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A MULTIPURPOSE MRI PHANTOM BASED ON A REVERSE MICELLE SOLUTION

Jo E. Roe

May 1996

A thesis submitted in partial fulfillment of the
requirements for the degree of
Master of Science of Chemistry.

Approved: Joseph Hornak
Thesis Advisor

Gerald A. Takacs
Department Head

Department of Chemistry
Rochester Institute of Technology
Rochester, NY 14423-5603

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A MULTIPURPOSE MRI PHANTOM BASED ON

A REVERSE MICELLE SOLUTION

Abstract

by

Jo E. Roe

Magnetic resonance imaging (MRI) phantoms are anthropogenic objects used for system performance testing of an MR imager. Current phantoms used today contain aqueous solutions of paramagnetic salts. When imaged, these phantoms produce a standing wave artifact. The presence of this artifact make it difficult to distinguish between inhomogeneities in the B_1 transmit and receive field of an imaging coil and the imaging phantom. This thesis describes a reverse micelle (RM) solution and its applicability for use in an MR imaging phantom. The RM solution consists of three components: H_2O , which is surrounded by the surfactant, AOT, to form a droplet, which is suspended in the hydrocarbon, and decane. Volume fractions (ϕ) of water and AOT from 0.50 to 0.90 were studied. The RM solution is easy to make up and is homogeneous. The RM solution was found to be phase-stable at temperatures ranging from 0 to 40°C at certain ϕ values. The dielectric constant of the RM solution was half the dielectric constant of H_2O . The resistivity of the RM solution was 5 times greater than an aqueous solution of 6 mM NiCl and 0.154 mM NaCl. The diffusion coefficient of the H_2O and AOT component increased as ϕ increased. The diffusion coefficient of decane decreased as ϕ increased. The viscosity of the RM solution increased as ϕ increased. The viscosity of the RM solution is 35 times greater than the viscosity of H_2O . The RM solutions containing 0.10, 0.25, and 0.50 mM Mn^{+2} in the aqueous phase produce overall T_1 and T_2 values similar to human tissue. The addition of either 3, 5, or 8 mM Ni^{+2} in the aqueous phase of the RM solution produces T_1 and T_2 values useful for system performance testing. The RM solution containing 8 mM Ni^{+2} in the aqueous phase displayed the least temperature dependent T_1 and T_2 of all six RM solutions studied. The RM solution phantom is effective in B_{1T} and B_{1R} field mapping. An image of the RM solution phantom displays minimal standing wave artifact and no skin-effect artifact. The spectrum of the RM solution phantom contains two peaks which correspond to the fat and H_2O

spectral peaks of human tissue. Consequently, the RM solution phantom could be used to test the fat saturation imaging sequence. Spectral T_1 values of the H_2O and decane component were measured. The individual T_1 values differed from the overall T_1 values of the RM solution, thus, studying the effects of magnetization transfer between the three components could be possible.

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This one is for you, Mom.

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List of Symbols and Abbreviations

AOT	aerosol OT
B_0	static magnetic field
B_1	applied magnetic field
B_{1R}	applied receive magnetic field
B_{1T}	applied magnetic transmit field
c	speed of light in a vacuum
CFA	center frequency adjust
[A]	concentration of species A
cp	centipoise
C_t	average capacitance
CI	95% confidence interval
CV	coefficient of variation
D	density
ϵ	permittivity
FID	free induction decay
γ	gyromagnetic ratio for H proton
G_ϕ	phase encoding gradient
G_f	frequency encoding gradient
G_s	slice selection gradient
GE	General Electric
HPLC	high performance liquid chromatography
ID	inside diameter
I_r	reflected incident light
J	joules
κ	dielectric constant
$k_{(1,2,3,4,5,6,,7,8)}$	spin exchange constant
K	degrees Kelvin
λ	wavelength
l	cell constant
L	inductance
L_p	probe inductance
μ	permeability
M	Molarity expressed in moles/liter
Mhz	megahertz
Mn^{+2}	manganese +2 ion
MW	molecular weight
MRI	magnetic resonance imaging
M_{xy}	magnetization in xy plane
M_z	magnetization vector in z direction
M_{z0}	initial magnetization vector in z direction
ν	Larmor frequency
ν	resonant frequency
ν_{ref}	standard frequency
$\Delta\nu$	change in the frequency of the bandwidth
η	index of refraction
η	viscosity
n	number of dipoles in a volume (ν)
Na^{+2}	sodium +2 ion
$N_{antiparallel}$	number of spins aligned aniparallel to magnetic field
Ni^{+2}	nickel +2 ion
NLLS	nonlinear least squares

NMR	nuclear magnetic resonance
N_{parallel}	number of spins aligned parallel to magnetic field
NS	normal saline solution
OD	outside diameter
ϕ	volume fraction of $\text{H}_2\text{O} + \text{AOT} / \text{H}_2\text{O} + \text{AOT} + \text{decane}$
ϕ	shorted cell phase
q	charge of an ion
Q factor	quality factor
Q_l	net dipole moment in volume (v)
Q_{coil}	quality factor in an imaging coil
Q_{loaded}	quality factor of a loaded imaging coil
Q_{sample}	quality factor of a sample
ρ	resistivity
P	polarization
R	resistance
r	distance between two charges (q)
RF	radio frequency
RG	receive gain
RM	reverse micelle
RPS	revolutions per second
δ	chemical shift
δ	skin-depth
slice thk	slice thickness
SNR	signal-to-noise ratio
T	Tesla
T_1	spin-lattice relaxation time
T_2	spin-spin relaxation time
T_2^*	inhomogeneous spin-spin relaxation time
TE	time of echo
TG	transmit gain
TR	time of repetition
V	volume
v	velocity
ω	angular frequency
Z	impedance of cell
z	magnitude of z from impedance meter

1.0 Introduction

The imaging modality called magnetic resonance imaging (MRI) is based on the principles of nuclear magnetic resonance (NMR) [Stark]. While NMR has been widely used by the chemist for the past fifty years, MRI is a relatively new technique in the field of medicine. NMR was first discovered by Bloch and Purcell independently in 1940. In 1952, they received the Nobel prize for this discovery. This work was used by chemists in a variety of applications until 1971, when Damadian found that an NMR parameter called the spin-lattice relaxation time, T_1 , was not the same for healthy and diseased tissue. In 1973, Lauterbur applied linear magnetic field gradients to obtain spatial information, which allowed an image of an object to be produced similar to images produced by computerized tomography [Kean]. These developments made the medical community aware of the importance of this imaging technique for the diagnosis of disease in the human body. MRI is a radiationless, nondestructive, and noninvasive technique, which has growing applications in the diagnosis of different diseases in the human body.

Since MRI is used as a diagnostic tool to produce high quality anatomical images of the human body, the imager, or the instrument which provides the images must be maintained periodically. This is done to ensure that the magnetic fields and gradients are operating optimally to generate accurate and clear images. Imaging phantoms, which are anthropogenic objects, are employed as standards for system performance testing.

The most common phantoms currently used for system performance testing contain aqueous solutions of paramagnetic salts. This type of phantom solution produces a standing wave artifact when imaged. The presence of this artifact makes it difficult to test the homogeneity of the transmit and receive radio frequency fields of an imaging coil. The thrust of this research was to determine the cause of the standing wave artifact and to develop a phantom solution which contained properties that would minimize or eliminate the standing wave artifact.

As previously mentioned, phantoms are anthropogenic objects which are used for system performance testing. The ideal phantom solution should possess the following properties. The solution inside the phantom should be easily prepared, homogeneous, and stable over a range of temperatures. To minimize the standing wave artifact, the solution should have a lower dielectric constant than H_2O .

[Tofts]. The solution should have a high resistivity to eliminate the *skin-effect* artifact [Bottomley], and a high viscosity to prevent convection currents from forming in the solution inside the phantom. The phantom solution should have spin-lattice relaxation time (T_1) and spin-spin relaxation time (T_2) values which could be varied and are reproducible. Varying the T_1 and T_2 values of the solution would allow ranges which are similar to human tissue or a range useful for system performance testing. The T_1 and T_2 's of the solution should be reproducible from one imaging site to the next regardless of temperature fluctuations.

Past studies of phantoms involve using acrylamide gels, hydrogels, agar, and aqueous solutions of paramagnetic salts [De Luca, Gore]. Each of these types of phantom solutions have disadvantages. In organic materials, indirect spin-coupling, or J-coupling produces echo modulation which affect the intensity of the signal in an image [Hinks]. The standing wave artifact is present in the phantoms containing an aqueous solution of paramagnetic salts [Tofts]. Other disadvantages of these solutions are their lack of stability, difficulty in preparation, and impurities. These solutions also may be conductive or have low resistivity which produces a *skin-effect* artifact.

We have attempted to design a phantom solution which contains many of the properties of an ideal phantom solution. The phantom solution that we propose is a reverse micelle (RM) solution. The RM solution is an example of a three component microemulsion. The H_2O component of the solution is surrounded by the surfactant component, which is suspended in the third component, a hydrocarbon. Determination of the applicability of the RM solution for use as an MR imaging phantom included studying the stability of the RM solution, its dielectric constant, resistivity, and viscosity. Further characterization of the behavior of the RM solution included studies determining the diffusion coefficients of the individual components, the overall T_1 and T_2 values at various temperatures, and spectral T_1 values of the individual components.

Presented in this thesis is the background theory for NMR and MRI, as well as background material about the RM solution. The following section describe the experiments that were performed to determine the physical, electrical, NMR, and MRI properties of the RM solution. The physical properties are characterized by stability and phase studies, and viscosity and diffusion measurements. The electrical

properties of the RM solution include the dielectric constant, resistivity, and conductivity measurements as well as the ability of the solution to load the imaging coil, which is called the Q factor. The NMR properties of the RM solution include the magnetic susceptibility of the solution and spectra of the components of the RM solution. Also included in the NMR section is the overall T_1 and T_2 studies and how the T_1 and T_2 values are affected by temperature. The spectral T_1 values of the individual components of the RM solution conclude the NMR section. The MRI properties of the RM solution contain the transmit and receive field mapping and the fat-saturation imaging sequence studies. The results and discussion section are organized in a similar manner as the experimental section. A summary of the results follows and contains the last section of this thesis.

2.0 Background

This background section contains the theory which is essential for understanding the results of this thesis. It comprises six main topics; spin physics, nuclear magnetic resonance (NMR), magnetic resonance imaging (MRI), the MRI phantom, artifacts in MR images, and the reverse micelle (RM) solution. Spin physics relates NMR to the atomic level, while the MRI section applies the theory of NMR to produce an image and explains imaging parameters for use in the experiments. The MRI phantom section explains the purpose and construction of a phantom, and the problems which can occur with phantoms are explained in the artifact portion. Finally, the RM segment presents information concerning the RM solution as well as reasons it could be useful as an MRI phantom solution.

2.1 Spin Physics

In nuclear magnetic resonance, the signal originates from the spins in the nucleus of the atom and its interaction with a magnetic field. From a rather simplified view, the proton and neutron possess a fundamental property of nature called spin or intrinsic angular momentum. The origin of the spin is controversial and not well understood [Jaffe] and will not be discussed in this thesis. It is difficult to define what exactly spin is; therefore, it is most often defined in terms of the behavior it exhibits in an externally applied field [Schwartz]. A classical analogy found in many textbooks compares the spin angular momentum to a spinning top in a gravitational force field. The angular momentum of a spinning top will not change as long as there is no torque acting on the system. The top continues to spin about its vertical axis. Once a torque or external field, such as, gravity, is applied, the angular momentum of the top is no longer constant. The angular momentum continually changes in the direction of the torque [Kotlarchyk, 1995]. The spinning top no longer spins about its axis, but will spin in a cone-shaped path. This phenomena of following a cone-shaped path is termed precession [Chakeres].

In the nucleus, the protons and neutrons pair up or align in the nuclear orbitals in a manner similar to electrons in electronic orbitals. When the spins of individual nucleons pair up and completely cancel each other, there is no angular momentum, or the nuclear spin is equal to zero. When the nucleons do not pair up, the nucleus contains a nonzero net spin and the nucleus possesses spin angular momentum. The sum of the unpaired protons and neutrons gives rise to the spin quantum number and

tells the maximum amount of spin that the nucleus contains. Quantum mechanically, the spin quantum number is a multiple n , of $1/2$. The $n/2$ net spin has a magnetic moment associated with it. The magnetic dipole moment causes the nucleus to behave similarly to a small bar magnet with a north and south pole. The magnetic dipole moment is a vector containing both direction, which is the sense of spin using the right hand rule and magnitude, which is the strength of the bar magnet. The magnetic dipole moment can be represented by a vector along the axis of spin with the dipole moment pointing from the south pole to the north pole of the magnet [Morgan]. The nucleus possesses many possible $(n + 1)$ energy states in an external magnetic field. It is this property which allows the resonance phenomena to occur, and produce the NMR signal.

^1H is the most abundant and simplest isotope found in the human body. The nucleus consists of a single proton and its net spin is $1/2$. ^1H is widely used in MRI because it produces the strongest NMR signal.

In nature, the spin vectors of the ^1H protons are randomly oriented. Once the spin vectors are placed in a static magnetic field (B_0), they will behave similarly to the spinning top in a gravitational field. Since the ^1H proton possesses a spin of $1/2$, two spin vector orientations can be found - parallel and antiparallel to the static external magnetic field. A parallel orientation occurs when the magnetic dipole moment of the spin vectors are aligned with the static magnetic field. This is the lower energy configuration and most common. The antiparallel orientation, where the magnetic dipole moment of the spin vectors are aligned opposing the static magnetic field, is a higher energy configuration. In nature, there are more parallel oriented protons, thus creating a net magnetic moment vector or Magnetization (M_z).

The ratio of the number spins in the parallel orientation (N_{parallel}) to the number of spins in the antiparallel orientation ($N_{\text{antiparallel}}$) can be represented by the Boltzmann equation:

$$N_{\text{antiparallel}} / N_{\text{parallel}} = e^{-\Delta E / kT} \quad (1)$$

where ΔE is the difference in energy between the two orientations, k is the Boltzmann constant of $1.38066 \times 10^{-23} \text{ J K}^{-1}$ and T is temperature in Kelvin. This ratio becomes zero when the temperature nears absolute

zero, which means all of the protons are aligned parallel to the static magnetic field. As the temperature increases, the thermal motions of the molecules cause the number of protons in the higher energy antiparallel state to increase until they are almost equal to the number of those in the lower energy state, and the ratio approaches unity.

The NMR signal is proportional to the difference between N_{parallel} and $N_{\text{antiparallel}}$. Since at room temperature the ratio of $N_{\text{antiparallel}}/N_{\text{parallel}} \approx 1$ and $N_{\text{parallel}} - N_{\text{antiparallel}} \approx 0$, it is necessary for NMR spectroscopy to be a very sensitive technique [Dixon].

The transition of the protons in the low energy parallel state to the higher energy antiparallel state requires the absorption of a photon of energy. The energy of a photon must exactly match the energy difference between the two states for this transition to occur. Once the proton is excited to the higher energy state, it will return to the lower energy state and does this by emitting a photon of energy.

The protons in the antiparallel state will emit a photon only when another photon is absorbed and stimulates the proton to emit a photon. So one photon of energy is absorbed and two photons are emitted and the proton is left in the lower energy state. NMR spectroscopy is dependent on three phenomena: 1) the absorption of energy, 2) the exciting of the low energy proton to a higher energy state, and 3) the return of the proton to the low energy state.

2.2 Nuclear Magnetic Resonance

On a microscopic level, placing the protons in a static magnetic field causes the spin vectors of the ^1H protons to begin to precess, or spin about the z axis in a cone-shaped path. The Larmor frequency (ν), or the frequency at which they precess, is dependent upon the strength of the static magnetic field, B_0 . It can be written:

$$\nu = \gamma B_0 \quad (2)$$

where γ , the gyromagnetic ratio, is the ratio of the magnetic dipole moment to its spin and is unique for each type of nucleus [Smith]. The Larmor frequency is important because it is the frequency at which the protons can absorb photons of energy.

A group of protons experiencing exactly the same B_0 field is called a spin packet. Spin packets can be observed precessing about the applied magnetic field.

Once the protons are aligned in the static B_0 field, a second magnetic field from an alternating or oscillating radio frequency (RF) pulse is applied perpendicular to the B_0 field. The frequency of the RF pulse is at the Larmor or precessional frequency of the ^1H protons, which is approximately 63 MHz in a 1.5 Tesla B_0 field. At the atomic level this field causes transitions between the two energy states to occur due to the presence of photons at the resonance frequency. The protons in the lower (higher) energy state are excited to the higher (lower) energy state. This phenomenon is termed resonance and must occur for the NMR signal to be detected.

