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High-Resolution MRI for 3D Biomechanical Modeling: Signal Optimization Through RF Coil Design and MR Relaxometry

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ABSTRACT

High-Resolution MRI for 3D Biomechanical Modeling: Signal Optimization Through RF Coil Design and MR Relaxometry

James Badal
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Master of Science

Computed Tomography (CT) is often used for building 3D biomechanical models of human anatomy. This method exposes the subject to a significant x-ray dose and provides limited soft-tissue contrast. Magnetic Resonance Imaging (MRI) is a potential alternative to CT for this application, as MRI offers significantly better soft-tissue contrast and does not expose the subject to ionizing radiation. However, MRI requires long scan times to achieve 3D images at sufficient resolution, signal-to-noise ratio (SNR), and contrast-to-noise ratio (CNR). These long scan times can make subject motion a problem. This thesis describes my work to reduce scan time while achieving sufficient resolution, SNR, and CNR for 3D biomechanical modeling of (1) the human larynx, and (2) the human hip. I focused on two important strategies for reducing scan time and improving SNR and CNR: the design of RF coils optimized to detect MRI signals from the anatomy of interest, and the determination of MRI relaxation properties of the tissues being imaged (allowing optimization of imaging parameters to improve CNR between tissues).

Work on the larynx was done in collaboration with the Thomson group in Mechanical Engineering at BYU. To produce a high-resolution 3D image of the larynx, a 2-channel phased array was constructed. Eight different coil element designs were analyzed for use in the array, and one chosen that provided the highest Q-ratio while still meeting the mechanical constraints of the problem. The phased array was tested by imaging a pig larynx, a good substitute for the human larynx. Excellent image quality was achieved and MR relaxometry was then performed on tissues in the larynx.

The work on the hip was done in collaboration with the Anderson group in orthopedics at the University of Utah, who are building models of femoral acetabular impingement (FAI). Accurate imaging of hip cartilage requires injection of fluid into the hip joint capsule while in traction. To optimize contrast, MR relaxometry measurements were performed on saline, isovue, and lidocaine solutions (all typically injected into the hip). Our analysis showed that these substances actually should not be used for MR imaging of the hip, and alternate strategies should be explored as a result.

Keywords: biomechanical modeling, MRI, MRI coils, T1 relaxation, T2 relaxation, pig larynx, high resolution MRI images, MR relaxometry, human hip
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I would like to thank Josh Kaggie, Rock Hadley, and Rob Merrill who work at the MRI Coil Lab at the University of Utah for teaching me, facilitating my education with MRI coils, and answering my questions regarding MRI circuitry. I would especially like to thank Josh Kaggie for finishing the tune of the hip phased array. I would like to express my gratitude to the BYU MRI Research Group who helped me with my research, especially Danny Park, whom I was able to exchange and formulate ideas.

I would also like to thank Dr. Mazzeo and Dr. Long for their willingness to serve on my thesis committee, and the time and effort they put into reading my thesis, providing feedback, and helping me through the process.

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Chapter 1

Introduction

Biomechanical models are helpful in aiding our understanding of the function of organs and joints. These biomechanical models have traditionally been based on dissections of cadavers, and more recently through high-resolution 3D imaging using x-ray computed tomography (CT). However, CT exposes the subject to harmful ionizing radiation in the form of a significant x-ray dosages. Furthermore, soft tissue contrast in CT is limited. Recently, high-resolution magnetic resonance imaging (MRI) has begun to yield promising results for some biomechanical modeling. MRI yields excellent soft tissue contrast, allowing a richer differentiation between tissues compared to models based on CT. Unfortunately, there are still many challenges associated with obtaining high-resolution 3D images suitable for modeling some organs and joints. 3D MRI scans are long, and the signal-to-noise ratio achieved in the images is proportional to voxel volume (meaning higher resolutions yield lower SNRs) and the square root of the acquisition time. In areas of the body where motion is normal, it is often difficult to achieve adequate resolution with sufficient SNR in the short times required to mitigate the effects of motion.

The focus of my work over the past two years (and of this thesis) is the improvement of high-resolution 3D MRI techniques for two specific modeling applications: biomechanical modeling of the larynx, and biomechanical modeling of the hip. First, I have worked to boost SNR through better signal detection. This was done by designing improved MR coil topologies for detecting the NMR signal to each application. Second, I have measured MR relaxation properties of tissues and substances critical for each application, enabling optimization of pulse sequence acquisition parameters to boost SNR and contrast.
1.1 Motivation

1.1.1 Biomechanical Modeling of the Larynx

Dr. Thomson’s group in mechanical engineering at BYU is conducting research to improve understanding of human vocal fold function. Using high-resolution images of the larynx, Dr. Thomson’s team has been able to develop more accurate models of vocal fold vibration. These models can be used to study voice physics and to guide clinicians in their choice of treatment options as they attempt to restore or preserve voice quality in people suffering from vocal or speech disorders. The development of engineered tissue to replace damaged vocal fold tissues and the development of voice prostheses could benefit from this research. Post-operative functions and voice quality can be studied with new models in order to guide phonosurgical interventions. Dr. Thomson’s research could have an enormous impact on quality of life for those suffering from acute or chronic vocal and speech disorders.

Of fundamental importance to Dr. Thomson’s research are the biomedical images that feed models of laryngeal function. High spatial resolution in all three dimensions is desirable to provide accurate models of laryngeal anatomy. Furthermore, the ability to image at high temporal resolutions would facilitate the study of vocal fold dynamics, allow the capture of snapshots of the larynx during different vocal postures and voicing activities, and enable image acquisition during a single breath hold.

CT imaging can provide high spatial and temporal resolution images of the larynx, and has been the source for some of Dr. Thomson’s 3D models of the laryngeal respiratory airway. However, CT imaging provides very limited soft-tissue contrast for differentiating the component tissues of the larynx. Furthermore, CT exposes volunteers to significant x-ray dosages, and is becoming increasingly difficult to justify in pure research studies where the health benefit of heavy x-ray exposure is questionable.

