Physics Tutorial 4: Image Distortions

Originally Written by James Meakin and Alex Gardener
Last Modified: Mark Chiew, Robert Frost, Nadine Graedel, Alan Stone, Tom Okell, Oct 2014

The purpose of this tutorial is to provide an understanding of how B0 field inhomogeneity causes distortion in EPI images, and how acquisition parameters can be chosen to acquire higher quality data. Questions without any asterisks are those that should be attempted by everyone, whereas more challenging questions are marked with one (*) or two (**) asterisks, and should be considered optional. Take these opportunities to think about these questions, and then discuss your answers with your tutor.

At the end of the tutorial period, you will receive a "take-home tutor", which is an annotated version of this tutorial guide that will help you complete the tutorial at home if you don't manage to make it all the way through with your tutor, or if you missed the tutorial session.

Attendance will be taken by the tutors, and marks will be given by participation. If you would like additional feedback or clarification on the tutorial material, you are welcome to submit your questions or comments to Weblearn, and a tutor will provide written feedback for you. Those unable to attend the tutorial must submit answers to all unstarred questions to receive credit for the tutorial.

Part 0 – Getting Started

0.1 Starting MATLAB

Download and unzip the file containing the tutorial resources from Weblearn, or if you have access to the FMRIB internal network, copy them into your current directory from here:

~mchiew/GradCourse/4_Image_Distortions

Start MATLAB, and make sure you're inside the tutorial directory (i.e., the folder containing all the tutorial files).

Note to jalapeno00 users: please start with the -nojvm option, "matlab -nojvm"; *this should reduce server CPU load if lots of people are trying to use jalapeno00 simultaneously*

Part 1 – Distortion Simulator *[5 minutes]*

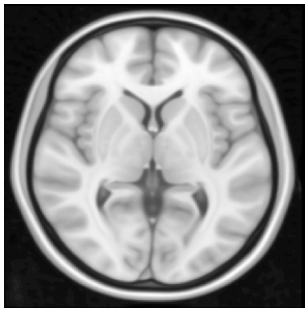

Figure 1

`simDistortion()` is a simplified simulator for the acquisition of echo-planar imaging (EPI) data. This simulation uses a slice of the MNI 0.5mm template (Fig. 1) as source data, and simulated the effects of magnetic field offsets at different spatial locations.

The options struct can be used to modify many aspects of the simulation (for a complete list of options, you can run "`help simDistortion`"). For example, to set field strength type:
```
>> options.FieldStrength = 3.0;
```

Any number of options can be combined to investigate different acquisitions and what we're actually simulating is outputted to the command line. If you need to reset the options at any point type:
```
>> clear options
```

1.1 Chemical Shift *[30 minutes]*

One of the most fundamental relationships in MRI is the correspondence between magnetic field strength and resonance frequency. There are two ways we can measure signals with the *wrong* resonance frequency:

- The magnetic field is not what it should be
- The resonance frequency is wrong because the signals are coming from non-water protons

The former case is often called *magnetic field inhomogeneity* (covered later in this tutorial), whereas the latter is called *chemical shift*. Fat molecules, for instance, contain protons just like water, but their resonance frequencies are slightly shifted (by 3.5 parts-per-million) relative to water protons. However, when signals are measured, it is difficult to distinguish between water that is in the right place, and fat that is in the wrong place, which results in chemical shift artefacts.

Launch the simulator – in the MATLAB command line type:
```
>> simDistortion();
```

Discuss the components of the output figure (top left: source brain, bottom left: GRE image, top right: field map, bottom right: EPI image), and discuss the fat shift in the PE direction (A/P).

Show by clicking on the field map that the frequency difference in the fat ring around the skull is 447 Hz. The field map shows differences relative to the main magnetic field (~128 MHz at 3 T).

The ppm shift between fat and water is 3.5 ppm (ppm = parts per million). The ppm notation is a B0 field invariant way of expressing differences in resonance frequency. Quickly show how to calculate the frequency shift in MATLAB at 3 and 7 T.

*Frequency of main field = $\gamma/2\pi$ * B0 = 42.577 * 10^6 Hz/T * 3 T = 127.7 MHz*

*Chemical shift frequency = $\gamma/2\pi$ * B0 * ΔCS_{ppm} = 127.7 * 10^6 Hz * 3.5 * 10^-6*

Chemical shift frequency = 447 Hz

The ppm measure is convenient because B0 frequencies are in MHz.

Question 1.1.1
Why do off-resonance spins shift in the Phase-Encode direction in EPI? Why isn't there (much of) a shift in the readout direction? Why does this not occur in the simple GRE image?

