MRI lab - Mandatory Assignment FYS4740

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(Dated: March 3, 2021)

I. INTRODUCTION

MRI is a medical imaging technique used to obtain detailed images of the inside of different parts of the body. This image technique consists of using a strong magnetic field and radio frequency (RF) waves to view bones, organs and tissues, and is thus a very important tool for pathology and diagnostic medicine.

The MR-scanner consists of a large cylindrical magnet where the patient is placed inside the center. When turning on the magnetic field the hydrogen nuclei within the patient's body go from spinning in random directions to aligning the spin directions parallel to the magnetic field. When RF-pulses are sent into the part of the body which we want to image, the nuclei are flipped into another plane. After the nuclei is flipped, the nuclei will proceed to "relax" back again into the direction of the external magnetic field – this is often referred to the equilibrium state. In order for the pulse to flip the nuclei, it needs to be in resonance with the spins of the nuclei. Having a magnetic field and obtaining resonance is two crucial tools in order to perform imaging, thus giving rise to the name we all know: Magnetic Resonance Imaging.

Getting a MRI will be double the cost as taking a CT scan, but it has the advantage of producing mores sharper and detailed images of high resolution. Tendons and ligaments does not show up in a CT scan.

Each imaging sequence in a MRI is a combination of RF pulses and gradients. The aim for every sequence is to obtain tissue contrast as quickly as possible in order to limit artifacts but while also maintaining a sufficient signal to noise ratio. There are a large number of sequence types. Each sequence tend to have their own acronym depending on the manufacturer of the MR-scanner, and this is why specialists within the field tend to focus on only one type of scanner, for example, a Philips or a Siemens scanner.

This paper will be parted in three sections. The first consists of a short theory section, the next section will consist of a explanation of the methods used to obtain the results. The last section will consists of a presentation of the results followed by discussions.

II. THEORY

NordicICE

The MRI software called nordicICE is used as a tool in order to visualize and analyze images obtained from MRI. It can be used to measure the intensity signal of different region-of-interest (ROI). The software allows for using a ROI in a shape which can be determined by the user, and further allows for different analysis in the ROI where the software can quantify mean signal/pixel intensity, standard error of pixel intensity, median pixel intensity in the ROI etc. This tool allows for estimating important parameters from the imaged object of interest. These parameters include T1- and T2 relaxation in different ROIs, as well as signal-to-noise ratio (SNR) of the image. The latter is important in order to quantify the quality of the obtained image, and the relaxation times is important in order to classify the different substances in the object of interest. Differentiation between substances can especially be important for diagnostic purposes especially when wanting to differentiate between malignant tumor tissue and normal tissue in the imaged object.

T2 Relaxation

After the RF excitation pulse, the net magnetization will be flipped down into the transverse plane. This precessing net transverse magnetization will induce current in a receiver coil. This current is known as the MR signal, which is the only observed signal in a MRI-scanner.

The magnetization will precess with the same frequency if the external field is homogeneous. But this is not the case since each magnetization exists in a microscopic environment which is unique to the local tissue. The effective field in the tissue is not homogeneous since each spinning charged nuclei will generate small magnetic fields. Due to the fact that the Larmor frequency is proportional to the effective field in the given area, the spins in different spatial locations will precess at different frequencies. This process of dephasing continues until the frequencies of each spin have completely randomized and the net magnetization in the transverse plane will reduce to zero.

This means that the MR signal detected in the coil will follow a decaying sine wave, and decay with time until it reaches zero. This is known as the free induction decay (FID) signal. The time constant where the signal have

reduced to $1/e$-th of its original magnitude is known as the T2-relaxation. This relaxation effect can be described by the following equation

$$M_{xy}(t) = M_{xy}(0)e^{t/T2}, \tag{1}$$

where $M_{xy}(0)$ is the signal at the start of the acquisition (t=0).

T1 Relaxation

Without an external magnetic field, each nucleus in a given substance will have a random orientation of its magnetization in a state of thermal equilibrium. When this substance is inside a static homogeneous magnetic field, the orientation of each magnetization will be parallel to the magnetic field.

Let's assume that the magnetic field is in the z-direction, and when the magnetic field is turned on, we obtain a net magnetization in the z-direction. This is also called net longitudinal magnetization. When a RF-pulse applied into the system, some of the nuclei will "flip" its magnetization down in the transverse plane. When the net magnetization is flipped into the transverse plane, the longitudinal magnetization will thus be non-existent. This precessing transverse magnetization will eventually decay (T2 relaxation) and give rise to an induced current in the receiver coils placed near the part of the body which we want to image. The current is the actual MR signal.

Applying a 90 degree RF-pulse will give rise to net magnetization in the transverse plane. A flip angle lower than 90° will decrease the amount of magnetization flipped onto the transverse plane. Having e.g. 45° RF pulse will give us only half of the maximum signal because it allows for a faster recovery of longitudinal magnetization.

After sending in the RF-pulse, the system will go back into its equilibrium state thus aligning with external magnetic field, and the longitudinal magnetization will eventually recover. T1 is the time where (1/e) of the total longitudinal magnetization has recovered.

As mentioned earlier, estimating important parameters in T1 relaxation times of different substances is quite useful. This parameter is often used because different substances have different T1 relaxation times, and thus estimating this parameter at different regions can be used in order to classify the different substances in T1-weighted MRI sequences.

Let's observe the Bloch equation of the longitudinal magnetization component,

$$M_z(t) = M_{z,0}e^{t/T1} + M_0(1 + e^{t/T1}), \tag{2}$$

when in the time where t=T1, 36% of the initial longitudinal magnetization $M_{z,0}$ has recovered.

Inversion Recovery

In order to obtain increased T1 contrast in T1-weighted MR images, we can use a magnetization preparation technique called inversion recovery (IR). This method contains sending a 180° RF excitation pulse followed by an imaging sequence.

The 180° RF-pulse allows the longitudinal magnetization to be flipped down to the negative z-direction. The longitudinal magnetization will recover due to relaxation and will thus increase with time. The longitudinal relaxation will recover with different speed depending on the type of substance. And at some point in time, the longitudinal relaxation will pass through a value of zero. If a specific substance, have a zero value magnetization at the time where a RF excitation pulse is given, the substance will give no signal since there is no net magnetization to flip. This is often referred to as "silencing" a specific substance, and will thus appear dark in the MRI.

