

A REVIEW ON EXPERIMENTAL FINDINGS ON PROTON RELAXATION IN SEVERAL LABORATORY MAGNETIC RESONANCE IMAGING (MRI) PHANTOMS

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ABSTRACT

Alternative materials that are potentially useful for magnetic resonance imaging (MRI) phantom have been studied to investigate their proton relaxation properties. The laboratory prepared materials should first fulfil characteristics such as inexpensive, non-toxic, easy to handle, easily prepared, stable over a long period of time and should possess properties of tissues. Two important quantities of interest that were investigated are the longitudinal (T_1) and transversal (T_2) relaxation times. In this work, the SNR relaxations of three laboratory-made MRI phantoms which were agar gel, agarose gel and poly[vinyl] alcohol (PVA) slime were measured using a 3-T MRI system. A standard spin-echo pulse sequence was used. The data obtained fulfilled the exponential behaviour of the form $\text{SNR} \propto 1 - e^{-TR/T_1}$ for T_1 curve and of $\text{SNR} \propto e^{-TE/T_2}$ for T_2 curve. By using a curve-fitting method, it was found that the addition of relaxation modifier has significantly shortened the T_1 relaxation time of the pure agar, agarose and PVA slime. Their T_2 relaxation time was less affected. This indicated the sensitivity of the longitudinal relaxation of the protons to changes in their surroundings (or lattice). The shortening of T_1 relaxation time has brought these laboratory-made MRI phantoms closer the properties of a standard MRI water phantom.

Keywords: lattic;, magnetic field; saturation; spin; SNR uniformity

INTRODUCTION

Proton relaxation is a phenomenon experienced by proton spins when they release their energy to the surrounding or lattice [1]. It is also the effect that resulted from the interaction among the proton spins [2]. In magnetic resonance imaging (MRI), energy is gained by the proton spins during the delivery of the radiofrequency (RF) field [3]. At a specific frequency determined by the tissue-specific properties, the precession of the proton spin is said to be at resonance with the oscillation of the RF field, hence magnetic resonance [3]. The termination of the RF field thus left the protons at a non-equilibrium high-energy condition resulting in an energy tranfer from the proton spins

to the lattice. Proton spins relaxation occurs during this energy transfer and is known as spin-lattice or longitudinal or $T1$ relaxation [1]. In the same time interval as $T1$ relaxation, spins interact among them giving rise to another type of proton spins relaxation known as spin-spin or transversal or $T2$ relaxation [2]. As opposed to the energy being transferred to the lattice in $T1$ relaxation, the energy is being exchanged among spins in $T2$ relaxation [4].

MRI phantoms are designed to possess the properties of human tissues [5]. Human tissue mimicking materials are gaining increasing attention in this field of research [5, 6]. Human tissue mimicking materials should be easy to handle, cost-effective, robust, non-hazardous and stable and must be able to test the accuracy, stability and comparability of MRI scanners [7, 8]. Water phantom is easy to handle [9] but it is also affected by vibration, producing air bubbles that can cause non-uniformity. Another concern when using a water phantom is the relaxation time of water that is much longer than human tissue [9]. Materials that have relaxation times closer to human tissue were gaining more attention. A phantom made from human tissue mimicking material is reliable as its $T1$ and $T2$ relaxation time can be modified [1, 3] to fall in a range possessed by most human tissues, which makes the experimental result closer to the result performed on human subjects which can be utilized as biomarkers [10].

This study used a 3-T MRI system to investigate several MRI phantom materials fabricated in the laboratory. The effects of the presence of relaxation modifier on the $T1$ and $T2$ relaxations are studied. The present MRI phantom study rests on three main objectives; (i) to measure the intensity of the MRI phantom images obtained from $T1$ and $T2$ measurements, (ii) to calculate the signal-to-noise ratio (SNR) of the images and (iii) to determine the $T1$ and $T2$ values of the phantoms by fitting the experimental data to the relaxation models. It was hypothesised that water molecules play important roles in proton relaxations. Due to the lattice-dependent nature of longitudinal relaxation, the use of relaxation modifier is useful only for $T1$ relaxation.