Magnetic Resonance Imaging (MRI) is an appealing alternative to CT; it provides excellent soft tissue contrast and very flexible contrast mechanisms without any ionizing radiation. However, achieving adequate resolution (both spatial and temporal) is a much larger challenge in MRI. Conventional MR imaging methods, with the tools available on most commercial scanners, achieve images of only marginal utility for modeling of vocal function.
This is because commercial RF coils are designed to accommodate most patients, and are
not optimized for imaging the larynx and hip joint.

Part of this thesis is focused on the design of better RF coils for optimizing MR signal
detection from the human larynx, and measurement of the relaxation parameters ($T_1$ and
$T_2$) of tissues in the human larynx. Improved coils will allow for increased spatial resolution
and shortened scan times. Knowledge of the $T_1$ and $T_2$ relaxation parameters in tissues in
the larynx will allow optimization of MRI sequence acquisition parameters to achieve higher
signal levels and better contrast between tissues of interest.

1.1.2 Biomechanical Modeling of the Human Hip

Additional motivation for my work came from our collaboration with a group in
orthopedics at the University of Utah on biomechanical modeling of the human hip. Bony
hip pathologies such as femoral acetabular impingement (FAI) and hip dysplasia are believed
to be primary initiators of hip osteoarthritis (OA) in young adults. Surgical treatments that
seek to preserve the native hip joint have been developed to prevent or delay hip replacement
in patients with minimal joint damage. The success of hip preserving surgeries is highly
dependent on a comprehensive pre-operative assessment that identifies the extent of the bony
pathology and cartilage damage.

The Anderson group in orthopedics at the University of Utah has been using x-ray
computed tomography arthrography (CTA) to build high-resolution 3D biomechanical models
of the hip for pre-operative assessment of the hip joint. However, MR imaging would be
preferred for this application because of the improved soft-tissue contrast and absence of
damaging ionizing radiation. Unfortunately, most MR imaging of the hip to date has been
done with 2D multi-slice sequences that do not lend themselves to the construction of accurate
high-resolution 3D biomechanical models. One limitation to these high-resolution 3D images
in MRI is the difficulty achieving adequate SNR and contrast.

The second part of my work has focused on applying the same techniques I applied
to 3D modeling of the larynx to the hip. Specifically, I have performed MR relaxometry on
fluids that are typically present in the hip during imaging, yielding some very interesting
insights into the required MRI protocol for imaging the hip.
During hip imaging, the hip is kept under traction in order to increase the separation between cartilage layers and allow accurate modeling of cartilage surfaces on both the femur and the acetabulum (the hip socket). It is desirable to inject saline into the expanded joint space created by traction, as saline creates a very distinct border with cartilage in MRI images. However, both traction and the saline injection can be painful. Thus, it is typical for the orthopedists to inject lidocaine along with the saline into the subject’s hip to reduce pain from putting the hip joint under traction. In order to guide the injections, an x-ray contrast agent called isovue is also injected, and the injection performed under x-ray fluoroscopy (real-time x-ray imaging). The subject is taken to the MRI machine, with the fluids still in the joint. MR relaxometry was performed to understand the effects of the lidocaine and isovue on MR images.

1.2 Contributions

This thesis documents the work done to improve MRI images of the larynx and the hip. For the larynx project, MRI coils were optimized by comparing the Q-ratio values for eight different designs. Two different larynx coils, a single-loop surface coil and a 2-channel phased array, were built for the larynx. These custom coils were compared to an off-the-shelf Siemens’ neck coil to determine which coil configuration yields the best SNR and field of view. Both coil geometries that I designed and tested yielded improved SNR over the Siemens’ neck coil, although the field of view of the single loop coil was a bit limited. The 2-channel phased array coil yielded both superior SNR compared to the Siemens coil, and an adequate field of view for imaging the larynx. This coil was tested on an excised pig larynx, and I was able to achieve higher resolution images of the larynx than previously reported using MRI. High resolution relaxometry measurements were then performed on tissues in the pig larynx. These relaxometry values have not been previously reported in the literature, and are expected to enable optimization of tissue contrast in the larynx for building better 3D biomechanical models.

For the hip project, a dedicated unilateral 4-channel phased array hip coil was built that hugs the hip and positions the coil elements in close proximity to the femoral acetabular joint, achieving very good detection of the NMR signal from the hip joint. Experimental MRI
images of the hip showed that the mixture of lidocaine and isovue yielded an MR signal that was almost isointense as cartilage, which made it very difficult to distinguish cartilage from fluid when building the model of the joint. Thus, I performed MR relaxometry on different combination mixtures of saline, isovue, and lidocaine. A major contribution to this thesis came as a result of these measurements. The measurements indicated that good MR contrast between cartilage and these fluids was unlikely, as the $T_1$ and $T_2$ were too closely matched. Thus, the orthopedists are now exploring ultrasound-guided injection of saline into the joint, to avoid introduction of isovue or lidocaine into the joint during MR imaging.

1.3 Organization of the Thesis

Chapter 2 provides a brief overview of magnetic resonance physics, and the concepts needed to understand the remainder of my work,

Chapter 3 details the design and testing of the larynx coils, as well as the high-resolution images of the excised pig larynx achieved with the 2-channel phased array coil design,

Chapter 4 describes the $T_1$ and $T_2$ relaxometry measurements performed on the excised pig larynx,

Chapter 5 presents the relaxometry experiments performed on mixtures of saline, lidocaine, and isovue, and the significance of the results.
Chapter 2

MRI Physics Background

Magnetic Resonance Imaging (MRI) is a noninvasive imaging modality that can be used to generate images of the body. A very strong and uniform magnetic field is used to induce a net magnetic polarization in tissues in the body, which then allows the excitation and reception of a nuclear magnetic resonance signal from certain nuclei. Most clinical MRI machines create a static magnetic field of between 1.5 Tesla and 3 Tesla. Our experiments were performed on a 3 Tesla whole-body MRI scanner from Siemens Healthcare.

The following sections provides a brief introduction to MRI physics, and some of the basic terms and concepts in MRI.

