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Labreport

Contrast in MRI

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1. Introduction

Contrast imaging in MRI is a technique that can be used for better visualizing and distinguishing different tissue types in the body. It allows the identification of structural abnormalities and functional changes in tissues. The contrast in MRI images arises from differences in how tissues respond to magnetic fields and radiofrequency pulses, which is determined by their unique physical and chemical properties (Azhar & Chong, 2023). By changing the imaging parameters and introducing contrast agents it is possible to get highly detailed and diagnostically valuable images.

The most important factors that influence MRI contrast are proton density, T1 relaxation time, and T2 relaxation time. Proton density refers to the concentration of hydrogen nuclei in a tissue and determines the overall signal strength in an MRI image. However, it is the relaxation properties, T1 and T2, that introduce contrast by highlighting differences between tissues (Nyman et al., 1986). T1, measures how quickly the longitudinal magnetization recovers after it has been disturbed by an RF pulse. Tissues with a shorter T1, such as fat, recover more quickly and appear brighter in T1-weighted images, whereas tissues with a longer T1, like cerebrospinal fluid, appear darker. T2, describes the decay of transverse magnetization due to interactions between neighboring spins. Tissues with a longer T2, such as fluid-filled spaces, retain their transverse magnetization longer and appear brighter in T2-weighted images.

The ability to control and enhance contrast in MRI depends on the selection of imaging sequences and their parameters. Repetition time, echo time, and flip angle are three key parameters that determine the degree of T1, T2, or proton density weighting in an image. Repetition time (TR) is the time between consecutive RF pulses in an imaging sequence. When TR is short, the differences in T1 relaxation times between tissues are accentuated, making T1-weighting more prominent. Long TR values allow all tissues to fully recover their longitudinal magnetization and reduce T1-weighting. Echo time (TE) refers to the time between the RF pulse and the collection of the MRI signal. Longer TE values allow more time for transverse magnetization to decay and enhance T2-weighting, while shorter TE values suppress T2 effects. Flip angle, the angle by which the magnetization vector is rotated by the RF pulse, is another important parameter. In gradient echo sequences, adjusting the flip angle allows control over the balance between T1 and proton density weighting, with larger flip angles enhancing T1 contrast. The flip angle is typically limited

by the Ernst angle, beyond which the signal intensity decreases due to saturation effects and increasing it further than 90° can lead to near-complete depletion of longitudinal magnetization, reducing contrast and image quality (Helms et al., 2008). In some cases, tissues do not provide sufficient contrast for diagnostic purposes. In such situations, contrast agents are introduced to enhance image contrast. Gadolinium-based contrast agents are widely used for this purpose due to their paramagnetic properties. Gadolinium shortens both T1 and T2 relaxation times of nearby water molecules, making tissues containing the agent appear brighter on T1-weighted images or darker on T2-weighted images, depending on the sequence used. The effect of a contrast agent on relaxation times can be mathematically described by the equations:

$$\frac{1}{T_1} = \frac{1}{T_{1_0}} + r_1 \cdot [Gd^{3+}] \quad (1)$$

$$\frac{1}{T_2} = \frac{1}{T_{2_0}} + r_2 \cdot [Gd^{3+}] \quad (2)$$

T_{1_0} and T_{2_0} represent the relaxation times of the tissue without the agent, r_1 and r_2 quantify the agent's efficiency, and $[Gd^{3+}]$ is the concentration of the contrast agent. For gradient echo sequences, the signal intensity can be expressed using the equation:

$$GE = A \cdot \sin(FA) \cdot \frac{1 - e^{-\frac{TR}{T_1}}}{1 - e^{-\frac{TR}{T_1} \cdot \cos(FA)}} \quad (3)$$

In this equation, GE is the signal intensity, A is a scaling factor, FA is the flip angle, and TR, TE, T1, and T2 are the repetition time, echo time, longitudinal relaxation time, and transverse relaxation time. This equation illustrates how varying parameters such as TR, and FA can influence the contrast and intensity of the MRI signal.