EPI has a low effective bandwidth in the phase encode direction as a consequence of acquiring all the phase-encode lines consecutively in a single shot. The time between these EPI "lines" or "echoes" is long (low PE bandwidth), on the order of 1

ms whereas the time between sampled points in the readout direction is much shorter (high RO bandwidth), on the order of 10 µs. The long time between sampled points in the PE direction allows differences in phase to build up between fat and water. We expect that all the signal is at the water resonance and any phase difference is caused by the position of the spins in the phase-encoding gradient. However fat has a different resonant frequency, so the signal from it is mapped to the wrong location. In the readout direction this shift is not noticeable because the shift is sub-voxel but in the phase-encode direction it is larger as shown in the example.

The measured frequency of a spin in a gradient G_x at position x, with field inhomogeneity ΔB_0 and chemical frequency shift δf_{CS} is (this is for a constant amplitude readout gradient but analogous expressions apply for the phase-encoding gradient)

$$f = \frac{\gamma}{2\pi}(B_0 + G_x x + \Delta B_0) + \delta f_{CS}$$

Rearrange for position:

$$x = \frac{2\pi f - \gamma B_0}{\gamma G_x} - \frac{\Delta B_0}{G_x} - \frac{2\pi \delta f_{CS}}{\gamma G_x}$$

- *1st term: RO gradient encoding of frequency to position*
- *2nd term: error due to B0 inhomogeneity*
- *3rd term: error due to chemical shift*

For a given FOV and bandwidth (or equivalently sampling time Δt) the gradient strength is chosen to satisfy this equation:

$$BW = \frac{1}{\Delta t} = \frac{\gamma}{2\pi} G_x FOV_x$$

Hence position error due to chemical shift can be expressed as:

$$\Delta x_{CS} = FOV_x \cdot \Delta t \cdot \delta f_{CS}$$

At 3T, with δf_{CS} = 447 Hz, and readout sampling time Δt = 10 µs (BW = 100 kHz) and phase-encode sampling time 1 ms (BW = 1 kHz)

- *RO shift = 0.0044 FOV_x*
- *PE shift = 0.44 FOV_x*

The separation between the fat and water resonance frequencies changes with the strength of the main magnetic field.
Simulate the effect of reducing the field to 1.5 T (from 3.0 T):

```
>> options.FieldStrength = 1.5;
>> simDistortion(options);
```

Question 1.1.2
How does changing field strength change the shift of the fat? How does the fat shift at 7T compare to 3T or 1.5T?
Higher field strength means there is a greater separation of frequency between fat and water (3.5 ppm, or 149 Hz/T). Therefore the spatial shift of the fat signal increases at higher field strength due to the larger difference in resonance frequency, assuming everything else is the same.

One solution to fat-related chemical shift artefacts is the use of fat saturation ("fat sat"), which destroys the magnetisation of protons in fat by applying an RF pulse at the fat resonance frequency.

Consider the following analogy:

- Imagine a race (scan) between neighbourhood cats (water), but to which unwanted dogs (fat) show up to participate. If you blow a normal whistle (RF excitation), all the animals start running and the dogs end up disrupting the race. Instead, if you first blow a whistle that only dogs can hear (fat-selective RF excitation), the dogs go off and they run first. Only after the dogs are gone (fat saturated), do you blow the regular whistle and run the race with cats only (water only image, no fat).

Simulate fat saturation and discuss the concept of removing fat signal prior to imaging. The above analogy may or may not be useful in making things a little more concrete.

```
>> options.FieldStrength = 3;
>> options.FatSat = true;
>> simDistortion(options);
```

Note to students that fat-sat does not always work in practice (usually because of field inhomogeneity). In this case, fat chemical shift artefacts may be evident in the output images.

Question 1.1.3*
Can you think of other ways of suppressing fat signals from your images?

1. *Spectrally selective pulses – the reason we see signal from fat is because we excite a slice with a range or bandwidth of frequencies. We can design "spectral-spatial" pulses that excite protons at the water resonance, but have low flip angle at the fat resonance (a null point on a sinusoidal plot of flip angle vs resonance frequency).*
2. *Refocusing slice-selection gradient reversal for spin echo sequences (see Gomori et al., Radiology 1988). If, in a spin-echo sequence, the polarity of the slice selection gradient in the 180° pulse is opposite to that of the excitation, then the slice-select chemical shift will be opposite, and fat refocusing will be minimised.*

1.2 Distortions [20-30 minutes]

In the last section, the simulated system assumed perfect shimming (perfectly homogeneous magnetic field). In reality, shimming a brain is difficult due to its shape and the presence of air-tissue boundaries. The simulation can additionally model the effects sinus cavities have on the magnetic field.

Add in some sinuses to our model by running:

```
>> options.AddSinus = true;
>> simDistortion(options);
```

Discuss the resulting images, noting how the tissue is now distorted in the phase encode direction.