2.1 Spins

The phenomenon of nuclear magnetic resonance (NMR) can only be observed in atoms that have an odd number of nucleons (protons and neutrons). In MRI, we loosely use the term “spin” to describe the nucleus of such an atom, as it has an associated spin angular momentum [1]. The concepts described below are a classical representation of MR physics, and are really only valid across large collections of spins. However, the concept of a spin is a useful classical representation, and is adopted for our explanation. We often diagram a spin as shown in Figure 2.1.

2.2 Excitation

When placed in a strong magnetic field, the nuclear spins tend to align either parallel or anti-parallel to the direction of the applied magnetic field. In thermal equilibrium, there is a slight preponderance of spins aligned parallel to the applied field (as opposed to anti-parallel). This produces a net magnetization across tissues in the body. These aligned spins can be
(figuratively) knocked out of thermal equilibrium by a radiofrequency (RF) pulse tuned to a certain resonant frequency called the Larmor frequency. This is referred to as “excitation”. We can think of excitation as “tipping” the spin out of alignment with the main polarizing magnetic field. The RF pulse needs to be polarized in a plane transverse to the direction of the main applied magnetic field (the polarizing field) in order to induce excitation.

The Larmor frequency is linearly proportional to the applied polarizing magnetic field, which we shall call $B_0$. The proportionality constant is dependent on the nucleus being excited. In clinical MRI, the hydrogen (1H) nucleus is excited. At 3 Tesla, the Larmor frequency of 1H is approximately 123 MHz. Our RF coil used for signal excitation must be tuned to transmit an RF pulse at this Larmor frequency.

More precisely, the Larmor frequency is defined as,

$$f = \frac{\gamma}{2\pi} B$$  \hspace{1cm} (2.1)

where $f$ is the Larmor frequency, $\gamma$ is nucleus-dependent constant called the gyromagnetic ratio, and $B$ is the applied polarizing magnetic field.
2.3 Orientation Conventions

To better understand the descriptions that follow, it is useful to adopt a standard coordinate frame of reference for an MRI machine. We define the z-axis as the direction of the main polarizing magnetic field $B_0$. In an MRI machine, this is the direction parallel to the bore of the scanner. We also call this the “longitudinal” direction, since it is oriented along the direction of $B_0$. This main polarizing field is produced by a superconducting magnetic solenoid.

The x-y plane is then referred to as the “transverse plane”. As described above, in order to excite a nuclear spin, we apply an RF pulse that is either linearly or circularly polarized in the transverse (x-y) plane.

This standard orientation is illustrated in Figure 2.2. Images acquired in the x-y plane are called “axial” images. Images acquired in the y-z plane are called “coronal” images. Images acquired in the x-z plane are called “sagittal” images.

2.4 Precession and Relaxation

Once excited (“tipped” out of alignment with the main magnetic field), a spin begins to “precess” around the z-axis at the Larmor frequency. This generates a faint RF signal that is circularly polarized in the x-y, or transverse, plane. This faint RF signal, at the Larmor frequency, is the NMR signal. Precession continues until the spin returns to thermal equilibrium and is again aligned in the z-direction (the longitudinal direction), with no remaining transverse component.

The magnitude of the NMR signal is proportional to the projection of the spin onto the transverse (x-y) plane after it is tipped, or excited. After excitation, the transverse component of the spin decays exponentially with a time constant $T_2$. Thus, the observed NMR signal from any excited spin decays exponentially with time constant $T_2$. We call this transverse decay process “$T_2$ relaxation”. (Note that we are ignoring other effects that might cause the signal from a collection of spins to decay more rapidly due to dephasing of the spins.) $T_2$ is tissue dependent. Differences in $T_2$ between tissues and other substances being imaged can be used to generate contrast in MR images.
Figure 2.2: MRI image coordinate system

$T_2$ relaxation can be modeled as a decaying exponential function. Assuming a 90-degree excitation, the decaying exponential function is modeled by

$$M_{xy} = M_0 e^{-\frac{t}{T_2}}$$  \hfill (2.2)

where $M_{xy}$ is the magnetization in the transverse plane, $M_0$ is the magnetization before excitation, $t$ is the time since excitation, and $T_2$ is the transverse relaxation time constant. $T_2$ values for some common tissues are found in Table 2.1 [2].

In addition to $T_2$ relaxation, the projection of the spin onto the longitudinal axis “recovers” exponentially, growing back to its magnitude at thermal equilibrium. This process is termed “$T_1$ recovery”. $T_1$ is also tissue dependent, and represents, loosely speaking, the amount of time we have to wait until we can tip or excite the spin again. Mathematically, $T_1$ recovery can be expressed as a recovering exponential function. Looking at a sample spin
having magnetization $M$ (and z magnetization component $M_z$) the longitudinal relaxation for a 90-degree excitation can be modeled as:

$$M_z = M_o \left( 1 - e^{-\frac{t}{T_1}} \right) \quad (2.3)$$

where $M_o$ is the initial magnetization before excitation, $t$ is time since excitation and $T_1$ is the longitudinal recovery time constant. In biological tissues, $T_1$ is always greater than $T_2$. Some $T_1$ values for common tissues are found in Table 2.0 [2].

<table>
<thead>
<tr>
<th>Tissue Name</th>
<th>$T_1$ (ms)</th>
<th>$T_2$ (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Skeletal muscle</td>
<td>1412</td>
<td>50</td>
</tr>
<tr>
<td>Heart</td>
<td>1471</td>
<td>47</td>
</tr>
<tr>
<td>Kidney</td>
<td>1194</td>
<td>56</td>
</tr>
<tr>
<td>Cartilage*</td>
<td>1168</td>
<td>27</td>
</tr>
<tr>
<td>White matter</td>
<td>1084</td>
<td>69</td>
</tr>
<tr>
<td>Gray matter</td>
<td>1820</td>
<td>99</td>
</tr>
<tr>
<td>Optic nerve</td>
<td>1083</td>
<td>78</td>
</tr>
<tr>
<td>Blood</td>
<td>1932</td>
<td>275</td>
</tr>
</tbody>
</table>

*Relaxometry characteristics can vary depending on orientation to $B_0$ field

MR relaxometry is the study of measuring and calculating $T_1$ and $T_2$ values. MR relaxometry is important, as knowledge of the $T_1$ and $T_2$ values of various tissues allows optimization of MR pulse sequence acquisition parameters.

