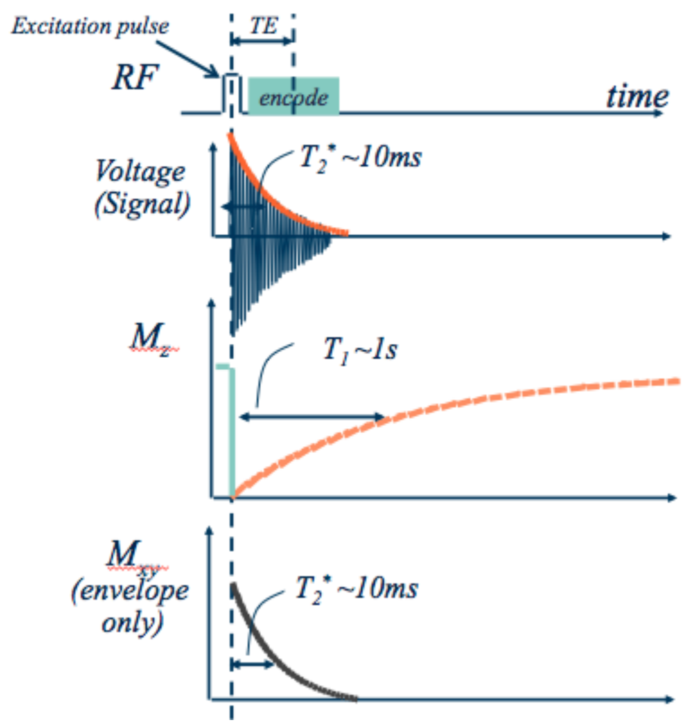
Excitation To see a signal, the spins must have a moving component; a component in the xy plane ($M_{xy}(\mathrm{t{=}0}) \neq 0$). But we also know that the aligned state $M_z = M_0$ and $M_{xy} = 0$ is the thermal equilibrium state that we are likely to find the system in. How do we tilt the spins away from the z axis to get them going? One way would be to allow them to align (achieve equilibrium) and then suddenly turn the magnet (or the sample) by 90 degrees. If this were done non-adiabatically (which means quickly compared to the time constant of energy exchange and comes from the Greek; "doesn't walk"), then the magnetization would suddenly find itself oriented orthogonal to B. It would "wake up" and start precessing. This needs to be done in about 1ms for our system. Even for the small magnet this would be difficult to achieve, corresponding to an angular rotation rate of the magnet of $\pi/2\,\frac{radians}{1ms} \approx 1570\,\frac{rad}{s}$. It is, of course, even harder for a 10 ton human magnet. Rotating the patient this fast isn't a great idea either.

Pre-Lab Problem #1; Estimate the g-force experience by the patient rotated at the rate sufficient to non-adiabatically induce a M_{xy} component.

A smarter way uses a second magnetic field B_1, in the xy plane to torque the magnetization away from the z axis. We will only describe this qualitatively. This field rotates M away from the z axis. Imagine we succeed in rotating the vector down by 1°. Then it will start precessing in a tight little 1° cone about the z axis. If we leave this field stationary, after a half-cycle of precession it will torque it back up instead of torqueing it down more. Thus if the torqueing process is not fast compared to the precession frequency, we need for this field to move and keep pace with the rotating vector so that it always pulls it down. What is needed is a B1 field in the xy plane that exactly tracks the magnetization, always torqueing it down the x axis. If the rotational frequency of this field does not exactly match the Larmor frequency, then over several thousand cycles (or less) it will get out of sync and start un-doing its work. This is the resonance phenomena; the prescription for adding energy to any resonator which has a natural frequency. The perturbing force must have the correct frequency and phase so that it adds energy over every cycle. If they get out of sync, then it will start subtracting energy from the resonator. Think about pushing a child on a swing. The applied force must have the proper frequency and phase. If not it gets out of sync and starts opposing the motion of the child. This is where the “Resonance” in the name “Magnetic Resonance Imaging” comes in.