Recall from the lectures how the direction of distortion changes with the phase encode direction:

- in orientation or angle of the phase encode direction, e.g. anterior/posterior or left/right
- in order of traversal of k-space, e.g. bottom to top (blip up) or top to bottom (blip down)

Demonstrate the effects of changing PE direction and blip direction:

```
>> options.PhaseEncodeLR = true;
>> simDistortion(options);
>> options.BlipUp = true;
>> simDistortion(options);
```

Discuss the fact that the PE direction is completely arbitrary. Not only can it be A/P or L/R, but any arbitrary angle.

Question 1.2.1
In this example, why would it be preferable to use blip-up rather than blip-down? What is a possible advantage to interleaving the blip-up and blip-down acquisitions?
In the blip up acquisition the tissue in the frontal lobe is distorted away from the brain, so avoids signal pile-up, which can reduce BOLD sensitivity (among other things) as this is difficult to correct. Interleaving blip-up and blip-down allows for an inherent correction of distortion and does not require a separate field map acquisition – it is therefore more robust to motion. Another advantage is that tissue piled up in blip-down is stretched out in the blip-up and vice-versa.

Question 1.2.2
Why do you think that Posterior-Anterior is the favoured PE direction, over Left-Right? Can you think of a case when we'd want to use a left-right PE?
A/P is favoured because distortions in each hemisphere are identical. This can be particularly important in applications like FMRI, where differences between unilateral and bilateral brain activity can have a significant impact on the interpretation of results. If we're particularly interested in one hemisphere around the sinuses we may choose to use left-right encoding to reduce the pile up on the structures we are interested in.

The level of distortion that any given field inhomogeneity will produce will depend on the **bandwidth** or **bandwidth-per-pixel** in any given direction. The parameter that is set on the (Siemens) scanner is the bandwidth-per-pixel in the readout direction. This value specifies the range of frequencies present in one pixel, and is equal to the total bandwidth across the field-of-view divided by the number of pixels in the readout direction:

$$BW\text{-}per\text{-}px = \frac{BW}{N_{READ}}$$

The time to acquire a single line of k-space is also given by 1/bandwidth-per-pixel:

$$\frac{1}{BW\text{-}per\text{-}px} = \frac{N_{READ}}{BW} = N_{READ}\Delta t$$

In EPI, the magnetisation is 'recycled' by acquiring multiple phase encode lines after a single RF excitation. The time between phase encode lines is therefore limited by the time to acquire a single line of k-space plus the time it takes to switch the gradient from positive to negative. This is referred to as the phase encode spacing.
With the students, explore the effects of changing readout bandwidths on the image distortion and chemical shifts:
>> options.Bandwidth = 1000;
>> result = simDistortion(options)
Note the phase encode spacing in the result output, and how that decreases with increasing bandwidth. Advanced students may benefit from a discussion about the interpretation of distortion as mislocalisation of magnetisation due to improper phase accrual due to inhomogeneous magnetic fields. Because phase errors become more egregious when the magnetisation has more time to evolve, distortions and mislocalisation errors increase with increasing phase encode spacing. Alternatively, a discussion about "effective gradient strength" relating to bandwidth and distortion may be useful when attempting to avoid explicit discussion of phase.

Question 1.2.3*
For a 64x64 matrix our system is capable of producing a maximum readout bandwidth of 7500 Hz/pixel. On our system what is the minimum phase encode spacing? At which readout bandwidth (to the nearest 500 Hz/px) does this occur? What is the disadvantage of using higher bandwidths (think of at least two)?
Approx. 4000 Hz/pixel; this is not the system maximum, as in reality we have to

spend more and more time switching the gradients as we increase the bandwidth (but less time acquiring a readout line). This is because gradients have maximum "slew-rates", which is the how quickly the gradient can ramp up its amplitude. Disadvantages include lowered SNR, higher peripheral nerve stimulation, expensive gradient hardware and non-linearity etc.

Another way to reduce the spacing between phase encode lines is to use parallel imaging. In parallel imaging, phase encode lines are skipped and coil sensitivity information is used to fill in the missing lines. Skipping lines greatly decreases the phase encode spacing, since multiple lines are effectively "acquired" in the time it takes to measure one.
One implementation of parallel imaging is called GRAPPA (stands for Generalized Autocalibrating Partially Parallel Acquisitions), and on Siemens scanners it is often referred to by the terms PAT or iPAT.
Explore the effects of GRAPPA (using integer acceleration factors only) on the output image quality.

```
>> options.GRAPPA = 2;
>> result = simDistortion(options)
```

Question 1.2.4*
Consider acquiring data using a 32 channel receive coil (a receive coil consisting of 32 individual radiofrequency coils). What is the theoretical maximum acceleration factor you could achieve with 32 coils (assuming best-case scenario)? Why do you think this cannot be attained in practice?
If the 32 RF coil sensitivities are linearly independent (no coil sensitivity can be formed from a linear combination of the others), and the measurement of the coil sensitivities is perfect, in theory an acceleration factor of 32 is possible. In practice, many factors contribute to the impracticality of extremely high acceleration factors:

1. *Coils sensitivities may not be linearly independent along the direction of acceleration, which causes the mathematical problem to be unsolveable*
2. *Even if coil sensitivities are linearly independent, but exhibit some large degree of similarity, noise amplification will cause extremely poor image reconstruction*
3. *It is very difficult to measure reference/calibration data or coil sensitivities that are perfectly accurate (due to measurement error, motion, physiological noise etc…)*

Often, the degree of coil sensitivity variation and the amount of tolerable noise amplification on current hardware limit acceleration factors to a maximum of ~ 4x.