The time constant between the inversion recovery pulse and the first RF excitation pulse is called TI. The time evolution of the longitudinal magnetization after a 180° inversion pulse is given by

$$M_z(t) = M_0\left(1 - 2e^{-t/T1}\right), \tag{3}$$

and note that when t=T1, the longitudinal magnetization reduces to zero.

Spoiled Gradient Echo and Ernst angle

One of the main characteristics of a gradient echo (GE) sequence is usually applying a FA lower than 90° (partial FA) and using the frequency encoding gradient (GFE) in order to create an echo. Since no refocusing 180-degree excitation pulse is applied in order to eliminate static inhomogenities of the magnetic field, the images is therefore T2*-weighted instead of T2-weighted.

Using a partial FA gives the consequence of giving a less stronger signal due to having partial magnetization flipped into the transverse plane. This also gives the advantage of having decreased recovery time (T1) and thus allows for shorter repetition time (TR - time between each RF excitation pulse). This results in shorter acquisition time when using GE sequences.

The formation of an echo in the GE sequence is due to applying a dephasing GFE (negative polar)

before a positive rephasing GFE. The dephasing lobe of the gradient is half of the lobe of the positive rephasing GFE. During readout of the signal, the first half of the positive lobe reverses the earlier dephasing and the last half of the positive lobe will induce again an dephasing of the spins. The peak of the echo corresponds to where the spins are maximally in phase with each other.

The transverse magnetization will not fully decay when TR is reduced. Therefore, we will obtain a residual transverse magnetization before the start of the next excitation pulse. Having a permanent residual transverse due to short TR is referred to as obtaining a steady state. Having redisual transverse magnetization can be avoided by spoiling. Having a spoiled GE sequence utilizes RF-pulses at different phases at the beginning of each TR (RF-spoiling), or gradients to completely eliminate the residual transverse magnetization at the end of each TR (gradient spoiling).

In a spoiled GE sequence, the image contrast will depend on

- The FA for T1-weighted images. The greater FA, the more T1-weighting.
- The value of TE for T2*-weighted images. Shorter TE result in less T2*-contrast.

The expression of the signal equation for a spoiled GE sequence is given by the following equation

$$M_T(TE, TR, \alpha) = M_0 \frac{\sin\alpha(1 - e^{-TR/T1})}{1 - e^{-TR/T1}\cos\alpha} e^{-TR/T2^*}, \tag{4}$$

where α denotes the FA, and $T2^*$ is the relaxation of the transverse magnetization due to field inhomogeneities in the static magnetic field.

As earlier mentioned, we usually want a partial FA in a GE sequence. Consequently, a smaller angle reduces T1-contrast. For a given TR and T1, we want to determine the FA which gives the maximum signal. Taking the partial derivative of the signal equation of a perfectly spoiled GE sequence (eq. **??**) with respect to FA and set the equation to zero in order to find the FA which gives the maximum signal. Solving for FA will give us the optimum angle. This angle is also referred to as the Ernst angle, given by

$$\alpha_E = \arccos(e^{-TR/T1}). \tag{5}$$

SNR

Signal-to-noise ratio is used as a measurement of the relative ratio between noise and signal. Having a SNR higher than 1 would in fact imply having more signal than noise. It is important that a signal does not disappear into the random background noise in order to properly analyse the signal. SNR is used to quantify the quality of the signal many different circumstances, and is as commonly used in the context of studying MRI.

The random noise in the MRI can be due to effective coil resistance arising from the random thermal motion of electrons in the RF coil [2]. Noise can also be due to radiation being emitted and absorbed by RF coils, and will thus cause random fluctuations in voltage (noise)[2]. Eddy currents due to changing magnetic fields will also induce voltages in the RF coil, and thus cause noise[2].

SNR is also dependent on the type of sequence and the predetermined sequence parameters[1].

NSA is the number of excitation or the average number of phase encoding steps used in a slice in order to fill the k-space. Doubling the NSA will also double the scan time because the NSA determines how many times a line of k-space is filled (no. of phase encoding steps).

The signal is independent of the NSA, but the noise is random for each measure. Thus, SNR is theoretically expected to increase with the square root of NSA. Increasing the NSA is favourable in terms of increasing the SNR, but the major disadvantage will be to increase the total acquisition time. Increasing the scan time is also known to give rise to other artifacts.

Echo Planar

Echo planar (EPI) is the fastest acquisition method in MRI with an acquisition time of less than 100 ms per slice. There are four main types of EPI sequences:

- GE-EPI: Sequence starts with a single RF pulse with no magnetization preparation. The image is T2*-weighted.
- SE-EPI: Starts with a 90° excitation-pulse followed by a 180° refocusing pulse eliminating dephasing effects due to static field inhomogeneties. The image is thus T2-weighted.
- IR-EPI: 180° inversion pulse is applied, followed by a RF excitation-pulse. The image is thus T1-weighted with increased T1 contrast due to magnetization preparation (IR-pulse).
- DW-EPI: Used for diffusion weighting.

The cost of the high acquisition speed of an EPI-sequence is demand for intense high-performance gradients for fast signal readouts, with short ascent times. The artifacts in EPI sequences are often due to imperfect gradients and sensitivity due to magnetic susceptibility.

The main difference between an EPI-sequence and

standard GE-sequence is the acquisition time. With standard GE, the acquisition is segmented, obtaining only one line in k-space in the k_x-direction with each TR. Thus the total scan time is a total of $N_x \cdot TR$ where N_y denotes the number of steps in the k_y-direction. For EPI-sequences, we can obtain multiple echoes in one single TR. Obtaining enough echoes to fill the entire k-space in a single TR, a so-called single-shot acquisition. With this sequence it is possible to acquire signal from the total or partial k-space with only one TR.

The main difference between GE-EPI and SE-EPI is image-weighting. The former does not utilize a 180° refocusing pulse which corrects for the inhomogenities in the static magnetic field. The resulting image is thus T2*-weighted. The SE-EPI is T2-weighted since it utilizes a 180 pulse which refocuses the dephasing of spins caused by the inhomogenities in the static magnetic field, and thus give existence to a spin echo. T2* corresponding to a substance will always be shorter or equal to T2 of the given substance because decay of transverse magnetization is much faster due to T2*-effects than only due to local microscopic susceptibility in the substances. Both sequences utilize long TR time in order to acquire many echoes in the same TR, and thus eliminate T1-effects in the resulting image.