EXPERIMENTAL DETAILS

Materials

All phantoms reported in this paper were prepared by means of conventional methods and were published elsewhere; see Tee et al. [11] for poly(vinyl) alcohol (PVA) slime phantom with different PVA/borax ratio, Yee et al. [1] for poly(vinyl) alcohol slime phantom with different relaxation modifier concentrations, Azhar et al. [3] for agarose gel phantom with different gadolinium oxide concentration as relaxation modifier, Er et al. [12] for Agar Gel Phantom with Iron (III) Oxide as Relaxation Modifier and Tee et al. [13] for poly(vinyl) alcohol (PVA) MRI slime phantom with different PVA/borax ratio.

Magnetic Resonance Imaging Scans

The MRI data were acquired using Siemens Magnetom Verio 3-Tesla MRI system at the Department of Radiology, Hospital Canselor Tuanku Muhriz. The plastic container containing phantoms and distilled water (Figure 1(Left)) were inspected for leakage and

any presence of air bubbles and foreign particles before being scanned. They were then positioned horizontally in the 32-channel head coil by using a home-made sample holder at the iso-center of the magnet bore (Figure 1(Middle)). All the phantoms were scanned simultaneously. The spin echo (SE) and turbo spin echo (TSE) sequences were applied to obtain the axial $T1$ and $T2$ weighted images respectively. $T1$ weighted images were obtained using a constant TE = 15 ms and TR = 100, 150, 200, 300, 600, 800, 1000, 1200, 1800, 2400, 3600, 4800, 5600, 6400 and 8000 ms. The $T2$ weighted images were obtained using a constant TR = 4970 ms and TE = 69, 82, 96, 110, 123, 137, 151, 165, 178, 192, 206, 220, 233, 247 and 261 ms. Example images are shown in Figure 1(Right).



Figure 1: MRI phantoms (Left) when positioned horizontally inside the RF head-coil (Middle) and the images obtained (Right)

Other imaging parameters include slice thickness = 3.0 mm, matrix size = 64×64 , field of view (FOV) = 230×230 mm and voxel size = $0.6 \times 0.4 \times 3.0$ mm. All the images acquired were stored in a CD-ROM in DICOM format.

Data Analysis

Image-J software (The National Institutes of Health, NIH) was used to measure the signal-to-noise ratio (SNR) for all phantoms. The middle slice from each volume measurement of the phantoms was chosen for SNR evaluation. The image selected was visually ensured to have a spatially homogenous distribution of signal intensity [11]. From the acquired images, the mean signal intensity of the phantom (I_p), signal intensity of the background noise (I_b) and standard deviation of the background noise (σ_b) were computed. Three circular ROIs with the size of 5 mm^2 were specified on the phantom images and I_p were obtained and averaged. Another circular ROI with the size of 40 cm^2 was drawn at the background, outside the phantom images in order to obtain I_b and σ_b . The SNR value for each phantom across all TEs and TRs were then calculated using the formula $\text{SNR} = (I_p - I_b) / \sigma_b$.

The $T1$ and $T2$ values of the phantoms were determined based on the SNR vs TR plot ($T1$ curve) with the equation of $\text{SNR} \propto 1 - e^{-\text{TR}/T1}$ and SNR vs TE plot ($T2$ curve) with the equation of $\text{SNR} \propto e^{-\text{TE}/T2}$ respectively [11]. The experimental data were fitted to both the equations using MATLAB® R2018b (The Math Works, Inc., Natick) curve

fitting toolbox. The $T1$ and $T2$ values were obtained upon achieving the minimum sum of squared difference between the observed and fitted data. Likewise, by referring to the fitted $T1$ curve, the saturation value (SNR_0) was also determined. The 95% confidence interval (CI), standard sum of errors (SSE) and the R^2 of fit were also computed. The R^2 of fit is the correlation coefficient between the observed and fitted data and the ideal value of R^2 is 1. The $T1$ and $T2$ curves in this study were fitted in such way that the R^2 of fit is maximised and the SSE is minimised.