T1 relaxation can be manipulated in several ways. One method is to vary TR in a gradient echo sequence. By acquiring images with different TR values, the contribution of T1 relaxation to the signal can be isolated and analyzed. Short TR values enhance T1 contrast, as tissues with shorter T1 times recover faster and contribute more to the signal. Another method involves varying the flip angle. By adjusting the flip angle, the balance between T1 and proton density weighting can be controlled, allowing for selective enhancement of tissues with specific T1 values. The optimal flip angle, known as the Ernst angle, maximizes the signal for a given TR and T1. A third approach is inversion recovery, a

technique that applies an initial 180° inversion pulse before the imaging sequence begins. This pulse flips the longitudinal magnetization, and the signal is measured at various inversion times (TI) as the magnetization recovers. In this lab report we will focus on the variation of the flip angle.

T2 relaxation can be measured using spin echo sequences. In a standard spin echo sequence, a 90° pulse flips the magnetization into the transverse plane, followed by a 180° refocusing pulse. The signal decays over time due to T2 relaxation, and the intensity follows the equation:

$$SE = A \cdot e^{-\frac{TE}{T_2}} \quad (4)$$

By acquiring images at different TE values, the decay curve can be fitted to estimate T2. Multi-echo spin echo sequences, such as the Carr-Purcell-Meiboom-Gill (CPMG) technique, acquire multiple echoes within a single sequence, providing more efficient T2 quantification.

2. Materials and Methods

The experiments were conducted using a 3 Tesla (3T) MRI scanner. The high magnetic field strength of 3T scanners allows for improved signal-to-noise ratio (SNR), which enhances image quality and resolution compared to lower-field systems. A RF-body coil was used for signal transmission and an RF-head coil for signal reception. The imaging phantom used in this experiment consisted of four test tubes filled with Dotarem solutions with varying concentrations (0.5, 1, 2, and 3 mM).

The imaging sequences used were gradient echo and spin echo, each tailored for specific contrast mechanisms and relaxation properties. The gradient echo sequence, a rapid imaging method, was chosen to highlight T1-weighted contrast. In this sequence, a single radiofrequency pulse is applied, followed by gradient reversals to generate the signal. Key parameters for this sequence include the flip angle, repetition time, and echo time.

The spin echo sequence, widely used for T2-weighted imaging, has reduced artifacts caused by magnetic field inhomogeneities. In this sequence, an initial 90° RF pulse excites the spins, followed by a 180° refocusing pulse to generate a spin echo. Key parameters for this sequence include TR and TE.

First, the phantom (filled test tubes with Dotarem solutions at concentrations of 0.5, 1, 2, and 3 mM) was positioned at the isocenter of the 3T MRI scanner to minimize distortions and ensure uniform field strength. Then with the gradient echo sequence, FA was systematically varied (1, 5, 10, 20, 30, 45, 60, 75, 90deg) to observe the impact on T1-weighted contrast. Images were acquired using the GE_VarFA protocol with axial slices and consistent field-of-view (210 mm), voxel size (0,5x0,5x2 mm), TE (4,70 ms) and TR (10 ms) to allow direct comparisons. For the spin echo sequence, TE was gradually increased (TE = 15, 30, 60, 100) to measure T2-weighted signal decay.

Following data acquisition, the images were analyzed using ImageJ to extract signal intensities from regions of interest (ROIs) within each test tube. Using python, these intensities were plotted against sequence parameters, such as TR for GE and TE for SE, to calculate T1 and T2 relaxation times. Linear regression was applied to determine the relaxivity values (r1 and r2) of Dotarem by plotting the inverse relaxation times (1/T1 and 1/T2) against the concentration of the contrast agent. This process provided insights into how Dotarem modifies relaxation properties to enhance image contrast

3. Results

Experiment 2: T1 using gradient Echo MRI and variation of flip angles

Experiment two was performed by our group and therefore this data was chosen for the data evaluation.

