EXPERIMENTAL COMPARISON OF ACR AND ICAMRL MAGNETIC RESONANCE IMAGING ACCREDITATION PROTOCOLS

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

By

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B.S., Ohio Northern University, 2009

2010
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Two primary accrediting bodies exist for magnetic resonance imaging systems: the American College of Radiology (ACR) and the Intersocietal Commission for the Accreditation of Magnetic Resonance Laboratories (ICAMRL), each of which defines specific standards for specific image quality criteria at which MRI images must be produced. An MRI clinic that wishes to show a commitment to image quality may do so by becoming accredited by one of these organizations of their choosing. The limits of these image criteria were compared to demonstrate the standards of each accrediting body. Images were produced that fell well within the standards of both accrediting organization, and subsequent images were produced at the limits of ACR and ICAMRL standards respectively. These images were first produced using a phantom to quantify a difference in criteria standards, then images were produced using a human patient to show a qualitative difference in criteria standards for clinical applications.
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I would like to acknowledge and thank Dr. Jason Parker for all of his help and guidance, as well as allowing me to perform scans at the Kettering Innovation Center to complete my research. His supervision throughout the project was extremely helpful and allowed for me to complete my venture in the amount of time I did, and with a much greater understanding for how MRI worked. I would also like to thank him for being a part of my committee and evaluating the work I put in.

I would also like to acknowledge Eric Zalusky for his help performing scans, and guiding me through the intricacies of the MRI interface that was used.

Finally, I would like to thank my other committee members, Dr. Brent Foy, who was also my thesis advisor, and Dr. Gary Farlow. They both helped me greatly with the construction and revision of my paper, as well as with the understanding the underlying concepts in my project.
I: INTRODUCTION

In clinical magnetic resonance imaging, image quality is important to help ensure proper diagnosis of patients. For this reason, different organizations have begun to construct sets of procedures that allow the images produced with an MRI system to be evaluated. These organizations decide what aspects of an image need to be evaluated to ensure good image quality, and create limits for what the acceptance criteria for these aspects of the image should be. These sets of procedures and acceptance criteria that are constructed can be performed on any MRI system, and if the aspects being tested fall within the acceptance ranges of the organization, they grant you with an official approval that shows you have passed their protocol and your commitment to image quality. This process is known as accreditation.

Clinics that wish to show a dedication to image quality, and therefore wish to become accredited, can choose between several organizations that offer accreditation services. The two main organizations that provide accreditation services are the American College of Radiology (ACR) and the Intersocietal Commission for the Accreditation of Magnetic Resonance Laboratories (ICAMRL). It is at the discretion of the medical physicist and/or the lead MR scientist at the clinic which organization they wish to be accredited through. Since the acceptance criteria for each accrediting organization are different, it is possible to produce images that may pass one protocol, without passing the other protocol. It is therefore the objective of this paper to produce
images that represent the limits of standards for each accreditation protocol and compare the results.

The following is a comparison of images produced at the limits of the acceptance criteria for both ACR and ICAMRL accreditation protocols. Specific criteria that exist in both accreditation procedures will be evaluated to quantify the differences in the standards of each organization. The criteria will also be evaluated in patient scans to determine the effects the acceptable range of these limits may have on patient diagnosis. The criteria chosen to be tested were; signal-to-noise ratio, spatial resolution, and slice thickness accuracy.

These individual criteria will be compared on an individual basis to give a comparison of the limits of each accrediting body for one acceptance criteria at a time. It should be noted that in a typical system, it is possible for multiple criteria to be outside of an acceptable range of a more stringent criteria and degrade the image through multiple criteria variations. However, for this study, a case by case comparison was done for each acceptance criteria to demonstrate the effects that each of the ranges for individual criteria would affect the quality of the image produced.
A charge $q$ moving at a velocity $v$ in a circular path with a radius $r$, will have a period of $T = \frac{2\pi r}{v}$. The charge flowing past any point of the circle per second, known as the current $I$, is given as $q/T$, resulting in:

$$I = \frac{qv}{2\pi r}$$

For the circular path area $A = \pi r^2$, the magnetic moment of $\mu = IA$ is given as:

$$\mu = \frac{qvr}{2}$$

The angular momentum of the charge is $L = mvr$, which gives us a magnetic moment of:

$$\mu = \frac{qL}{2m}$$

proportional to angular momentum. This is also written as,

$$\mu = \gamma L$$

where $\gamma = q/2m$ is the gyromagnetic ratio, classically.

Experimentally, a proton has been found to possess an intrinsic, internal magnetic moment, $\mu$. By analogy to classical mechanics, we might also expect the magnetic moment for a proton in quantum mechanics, $\mu_p$, to be proportional to an intrinsic “spin” angular momentum, $S$. We can then write

$$\mu_p = \gamma S$$
and we consider
\[ \gamma = \frac{ge}{2m} \]
where \(e\) is the fundamental unit of charge for a proton \((1.602 \times 10^{-19}\text{C})\), \(m\) is the proton mass \((1.67 \times 10^{-27}\text{kg})\), and \(g\) is a factor allowing for the fact that the \(\gamma\) in quantum mechanics may not correspond to classical expectations. Experimentally, we obtain \(g = 5.6\), which results in
\[ \frac{\gamma}{2\pi} = 42.69\text{ MHz/Tesla} \]
for a proton.

Spins at equilibrium will reside in one of the spin states \((S_z = S_{max}, S_{max}-1, \ldots, -|S_{max}|)\). When a particles of spin \(S\) is placed in an external magnetic field, the spins will try to align themselves with the field, in a parallel or an anti-parallel fashion. Since the proton has \(S = \frac{1}{2}\), there are only two states available, \(S_z = \pm \frac{1}{2}\), representing the orientation parallel (+) or antiparallel (-) to the magnetic field. The orientation results in two separate energy states, labeled \(\pm \frac{1}{2}\) respectively, with the parallel alignment being the lower energy state. As the strength of the magnetic field, \(B_0\), is increased, the energy, \(\Delta E\), needed to cause a transition from one state to the other will increase (Figure 1). The energy gap is linearly dependent on the type of nucleus and the field strength as follows:
\[ \Delta E = \frac{h}{2\pi} \omega_0 = \frac{h}{2\pi} \gamma B_0 \]

The protons will occupy either the ground state, with spin up and number density \(N_{+1/2}\), or the excited state, with spin down and number density, \(N_{-1/2}\). At equilibrium, the number of spins in a given state is given by the Boltzmann distribution, which has a
dependence on the energy gap and the absolute temperature, T in the following fashion:

\[ N_{-1/2} = N_{+1/2} e^{-\Delta E/kT} \]

Figure 1: As \( B_0 \) strength increases, the \( \Delta E \) between spin-up and spin-down states increases

Where \( k \) is the Boltzmann constant (1.381 × 10\(^{-23}\) J/K). The vector sum of the spin magnetic moments in the excited ground states yields the bulk magnetization vector, \( M_0 \) (Figure 2). For room temperature (approximately 295K) in a 1 Tesla field, \( kT \) is then 4.1 × 10\(^{-21}\) Joules, and \( \Delta E \) is then 2.8 × 10\(^{-26}\) Joules. This results in a net polarization, \( N_{+1/2} / N_{-1/2} \), of the ground state of approximately 1.0000075. The low value for net polarization shows that nuclear magnetic resonance is an insensitive technique, but the large \( \gamma \) of the proton (2.68 × 10\(^8\) rad/(sec · T)), as well as the vast number of hydrogen atoms in the body tissue help make even this small polarization an easily measurable quantity.
Classically, we know that a magnetic moment, $\mu$, in a static magnetic field, $B_0$, experiences a torque, $\tau$, such that:

$$\tau = \mu \times B$$

Since torque is the time rate change of the angular momentum, we can arrive at,

$$\frac{dS}{dt} = \frac{l}{\gamma} \frac{d\mu}{dt} = \mu \times B$$

Therefore, using the fact that bulk magnetization of the sample, $M_0$, is the sum of the individual magnetic moments of the protons gives

$$\frac{dM_0}{dt} = \gamma (M_0 \times B_0) = \omega_0 \times M_0$$

for $\omega_0 = -\gamma \cdot B_0$ ($B_0$ is typically chosen to be along the positive z-direction by convention). This equation is recognized as the equation of motion for $M_0$ precessing about the z-axis in Figure 3 at the Larmor frequency $\omega_0$. In component form we can arrive at

$$\frac{dM_x}{dt} = \omega_y M_z - \omega_z M_y = \omega_0 M_y$$
\[ \frac{dM_y}{dt} = \omega_z M_x - \omega_x M_z = -\omega_0 M_x \]
\[ \frac{dM_z}{dt} = 0 \]
for \( \omega_0 = (0,0,-\omega_0) \). Therefore, \( M_z \) is a constant of the motion, which is equal to \( M_0 \cos \theta \), and

\[ \frac{d^2 M_x}{dt^2} = \omega_0 \frac{dM_y}{dt} = -\omega_0^2 M_x \]
with a general solution \( A \cdot \cos(\omega_0 t) + B \cdot \sin(\omega_0 t) \). For initial the initial condition

\[ M_x(0) = M_0 \cdot \sin \theta, M_y(0) = 0 \]
we obtain

\[ M_x(t) = (M_0 \sin \theta) \cos \omega_0 t \]
\[ M_y(t) = -(M_0 \sin \theta) \sin \omega_0 t \]
which is a clockwise precession in the transverse plane at the angular frequency, \( \omega_0 \), as shown in the figure on the right side of Figure 3. In words, a magnetic moment in an outside magnetic field precesses about the instantaneous direction of the magnetic field.
Figure 3: In a static magnetic field $B_0$ along the positive z-axis, a net magnetic moment, $M_0$, will precess about the z-axis with angular frequency $\omega_0$.