RESULTS AND DISCUSSION

Baseline measurements were conducted on distilled water and standard MRI phantoms as controls. Comparing distilled water phantom with standard MRI phantom, a shorter $T1$ relaxation time was obtained for standard MRI phantom; $T1_{\text{distilled-water}} = 5051$ ms (TE = 15 ms) [1], $T1_{\text{MRI-phantom}} = 419$ ms (TE = 10 ms) [4]. The reduction in $T1$ values for MRI phantom was contributed by the presence of relaxation modifier usually CuSO_4 – based additive added to the standard MRI phantom. $T2$ relaxation was barely influenced by the relaxation modifier. The $T2$ values for the distilled water and MRI standard phantoms were 246 ms (TR = 4970 ms) [1] and 202 ms (TR = 4970) [4] respectively. Furthermore, the standard MRI phantom also produced a higher $T1$ SNR at any TE then the distilled water phantom while the $T2$ SNR at any TR is incomparable for both [1, 4].

For poly(vinyl) alcohol (PVA) MRI slime phantom with different PVA/borax ratio [13], $T1$ is always longer than $T2$ irrespective of borax proportion. The average $T1$ value (3423 ms) was about 3.4 times the average of $T2$ value (276 ms). The $T1$ and $T2$ values of PVA slime phantoms are similar to human tissues e.g. synovial fluid and blood respectively [14]. Both the $T1$ and $T2$ values increase slightly after borax proportion is increased beyond 3. An increase in $T1$ or $T2$ values means that the $T1$ and $T2$ curves relaxed slower and the curve are shifted to the right. A longer $T1$ relaxation time means that the oscillation frequency of the spins in the surrounding molecules is much slower or faster than the Larmor frequency of the spins [4]. More time is needed for the spins to transfer their energy to the surrounding [4]. A longer $T2$ relaxation means that the spins stay in precession (minimal relaxation) a little bit longer. A comprehensive discussion on $T1$ and $T2$ relaxations of the PVA slime phantom based on dipole-dipole interaction can be found in our separate communication [4]. Recent findings on the use of $T2$ weighted images to assess inhomogeneity [2] indicate the importance of similar study to be conducted in the future. Figure 2 shows the $T1$ and $T2$ curves for different PVA to borax ratio.