Non-linear curve fitting was used to obtain plots of T1 with varied flip angles (Image 1) and the T1 relaxation time was calculated. The Fitted T1 Values are:

Tube1: 2.4233 s

Tube2: 3.1846 s

Tube3: 4.4932 s

Tube4: 5.8326 s

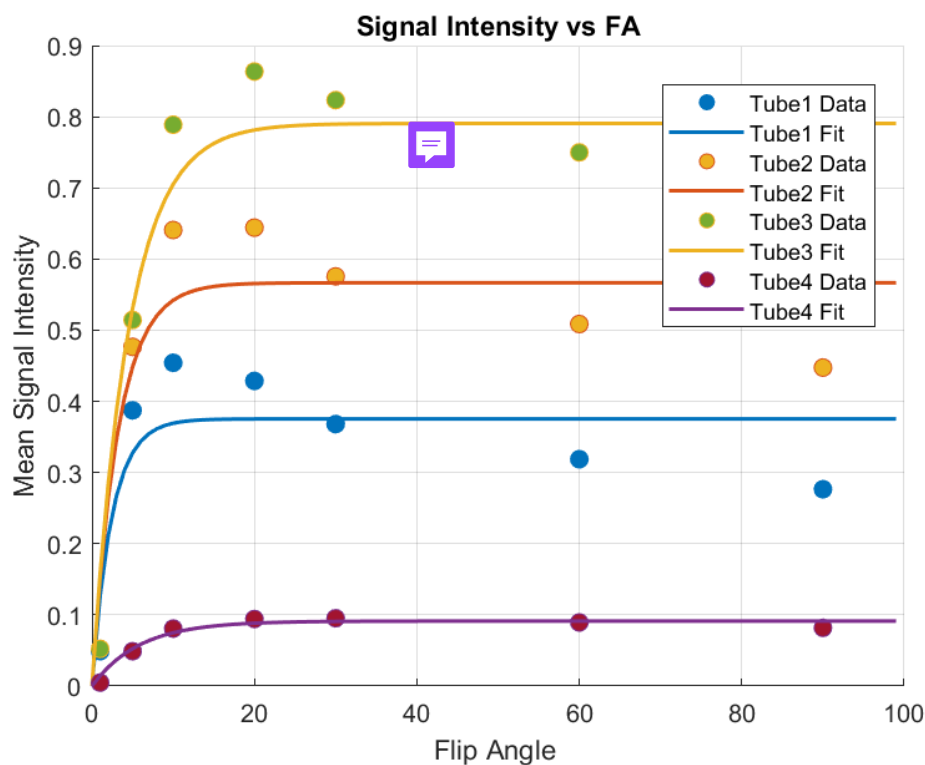


Figure 1: plot of the normalized signal intensity vs the flip angle

The signal intensity was normalized to obtain a better fit. It becomes visible that the signal intensity decays depending on the concentration in the tube. The ROI for each tube was selected once and kept consistent throughout all intensity measurements using a python script. The fitting was done in matlab. Here the measurements for FA=45° were excluded

because they are outliers.