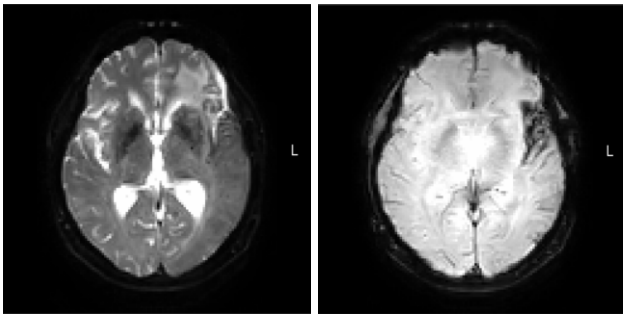

FIG. 1. left: SE-EPI, right: GE-EPI. Both are MR images of the same tumor patient.

MR images in figure 1 represent images of the same human brain obtained from SE-EPI sequence (left image) and a GE-EPI sequence (right image). The right image is brighter because longer echo train causes greater T2*-weighing.

Magnetic Susceptibility Artefacts

Artifacts can arise from the presence of substances of different magnetic susceptibility that cause local variations in the magnetic field. Protons are paramagnetic ($\chi > 0$) meaning that their magnetic moments align in the direction of (parallel) the external magnetic field. Diamagnetic materials such as human tissue ($\chi < 0$) align their magnetic moments anti-parallel in the presence of a external magnetic field. The susceptibility of the protons are thus much smaller than the total susceptibility of the surrounding tissue, and the net effect will be a diamagnetic effect. The local effective field due to the local susceptibility is given by

$$B_{eff} = (1 + \chi)B_0, \tag{6}$$

and thus the effective field paramagnetic materials is larger compared to diamagnetic materials.

When ferromagnetic materials are placed in a external magnetic field the material gets magnetized. When the field is turned off, the material will remain magnetized, and thus a permanent magnet is created. Iron is an example of a ferromagnetic material.

Areas with local susceptibility differences give rise to imaging artefacts. The severity of these artifacts depend on the magnitude of the local field variations relative to the pixel dimension. When local field distortions are small compared to the voxel size the net effect will be dephasing within the voxel volume. This is referred to as intravoxel dephasing, and the effect of this is the artefacts in the image in a form of signal loss in the MR image.
If the local field distortions are large compared to the voxel size, the effect will be frequency shifts of spins within the voxel. This is referred to as off-resonance effects, and cause artifacts in the form of image distortions.

Patients with braces, surgical staples, mascara containing iron will most commonly cause signal loss due to intravoxel dephasing.

In spin echo (SE) sequences, these susceptibility artifacts are less severe than in in GE sequences. This is because SE sequences contain 180 pulses which eliminates susceptibility-induced dephasing of spins which is also know as T2*-effects. Therefore, the susceptibility artifacts are less visible in SE sequences than in GE sequences.

Magnetization prepared rapid Gradient Echo (MP-RAGE)

Magnetization prepared rapid gradient echo is sequence of the GE type and is accelerated in order to reduce acquisition speed by using a small FA and very short TR. Small FA and short TR reduces T1-contrast. One way to preserve this T1-weighting is to utilize the magnetization preparation technique of a IR-pulse before the accelerated GE sequence.

Due to short TR, there will exist residual transverse magnetization at the end of each TR. This residual magnetization may become permanent when TR is

shorter than T2. This steady state effect can worsen the T1-weighting, and thus gradient spoilers are used in order to completely eliminate this residual transverse magnetization.

The entire k-space can be filled after a single inversion pulse (single-shot), or part of the k-space can be filled (segmented acquisition).

MP-RAGE have shorter acquisition time compared to a standard GE sequence. MP-RAGE is also T1-weighted, while the GE sequence can either be T1 weighted (when FA is larger) or T2*-weighted (in case of a shorter TE). If a standard GE sequence is T1-weighted, the TR increases for larger FAs in order give time for transverse magnetization to completely decay. Standard GE uses no IR pulse in order to obtain T1-weighting compared to a MP-RAGE sequence. The use of an IR pulse can silence other tissue types of different T1 relaxation by determining the parameter TI as explained earlier.

3D vs 2D imaging

In regular 2D sequences, a slice selective gradient (SSG) together with an excitation pulse is used to select the slice of interest. The excitation pulse match the frequencies in the given slice, ensuring excitation of only the protons in the slice. Horizontal position localization of MR signal is done by applying a frequency encoding gradient (GFE), while vertical localization is done by a phase encoding gradient (GPE), such that each phase encoding step fills a line of k-space in the y-direction.

In contrast to standard 2D acquisition, 3D acquisition excite a volume of interest with the SSG and a excitation pulse. This type of acquisition method is possible when adding a phase encoding gradient in the third direction (added to the SSG) together with the GFE and GPE. 3D Fourier transform is used in order to reconstruct the image.

3D acquisition does have consequences. Sequences of short TR (GE sequences) or accelerated sequence methods needs to be used given the large amount of data sampled from the volume element. This gives restrictions on sequence parameters and the choice of sequence in order to reduce total scan time.

Another consequence is the rise of artifacts due to the two phase encoding gradients. This cause wrap-around/fold-over and truncation artifacts in the same two directions as the two phase encoding gradients.

Since the signal comes from the whole volume at each repetition rather than a single slice, the SNR increases with the number of slices included in the volume element. Because more signal is recorded with less noise, the SNR is higher for a slice acquired with 3D acquisition compared with a slice of same thickness acquired with 2D.

Gibbs Ringing

Gibbs artifacts – often referred to as Gibbs ringing, is due to the use of Fourier transformations in order to reconstruct the image. According to the Fourier theorem, any signal can be represented as a sum of sine waves with different phases, amplitudes and frequencies. This sum is referred to as the Fourier series. However, in MRI we only sample a limited range of frequencies. Therefore only a limited variations of sine waves is used to approximate the image. The Fourier series is thus shortened or truncated.

Only a small number of terms are needed in the Fourier sum when the signal intensity varies gradually in space, and thus the Gibbs artifacts are not noticeable. However, at high-contrast interfaces such truncation of the Fourier series result in severe ringing artifacts. Widening of edges at these high-contrast interfaces can also occur depending on the number of pixels the interface consists of.