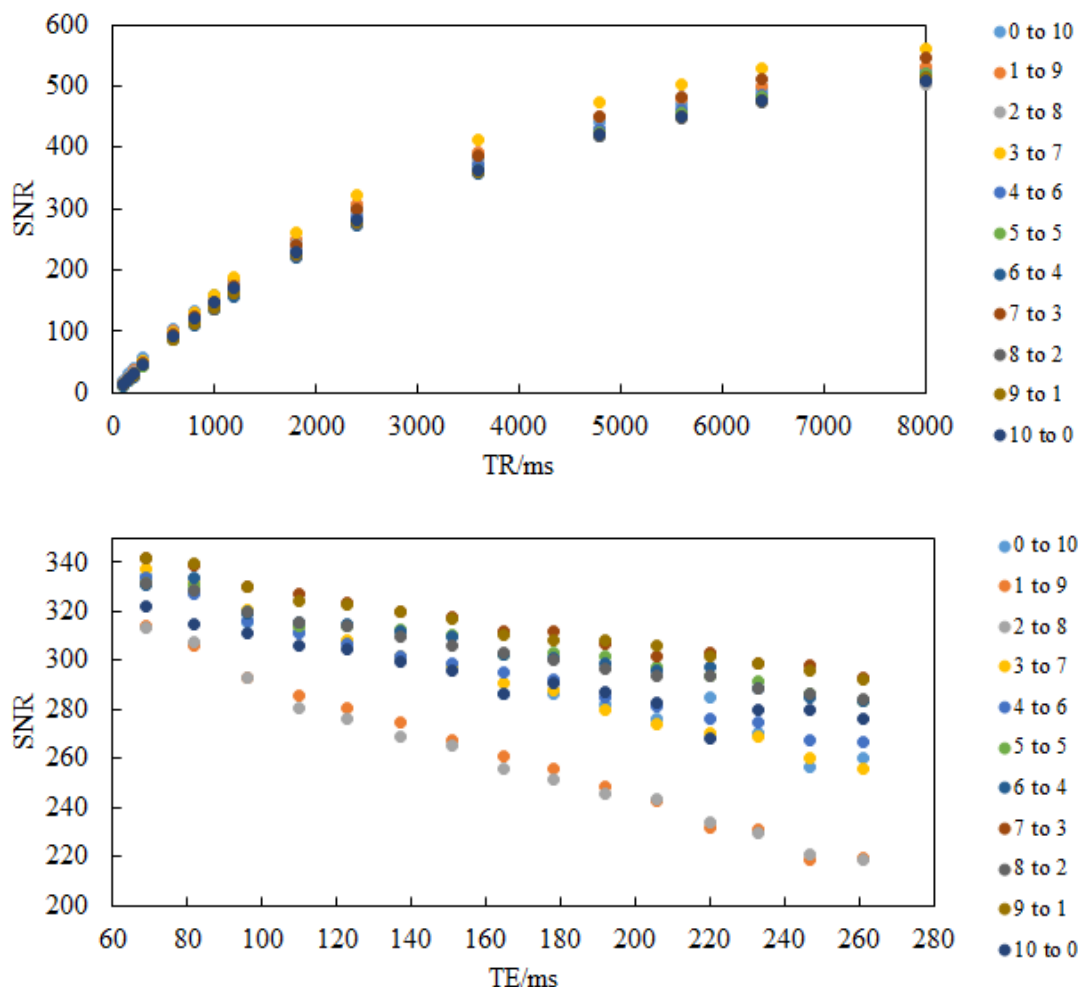


Figure 2: T_1 (top) and T_2 (bottom) relaxations for several content ratios of PVA to borax

In a separate study [4], the prepared PVA slime phantoms were scanned at four different time points (TP). The T_1 and T_2 values were found to be stable over time and in good agreement with the average values of human tissues. The results are promising at least in the context of phantom homogeneity and stability. Further studies are necessary in order to obtain suitable slime phantoms for MRI in actual scale which homogeneity and stability would withstand a longer period of time.

For PVA slime phantom with gadolinium oxide as relaxation modifier, similar to the comparison results between distilled water and MRI standard phantoms, relaxation modifier has significant effect in reducing T_1 relaxation time [1]. T_2 relaxation barely changed except for intensity fluctuation. T_1 value of the PVA slime phantom changed markedly in the presence of relaxation modifier while the change in T_2 values for all phantoms is smaller as compared to the change in T_1 value, see Figure 3 (Top) and (Bottom). Gd was found to affect the T_1 but not the T_2 curves. T_1 and T_2 relaxation times decrease with increasing Gd_2O_3 concentration but the effect is more prominently on T_1 . It can be concluded that with a systematic increase in the quantity of relaxation

modifier, the $T1$ and $T2$ relaxation times decrease in a systematic manner. Further studies are necessary to investigate the stability of the PVA slime phantom over a long period of time.