### 2.5 Bloch’s Equation

The processes of precession, $T_2$ relaxation, and $T_1$ recovery are modeled by a phenomenological differential equation. The time evolution of a magnetic moment, or spin, $(M)$ is given by:

$$\frac{dM}{dt} = M \times \gamma B - \frac{M_x i + M_y j}{T_2} - \frac{(M_z - M_o) k}{T_1} \quad (2.4)$$
where $\mathbf{M}$ is the magnetization of the spin, $\mathbf{B}$ is the applied magnetic field on the spin, $\gamma$ is the gyromagnetic ratio, $M_x, M_y, M_z$ are the magnetization of the spin in the corresponding axis, $M_o$ is the initial magnetization before excitation, $T_1$ is the longitudinal relaxation time constant, and $T_2$ is the transverse relaxation time constant [3, 4].

### 2.6 Receiving the NMR Signal

Faraday’s law of induction states that an oscillating magnetic field will generate electromotive force (EMF) in an electric circuit. When a spin is excited it produces a small rotating magnetic field that can be received through an RF antenna or coil.

$T_2$ (transverse) relaxation determines how quickly the signal decays, and thus dictates how long the signal can be effectively sampled. Since the NMR signal is circularly polarized in the transverse plane, we require RF coils that are sensitive to such signals to detect it. Single-loop coils or phased arrays of single-loop coils are often used to detect the faint NMR signal. These loops need to be oriented such that the normal to the plane of the loop lies in the transverse plane. The MR signal can only be sampled in the transverse plane because the MRI coils can only receive an oscillating signal. A major focus of my work has been on designing, building, tuning, and testing these receive coils for better NMR signal reception for specific applications.

### 2.7 Inversion Recovery and Spin Echo Pulse Sequence

A pulse sequence is a program that controls the MRI hardware. Each pulse sequence has a repetition time (TR) and a echo time (TE). TR denotes when the sequence can be repeated to sample the NMR signal. TE determines when the NMR signal can be sampled. To understand how the $T_1$ and $T_2$ properties are measured, a review of the inversion recovery and spin echo pulse sequences is given in this section.

To measure the $T_1$ property of materials, the inversion recovery pulse sequence is used. To understand the basics of the inversion recovery pulse sequence, we will only look at one spin with a magnetization of $M_o$. The pulse sequence begins with an excitation pulse that flips the magnetic moment by 180-degrees. After the excitation pulse, the spin will relax back to the equilibrium state. After some time, inversion time (TI), a 90-degree excitation pulse is
used to tip the remaining magnetic moment to the transverse plane where the signal can be sampled. The magnetic moment measurement from this pulse can be used to calculate the $T_1$ property of the material by applying a nonlinear fit to equation

$$M_z = M_0 \left( 1 - 2e^{-\frac{t}{T_1}} \right).$$  \hspace{1cm} (2.5)

It should be noted that equation 2.5 and equation 2.3 are different because equation 2.3 assumes a 90-degree excitation pulse and equation above assumes a 180-degree pulse.

The spin echo sequence is used to measure the $T_2$ properties of materials. It uses a 90-degree excitation pulse to tip the magnetic moments to the transverse plane. The spins will dephase in the transverse plane due to field inhomogeneity and chemical shift [4]. To solve this problem a 180-degree inversion pulse is use to “pancake flip” the magnetic moments. The magnetic moment will then dephase back to the original position on the transverse plane were the signal can be sampled. The magnetic moment measurement can be used to calculate the $T_2$ property of the material by applying nonlinear fit to Equation 2.2.
Chapter 3

RF Coils for Imaging of the Larynx

This chapter details the design, construction, and testing of coils for high-resolution 3D imaging of the larynx. Eight different surface coil designs were considered for the larynx coils. Each design varied depending on wire gauge and number of capacitors used and was compared with the Q-ratio metric, a measurement determining whether the coil was circuit noise dominated or sample noise dominated. The surface coil design with the best flexibility and Q-ratio was used to build a single-loop surface coil and a 2-channel phased array. In addition, these coils were compared to an off-the-shelf Siemens neck coil. Both coils achieved a much higher focal SNR than the Siemens neck coil over the larynx, but the 2-channel phased array performed better than the single-loop surface coil and achieved a better coverage of the larynx. The 2-channel phased array was then used to image a pig larynx with an extremely high resolution of 0.2 x 0.2 x 0.9 mm voxel sizes.

3.1 Methods

3.1.1 Coil Design

Surface coils are circular or square MRI antennas that can be used for detecting the NMR signal (and in some cases, exciting the signal). A surface coil that can excite and detect the NMR signal is called a transmit receive surface coil. In order to receive the NMR signal, the MRI surface coils must be tuned to the Larmor frequency of the desired nucleus. The resonance frequency of a surface coil is given by:

$$\omega = \frac{1}{\sqrt{LC}}$$

(3.1)
where $\omega$ is the resonance frequency in radians/s, $L$ is the coil’s total inductance, and $C$ is the coil’s total capacitance. The coil’s total capacitance is separated into two different capacitors in the MRI coil, the coupling capacitor and the tuning capacitor. The coupling capacitor $C_c$ matches the coil to the coaxial cable and the tuning capacitor $C_T$ determines the coil’s frequency. The orientation of the coupling capacitor and the tuning capacitor can be seen on Figure 3.1.

![Figure 3.1: Transmit Receive Surface Coil Schematic](image)

The tuning capacitance can be separated across multiple capacitors to reduce stray electric fields from penetrating the object being imaged, which is detrimental to coil performance. However, increasing the number of capacitors will increase the resistance of the surface coil because of additional soldering and copper plates needed.