The magnetization vectors from the individual spin packets can be represented using an x,y, and z coordinate system. The xyz coordinate system is a perspective of the magnetization vectors called a frame of reference. In NMR, a rotating frame of reference is used to depict what occurs to the net magnetization once it is perturbed by the B_1 field. In the rotating frame of reference, the x and y axes are rotating about the z axis at the Larmor frequency [Stark]. To differentiate the rotating frame of reference from a static or laboratory frame of reference, the x and y axis are primed. Figure 1 illustrates how the spin vectors of the ^1H protons in a rotating frame of reference are aligned with the +z-axis in a static magnetic field. Figure 2 shows the application of a second magnetic field (B_1) from a RF pulse at the Larmor frequency. By convention in NMR, the B_1 field is placed along the +x-axis in the rotating frame of reference.

Figure 1. Net magnetization (M_{z0}) of the spin vectors in a static magnetic field in a rotating frame of reference. The spin vectors are aligned with the +z-axis and in the same direction as the B_0 field.

Figure 2. Application of RF pulse or B_1 field along the +x-axis in the rotating frame of reference.

The applied RF pulse causes the precessing net spins to align away from the +z-axis as shown in Figure 3. The rotation away from the +z-axis depends on the magnitude and duration of the RF pulse.

Figure 3. Perturbation of the net spins away from the +z axis. The spins will begin to precess in a cone-shaped path.

If the RF pulse is left on for long enough, the net spins are rotated 90° down into the xy-plane. Once this pulse is turned off, the equilibrium configuration wants to be re-established. The equilibrium configuration has zero xy magnetization and is aligned with the +z axis. The NMR signal is produced by the relaxation of the xy magnetization to zero which induces an electric current in a receiver coil.

Chemical shift, longitudinal or spin-lattice relaxation time (T_1), and transverse, or spin-spin relaxation time (T_2) are three NMR parameters unique to each type of nucleus. These parameters are discussed below.

Electrons orbiting around the nuclei of molecules possess a moving negative charge, therefore, a magnetic field is produced. The environment of the electrons is not the same for each type of molecule due to different chemical structures, thus the electron-induced magnetic fields surrounding the protons vary. When a molecule is placed in an external B_0 field, the magnetic field from the electrons opposes the B_0 field and shields the protons from the B_0 field. This causes the magnetic field at the nucleus to vary with the resonant or Larmor frequency. These variations in frequency make up the chemical shift [Chakeres]. The chemical shift (δ) in parts per million of a resonance frequency, can be defined by the following equation:

$$\delta = [(\nu - \nu_{\text{ref}}) / \nu_{\text{ref}}] \times 10^6 \quad (3)$$

where ν is the resonance frequency and ν_{ref} is the standard frequency [Morris]. The chemical shift allows identification of a molecular compound since each type of chemical proton produces an NMR peak at a specific frequency.

Just as the electron's magnetic field affects the nucleus, the magnetic field of a nucleus with a certain chemical shift affects another nucleus' magnetic field if its chemical shift is different, and if the two nuclei have less than three bond lengths separating them. The two nuclei will have more than one Larmor frequency where energy can be absorbed, thus, the NMR spectrum will show splitting of the peaks. The distance between this splitting of the peaks is termed the spin-spin splitting or J-coupling constant and it is a measure in Hz of the interaction between two nuclei. As the number of nuclei less than three bond lengths away increases, there is more splitting in the NMR spectral peak. More than one J-coupling constant may be present when this occurs [Morris].

Another inherent NMR parameter is the longitudinal relaxation or spin-lattice relaxation time (T_1). Recall that the spin vectors are aligned along the +z axis in the B_0 field. Once an RF pulse is applied, the spin vectors rotate away from the +z axis. Over time, the spin vectors will return to equilibrium and re-align with the +z-axis in the B_0 field. The time it takes for this relaxation of the spin vectors from the nonequilibrium state back to equilibrium occurs exponentially, and is the T_1 time constant. The molecules in the nonequilibrium state will give off its energy to other molecules and into the lattice (in this thesis, the solution) in order for the relaxation to occur, hence, the term spin-lattice relaxation. T_1 's will vary due to the fluctuating magnetic or dipole fields from neighboring magnetic nuclei. In liquids, the fluctuating magnetic field from the neighboring magnetic nuclei, or the lattice, is caused by the Brownian motion of the molecules. The longitudinal relaxation is depicted in Figure 4.

Figure 4. Longitudinal relaxation of spin vectors back to equilibrium . The spin vectors have been rotated away from the +z axis and over time will re-align themselves with the +z-axis at equilibrium.

The relaxation behavior can be shown mathematically with the equation:

$$M_z = M_{z0} (1 - e^{-t/T_1}) \quad (4)$$

Put simply, the longitudinal relaxation time is the time it takes the longitudinal magnetization to return to equilibrium by a factor of “e” [Dixon].

Since T_1 is the longitudinal relaxation time, T_1^{-1} can be described as the relaxation rate. It is the rate of change of the longitudinal relaxation over time. This is proportional to the number of molecular motions at the Larmor frequency. Figure 5 depicts the molecular motions, $J(\nu)$, of a nonviscous, slightly viscous and viscous liquid as a function of frequency.

Figure 5. The molecular motions of either nonviscous, slightly viscous, and viscous liquid at a particular frequency is found by intersection of the frequency with the type of liquid and drawing a straight line across to the number of molecular motions. T_1 can be predicted since there is an inversely proportional relationship to the molecular motions and T_1 .

Another NMR parameter is the transverse or spin-spin relaxation time (T_2). When a 90° RF pulse is applied along the $+x$ -axis, the net spins are rotated away from the $+z$ -axis and into the xy -plane as shown in Figure 6. Figure 6 illustrates what occurs once the RF pulse is turned off. The spins begin to precess at different rates due to inherent inhomogeneities in the static B_0 field and variations in local magnetic fields caused by molecular motions. This produces a transverse magnetization (M_{xy}). There are many spins precessing at different rates which will eventually cancel each other. This is shown in Figure 8. Once the transverse magnetization is no longer present, or zero, the spins begin to relax longitudinally back to equilibrium along the $+z$ -axis. The time it takes the spins to cancel the transverse magnetization is related to the transverse or spin-spin relaxation time, T_2 .

Figure 6. The magnetization (M_{xy}) into the xy -plane after application of a 90° RF pulse along the $+x$ -axis.

Figure 7. The spin vectors begin to precess at different rates due to inhomogeneities in the B_0 field and variations in the local magnetic fields. In the rotating frame of reference, the precession is shown with the spin vectors travelling in a counter-clockwise and clockwise direction.

Figure 8. The spin vectors are precessing at different rates and over time cancel each other; therefore, the transverse magnetization becomes zero.

Mathematically, the loss of transverse magnetization caused by spin interactions can be represented by the equation:

$$M_{xy} = M_{xy0} e^{-t/T_2} \quad (5)$$

The spin-spin relaxation time can be described as the time to reduce the transverse magnetization by a factor of “e” [Dixon]. The T_2 is almost always shorter than T_1 , causing transverse magnetization to disappear before longitudinal relaxation can occur.

T_2^{-1} is the transverse relaxation rate and is proportional to the number of molecular motions at and less than the Larmor frequency. Energy is not transferred in this relaxation process, only a loss of phase of the individual spins occurs. The loss of phase happens due to the the molecular motions causing fluctuating magnetic fields [Chakeres].

T_2 from the variations in the magnetic fields from molecular motions and T_2 from the inhomogeneities in the B_0 field comprise the T_2^* , or the inhomogenous T_2 [Morris]. The transverse magnetization decays as a function of T_2^* .

Some T_1 and T_2 values for the human body are listed below in Table 1. The values of the tissues differ because the density and molecular motions of the ^1H protons vary from tissue to tissue. The T_1 and T_2 of water is 2500 ms for both relaxation values in a 1.5 Tesla magnetic field.

Table 1. T_1 and T_2 values in a 1.5 Tesla magnetic field [Fletcher].

Tissue	T_1 (ms)	T_2 (ms)
Cerebral spinal fluid	800-2000	110-2000
White matter	760-1080	61-100
Grey matter	1090-2150	61-109
Meninges	500-2200	50-165
Muscle	950-1820	20-67
Adipose	200-750	53-94

An increase in temperature causes an increase in the motion of molecules, consequently, the T_1 and T_2 values are changed. Usually, the T_1 and T_2 will become shorter. Conversely, a decrease in temperature will decrease the molecular motions, and generally, the T_1 and T_2 values are longer.

One method of adjusting the T_1 and T_2 values is a technique called paramagnetic doping. Paramagnetic substances contain unpaired electrons which makes them more susceptible to magnetization [Stark]. The unpaired electrons cause the local magnetic field surrounding the protons to change and dipole-dipole interactions are enhanced [Lufkin]. The T_1 relaxation time of water in an aqueous solution of paramagnetic ions is shortened because there is an increase in the strength of the alternating magnetic field of the paramagnetic ion, therefore, more spins will be excited and begin to resonate, producing an increase in the number of molecular motions. T_2 relaxation is shortened due to the increased inhomogeneity of the magnetic field causing transverse magnetization to relax at a faster rate [Chakeres]. Some examples of paramagnetic ions are Ni^{+2} , Mn^{+2} , and Cu^{+2}

In NMR, the signal is obtained from the T_2^* decay. An oscillating curve represents the signal from the transverse magnetization after application of an RF pulse at the Larmor frequency. Once the RF pulse is discontinued, the T_2^* decays to zero. This exponentially decaying oscillating wave is called a free induction decay (FID) and is shown in Figure 9.

Figure 9. An exponentially decaying oscillating wave over time along a direction in the xy -plane. The decaying oscillating wave is called an FID.

The FID is converted from the time domain to the frequency domain by a mathematical technique called Fourier transform. This technique allows absorptions associated with an NMR spectral peak to be seen as a function of the frequency. The amplitude of the FID determines the amplitude of the spectral peaks.

A spin-echo pulse sequence is one method of obtaining signal from a sample in NMR. A 90° RF pulse is applied along the $+x$ -axis causing the spins to precess in the xy -plane as shown in Figure 9.

Figure 10. The spin vectors begin to precess in the xy plane after a 90° pulse is applied in the $+x$ -axis in a spin-echo pulse sequence.

A 180° pulse is now applied along the $+x$ -axis, the spins are *flipped* 180° about the $+x$ -axis (along which the B_1 is applied), and a converging or refocusing of the spins occurs as shown in Figures 11 and 12.

Figure 11. In the spin-echo pulse sequence, the spins are flipped 180° by the application of a second pulse - a 180° pulse along the $+x$ -axis. Now the spins will converge or refocus.

Figure 12. The spins are refocused and will begin to precess at different rates until M_{xy} is equal to zero.

Figure 13 illustrates a timing diagram of the spin-echo pulse sequence.

Figure 13. Timing diagram for spin-echo pulse sequence illustrating the application of a 90° RF pulse followed by a 180° RF pulse and the FID and echo signals which are generated by these RF pulses. The TE and TR are adjustable experimental parameters.

The FID from the 90° RF pulse is shown as well as the *echo* or signal from the 180° RF pulse. The repeating of the pulse sequence from the 90° RF pulse to the next 90° RF pulse is called the repetition time (TR). The time from the 90° RF pulse to the maximum amplitude of the echo is called the time of the echo (TE). These times can be varied, depending on the NMR experiment.

Once the signal is obtained from the sample, the T_1 and T_2 can be calculated. The following signal equation from a spin-echo pulse sequence calculates these values:

$$S = (1 - e^{-TR/T_1}) (e^{-TE/T_2}) \quad (6)$$

Algorithms then find the best fit of the signal equation to the data and produce a T_1 and T_2 value [Gong, Li].

2.3 Magnetic Resonance Imaging

Magnetic resonance imaging applies magnetic field gradients and the principles of NMR to create an image of the NMR signal. Consequently, the MRI hardware is more elaborate than the NMR hardware. MRI hardware includes a super-conducting magnet, gradient coils, RF transmit and receive coils, computer hardware, image and array processors, and RF and magnetic field shields.

The super-conducting magnet is composed of niobium-titanium alloy and is enclosed in a copper matrix. The magnet produces the static external B_0 field. An advantage of this type of magnet is higher signal-to-noise ratio, which produces better soft tissue contrast and resolution. The gradient coils allow

spatial information of an image to be obtained, while the RF transmit and receive coils detect the signal and allow the computer and processors to produce the image as we see it. A lining of aluminum or copper mesh around the magnet room protects the imager from outside RF electromagnetic interference. An iron shield surrounding the MRI center protects the imager from distortion caused by ferromagnetic materials outside the magnet room [Morgan].

The ability of the MR imager to locate and separate signals from different tissues differentiates MRI from one dimensional NMR spectroscopy. As previously mentioned, the gradient coils used in MRI encode spatial information about a sample into the signal. The magnetic field of the gradient is varied linearly along either an x, y, or z direction. For example, an RF pulse excites the protons in a certain imaging volume. A single FID is generated for all the protons because they experience the same magnetic field and are precessing at the same frequency. Once a gradient is applied in the x direction, the magnetic field the protons experience is varied due to their locations and the strength of the field at that particular location. The frequency of precession of the protons is not the same and FIDs are generated for each of these frequencies [Stark]. There are three field gradients--slice selection, phase encoding, and frequency encoding.

The slice selection gradient (G_z) is applied perpendicularly to the chosen slice, where rotation of the selected spins into the y-axis and precession in the xy-plane takes place. This gradient determines the location and thickness of the slice. Both the location of the slice and thickness can be adjusted by changing the width and frequency of the RF pulse or slope of the gradient, G_z [Stark].

Phase encoding of the spin packets occurs when the phase encoding gradient (G_y) is turned on perpendicularly to G_z . Application of G_y along y causes spins at certain locations, x, to rotate at a frequency dependent on the static B_0 field at that location, and G_y field strength. The spin packets at the different x locations will have a particular phase angle along the y-axis and are now phase encoded.

The third or frequency encoding gradient (G_x) is applied orthogonally to G_y and G_z . Application of G_x along x will cause all the spins to rotate at a different frequency depending upon the static B_0 field

strength, y location, and G_r field strength. These spins are frequency encoded. The application of the G_ϕ and G_r gradients help determine the resolution of the image [Stark].

Imaging coils transmit the RF pulses into and receive signals from the sample being imaged. Two common coils of the imager are the head and body coil. The body coil is the largest of the coils and is built into the bore of the magnet. It transmits the RF pulses and will receive the RF signal from the sample. Because this coil is so large, it causes a poor signal-to-noise ratio (SNR) in an image of a small object. The head coil is a smaller transmit and receive coil. The SNR is improved with this type of coil due to the closer proximity of the small sample (or anatomy) being imaged. Figure 14 is a schematic of a side view of the head coil.

Figure 14. Birdcage head coil side view. The are capacitors placed on the coil. A sample or the human head is placed inside the head coil in the direction of the B_0 field.

There are three standard imaging planes which are obtained from an MR imager. These are the axial, coronal, and sagittal planes with respect to a body lying in the magnet. The plane is chosen depending on which view of the sample or anatomy is desired. Figure 15 illustrates these planes through a sample bottle standing up in the head coil.

Figure 15. The axial plane (----) is the cross-section of this sample bottle, the coronal plane (—) slices the top from the bottom of the sample bottle, and the sagittal plane (---) slices the sample bottle from right to left.

2.3.1 Other imaging parameters

There are several imaging parameters which can be adjusted to obtain the desired image containing the optimum contrast and resolution. In an image, the signal originates from a voxel or volume element of a sample. It is defined by its location within the magnet and its dimensions. Voxels are mostly rectangular in shape. Each voxel corresponds to a pixel, or picture element, and the smallest part of a digital image display. A higher signal from a voxel causes the intensity of the pixel to be brighter in the image. The pixel data points, composing each side of the image, are then placed in columns and rows, or a matrix. The size of the matrix is typically 128 x 256, 192 x 256, or 256 x 256. The details of the image are enhanced when there is a higher number of pixels or larger matrix [Chakeres].

MRI is a tomographic imaging technique. It creates two dimensional images of the hydrogen NMR signal inside a slice through the object with a thickness (thk). The slice thickness is dependent on the gradient strength and frequency of the RF pulse. The slice thickness can vary from submillimeter to centimeters, therefore the number of protons, which produce signal, will vary from slice thickness to slice thickness. Usually, the thinner the slice thickness the greater the resolution and image detail. While thicker slices improve the SNR, they decrease the spatial resolution.

The field of view (FOV) is the size in centimeters of the area to be imaged. The size of the FOV is determined by the size of the object to be imaged and must be as large as or larger than the imaged object [Smith]. The size of the voxel is determined by the FOV and slice thickness. Increasing the slice thickness while keeping the FOV constant, will increase the size of the voxel. This procedure can be followed to improve SNR levels.

Other imaging parameters can be varied to produce greater signal in the images, or optimal spectral information. These are the transmitter gain (TG), flip angle, receive gain (RG), TR, and TE. The TG determines the amplitude of the delivered RF pulse (B_1). This parameter can be varied to obtain an optimal value or 90° RF pulse. The flip angle, or amount of rotation of the RF pulse into the xy-plane, is controlled by the TG. The TG is increased up to a certain point, which increases the flip angle, and allows the spins to be rotated 90° and into the xy-plane. This produces a better signal in the image. If the TG is increased too much, the spins are rotated past 90° and less signal is produced. The contrast of an image can be altered by adjusting the TR and TE values. Most often, a longer TE and shorter TR affords the best contrast in an image.