Since the correct frequency is the Larmor frequency, we have supplied a small solenoid coil around the sample that we can drive with RF current to create an oscillating field at the resonant frequency. Recall the Larmor frequency is 8.1MHz, this is in the HF radio region of the electromagnetic spectrum, near the popular 40meter wavelength HAM radio band (40m corresponds to 7.1MHz). Our exact frequency is in a Maritime mobile band (apologies to ship captains in Boston Harbor…) Anyway, we will apply a burst of current to this coil to torque the spins from the z axis, to the xy plane. This 90deg rotation can be done in about 0.2ms with about 1W of RF power provided by the so-called RF power amplifier. On a commercial MR, the volume is ~10^4 x bigger (1m samples instead of 1cm), and about 10kW is needed.



We can control where we leave the magnetization by controlling either the amplitude or durction of the RF pulse. If we tip it 90 degrees in to the xy plane, we call it a “90 degree pulse”. If we choose to leave some component along the z axis we can tip it less. If we tip it 180 degrees, there is also no M_{xy} and no signal.

The Figure to right summarizes the scheme so far. I have included a green box labeled “encode” to denote where we plan to encode the image.

FID and Spectrum Since the precessing magnetization induces a decaying AC voltage at the Larmor frequency across our coil, we can digitize this into a discrete vector; x[n]. The UI for the scanner controller will allow you to set a readout duration and the sampling interval (dwell time) is fixed and is related to the bandwidth (BW = 31250Hz). It does not tell you the number of points recorded, N. You can determine this by either saving and reading the complex vector saved into MatLab and interrogating its size or by noting that the bandwidth, dwell time relationship and the readout length. Thus to obtain a readout time of 25ms requires N = 781 pts.

When x[n] is the recorded time domain signal (i.e. the FID), and X[k] is the frequency domain discrete signal. The two are related by the DFS:

$$\boldsymbol{X[k]} = \frac{1}{N}\sum_{n=0}^{n=N-1} x[n]\, e^{-j\frac{2\pi}{N}kn} \qquad\qquad x[n] = \sum_{k=-\frac{N}{2}}^{\frac{N}{2}-1} X[k]\, e^{j\overbrace{\underbrace{\frac{2\pi}{N}kn}_{\Omega}}^{\Omega_0}}$$

In this case, n is the integer index for time, and x[n] is our Free Induction Decay, and k is the integer index for temporal frequency and X[k] is the spectrum.

Demodulation Since The scanner does not actually digitize the full FID with its oscillations at 8.1MHz (the Larmor frequency), but the signal with the 8.1MHz carrier removed. For example, assume the FID is:

$$x[n] = C_0 \exp(-j\ \Omega_0 n)\exp(-n/T_2^*)$$

where C_0 is a constant (the initial signal voltage amplitude) and T_2^* is a constant (the decay constant expressed in the units of time samples) and Ω_0 is the Larmor frequency. Imagine that we had some way of multiplying this signal by another analog signal, exp(j Ω_0n), before we even digitize it. The result would be x'[n] = C_0 exp(-n/T_2^*) which is just the signal with the rapid Larmor frequency oscillations missing. This is the M_{xy} with "envelope only" plot show in the figure above. This process is called "removing the carrier" or homodyne reception or homodyne demodulation and is what a simple radio tuner does in analog circuitry. The exponential that we multiply the real and imaginary components by are sine and cosine waves from a local oscillator and the multiplication process is done electronically by a mixer. Alternatively, we could do this all in the digital domain by recording the signal fast enough (sample at least at 2x Ω_0) and then multiply it by the complex exponential digitally.

What happens if we don't know the Larmor frequency exactly and end up multiplying by exp(j (Ω_0 + $\Delta\Omega$) t)? Then the resulting signal is simply x'[n] = C_0 exp(-j $\Delta\Omega$ n) exp(-n/T_2^*). This is the same sort of signal but with slower oscillations ($\Delta\Omega$ instead of Ω_0). The oscillations are the different between the Larmor frequency and the applied demodulation frequency. On our scanner we can set the frequency that we think is the Larmor frequency into the GUI. The console GUI will display the demodulated FID. If we get the frequency exactly right, there will be no wiggles. and the spectrum of the FID, X[k] is centered in the window.