The size of the coil determines its inductance, and is related to the focal SNR the coil can achieve. The sensitivity of the coil is clearly related to the distance of the coil from the object producing the NMR signal. The maximum SNR is achieved at a distance of 0.25x radius to 0.7x the radius.[5]

For a receive surface coil, detuning circuitry is needed so the circuit is not resonant during excitation. For receive-only coils, signal excitation is produced by a large RF coil built into the bore of the MRI scanner called the “body coil”. The detuning circuitry is shown in Figure 3.2. When a DC bias is applied to the circuit the diode turns on and applies a short to the cable. This changes the tuning of the coil so that it does not resonate at the Larmor frequency. The bias is usually applied during the transmit phase so the receive coil does not
resonate during excitation. If it did, the coil would absorb RF energy, and could become hot enough to potentially cause the subject to burn.

![Receive Surface Coil Schematic](image)

**Figure 3.2:** Receive Surface Coil Schematic

To improve both the field of view of a surface coil and its overall sensitivity across the field of view, a phased array of surface coils can be constructed. For a two channel phased array the SNR improves by a factor of $\sqrt{2}$ [6]. The main problem with building phased arrays is that surface coils tuned to the same frequency couple when they are near to each other due to the mutual inductance between the coils.

To reduce mutual inductance between surfaces coils, two methods are utilized: (1) overlapping and (2) preamp decoupling. Overlapping reduces the shared inductance between adjacent coils. For surface coils that are placed on a flat surface the optimal overlap from center to center is 0.75 times the diameter of the coils [7]. If the coils are placed on a slightly curved surface, the optimal overlap is around the same distance. Preamp decoupling reduces the current flow between coils by reducing the mutual inductance between coils. Preamp decoupling is mainly used to reduce mutual inductance between nonadjacent coils in the phased array.

### 3.1.2 Q-ratio Test

**Quality Factor**

Quality factor, or Q-factor, is a measurement of how well an RLC electric circuit contains an oscillating signal during resonance. Looking at an RLC circuit at the resonance frequency, the definition of Q-factor is:
\[ Q = \frac{\omega L}{r} \] (3.2)

where \( \omega \) is the frequency of interest in radians, \( L \) is the inductance of the circuit, and \( r \) is the resistance of the circuit [8]. Measuring the resistance and the inductance of the circuit can be difficult, so the Q-factor is determined by using this alternative definition:

\[ Q = \frac{f}{\Delta f} \] (3.3)

where \( f \) is the frequency of interest in Hertz and \( \Delta f \) is the half-power bandwidth. The value of \( \Delta f \) is determined by measuring the -3 dB bandwidth around the frequency of interest.

To compare the Q-factor of different coil designs would be difficult because these values can vary depending on the inductance of the coil. To improve the comparison between coils the ratio of \( Q_{\text{loaded}} \) and \( Q_{\text{unloaded}} \) are used, which is called the Q-ratio. To understand Q-ratio for MRI coils, the assumption is made that the circuit resistance and the electric and magnetic field loss through a sample load are the only resistances affecting the MRI coil.

Using the definition of Q given by equation (3.2), let \( Q_{\text{loaded}} \) and \( Q_{\text{unloaded}} \) be defined as:

\[ Q_{\text{loaded}} = \frac{\omega L}{r_{\text{circuit}} + r_{\text{sample}}} \] (3.4)

\[ Q_{\text{unloaded}} = \frac{\omega L}{r_{\text{circuit}}} \] (3.5)

The Q-ratio is then defined as the ratio between \( Q_{\text{unloaded}} \) and \( Q_{\text{loaded}} \), i.e.,

\[ Q_{\text{ratio}} = \frac{Q_{\text{unloaded}}}{Q_{\text{loaded}}} = \frac{r_{\text{circuit}} + r_{\text{sample}}}{r_{\text{circuit}}} \] (3.6)

Referring to Equation (3.6), a higher Q-ratio means that the coil noise is more dominated by electrical noise from the sample being imaged than noise from the circuit itself. Sample-noise-dominated coils are desirable for MR imaging.
Q-ratio Test

There are multiple considerations in building a MRI surface coil. Surface coil designs can vary in size, wire gauge, detuning circuitry, and the number of capacitors used as tuning capacitors. To reduce the complexity of the coil design comparison, each surface coil design was built the same size (5.2 cm) and included the same detuning circuitry. Eight designs were considered in this thesis with each design varying in wire gauge and number of capacitors used as tuning capacitors. The design that had the highest Q-ratio and flexibility was used for the larynx coils.

Each coil was tuned to the Larmor frequency of 1H at 3 Tesla and at least measured a -30 dB. The Q-value was measured with a double loop probe that measured the resonance of the coils through $S_{21}$, a transmission coefficient. The cables of each coil were removed to prevent any interference with the double loop probe.

The coils were mounted on a plastic plate which was then placed on a stand to maintain equal distance for all measurements. $Q_{\text{loaded}}$ was first measured using a phantom (a bottle of doped water) to load the coil. The phantom was then removed to measure $Q_{\text{unloaded}}$. Each coil was loaded by the same uniform phantom to simulate loading from a human subject. The distance from the coil to the phantom was 6.5mm and the distance between the double loop probe and the coil was 3.2cm. A picture of the set-up is shown in Figure 3.3.

![Set-up for measuring $Q_{\text{unloaded}}$](image1.jpg) ![Set-up for measuring $Q_{\text{unloaded}}$](image2.jpg)

**Figure 3.3:** Q-ratio Test Set-Up
3.1.3 MRI Coil Comparison Test

The design of the surface coil that was used for the larynx coil had 18-gauge wire and two tuning capacitors. This design was used to construct a single loop larynx coil and a 2-channel larynx coil. These two custom larynx coils were then compared to the Siemens off-the-shelf neck coil.

A simple spin echo sequence was used to compare SNR and field of view of the coils. The scan parameters for the spin echo sequence were as follows: 0.69 mm x 0.59 mm voxel size, base resolution 256 x 218, slice thickness of 5.0 mm, 1 average, the corresponding TR of 1500, TE of 25 ms TE and scan time of 4:51 (minutes : seconds).

The Siemens uniform phantom used for loading the coils for the Q-ratio test was also used in this comparison test. The phantom was slightly tilted to prevent the bubble inside the phantom from showing in the picture. For the neck coil and the 2 channel phased array, constituent images from each coil were combined using the root sum of squares approach.