These artifacts can occur in the phase encoding- and the frequency encoding direction. Most commonly, the number of samples taken in the phase encoding direction are smaller compared to the number of samples taken in the frequency encoding direction. Because the Fourier series is more truncated because due to fewer samples, the ringing artifacts is more severe in the phase encoding direction. Increasing the number of steps in the phase encoding direction or by reducing the FOV can minimize these ringing artifacts.

Corsmed

Corsmed is a simulation tool used to simulate different MRI sequences. The user actively learns how to plan MR scans and can choose whether to scan a anatomic model of the body or the head. The user can predetermine sequence parameters and analyze the resulting images. The images obtained are synthetic MR images. The simulated images gives the same results with artefacts and noise as in real life.

The interface presented is similar to the one of a real MRI scanner. The purpose of the software is to allow students to practice MRI scanning.

The user can choose to perform the scan on either a 1.5T or a 3T scanner. Field of view (FOV), as wee as voxel parameters can be determined by the user.

The type of sequences including SE, GE, EPI, IR, Fast SE, MP-RAGE can be predetermined along with their respective sequence parameters TR, TE, TI, FA and bandwidth. Slice thickness and the number of slices can also be chosen by the user.

III. METHOD

Brain Phantom

The imaged object of interest which will be used for T1- and SNR estimations is of a custom-made phantom containing five circular compartments with specific proton T1 relaxation values. An MR image of this phantom is given in figure 2. Each compartment has also been renamed with a given number. Upper left compartment is compartment 1, lower left compartment is referred to as compartment 2, etc.

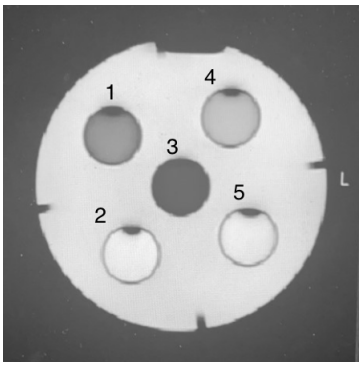

FIG. 2. Image of MR image of phantom from a IR sequence. The numbers is what the compartment will be referred to. E.g. middle compartment is compartment 3.

Extracting Data using nordicICE

The most often used tool from the nordicICE software was measuring the ROI statistics and using the mean signal/pixel intensity for T1-analysis, and the standard deviation of the signal of signal intensity when estimating SNR.

T1 Measurements

Different MR images taken of a phantom brain with five different compartments with different substances were used in this exercise. Each compartment had their own respective T1 relaxation times. For this exercise, we use two separate sets of images. The first data set containing MR images using a 2D IR multiple-inversion sequence with a flip angle (FA) of 4° and a total of 13 different inversion times (T1) of (50, 75, 100, 150, 200, 300, 400, 500, 750,1000,1500,2000,3000) ms. The second data set contains T1-weighted 2D dynamic spoiled gradient echo (GE) sequence with 12 different FAs of (1,5,10,15,20,30,40,50,60,70,80,90)°. Thus the first data set contains 13 images whereas the second contains 12 images.

Ernst Angle

We want to estimate the T1 based on the Ernst angle, this is done by simply solving eq. 5 for T1. We thus obtain the following formula

$$T1 = \frac{-TR}{\ln(\cos\alpha_E)}. \tag{7}$$

We find the Ernst angle from the data sample by finding the FA which corresponds to the maximum obtained signal intensity. Thus using eq. 7 for all five compartments.

Ratio of signal intensity

By using the expression of signal intensity in eq. **??**, we can use two data points from the spoiled GE-sequence of a specific compartment and two different FAs in order to estimate the T1 for the given compartment. We do so by taking the ratio of two signal intensities and solving this equation for T1. We thus obtain the following equation

$$T_1 = \frac{-TR}{\ln(\frac{a-1}{a\cos\alpha_1 - \cos\alpha_2})}, \tag{8}$$

where $a = \frac{M_{T,1}\sin\alpha_2}{M_{T,2}\sin\alpha_1}$, and $M_{T,1}$ denotes the signal intensity of using FA=α_1 and similarly for signal intensity $M_{T,2}$ by using FA=α_2.

We needed to use two FA's which var relatively largely different from each other, the FA of 5° and 20° in order to obtain reasonable result. Using two FAs which does not hugely differ, for instance 1° and 5° will result in unreasonable results, e.g. estimated T1 times of negative values. This is because adjusting the degree of the excitation pulse is a matter of precision. A 90°-pulse will in theory excite all the protons in the given substance, but not in reality due to microscopic and macroscopic field inhomogenities.

Estimating T1 with TI

We can use the signal intensities used obtained form the IR sequence using 13 different TI times. By using eq. **??** and evaluation the signal at point t=TI, and knowing that the signal intensity $M_z(t = TI)$ reduces

to zero as explained in the theory section, we arrive with the following expression

$$1 - 2e^{TI/T1} = 0 \Rightarrow T_1 = \frac{TI}{\ln(2)}. \quad (9)$$

Since there is no data point where the the signal intensity is zero for each of the four compartments, but in order to estimate T1 we need to "guess" where this potential data point is located by observing the curve behaviour.

Curve Fit using Signal Equations

By using the following curve fit function from Scipy:

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scipy.optimize.curve_fit(func, x, y)
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and knowing the signal equations of the IR (eq. 3) and GE sequence (eq. 4), we thus use Scipy to find the T1-parameter in the function using the raw data extracted from nordicICE.

This Scipy function estimates function parameters, including T1 for all compartments for both the IR and the spoiled GE sequence.

Effect of NSA on SNR

In this section we're going to study the effect of NSA on the SNR. Four images with four different NSA values of (1, 2, 3, 4) have been used for the four different MR images. For each image, we studied two specific ROIs. The first being in the tissue of interest, more specifically a small ROI in the brain to determine the signal intensity. The second ROI being at the background of the image which consist of air (a region which produces no signal). We use the mean intensity of the ROI at the tissue of interest, and the standard deviation of the signal intensity of the ROI of the background to compute the noise which is the variance (squared standard deviation). Taking the ratio of the signal intensity and variance act as a measure of the SNR.

By analyzing the estimated SNR by the MR images with their respective NSA value, we can consider if the SNR increases with the square root of NSA as expected from the theory.