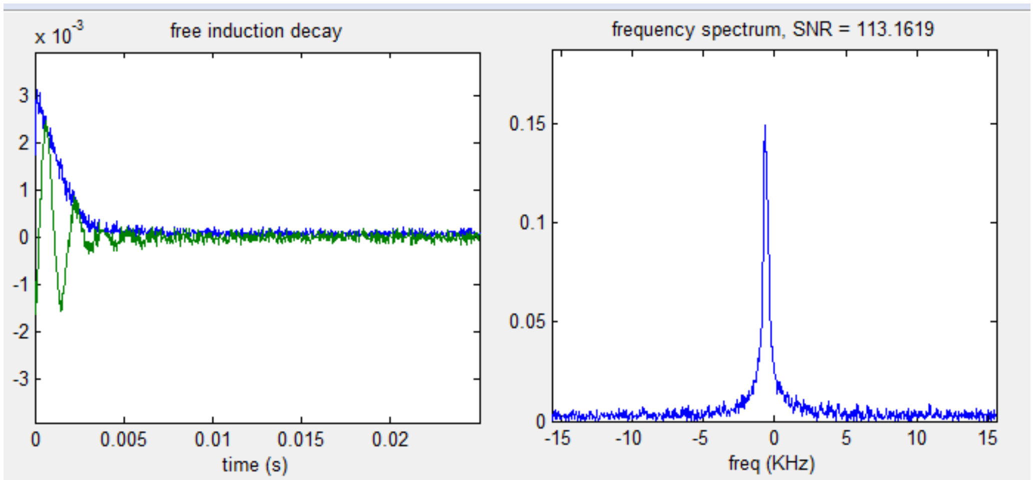


Fig. 7 The FID (left) and spectrum (right). The green trace of the FID shows the wiggles that result from the difference between the demodulation frequency and the Larmor Frequency:

$(x'[n] = C_0 \exp(-j\, \Delta\Omega\, n) \exp(-n/T_2^*))$.

The blue trace of the FID is the magnitude of the FID. The spectrum is also the magnitude of the spectrum. The offset of the spectrum (~2kHx) is $\Delta\Omega$.

PreLab problem #2: how many points are expected in the FID is BW = 31250Hz and the readout length is 25ms?

PreLab problem #3: When the spins dephase and the signal decays away, it tends to do so with an exponential decay envelope as described above. What will be the shape of the resulting spectrum. Plot X[k] for a signal; $x[n] = C_0 \exp(-n/T_2^*)$.

Another alternative is a Gaussian function;

$$x[n] = C_0 \exp(-n^2/4\sigma^2)$$

What does X[k] look like for this?

<u>Signal to Noise Ratio</u> You will be asked to measure the Signal to Noise Ratio (SNR) of your signals and images. SNR is a useful experimental metric that can be defined a number of ways. The basic point idea is to measure the ratio of the amplitude of the signal to the amplitude of the noise. Signal can be parameterized many ways; we usually take the peak voltage magnitude (for FID) or the peak amplidued of the spectrum. In other fields you might measure a waveforms rms, or mean. It might be either a voltage signal (which is the meaningful one for MRI) or a signal power. For the FID, we usually define its amplitude as the amplitude of the largest point in the timeseries x[n]. Noise can also be characterized a number of ways. For Gaussian white noise (such as ours) we will characterize the noise distribution by computing the Standard Deviation (SD) in a subset of the time series that contains no signal. For the FID, the last 100 points is usually safe since the signal decays away (but check).