Exploratory scans of the human larynx determined that the field of view required for good imaging of the human larynx in vivo was 1.5 cm to 3.2 cm in depth, and approximately 6 x 6 cm in the other directions. This field of view allows full visualization of the vocal fold from front to back.

3.1.4 MRI Image of the Pig Larynx

A pig larynx was procured from a local butcher. The larynx was cleaned of unnecessary tissue and mounted onto a wax base for support. A 2% agarose solution was prepared, and the larynx and base inserted into a plastic container. The agarose solution was used to completely surround the larynx, thus reducing any susceptibility-induced variations in magnetic field when the larynx was imaged. The agarose solution was initially heated to facilitate the dissolving of the agarose into a solution, then cooled to room temperature before being used. This prevented any damage or change to the larynx. A major concern was that a heated sample could partially or fully cook the Larynx, thus changing the molecular structure of the larynx and the $T_1$ and $T_2$ properties of tissues.

The agarose solution was placed first in the beaker and the pig larynx was lowered into the agarose prior to scanning. Agarose was also poured into the esophagus of the larynx...
to reduce susceptibility between air and tissues. It was then necessary to support the larynx
in the agarose solution until the agarose solution solidified. A picture of the prepared pig
larynx phantom is shown in Figure 3.4.

A spin echo pulse sequence was used to image the pig larynx. The scan parameters
for the spin echo sequence were as follows: coronal plane, 0.2 mm x 0.2 mm voxel size, base
resolution 512 x 512, slice thickness of 0.9 mm, 1 average, the corresponding TR of 1500 ms,
TE of 45 ms and scan time of 11:29 (minutes : seconds).

3.2 Results

3.2.1 Q-ratio Results

The results of the Q-ratio test are shown in Table 3.1. The coil using 14-gauge wire
had the highest Q-Ratio design as the evidence suggests. Building a 14-gauge surface coil
phased array on a curved surface proved to be very challenging because 14-gauge wire is
not very flexible. Thus, we ended up opting for 18-gauge wire, which afforded the necessary
flexibility while still achieving an acceptable Q-Ratio.

Note that the Q-Ratio values varied depending on the amount of solder and the
soldering quality of the coil. Prior to removing excess solder from the circuit the Q-Ratio for
Table 3.1: Q-ratio Test Results

<table>
<thead>
<tr>
<th>Wire Gauge</th>
<th>Number of capacitors</th>
<th>Q-ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>14</td>
<td>3</td>
<td>4.3279</td>
</tr>
<tr>
<td>16</td>
<td>3</td>
<td>3.3991</td>
</tr>
<tr>
<td>16</td>
<td>4</td>
<td>3.5303</td>
</tr>
<tr>
<td>18</td>
<td>3</td>
<td>3.7769</td>
</tr>
<tr>
<td>18</td>
<td>4</td>
<td>3.2800</td>
</tr>
<tr>
<td>20</td>
<td>3</td>
<td>3.2163</td>
</tr>
<tr>
<td>20</td>
<td>4</td>
<td>3.4151</td>
</tr>
<tr>
<td>22</td>
<td>3</td>
<td>2.8269</td>
</tr>
</tbody>
</table>

A 20-gauge 3 capacitor coil was 2.9858. Once excess solder was removed the Q-Ratio jumped by 0.2.

3.2.2 MRI Coil Comparison Results

MRI images using the surface coil, 2 channel phased array, and neck coil can be seen in figure 3.4. The region of interest for each coil is 1.5 cm to 3.2 cm into the sample. A histogram plot comparing the SNR of each voxel in this region of interest of the single-loop surface coil to the neck coil is shown in Figure 3.6. Figure 3.7 shows a histogram plot comparing the 2-channel phased array to the neck coil, and Figure 3.8 shows a histogram plot comparing the single-loop surface coil to the 2-channel phased array.

The Siemens off-the-shelf neck coil does provided more uniform SNR throughout the sample compared to the single surface coil and the phased array. However, the 2 channel phased array did provided the highest focal SNR in the region of interest, and still achieved adequate coverage of the entire region. These results suggested that the 2-channel phased array coil would probably yield the best images, and this coil was used in subsequent experiments.

3.2.3 MRI Image of the Pig Larynx

The scan of the pig larynx produced 22 slices (images) of the pig larynx. Figure 3.9 shows one of slices that contains pertinent larynx tissues. The other slices can be seen in Appendix A. The tissues that make up the vocal folds can be clearly visualized. A remarkable 0.2 mm in-plane resolution, with slice thickness of only 0.9 mm can been seen in these MRI
Figure 3.5: MRI Image Comparison between a Single-Loop Surface Coil, 2 Channel Phased Array, and Neck Coil

Figure 3.6: Histogram Plot Comparing Siemens’ Neck Coil and Surface Coil Voxel SNR in the Region of Interest
Figure 3.7: Histogram Plot Comparing Siemen’s Neck Coil and 2-Channel Phased Array Voxel SNR in the Region of Interest

Figure 3.8: Histogram Plot Comparing Single-Loop Surface Coil and 2-Channel Phased Array Voxel SNR in the Region of Interest
images. These images are of better quality than anything reported in the literature, and provide a good starting point for producing high SNR, high resolution images of the human larynx in vivo.

Figure 3.9: Axial Slice of a Pig Larynx

3.3 Discussion

The purpose of this chapter was to provide a high-resolution image of a pig larynx through improvement of custom MRI coils. Eight MRI surface coil designs were tested and compared using the MRI surface coil’s Q-ratio. The design consisting of a 14-gauge wire and
3 capacitors had the highest Q-ratio but curving the surface coil for tuning and placement unto the MRI coil support structure proved difficult. The design that had the second highest Q-ratio was the one that utilizes the 18-gauge wire and 3 capacitors. This design was used to build a single-loop surface coil and the 2-channel phased array.

The single-loop surface coil and 2-channel phased array was compared to a stock Siemen’s neck coil to determine which coil would provide the highest SNR in the region of interest. The results in this chapter showed that the 2-channel phased array would be the best coil to use to image an excised pig larynx because it achieved excellent SNR over an adequate field of view for high resolution imaging of the larynx. These images of a pig larynx were be used to measure the relaxation characteristics of tissues inside the larynx. This is noted in the next chapter.