Head scan: Echo Planar Imaging

We're now studying the MR images obtained from a SE-EPI and a GE-EPI sequence of the same tumor patient. Each sequence obtained 11 slices of the brain of the patient. We study the differences of these images, and especially focus on the differences of the artifacts in the GE-EPI images and the SE-EPI images.

Next, we study the mean signal intensity of four different circular ROIs at 10 different values of TE in a multi-echo GE-EPI sequence. The four different ROIs used to obtain singal intensity data is pictured in the four images below (3).

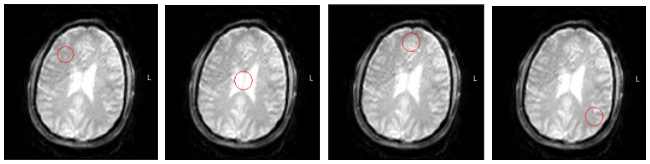

FIG. 3. Four ROIs of a brain obtained from using multi-echo GE-EPI. These four ROIs were used for obtaining mean signal intensity for 10 different values of TE.

The peak of the echoes in a GE-EPI sequence follows the FID signal behaviour, we can use the transverse magnetization signal equation 1 in order to estimate the $T2^*$ relaxation times for each ROI. We use the same curve fit function by Scipy as used when estimating T1 relaxation times.

3D MP-RAGE sequence

We will use the Corsmed software in order to simulate a 3D head scan by a MP-RAGE sequence.

First we use the MP-RAGE with slice thickness of 3 mm and obtain a total of 31 slices of the brain, and the other being the default settings. We thus simulate two additional head scans using TI times of 300 ms and 400 ms and compare these results with the scans with the longer default TI time.

We find the TI time which will give best contrast between white matter (T1=900ms) and grey matter (1200 ms). We do so by analysing the time evolution of the longitudinal magnetization using the IR sequence equation 3. The time point where the difference between the relaxation curve of grey and white matter differ the most is where we get the maximum contrast. Because the signal intensities between the tissue types differ the most at this time, we thus obtain better contrast.

We can also find the TI time which give the maximum contrast between grey and white matter by subtracting the signal equations of each tissue type (eq. 3), and taking the partial derivative of the time and setting the equation to zero. The following equation obtained:

$$TI = -\frac{T1_{white}T1_{grey}}{T1_{white} - T1_{grey}} \ln\left(\frac{T1_{grey}}{T1_{white}}\right). \quad (10)$$

Gibbs Ringing

By using the Corsmed simulation software, we will run a number of GE sequences with a TR of 6 ms and a TE of 4 ms. In the first simulation, we use a large FOV of 400x400 and a small matrix of 128x80.

We copy the earlier sequence but adjust the matrix size to 256x80 and run the simulation. In the next sequence the matrix is adjusted to 80x256. In the last simulation we will increase the matrix size to 256x256 and the FOV to 250x250.

IV. RESULTS AND DISCUSSIONS

In this subsection we will present the results for each of the exercises and then discuss the results.

T1 Measurements

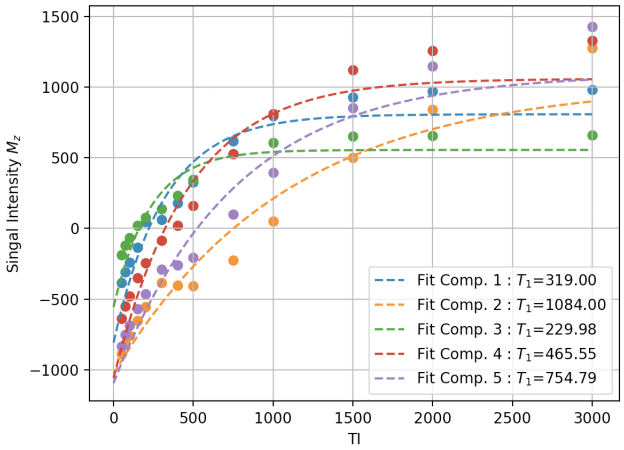

FIG. 4. Resulting signal intensity from a IR sequence using 13 different TI times. The curves corresponds to the measured signal intensity for each of the five compartments using a small circular ROI by the nordicICE software tool.

The image (fig 4) shows the resulting signal intensities of each of the five compartments in the phantom brain using 13 different IR sequences with a different corresponding inversion time (TI). The discontinuous circular points represent the actual measured data (multiplied with the scaling factors) using small circular ROIs with the nordicICE software. The dotted line represents the curve fit, and estimated T1 relaxation for each of the compartment by the curve fit.

From observing figure 4 it is evident that the substances in compartment 1, 3, 4 have regained full longitudinal magnetization at approximately 1000, 1500, 2500 ms respectively due to the fact that the longitudinal magnetization does not seem to increase after these points. Whereas the other compartments have not regained full longitudinal magnetization before 3000 ms according to the observations given in the plot. We thus expect compartment 1 to have the shortest T1 relaxation time, compartment 3 to have the second shortest T1, and compartment 4 to have the third shortest T1 relaxation time.

The curve fit that we observe in figure 4 (dotted line) is not of a high performance. This is because many data points differ highly from the fitted curve which is also reflected in the MSE of each of the curves. The total MSE of the curve for each compartment is given in table I. It is important to note that the curve fit do perform much better without the scaling factors. With the scaling factors, the slope of the data points flatten in the area around 100 to 500 ms as observed in figure 4.

TABLE I. Table contains computed MSE for the Scipy curve fit and the measured signal intensities for the 13 TI times for the IR sequence.

Compartment	MSE
1	230964.42
2	333575.77
3	105028.95
4	403878.93
5	417586.21

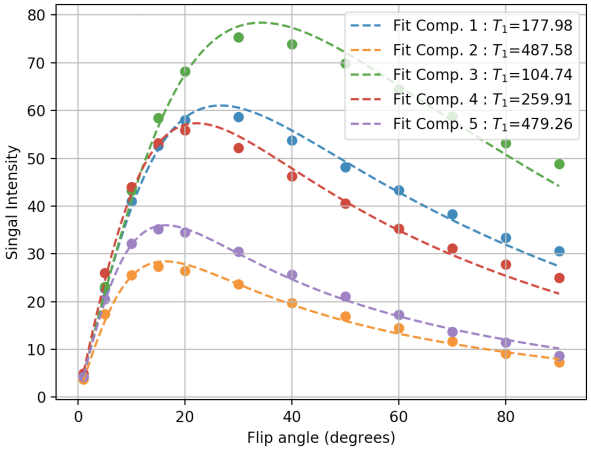

FIG. 5. The measured signal intensities for each compartment in the brain phantom of a spoiled GE sequence. Signal intensities were measured using the nordicICE software.