2.3.2 Imaging techniques

Each tissue has a unique T_1 , T_2 , and proton density which produces the signal intensities. Consequently, these unique values can be used to differentiate tissue types. Two types of signal, from the different chemical shifts of fat and water, are present in the human body. Sometimes it is useful to suppress the fat (or water signal) which will increase the water (or fat) signal so the contrast between

certain tissues is increased. For example, the water signal in tumors located in fatty tissue, can be enhanced, allowing the tumor to be more readily diagnosed [GE Manual-chemsat sequence].

When an RF pulse is applied, the fat protons precess slower than the water protons, resulting in a 220 Hz separation between their spectral peaks. This separation of frequency permits the suppression of the water or fat peak to occur. The GE chemsat technique presaturates the fat signal with a frequency selective pulse, prior to the normal 90° RF pulse in a spin-echo pulse sequence. The spins of the protons of the fat are rotated into the xy-plane causing M_z to be equal to 0. Spin-spin relaxation occurs causing $M_x = M_y = 0$ for the fat ^1H . Next, the 90° RF pulse from the spin-echo pulse sequence is applied.

Assuming the T_1 of fat is much larger than the time between the saturation pulse and the 90° pulse, only the spins of the water are rotated into the xy-plane because M_{xy} of fat should still equal zero. Now only the water spins are flipped by the 180° pulse and refocused. An echo or signal from the water protons is thus recorded. If the T_1 of the fat protons is short, the spins have relaxed back to equilibrium along the z-axis. The spins can now be excited by the 90° RF pulse of the spin echo pulse sequence, and some fat signal can be produced. The fat signal is no longer suppressed completely, and the contrast between the water and fat tissue is not as great. This technique can be applied to water as long as the presaturation pulse is the same precessional frequency as the water protons and the T_1 of water is longer than the time between the saturation and 90° pulse.

Magnetization transfer or magnetization exchange between different types of protons, such as the protons in one component (A) and those in another component (B) of a solution, affect the overall T_1 of a system. The exchange pathway can occur by chemical exchange or through space dipolar interactions [Balaban]. The antiparallel protons of one component gives off a photon at the Larmor frequency. This photon may excite a proton of another component at the Larmor frequency, changing it from the parallel to antiparallel state. Therefore, an energy exchange between the two components occurs. The probability of the emitted photon exciting a proton of another component is proportional to the square of the distance between the two protons. Thus, the closer the two protons are, the more likely an exchange of energy will take place. The overall T_1 of a three component system is produced from the T_1 of component A, B, and

C and the exchange rate k between the components. The overall T_1 can be either monoexponential or multiexponential depending on the magnitude of the exchange rates k . The overall T_1 of the three component system can be represented by a spin bath model, Figure 16.

Figure 16. Three component (A,B,C) spin-bath model. The * denotes spins in the excited state. Each component has its own spin-lattice relaxation rate ($1/T_1$) process, which corresponds to the k_1 , k_2 , and k_3 value and opens into the lattice. The exchange rates k_4 and k_5 are the energy pathways between the A* and B* components. The exchange rate k_6 and k_7 are the energy pathways between components B* and C*. No energy pathway is found between components A* and C*

If the k values are large, the overall T_1 will be monoexponential. If k values are small, the overall T_1 will be multiexponential, thus dependent on the T_1 of the individual components. To determine whether the overall T_1 is monoexponential or multiexponential, a magnetization-transfer pulse sequence can be utilized. This pulse sequence applies the center frequency adjust (CFA) variable control of the prescan mode, where only the slice selection and phase encoding gradients are turned on. T_1 data from the signal of the spectrum of each component is collected by adjusting the TE values. This is possible as long as the components absorb at different frequencies, so their peak's can be differentiated.

Determination of the k values can be found through the following equations and by plotting a curve of best fit to the data.

Initially, there are spins in the excited state (A^* , B^* , C^*) that give off their energy and relax back to the ground state (A, B, C). This can be expressed in the following manner:

Since there are three components making up the solution, the following energy exchange between components will occur:

Assuming first order reactions of equations 7 through 9, the change in the concentration of the excited spins over time can be expressed as:

$$dA^* / dt = -k_1 [A^*] \quad (14)$$

$$dB^* / dt = -k_2 [B^*] \quad (15)$$

$$dC^* / dt = -k_3 [C^*] \quad (16)$$

where k_1 , k_2 , k_3 , respectively, are first order rate constants for the spin-lattice relaxation processes of the individual components. The rates are negative since the concentration of the excited spins is decreasing in concentration.

Energy is exchanged between the components as depicted in equations 10 through 13. Now the reaction becomes dependent on the concentration of another in the system and the reaction is second order. The rate equations are expressed in the following manner:

$$dA^* / dt = -k_4 [A^*] [B] \quad (17)$$

$$dB^* / dt = -k_5 [B^*] [A] \quad (18)$$

$$dB^* / dt = -k_6 [B^*] [C] \quad (19)$$

$$dC^* / dt = -k_7 [C^*] [B] \quad (20)$$

where the k values are the exchange rate constants between components. If the concentration of unexcited spins is very large and therefore remains constant, the above equations can be modified to form pseudo first order rate equations. Now the exchange rate constants include the concentration of the unexcited spins as well as the exchange rate to form new exchange rate constants. The reactions can be expressed as:

$$dA^* / dt = -k_8 [A^*] \quad (21)$$

$$dB^* / dt = -k_9 [B^*] \quad (22)$$

$$dB^* / dt = -k_{10} [B^*] \quad (23)$$

$$dC^* / dt = -k_{11} [C^*] \quad (24)$$

Combining all the terms to form coupled differential equations, the exchange rate constants can be found. This is expressed in the following equations:

$$dA^* / dt = -k_1 [A^*] - k_8 [A^*] + k_9 [B^*] \quad (25)$$

$$dB^* / dt = -k_2 [B^*] - k_9 [B^*] - k_{10} [B^*] + k_8 [A^*] + k_{11} [C^*] \quad (26)$$

$$dC^* / dt = -k_3 [C^*] - k_{11} [C^*] + k_{10} [B^*] \quad (27)$$

2.4 MRI Phantom

Phantoms are nonliving objects which have a variety of uses in MRI. They are used for evaluating system performance of the imager, the development of pulse sequences, testing of the reliability of T_1 and T_2 measurements, and as test objects for tissues [Kraft,Gore]. The phantoms are constructed of different sizes and shapes depending on their use. The solutions inside the phantoms have been made of aqueous solutions of paramagnetic salts, polyacrylamide gels, agarose gels, gelatin, oils, or other tissue-mimicking organic solutions [DeLuca, Mitchell]. The most common phantom for system performance testing is a spherical phantom composed of an aqueous solution containing 14 mM $NiCl_2$. The availability and low cost of the material is a major advantage of this type of phantom. Disadvantages of this phantom include a standing wave artifact caused by the large dielectric constant of the solution, and an RF

penetration effect artifact produced by the increased ionic strength of the solution. As a result, the transmit and receive RF magnetic fields B_{1R} and B_{1T} of an imaging coil, and the homogeneity of the main B_0 field can not be accurately measured.

2.5 Artifacts in MR images

Artifacts are distortions in an image caused by inhomogeneities in the static B_0 field, nonuniformities of applied B_1 field, variations in the sensitivity of the RF coils, a FOV that is smaller than the object being imaged, and motion of the object. These artifacts may cause a decrease in the resolution and contrast in an image or a distorted image. Correction of the wrap-around artifact from too small a FOV, can be made by increasing the FOV on the imager. The motion artifact is eliminated when the imaged object is held in place and not allowed to move. The other types of artifact are more challenging and difficult to correct, since they can be caused by inherent properties of the imager and RF coils. Therefore, the use of a uniform phantom solution is significant in understanding and for correcting these artifacts.

A dielectric standing wave artifact is produced in phantoms containing aqueous solutions of paramagnetic salts due to the solution's high dielectric constant. The dielectric constant of the aqueous solution can be related to the index of refraction of a substance (η_i) through a series of calculations. The index of refraction is defined in the following equation:

$$\eta_i = c / v_i \quad (28)$$

where c is the velocity of light in a vacuum and v_i is the velocity of light in a medium i .

The velocity of radiation can be defined as :

$$c = v_{vac} \lambda_{vac} \quad (29)$$

where v_{vac} is the frequency in a vacuum and λ is the wavelength in a vacuum, and

$$v_i = v \lambda_i \quad (30)$$

where v is frequency in a medium and λ_i is the wavelength in a medium, Eqn 28, now becomes:

$$\eta_i = v_{vac} \lambda_{vac} / v \lambda_i \quad (31)$$

Rearranging the equation, the wavelength in a medium now becomes:

$$\lambda_i = \lambda / \eta_i \quad (32)$$

The index of refraction can also be related to the permeability and permittivity of a solution through this equation:

$$\eta_i = (\mu_{ir} \epsilon_{ir})^{-1/2} \quad (33)$$

where μ_{ir} is the relative permeability of a medium (μ_i / μ_o), which is 1 in most substances, except metals, and ϵ_{ir} is the relative permittivity of a medium (ϵ_i / ϵ_o) [Moore]. The relative permittivity is also called the dielectric constant κ_i [Gettys]:

$$\epsilon_i = \epsilon_o \kappa \quad (34)$$

For many nonmetallic materials, such as those used in this thesis, $\mu_{ir} = 1$. Combining equations 28.30, 33, and 34 yields

$$\lambda_i = c (\kappa)^{-1/2} / \nu \quad (35)$$

When an RF wave tries to enter or leave a phantom, a fraction of its intensity is lost. If I_o is the incident intensity and I_r the reflection intensity, the fraction reflected, I_r / I_o , is

$$I_r / I_o = (\eta_1 - \eta_2)^2 / (\eta_1 + \eta_2)^2 \quad (36)$$

where η_1 and η_2 , are the index of refraction of two mediums [Skoog]. Since the dielectric constant of the aqueous salt solution inside the phantom is ≈ 80 , the index of refraction is 9. The index of refraction of air is 1, making the amount of reflected waves $\approx 64\%$.

The electromagnetic wave traveling in the direction of the B_1 field will change after reflection. It reflects 180° out of phase and in the opposite direction. As the wave reflects, both constructive and destructive interference occurs. When the object to be imaged is the same diameter as one-half of this wavelength, a standing wave results.

The wavelength in air at 63 MHz, the operating frequency of the 1.5 T imager is 4.7 m, and when this passes into the aqueous solution phantom, the wavelength is reduced to 0.52 m. The 27 cm diameter sphere of the phantom, used in system performance testing, is approximately one-half this wavelength. The maximum amplitude of the B_1 field caused by constructive interference will be $\lambda/4$ from the outer edge of the phantom, which coincides with the center of the phantom. Therefore, the signal

intensity is greater in that region. Destructive interference occurs along the outer edges of the phantom resulting in underflipped spins with a subsequent decrease in the signal intensity. Consequently, the image of the sphere will appear to have a brighter intensity in the center than at its outer edges. This phenomenon is the standing wave artifact. The use of oils, such as, vegetable oil, have been proposed as alternative types of phantom solutions to minimize the standing wave artifact [Tofts, 1993]. The oils have a lower dielectric constant than water which results in a longer wavelength than the aqueous solutions, and thus, does not form a standing wave inside the phantom. The constructive and destructive interference from reflected waves inside the phantom do not cause a maximum and minimum in the wave and large variations in the signal intensity of the image are not seen. Disadvantages of oils are that they contain impurities, and do not have properties similar to human tissue.

In a similar manner, the size of the phantom can be decreased, which will minimize the standing wave artifact. Now the constructive and destructive interference will not cause a maximum in the wave when the shortened wavelength is reflected. The disadvantage of decreasing the size of the phantom is that it also reduces the desired large imaging volume. Therefore, reducing the size of the phantom and using oils for phantom solutions, produce a phantom that is less beneficial for system performance testing.

Ideally, the phantom solution should load an imaging coil similarly to the human body. In order for a phantom containing an aqueous solution of paramagnetic ions to perform in the same manner as the human body, NaCl is added to increase the ionic strength of the solution [Hornak, Smith]. An imaginary component is added to the dielectric constant as the ionic strength is increased, and an RF penetration or *skin-depth* effect is produced. This *skin-depth* effect may prevent the RF transmitter field from penetrating the object. The flip angle of the B_1 magnetic field is less than 90° at the center of the phantom, causing a decrease in transverse magnetization, therefore, the signal detected from the center is less. The phantom image will appear brighter at the edges of the sphere than in the center. This is called the *skin-effect* artifact. This effect is increased as the frequency, size of the phantom, and electrical conductivity of the phantom solution increases [Bottomley].

Convection currents caused by thermal gradients due to the deposition of RF energy in a liquid may create an artifact. The artifact is produced in the following manner. A solution may contain two regions, for example, an upper and a lower. As the temperature of the lower region of the solution increases, the viscosity of the lower region is decreased. Once the temperature between the regions reaches a critical difference, the Rayleigh number, the lower viscosity layer pushes upward while the upper layer with the increased viscosity flows to the bottom. A convection current emerges. This pattern becomes oscillatory and finally completely disordered. The artifact is manifested by circulatory patterns in the image of the solution [Gibbs]. As the viscosity of a solution increases, the temperature needed to produce the convection currents will rise. Therefore, when the temperature of the energy released is held constant, these convection currents will not appear in a solution with a higher viscosity.

To test for inhomogeneities of the B_1 transmit and receive fields of an imaging coil, these artifacts must be minimal or not present in a test phantom. If the phantom solution itself causes artifacts to occur, it would be difficult to distinguish the origin of the variations of the image intensity from the imaging coil or the phantom. In some cases, this may leave the imaging system at less than its optimal operating performance.

2.6 The reverse micelle solution

A reverse micelle solution offers many advantages when compared to an aqueous solution of paramagnetic ions for use in an MRI phantom. Its physical properties will be discussed in the following paragraphs.

The reverse micelle (RM) solution is an example of a *water-in-oil* microemulsion. Separating the water from the oil is a surfactant which contains a hydrophilic and hydrophobic end. The hydrophilic polar head is in contact with the water, and surrounds the water in a single layer to form a water droplet, or water pool. The hydrophobic end, a long carbon chain, extends away from the polar head and into the hydrocarbon or oil. These three components can also form another phase called the lamellar phase. In this phase, the three components are separated into sheets or layers. A sandwich of the water layer, a layer of the polar heads of the surfactant, the hydrophobic tails of the surfactant, and then the decane, is formed.

The phase change of the RM to the lamellar phase is due to an increase in the attractive interparticle interactions when the RM concentration is increased as ϕ increases and by increasing the temperature. These attractive interparticle interactions cause a phase change from a homogeneous droplet phase, that is, the RM phase, to a phase containing the RM and lamellar phase, or the coexistence phase [Vollmer]. The coexistence phase will form the lamellar phase at higher temperatures.

The particular RM solution used in this study consisted of water, the hydrocarbon decane, and the surfactant bis(2-ethylhexyl) sulfosuccinate sodium salt, or aerosol octyl (AOT). The AOT has the following chemical composition:

The Na^+ readily dissolves into the water inside the RM while the SO_3^- group comprises the polar head.

The radius of the RM can be adjusted by changing the molar ratio of water-to-AOT. In this study, the molar ratio of H_2O / AOT was held constant at 40.8, or the mass ratio of water to AOT was constant at 0.6 in all of the solutions. This fixed molar (or mass) ratio produced an RM with a mean radius of 50 Å [Kotlarchyk, 1992]. The phi (ϕ) value, or ratio of the volume of AOT and water to the total volume of AOT, water, and decane, can be adjusted to produce varying concentrations of RMs per volume in a solution. Therefore, the amount of decane present decreases, causing the concentration of RMs to increase as the ϕ value increases.

Inside the reverse micelle, some of these Na^+ ions dissociate into the water forming cations inside the water pool, leaving the polar head of the AOT negatively charged. Strong hydrogen bonds form between the polar heads of the AOT and the hydrogen of the water inside the RM [D'Aprano]. Consequently, there is a layer of structured or "bound" water of approximately 5 Å thick along the outside edge of the water pool [Kotlarchyk, 1985]. In the center of the RM, the water exhibits characteristics of free or "bulk" water [Hauser]. The tails of the AOT are approximately 8 Å in length. Only approximately

2.4 Å of the end of the tail extends into the hydrocarbon [Kotlarchyk, 1985]. The total volume of a single RM is approximately 5.2×10^{-22} ml.

The molecular weights, density (D), number of H atoms of the RM of the three components are listed in Table 2. The number of Ni^{+2} , Mn^{+2} , and Na^{+} in an RM for each concentration is represented in Table 3.