Question 1.2.5
What happens to the distortions and the noise as the GRAPPA factor is increased? What is a sensible maximum GRAPPA factor for this simulation?
When increasing the GRAPPA factor the time between phase encode lines is reduced, which reduces the distortions.
With higher GRAPPA factors there is a reduced SNR. This increased noise has a spatial pattern (i.e. the noise increases rapidly in the middle of the brain where there is little sensitivity difference between coils). We can get what looks like an image up to a GRAPPA factor of 4 on our simulator, but what you can actually use practically will depend on expected BOLD response size and your signal to noise/sensitivity to motion.

Extra Information – The cost of parallel imaging
When parallel imaging is used to accelerate the acquisition of an image by a factor R, the noise in the image is expected to increase by a factor of g*sqrt(R), which can significantly impact the SNR of the image.
Two different factors contribute to this noise increase:

1. Unfavorable coil geometries can lead to spatially varying noise amplification factors, called "geometry-factors" or g-factors for short. The g-factor depends on the acceleration factor and the configuration of the receive coils, and can range from 1 (ideal, no amplification) to >1 (noise amplification).
2. When N/R lines of k-space are acquired instead of the full N, this reduces the total readout time by a factor of R, and recall that SNR is proportional to the sqrt(readout time). This is a fundamental relationship, and no degree of coil rearrangement can reduce the impact of this factor. Note however, that this reduction is relative to a fully-sampled acquisition with the same coil, so if measurement quality were improved significantly in a new coil, the relative loss in SNR due to acceleration may be well accommodated by an overall increase in the absolute SNR levels provided by the coil.

1.3 Dropout [15-20 minutes]

From the lectures, recall that field inhomogeneity causes dropout in all gradient-echo sequences, but that distortions primarily occur in EPI (due to extended readout durations). However, distortions and dropout are linked as they are caused by the same effect, B_0 inhomogeneity. It may be useful to think of distortion relating to *inter-voxel* field inhomogeneities, and dropout relating to *intra-voxel* field inhomogeneities.
Introduce changes in dropout by changing the echo time and/or reducing the slice thickness. You might want to reset the simulation first.

```
>> clear options;
>> options.AddSinus = true;
>> options.TE = 100;
>> options.SliceThickness = 2;
>> simDistortion();
```

Discuss the difference between distortion and dropout, by reinforcing the concepts of distortion being the result of mislocalisation due to inaccurate average field in a voxel, and dropout being the result of dephasing or destructive interference from the presence of a spread of frequencies.

Question 1.3.1
Why does changing the TE affect the dropout? What about slice thickness? What will this do to signal levels? How does changing the TE or the slice thickness affect the distortions in EPI?

- *Longer TE: more time spent at an off-resonant field -> increased distribution of phase within the voxel -> more dropout.*
- *Slice thickness: thinner reduces the distribution of field offsets within the voxel and so reduces the dropout for a given TE. Dropout can reduce the optimal time for BOLD contrast, so our ideal TE changes with brain region.*
- *None of these changes affect distortion (you can think of distortion as being due to the average of the field inhomogeneity, whereas dropout is the range of field inhomogeneity within a single voxel).*

Question 1.3.2**
Introducing spin-echoes almost completely reduces the effect of signal dropout. Explain why this is the case.
Assuming things are completely static, 180° refocusing pulses completely reverse the effects of magnetic field inhomogeneity. To understand how this works, recognize that the effect of the 180° pulse acts to negate all the phase accumulated

by magnetisation between the excitation and refocusing pulse (i.e. for a duration of TE/2).

Consider a field offset of $\Delta\omega$ *affecting the signal* S *in a given voxel:*

$$\angle S_{TE/2} = \Delta\omega \frac{TE}{2}$$

now the spin echo occurs, and the current accumulated phase is reversed, and it accrues another TE/2 worth of new off-resonance phase:

$$\angle S_{TE} = -\angle S_{\frac{TE}{2}} + \Delta\omega \frac{TE}{2} = 0$$

So because at time TE, the net off-resonance phase is 0, the image is produced as if no off-resonance term (i.e. no inhomogeneity) was present at all.