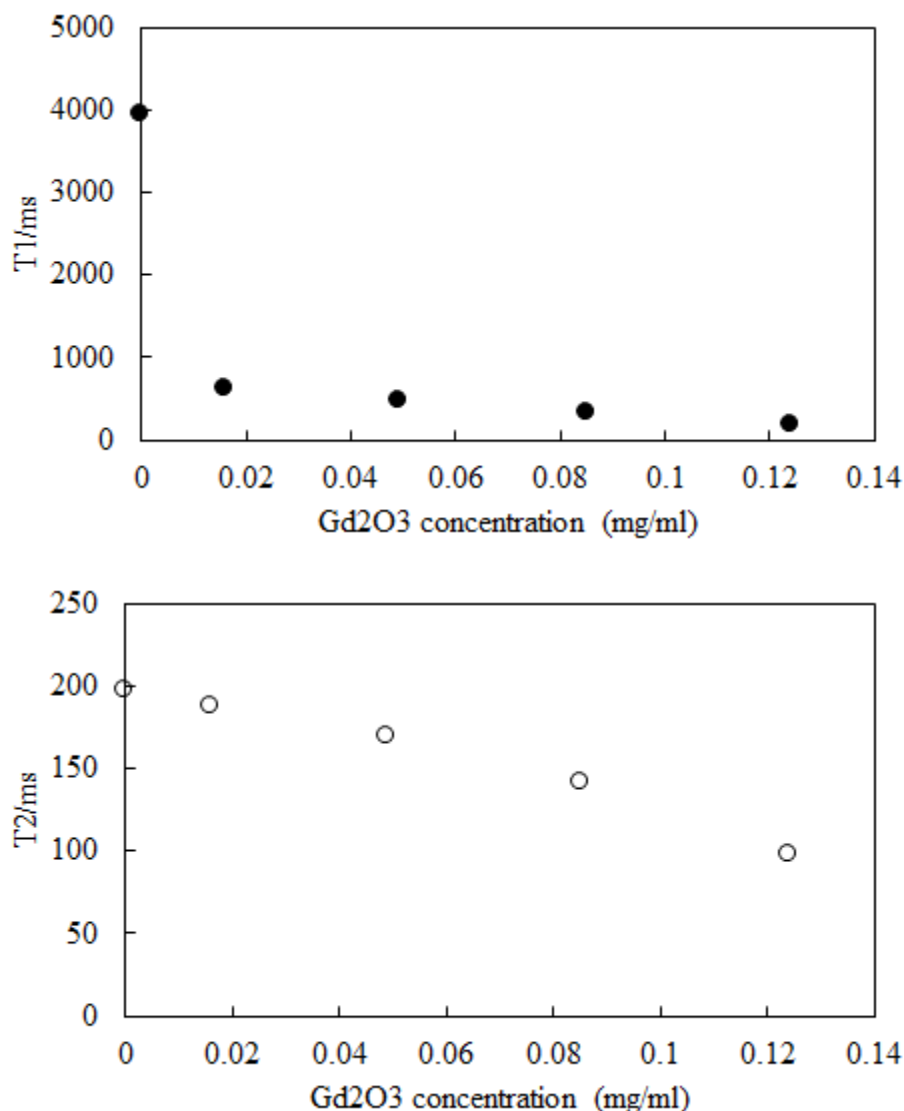


Figure 3: Change in $T1$ (Top) and $T2$ (Bottom) values of the PVA slime phantom as Gd_2O_3 is added