If in addition to the static field, $B_0$, a circularly polarized radio-frequency (RF) field, $B_1$, given by $B_1 = B_1(\cos \omega_0 t \cdot x - \sin \omega_0 t \cdot y)$ rotating at the Larmor frequency in the transverse plane, can be applied to the sample. The net effective field $B_e$, is

$$B_e = B_0 + B_1$$

and is thus offset from the z-axis. $M_0$ precesses about the instantaneous direction of $B_e$.

If we consider a rotating frame of reference in which $B_1$ will be a constant in the rotating frame, we know that

$$\left(\frac{dM_0}{dt}\right)_{lab} = \left(\frac{dM_0}{dt}\right)_{rot} + \omega_0 \times M_0 = \left(\frac{dM_0}{dt}\right)_{rot} + \gamma M_0 \times B_0$$

Therefore,

$$\left(\frac{dM_0}{dt}\right)_{rot} = \gamma M_0 \times B_0 - \left(\frac{dM_0}{dt}\right)_{lab} = \gamma M_0 \times B_0 - \gamma M_0 \times B_e$$

$$= \gamma M_0 \times B_0 - \gamma M_0 \times (B_0 + B_1) = \gamma M_0 \times B_1$$

and $M_0$ precesses in the rotating frame about the fixed direction of $B_1$ at an angular frequency of $\omega_0 = -\gamma B_1$. The strength and duration of $B_1$ will determine the extent that
the vector is rotated about $B_1$. If $\omega_1 t = \pi/2$, the magnetization will be tipped into the transverse plane by this $90^\circ$, or $\pi/2$ $B_1$ pulse.

The transverse plane is the same in both the lab and the rotating frame of reference, with only differing x- and y-axes. Once the magnetization is in the transverse plane, $B_1$ is turned off, and the vector precesses about $B_0$ at the Larmor frequency. This precessional magnetic field may be detected by placing a coil of wire perpendicular to the transverse plane. The oscillation of the magnetic field created by the precession of $M_0$ will produce a time dependent flux in the current loop, which by Lenz’s law will induce a current in the wire. This induced current may be detected as the NMR signal with the amplitude proportional to the magnitude of the net polarization, and therefore the total number of spins in the sample. The signal will decrease in amplitude with time as the $z$-magnetization returns to the equilibrium state it was in before the pulse, and there is a loss of magnetization in the transverse plane. This sinusoidal signal is referred to as the free induction decay (FID). The mechanisms for the return of the spins to the equilibrium state are described in the following, and we will then reconsider the oscillating FID afterwards.

At room temperature (295K), the quantity $kT$ is approximately equal to the energy gap, $\Delta E$, so there is sufficient energy for the relatively small number of polarized spins, which are now orthogonal to $B_0$, to make transitions back to the ground state due to the thermal motion and collisions of the protons. Their energy is lost to their surroundings and the $z$-magnetization returns to its equilibrium state. The amount of time it takes for the magnetization to return to about two thirds of its original value along the $z$-axis is known as the “spin-lattice”, or the longitudinal relaxation rate, $T_1$. 


The local field experienced by a proton determines its precessional frequency (resulting from the Larmor equation). Through molecular motions, any particular spin will randomly encounter other spins, and also the spin magnetic moments associated with those other spins, and therefore the local field for the particular spin will change. These local differences in the field cause a diffusion of rotational frequencies and a loss of phase coherence in the transverse plane. $T_2$ is the spin-spin or transverse relaxation rate, and is the time it takes for the transverse magnetization to drop to 37% of its initial value after $B_1$ is removed.

The dephasing of spins in the transverse plane can occur in a matter of milliseconds due to both the $T_2$ effects and inhomogeneities in the polarizing magnetic field, $B_0$. Imaging methods require the magnetic field gradients to be switched on and off (which will be discussed later in this section) before signal acquisition, and the signal can decay significantly during this period of switching on and off. One common imaging sequence is to first apply a 90° pulse and wait for a time, $\tau$, and then acquire the FID. This “spin-echo” sequence provides sufficient time for application of the switched magnetic fields during the period of $2\tau$ and still allows for signal acquisition at near maximum amplitude (Figure 4).
Figure 4: After a 90° pulse, all the magnetization is in the transverse plane. After waiting a time, $\tau$, the spins begin to dephase. Application of a 180° pulse flips the spins around the x-axis and though travelling in the same direction, the spins are now travelling towards phase coherence. After another $\tau$, the spins are in phase and the FID will be a maximum.

Two separate coils perpendicular to the transverse plane detect the FID; one for the plane of the coil in the x-z plane, and one for the plane of the coil in the y-z plane. The two channels of the spectrometer detect the real and imaginary portions of the signal. The real portion of the signal is represented in the x-channel as

$$ M_x = M_0 \cdot \cos \omega t $$

And the imaginary portion of the signal is represented in the y-channel as

$$ M_y = M_0 \cdot \sin \omega t \quad (\text{Figure 5}) $$
Figure 5: As $M_0$ precesses in the transverse plane at $\omega$, the signal in both the x and y-channels are components of $M_0$

Combining the real and imaginary portions of the signal gives the time dependent signal for the proton as:

$$f(t) = M_0 \cdot e^{i\omega t}$$

A proton that is part of a water molecule is attached to another hydrogen as well as being attached to an oxygen atom. A proton that is part of a fat molecule is attached to other hydrogens and a chain of carbon atoms. The local magnetic field felt by these two protons are inherently different. The signal received in the coil is a superposition of all the separate frequencies within the sample, given as

$$f(t) = \sum M_i e^{i\omega_i t}$$

Taking the Fourier transform of the time domain signal yields the frequency domain signal, which consists of peaks at the frequencies, $\omega_i$, with amplitude proportional to $M_i$. The frequency spectrum this represents how many different frequencies (protons in different molecules) exist within the sample, their shift with respect to the rotating frame of reference frequency, and their amplitude. The separation between any two frequencies
is referred to as the chemical shift.

If a spectrum of a water sample is taken in a homogenous $B_0$ field, all the water protons will precess at the same frequency and produce a single peak in the frequency domain. To transform spectroscopy into imaging, the frequency of the protons is varied in a known manner as a function of position in the sample by applying linear magnetic field gradients $G_x$ and $G_y$ (T/m) along the $x$- and $y$-axes. The respective magnetic field gradients add to the $B_0$, or z-axis, field as a linear function of $x$ or $y$, and vary symmetrically between the negative and positive increments to the $B_0$ field. Thus, with a gradient $G_x$ existing, the precessional frequency of a proton at a point, $x$, is $\gamma (B_0 + G_x x)$, which relates frequency to position within the sample.

To illustrate this, we will imagine a cylinder of water as shown in Figure 6a. In the absence of a magnetic gradient, a spectrum is achieved with one peak at a rotating frame frequency, as illustrated in Figure 6b. Now, if you consider applying only an $x$-gradient, $G_x$ (Figure 6c), with the field increasing linearly along the $x$-axis. At a given $x$-position, $x_0$, the field is the same for all points along the $y$-axis. Then, ignoring the constant frequency common to all of the points, $\gamma B_0$, the signal from each point $(x_0, y)$, where $x_0$ is fixed and $y$ ranges over the sample, is given as

$$f(t) = M_0(x_0, y) \cdot e^{i \omega(x_0)t}$$

where the precessional frequency $\omega$ is a function of the $x$-position in the sample, given by the relation

$$\omega(x) = \gamma G_x x.$$

The signal for the entire cylinder within the sample is then the sum of the individual signals at each point $(x,y)$, or:
\[ f(t) = \int \int M_0(x, y) e^{i\omega(x)t} dx dy \]

which gives the projection of the cylinder on the x-axis, as shown in Figure 6d. Defining a k-space through the relation \( k_x(t) = \gamma G_x t \), the signal may be written as

\[
\int \int_{-\infty}^{\infty} M_0(x, y) e^{i k_x(t)x} dx dy = F(k_x, k_y = 0)
\]

for a single FID with no \( G_y \) gradient. By then sampling \( k_y \) space with additional FID’s acquired after applying a \( y \)-gradient which is varied between FID’s, a 2-D function \( F(k_x, k_y) \) may be constructed such that the FT of \( F(k_x, k_y) \) yields our image, \( M_0(x, y) \). Just as the \( x \)-gradient encodes the frequencies of the spins during the acquisition, an additional \( y \)-gradient, \( G_y \), encodes the precessional phase of the spins during the time \( G_y \) is present.
Figure 6: a) Water cylinder with longitudinal axis along x direction. b) Frequency spectrum with x-gradient applied. c) Gradient that’s applied along x-axis. Gradient starts as a negative value and ends as a positive value. d) Frequency profile of water cylinder with the gradient applied. Amplitude represents the number of spins in each frequency-encoded region. Frequency shows the spatial localization of spins.