Table 2. Characteristics of the RM Components

Component	M.W. (g/mol)	D (g/cm ³)	No. of H atoms
AOT	444.57	1.2060	37
Decane	142.29	0.7300	22
H ₂ O	18.02	0.9970	2

Table 3. Ions per reverse micelle in different concentrations of the aqueous phase

Ion	Concentration (mM)	Average number of ions per RM
Ni^{+2}	8	2.51
Ni^{+2}	5	1.58
Ni^{+2}	3	0.95
Mn^{+2}	0.5	0.16
Mn^{+2}	0.25	0.08
Mn^{+2}	0.10	0.03
Na^{+}	1340	405

The concentration of Na^{+} from the AOT inside each reverse micelle is 1.34 M, and dependent only on the concentration of the AOT. Therefore, 405 Na^{+} are contained inside each of the RM, regardless of the concentration of the paramagnetic ion. The paramagnetic ions are located in the aqueous phase of the RM and do not affect the size and shape of the RM.

Figure 17 displays a cross-section diagram of the RM containing water with a paramagnetic ion, AOT and decane. The hydrophilic polar heads of the AOT encapsulate the water while the hydrophobic tails of the AOT extend out into the hydrocarbon decane. The structure of the lattice of RMs in a solution corresponds to a face or body-centered cubic [Kotlarchyk, 1984]. Figure 18 shows the RMs in a solution. At lower ϕ values, more decane is present, consequently, the RMs are further apart in solution.

The RMs are not static in solution. In solution, the spherical shape of the RMs may fluctuate depending on the bending energy of the AOT interface. These fluctuations are manifested by a crimping of the spheres or an ellipsoidal shape. The fluctuations in the shape of the RMs are dependent on the concentration of the RMs and the temperature of the solution [Safran].

There are attractive interactions between the RMs which cause the RMs to collide and merge. Thus, the RMs are able to exchange contents of the H₂O, ions, and AOT polar heads from the inside of the RM between themselves. It is the forming of aggregates or *clustering* of the RMs that allows percolation, which is the ability of the solution to conduct an electrical current [Paul]. Literature describes at least two mechanisms for the conduction of current in the RM solution. One mechanism is attributed to the *hopping* of the negatively-charged ion of the AOT from droplet to droplet within the droplet clusters. Another mechanism for percolation takes place once the RMs merge. The formation of clusters from collisions allows ions to be transferred from one RM to another because of an open water channel [Jada]. It is these properties that make the RM solution desirable for use in an MRI phantom. The experiments designed to characterize the RM solution follow in the experimental section.

Figure 17. Cross-section diagram of a reverse micelle in decane.

Figure 18. Cross-sectional view of RM in a solution of decane. As the ϕ increases, the amount of decane for the RM to be suspended in decreases, thus the concentration of RM increases per unit volume.

3.0 EXPERIMENTAL

3.1 Introduction

This section describes the experiments performed to determine the properties of a reverse micelle solution and its effectiveness for use as an MRI phantom. The experimental section has been broken down into four parts. The first part describes the solution preparation of the Mn^{+2} solutions to determine which concentrations of Mn^{+2} will give the desired T_1 values, the preparation of the reverse micelle solutions which were used in the experiments, and phantom preparation. In the second part, the physical properties of the reverse micelle solution were determined by phase and stability studies, viscosity measurements, and diffusion studies. Dielectric constant studies and RF conductivity define the third part, or the electrical properties of the reverse micelle solution. The T_1 and T_2 of the aqueous Mn^{+2} solutions were measured. NMR properties, the effects of variable temperatures on the T_1 and T_2 , reproducibility and stability of T_1 and T_2 of the reverse micelle solution, and magnetic susceptibility, characterizes the magnetic properties of the reverse micelle solution. The fat saturation imaging sequence, magnetization-transfer of the three component reverse micelle system and the study of the transmit and receive B_1 fields conclude the fourth part, the study of magnetic properties of the reverse micelle.

3.2 Solution preparation.

3.2.1 Initial manganese solution preparation.

The T_1 and T_2 values of aqueous solutions can be modified so they more closely match the T_1 and T_2 of the tissues in the body. These values can also be shortened to decrease the scanning time needed to image the sample. This is accomplished by the addition of the paramagnetic ions Mn^{+2} or Ni^{+2} . To determine the appropriate concentration of Mn^{+2} for the desired T_1 and T_2 range, solutions of 0, 0.05, 0.1, 0.25, 0.5, 1.0, 2.5, 5.0, 7.5, 10.0, 15.0 mM $MnCl_2$ were made with $MnCl_2$ (Sigma, St. Louis, Mo) and H_2O (J.T. Baker Analyzed HPLC, Phillipsburg, NJ). These solutions were placed in 20 ml polyethylene bottles. T_1 and T_2 values of the solutions were measured using the 1.5 T GE (Milwaukee, WI) Signa magnetic resonance imager located at University of Rochester, Strong Memorial Hospital. The procedure for measuring T_1 and T_2 values of these solutions will be outlined in the magnetic properties portion of the experimental section.

3.2.2 Reverse micelle preparation.

Large batches for each Mn^{+2} or Ni^{+2} concentration of the reverse micelle solution were made up by placing 60.0 grams of AOT (Fluka Chemica #86140, 98%, Ronkonkom, NY) in a glass beaker and 100 ml of water doped with either 0.1, 0.25, and 0.5 mM $MnCl_2$ or 3, 5, and 8 mM $NiCl_2$ (Fisher Scientific, Fair Lawn, NJ). Once the AOT had dissolved in the water, 16 grams of the AOT/water mixture was placed into 25 ml glass bottles with a diameter of 2.5 cm and 8.5 cm length. Various amounts of *n*-decane (Fluka Chemica #30560, 95%) were added, depending on the desired ϕ value. The mixture was shaken vigorously to completely mix the *n*-decane with the AOT and water to form the reverse micelle. Table 4 shows the amount of H_2O , AOT, and decane needed for ϕ value.

Table 4. Amounts H_2O , AOT, and Decane Needed for ϕ Values in 25 ml containers.

$\phi =$	0.5	0.55	0.6	0.65	0.7	0.75	0.8	0.85	0.9
HPLC H_2O (ml)	10	10	10	10	10	10	10	10	10
AOT (g)	6	6	6	6	6	6	6	6	6
Decane (ml)	14.98	12.26	9.98	8.06	6.42	5.00	3.74	2.64	1.66

3.2.3 Phantom preparation

Three different-sized high density polyethylene spheres were used as phantoms in the experiments. Table 5 contains volume and dimensional statistics of each phantom size. The smaller phantoms were used for the magnetic susceptibility, stability, and fat-saturation imaging sequence studies. The larger phantoms were developed to test the transmit and receive B_1 fields of the imaging coil, since they provided the most imaging volume for measurements of spatial variations.

Table 5.. Volume and diameter of the small, medium, and large phantoms.

Phantom Size	Volume (L)	Diameter (cm)
Small	0.700	11.0
Medium	2.572	17.0
Large	10.31	27.0

The phantom reverse micelle solution was made according to the ϕ value desired and the total volume of the phantom. The following equations were used to determine the amounts of decane, AOT, and H_2O needed for making up the reverse micelle solution for individual phantoms:

$$\phi = V_{(AOT + Water)} / V_{(AOT + Water + Decane)} \quad (37)$$

$$V_{AOT} = 0.6 V_{(Water D_{Water})} / D_{AOT} \quad (38)$$

$$V_{Water} = \phi / (0.6 D_{Water} / D_{AOT} + 1) \quad (39)$$

$$V_{Decane} = V_{Water} (0.6 D_{Water} / D_{AOT} + 1) (1 - \phi) / \phi \quad (40)$$

Table 6 lists all of the spherical phantoms made and the solution contained inside them. These phantoms will be referred to by number throughout the remainder of the thesis.

Table 6. Phantom Solutions and Sizes

Phantom	Solution	Size
1	Aqueous 0.25 mM $MnCl_2$	Small
2	$[Mn^{+2}] = 0.25$ mM $\phi = 0.5$	Small
3	$[Mn^{+2}] = 0.25$ mM $\phi = 0.6$	Small
4	$[Mn^{+2}] = 0.25$ mM $\phi = 0.7$	Small
5	<i>n</i> -decane	Small
6	Aqueous 2.3 mM $NiCl_2$	Small
7	$[Ni^{+2}] = 8$ mM $\phi = 0.6$	Large
8	$[Ni^{+2}] = 6$ mM $[NaCl] = 0.154$ M	Large
9	$[Ni^{+2}] = 6$ mM $[NaCl] = 0.038$	Large
10	$[Ni^{+2}] = 6$ mM $[NaCl] = 0.077$	Large

Phantoms 1 through 5 were used for the reproducibility, susceptibility studies, and fat saturation imaging studies. Phantoms 7 through 10 were used for the study of the B_1 transmit and receive field studies.

3.3 Solution Characteristics

3.3.1 Stability and phase studies of the reverse micelle at $0.5 < \phi < 0.9$ values

Samples of the 0.1, 0.25, 0.50 mM Mn^{+2} reverse micelle solutions with ϕ values ranging from 0.50 to 0.9 were placed in a Haake FK water bath at 5° C. After equilibrating for one hour, the samples were taken out of the water bath and vigorously shaken. They were placed back into the water bath and allowed to equilibrate for one and one-half hours. The samples of the ϕ values of each $[Mn^{+2}]$ were taken out of the water bath and observed for color, opacity, and phase separation. The length of the phases were measured in cm. This was repeated for temperatures of 10, 15, 20, 25, 30, 35, and 40° C. A temperature - concentration stability diagram was constructed by plotting temperature versus ϕ and whether phase separation occurred.

3.3.2 Viscosity measurements

All viscosity measurements of the RM solutions were taken on a Brookfield Digital Viscometer (Stoughton, MA) model RVTDV-II with the UL adaptor spindle. Rotational speed of the spindle was 50 RPM. Sample temperatures were between 29 and 30° C. Measurements were taken of 0.5 mM $MnCl_2$ samples with ϕ values ranging from 0.5 to 0.80. All aqueous solution viscosities were measured using a Ubbelohde viscometer because their viscosities were so low. Densities of each of the samples were calculated and viscosity measurements in centipoise (cp) were taken.

3.3.3 Diffusion studies

Diffusion coefficients of AOT, water, and decane of the reverse micelle solution samples of 8 mM Ni^{+2} with ϕ values of 0.5, 0.6, 0.7, and 0.8 were measured using a 500 MHz Varian Unity narrow bore NMR spectrometer (Palo Alto, CA) located at Eastman Kodak Analytical Technology Division NMR lab. A pulsed field gradient spin-echo (PFGSE) NMR diffusion technique, where the width and separation of the gradients were fixed, was used to take the measurements. The amplitude of the gradient

varied by ten steps and the gradient values depended on the component, AOT, H₂O or decane, being measured. Plots of the square of the gradient and natural log of the amplitude of the spectral signal were made, and the diffusion coefficient was determined from the slope of this line [Antalek]. Measurements were taken at temperatures of 10 and 23° C.

3.4 Electrical properties

Three different experiments; dielectric studies, RF conductivity, and Q measurements help characterize the electrical properties of the reverse micelle solution. The experiments are described below.

3.4.1 Dielectric studies and RF Conductivity

In an electric field, such as a uniform parallel-plate capacitor, the ions in a solution will tend to migrate to the surface of an oppositely-charged plate. There is a force of attraction between charges that allows the two oppositely charged ions to either separate or stay together. The force of attraction can be explained using the equation

$$\text{Force} = - q_1 q_2 / (4\pi\epsilon_0\kappa r^2) \quad (41)$$

where the force is the attractive force between charges q_1 and q_2 of opposite sign, ϵ_0 is the permittivity of a vacuum, κ is the dielectric constant, and r is the distance between the charges. The higher the dielectric constant of the solution, the weaker the force of attraction holding the two charges together. Sometimes the charges in the solution are only partial charges of the molecules. The partial charges do not completely migrate to the oppositely-charged parallel plate capacitor, but become displaced. The partial charges behave like very small dipoles in the solution. The reorientation of the small dipoles in the solution is called the polarizability of the solution. As the number of the tiny dipoles increases in the solution, the solution becomes more polarized. The polarization effect can be described by the following equation

$$P = nql/v = Ql/v \quad (42)$$

where P is the polarization, n is the number of dipoles in volume v , $Q = nq$, or the charge of all dipoles, and Ql is the net dipole moment in volume v .

A general purpose Hewlett-Packard 4815A RF vector impedance meter was used to measure the complex impedance of several solutions at 22° C. The complex impedance was used to calculate the dielectric constant and resistivity for each of the solutions. The cell used in this study was constructed from a 25 ml glass bottle containing a 2.5 cm diameter and 4.5 cm length. The dimensions of the BNC female connector were 9 mm diameter and 12 mm length. The leads between the BNC connector and cell plates were kept at a minimum length and insulated with PVC to minimize their influence on the measurements. The area of the plates was 25 mm² with a distance of 2 mm between the them. Figure 19 shows a diagram of the cell used for this experiment.

Figure 19. Diagram of the conductivity cell for the dielectric and RF studies.

The dielectric constant and RF conductivity were calculated from the measured impedance and the phase angle of the solution and the following constants in Table 7 specific to this particular RF vector impedance meter and the equations:

$$Z = R / (1 + R^2\omega^2C_t^2) + i\omega[L_p - R^2C_t / (1 + R^2\omega^2C_t^2)] \quad (43)$$

$$\rho = (1 / (z^2 - 2\omega L_p z \sin\phi + \omega^2 L_p^2)) / z \cos\phi \quad (44)$$

$$L_p = 1/\omega [z \sin\phi + R^2 C_t \omega / (1 + R^2 \omega^2 C_t^2)] \quad (45)$$

Where R is the specimen resistance, ρ is the resistivity, and z is the magnitude of z from the impedance meter [Bottomley]. The dielectric constant and RF conductivity were determined for the following solutions listed in Table 8.

Table 7. Determination of the dielectric constant and RF conductivity

Cell Constant (l)	0.0388 meter
Probe Inductance (L_p)	3.23×10^{-08} Henries
Angular Frequency (ω)	$4.02 \times 10^{+08}$ Rad/s
Average Capacitance (C_t)	4.57×10^{-12} Farads
Shorted Cell Phase (ϕ)	89 degrees
Impedance of the cell (Z)	13 Ohms

Table 8. Solutions for dielectric constant and RF conductivity measurements

1) Air	14) Decane
2) HPLC H ₂ O	15) Ethylene Glycol
3) Tap H ₂ O	16) Silicon oil
4) Aqueous 8 mM NiCl ₂	17) RM, [Mn ⁺²] = 0.25 mM, ϕ = .5
5) Aqueous 0.10 mM MnCl ₂	18) RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.55
6) Aqueous 0.0385 M NS	19) RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.60
7) Aqueous 0.077 M NS	20) RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.65
8) Aqueous 0.154 M NS	21) RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.70
9) Aqueous 0.5 M NS	22) RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.75
10) Aqueous 1.0 M NS	23) RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.80
11) Aqueous 1.35 M NS	24) RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.85
12) Aqueous 1.5 M NS	25) RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.90
13) Toluene	26) [Mn ⁺²] = 0.25 mM, ϕ = 1.0

The solutions containing normal saline (NS) were chosen to determine how the dielectric constant is affected by the ionic strength of a solution. To determine the effect of increasing the ϕ value on the dielectric constant, several RM solutions containing the paramagnetic ion Mn⁺² were studied. The

organic solutions were chosen for comparisons of a nonconducting solution to those of the aqueous solutions and the RM solutions.

3.4.2 Q (Quality factor) Measurements

The quality factor (Q) is expressed by the following equation

$$Q = \text{energy stored in the inductor} / \text{energy dissipated in the circuit per RF cycle} \quad \text{Eqn. (45)}$$

[Kraus]. The higher the Q factor, the less power required to set up a B_1 field within the coil, and there is more signal obtained.

The signal-to-noise ratio (SNR) is a critical factor in MR image quality. An improvement in SNR may decrease scan times, increase spatial resolution, or decrease volume averaging errors [Lufkin].

The SNR can be related to the Q factor with the following equation

$$\text{SNR} \propto (Q_L n)^{-1/2} \quad (46)$$

where Q_L is the Q factor of the loaded coil, and n is the filling factor, which is the portion of the coil volume occupied by the sample. Coil loading and filling factors are a measure of how well the sample is matched to and fills the coils.

In a coil, the Q factor is defined by the equation

$$Q_{\text{coil}} = \omega_o L / R \quad (47)$$

where ω_o is the resonance frequency, L is the inductance, and R is the resistance of the coil [Poole]. The Q_{loaded} is represented by the equation

$$1 / Q_{\text{loaded}} = 1 / Q_{\text{coil}} + 1 / Q_{\text{sample}} \quad (48)$$

so, it takes into account the loading of the coil by the sample, or Q_{sample} , as well as the loading effects from the coil itself.

The definition of Q_{sample} is more complicated than the definition of the Q_{coil} ; therefore, the original definition for Q factor, that is eqn. (44) is used for simplicity and as a close approximation to Q_{sample} .

When the inductor of the coil is loaded by the insertion of a sample, Q_{loaded} will drop because of the additional absorption of the RF energy by the sample [Morris]. Translational energy of the molecules

in the solution is expended when the molecules attempt to align their charges and re-orient themselves in the electrical field. The translational energy of the molecules in the sample is part of the energy that is dissipated per RF cycle. The more energy absorbed by the rearrangement of the charges of the molecules in the solution, the more the energy is dissipated in the RF cycle, and the lower the Q_{loaded} factor. The Q factor for the samples was determined by the following equation

$$Q = \Delta\nu / \nu \quad (49)$$

where $\Delta\nu$ is the change in the frequency of the bandwidth of the peak at one-half height and ν is the frequency at the peak of absorption.