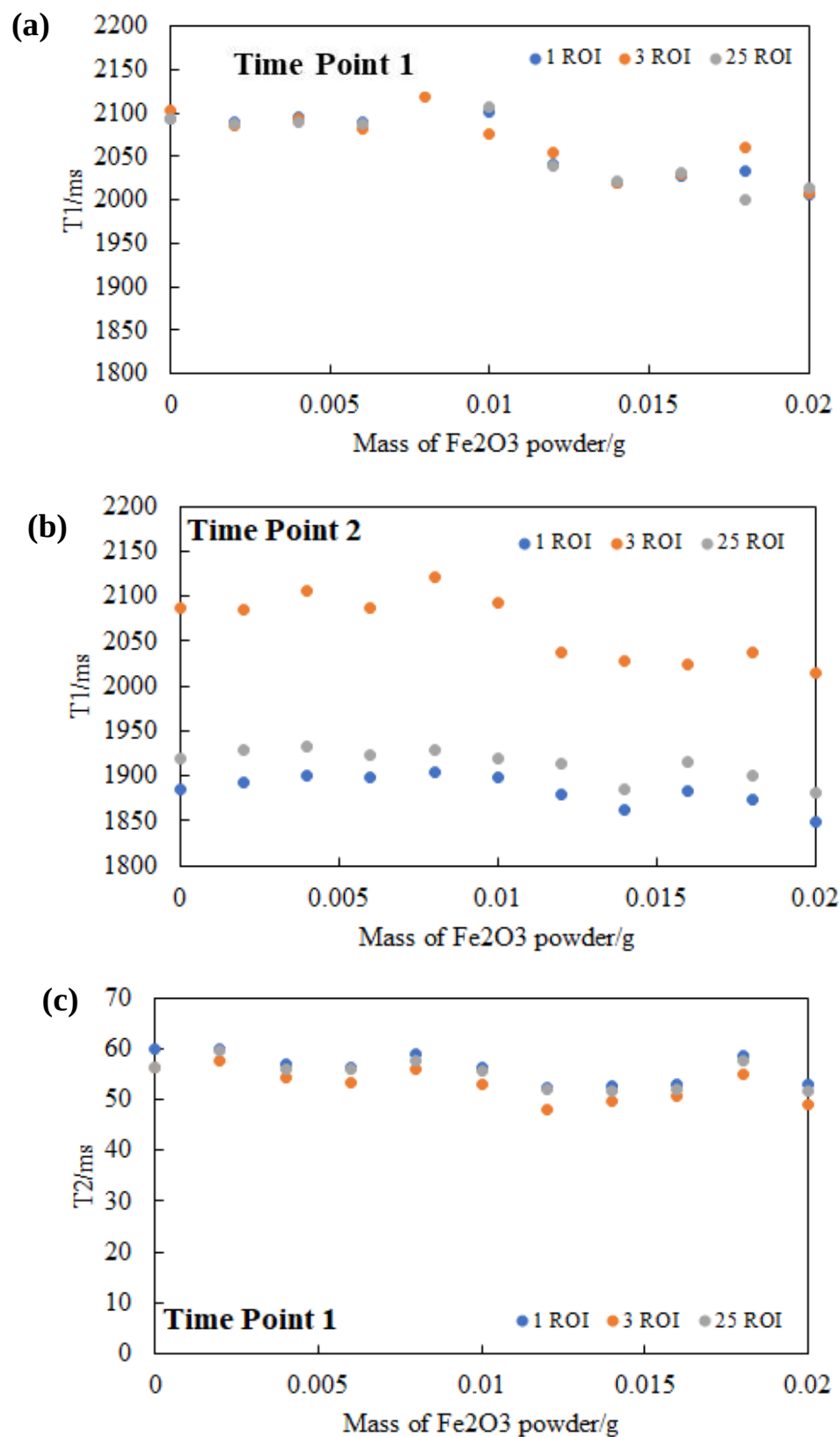
For the agarose gel phantoms [3], the addition of Gd_2O_3 into the matrix of agarose gel phantom reduced the $T1$ relaxation time significantly and stabilised the signal intensity in a wide range of TR values. While Gd_2O_3 has significant effects on $T1$ relaxation parameters as well as on the saturation of signal intensity, $T2$ relaxation parameters were minimally affected [3]. Gd_2O_3 was found to successfully act as the relaxation modifier for the $T1$ but not the $T2$ curves. The $T1$ curve started to show saturated SNR (SNR_0) and increasing SNR_0 for $TR > 1000$ ms and $[Gd_2O_3] = 0.005$ g/ml or higher.