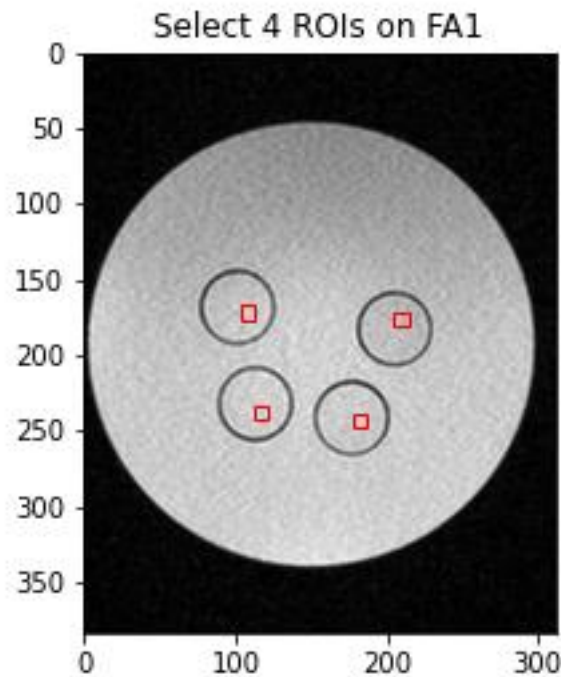


Figure 2: ROI selection for signal intensity selection

Experiment 4: T2 using spin echoes

Multi-echo (CPMG) and single SE sequences were both run to collect T2 data. CPMG scans takes less time than single SE scans, since CPMG is more efficient due to the shorter echo spacing and sampling more k-space in one repetition (Ben-Eliezer et al.). Multi-echo SE collects more measurements of T2 decay than single SE, so the former SE is more sensitive and less vulnerable to dropouts. However, multi-echo SE has worse temporal and spatial resolution as a result of the acceleration required to make the multiple measurements (Fazal et al.). One of the two sequences were further analyzed to calculate T2 relaxation time. Our group chose single SE. Data for multiple echo SE is in the supplemental section.

Using ImageJ, an ROI was placed in each tube labeled 1-4 from left to right, top to bottom (Figure 3). The concentrations for each tube are increase clockwise from the bottom left position (0.5, 1, 2, and 3 mM), so the tube order from least to greatest concentration is 3, 1, 2, then 4.

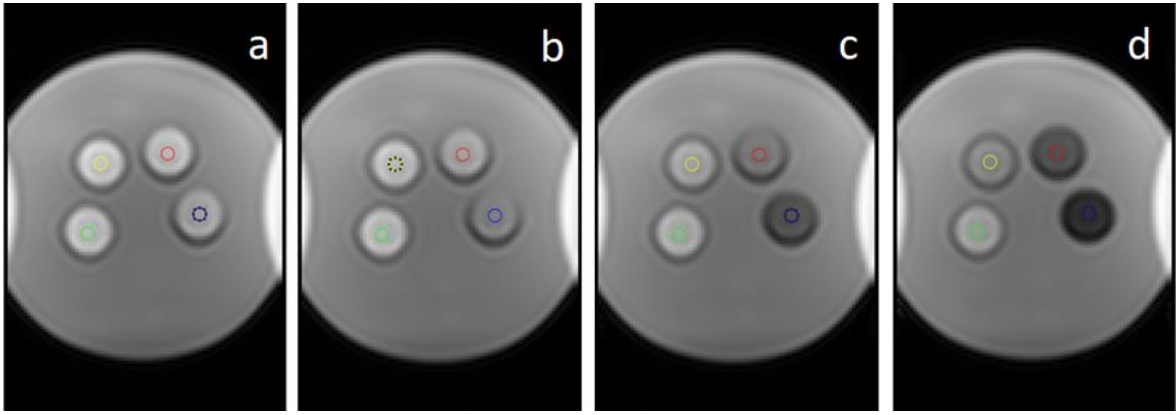


Figure 3: single spin echo protocol images with TE of 15 (a), 30 (b), 60(c), and 100 (d) ms respectively. ROIs are yellow for Tube 1, red for Tube 2, green for Tube 3, and blue for Tube 4. The TE with the best contrast between each of the tubes is at 100 ms.

	Signal Intensity at TE (s)			
Tube#	15	30	60	100
1	2656.667	2451.083	2078.417	1668.417
2	2463.167	2117.833	1546.667	1023.5
3	2491.25	2384.417	2176.083	1933.25
4	2124.167	1685.417	1057.083	565.083