The Q measurements were done to determine how the solutions load the imaging coil. An ideal phantom should load the coil similar to the anatomy typically imaged with the coil [Hornak]. The sample solutions were contained in 25 ml glass bottles and placed inside a single turn solenoid (STS) coil. The Q value was measured using a Return Loss Bridge Tuning Apparatus consisting of an RF sweeper, oscilloscope, and 180° hybrid tee. The center frequency (ν) and width of absorption at half height ($\Delta\nu$) of the solution was measured.

Q measurements were calculated for the following solutions listed in Table 9. These solutions were used to compare the ability of the RM to load the imaging coil to the same extent as body tissue and ionic solutions.

Table 9. Solutions used for Q measurements

1) RM, $[\text{Mn}^{+2}] = 0.5 \text{ mM}$, $\phi=0.5$	6) RM, $[\text{Mn}^{+2}] = 0.5 \text{ mM}$, $\phi = 0.7$
2) Decane	7) Aqueous NaCl = 0.38 M
3) Aqueous NaCl = 0.077 M	8) Aqueous NaCl = 0.154 M
4) Arm (proximal)	9) Arm (distal)
5) Aqueous NaCl = 1.34 M	10) Air

3.5 Magnetic Properties

Several experiments were performed to characterize the magnetic properties of the reverse micelle solution. These experiments are discussed below.

3.5.1 NMR studies

Low resolution NMR spectra were recorded for the reverse micelle solutions with $[\text{Mn}^{+2}] = 0.25$ mM at all ϕ values, decane, and H_2O , using a 1.5 T GE Signa magnetic resonance imager and single turn solenoid coil. The sample solutions were placed in 25 ml glass bottles. A spin-echo pulse sequence with the imaging parameters listed in Table 10, was utilized.

Table 10. Imaging parameters for spectral study

TR	1000 ms
TE	20 ms
Slice Thk	3 mm
Slice	axial

The spectra were recorded using the center frequency adjust (CFA) fine control variable of the precscan mode. The spectra from a reverse micelle solution, decane, and H_2O will be presented in the results and discussion section of this thesis.

The AOT peaks of the reverse micelle spectra could not be seen and were thought to be buried in both the H_2O and decane peaks. In order to record the peaks of the AOT, ≈ 4 grams of AOT was dissolved in ≈ 16 grams of CCl_4 and placed in a 25 ml glass bottle. The spectra of AOT was obtained using the same imaging hardware and parameters described above. Figure 20 displays the spectra from the AOT.

Figure 20. Low resolution spectrum of AOT in CCl_4

3.5.2 Echo modulation studies

Echo modulation can be present in solutions containing compounds with spin-spin splitting. The presence of echo modulation can cause changes in intensity of the phantom's image with the lengthening of the TE value. Since decane is a 10 carbon chain, the hydrogens present will have different environments or chemical shifts from hydrogens that are three bond lengths away. Spin-spin splitting will take place. The RM solutions contain decane, therefore, spin-spin splitting should occur, and echo modulation would be present. Because aqueous solutions contain only H₂O, echo modulation should not be present. The effect of echo modulation on the RM solution was measured and compared to the echo modulation effects from decane and the aqueous solutions. The solutions studied are listed in Table 11.

Table 11. Solutions for echo modulation studies

-
- 1) RM, [Ni⁺²] = 8 mM, ϕ = 0.5
 - 2) RM, [Ni⁺²] = 8 mM, ϕ = 0.6
 - 3) RM, [Ni⁺²] = 8 mM, ϕ = 0.7
 - 4) decane
 - 5) aqueous 2.3 mM NiCl₂
 - 6) 8 mM NiCl₂
-

These solutions were placed in 100 ml glass bottles. Signals from each of these solutions were obtained using a single spin-echo pulse sequence, head coil, and the following imaging parameters in

Table 12.

Table 12. Imaging parameters for echo modulation studies

TR	1000 ms
TE	12,15,20,25,30,...,340,360,...,420 ms
FOV	24 cm
Matrix	256 X 128
Slice thk	2 cm
Slice	coronal

3.5.3 Determination of T₁ and T₂ values for different concentrations of paramagnetic ions.

In order for the T₁ and T₂ values to be within a desired range, the T₁ and T₂ values of different paramagnetic ion concentrations must be studied. Aqueous solutions of NiCl₂ and MnCl₂ were made up

(See section 3.21). The T_1 and T_2 's of these solutions were found using a spin-echo pulse sequence, the head coil and the following imaging parameters for the T_1 values listed in Tables 13 and 14. The T_2 values were obtained by keeping the TR value constant and varying the TE. The software on the GE Signa imager calculated T_1 and T_2 by a nonlinear least squares method (NLLS) [Gong, Li].

Table 13. Imaging parameters for T_1 determination

TR	6000,2000,1000,750,500,167,150 ms
TE	15 ms
FOV	24 cm
Matrix	256 X 128
Slice thk	3 cm
Slice	axial

Table 14. Imaging parameters for T_2 calculation

TR	1000 ms
TE	15,20,25,40,50,60,70,80,100,105,140,150,200,300,400 ms
FOV	24 cm
Matrix	256 X 128
Slice thk	10 cm
Slice	axial

3.5.4 Variable temperature studies

The T_1 and T_2 values of the reverse micelle solution must be independent of temperature to be effective as a mass produced MRI phantom, which might be used at several different temperatures. The variable temperature studies were performed to find the paramagnetic ion concentration and phi value containing the least temperature-dependent T_1 and T_2 values. The procedure for determining the optimum paramagnetic ion concentration and phi value is discussed below.

A variable temperature apparatus was designed to be used in the imager. It was made out of a nonmagnetic material to avoid being pulled into the imager's magnet. Since leaking of water from the apparatus could damage the imager, the apparatus contained a water-tight seal. The apparatus allowed the samples to be cooled or heated during the scanning process by flowing water from a water bath located outside the magnet room through the apparatus and back into the water bath. A motion artifact, which is produced by the water flowing during the scan, was eliminated by clamping off the water flow while

scanning. The water was allowed to flow between scans and hence keep the samples at the desired temperature. Another method, which allowed the water to flow during the scan was devised. The flowing water was doped with 10 mM MnCl_2 causing a short T_2 and eliminating the signal from the water.

A conventional Haake FK water bath, located outside the magnet room of the imager, was used to regulate the temperature of the circulating water to keep the samples at the desired temperature. The water bath was connected to the inflow and outflow ports of the sample container by six meters of 1.3 cm I.D. Tygon tubing.

The cylindrical sample container was made of polyvinyl chloride. The dimensions of the cylindrical bottom were: 15.2 cm I.D., 16.8 cm O. D., 12.6 cm height. A 3 mm sheet of PVC was glued with a PVC adhesive to the bottom of the cylinder to close it. The cover of the apparatus was 1.5 cm in height and was closed by a 3 mm sheet of PVC glued with PVC adhesive to the top. A water tight seal was maintained by placing an o-ring and silicone stopcock grease around the base of the cover. The cover was connected to the bottom by six nylon screws. Figure 21 shows a cross-sectional view of the variable temperature apparatus containing sample bottles. The water flowed at a rate of 2.6 liters per minute and the temperature difference between the water bath and the sample container varied $\approx \pm 0.7^\circ\text{C}$.

Figure 21. Cross sectional view of variable temperature apparatus containing four samples. The water flows from the water bath outside the imager into the in port. The water returns to the conventional water bath via the out port.

Samples of the reverse micelle solution of $[\text{Mn}^{+2}] = 0.5, 0.25, \text{ and } 0.1 \text{ mM}$ with ϕ values ranging from 0.5 to 0.85 contained in 25 ml glass bottles were placed inside the variable temperature apparatus. Samples containing $[\text{Ni}^{+2}] = 8, 5, \text{ and } 3 \text{ mM}$ with ϕ values ranging from 0.5 to 0.75 were also studied in the same manner.

The variable temperature sample container was placed inside the head coil of the imager. A multislice spin-echo pulse sequence was used to obtain the images of the samples and signal for T_1 calculation from each of the samples. The imaging parameters used in this procedure were the same as in Table 13 except the FOV was 14 cm.

For the calculation of T_2 , signal and images for the samples were obtained using the imaging parameters outlined in Table 14 with the exception of the 14 cm FOV.

The above procedure was used for acquiring images and signal at the following temperatures: 5, 13, 22, 28, and 35° C.

T_1 and T_2 values were calculated by two different nonlinear least squares (NLLS) algorithms and three different methods.[Gong, Li] The three methods will be referred to as the GE Signa imager, XMRITools, and Vax. The GE Signa Imager used a NLLS algorithm located on the imager. XMRITools was an X-windows based C-program operating on UNIX platforms. The Vax method was a Fortran program running under the Vax VMS operating system. Both XMRITools and the Vax method used an NLLS algorithm.

3.5.5 Stability/Reproducibility studies

The T_1 and T_2 of an MRI phantom must be constant regardless of temperature changes between use, mixing, and age of the solution inside the phantom. Therefore, the effects of temperature and mixing on the T_1 and T_2 of the reverse micelle solution was studied. The procedure for studying these effects is described below.

Seven samples of the reverse micelle solution with $[Mn^{+2}] = 0.25$ mM and ϕ value of 0.6 were made up and placed in 25 ml glass bottles. Two of the samples were sonicated with an ultrasound wand for 10 minutes and left at room temperature. Two samples were placed in a water bath at 30 °C and shaken after one hour. After 24 hours, the samples were shaken again and allowed to equilibrate. After an additional 24 hours, the samples were taken out of the water bath and allowed to settle at room temperature. Two samples were placed in a water bath at 5° C and the same procedure for the samples at 30° C was followed. Three samples were left undisturbed at room temperature.

The samples were placed inside the head coil and imaged at room temperature. The imaging parameters described in Table 13 and 14 were used to image the samples. The signal amplitude was measured using the CFA fine variable control of the manual prescan mode on the imager. The signal amplitude, TR, and TE values were used to calculate the T_1 and T_2 values by the Vax method. The T_1 and T_2 values of each of the samples were compared to the T_1 and T_2 values previously measured at room temperature and to each other.

Reproducibility of the T_1 and T_2 over time was studied using four different small phantoms. The phantoms contained solutions of aqueous 0.25 mM MnCl_2 , decane, reverse micelle with $[\text{Mn}^{+2}] = 0.25$ and ϕ of 0.5, or reverse micelle with $[\text{Mn}^{+2}] = 0.25$ and ϕ of 0.7. The phantoms were placed inside the head coil and imaged using the same imaging parameters described in Tables 13 and 14. The signal intensity was measured using two different procedures. Images of the phantoms were taken and the T_1 and T_2 were calculated using the signal intensity of the image by the XMRITools method. Signal intensity was also found by measuring the signal amplitude of the phantom using the CFA fine variable control of the manual prescan mode. T_1 and T_2 were calculated using the Vax methods. The phantoms were imaged on different days over a period of 8 months.

3.5.6 Magnetic susceptibility studies

Distortion of the image of a standard sample could occur due to changes in magnetic susceptibility of samples which are placed adjacent to the standard sample. These distortions should be most apparent at the corners of the rectangular-shaped standard samples. The amount of distortion of one sample caused by another sample was measured. Two large clear plastic culture bottles and six 20 ml polyethylene bottles contained the following solutions listed in Table 15. The samples of decane, normal saline, and HPLC H_2O and AOT/ H_2O mixture were used since they were components of the RM. The two RM samples were chosen because they did not phase separate until greater than 35°C . The large bottles contained the references decane and aqueous 8 mM NiCl_2 .

Table 15. Solutions for susceptibility studies

large culture bottles	small bottles
1) decane	1) decane
2) aqueous 8 mM NiCl_2	2) normal saline
	3) HPLC H_2O
	4) RM, $[\text{Mn}^{+2}] = 0.25\text{ mM}$, $\phi = 0.5$
	5) RM, $[\text{Mn}^{+2}] = 0.25\text{ mM}$, $\phi = 0.7$
	6) AOT/ H_2O mixture, $\phi = 1.0$.

A stand to hold the bottles in place was made of PVC, which allowed a large bottle filled with decane or aqueous NiCl_2 to be placed at a reproducible location next to a smaller bottle containing one of

the six solutions. Figure 22 shows a diagram of the stand and the placement of the large and small bottles.

Figure 22. Arrangement of the large bottle next to the small bottle in the support stand.

The sample bottles were placed in the head coil of the imager and imaged using the imaging parameters described in Tables 13 and 14, except the slice selected was coronal. Images of the large bottle filled with decane and each of the six samples were obtained. These images were compared to an image of only the large bottle to determine if any distortion to the image of the large bottle by the smaller bottle occurred. Comparisons of the amount of distortion from each of the small bottles on the large bottle were made. The difference in the susceptibility of the material in the or small bottle is expected to distort the shape of the adjacent corner of the standard sample relative to the opposite corner.

3.5.7 Transmit and receive B_1 field mapping.

System performance testing of the MR imager relies on using a phantom that produces a uniform image. The aqueous phantoms currently used show a low frequency intensity shading when imaged in the body and surface coils. Transmit and receive B_1 field mapping is used by the manufacturer of the imager to study the effects of transmit/receive coil coupling in both the transmit and receive mode, B_1 homogeneity/penetration versus coil design and patient loading, and spatial variation in receive sensitivity profile.[Maier] A reverse micelle solution phantom was studied to determine its effectiveness as a tool to accurately assess RF coil effects.

The following large phantoms in Table 16 were included in this study. Phantom solution 3 is the GE standard phantom. Phantoms 4 through 6 were studied to approximate the loading of the coil by the

body tissue and to see the effects of increasing the ionic strength of an aqueous solution. The RM solution was used for comparison to the GE standard phantom and the other phantoms to test its performance in this study.

Table 16. Phantom solutions for TR mapping.

1) RM $[\text{Ni}^{+2}] = 8 \text{ mM}$ $\phi = 0.6$
2) Acetone
3) Aqueous 14 mM NiCl_2
4) Aqueous 6 mM NiCl_2 with $[\text{NaCl}] = 0.154 \text{ M}$
5) Aqueous 6 mM NiCl_2 with $[\text{NaCl}] = 0.038 \text{ M}$
6) Aqueous 6 mM NiCl_2 with $[\text{NaCl}] = 0.077 \text{ M}$

A GE Signa 1.5 T magnetic resonance imager located at University Medical Imaging, Rochester, NY, was used for this study. A procedure called TR mapping was used to obtain images, transmit and receive maps, and profiles of the transmit and receive maps [Maier]. The imaging parameters for this study are listed in Table 17.

Table 17. Imaging parameters for TR mapping study.

Imaging coil	low pass body coil
TR	2240 ms
TE	20 ms
FOV	48 cm
Matrix	256 X 128
Scan thk	3 mm
Slice	Axial

3.5.8 Fat saturation studies.

The reverse micelle solution contains two NMR spectral peaks that coincide with the water and fat peaks of the human body. The following experiment was done to determine the suitability of the reverse micelle solution for testing the fat-saturation imaging sequence.

Five solutions were made and placed in 125 ml glass bottles. These solutions are listed in Table 18. The H_2O solutions were used as standards and the three RM solutions were chosen to study the effects of increasing the ϕ value on the fat saturation.

Table 18. Solutions for fat saturation studies.

-
- 1) H₂O
 - 2) decane
 - 3) RM, [Ni²⁺] = 8 mM, ϕ = 0.5
 - 4) RM, [Ni²⁺] = 8 mM, ϕ = 0.6
 - 5) RM [Ni²⁺] = 8 mM, ϕ = 0.7
-

These solutions were arranged in the variable temperature sample container and bathed with aqueous 2 mM NiCl₂ at room temperature. The variable temperature sample container was placed in the head coil and the following imaging parameters of a spin-echo pulse sequence and the standard GE Chemsat sequence listed in Table 19 were used.

Table 19. Imaging parameters for fat saturation study.

TR	1000 ms
TE	15 ms
FOV	18 cm
Matrix	256 X 128
Slice thk	10 mm
Slice	Axial

The CFA fine control of the manual prescan mode of the imager was used to obtain NMR spectra of the five samples and the aqueous 2 mM NiCl₂ solution. The observe frequency was centered on the resonance peak of water. The offset frequency of the saturation pulse was adjusted by changing the offset frequency control variable *off_rfcssat* on the imager. The frequency offset ranged $\pm$ 700 Hz in steps of 20 Hz plus two values at $\pm$ 1200 Hz from the center observe frequency. Signal amplitude of the water or fat was measured for each sample. Plots of signal intensity versus frequency offset were made for each of the solutions.