These behaviours are explained based on the dipole-dipole interaction that increases in phantoms with higher $[\text{Gd}_2\text{O}_3]$, thus shortening the $T1$ relaxation. However, a systematic change in the $T2$ parameters with increasing $[\text{Gd}_2\text{O}_3]$ was not observed. While Gd_2O_3 has significant effects on $T1$ relaxation parameters, the $T2$ relaxation parameters were minimally affected [3]. With a shorter $T1$, the Gd_2O_3 added agarose gel can potentially be used as test phantom in fast imaging sequence, e.g. gradient echo pulse sequences.

For agar gel phantoms with Iron (III) Oxide as relaxation modifier, pure agar gel shows significant different in $T1$ and $T2$ relaxation as compared to those containing relaxation modifier [12]. The $T1$ and $T2$ of the agar gel phantoms were minimally affected although there was a systemic increase in the mass of the iron oxide powder (Figure 4). This can be observed especially for $T2$. Comparing between different ROIs, only significant different in $T1$ was observed which occurred at TP2 (Figure 4). Comparing between Time Point 1 (TP1) and Time Point 2 (TP2) of 2 weeks difference, agar gel phantoms exhibited the characteristics of SNR uniformity but they showed instability of SNR at TP2 [12]. It can be concluded that iron oxide in powder form is not an effective relaxation modifier to reduce the $T1$ and $T2$ when it is introduced into the agar gel phantoms. For further studies using iron oxide as the relaxation modifier, nano-sized particles should be used and the particles should be dissolved properly before being added to the agar gels.

CONCLUSION

In conclusion, the shortening of $T1$ relaxation time of lab-based phantoms can be achieved with the addition of relaxation modifier. However, the evidences collected so far show that the relaxation modifier has a prominent influence over the $T1$ relaxation but not on $T2$ relaxation. It was also found that the relaxation modifier (such as metal oxide) in powder form would not be able to reduce the relaxation time. Metal oxides powders must be dissolved in order to obtain a desired shortening of relaxation time. The difference effects of the size of the relaxation modifier powder grains on the relaxation of the MRI phantoms are still being investigated. Future studies should be focused on developing a prototype phantom for relaxation studies.



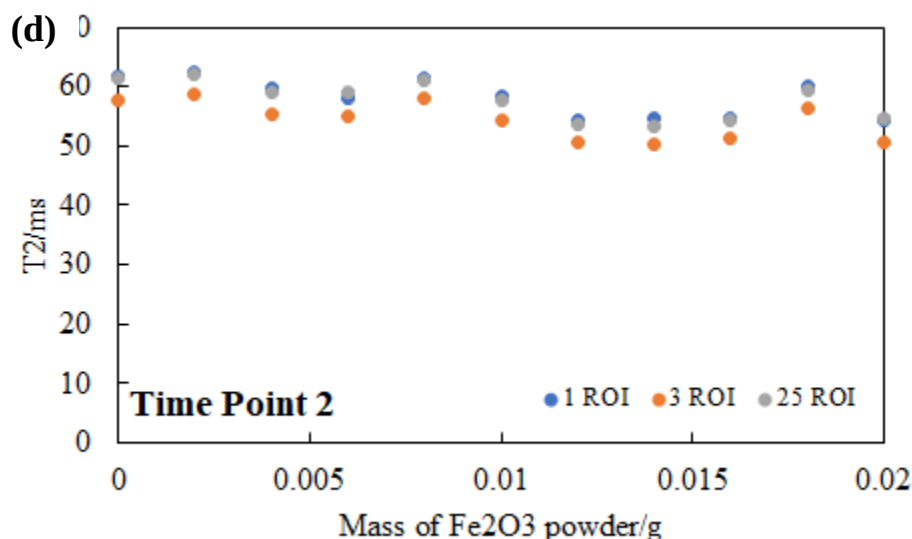


Figure 4: (a) T_1 at Time Point 1, (b) T_1 at Time Point 2, (c) T_2 at Time Point 1 and (d) T_2 at Time Point 2 for agar gel phantoms with different masses of Fe₂O₃ content calculated from three different ROIs

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