The y-gradient is applied prior to each signal acquisition as shown in Figure 7. As $k_y(T) = \gamma G_y T$, each increment in time yields a different position in k-space with $T=0$ equal to $k_y = 0$. Before each FID, if $T$ is varied for each acquisition, k-space will be mapped. More practically, $T$ can be kept fixed and $G_y$ incremented during each sequence prior to acquisition of the FID. This provides us with the 2-D map
Up to this point, the resulting 2-D image in the x-y plane contains signals from all the individual spins along the z-axis. In order to limit the image to a specific width, a “slice” along the z-axis of the sample is chosen by selective application of a $G_z$ gradient during the RF pulse, which is amplitude modulated as a sinc function

$$\text{sinc}(t) = \frac{\sin x}{x} = \frac{\sin \pi t}{\pi t}$$

The Fourier transform of a sinc pulse is rectangular in the frequency domain, with a constant amplitude over a fixed bandwidth. Therefore, only frequencies within that bandwidth are tipped to the transverse plane and the $G_z$ gradient selects the width of the sample in which the magnetization is tipped. Spins not in the slice remain along the z-axis. The basic elements of the standard 2-D Fourier transform spin-echo imaging sequence are illustrated in Figure 7. The actual sequence used in practice is somewhat more complicated than the sequence depicted, but is beyond the level of detail required to address the topic of this thesis.

$$F(k_x, k_y) = \int \int M_0(x, y) e^{ik_x x} e^{ik_y y} dx dy$$
Figure 7: Illustration of a standard 2-D Fourier transform spin-echo imaging sequence.
ACCREDITATION AND CURRENT REQUIREMENTS

Accreditation of an MRI system is an independent testing procedure, different from the vendor’s testing protocols, that requires both an annual quality assurance evaluation as well as a daily and weekly quality control program to be established. Acquiring accreditation is verification that the MRI site is maintaining a specific level of performance for their MRI system. It is the job of a qualified MRI physics expert to perform an unbiased annual quality assurance evaluation, as well as making sure the quality control procedures are being followed and performed properly by the MRI technologists, or other qualified person who is conducting these tests.

At the current time in the United States, there is no requirement for sites which perform MRI scans to be accredited by any governing body. Accreditation is a voluntary process for those MRI sites who wish to show a commitment to quality assurance and control, but is not necessary for a site to conduct MRI scans on patients. Due to this lack of requirement for accreditation, it is possible for unaccredited sites to unknowingly have lower image quality than sites that meet the accreditation criteria, possibly affecting the diagnosis of the patient.

Another issue that arises in the MRI accreditation process is that there are two major accrediting bodies for MRI sites. These are the American College of Radiology (ACR) and the Intersocietal Commission for the Accreditation of Magnetic Resonance Laboratories (ICAMRL) (1, 2). Each organization has its own criteria to pass the
accreditation evaluation, meaning one accredited MRI site could produce images of a significantly lower quality than another MRI site who is also accredited, even though both sites pass their respective accreditation requirements.

Recently there has been a push for MRI sites to become accredited. This push was helped at the start of 2001, when healthcare provider, Aetna announced that after January 1, 2001, they would no longer pay for MRI scans performed at unaccredited MRI sites (3). Other healthcare providers have begun to follow suit, including UnitedHealthCare, who will no longer reimburse outpatient MRI scans at unaccredited sites after 2009 (4). In 2008, the Medicare Improvements for Patients and Providers Act was signed into law (5, 6). It included a provision that stated that providers of advanced imaging services seeking federal reimbursement are required to be accredited by January 1, 2012.

While most of these provisions being instated by healthcare providers do not specify which accrediting body the MRI site needs to be approved by, there are instances where a specific test must be passed. This is the case for a law that came into effect in Rhode Island as of Jan 1, 2000 (7). This law requires that MRI sites within Rhode Island must be accredited by the ACR in order to perform MRI scans on patients. However, the majority of MRI sites across the United States may choose which organization by which they wish to be accredited, assuming they pass the accreditation process.

Due to the fact that most MRI sites in the United States can choose which institution they would like to apply for accreditation through, it has become a necessary concern to compare the quality of images produced at the limits of the following criteria for each accrediting processes.
SIGNAL-TO-NOISE RATIO

Signal-to-noise ratio is the relationship between the MR signal and the amount of image noise present in a given region of interest. Specifically, the signal-to-noise ratio is the quotient of the signal intensity measured in a region of interest and the standard deviation of the signal intensity in a region outside of the object being imaged. The primary sources of noise come from the sample and the MRI system.

The sample, in our case the phantom or patient, contains a number of potential sources of noise. The first and most basic source of noise from a sample is thermal noise. Due to the fact that we are not conducting scans at absolute zero, the molecules of the sample are moving. Molecular movement leads to energy exchange between molecules, including water molecules, with which we are concerned. These energy exchanges between water molecules modulate the MR signal randomly. If the sample we are dealing with is a patient rather than a phantom, physiologic changes in tissue content or location, such as through respiration, blood flow, and diffusion also create variations in the MR signal. While some of these disturbances are predictable, such as cardiac or respiratory cycles, the fact that we are not interested in these events make them signal that is of no interest to the scan, and therefore are considered noise.

The MRI system also contributes random perturbations to the signal which result in noise. The electronic system used to generate the RF signal, gradient magnetic fields, and run the computer systems are less than 100% efficient. As a result of this, these
systems generate electromagnetic fields that either alter the signal within the sample or are themselves detected as MR signal. An example of noise arising from the MRI system is the effect of eddy currents. An eddy current is an electrical current induced in a conductor that is subjected to a time-varying magnetic field. During imaging, gradient magnetic fields are turned on and off at high rates, causing a time-varying magnetic field. From Faraday’s law of induction, we can predict that this will induce a current in any nearby conductor, creating these eddy currents. Once current is flowing through this conductor, it will in turn generate a unique magnetic field. Therefore, we have our gradient magnetic field, as well as an additional magnetic field induced by the gradient magnetic field. The combination of these fields results in a net gradient magnetic field that is different from the linear gradient we expect to be applied to the sample. This variation from the expected gradient magnetic field results in perturbations that cause noise as well, although their effects can be minimized through shielding or factoring the induced magnetic fields from the eddy currents into the determination of the gradient magnetic field.

When acquiring an MRI image we want signal, but don’t want noise. Although it is impossible to completely eliminate noise, there are ways to maximize the signal-to-noise ratio. The signal-to-noise ratio can be described as being dependent on a few different variables. These variables are as follows:

- Voxel volume = $\Delta x \cdot \Delta y \cdot \Delta z$
- Number of excitations/acquisitions ($N_{EX}$)
- Number of phase-encoding steps ($N_y$)
- Bandwidth (BW)
These variables affect the signal-to-noise ratio as follows:

\[
SNR \propto (\text{voxel volume}) \sqrt{(N_y)(N_{EX})/BW}
\]

If we increase the voxel volume, we also increase the number of proton spins in each voxel, therefore, increasing the signal coming out of the voxel. The voxel size is given by

\[
Voxel \, Volume = \Delta x \cdot \Delta y \cdot \Delta z
\]

where \(\Delta x\) is the pixel size in the x direction, \(\Delta y\) is the pixel size in the y direction, and \(\Delta z\) is the slice thickness. To achieve optimal image resolution, very thin slices with a high signal-to-noise ratio are desirable. However, thinner slices decrease the voxel size and therefore the number of proton spins recorded, leading to less signal per voxel and a higher signal-to-noise ratio.

Increasing the number of times the sample is scanned \((N_{EX})\) also increases the signal-to-noise ratio. When considering a slice in the sample, there is a constant noise associated with each signal \((N_1 = N_2 = N)\), as well as the signal from each slice \((S_1 = S_2 = S)\). If we add up the signals for two acquisitions, we get:

\[
S_1 + S_2 = 2S
\]

The noise however cannot be added this way, as it needs to be considered as a
variance of a Gaussian distribution ($\sigma = \text{standard deviation}$). Therefore, the sum of noise distributions for two acquisitions would be given by:

$$\sigma_1^2 + \sigma_2^2 = \sigma^2 + \sigma^2 = 2\sigma^2$$

From this equation, the standard deviation is calculated to be:

$$\sqrt{(2\sigma^2)} = (\sqrt{2})\sigma = (\sqrt{2})N = 1.41N$$

Combining the equations for signal and noise when two acquisitions are taken, we arrive at the equation:

$$\frac{S_1 + S_2}{N_1 + N_2} = \frac{2S}{\sqrt{2N}}$$

This equation shows us that for two acquisitions, the resulting signal will be twice the signal of one acquisition, while the resulting noise will be the square root of two times larger than the noise of one acquisition. Therefore, if we were to increase the number of acquisitions ($N_{\text{EX}}$) by a factor of two, the signal-to-noise ratio will increase by a factor of $2$ divided by the square root of two, or simply, the square root of 2. This process is called “signal averaging”, and is implemented to reduce the noise of images with a small voxel size.

It should be noted however, that increasing the number of acquisitions ($N_{\text{EX}}$) by a factor of two, will also increase the scan time by a factor of two. This needs to be taken
into consideration, as total scan time can increase dramatically when taking multiple averages of a sample. The fact that scan time is increased at a much higher rate than SNR is increased means that the image is improved in a manner that may not be cost effective to the MRI clinic.

The same concept for the increase in signal-to-noise ratio holds true for the number of phase-encoding steps ($N_y$). That is, there is an increase in signal-to-noise ratio by a factor of the square root of two as the number of phase-encoding steps is doubled. However, an increase in the number of phase-encoding steps increases the scan time proportionally.

There is an inverse relationship between bandwidth (BW) and signal-to-noise ratio. As bandwidth is increased (widened), we include more noise, and the signal-to-noise ratio decreases. Conversely, if the bandwidth is decreased (narrowed), we allow less noise to come through, and the signal-to-noise ratio increases. The bandwidth is the range of frequencies of the signals received by the MRI system. The bandwidth can be given by:

\[
\text{Bandwidth} = \gamma \cdot (\text{Magnetic Gradient}) \cdot (\text{Field of View})
\]

where $\gamma$ is the gyromagnetic ratio for a hydrogen proton. Decreasing the bandwidth causes an increase in scanning time however, and costs the clinic more to perform. The decrease in bandwidth is also limited by the abilities of the MRI system being used and cannot always be lowered.