3.5.9 T₁ values of the individual components of the reverse micelle system.

The main goal of this study was to measure spectral T₁ values of the individual components of the RM system and to determine if spin exchange would occur between the H₂O and AOT reservoir and AOT and decane reservoir. The following solutions illustrated in Figure 23 were made and placed in 125 ml glass bottles. The 0 mM and 8 mM are the two extremes of the solutions used in all of the studies. The ϕ values of 0.6 more closely matches the desired T₁ values.

[Ni ²⁺]	$\phi = 0.50$	.55	0.60	0.65	0.70
0 mM	x	x	x	x	x
3 mM			x		
5 mM			x		
8 mM	x	x	x	x	x

Figure 23. RM solutions used in T_1 of the individual component study. The x areas indicate solutions in the study.

The individual sample bottles were placed in the head coil of the imager. A single slice spin-echo pulse sequence and a PSD software program and the following imaging parameters listed in Table 20 were used for calculation of the T_1 of water and decane.

Table 20. Imaging parameters for individual spectral T_1 study.

TR (ms)	6000,5000,4000,3500,3000,2750,2500,2250,2000,1750,1250,1000,900,800,700,600,500,400,300,150
TE (ms)	100
FOV (cm)	12
Matrix	256 X 128
Slice thk (mm)	10
Slice	Axial

The T_1 of the AOT was found by changing the TE to 15 ms and the above imaging parameters. Signal amplitude was measured of the water, decane, AOT using the CFA fine variable control of the manual prescan mode on the imager. The T_1 was calculated for the three components using the Vax method. The calculated T_1 of each component was used to determine a signal for each TR value. Plots of measured signal amplitude versus TR and calculated signal amplitude versus TR were made. These plots were used to attempt to determine the exchange rate between the components.

4.0 Results and Discussion

The results obtained from the experiments and a discussion of the results are presented in this section. The format of this section is similar to the experimental section; therefore, the results from the completed experiments are placed into four categories: the Mn^{+2} and Ni^{+2} solutions used in this study, the physical properties, electric properties, and the magnetic properties of the reverse micelle solution. The results of the experiments will be arranged in tabular or graphical form and a discussion of the results will follow.

4.1 Concentration of the paramagnetic ions for control of T_1 and T_2 of the reverse micelle solution

One of the goals is to develop a phantom solution that produced T_1 and T_2 's similar in value to human tissue. The following solutions were equilibrated so that atmospheric oxygen was present in solution. Table 21 displays the T_1 and T_2 values obtained for an aqueous solution containing the paramagnetic ion Mn^{+2} . The inverse of the T_1 and T_2 is also included for the purpose of developing a graph for the estimation of T_1 or T_2 values within the range of 0 to 2.5 mM Mn^{+2} . Figures 24 and 25 depict the graphs of either $1/T_1$ or $1/T_2$ for a range of Mn^{+2} concentrations.

Table 21. T_1 and T_2 values of aqueous solutions of the paramagnetic ion Mn^{+2}

$[\text{Mn}^{+2}](\text{mM})$	T_1 (ms)	$1/T_1$ (s^{-1})	$T_2(\text{ms})$	$1/T_2(\text{s}^{-1})$
2.5	68.5	14.60	6.5	153.85
1	185	5.41	15.2	65.79
0.5	374	2.67	29.4	34.01
0.25	672	1.49	52.2	19.16
0.1	1311	0.76	106.1	9.43
0.05	1911	0.52	160.4	6.23
0	3173	0.32	364.9	2.74

The following equations were derived for aqueous Mn^{+2} solutions from the T_1 and T_2 measurements and Figures 24 and 25:

$$1/T_1 = 5722 \text{ s}^{-1} \text{ M}^{-1} [\text{Mn}^{+2}] + 0.0846 \text{ s}^{-1} \quad (50)$$

$$1/T_2 = 60386 \text{ s}^{-1} \text{ M}^{-1} [\text{Mn}^{+2}] + 3.644 \text{ s}^{-1} \quad (51)$$

From these equations, T_1 and T_2 values for a certain $[Mn^{+2}]$, or for a $[Mn^{+2}]$, a particular T_1 or T_2 can be predicted. This is important when certain T_1 and T_2 values are needed, or the $[Mn^{+2}]$ is unknown, but the T_1 and T_2 's are known.

Figure 24. A graph of the inverse of the T_1 in s^{-1} versus the concentration of Mn^{+2} . From this graph, the T_1 could be determined for a Mn^{+2} concentration ranging from 0 to 2.5 mM. The higher the concentration of Mn^{+2} , the shorter the T_1 becomes.

Figure 25. A graph of the inverse of the T_2 in s^{-1} versus the concentration of Mn^{+2} . From this graph, the T_2 could be determined for a Mn^{+2} concentration ranging from 0 to 2.5 mM. The higher the concentration of Mn^{+2} , the shorter the T_2 becomes.

For system performance testing, another range of T_1 and T_2 values was needed. This range contained a T_1 between 400 and 500 ms, and a T_2 between 85 and 100 ms. This range could not be achieved with the addition of Mn^{+2} ion in the aqueous phase, so instead the paramagnetic ion Ni^{+2} was chosen. The T_1 and T_2 values for the aqueous Ni^{+2} concentrations were determined from the following equations:

$$1/T_1 = 632 \text{ s}^{-1} \text{ M}^{-1} [Ni^{+2}] + 0.337 \text{ s}^{-1} \quad (52)$$

$$1/T_2 = 691 \text{ s}^{-1} \text{ M}^{-1} [Ni^{+2}] + 1.33 \text{ s}^{-1} \quad (53)$$

A comparison in tabular form of the T_1 and T_2 for human tissue [Fletcher], the three aqueous solutions containing Mn^{+2} , including the calculated T_1 and T_2 for the aqueous Ni^{+2} solutions, are shown in Table 22.

Table 22. T_1 and T_2 values of $[Mn^{+2}]$ and $[Ni^{+2}]$ solutions and human tissue

Solution/Tissue	T_1 (ms)	T_2 (ms)
$[Ni^{+2}] = 0 \text{ mM}$	2967.4	882.6
$[Ni^{+2}] = 3 \text{ mM}$	447.8	311.9
$[Ni^{+2}] = 5 \text{ mM}$	285.9	217.9
$[Ni^{+2}] = 6 \text{ mM}$	242.2	189.4
$[Ni^{+2}] = 8 \text{ mM}$	185.4	150.1
$[Mn^{+2}] = 0 \text{ mM}$	3173	364.9
$[Mn^{+2}] = 0.10 \text{ mM}$	1311	106.1
$[Mn^{+2}] = 0.25 \text{ mM}$	672	52.2
$[Mn^{+2}] = 0.5 \text{ mM}$	374	29.4
white matter	760-1080	61-100
grey matter	1090-2150	61-109
meninges	500-2200	50-165
muscle	950-1820	20-67
adipose	200-750	53-94

Aqueous solutions containing 0.5, 0.25, and 0.1 mM Mn^{+2} were chosen to dope the aqueous phase of the reverse micelle because of their similarity in T_1 and T_2 to human tissue. The concentrations of Ni^{+2} chosen to dope the aqueous phase of the reverse micelle for use in the phantom solution for system performance testing were 3, 5, and 8 mM. These concentrations were chosen for their T_1 and T_2 values within the desired range for system performance testing.

4.2 Physical properties of the reverse micelle solution

The physical properties of the reverse micelle include the stability, viscosity, and diffusion of the RM and reference solutions. The results of these studies are discussed in the following paragraphs.

4.2.1 Phase and stability studies

Phase diagrams of this particular reverse micelle solution have been described in detail elsewhere [Kotlarchyk, 1985]. The results of that previous study were confirmed by the stability diagrams below. Figure 26 represents the stability diagrams for the reverse micelle containing Mn^{+2} in the aqueous phase. Figure 27 represents the stability diagrams for the reverse micelle containing Ni^{+2} in the aqueous phase. The ϕ values ranged from 0.5 to 0.95, and the temperatures ranged from 0 to 40 °C. The staircase line designates the boundaries between the RM phase, and coexistence region and lamellar phase of the solution. The stability diagrams for all concentrations of paramagnetic ions are similar, with only a few exceptions. The slight variations found in the delineation line between the Mn^{+2} or Ni^{+2} concentrations are not attributed to the paramagnetic ion concentration but to the limitations of the experimental procedure used to determine these boundaries.

The delineation line was constructed from the following observations. The solution in the reverse micelle phase was clear and homogeneous throughout the sample. At higher temperatures, the sample solution was still clear, but two distinct upper and lower layers were observed. The upper layer had an oily appearance, while the lower layer resembled water. Some sample solutions had only a small upper layer or the upper layer looked as though it was mixed into the lower layer. At ϕ values greater than 0.8, the solution appeared opaque or gel-like at higher temperatures. Since there was difficulty distinguishing the coexistence phase of the solutions by observation, the stability diagrams for the six solutions are only rough approximations.

Figure 26. Stability diagram for RM solutions contain $[\text{Mn}^{+2}] = 0.1, 0.25, \text{ and } 0.5 \text{ mM}$. The ϕ values of the RM solution ranged from 0.5 to 0.90, while the temperatures ranged from 5 to 40 °C.

Figure 27. Stability diagram for RM solutions contain $[\text{Ni}^{+2}] = 3, 5, \text{ and } 8 \text{ mM}$. The ϕ values of the RM solution ranged from 0.5 to 0.90, while the temperatures ranged from 5 to 40 °C.

For both the Mn^{+2} and Ni^{+2} RM solutions, ϕ values of 0.5 and 0.55 were RM phase-stable over the entire range of temperatures. As the ϕ value increased, the range of temperatures where the RM phase is stable becomes smaller. This is shown in the stability diagrams, where the region to the left of the staircase line contains the RM-stable-phase. The region to the right of the staircase line is the region of the coexistence and lamellar phase of the RM solution as shown in Figures 26 and 27.

4.2.2 Viscosity measurements

Table 23 contains the viscosity measurements for a RM solution containing $[Mn^{+2}] = 0.25$ mM in the aqueous phase as well as the viscosity measurements for aqueous 8 mM Ni^{+2} , 14 mM Ni^{+2} , 6 mM Ni^{+2} with 0.154 mM NS, HPLC H_2O , and decane.

Table 23 Viscosity results of RM solutions and reference solutions at 24° C

Phi/solution	Spindle speed (RPS)	Viscosity (cP)
0.5	50	25.0
0.55	50	29.1
0.60	50	34.8
0.65	50	43.0
0.70	50	56.4
0.75	20	35.5
8 mM Ni^{+2}		0.8924
14 mM Ni^{+2}		0.8893
6 mM Ni^{+2} with 0.154 M NS		0.9045
H_2O^*		0.8900
<i>n</i> -decane*		0.9200

*Literature value - [CRC]

Observations of the RM samples establishes that the solutions become thicker as the ϕ increases. When ϕ increases, the concentration of the RM's increase, while the amount of the lower viscosity component - decane, decreases. Therefore, one would expect the viscosity of the RM solution to increase as the ϕ value increases.

In a liquid, viscosity (η) can be expressed as

$$\eta \propto e^{\Delta E_{vis}/RT} \quad (54)$$

where ΔE_{vis} is an activation energy for the rate of viscous flow, R is the gas constant, and T is the temperature in Kelvin [Atkins]. The term $\Delta E_{vis}/RT$ is the Boltzmann factor for the fraction of molecules having enough energy to surmount the barrier for flow processes to occur.

When taking viscosity measurements, the moving layers of fluid in the solution produce a viscous force, and a transfer of momentum occurs. Intermolecular attractions cause the momentum transfer, which creates a frictional drag between the moving layers [Moore]. The frictional drag from the intermolecular attraction in the RM solution, due to partial overlap of the AOT tails, is increased as the ϕ , or concentration of RMs increases, resulting in an increased viscosity [Kim].

The viscosity was found to be independent of the paramagnetic ion or salt concentration for the aqueous solutions. Comparisons of the RM solution to H_2O or decane show the RM solution to be 25 to 56 times greater in viscosity than either the H_2O or decane. The viscosity of the RM solution with a ϕ of 0.6 is similar to that of 30W motor oil.

4.2.3 Diffusion measurements

Table 24 is comprised the diffusion coefficients of the H_2O , decane, and AOT components of the RM solution with ϕ values of 0.5, 0.6, and 0.7 containing $[Mn^{+2}] = 0.25$ mM in the aqueous phase at both 22 and 10° C.

Table 24. Diffusion coefficients for components of an RM solution

Temperature (° C)	ϕ	Water (cm ² /s)	Decane (cm ² /s)	AOT (cm ² /s)
22	0.5	6.39×10^{-7}	5.04×10^{-6}	8×10^{-8}
	0.6	7.29×10^{-7}	3.96×10^{-6}	6×10^{-8}
	0.7	1.69×10^{-6}	2.89×10^{-6}	1.06×10^{-7}
10	0.5	4.04×10^{-8}		2.71×10^{-8}
	0.6	4.3×10^{-8}		1.71×10^{-8}
	0.7	5.22×10^{-8}		1.27×10^{-8}
22		1.7×10^{-5}	$1.56-1.37 \times 10^{-5}$	

At 22° C, general trends for the diffusion coefficient show decane has the largest diffusion coefficient followed by H_2O , then AOT having the smallest diffusion coefficient. The explanation of these values is not completely understood.

At ϕ values of 0.50 to 0.70, the RM are close together in solution which allows the RM to collide and emerge to form aggregates momentarily. As ϕ increases, the more likely the RM will form the aggregates and the merged RM will produce open H₂O channels varying in size depending on how large the aggregate is and may become bicontinuous in nature. The H₂O will be able to move more freely inside the larger RM aggregates and the H₂O will be less viscous [Chen, 1989].

Although H₂O is relatively immiscible, there is still a small amount of H₂O that will be soluble in the decane. The diffusion coefficient is determined by the weighted average of the RM moving in the solution and the movement of the H₂O molecules which are partially soluble in the decane.

Similarly, the diffusion coefficient of the AOT is determined from the weighted average from the AOT molecules in the decane that have not encapsulated the H₂O to form a droplet and the movement of the entire RM itself. Rotational movement of the tails of the AOT must also be taken into account.

A quantitative diffusion coefficient of $\approx 3-6 \times 10^{-7}$ cm²/s has been reported for the RM [Antalek, 1996]. Predicted diffusion coefficient for the H₂O and AOT should be in this range because the AOT is present on the surface of the H₂O and makes up the RM [Strius]. Although our diffusion coefficients are not the same for the H₂O and AOT component of each ϕ value, we are unable to determine the behavior of our system without taking the measurement of the diffusion coefficient of H₂O in pure decane and AOT in pure decane. This may lead to understanding the formation of the aggregates to a more bicontinuous phase as ϕ is increased in our system [Chen, 1989].

When the temperature is decreased to 10° C, the diffusion coefficient of the H₂O and AOT also decreases. The first conclusion is that the molecular motions are decreased due to the lowered temperature, but the decrease in temperature is not significant enough to warrant such a decreased diffusion coefficient. At 10° C, the RM solution may be going through another phase change as evidenced by the appearance of the clear solution to an opaque milky solution as the temperature is decreased. The diffusion coefficient of the two components are within a range of $1.27 - 5.22 \times 10^{-8}$ cm²/s which suggests the movement of the entire RM at this temperature for both components.

The diffusion coefficient of decane in the RM solution is an order of magnitude less than the diffusion coefficient for pure decane. The decane in solution may be more viscous than pure decane, thus it will have a decreased diffusion coefficient. A slight decrease in the diffusion coefficient is seen as ϕ is increased. This may be attributed to the smaller amount of decane present and an increase in viscosity as ϕ is increased.

Further study is needed in this area to better understand what is happening in the system and how temperature and increasing ϕ affect the diffusion coefficient.

4.3 Electrical properties of the RM solution

The dielectric constant, resistivity, and the ability of the RM solution to load the imaging coil were determined for ϕ values of 0.5 to 0.9 of the RM solution. The results of these studies are explained below.

4.3.1 Dielectric constant of the RM solutions

The dielectric constant of a solvent is a measure of how well the solvent can separate oppositely charged ions or how polarizable a solution is [Harris]. The dielectric constant of HPLC H_2O is 81.07, which means the attractive forces between the charges is weak and it can separate ions very well in solution. Upon the addition of 8 mM Ni^{+2} or 0.1 mM Mn^{+2} the dielectric constant does not change. With the addition of NaCl in concentrations of 0.0385 M to 1.5 M, the ionic strength of the solution increases, which resulted in a larger dielectric constant than H_2O . Each of the dielectric constants from the above solutions were used as a baseline for comparison to the RM solution. The dielectric constants of the aqueous 8 mM Ni^{+2} , 0.154 M NaCl, and decane solutions were especially important since they are components of the RM solution. The dielectric constant of the aqueous Ni^{+2} and NaCl solutions ranged from 82 to 88. The literature value for the dielectric constant of decane is 1.99 [CRC]. Decane is nonpolar in nature and does not separate charges well in solution, thus its dielectric constant is low. The dielectric constant of the solutions are listed in Table 25. Also, seven reference solutions are listed with the dielectric constants taken from the *Handbook of Chemistry and Physics*.