Future work includes exploring the improvement of a 3-channel phased array can make on image homogeneity across the larynx. This phased array can improve the SNR and homogeneity of image, which will allow shorter scan times while achieving a descent image. The phased array that provides the highest SNR will be used to image an in vivo scan of a human larynx.
Chapter 4

Relaxometry of Tissues in Pig Larynx Model

Once the larynx coil was built and tested as described in chapter 3, the coil was ready for ex vivo images of a larynx. Due to availability and its close similarity to the human larynx in shape and size a pig larynx was chosen to perform the MR Relaxometry of the larynx tissues. The goal of the experiments described below was to make accurate measurements of $T_1$ and $T_2$ of various tissues in the larynx. These values allow us to optimize MR pulse sequence parameters for better SNR and differentiation of tissues of interest in the larynx. While pulse sequence optimization is beyond the scope of this thesis, the relaxation measurements will lay the groundwork for future optimization work.

The relaxation properties of laryngeal tissues have not been previously calculated, possibly because the tissues that make up the vocal folds are only 50 µm to 7.3mm in thickness [9]. It is difficult to achieve adequate SNR to measure the relaxation properties of structures this small. However, our coil geometry and pig model of the larynx enabled the measurements to be performed.

4.1 Methods

In order to measure the $T_1$ properties of the pig larynx an inversion recovery MR sequence was used to obtain these measurements. The scan parameters for the sequence were as follows: 0.2 mm x 0.2 mm voxel size, base resolution of 512 x 512, slice thickness of 1.5 mm, 1 average, repetition time (TR) of 5000 ms, echo time (TE) of 5.23 ms and scan time of 1:11:40 (hours : minutes : seconds). The inversion time (TI) value of 50 ms, 300 ms, 700 ms, 1200 ms, and 2000 ms were used to measure the $T_1$ properties of the pig larynx.
The scan parameters for measuring the $T_2$ properties of the pig larynx is as follows: 0.2 mm x 0.2 mm voxel size, base resolution of 512 x 512, slice thickness of 0.9 mm, 1 average, the corresponding TR, TE and scan time are shown in Table 4.1

<table>
<thead>
<tr>
<th>TE (ms)</th>
<th>TR (ms)</th>
<th>Scan Time (min:sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>17</td>
<td>1500</td>
<td>11:29</td>
</tr>
<tr>
<td>30</td>
<td>1500</td>
<td>11:29</td>
</tr>
<tr>
<td>45</td>
<td>1500</td>
<td>11:29</td>
</tr>
<tr>
<td>75</td>
<td>1830</td>
<td>14:00</td>
</tr>
<tr>
<td>90</td>
<td>2160</td>
<td>16:29</td>
</tr>
<tr>
<td>120</td>
<td>2820</td>
<td>21:30</td>
</tr>
<tr>
<td>200</td>
<td>4580</td>
<td>34:55</td>
</tr>
</tbody>
</table>

4.2 Results

The various tissues of the larynx were classified and labeled in Figure 4.1. The $T_1$ and $T_2$ values of the pig larynx tissues are given in Table 4.2. The relaxometry properties of Cricoid Lamina, which is cartilage in the larynx, did fall in the range of cartilage relaxometry properties. The characteristics of the vocal fold anatomy are made of 3 layers of muscle and mucosa [10]. The $T_1$ characterisitcs of the vocal folds are similar to cartilage but $T_2$ characteristics are similiar to skeletal muscle (1412 ms).

<table>
<thead>
<tr>
<th>Tissue</th>
<th>$T_1$ (ms)</th>
<th>$T_2$ (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mucosa</td>
<td>1266.6 +/- 57.8</td>
<td>110.2 +/- 16.3</td>
</tr>
<tr>
<td>Comus Elasticus</td>
<td>1318.8 +/- 59.9</td>
<td>79.3 +/- 9.8</td>
</tr>
<tr>
<td>Cricoid Lamina (Cartilage)</td>
<td>1023 +/- 74.6</td>
<td>33.4 +/- 8.2</td>
</tr>
<tr>
<td>Posterior Cricoarytenoid</td>
<td>965.9 +/- 39.2</td>
<td>33.5 +/- 5.4</td>
</tr>
<tr>
<td>Vocal Folds</td>
<td>1130.5 +/- 128.2</td>
<td>59.0 +/- 7.0</td>
</tr>
<tr>
<td>Thyroide Cartilage</td>
<td>991.2 +/- 13.5</td>
<td>67.7 +/- 8.7</td>
</tr>
</tbody>
</table>
4.3 Discussion

The larynx coil constructed in this thesis provided high resolution images of an excised pig larynx. One of these images was used to calculate the $T_1$ and $T_2$ characteristics of tissues inside the pig larynx due to the fine detail image provided. The Cricoid Lamina, which consist of cartilage, has a relaxation values that are similar to the cartilage’s relaxation values. These results helps bring validity to the relaxation values even though some of the $T_1$ and $T_2$ values do have a moderate standard deviation. The standard deviation can be improved by reducing the noise in the image and excluding any voxels that contain two or more tissues from future relaxation calculation.

The pig larynx tissue relaxometry values can be used to design pulse sequences that can give better contrast between tissues. These images could be used to better understand the muscles and cartilages inside the larynx. After fine tuning this pulse sequence, rapid imaging techniques can be used to reduce the scan time while sustaining good SNR value. By reducing the scan time, an in vivo image of a human larynx with good SNR can be imaged in a single breath hold, thereby reducing any distortion caused by breathing or reflex motion. With a high resolution MRI image of larynx, Dr. Thomson’s research group can use these MRI images to better design the biomechanical model of a larynx.
Future work should include measuring the epithelium, which has a thickness of 50 µm. In order to produce a higher resolution image that can capture the epithelium, the image SNR needs to be improved. The approach that can be used to improve the SNR of the MRI image is to build a 3-channel phased array and increasing the averages in the pulse sequence.
Chapter 5

Relaxometry of Fluids Injected in the Hip

Bony hip pathologies are believed to be primary initiators of hip osteoarthritis in young adults. It is believed that better modeling of the hip can help with pre-operative assessment of the hip which can assist with hip surgery. Orthopedists at the University of Utah are exploring CT and MRI to produce a biomedical model of the hip. To produce images of the hip, the patient’s leg is put under traction before being imaged by a CT and MRI scanner. To better visualize the cartilage in the hip, the contrast agents isovue and saline are used for CT and MRI respectively. To help with pain caused by the traction of the leg, lidocaine is administer to the patient through ejection.