In the above figure 5, we observe the signal intensity at the five compartments using a spoiled GE sequence with a 12 different FA's of (1, 5, 10, 15, 20, 30, 40, 50, 60, 70, 80, 90) degrees. The maximum signal intensities for each of the compartments is obtained with relatively smaller FAs. These angles range from approximately 15° to 30 °.

The circular datapoints represents the actual measured signal intensity measured from nordicICE, while the dotted line represents the curve fit and its estimated T1 relaxation times for teach of the compartments.

Due to the slope of each curve after it has reached its maximum signal intensity, we can thus expect compartment 3 to have the shortest T1 relaxation due to it having the steepest slope. Compartment 2 has the second shortest T1 time, and compartment 4 has the thirst shortest T1 time. Lastly, we would expect compartment 2 and 5 to have approximately similar (but not identical) T1 times due of the fact that the curves are being very similar to each other.

The curve fit have a somewhat higher performance for these data points (fig 5) compared to the estimated curve fit of the IR-sequence (fig 4). The MSEs from the spoiled GE sequence in table II are many magnitudes smaller than the MSEs of the IR sequence i table I.

TABLE II. Table contains computed MSE for the Scipy curve fit and the measured signal intensities for the 12 FAs for the spoiled GE sequence.

Compartment	**MSE**
1	1412.22
2	215.59
3	2657.17
4	1107.20
5	350.50

Using the curve fits to estimate the T1 will give rise to errors due to the small number of data points for each compartment in both figure 4 and figure 5. Systematic differences in these two results is the fact that the estimated T1 times for the IR-sequence is approximately double the values of the estimated T1 times using the curve fit for the spoiled GE-sequence.

TABLE III. Table contains computed T1 times for the different compartments of the brain phantom. T1 was estimated using eq. 7 and inserting the FA which corresponds to the maximum signal intensity, that is, the estimated Ernst angle.

Compartment	α_E [°]	**T1 [ms]**
1	30	139.0
2	15	576.9
3	30	139.0
4	20	321.5
5	15	576.9

The values in table III represent the maximum data point for each of the compartment which correspond to the estimated Ernst angle. Many compartments share the same Ernst angle since the true Ernst angle lies around these areas, but becomes the same for many of the compartments since we're only dealing with very few data points, 12 to be precise. If we obtained more data points around this maximum, we would obtain the true Ernst angle, and will slightly differ from the estimated angles given in table III.

Since the estimation of the Ernst angle will not be true for each of the compartments due the fact that the data set contains very few number of data points. This will give rise to errors in the estimated T1 relaxation times for each of the compartments. For instance, the curves of compartment 2 and 5 is similar as observed in figure 4, but not exactly identical. This will indicate that the T1 times will not be similar as estimated by the Ernst angle in table III.

TABLE IV. Table contains computed T1 times for the different compartments of the brain phantom. T1 was estimated using the ratio two data points of FA's of 5° and 20° with their respective signal intensity for every compartment in the phantom brain.

Compartment	**T1 [ms]**
1	209.2
2	634.3
3	128.9
4	319.2
5	527.8

The estimated T1-times using the ratio-method is given in table IV. These estimations is most likely not exact due to the precision of the excitation pulse. For instance, for a FA of 90° is not in reality a perfect 90°-pulse because the pulse is truncated. This is a possible reason to why this method may not give accurate results when estimating the T1 for each of the compartments.

TABLE V. Table contains computed T1 times for the different compartments of the brain phantom. T1 was estimated using eq. 9 and inserting the estimated time points where the curve for each compartment is at zero.

Compartment	**T1 [ms]**
1	288.54
2	1308.74
3	184.87
4	523.67
5	933.44

The results by using eq. 9 by setting $M_z = 0$ and estimating by eye where the net magnetization will be at zero for each of the curves in figure 5 is listed in table V. This method will give rise to inaccurate results due to the fact that there is not enough data points to correctly observe where the magnetization is at a value of zero. If the number of data points were dense around where $M_z = 0$ then this method would give more accurate estimations of T1.

Effect of NSA on SNR

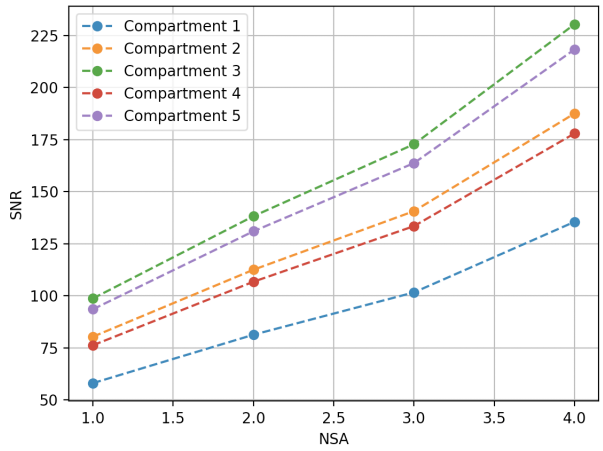

FIG. 6. Illustrating data points of measured SNR for four different images where each image correspond to a different NSA value.

The figure above (6) shows a plot of the measured SNR for the four images corresponding to NSA values of (1,2,3,4), respectively. We observe that SNR is increasing when NSA is increasing, as expected from the theory.

Other parameters which will influence SNR is the slice thickness. Reducing slice thickness will also reduce signal intensity, and thus the SNR. But increasing the NSA will compensate for this signal loss. Increasing the NSA has a consequence of also increasing the total scan time. In this way, increasing NSA will maintain image quality.

Decreasing the FOV will also decrease pixel size, which causes decrease in SNR. NSA can be increased in order to compensate for the signal loss when decreasing FOV.

Since SNR is proportional to $\frac{1}{\sqrt{BW}}$ [1], increasing BW will decrease the SNR. Increasing the NSA can again compensate for this, but increasing the NSA will also increase the total scan time.