Table 25. Dielectric constant, resistivity, and conductivity values

Sample solution	Dielectric constant	Resistivity (Ω m)	Conductivity (Ω m ⁻¹)
Air	1*	372	0.0027
HPLC H ₂ O	81*	39.6	0.0253
Toluene	2.38*	324	0.0031
Decane	1.99*	404	0.0025
Ethylene glycol	37*	60.6	0.0165
Silicon oil	4*	473	0.0021
Glycerin	56*	17.6	0.0568
HPLC H ₂ O	81.07	39.6	0.0253
Tap H ₂ O	80.73	24.8	0.0403
0.1 mM Mn ⁺²	79	50.5	0.0198
8 mM Ni ⁺²	81.69	5.39	0.1856
0.0385 M NaCl	83.64	1.89	0.5291
0.077 M NaCl	84.01	1.16	0.8621
0.154 M NaCl	88.18	0.9	1.1111
0.5 M NaCl	160.2	0.29	3.4483
1.0 M NaCl	276.6	0.19	5.2632
1.35 M NaCl	345.6	0.16	6.2500
1.5 M NaCl	355	0.14	7.1429
RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.5	27.9	8.04	0.1244
RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.55	31.48	7.25	0.1379
RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.60	34.69	5.61	0.1783
RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.65	37.58	4.29	0.2331
RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.70	42.26	3.37	0.2967
RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.75	42.64	2.74	0.3649
RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.80	49	2.21	0.4525
RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.85	55.94	1.76	0.5682
RM, [Mn ⁺²] = 0.25 mM, ϕ = 0.90	64.67	2.15	0.4651
AOT + H ₂ O	103.8	1.74	0.5747

*CRC values

The dielectric constant of the RM solution ranged from 27.9 for a ϕ of 0.5 to 42.64 at a ϕ of 0.75.

The RM solution phase separated above a ϕ of 0.75, but the dielectric constant continued to increase with ϕ .

The RM is encapsulated by the AOT; therefore, the H₂O, Na⁺, paramagnetic ions, and the negatively charged polar heads of the AOT are not able to separate in the solution. Instead, the positive and negative charges of the Na⁺, Ni⁺², or Mn⁺² and AOT may become displaced and behave like small dipoles. As ϕ is increased, there are more of the small dipoles and the polarization of the solution increases, consequently, the dielectric constant increases.

When the RM solution phase separates into the lamellar phase, the migration of the partial charges from the H₂O and AOT can be achieved because there are no boundary restrictions of either the H₂O or AOT. Therefore, the force of attraction between two charged particles may decrease, allowing the charges to be separated in solution, and the dielectric constant to increase [Harris].

The dielectric constant for a mixture of AOT and H₂O with a mass ratio of 0.6 was found to be 103.8 and is greater than that of HPLC H₂O. The AOT and H₂O are freely moving about in the solution since the reverse micelles are not formed; consequently, the migration of the charges from the AOT, H₂O, and paramagnetic ions is not restricted. These charges will be able to align in an electrical field of a parallel-plate capacitor and a larger dielectric constant is produced [Kraus]. The role of the dielectric constant will be explained further in the magnetic properties section of the thesis.

4.3.2 Resistivity and conductivity measurements

Table 25 contains the resistivity and conductivity measurements for the reference and RM solutions. A comparison of these values to the different reference values is discussed below.

The resistivity of the HPLC H₂O is 39.6 $\Omega \cdot m$ while the resistivity of tap H₂O, which contains many impurities, is 24.8 $\Omega \cdot m$. The resistivities of the 0.1 mM Mn²⁺ and 8 mM Ni²⁺ were 50.5 and 5.39 $\Omega \cdot m$. These two resistivities did not seem valid when compared to the HPLC and tap H₂O resistivities. Since the resistivities of the 0.1 mM Mn²⁺ and 8 mM Ni²⁺ were not essential for this study, repeat measurements were not taken. When the ionic strength of the solution is increased by adding NaCl, as shown by the solutions containing NaCl in concentrations of 0.0385 M to 1.5 M, the resistivity is decreased from 1.89 $\Omega \cdot m$ to 0.14 $\Omega \cdot m$.

Literature states the conductivity of a RM solution depends upon the concentration of reverse micelles per unit volume in the solution, and also that as the concentration of reverse micelles increases, the conductivity increases [Eicke]. The increase in conductivity, as stated in the literature, can be attributed to electrostatic energy between the droplets or the charge fluctuation model [Tomic]. Our results confirm the literature. The resistivity showed an overall decrease and the conductivity an increase as ϕ increased. At ϕ values of 0.5 in the RM solution, a percolation transition must occur to some extent.

allowing conduction of electrical current. As ϕ increases, the RM's are closer packed and begin to touch. Then, percolation is more pronounced, providing a pathway for electrical current to be conducted.

Upon comparison of the RM solutions to 1.35 M NaCl, HPLC H₂O, and decane, which are components of the RM solutions, the RM solutions are seen to be more conductive and less resistive than HPLC H₂O and decane. The RM solutions are less conductive and more resistive than the 1.35 M NaCl. The significance of the conductivity and resistivity measurements will be discussed in further detail in the magnetic properties results section of this thesis.

4.3.3 The quality (Q) factor

The frequency at the peak of absorption (ν), the change in the frequency of the bandwidth of the peak ($\Delta\nu$), and the calculated Q factor of the STS imaging coil containing air, seven solutions, and distal and proximal anatomy of the human arm are listed in Table 26.

Table 26. Quality measurements of several solutions and human anatomy in an RF coil

Solution	ν (MHz)	$\Delta\nu$	Q factor
Air	63.82	0.12	531.8
Decane	63.56	0.20	317.8
[NaCl] = 0.0385 M	64.10	3.0	21.37
[NaCl] = 0.077 M	64.76	5.2	12.45
[NaCl] = 0.154 M	65.5	7.2	9.1
RM, $\phi = 0.5$	63.45	0.88	72.1
RM, $\phi = 0.7$	63.92	1.48	42.19
Arm, proximal	62.51	7.0	8.93
Arm, distal	64.31	3.4	18.91

The Q factor for decane is 317.8 and much higher than the Q factor of 9.1 for the 0.154 M NaCl solution. Since decane does not contain ions in solution and is electronically neutral, very little or no translational energy is used to make the ions in solution align with an electrical field. Consequently, little energy would dissipate in each RF cycle and the Q factor is higher than a solution containing ions. Phantoms containing NaCl can be used to mimic the loading of the coil by human tissue [Morris]. The three NaCl solutions studied had Q factors between 9.1 and 21.37. These values are within the range of the Q factor measurements for the proximal arm and distal arm, which are 8.93 and 18.91, respectively.

The Q factor for these solutions is decreased because translational energy is used to align the ions in an electric field. The energy dissipated per RF cycle is increased, thus, the Q factor decreases.

The Q factors for the RM solutions can be explained by the restricted Na^+ ions inside the RM, which are not able to easily align themselves in the electric field, and by the encumbered mobility of the negatively charged polar heads of the AOT surrounding the H_2O . Since the Na^+ and the polar heads of the AOT are oppositely charged, they compete with each other to align in the field, so there is a decrease in the translational energy. There is little or no movement of charges in the nonpolar decane or air; therefore, very little energy is lost, and the Q factor increases.

It was hoped that the RM solutions would produce a Q factor similar to human tissue, so the RM solution phantom would load the coil the same as the human body and the coil would not need to be retuned to a different resonant frequency. While the Q factors of the RM solutions and the proximal and distal arm did not match, we were able to qualitatively measure and compare Q factor measurements for the RM solutions and human tissue. Further studies to quantify the Q factor and develop an RM solution, which would load the coil in the same manner as the human body, could be done.

4.4.0 Magnetic properties of the RM solution

Described below are the results and discussion from the NMR, magnetic susceptibility, transmit and receive field mapping, fat-saturation imaging sequence and magnetization transfer studies.

4.4.1 NMR studies of the RM solution

Low resolution spectra were obtained for the reference solutions decane, AOT, and H_2O . Figures 28 and 29 display the spectra of the decane and H_2O , respectively. The AOT spectra was previously shown in Figure 20 in Section 3.5.1. The low resolution spectra of the RM solutions for the ϕ values ranging from 0.5 to 0.7 were similar to, and an example of an RM solution with a ϕ value of 0.6 is displayed in Figure 30. At ϕ values greater than 0.70, the RM solutions phase separated.

Figure 28. Low resolution NMR spectra of *n*-decane.

Figure 29. Low resolution NMR spectra of H₂O containing 0.25 mM Mn²⁺ in the aqueous phase.

Figure 30. Low resolution NMR spectra of a RM solution containing 0.25 mM Mn^{2+} in the aqueous phase and ϕ value of 0.6.

The RM solutions contained peaks similar to those of both the H_2O and decane. As mentioned in Section 3.5.1, the peaks from the AOT are buried in the H_2O and decane peaks. No correlation between the increasing ϕ values and peak height of the H_2O and decane could be made. Furthermore, calculating the amount of H_2O and decane in each of the phases of the phase separated RM solutions was unsuccessful, other than making note that the area under the peak of decane decreases for the lower phase regardless of whether ϕ increases.

4.4.2 Echo modulation studies

The effect of echo modulation from the decane, H_2O , and the decane in the RM solution containing 8 mM Ni^{2+} in the aqueous phase with a ϕ value of 0.6, is displayed in Figure 31. The signal from the spin-echo pulse sequence as a function of TE was measured. Comparisons to H_2O and decane with the RM solution will be discussed.

Figure 31. Signal obtained from a spin-echo pulse sequence for an RM solution containing 8 mM Ni²⁺ in the aqueous phase with ϕ value of 0.6, an aqueous 8 mM Ni²⁺ solution and decane as a function of varying TE.

The amplitude of the signal for TE values ranging from 0 to 150 ms for decane varied due to echo modulation, while the amplitude of the signal from the aqueous 8 mM Ni²⁺ was a pure exponential decay and did not exhibit echo modulation. Comparing the RM solution to both the decane and aqueous solution, the RM did not display a pure exponential decay as that of the aqueous solution, but did exhibit a variation in signal amplitude similar to decane. The variation in amplitude of the signal from the echo modulation for the RM solution was not as great as the decane. As ϕ increases, the amount of decane

present will decrease, which means the effects of echo modulation should not be as great at the higher ϕ values. The effects of echo modulation are minimal even at a ϕ of 0.6, and do not affect the image signal intensity as will be shown later in the transmit and receive field mapping results section of this thesis. The effects of echo modulation of the AOT were not studied since the AOT peaks were hidden in the decane and H₂O peaks and thought to be minimal.

4.4.3 Overall T_1 and T_2 values for the RM solutions containing Mn^{+2} and Ni^{+2} ions

Overall T_1 and T_2 values were measured at room temperature for the RM solutions containing either 0.1, 0.25, or 0.5 mM Mn^{+2} ; or 3, 5, or 8 mM Ni^{+2} . Results of the measurements are shown in Figures 32 and 33. The relaxation rates $1/T_1$ and $1/T_2$ were plotted as a function of ϕ .

Figure 32 a) $1/T_1$ in s⁻¹ as a function of ϕ for the RM solution containing 0.1 (Δ), 0.25 ($\blacksquare$), 0.5 ($\blacklozenge$) mM Mn^{+2} in the aqueous phase. b) $1/T_2$ in s⁻¹ as a function of ϕ for the RM solution containing 0.1, 0.25, 0.5 mM Mn^{+2} in the aqueous phase.

Figure 33. a) $1/T_1$ in s⁻¹ as a function of ϕ for the RM solutions containing 3 (Δ), 5 ($\blacksquare$), 8 ($\bullet$) mM Ni²⁺ in the aqueous phase. b) $1/T_2$ in s⁻¹ as a function of ϕ for the RM solutions containing 3, 5, 8 mM Ni²⁺ in the aqueous phase.

The overall T_1 's and T_2 's of RM solutions containing either the Mn²⁺ or Ni²⁺ in the aqueous phase shortens as ϕ increases. As ϕ is increased, the amount of the decane decreases and the amount of the paramagnetically-doped H₂O is increased; therefore shortening of the overall relaxation times is expected. Much less of the paramagnetic ion Mn²⁺ is needed than the Ni²⁺ ion to produce the shortened T_1 's and T_2 values. Ni²⁺ has a calculated magnetic moment of 2.83 while the calculated magnetic moment of Mn²⁺ is 5.92 [Huheey]. This means that the magnetic field generated by the Mn²⁺ ions is at least twice that of the Ni²⁺. Consequently, the total effect of the local fields from the Mn²⁺ ions will be greater than the Ni²⁺ ions as long as no additional ions are present, which would cause a shortening of the relaxation times.

4.4.4 T_1 and T_2 's from variable temperature studies

The T_1 and T_2 values of a solution used in an MRI phantom should be reproducible from week to week, as well as from one imaging site to another imaging site. Since there are temperature fluctuations when the phantoms are shipped to an imaging site and slight variations in temperature found at different imaging sites, the T_1 and T_2 values of a solution should be independent of temperature. Consequently, T_1 and T_2 values should be similar regardless of large or small temperature variations. Figures 34, 35, 36, and 37 are contour maps depicting the

effect of temperature on the RM solutions with ϕ values between 0.5 and 0.8 and containing three different concentrations Mn^{+2} or Ni^{+2} in the aqueous phase. The temperatures studied ranged from 0 to 40° C.

Figure 34. T_1 as a function of ϕ and temperature for a RM solution containing Mn^{2+} in the aqueous phase. The staircase line is the boundary between the RM phase and the lamellar and coexistence phase.

Figure 35. T_2 as a function of ϕ and temperature for a RM solution containing Mn^{2+} in the aqueous phase. The staircase line is the boundary between the RM phase and the lamellar and coexistence phase.

Figure 36. T_1 as a function of ϕ and temperature for a RM solution containing Ni^{2+} in the aqueous phase. The staircase line is the boundary between the RM phase and the lamellar and coexistence phase.

Figure 37. T_2 as a function of ϕ and temperature for a RM solution containing Ni^{2+} in the aqueous phase. The staircase line is the boundary between the RM phase and the lamellar and coexistence phase.

The RM solutions containing Ni^{+2} produced T_1 values that are less temperature dependent than the RM solutions containing Mn^{+2} . This was supported by the literature which states that the Ni^{+2} aquaion has a correlation time, which is determined by the ion's tumbling rate in solution, is dominated by a fast electron-spin relaxation [Kraft]. The temperature dependence of the Ni^{+2} aquaion is reduced because the reorientation of the aquaion is masked by the faster electron relaxation rate. The RM solution containing 8 mM Ni^{+2} in the aqueous phase is the least temperature dependent of all of the RM solutions, as evidenced by the nearly vertical lines in Figure 38, showing the T_1 values of the solution as the temperature is increased. The T_1 values of the 5 mM Ni^{+2} RM solution are also temperature independent but not to the same extent as those of the 8 mM Ni^{+2} solution. The RM solution containing 0.5 mM Mn^{+2} in the aqueous phase has T_1 values that are the least temperature dependent of the three RM solutions containing Mn^{+2} . As the concentration of the paramagnetic ion increases, the T_1 values of the RM solutions containing either Mn^{+2} or Ni^{+2} become less temperature dependent.

The same trend is seen with the T_2 values of the six RM solutions. The T_2 values of the 0.5 mM Mn^{+2} RM solution is the least temperature dependent of the Mn^{+2} RM solutions, while the 8 mM Ni^{+2} RM solution displays T_2 values that are constant at approximately 90 ms for the entire range of the ϕ values.

4.5 Reproducibility and stability of the RM solution

To test the effects of mixing and temperature on the T_1 and T_2 values of the RM solution, nine samples were treated either to sonication or temperature changes. The first three samples were the control samples and left undisturbed at room temperature. The 4th and 5th samples were sonicated for ten minutes and left at room temperature. Samples 6 and 7 were placed in an H_2O bath at 5° C for 24 hours. The last two samples, numbered 8 and 9, were placed in an H_2O bath at 30° C for 24 hours. After equilibrating each of the samples for 24 hours at room temperature, the T_1 and T_2 values of the samples were determined. The overall T_1 and T_2 values for each of the nine solutions are listed in Table 27.

Table 27. Overall T_1 and T_2 stability after sonication or temperature changes

Sample	Procedure	T_1 (ms)	T_2 (ms)
1	control	572.2	53.0
2	control	572.0	52.6
3	control	581.6	54.9
4	sonicated	567.0	53.0
5	sonicated	564.0	52.9
6	$T = 5^\circ$	553.0	53.0
7	$T = 5^\circ$	552.3	53.3
8	$T = 30^\circ$	571.1	54.5
9	$T = 30^\circ$	548.8	52.7

There are slight variations in the T_1 C of each of the samples due to inconsistencies from sample to sample in making the samples. The mean, coefficient of variation (CV), and 95% confidence interval (CI) of the T_1 and T_2 values for the nine solutions are listed in Table 28.