Another factor that needs to be considered is the interslice gap. The interslice gap
is a small distance of space in the subject between two adjacent slices. It would be desirable to attain contiguous slices when imaging a subject, but interslice gaps are necessary due to imperfections of RF pulses. Because the resultant slice profiles are not perfectly rectangular, but rather resemble a Gaussian distribution, two adjacent slices overlap at their edges when closely spaced. Under these conditions, the RF pulse for one slice will also excite protons in the adjacent slices. This interference of adjacent slices is known as “cross-talk”. Cross-talk produces saturation effects and thus reduces the signal-to-noise ratio.

When selecting an appropriate interslice gap for imaging, it is necessary to find a compromise between an optimal signal-to-noise ratio, which requires a large enough gap to completely eliminate cross-talk, and the desire to reduce the amount of information that is not recorded when the interslice gap is too large. This range is typically in the range of 25-50% of the slice thickness implemented. When calculating the signal-to-noise ratio for accreditation purposes however, the interslice gap is typically much larger than this range (70-100% slice thickness) to eliminate any possibility of cross-talk occurring between slices.

**Spatial Resolution**

Spatial Resolution is the minimum distance that we can distinguish between two points on an image. While in MRI we are concerned with voxels in three dimensions for a total spatial resolution, we are looking at the resolution of the 2-D image produced for each slice. While it would be most beneficial to have an infinitely small slice thickness,
this is not possible, and would contribute to other image degrading factors (such as signal-to-noise ratio). If the same slice thickness is maintained throughout the production of MR images, we consider the 2-D images produced to have nearly the same amount of slice averaging, and the 2-D plane through the slice has a varying spatial resolution based on the field of view and the size of the image matrix. Therefore, the spatial resolution needs only be considered in the two-dimensional plane that the slices are imaged.

The matrix of an MR image is a two-dimensional grid of rows and columns. Each square of the grid is a pixel, with a specific height (y-value) and width (x-value), is assigned a value that corresponds to the average signal intensity of the section of a slice being imaged. Each pixel in the two-dimensional MR image therefore provides information for a corresponding volume element, known as a voxel, which is the product of the pixel dimensions in the x and y-direction and the slice thickness in the z-direction. The size of this voxel determines the resolution of an MR image, and specifically in the directions parallel to the slice plane the spatial resolution of the image is defined. The pixel size of the image is then defined as

\[
\text{Pixel Size} = \frac{\text{FOV}}{\# \ of \ pixels}
\]

where FOV is the field of view, which is the overall length of the image in the x or y-direction.

When the matrix size is held constant, the field of view determines the size of the pixels. Conversely, when the field of view is held constant, the matrix size determines the size of the pixels. The pixel size in the phase-encoding direction (x-direction) is
calculated as the field of view in mm divided by the matrix size in the phase-encoding direction. Similarly, the pixel size in the frequency-encoding direction (y-direction) is calculated as the field of view in mm divided by the matrix size in the frequency-encoding direction.

Due to the fact that the size of the matrix is a determining factor in the size of the pixels of an image, a larger matrix will improve the spatial resolution by allowing smaller elements to be resolved. This matrix size is limited however by the signal-to-noise ratio also being dependent on the pixel size. As described previously, signal-to-noise ratio is affected by voxel size, so decreasing the voxel in the x and y-directions (the pixel dimensions) will also decrease the signal-to-noise ratio. This decrease in signal-to-noise ratio will eventually cause degradation of the image and effect spatial resolution.

Another limiting factor for the size of the matrix is the scan time, which increases in direct proportion to the matrix size. As economic efficiency is always a priority for an MRI clinic, increasing the matrix size to improve spatial resolution must always be weighed with the amount of image improvement versus the additional time the scan will take to perform. The scan time of an image acquisition can be calculated by using the equation

\[
Scan Time = TR \times N_x \times N_{EX}
\]

where TR is the relaxation time, \(N_x\) is the number of matrix elements in the phase-encoding direction (x-direction), and \(N_{EX}\) is the number of excitations recorded, also known as averages. This equation shows that increasing the number of matrix elements
in the phase-encoding directions increases overall scan time proportionally.

**SLICE THICKNESS**

When the sample is in the MRI scanner, it is in an external magnetic field, $B_0$, which is oriented along the z-axis. If we were to transmit a radio frequency (RF) pulse and receive an echo back, the signal would be from the entire sample. The required frequency of the RF pulse is given by the Larmor frequency:

$$\omega_0 = \gamma B_0$$

If the transmitted RF pulse’s frequency did not match the Larmor frequency, none of the protons inside the sample would be excited.

If we were to vary the magnetic field from point to point however, each position would have its own resonant frequency. This varying magnetic field can be created through the use of a **gradient coil**, which causes the magnetic field strength to increase in one direction (x, y, or z). Typically this gradient is a linear increasing in one direction. Let’s consider a patient in a 1.5 Tesla MRI scanner, with a magnetic field gradient in the z-direction (axial), so that the magnetic field strength is 1.4 Tesla at the patient’s feet, and 1.6 Tesla at the patient’s head. If a radio frequency pulse with a single frequency is then transmitted into the patient, we will only excite the protons in the patient at the level of the magnetic field corresponding to that frequency, but only at an infinitesimally thin line along the gradient. We therefore must transmit an RF pulse with a range of frequencies,
or a bandwidth, to excite the protons along a measurable thickness of the magnetic field gradient.

Considering how a slice and its thickness are selected is a matter of choosing a frequency range for the radio frequency pulse. We must first note that the Larmor frequency for hydrogen protons in a 1.5 Tesla magnet is approximately 64 MHz, determined by the Larmor frequency equation:

$$\omega_0 = \gamma B_0 \Rightarrow \omega_0 = \left(\frac{42.6 \text{ MHz}}{T}\right) \times (1.5 \ T) = 64 \text{ MHz}$$

where $\gamma$ is the gyromagnetic ratio for a hydrogen proton.

Since we have the magnetic field gradient set up in such a way that the magnetic field strength is 1.4 Tesla at the patient’s feet and 1.6 Tesla at the patients head, we get a range of Larmor frequencies throughout the length of the patient as well, that are approximately as follows:

$$1.6 \ T \sim 68 \text{ MHz}$$

$$1.5 \ T \sim 64 \text{ MHz}$$

$$1.4 \ T \sim 60 \text{ MHz}$$

We can then select a slice of a certain thickness to have its hydrogen protons excited by picking a frequency range of a certain bandwidth. For example, there exists a plane orthogonal to the magnetic field gradient direction (z-direction) in which the magnetic field is 1.55 Tesla at all points, and there is also a plane orthogonal to the magnetic field gradient direction (z-direction) in which the magnetic field is 1.57 Tesla at all points. We
can excite the slice between these two planes by transmitting a radio frequency pulse with a bandwidth that ranges between the corresponding Larmor frequencies for these magnetic field strengths, which are approximately 66 MHz and 67 MHz. This causes only the hydrogen protons between these two planes to be excited and produce signal to be recorded. The range of frequencies selected in the bandwidth determines the range of magnetic field strengths that will be excited, and therefore the slice thickness that will be measured.
III. METHODS

SELECTION OF TESTING PARAMETERS

When selecting the specific acceptance criteria of each accreditation process to be tested and compared, first we need only consider the testing criteria that are present in both qualification procedures (ACR and ICAMRL) and are different. It was then decided that the acceptance criteria being tested should be those comparing the accuracy of parameters used in the scanning process and image reproduction, rather than those criteria that measure the status of the system itself, such as magnetic field homogeneity and monitor/processor quality control.

It was also important to select acceptance criteria that could be tested and compared both quantitatively and qualitatively. This would give us the ability to analyze the differences in the requirements of each accreditation procedure both numerically and also be able to see how these different criteria may affect image quality and possible diagnosis. Finally, the criteria to be tested needed to be those which could be manipulated through the program interface, as physical manipulations would have caused problems with the system, and may have affected multiple quantities at the same time.

After deciding on these qualifications, the acceptance criteria that were selected to be tested and compared in both accreditation protocols were Signal-to-Noise Ratio (SNR), Spatial Resolution, and Slice Thickness Accuracy. The effect of differing acceptance criteria was then evaluated by adjusting the MRI system and acquisition
protocol so that the resulting images met both accreditation standards, and comparing these images with images obtained after manipulating the MRI system so that it met each accreditation protocol’s standards by a small margin.

If we set the acceptance criterion at or near the worst acceptable value for the protocol which was less stringent in the criterion being tested, we could impute to what extent image quality could differ between two MRI sites, who were both accredited (albeit, by different accrediting bodies); when only one of these criteria was outside one protocol’s acceptance range but within the other protocol’s acceptance range.

SELECTING TESTING PROCEDURES AND ACTION LIMITS

After selecting the acceptance criteria that were present in both accreditation processes, it was necessary to choose a standard way to test each variable so that different testing methods would not influence the results.

The American College of Radiology (ACR) has a very well defined procedure for testing the criteria that must be evaluated for accreditation protocols, including a specific phantom provided by the ACR that allows for all of the required tests to be performed. The ACR MRI accreditation phantom is not required to be used when applying for accreditation, but it is recommended.

The Intersocietal Commission for the Accreditation of Magnetic Resonance Laboratories (ICAMRL) however, does not have a specific test procedure for the acceptance criteria that must be evaluated for accreditation, as well as not having a phantom available directly from the ICAMRL for use during testing. Due to the fact that
the ICAMRL does not have a specific phantom recommended for use during accreditation testing, it is up to the discretion of the medical physicist to choose a phantom, or phantoms, that are suitable to perform all the tests necessary in the ICAMRL accreditation process.