Table 1. Mean Signal Intensity at varying TEs in each tube

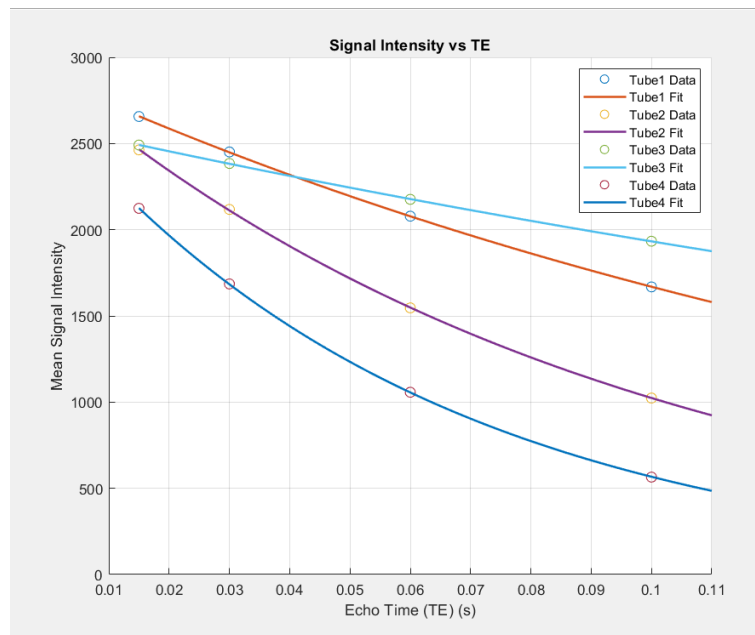


Figure 4: T2 Relaxation times for the test tubes at different TE fitted against the equation $SE = A \cdot e^{-TE/T2}$, where the MR-intensity scaling factors and T2 relaxation times for each tube are listed in Table 2.

Tube	A	Calculated T2 relaxation times (s)	Calculated T2 relaxivity, r2 (mmol-1 s-1)
1	2886	0.1827	6.549
2	2881	0.09668	10.343
3	2606	0.3343	2.991
4	2684	0.06433	15.55

Table 2. Calculated values MR-intensity scaling factor, A, and T2 relaxation times using python.

4. Discussion and conclusions

To understand the effect of varying MR sequences and sequence parameters on contrast measurement, we performed several experiments on a container with 4 test tubes with differing concentrations of Dotarem, . To understand the changes in relaxation time and rate due to the aforementioned variables, we analyzed the experiment measurements using python. Two of the experiments, 1 and 4, are covered in this report.

In Experiment 2, the overall trend of the relaxation time with varied concentration matches the expectation. The signal intensity increases until the maximum is reached and then slowly decreases again indicating that the saturation is reached. Therefore, the optimal flip angle for this setup would be between 30° and 60° to reach maximum intensity. Overall, it becomes clear that an increased concentration of contrast agent leads to a longer relaxation time.

In Experiment 4, a graph was generated using the data from single spin echo measurements of signal intensity in the four tubes with different concentrations at different echo times. Figure 4 demonstrated close fit of the experimental values to the expected exponential decrease of signal intensity, with tubes 3 and 4 showing the lowest and highest exponential decreases respectively. The calculated r2 increases with increasing concentrations (tubes 3, 1, 2, then 4). One analytical error that may have led to the measurements of tube 3 to be

lower than expected (eg. the signal intensity for tube 3 is lower than tube 1 in the shortest TE measurement) is that the ROI for tube 3 is placed further from the center of the tube. The variation in ROI placements could have been better accounted for by creating larger ROIs for each tube, which would prevent skewing from the signal intensity variation within each tube.

T1 is expected to be larger than T2; the values in our experiment supports this.

The relaxivity measured in vivo would differ from in vitro measurements because the liquid that the contrast agent would be dissolved in would have very different properties, which would change the offset. The difference in temperature between in vivo and in vitro would also lead to differences in relaxivity, as human body temperature (37°C) is significantly higher than the temperature our experiments were performed at (not measured). The literature suggest that MRI contrast behaves differently in corpses depending on their temperature suggesting this might also influence phantom measurements. (Ruder et al.)