Increasing NSA in order to maintain image quality to compensate for increasing eg. FOV, BW or slice thickness will as mentioned also increase the scan time. Increasing scan time allows for motion artefacts. These artefacts is caused by physical movements, respiration or due to cardiac/vascular pulsation. Increasing scan time will increase the probability of obtaining such artifacts, and these are commonly referred to as ghosting artifacts. Reducing the NSA can also reduce the probability of the occurrence of ghosing artifacts.

EPI

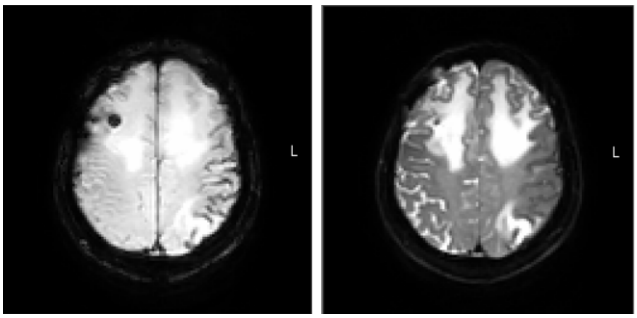

FIG. 7. Slice 8 of MR image of a tumor patient. GE-EPI was used in the left image and SE-EPI was used in the right image.

Observing the above figures 7 which are obtained slices (slice no. 8) of a GE-EPI (left image) and a SE-EPI (right image). The dark circle obtained in the upper left frontal lobe seen in both images are a susceptibility artefact. This is most likely due to a surgical staple placed from a earlier surgery of the tumor patient. Magnetic susceptibility artifacts can give rise to two types of artifact as mentioned in the theory. Since we observe complete signal silencing due to the surgical clip (containing magnetic material) this is of the type intravoxel dephasing as explained in the theory. Such artifacts occur when the local magnetic field variations is smaller than the voxel volume.

These types of artifacts are more severe in sequences of the GE type, due to the image being T2*-weighted and thus missing a 180-refocusing pulse which corrects for static field inhomogenities. Therefore, the artifact appear larger in the GE-EPI image than in the SE-EPI image as we observe in figure 7 as expected from the theory.

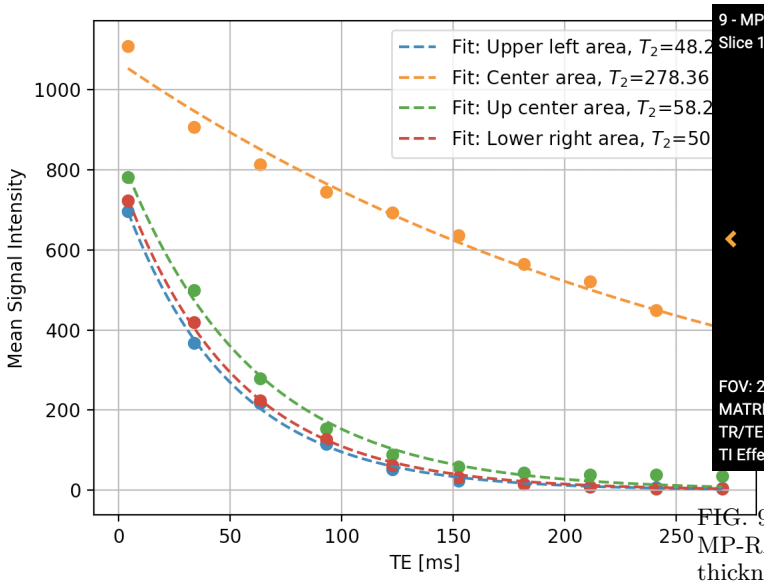

FIG. 8. Obtained mean signal intensities for four ROIs from the images obtained with multi echo GE-EPI sequences.

In the above image we observe the signal intensities of four different ROIs in a multi-echo GE-EPI with 10 different TE times. Since the peak of the echoes in a GE-EPI sequence follows the FID signal, using the transverse magnetization signal equation 1 will give a estimate of T2* for each of the four ROIs. This is the estimated T2* and not T2 since this is a sequence of the GE type.

3D MP-RAGE

In a MP-RAGE sequence we use a IR pulse before the excitation pulse in order to obtain increased T1-contrast, but also silence signals from specific tissue types when sending the RF pulse at the time where the longitudinal magnetization is at zero. When looking at the recovery curves of the longitudinal magnetization of two different T1-times such as grey and white matter, we can thus plot the resulting curves and send in a RF pulse at the time point where the two curves differ the most, and thus obtaining maximum contrast between the tissue types. We use eq. 10 in order to obtain TI which give maximum contrast between grey and white matter in the brain:

$$TI_{max} = -\frac{900 \cdot 1200}{900 - 1200} \ln\left(\frac{1200}{900}\right) = 1036 \text{ ms}. \quad (11)$$

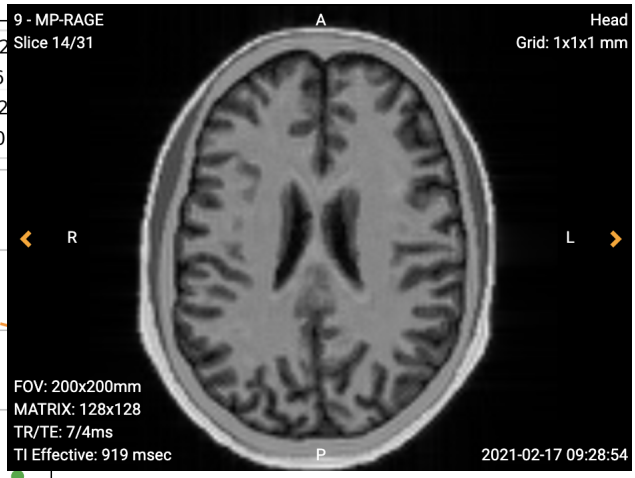

FIG. 9. Axial MRI obtained from Crosmed simulation tool. MP-RAGE was used with the default settings except the slice thickness of 3 mm and a 31 number of slices. The default TI time is of 828 ms.

In figure 9, we observe the obtained image using the Corsmed simulation tool. The sequence used was a MP-RAGE type using the default setting except slice thickness of 3 mm and a total of 31 slices. Compare the resulting images with those obtained with the longer default TI-time. The inversion time used is of 828 ms.