The definition of limits for the acceptance criteria being tested differs between the ACR and ICAMRL accreditation protocols as well. The ACR has a specific range and/or upper and lower bounds of limits for each criterion being tested during the accreditation process, which can be found in the accreditation manual. The ICAMRL however, does not have specific values listed for the limits of each criterion. The ICAMRL leaves the determination of the limits for each testing criterion up to the medical physicist, with some limitations. The ICAMRL accreditation guidelines state that, “The system parameters should be compared to the manufacturer’s system specifications or industry standards, and reviewed by the Quality Assurance Committee.” Upon contacting the ICAMRL, they stated that the limits for each acceptance criterion could be selected by the medical physicist, with the stipulation that documentation must be provided showing that the limits chosen were defined by either the manufacturer of the MRI system, or by an industry standard, which includes the American Association of Physicists in Medicine (AAPM). While it should also be noted that ACR standards may be used as an industry standards when defining limits for ICAMRL accreditation testing, it is possible to achieve accreditation with standards that are less stringent than those defined by the ACR, so long as they can be referenced.

The purpose of the comparison being performed was to analyze the quality of MRI images produced at or near the extreme limit ranges of each testing protocol. To
this point, the limits that need be considered for ICAMRL accreditation protocols are those limits which have the least stringent standards, yet still allow for accreditation. Due to this fact, limits for the criteria being tested were found in various sources from industry standards to demonstrate the extreme limits for which ICAMRL accreditation is still able to be achieved.

**SIGNAL TO NOISE RATIO**

The first acceptance criterion to be compared by manipulating the MRI acquisition protocol (field of view in this case), was the *signal-to-noise ratio*, or SNR. Signal-to-noise ratio is a measure value which indicates how much useful information is gathered from a volume and how much non-useful information we receive in the same volume during an image acquisition, also known as *noise*. In general, the higher the signal-to-noise ratio, the better; as a higher value for SNR indicates a higher value of useful signal being received in producing an image in relation to the amount of non-useful information received when producing an image.

Since the scans being performed were conducted on the same MR system, with either a phantom or the same, limited movement patient for each scan; we will consider the imperfections of the MRI system, as well as the patient related factors to be nearly the same in each acquisition, and not directly affecting the difference in image reproduction in either case. This leaves us with only factors associated with image production and processing as those we will consider to affect our image quality. There are also image production factors that are constant in all of the scans and therefore need not be
considered; including magnetic field strength and selection of the transmit and receive coils (as we will always be using the body coil to transmit, and the head coil to receive signals). Therefore, the factors we will consider to affect the signal-to-noise ratio are the following:

- Slice thickness
- Field of view
- Size of the image matrix
- Number of acquisitions
- NMR specific scan parameters

When performing the scan on the phantom at each specification, we used a T1-weighted, axial scan as described by ACR protocols. This means that the scans produced of the phantom in each comparative signal-to-noise ratio case have the same scan parameters, as well as having only one acquisition. While image matrix is not specified in the ACR axial scan protocols, the same matrix was used for each signal-to-noise ratio scan. This being the case means that the only factors manipulated in the scans of the phantom were the slice thickness, field of view, and image matrix size; which is effectively changing the size of the voxels in all three dimensions (x and y-directions via field of view and image matrix size, and z-direction via slice thickness).

Due to the fact that MRI systems differ greatly, both accrediting bodies require that we must first come up with a mean signal-to-noise ratio, as well as standard deviation values for the specific machine being used. To do this multiple signal-to-noise measurements were taken using ACR testing protocols. These baseline measurements are made using a typical T1-weighted axial scan that will be performed on the head of a
patient. The manufacturer has a specific field of view for they believe to be satisfactory for their head coil to image a patient, and that scan is what is used when creating these baseline measurements. The ACR testing protocol for signal-to-noise ratio requires that a slice image is acquired through the homogenous portion of the ACR phantom, which is a portion where the only material throughout the cross section of the phantom is the saline that fills the volume. The T1-weighted axial scan used for signal-to-noise ratio measurements does not have a specified slice thickness, however it is typical by ACR standards and by the standards of the MRI system that was used to perform scans with a 5mm slice thickness and a 5mm interslice gap when using a head coil for receiving the RF signal. Therefore, when signal-to-noise ratio measurements were taken to determine the mean and standard deviations of the specific machine, these are the specifications that were used. A field of view was also selected so that the whole cross sectional area of the phantom was inside this field and a small amount of empty space around the phantom was also present to calculate the noise standard deviation.

The scan was then performed at the above specifications, and the image of the homogenous cross section of the phantom was analyzed for signal-to-noise ratio values. This was done by choosing a region of interest (ROI) that covers approximately 80% of the cross sectional area of the phantom, as viewed in the image. This region of interest is known as the “Mean Signal ROI”. Next, a region of interest was created that is approximately 0.15% of the area of the field of view. This second region of interest was then placed in the empty space outside the phantom on the image in the frequency-encoding direction. This made sure that the signal received in this region of interest did not receive any “ghost signal”, which is a phenomenon that occurs across the phase-
encoding direction of the image outside the phantom. This second region of interest is known as the “Noise ROI”.

Figure 8: Example of regions of interest used in calculations during ACR accreditation protocols, with regions #1 (the majority of the volume of the phantom, with values listed inside the left of the phantom) and #4 (located outside the phantom at the top of the figure) being those used for SNR calculations. The lighter the color, the more signal; so the “black” space in region #4 is very low signal and is the noise.
The signal-to-noise ratio was then calculated by dividing the mean signal value in the Mean Signal ROI by the standard deviation value in the Noise ROI:

\[
SNR = \frac{\text{Mean Signal}}{\text{Noise Standard Deviation}}
\]

This procedure was then repeated five times to produce an average value and standard deviation for the specific MRI system used. The five calculated signal-to-noise ratios were then used to find the average signal-to-noise ratio.

This average signal-to-noise ratio is then used to calculate the standard deviation of the value using the following equation for small sample sizes:

\[
\text{Standard Deviation} = \sqrt{\frac{(\bar{x} - x_1)^2 + (\bar{x} - x_2)^2 + \cdots + (\bar{x} - x_n)^2}{n - 1}}
\]

Using these calculated values, we can determine the acceptable ranges required that the MRI system must operate within to pass the standards of each accrediting body. ACR standards specify that to pass accreditation evaluation, the signal-to-noise ratio must fall within one standard deviation of the baseline measurements, made with the manufacturer’s prescribed head scan specific to their listening coil. This gave us an acceptable range for ACR accreditation. However, only lower signal-to-noise ratio will cause a deterioration of the image quality, so we need only to concern ourselves with the lower bound of this range. Since ICAMRL does not specify a distinct value, we must
select one from the manufacturer, or industry standards. The standard chosen was defined by the American Association of Physicists in Medicine (AAPM) in their Task Group-34 report, which states that signal-to-noise ratio must be within 20% of the average baseline measurement. This gave us a range for ICAMRL accreditation. Again however, we are only concerned with the lower limit of the range found for image quality.

After the lower limits for signal-to-noise ratio were determined for each accreditation requirement, images were taken with each respective SNR. Since the scan protocol called for a 5mm slice thickness, this was maintained throughout all the scans, and the parameters that were changed were the field of view and the matrix size. This means that each voxel had a constant length of 5mm in the z-direction, while the length of the sides of each voxel in the x and y-directions were determined by the field of view in each direction (in mm) divided by the number of matrix elements in that direction. For the scans taken of the phantom, the image matrix size was held constant at \(256 \times 256\) elements. This means the only parameter that was changed was the field of view to affect the signal to noise ratio during scans of the phantom. For simplicity, the field of view was also given the constraints that the number of matrix elements in each dimension should be the same, giving us a square image matrix for the field of view in all scans of both the phantom and the patient.

The first scan was then performed with the parameters set such that the signal-to-noise ratio of the image produced would fall well above the lower limit of both accreditation protocols, and very near the average of the baseline measurements. The next scan that was conducted had the field of view manipulated in such a way that the
signal-to-noise ratio was reduced to a value that fell just within ACR accreditation standards, giving a visual representation of the lower limits of the ACR testing protocol. The final scan that was conducted was done so with a field of view that was manipulated in such a way so the signal-to-noise ratio would fall just within the lower limit of the ICAMRL accreditation protocol we chose, which is outside of ACR accreditation standards, and gave us a visual representation of the lower limits of the ICAMRL testing protocol.

After performing the preceding scans of the phantom to provide scan parameters that created a signal-to-noise ratio at the limits of the acceptance criteria of the accreditation protocols, scans of volunteer patient were then performed using these same parameters to give a perspective of how these limits would affect image quality in a clinical application. The patient’s knee was imaged with varying slice thicknesses that may be used in a clinical scan, and the images were qualitatively compared, including possible effects of diagnosis.

SPATIAL RESOLUTION

The second acceptance criterion to be compared by manipulating the MRI acquisition protocol (field of view in this case), was the spatial resolution. Spatial resolution is the minimum distance we can distinguish between two points in an image. In general, the smaller the minimum distance we can distinguish between two points in the image, the better; as a better spatial resolution allows us to be able to view smaller physical features of the tissues of the body. Spatial resolution is considered in the two-
dimensional plane that the slices produce, as the tissues present at the location of each slice are what is actually able to be viewed.

Considering that the factors affecting spatial resolution are the field of view (FOV) and the number of matrix elements in both the x and y-directions, or the phase-encoding and frequency-encoding directions respectively. Since we only consider spatial resolution as a two-dimensional parameter, we must consider the pixel size of the image, rather than the voxel size, as slice thickness does not directly influence spatial resolution. The pixel size in the phase-encoding direction is calculated as the field of view in mm divided by the matrix size in the phase-encoding direction. Similarly, the pixel size in the frequency-encoding direction is calculated as the field of view in mm divided by the matrix size in the frequency-encoding direction.