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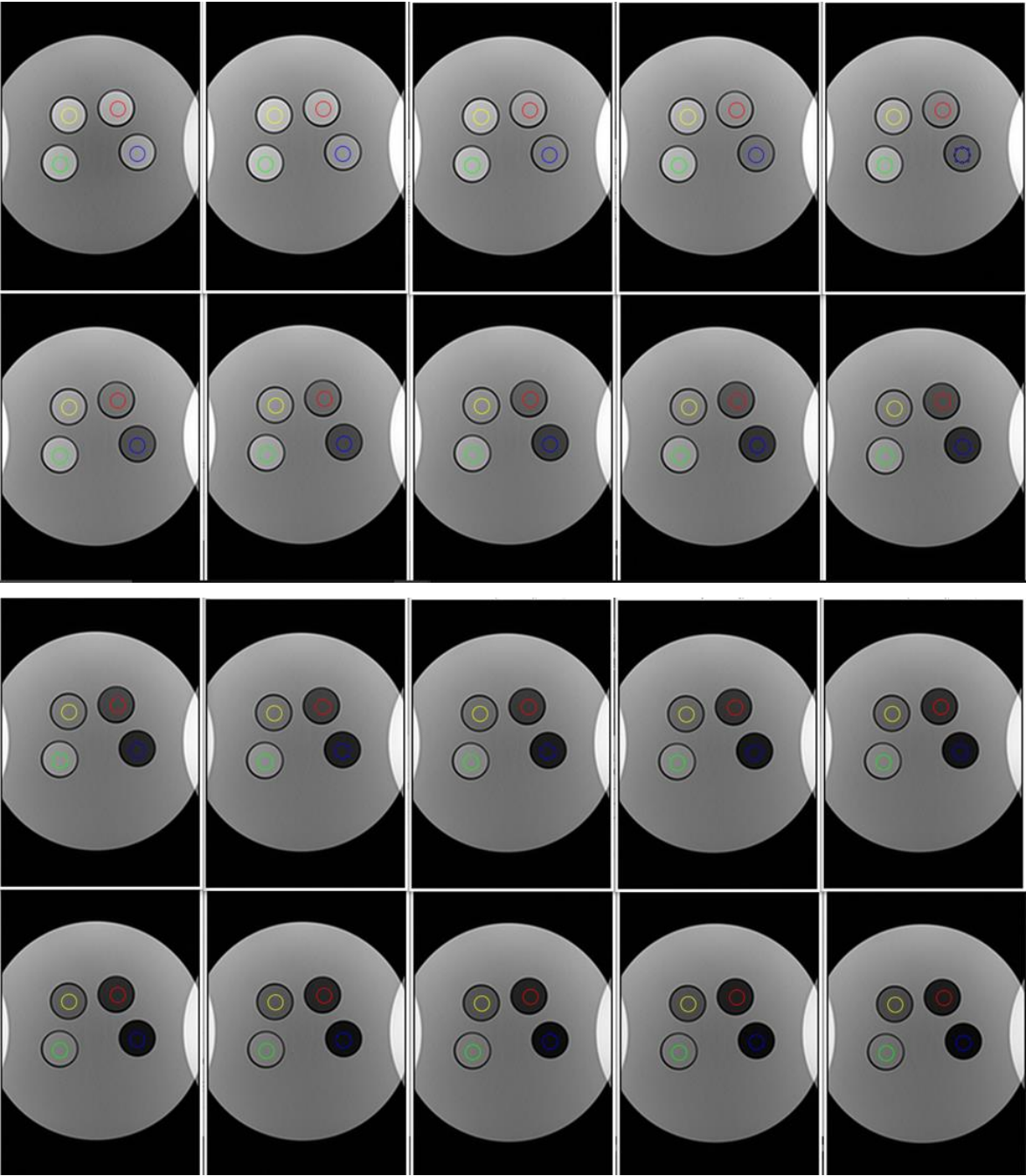
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6. Supplemental



