MRI exercise: Acquisition and reconstruction

Medical Imaging Systems, Lars G. Hanson, DTU and DRCMR, lgh@elektro.dtu.dk

1 Background

Linear field variations (gradients) are induced by electromagnets during scanning. Choosing the z-axis to be along the direction of the main field $B_0$, a gradient $G_x$ in the $x$-direction gives the following field dependence: $B_z = B_0 + G_x x$. There are independent electromagnets (gradient coils) that give linear field variations in all directions, so more generally $B_z = B_0 + G \cdot r$ where $G = (G_x, G_y, G_z)$ and $r = (x, y, z)$. A gradient $(G_x, 0, 0)$ of duration $\tau$ applied after excitation will, for example, induce a phase roll in the $x$-direction characterized by the wave number $k_x = \gamma G_x \tau$ (inverse wave length), where $\gamma = 42.58$ MHz/tesla is the gyromagnetic ratio for hydrogen nuclei. More generally, a gradient $G$ applied for a duration $\tau$ changes the $k$-space sampling position by an amount $\Delta k = \gamma G \tau$.

2 Part 1: Simulation of magnetic resonance phenomena

The Bloch simulator can be started from [http://www.drcmr.dk/BlochSimulator](http://www.drcmr.dk/BlochSimulator) where you will also find links to animations shown in the lectures. Use “hard” radio wave pulses for this exercise, i.e. short, strong bursts.

The Scene/Gradient menu item starts a simulation where a row of nuclei experience a field gradient. Following excitation, the nuclei precess at different frequencies. In a frame of reference rotating at the center frequency, some spins precess clockwise, others anti-clockwise. Verify this using the “Change frame” button after excitation by a 90° RF pulse.

The changing phase-roll pattern reflects movement in $k$-space. A subsequent 180° pulse causes the immediate sampling position in $k$-space to be reflected in the origin (the trajectory continues there). Draw the trajectory in one-dimensional $k$-space followed during the spin-echo sequence that you thereby simulated.

In order to do imaging, the MR signal needs to be recorded as a function of $k$ and be Fourier transformed to estimate the distribution of transversal magnetization. Plot the signal as a function of $k_x$ and explain its relation to a “sinc”-function (the Fourier transform of a square function). Relate this to the spatial spin distribution.

3 Part 2: Image reconstruction for echo planar imaging

Echo planar imaging (EPI) is the fastest conventional MR imaging technique. It is used when image quality needs to be sacrificed for temporal resolution. An EPI brain image is acquired in a tenth of a second at the cost of spatial resolution, signal to noise ratio, artifacts and contrast loss as seen in figure 1. Despite the limited image quality, EPI images are widely used for mapping of brain functions (functional imaging, fMRI), since temporal resolution is of paramount importance for this application.
The present exercise gives insight into MR sequence design and image reconstruction using EPI as an example. The example data are acquired simultaneously and independently with an array of surface coils located around the head as shown in figure 1.

Figure 1: Left: An axial slice through the brain acquired with EPI in a tenth of a second. The EPI image quality is fairly bad but the technique is nevertheless very important, e.g. for mapping of brain functions. The arrow point to artifactual signal loss and distortions in the frontal part of the brain caused by magnetic field inhomogeneity near the air-filled sinuses. Right: For acquiring the example data, an 8-element head coil was used. the bars drawn on the conventional MR image indicate the coil element positions. Each of the eight elements is predominantly sensitive to the nearby region.

Doing echo planar imaging (EPI) the entire 2D $k$-space is mapped after a single excitation. An EPI sequence diagram is given in figure 2 together with the corresponding $k$-space trajectory. This sequence was used to measure the provided data.

1. Load EPI MR raw data into matlab using the commands

   ```matlab
   load('MRdata.mat');
   % Dimensions of raw are slices, coils, ky, kx:
   [nslc, ncoil, ny, nx] = size(raw);
   ```

   For each slice and coil, raw is a 128 x 128 matrix of complex measurements, $S = S_x + iS_y$, where the signals $S_x$ and $S_y$ are proportional to the components of the total transversal magnetization in orthogonal directions $x$ and $y$. Each line raw(slc,coil,Cy,:) is measured on a positive or negative gradient lobe (moving right or left in $k$-space).

   Do the following exercises for slice slc=3 and coil element coil=1 that covers the back of the head:

2. Reconstruct an image from the even lines only, by doing a two-dimensional FFT. The phase expressing the direction of the transversal magnetization is normally not of importance and can be discarded. Note that the low sampling density of the reduced data set give rise to an “aliasing” artifact: The part of the head that extends outside the “field of view” appears in the opposite side of the image, so there is an apparent overlap between tissues at two different locations.
3. To avoid aliasing, the sampling density needs to be maintained high by doing an image reconstruction from even and odd lines simultaneously. To do that, reorder the data into a $k$-space data matrix by reversing every second line.

4. Show the $k$-space raw data matrix with the right axis units (the “field of view” is 230 mm). Note that the signal is strongly peaked around $k_x = k_y = 0$ (center of the matrix). Hint: You may choose to view the logarithm of the absolute signal to enhance the visibility of the small signals. Note the asymmetry due to relaxation during the signal readout period.

5. For each coil element and slice, perform image reconstruction by doing a two-dimensional FFT of the complex $k$-space raw data. View the results for slice 3, all coil elements separately.

6. Add the reconstructed images for all coil elements to reconstruct a full view of slice 3. The best SNR is obtained by doing a “sum-of-squares” reconstruction where the resulting image is calculated as the square root of a weighted sum of images from each coil element. Each image is thereby weighted by itself to ensure that each part of the sum image is dominated by the coil having the best SNR in that part.

![Diagram](image.png)

Figure 2: Diagrams showing the sequence of events during the acquisition of a single image and the corresponding $k$-space trajectory. First, a slice selective RF pulse excites a slice of tissue in the $xy$-plane. Subsequently $k$-space is traversed by switching gradients in the $x$- and $y$-directions. The combined effect of the initial gradient lobes in both directions is a phase roll along a slanted direction (e.g. from right neck to left forehead). The $y$-component of this roll is gradually removed and later reversed by subsequent small gradient “blips” in the negative $y$-direction. Meanwhile, $k$-space is rapidly and repeatedly traversed in the $x$-direction. The signal $S(k_x, k_y)$ shown in the background is measured during each gradient plateau. The density of $k$-space samples, $\Delta k_x$ and $\Delta k_y$, determines the “field of view”, e.g. $\text{FOV}_y = 1/\Delta k_y$. Aliasing results, if $k$-space is not sampled sufficiently dense. The extend of the covered $k$-space region determines the spatial resolution.