With these things considered, during the testing of the spatial resolution each scan that was performed had the image matrix was held constant with dimensions of 256 x 256, and the field of view was manipulated to alter the spatial resolution of the MR image.

In the test for spatial resolution we use the T1-weighted axial scan defined by ACR protocols with 11 slices, a 5mm slice thickness, and a 5mm interslice gap distance. When using these scan parameters and the slices are positioned in such a way that slice #1 and slice #11 intersect the crossing of the 45 degree wedges at either end of the phantom, then the spatial resolution insert is displayed in the image of slice #1.

The spatial resolution insert is a block of plastic with sets of holes in square patterns which are open to the volume of the phantom, and therefore contain the same solution that fills the bulk of the phantom. Each square consists of 16 holes in a $4 \times 4$
pattern with varying hole sizes. The squares are arranged in pairs, with the first pair of squares with holes that are 1.1mm in diameter, a second pair of squares with holes that are 1.0mm in diameter, and a third pair of squares with holes that are 0.9mm in diameter. The resulting configuration is three pairs of squares, where both squares in each pair share one hole between them. The first square in each pair is located upper-left to the second square, which is therefore located lower-right in relation to the first square. This means that the first square’s lower-right hole is also the upper-left hole of the second square. The upper-left square matrix of holes in each array is used to measure the left to right resolution of the MR image, while the lower-right square matrix of holes is used to measure the top to bottom resolution of the MR image (Figure 9).

![1.1 mm Array 1.0 mm Array 0.9 mm Array](image)

**Figure 9: Arrays of holes used for spatial resolution measurement in the left-right and up-down direction.**

When determining the spatial resolution of the image collected, we must then analyze it in both the left-right and up-down directions using the appropriate square matrices of circles listed above. First we must adjust the window of contrast values for the image to a narrow band, making the circles more easily distinguishable individually.
The spatial resolution of the image in the left-right direction is determined using the upper-left square of each pair. With this being the case, we need to look at the rows of the upper-left matrix of each pair. We begin with the left most pair of hole arrays, which is the pair with the largest hole size, 1.1 mm. If all four holes in any single row are distinguishable from one another, the image is considered resolved left to right at this particular hole size, and in turn at this resolution. Continue this process for the middle and right most pairs of hole arrays (1.0 mm and 0.9 mm) and the array with the smallest hole size in which all four holes in any row can be resolved is the left-right spatial resolution of the MR image. This same process is then repeated in a similar manner for the lower-right matrix of holes in each pair to determine the spatial resolution in the up-down direction. In this case however, the columns of the lower-right square should be analyzed to see if all four holes can be distinguished from one another in an up and down manner. The array with the smallest hole size in which all four holes in any column can be resolved is the up-down spatial resolution of the MR image. In our experimentation, the field of view is maintained at square dimensions, so the spatial resolution in the left-right and the up-down directions should be very nearly the same for all the images collected.

It should also be noted that when determining whether or not all four holes in a row or column are “distinguishable”, the ACR protocol guidelines give specific instructions for determining what is needed to meet this criterion. It is not required that the image intensity drop to zero between the holes, even with the small contrast window selected. It is only necessary that with a narrow window and an appropriate level to distinguish the holes well from the surrounding material of the insert, that all four holes
are recognizable as points of brighter signal intensity than the spaces between them. Also, it is possible for holes from adjacent columns to blur together when deciding if a row is resolved, yet the spaces between holes in the row are all still distinguishable. In this case the spaces between holes in the row are all that we are concerned with and the columnar blurring does not affect the scoring for the row.

After the spatial resolution scans were performed on the phantom, the scan parameters were again recorded and applied to scans performed on a volunteer patient. The patient’s knee was imaged with scan parameters the same as those found to produce a spatial resolution that is at the limits of each accreditation protocol’s acceptance criterion. These images were again qualitatively compared and assessed for possible effects on diagnosis.

SLICE THICKNESS ACCURACY

The final acceptance criterion to be compared by manipulating the MRI acquisition protocol (magnetic field gradient in this case), was the slice thickness accuracy. Slice thickness accuracy is a test that determines the accuracy of the specified slice thickness during image acquisition. In this test, prescribed slice thickness is compared with a measured slice thickness.

According to ACR testing protocols, a slice thickness in the range of 3mm to 7mm is recommended. For simplicity and continuity, the T1-weighted axial scan defined by ACR protocols was used during the slice thickness accuracy testing. This scan has a slice thickness of 5mm and an interslice gap of 5mm also, which falls within the range
necessary for the slice thickness accuracy to be measured appropriately. When using the T1-weighted axial scan as prescribed by the ACR protocol with 11 slices, 5mm slice thickness, and 5mm interslice gap, with slices #1 and #11 positioned so they are aligned with the vertices of the 45 degree wedges at the superior and inferior end of the phantom respectively; slice #1 will be in the correct position to perform the slice thickness accuracy measurement. In slice #1, two thin opposed inclined ramps appear in a structure called the “slice thickness insert”, which will be used to determine the actual slice thickness imaged.

![Figure 10: Sagittal view of plastic ramps used in slice thickness accuracy calculation. One (Ramp 1) has a positive slope from left to right and is forward in the picture, the other (Ramp 2) has negative slope from left to right and is just behind the first ramp.]

If the phantom is not tilted, the two bright thick lines, which represent the ramps,
will appear on top of one another. These two ramps are crossed; one has a negative slope, and the other has a positive slope with respect to the plane of slice #1. The ramps are cut from a block of plastic and are filled with the same solution that fills the bulk of the phantom. The signal ramps have a slope of 10 to 1 with respect to the plane of slice #1, creating an angle of about $5.71^\circ$ with slice #1. Therefore, the signal ramps will appear in the image created of slice #1 with a length that is ten times the thickness of the slice. If the phantom is tilted, one ramp will appear longer than the other; however, this error is corrected by averaging the measurements from the two crossed ramps and using the slice thickness formula provided in the ACR testing protocol.

$$Slice\;Thickness = 0.2 \times \frac{(top \times bottom)}{(top + bottom)}$$

After the T1-weighted axial scan is completed, the image of slice #1 was displayed and magnified by a factor of two in order to increase the accuracy of the length measurements. The display level and window were then adjusted so that the signal ramps were well visualized. The on-screen length measurement tool was then used to measure the lengths of the top and bottom ramps. These values are then used in the slice thickness formula defined by the ACR testing protocol. It should be noted that the ends of the ramps appear to be scalloped or ragged, and one must estimate the average location of the ends of the ramps in order to measure the ramp lengths. This causes a degree of error in the ramp length measurement; however, due to the fact that the ramps have a slope of 10 to 1 with the plane of slice #1, a millimeter of error in the ramp length measurement corresponds to only a tenth of a millimeter error in the slice thickness, and turns out to be a small effect.
The measurements for both the top and bottom ramp were then entered into the slice thickness formula to determine the actual thickness of the slice imaged. According to the ACR accreditation standards, the measured slice thickness for the ACR T1-weighted axial imaging series should be 5.0mm ± 0.7mm. This gives us a range of acceptable values from 4.3mm-5.7mm for actual slice thickness.

For the ICAMRL standards a value from industry standards was used for the acceptable error in the slice thickness accuracy. This value for acceptable error in slice thickness accuracy was taken from the American Association of Physicists in Medicine (AAPM) Task Group – 28 report. The AAPM TG-28 report specifies that measured slice thickness should be within 20% of the prescribed slice thickness for the scan. Since the ACR T1-weighted axial imaging scan has a slice thickness of 5mm, this means the acceptable error is 1.0mm. This gives an acceptable range of 4.0mm-6.0mm for the actual slice thickness.

Three scans of the ACR phantom were then performed to achieve a visual representation of the limits of each accreditation protocol. The first scan was then performed so that the actual slice thickness would fall well within the limits of both accreditation protocols and be near the prescribed slice thickness. The next scan had the slice thickness manipulated in such a way that the measured slice thickness would fall just inside the upper limit of ACR accreditation standards for a T1-weighted axial scan series with a 5mm slice thickness, by altering the magnetic field gradient in such a way that the range of magnetic field strength that covered a distance of 5 mm in the first scan, would now cover a distance of 5.7 mm. The final scan of the phantom was then performed with the slice thickness manipulated in such a way that the measured slice
thickness was just inside the upper limit of the defined ICAMRL accreditation standards for a T1-weighted axial scan series with a 5mm slice thickness, by again manipulating the magnetic field gradient so that the range of magnetic field strengths that covered a distance of 5 mm in the first scan, would now cover a distance of 6.0 mm.

Once again, after performing scans to determine the different slice thickness accuracy limits for each accreditation protocol, the scan parameters were recorded and were applied to scans of a patient. In this instance, a volunteer patient known to have a brain tumor was selected to demonstrate how the different limits for the slice thickness accuracy could affect diagnosis and treatment analysis of the patient. During treatment of tumors, including chemotherapy and radiation therapy, the change in volume of a tumor is a major determining factor on the effectiveness of the treatment. The images collected demonstrate how the different limits for acceptance criteria may affect the calculated volume of a tumor, and therefore the assessment of how effective a treatment is.
IV. RESULTS

SIGNAL TO NOISE RATIO

First, it was necessary to establish baseline measurements and a standard deviation of the signal-to-noise ratio calculated from these measurements.