$$\sigma = \left(\frac{1}{N}\right) \sum_{n=0}^{N-1} (x[n] - \bar{x})^2$$

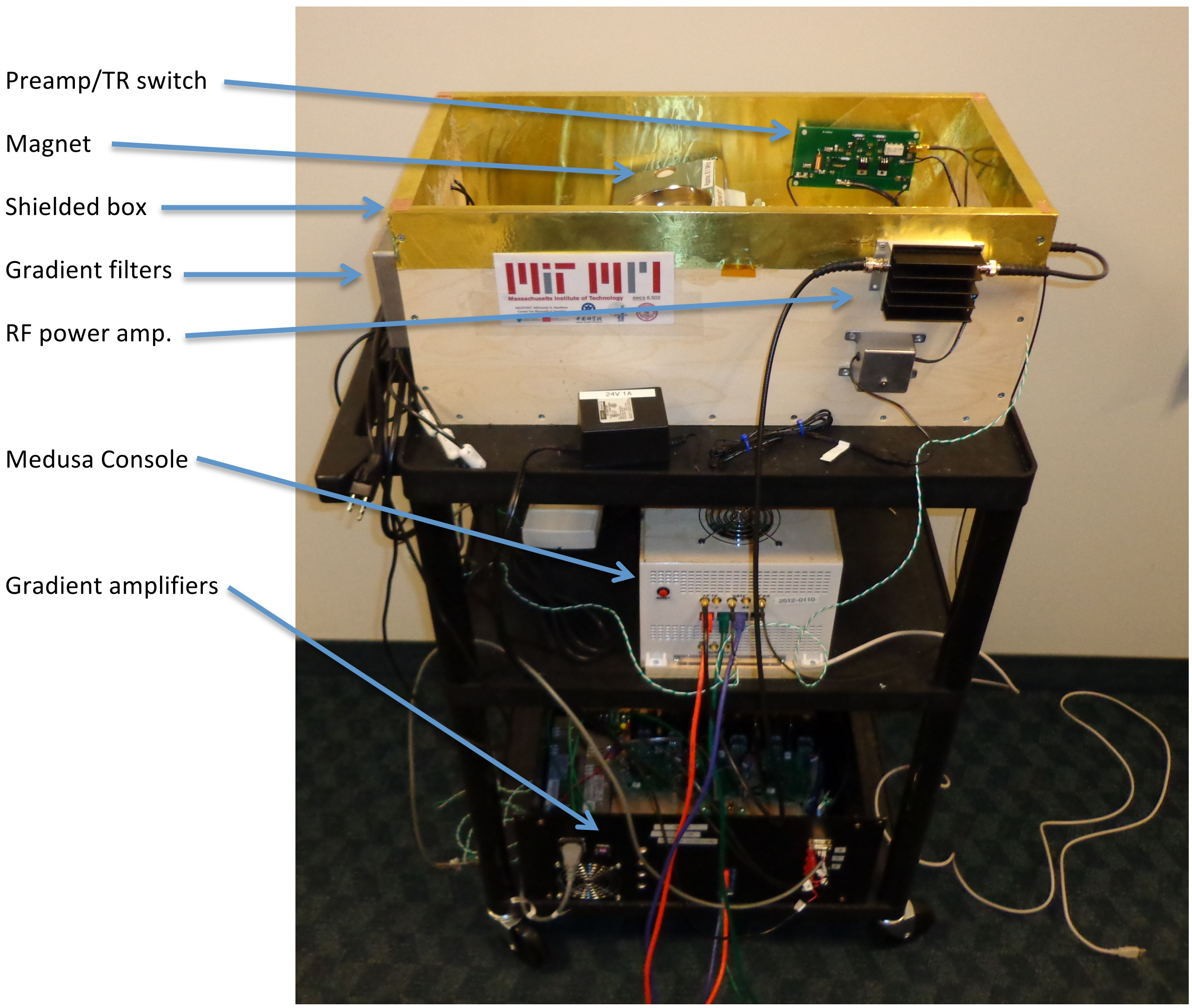
where $\bar{x}$ is the mean and N is the number of points you are summing over. For complex data (such as ours); $\sigma = (\frac{1}{N}) \sum (x - \bar{x})(x - \bar{x})^*$. If you decide that there is signal present in all of the FID, you can always acquire more data points (at the same bandwidth) such that its sure to be gone. Another trick is to reduce the signal (without changing the noise) by decreasing the amplitude of the RF excitation. With 0V excitation, there is no signal

We will also ask you to measure the SNR of the computed spectrum. You can do this the same way; the amplitude of the spectrum defined as the peak amplitude and the noise is the SD of a spectral region with no signal.