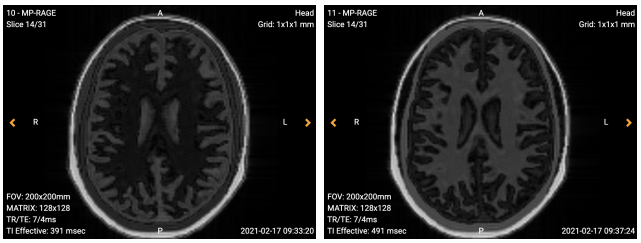

FIG. 10. Left image: Image obtained in Corsmed, using a MP-RAGE sequence of the brain anatomy model with a TI time of 300 ms. The left image is also obtained by using MP-RAGE sequence but now with a TI time of 400 ms.

In the figures in 10 we see have the same parameters are used as when obtaining the image in 9, but now with the non-default parameter of inversion time of 300 ms (left image) and 400 ms (right image). It appears that the left image, we have silenced the signal of the white matter of the brain, and thus it appears black. The signal coming from the grey matter appears grey or weak since this TI time is given when the grey matter have not completely recovered its longitudinal magnetization and thus giving weak signal. However in the right image (fig 10), the the signal coming from the grey matter is now silenced and thus appear black, whereas the only signal coming from the brain area is the white matter. The signal coming from the white matter appears weak, which is the same reason as for why the signal from the grey matter appears weak in the left image.

For 3D acquisitions, we're performing phase encoding in two directions (thus frequency encoding in the last direction), and thus the obtained images are susceptible to fold-over artifacts. These artifacts can occur in the phase encoding direction when the FOV is smaller than the imaged object. Signal from outside the FOV is projected across of the image.

These artifacts can be avoided when the FOV and the number of phase encoding steps are increased, or by simply switching the directions of the phase and frequency encoding directions. It is theoretically possible to obtain fold-over artifacts in the frequency direction, but is rarely happens due to of easy access of hardware that can sample the signal in the frequency direction at a rate twice the maximum frequency within the signal (Nyquist theorem).

FIG. 12. In the left image we have a matrix size of 80 in the read direction and 256 in the phase direction. Image was obtained using a GE sequence in the Corsmed simulation software.

Gibbs ringing

FIG. 11. Obtained sagittal head MRI using the Corsmed simulation tool. We used a GE sequence with a FOV of 400x400 and a matrix size of 128x80.

Observing from the above image (fig 11) the ringing artifacts more severely at areas near the high-contrast interfaces. For instance, we observe ringing clearly at the top of the head where there is air between the skull and scalp. The ringing artifacts appear at the areas which was expected from the theory.

In the image in figure 11, we have adjusted the sequence parameters in the Corsmed simulation tool to have a matrix size of 80 in the read direction and 256 in the phase direction. The FOV was 400x400. The contrast between tissue types has been increased when compared to the image in figure 11, and thus increasing the overall image quality, this is because we increased phase encoding steps from 80 to 256, and thus increasing image contrast, spatial resolution and decreasing ringing artifacts. The ringing artifacts along the vertical/phase direction appears to be less severe – almost non-existent in this image compared to figure 11 due to increasing the number of phase encoding steps as we expected from the theory. The ringing only appears to be along the frequency direction due to minimizing the matrix size in this direction, therefore the number of samples recorded in this direction is fewer and therefore the truncation of the Fourier series is more severe and thus we obtain more Gibbs ringing in this direction. Increasing phase encoding steps also reduces SNR (as the square root of the no. of phase encoding is proportional to SNR) which is also why this image has better contrast in comparison to figure 11 and figure 13.

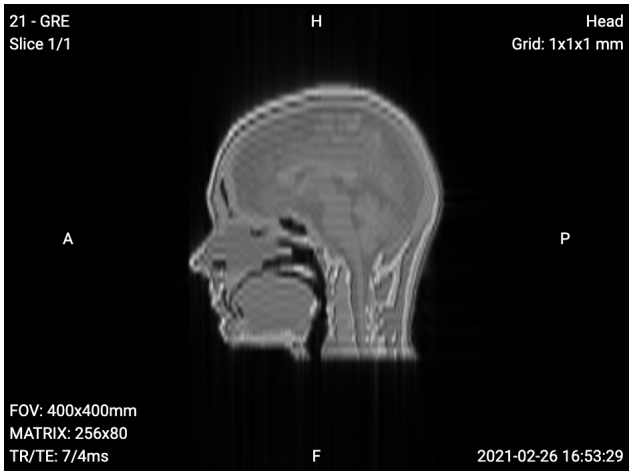

FIG. 13. Image obtained by GE sequence in Corsmed, using a matrix size of 256 in the read direction and 80 in the phase direction.

In this image above (fig 13) we increased the number of frequency encoding steps to 256 and decreased the number of phase encoding steps down to 80 by adjusting the matrix size to these values. The ringing artifacts in the frequency direction (horizontal) has almost been completely eliminated, and there only appears to be ringing artifacts in the phase direction as expected when we decrease matrix size in this direction as explained in the theory. Contrast is now severely reduced compared to figure 12 as a consequence to reducing the number of phase encoding steps and thus increasing SNR.

REFERENCES

[1] Bjørnerud, A. (2020). The physics of magnetic resonance imaging.
[2] K.M. Gilbert, T.J. Scholl, B. C. (2008). Rf coil loading measurements between 1 and 50 mhz to guide field-cycled mri system design. pages 179–181. https://onlinelibrary.wiley.com/doi/pdf/10.1002/cmr.b.20118.

V. APPENDIX A: GLOSSARY

A short glossary list is presented. The intention of table VI is to serve as a lookup table when needed.

TABLE VI. A glossary list containing abbreviations

Word	**Meaning**
MRI	Magnetic resonance imaging
TE	Echo time
TR	Repetition time
TI	Inversion time
GE	Gradient echo
SE	Spin echo
SSG	Slice selective gradient
GPE	Phase encoding gradient
GFE	Frequency encoding gradient
BW	Bandwidth
SNR	Signal to noise ratio
NSA	Number of signal averages
FOV	Field of view
GE-EPI	Gradient echo echo-planar imaging
SE-EPI	Spin echo echo-planer imaging
MP-RAGE	Magnetization-prepared rapid gradient echo

VI. APPENDIX B: CODE

Code is found by this link `https://github.com/feliciajacobsen/MRI/blob/master/oblig.py`.