<table>
<thead>
<tr>
<th>Scan</th>
<th>Mean Signal</th>
<th>Noise Standard Deviation</th>
<th>Signal-to-Noise Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Scan 1</td>
<td>1798.9</td>
<td>4.9</td>
<td>367.1</td>
</tr>
<tr>
<td>Scan 2</td>
<td>1797.7</td>
<td>5.2</td>
<td>345.7</td>
</tr>
<tr>
<td>Scan 3</td>
<td>1801.6</td>
<td>4.4</td>
<td>409.5</td>
</tr>
<tr>
<td>Scan 4</td>
<td>1797.8</td>
<td>4.6</td>
<td>390.8</td>
</tr>
<tr>
<td>Scan 5</td>
<td>1800.6</td>
<td>5.2</td>
<td>346.3</td>
</tr>
</tbody>
</table>

The five calculated signal-to-noise ratios were then used to find the average signal-to-noise ratio.

\[
Average \, SNR = \frac{367.1 + 345.7 + 409.5 + 390.8 + 346.3}{5} = 371.9
\]
These values then gave a standard deviation of

\[
\text{Standard Deviation} = 13.99
\]

The signal-to-noise ratio tested with the phantom was manipulated and measured to satisfy three different criteria. The first scan was given parameters to allow for the optimum signal-to-noise ratio from the MRI system, and measured to show that the SNR falls well within the limiting parameters for both ACR and ICAMRL accrediting bodies. The second scan was given parameters to create a signal-to-noise ratio that is just within the lower limit of the ACR accreditation protocol, giving us a reference for the lowest acceptable SNR allowed by ACR standards, and the scan parameters associated with this measured SNR. The third and final phantom signal-to-noise ratio scan was given parameters to create signal-to-noise ratio that was just within the lower limit of the ICAMRL accreditation protocol, giving us a reference for the lowest acceptable SNR allowed by ICAMRL standards, and the scan parameters associated with this measured SNR. The scan parameters and slice positioning used can be seen in the figures in the appendix.
Figure 11: Signal-to-noise ratio calculation well within both accreditation protocols.

Region #1 is the majority of the phantom cross-section and measures the signal from the solution inside, and region #2 is outside the phantom where the signal is very low (black) and is due to noise only.
The field of view in the image collected within both accreditation standards (Figure 11) was 230mm x 230mm, which is large enough to include the entire cross-section of the phantom. The calculated signal-to-noise ratio for the image produced was.

\[ \frac{1809.9}{2.8} = 646.4 \]
Figure 12: Signal-to-noise ratio calculation slightly above lower limit of ACR accreditation standards.

The field of view in the image collected just above ACR limits (Figure 12) was 135mm x 135mm, which was less than the cross sectional area of the phantom as seen in the figure. The calculated signal-to-noise ratio for the image produced was.
\[
\frac{1800.1}{5.0} = 360.0
\]

Figure 13: Signal-to-noise ratio calculation slightly above lower limit of ICAMRL accreditation standards.
The field of view in the image collected just above ICAMRL limits (Figure 13) was 115mm x 115mm, which was an even smaller portion of the total phantom cross-section. The calculated signal-to-noise ratio for the image produced was.

\[
\frac{1815.0}{5.7} = 318.4
\]

The signal-to-noise ratio was then altered in scans of a patient to illustrate how degraded images can be at the lower limits of each accreditation protocol. Scans were performed of the patient’s knee with parameters matching those for the calculated signal-to-noise ratios listed above. Three scans were performed with 3mm slice thickness. The first used the parameters that created an SNR calculation well within both accrediting limits. The second scan used the parameters found to produce an SNR slightly above the lower limits required during ACR accreditation. The third scan used the parameters found to produce an SNR slightly above the lower limits required during ICAMRL accreditation. Two more scans were then performed with the slice thickness reduced to 1mm to illustrate how much low limits of SNR can affect image quality for small slice thickness scans. The first scan with a 1mm slice thickness used the parameters found to produce an SNR slightly above the lower limits required during ACR accreditation. The second scan with a 1mm slice thickness used the parameters found to produce an SNR slightly above the lower limits required during ICAMRL accreditation. All of the following images are at a similar slice location that bisects the anterior cruciate ligament.
Figure 14: Sagittal scan of patient’s knee bisecting ACL with SNR well within both accreditation protocols, with 3 mm slice thickness.
Figure 15: Sagittal scan of patient’s knee bisecting ACL with SNR slightly above lower limit of ACR accreditation standards, with 3 mm slice thickness.
Figure 16: Sagittal scan of patient’s knee bisecting ACL with SNR slightly above lower limit of ICAMRL accreditation standards, with 3 mm slice thickness.
Figure 17: Sagittal scan of patient’s knee bisecting ACL with SNR slightly above lower limit of ACR accreditation standards, with 1 mm slice thickness.
Figure 18: Sagittal scan of patient’s knee bisecting ACL with SNR slightly above lower limit of ICAMRL accreditation standards, with 1 mm slice thickness.

The degradation of the image quality and ability to see the tissues in the slice is what is being compared. The main focus is the ACL, and the ability to distinguish its condition for diagnosis. Also, it should be noted that the images were cropped to maintain a somewhat similar area of the knee to be focused upon. The lowering of the SNR in these final scans degraded the image to a point that the ligaments and possible partial tears are not able to be seen. It is especially difficult to see the structures in the ICAMRL limit image when the 1 mm slice thickness is selected.
SPATIAL RESOLUTION

The spatial resolution tested with the phantom was manipulated and measured to satisfy three different criteria; well within both accreditation protocols, just within ACR acceptance limits, and just within ICAMRL acceptance limits.

Figure 19: Spatial resolution insert with resolution well within both accreditation protocols.
The field of view in the image collected just above with spatial resolution well within both accreditation protocols (Figure 19) was 200mm x 200mm, which was a field of view that was just large enough to include the entire phantom.

Figure 20: Spatial resolution insert with resolution slightly above the lower limit for ACR accreditation standards.

The field of view in the image collected just above for spatial resolution just within ACR limits (Figure 20) was 240mm x 240mm, which was a field of view that was larger than the diameter of the phantom, as the phantom has a diameter of 200 mm.
Figure 21: Spatial resolution insert with resolution slightly above the lower limit for ICAMRL accreditation standards.

The field of view in the image collected just above for spatial resolution just within ICAMRL limits (Figure 21) was 270mm x 270mm, which was a field of view that was larger than the diameter of the phantom, as well as larger than the field of view used for the ACR limit image.
It can be seen from the scans of the phantom testing spatial resolution how the
decrease in spatial resolution causes small objects to no longer be distinguished from
their surrounding material. In the first scan of the phantom we see all 9 holes in each
square of all three arrays. In the second scan, we can still see the different holes in each
square of all three arrays, however some blurring has begun to occur between holes, and
the signal does not drop to zero (or the level of only noise) in the middle and rightmost
array. In the third scan, we can still distinguish all the holes separately in the leftmost
and middle arrays, even though there is blurring. In the rightmost array however, we can
no longer distinguish four separate holes in either direction. Therefore, in the final scan
objects that are 0.9 mm or less cannot be resolved.

The spatial resolution was then altered in scans of a patient to illustrate how
degraded images can be at the lower limits of each accreditation protocol. Scans were
performed of the patient’s knee with parameters matching those for the calculated spatial
resolution listed above. Three scans were performed with 3mm slice thickness. The first
used the parameters that created a spatial resolution well within both accrediting limits.
The second scan used the parameters found to produce a spatial resolution slightly above
the lower limits required during ACR accreditation. The third scan used the parameters
found to produce a spatial resolution slightly above the lower limits required during
ICAMRL accreditation. The images produced were done so in a similar fashion as those
produced in the signal-to-noise ratio comparison, except the field of view in these images
was increased rather than decreased. This increase in field of view made the voxel sizes
larger and we are comparing the effects this larger volume covered by each pixel on the
2-D image have any effects on image quality.

Figure 22: Sagittal scan of patient’s knee bisecting ACL with spatial resolution well
within both accreditation protocols.
Figure 23: Sagittal scan of patient’s knee bisecting ACL with spatial resolution slightly above lower limit for ACR accreditation standards.
Figure 24: Sagittal scan of patient’s knee bisecting ACL with spatial resolution slightly above lower limit for ICAMRL accreditation standards.

The images in the spatial resolution comparison are not degraded as much as the signal-to-noise ratio images, and don’t cause as many problems with the ability to see structures in the knee. This is due to the fact that the field of view was not altered by as large an amount for the spatial resolution tests as it was for signal-to-noise ratio tests. While the field of view was changed by approximately a factor of two in the SNR comparisons, the field of view for spatial resolution was only increased by 20-35%. There is enough blurring present in the scan with spatial resolution standards near ICAMRL limits however to possibly cause partial tears or other small abnormalities to be missed, but they would have to be minor structural damage to be overlooked.
SLICE THICKNESS ACCURACY

The slice thickness accuracy tested with the phantom was manipulated and measured to satisfy three different criteria; well within both accreditation protocols, just within ACR acceptance limits, and just within ICAMRL acceptance limits.

Figure 25: Slice thickness accuracy measurement with slice thickness well within both accreditation protocols.
Figure 26: Slice thickness accuracy measurement with slice thickness just below upper limits for ACR accreditation standards.
Figure 27: Slice thickness accuracy measurement with slice thickness just below upper limits for ICAMRL accreditation standards.

It should be noted how the length of the ramps observed and measured in the phantom in each trial differed. The distances measured were used to calculate the distance a similar magnetic field gradient change covered in each trial. The magnetic field gradient change was such as to define a single slice. The field of view in each image was not changed, and instead these images focus on the size of the z-dimension.