About your Table-Top MRI…

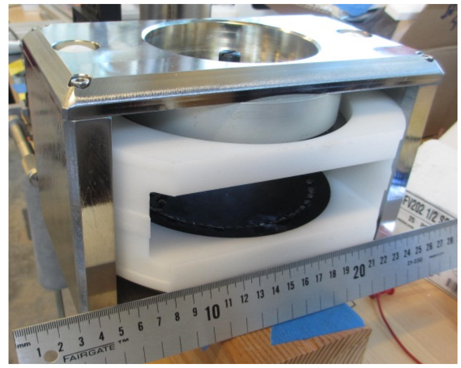
The system you are about to use came together across 3 continents and was the work of at least 20 people. It was coordinated and assembled by the MGH Martinos Center MR Physics group with key components coming from other MR groups around the world. The rare-earth magnet was designed and constructed by the group of Prof. Wenhui Yang at the Institute of Electrical Engineering of the Chinese Academy of Sciences in Beijing (http://english.iee.cas.cn/rh/rd/200907/t20090722_24888.html). The scanner's Medusa console has been a >5 year development project by Greig Scott and Pascal Stang at the Stanford EECS Dept (http://mrsrl.stanford.edu/~medusa/hardware/). The gradient coil current contours were calculated by Maxim Zeitsev and Feng Jia of Freiburg Univ. 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.

System Orientation



The System components

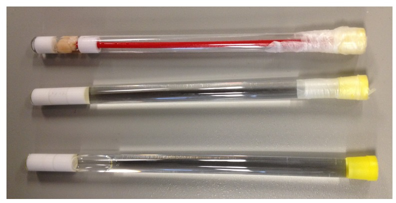
1.1 Magnet



The B_0 field is created by a small 0.19T 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).

1.2 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.

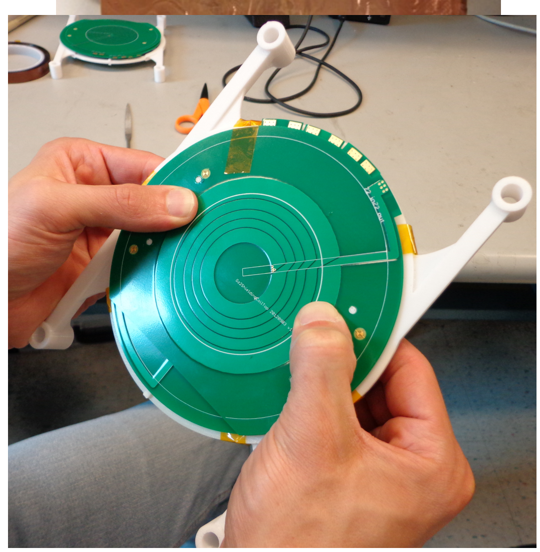


1.3 RF coil

The RF coil for this system is used for transmitting the excitation pulse to get the magnetization precessing and to detect the MRI signal thru 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 which the NMR tubes fit into snuggly; it is contained in a plastic case which is wrapped in copper foil to shield against external RF noise sources.

1.4 Gradient coils

The gradient coils generate the linear gradient fields that are used for imaging. They produce a magnetic field in the z direction which changes linearly with position with a slope of14G/cm when driven with 1 amp of current. With 2 A, the slope is 28G/cm so we refer to the sensitivity of the gradients as 14G/(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/exciter coil.



1.5 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.

1.6 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.