Table 28. Mean, CV, and CI for T_1 and T_2 reproducibility

	T_1	T_2
Mean (ms)	564.7	53.3
CV (ms)	1.96	1.56
CI (ms)	564.7 $\pm$ 9.7	53.3 $\pm$ 0.7

The average of the T_1 of all the samples is 564.7 ms. The CV for the T_1 values of the nine samples is 1.96. Most of the T_1 measurements are within the range of 554.9 and 574.4 ms. The average T_2 for all of the samples is 53.3 ms. The CV for the T_2 values of the samples is 1.56 ms. Most of the samples T_2 values are within the range of 52.59 and 54.1 ms. Since most of the T_1 and T_2 values were within the range for the 95% confidence limits, mixing the solution and varying the temperature of the solution had very little or no effect on the T_1 and T_2 values of the RM solution.

4.4.6 Susceptibility studies

A simple experiment was designed to examine the permeability of the RM solution. Insights into these properties were studied by determining the amount of distortion of an image from the magnetic susceptibility of a sample placed adjacent to the solution being studied. Distortion of the image of a small bottle containing one of the six solutions listed in Table 15 of the experimental section due to a large culture bottle containing either decane

or aqueous 8 mM Ni^{+2} was determined. The process of determining the amount of distortion by counting variations in the grey scale of the pixels, was not possible because variations were on the order of one pixel. This result was expected since the permeability of most liquids is the same.

4.4.7 Transmit and receive B_1 field mapping

One component of system performance testing of the MR imager is transmit and receive B_1 field mapping. The B_1 transmit and receive fields of an imaging coil are tested to ensure B_1 field homogeneity of the coil. It is important for the B_1 of the transmit and receive fields of a coil to be homogeneous so that variations in the intensity of the image are due to the imaged object rather than the coil itself. Because a larger imaging volume is desired to approximate the volume of a human body, a large phantom with a diameter of 27 cm is used for transmit and receive field mapping. For this study, three large phantoms - one containing aqueous 14 mM Ni^{+2} , another containing aqueous 6 mM Ni^{+2} and 154 mM NaCl, and a third containing RM solution with a $\phi = 0.6$ and 8 mM Ni^{+2} in the aqueous phase were used. These solutions are currently used for system performance testing, have the same NaCl concentration as the RM solution with $\phi = 0.6$, and had phase-stable, temperature-independent T_1 values in the desired range, respectively. Figure 38(a) and (b) displays the B_1 transmit and receive field map of the three phantom solutions. The B_{1T} and B_{1R} intensities across the diameter of the image from left to right were measured. The B_{1T} and B_{1R} were normalized to one in the center of the sphere. Table 29 lists the CV for the B_{1T} and B_{1R} fields.

Table 29. Coefficient of variation for B_{1T} and B_{1R} fields inside a 27 cm diameter phantom

Phantom solution	CV (B_{1T})	CV(B_{1R})
14 mM Ni^{+2}	0.132	0.225
6 mM Ni^{+2} and 154 mM NaCl	0.133	0.206
RM, $\phi = 0.6$ (8 mM Ni^{+2})	0.057	0.176

Figure 38. Normalized (a) transmit. and (b) receive RF magnetic fields from the GE low pass quadrature body coil. The fields were measured in a 27 cm diameter sphere containing (1) aqueous 6 mM Ni^{2+} with 154 mM NaCl, (2) RM with a $\phi = 0.6$ (8 mM Ni^{2+}) in the aqueous phase, and (3) aqueous 14 mM Ni^{2+}

The aqueous 14 mM Ni^{+2} phantom solution displayed a negative parabolic B_{1T} and B_{1R} profile. An increase in the ionic strength of the solution caused an increase at the internal surface of the sphere as shown by the aqueous 6 mM Ni^{+2} and 154 mM NaCl phantom solution B_{1T} and B_{1R} profile. The RM phantom solution B_{1T} and B_{1R} profile showed less of the negative parabolic tendency than the aqueous 14 mM Ni^{+2} solution phantom and no increase at the edges of the B_{1T} and B_{1R} . The CV for the B_{1T} and B_{1R} of the RM phantom solution was less than the CV of either the 14 mM Ni^{+2} phantom solution or the 6 mM Ni^{+2} and 154 mM NaCl phantom solution. Consequently, the RM phantom solution had a more uniform overall set of B_{1T} and B_{1R} fields. The slight decrease on the right of the B_{1R} seen in all three phantom B_{1T} and B_{1R} profiles is attributed to a property of the RF coil.

The negative parabolic of the B_{1T} and B_{1R} of the aqueous 14 mM Ni^{+2} phantom solution was caused by the standing wave artifact. Recall the wavelength in air at 1.5 T is approximately 4.7 m [Bottomley]. Once a wavelength passes through the aqueous solution inside the phantom, the wavelength is reduced to approximately 0.52 m. One-half of this wavelength is the diameter of the sphere of the phantom and a standing wave is formed inside the phantom. The RM solution displays less of the negative parabolic tendency of the B_{1T} and B_{1R} . This is attributed to the lower dielectric constant of the RM solution [Tofts]. The wavelength is reduced to 0.78 m when it passes through the RM phantom solution. One-half of this wavelength is approximately 0.39 m which is more than the diameter of the sphere of the phantom. Therefore, the standing wave does not completely form inside the phantom and the effects of the standing wave are minimized.

The parabolic tendency of the 6 mM Ni^{+2} and 154 mM NaCl phantom solution is called the *skin-effect* artifact. The B_{1T} and B_{1R} fields are not able to penetrate to the center of the phantom. The B_1 fields decrease by a factor of $e^{-X/\delta}$, where X is the distance from the surface in the phantom and δ is the skin-depth of the solution, as the B_1 field travels through the phantom. The skin-depth for the aqueous 6 mM Ni^{+2} and 154 mM NaCl is 6 cm compared to the skin depth of the RM solution of approximately 15 cm. The B_1 field of the aqueous 6 mM Ni^{+2} and 154 mM NaCl solution will decrease to a greater extent as it travels through the phantom solution than the RM solution phantom. The *skin-effect* artifact is not seen in the RM solution phantom because the effects from the *skin-effect* artifact are less significant than the standing wave artifact. Consequently, the standing wave artifact will dominate [Tofts].

4.4.8 Fat-saturation studies

Comparison of the NMR spectra of H_2O , decane, and RM solution with ϕ of 0.6, shows that the RM solution contains two spectral peaks that correspond to the spectral peaks of pure H_2O and decane. The H_2O and decane spectral peaks occur at frequencies similar to those of H_2O and fat found in the human body as described earlier in the background section of this thesis. Because the two peaks of the RM solution coincide with the peaks of the H_2O and fat of the human body, the RM solution could be used to test the fat-saturation imaging sequence. Figure 39 contains the signal from solutions of aqueous 8 mM Ni^{++} , decane, and an RM solution with $\phi = 0.6$ and 8 mM Ni^{++} in the aqueous phase as a function of the offset frequency of the saturation pulse in a GE Chemsat spin-echo pulse sequence.

Figure 39. Signal from solutions of aqueous 8 mM Ni²⁺, decane, and an RM solution with $\phi = 0.6$ and 8 mM Ni²⁺ in the aqueous phase as a function of offset frequency of the saturation pulse from the water resonance in a GE Chemsat spin-echo pulse sequence.

The signal from the aqueous 8 mM Ni^{+2} solution in Figure 38 showed a decrease from 0 to 150 Hz due to the saturation of the H_2O signal. The signal from the decane showed a decrease from 150 to 250 Hz due to the saturation of the fat/decane signal. Neither of the signals from the H_2O or decane were completely suppressed by the saturation pulse because the time between the saturation pulse and 90° RF pulse allowed some of the spins to return to equilibrium, thus producing signal.

When the H_2O signal is suppressed in the RM solution, a decrease in signal is seen between 0 to 150 Hz, which is similar to the response of the aqueous 8 mM Ni^{+2} suppressed signal. Application of the fat saturation signal on the RM solutions shows a decrease in signal similar to that of decane, that is, between 150 and 250 Hz. The signals from the RM solution were not completely suppressed to the extent of either the H_2O or decane due to the incomplete suppression of spins described above. The signals may not be completely suppressed because the AOT may provide a pathway for the transfer of magnetization between the AOT and decane.

4.4.9 T_1 values of the individual components

The spectral T_1 values of the individual components were measured at room temperature to provide information about both the molecular dynamics of the three component system and the overall T_1 values of the system. Table 30 contains the individual spectral T_1 values of the H_2O and decane components for the 0, 3, 5, and 8 mM Ni^{+2} RM solutions with ϕ values from 0.5 to 0.7. We were unable to determine the spectral T_1 values of the AOT since the AOT peaks could not be differentiated from the H_2O or decane peaks.

Table 30. Spectral T_1 values in ms of the H_2O and decane components of the RM solutions containing Ni^{+2}

	H ₂ O component					decane component				
	ϕ					ϕ				
Ni^{+2} (mM)	0.50	0.55	0.60	0.65	0.70	0.50	0.55	0.60	0.65	0.70
0	2575	2714	2512	2648	2574	1979	2070	1798	1683	1529
3			420					1830		
5			380					1805		
8	800	355	280	235	255	2060	1865	1820	1775	1595

As the concentration of Ni^{+2} is increased, the T_1 of the H_2O component decreases. This trend is expected since the Ni^{+2} in the aqueous phase has increased, so that the paramagnetic effects of the Ni^{+2} increases and

shortens the T_1 . There is a decrease in T_1 as ϕ increases. The mechanism for this trend is not well understood. As ϕ increases, the RM become more tightly packed and may form clusters. The formation of the clusters may change the diameter of the RM, allowing the H_2O to move about more freely in solution. Consequently, more molecular motions would shorten the T_1 . This data is supported by the diffusion data at room temperature for the H_2O and decane. There is an increase in mobility of the H_2O as evidenced by both the shortening of the individual T_1 and by the increase in the diffusion coefficient as ϕ increases.

The individual T_1 values of decane also decrease as ϕ increases, yet they are independent of Ni^{+2} concentration. The Ni^{+2} is contained inside the aqueous phase of the RM, therefore leaving the decane unaffected. As ϕ increases, the amount of decane present is decreased and the RMs become more tightly packed. Consequently, the mobility of the decane is restricted by the RMs in the solution, especially the AOT tails, and the T_1 is shortened. A decrease in the mobility is supported by the diffusion studies at room temperature which shows a smaller diffusion coefficient of the decane as ϕ is increased.

Plots for T_1 data were made to determine the spin exchange (k) between the components. These plots were based on the assumption that there is no exchange between the decane and H_2O components. It was not possible to solve for k since the number of variables present in the equations was too great. The calculation of the exchange rate was left for future studies.

Since the individual T_1 values of the H_2O and decane are not the same as the overall T_1 value, it is assumed that the spin exchange rate between the components is small. Therefore, the overall T_1 is representative of a monexponential best fit of multiexponential data.

The presence of a small spin exchange value opens up the possibility for magnetization transfer studies to be done using the RM solution. This would further characterize the behavior of the RM solution on a molecular level, as well as allow the RM solution to be used to test magnetization transfer imaging sequences on the MR imager.

4.4.10 Summary of spectral T_1 , molecular motions, and viscosity

It is possible to comment on the molecular dynamics as a function of frequency model using the diffusion coefficients, predicted viscosity, and spectral T_1 of the H_2O component of the RM solution. Table 31 contain the

diffusion coefficient, predicted viscosity, measured T_1 and predicted molecule motions of a RM solution with ϕ values of 0.50 and 0.70 and pure H_2O .

Table 31. Physical and NMR parameters for H_2O component of RM solutions

Solution	Measured Diff. Coeff	Predicted η	Measured T_1 (s)	Molecular motions $J(\nu)$
neat	<	low	0.19	>
$\phi = 0.7$	intermediate	intermediate	0.26	intermediate
$\phi = 0.5$	>	high	0.8	<

A model of molecular motions as a function of frequency can be designed from the information in Table 31. Figure 40 displays the model. Three $J(\nu)$ curves are predicted based on the perceived viscosities of the water. In a specific frequency band in the figure, the molecular motions of each solution must match that predicted by the T_1 values. The neat H_2O has a larger diffusion coefficient and lower predicted viscosity, therefore, the molecular motions will be greater and the T_1 should be shorter. This is what the results showed. The RM solution with a $\phi = 0.5$ has a smaller diffusion coefficient and higher predicted viscosity. The molecular motions will be less than than of pure H_2O and the T_1 will be longer. The RM with a $\phi = 0.7$ is in between these two. The operating frequency of the imager (63 Mhz) must therefore be in the range delineated by the rectangular area in Figure 40.

Figure 40. Molecular motions as a function of frequency model for the H_2O component of RM solution with $\phi = 0.5$ and $\phi = 0.7$ and pure H_2O . The approximate molecular motions can be predicted at specific frequencies for each solution. The rectangular box is the operating frequency band of the MR imager of ≈ 63 MHz.

5.0 Conclusion

In review, some of the properties of an ideal phantom solution include stability and phantom-to-phantom consistency. The phantom solution should produce an optimal signal with a short scan time, T_1 and T_2 values which mimic human tissue, and T_1 and T_2 values that are adjustable. The dielectric constant of the phantom solution should be less than the dielectric constant of H_2O so that the standing wave artifact is no longer present, or minimized. The resistivity of the solution which is increased by the addition of $NaCl$ to the solution should be high so the skin-effect artifact is not seen when the phantom is imaged. The solution should have a high viscosity to decrease the possibility of a convection current artifact forming.

The RM solution possesses all of the above mentioned properties of an ideal phantom solution. The RM solution is phase stable over a wide range of temperatures. It is easy to make and can be subjected to various changes in temperature and mixing conditions and consistently produces the same T_1 and T_2 values. The T_1 and T_2 values can be customized. This would allow the T_1 and T_2 to exhibit those values found in human tissue, produce an optimal signal and short scan time, or any desired value. This customization is done by the addition of the paramagnetic salt $MnCl_2$ or $NiCl_2$. The RM solution has a dielectric constant half that of H_2O . This result causes the standing wave artifact found in the current phantoms used during system performance testing to be minimized. The resistivity of the $\phi = 0.6$ RM solution containing 0.25 mM Mn^{2+} in the aqueous phase is larger than the aqueous solutions of paramagnetic ions currently used for system performance testing. Consequently, the *skin-effect* artifact is no longer present. The viscosity of the RM solution is similar to that of 30W motor oil. Therefore, the potential problem of a convection current artifact forming in the solution inside the phantom, is reduced.

Other properties found in the RM solution make it useful for a multipurpose phantom solution. Therefore, one phantom could be used to test the performance of the imaging system rather than many different phantoms. Because the RM solution consists of two NMR spectral peaks centered at 0 and 220 Hz, which correspond to the fat (220 Hz) and H_2O (0 Hz) signal in the human body, the RM solution phantom could be used to test the fat-saturation imaging sequence of the imager. Additional

characterization of the spectral T_1 and T_2 values may lead to applying the RM solution phantom to be used to test the magnetization transfer imaging sequence of the imager.

Other studies, such as, diffusion studies, gave us some understanding of how the H_2O , decane and AOT components of the RM solution functioned on a molecular level. The diffusion studies showed that as ϕ increased, the diffusion coefficient of the H_2O component increased. As ϕ increases, the RMs will cluster or aggregate in solution. This clustering allows the H_2O molecules to exchange more readily or move about more freely, consequently the diffusion coefficient is increased. The diffusion coefficient of the decane component decreased as ϕ increased, thus the mobility of the decane in the solution decreased because the reverse micelles were more tightly packed in the solution. The decane may possibly be trapped between the RM. This decreased mobility of the decane explains why the diffusion decreases with ϕ . The spectral T_1 values of the H_2O and decane helped confirm the above behavior mechanisms. As ϕ increased, the spectral T_1 of H_2O decreased and the spectral T_1 of decane decreased. These results are explained by the same mechanism as the diffusion coefficient results. Both the diffusion coefficients and spectral T_1 results increase our understanding of the behavior of the RM solution as it reaches the percolation threshold.

As with all research, more questions arose than those we originally thought to answer. Proposed work to develop or *fine-tune* the RM solution for use as an imaging phantom in system performance testing as well as studies to characterize the behavior of the RM solution include the following studies. The size of the RM itself could be varied by changing the molar ratio of AOT to H_2O to determine if the dielectric constant would decrease. Decreasing the dielectric constant should eliminate the standing wave artifact. The optimal concentration of Ni^{+2} and Mn^{+2} , which would be independent of temperature could be found. NMR and electrical studies may be performed using of a solvent other than decane. The effects of echo modulation could be increased or decreased, depending on the properties needed in a solution. The handling of a hazardous solution could be minimized, consequently making the shipping of the solution easier, and the evaporation of the solvent from the phantom container could be minimized. The amount of NaCl in the solution could be varied to produce a Q factor similar to human anatomy in

the coil. Diffusion coefficients could be studied in more depth by examining temperatures higher and lower than room temperature, to more closely characterize the behavior of each of the components in the RM solution. To prove the applicability of the RM solution phantom to fat-saturation imaging sequence testing, a software program could be developed to test the performance of the fat-saturation imaging sequence. Finally, an in-depth study of the spectral T_1 values could be done to determine the exchange rate between the components in the solution, which would further define the behavior of the RM solution and study the effects of magnetization transfer between the components of the solution.

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