Multiple T2-axial scans were then performed on a patient with a brain tumor located just posterior to the pons at varying slice thicknesses to illustrate how variations in slice thickness accuracy could affect produced images, and possibly diagnosis of conditions and treatment effectiveness. The first scan was performed with parameters
that created a slice thickness accuracy measurement well within both accreditation protocols. The second scan was performed with parameters that created a slice thickness accuracy just below the highest limits for ACR accreditation. The third scan was performed with parameters that created a slice thickness accuracy just below the highest limits for ICAMRL accreditation. The following images are those images that contained any portion or signal from the tumor within the slice.
Figure 28: Superior, middle and inferior slices with signal from tumor tissue for slice thickness accuracy well within standards for both accreditation protocols.
Figure 29: Superior and inferior slices with signal from tumor tissue for slice thickness accuracy just below upper limit for ACR standards.

Figure 30: Superior and inferior slices with signal from tumor tissue for slice thickness accuracy just below upper limit for ICAMRL standards.
It can be seen in the images above that the apparent volume of the tumor changes with the changing slice thickness accuracy limits as expected. It is distinctly obvious that the volume changes in the slice thickness direction (z-direction) because in the first set of images we can see signal from the tumor appear in a third slice, while signal from the tumor tissue only appears in two slices in the images produced at both accreditation limits.

For the case of the patient in these scans, the x and y- dimensions of the tumor in the slice with the most tumor signal present (middle slice in scan 1, and inferior slice on scans 2 and 3) are not drastically different. This is due to the fact that the tumor in the patient is relatively cylindrical throughout most of the length, with tapering at the ends.
V. DISCUSSION AND CONCLUSIONS

When analyzing the images taken for the signal-to-noise ratio we can see the differences in the measurements for the SNR well within both standards, at the lower limit of ACR standards, and at the lower limits of ICAMRL standards. One can see that the field of view must be made relatively small, 135 x 135 mm and 115 x 115 for ACR and ICAMRL respectively, with the MR system used to create a signal-to-noise ratio that is at the lower limits of both the accrediting bodies. From the equation for SNR with changing the changing variable

\[ SNR \propto (voxel \ volume) \sqrt{(N_y)(N_{EX})/BW} \]

we note that we are only changing the voxel volume by altering the field of view. With the original SNR of 646.4 and a field of view of 230 x 230 mm, we expect the SNR to be 379.4 at a field of view of 135 x 135 mm, and 323.2 at a field of view of 115 x 115 mm. The actual measured SNR values for the 135 x 135 mm FOV and the 115 x 115 mm FOV were 360.0 and 318.4 respectively.

The accreditation testing is performed using a imaging sequence with a 5 mm slice thickness, however many imaging techniques require a slice thickness that is smaller than 5 mm. The problem of accreditation protocols not taking varying slice thicknesses for different imaging sequences into account can be seen in the images of the patient’s anterior cruciate ligament. A typical knee scan to examine the overall anatomy of and
structures associated with the knee joint has a slice thickness of 3 mm. The first three images of the patient’s knee were performed at this slice thickness, and the degradation of the images is not at a point to drastically affect the ability to see most structures in the field of view. It is however becoming a bit of an issue in the ICAMRL limit image with a 3 mm slice thickness, while it is not as much of an issue in the ACR limit image with a 3 mm slice thickness.

When the slice thickness is lowered however, the signal-to-noise ratio does begin to become low enough that it affects the quality of the image and how well the structures can be distinguished. The final two images of the patient’s knee are performed with a slice thickness of 1 mm, which is performed in special cases, such as determining partial tears of ligaments in the body. When the slice thickness is at this level the signal-to-noise ratio at the lower limits of both accrediting protocols has degraded the image to a point that the ligaments and possible partial tears are not able to be seen. It is especially difficult to see the structures in the ICAMRL limit image when the 1 mm slice thickness is selected. While both accreditation protocols produce images that are not satisfactory for the diagnosis of tissue damage, the ICAMRL image demonstrates that the image quality with larger slice thicknesses than 1 mm would also be substantially degraded when a machine passes near the limits of ICAMRL accreditation standards.

These findings demonstrate that, unless the MRI clinic can ensure that they will never need to perform scans with a slice thickness on the order of 1 mm, the standards for both accreditation protocols should be more stringent. It is also can be seen by the drastic decrease in size of the field of view needed to lower the signal-to-noise ratio to the limits of the accreditation protocols, that an increase in the stringency of the standards of each
protocol would be a realistic request to ensure quality of images being produced by accredited clinics. However, it is not always possible for clinics to purchase equipment that can operate at an ideal level, so altering the range of acceptable values must also take into consideration the constraints of clinics that have lesser equipment, even though they maintain a commitment to creating quality images.

When analyzing the images of the spatial resolution inserts, it can be seen that the spatial resolution of the MR system can be made substantially better than the lower limits of each accrediting protocol’s standards. When looking at the varying sizes of the field of view to achieve spatial resolutions at the varying limits required, we see that the field of view does not need to be manipulated to a relatively high degree to affect the value measured for the spatial resolution. This leads to thinking that the ability of MR systems being evaluated may not be able to produce the necessary spatial resolution extremely easily, and it may not be reasonable to require the standards of the accrediting bodies to be drastically increased.

When analyzing the images taken of the patient’s ACL, we can also see that the spatial resolution does not dramatically decrease the ability to diagnose possible damage and tears, even partial tears, to the structures being imaged. While the quality of the image is noticeably diminished, and the ability to see possible damage is also diminished, the structures can still be distinguished relatively well. It may lend again to the fact that more stringent requirements for spatial resolution may be desirable, but are not the existing limits are at a reasonably appropriate level. It should be considered that as with the 3 mm ICAMRL limit SNR image, the ICAMRL limit spatial resolution image is near the range of affecting the ability to diagnose tissue abnormalities. There may exist the
possibility for misdiagnosis with the spatial resolution in the ICAMRL limit image, so it
would be desirable for the limits ICAMRL standards to be raised slightly, near ACR
standards.

When analyzing the slice thickness accuracy of the ACR and ICAMRL
accreditation protocols, the volume of a tumor has been considered due to the fact that the
change in size (typically volume) of a tumor is what is used clinically to determine the
effectiveness of treatment. The first three images (set one) of the patient’s brain tumor
show the slices which include some signal from the tumor, when a slice thickness of 5
mm is used, which yields a volume for the tumor.

The second and third sets of images are those slice images that had signal from
the tumor when the slice thickness was set at 5.7 mm and 6.0 mm, which are the upper
limits for slice thickness accuracy for ACR and ICAMRL accreditation standards
respectively. These images not only show how different the volume of the tumor can be
calculated to be, but also the fact that all the signal from the tumor that was originally
present in the third slice with 5 mm slice thickness, has been removed from the slice. So
not only is the calculation for the volume of the tumor being drastically changed in the z-
direction, but the location of the edge of the tumor is also being effected, which could
lead to improper positioning of the therapy beam, or the plan for invasive removal of the
tumor.

The limits for slice thickness accuracy have far too large of an acceptable range
for both accreditation protocols, especially considering the sensitive nature of the
interpretation of tumor volume and edge location that slice thickness accuracy carries.
The images show how these large ranges allowed by both ACR and ICAMRL
accreditation protocols are not satisfactory standards, and have been the criticism of multiple published reviews of the accreditation process (8, 9, 10). Due to these facts and the large discrepancies created by poor slice thickness accuracy, much more stringent requirements should be implemented in both accreditation protocols to ensure proper diagnosis and treatment planning.

Due to the fact that the tumor of the patient in the scans performed was relatively symmetrical about the z-axis, the volume calculation for this patient is therefore primarily effected by the fact that the length of the tumor in the z-direction would be miscalculated by the same amount the slice thickness was varied from the prescribed slice thickness. Therefore, in this case the z-dimension of the tumor would be miscalculated 14% at ACR limits and 20% at ICAMRL limits, leading to an overall miscalculation in the tumor volume.

While this patient had a tumor with a similar cross-section throughout most of the length in the z-direction, the volume calculation would be affected even more for a patient with a tumor that was not as symmetrical about the z-axis. Typically when volume calculations are done to compare the volume of a tumor before and after a treatment, it is desired to image the tumor at similar locations. For this reason it is normal practice to place the first slice of a scan at the apex, or superior most point, of the tumor, and image with a certain slice thickness. When the scan is then repeated at a later time, after treatments have been performed; the scan is performed in the same fashion, with the first slice positioned at the apex of the tumor and the same slice thickness.

If the slice thickness accuracy would be disturbed in the machine between the first scan and the second scan, the location of the subsequent slices would be assumed in the
same location, while they were actually in different locations, off by 14% and 20% of the slice thickness for each subsequent slice. This repositioning of the slices in the tumor would also account for volume miscalculation by taking a cross-section at a location in the tumor different from where it is believed to be. The amount of volume miscalculation the mispositioning of these slices creates would depend on the shape of the tumor being imaged on a case by case basis.
Figure 31: Scan parameters, slice location, and field of view for SNR scan well within both accreditation protocols.
Figure 32: Scan parameters, slice location, and field of view for SNR scan slightly above lower limits of ACR accreditation standards.
Figure 33: Scan parameters, slice location, and field of view for SNR scan slightly above lower limits of ICAMRL accreditation standards.
Figure 34: Scan parameters, slice location, and field of view for spatial resolution scan well within both accreditation protocols.
Figure 35: Scan parameters, slice location, and field of view for spatial resolution scan slightly above lower limits of ACR accreditation standards.
Figure 36: Scan parameters, slice location, and field of view for spatial resolution scan slightly above lower limits of ICAMRL accreditation standards.
VI: REFERENCES