1.7 Console

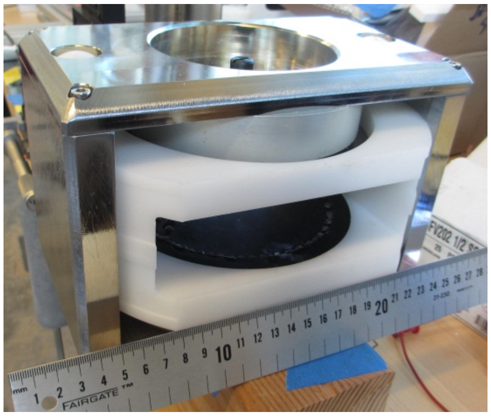
The console interfaces with the computer via MATLAB. It produces the RF transmit pulses and gradient waveforms based on vectors created in a MATLAB sequence. It also acquires the received MRI signal at a time specified in the sequence. The console samples the received signal and downconverts it to baseband. Some specifics are at: http://mrsrl.stanford.edu/~medusa/hardware/

1.8 MATLAB GUI

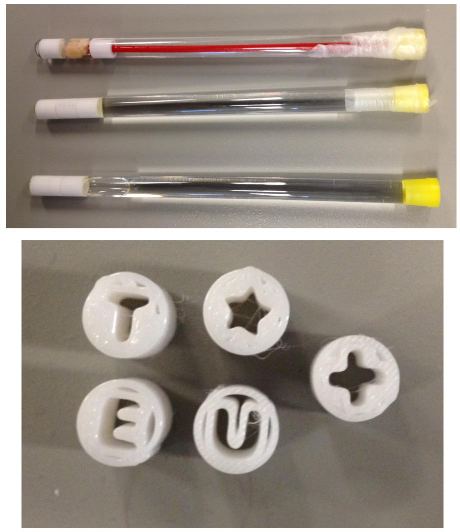
The MATLAB GUI is what you will use to edit and run sequences. You may run an FID sequence or a spin echo sequence. You can control the amplitude of the RF pulses and gradient waveform. You will also control the repetition time (TR), echo time (TE), and the read out time.

Appendix

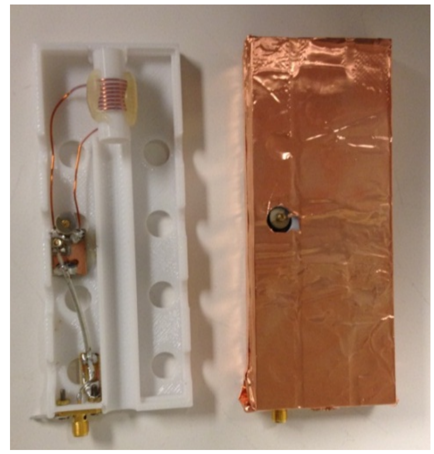
1 MRI System Components



B_0 Magnet



Imaging Phantoms

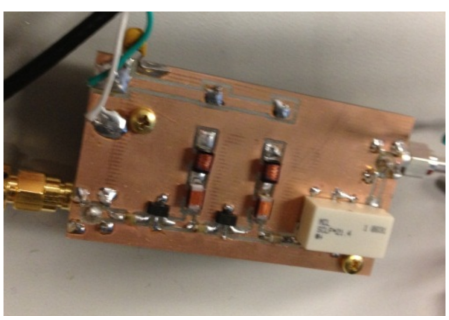


RF Coil and Enclosure

Gradient Coil PCB

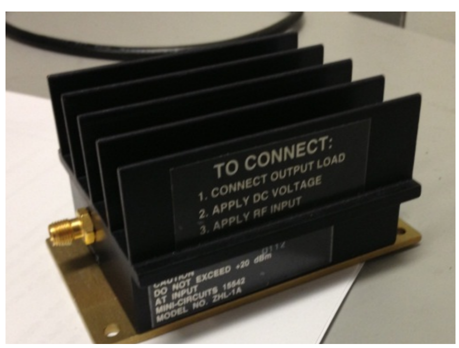
Gradient Amplifiers

Console



Receive Amplifier

Transmit/Receive Switch



Transmit Amplifier