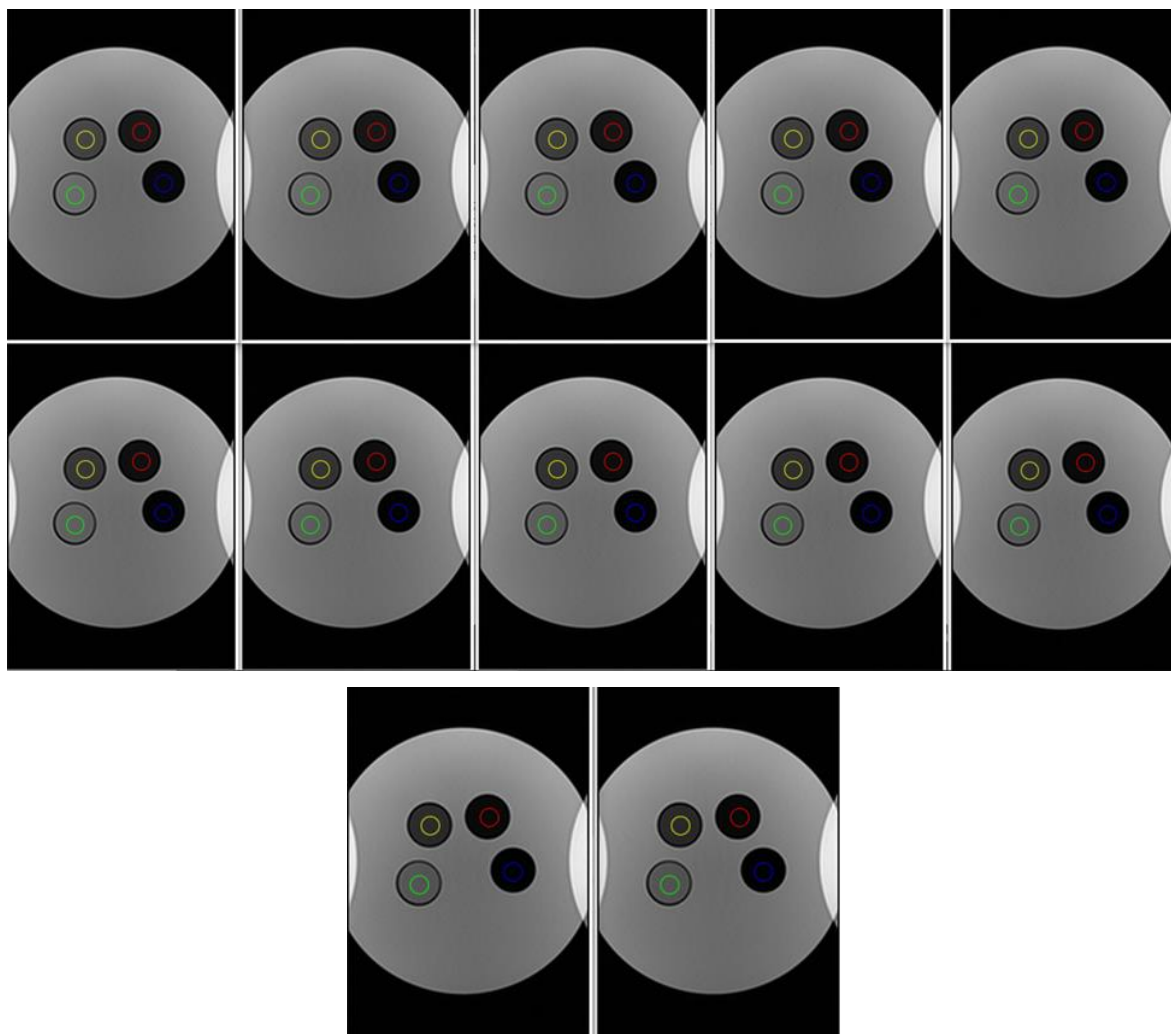


Figure S1. Multi-echo SE images for varying TE from 15 to 228 ms.