Initial experimental MRI images from these procedures showed that a mixture of lidocaine, saline, and isovue yielded an MR signal that was almost as isointense as cartilage. In essence, this made distinguishing the cartilage and fluid difficult. To determine the cause of the problem the $T_1$ and $T_2$ values from different solutions of saline, isovue, and lidocaine were tested. This chapter will cover the experiment that measured the MR relaxometry of these solutions.

5.1 Methods

Six different 4.2mL vials of varying concentrations of saline, isovue, and lidocaine were produced to determine which solution was producing the same relaxometry characteristics as cartilage. The concentrations of saline and isovue varied from 100% saline to 50% saline and 50 % isovue. These vials were epoxied to a plastic cylinder to give the vials support during scanning. The vials were placed in a container which was filled with distilled water to reduce image distortion by reducing magnetic field variations. Figure 5.1 shows pictures of the vials inside the container which is referred to as a phantom.
After producing the phantom, the $T_1$ and $T_2$ values of the solutions were measured. The inversion recovery sequence was used to measure the $T_1$ relaxation constant for saline, isovue, and lidocaine solutions. The scan parameters for the sequence were: 2.05 mm x 2.74 mm voxel size, a base resolution of 256 x 256, a slice thickness of 5 mm, 1 average, TR of 5000 ms, TE of 4.15 ms, and a scan time of 10:45 (minutes: seconds). The TI values of 100 ms, 300 ms, 900 ms, 1500 ms and 3000 ms were used.

The inversion recovery sequence was used to measure the $T_1$ relaxation constant for saline, isovue, and lidocaine solutions. The scan parameters for the sequence are: 2.05 mm x 2.74 mm in voxel size, base resolution 256 x 256, slice thickness of 5 mm, 1 average, TR of 5000 ms, TE of 4.15 ms, and a scan time of 10:45. The TI values of 100 ms, 300 ms, 900 ms, 1500 ms and 3000 ms were used.

The turbo spin echo pulse sequence, a spin echo sequence that accelerates image acquisition, was used to measure $T_2$ characteristics with these parameters: 2.05 mm x 2.74 mm voxel size, base resolution 256 x 256, slice thickness of 5 mm, 1 average, TR of 5000, and a scan time of 21:00. The TE that were measured was 15 ms, 59 ms, 117 ms, 161 ms, 220 ms, and 264 ms.

The magnitude images of the scans were used to calculate $T_1$ and $T_2$ constants. Between 50 and 144 pixels were averaged to increase the SNR of the signal and the $T_1$ and $T_2$ constants were calculated using those values.
To measure the effects of lidocaine on the isovue and saline solution, lidocaine was injected into two vials due to a lack of lidocaine available for this experiment. Lidocaine was added to the 100 % saline vial and the 50 % saline/50 % isovue vial. The first measurement was taken after replacing 2.1 mL of the solution with lidocaine. The extra liquids from the vials were stored for the second measurement. The second measurement was taken after replacing 2.1mL of the solution with the previously stored solutions. The same sequences that were used to measured \( T_1 \) and \( T_2 \) were then used to measure the effects of lidocaine on those two solutions.

5.2 Results

Contrast in MRI images is heavily influenced by differences in the relaxation parameters \( T_1 \) and \( T_2 \). While we can choose different pulse sequences to preferentially bring out contrast based on one or the other of these parameters (i.e., get more \( T_1 \) weighting or more \( T_2 \) weighting), many of the rapid high-resolution 3D imaging techniques suitable for the creation of accurate high-resolution 3D volumetric models yield a contrast that is a strong non-linear function of \( T_2/T_1 \).

<table>
<thead>
<tr>
<th>Solution</th>
<th>( T_1 ) value (ms)</th>
<th>( T_2 ) value (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>100% Saline</td>
<td>1465</td>
<td>1233</td>
</tr>
<tr>
<td>90% Saline 10% Isovue</td>
<td>1458</td>
<td>593</td>
</tr>
<tr>
<td>80% Saline 20% Isovue</td>
<td>1431</td>
<td>469</td>
</tr>
<tr>
<td>70% Saline 30% Isovue</td>
<td>1409</td>
<td>315</td>
</tr>
<tr>
<td>60% Saline 40% Isovue</td>
<td>1362</td>
<td>192</td>
</tr>
<tr>
<td>50% Saline 50% Isovue</td>
<td>1347</td>
<td>189</td>
</tr>
<tr>
<td>75% Saline 25% Lidocaine</td>
<td>1426</td>
<td>595</td>
</tr>
<tr>
<td>37.5% Saline 37.5% Isovue 25% Lidocaine</td>
<td>1398</td>
<td>254</td>
</tr>
<tr>
<td>50% Saline 50% Lidocaine</td>
<td>1418</td>
<td>405</td>
</tr>
<tr>
<td>25% Saline 25% Isovue 50% Lidocaine</td>
<td>1406</td>
<td>209</td>
</tr>
</tbody>
</table>

Table 5.1: \( T_1 \) and \( T_2 \) Values for Different Levels of Concentration of Saline, Isovue, and Lidocaine

At 3 Tesla, the field strength of our imaging experiments, cartilage has a \( T_1 \) of 1100 ms and a \( T_2 \) of 40 ms. As measured and shown in the table, pure saline has a \( T_1 \) of 1500
ms and a $T_2$ of 1200 ms. Thus, we get truly excellent contrast between cartilage and pure saline in a $T_2/T_1$ weighted sequence, due to the very different ratios of $T_2/T_1$. However, as lidocaine and/or isovue are added to the saline, the $T_2$ value drops dramatically, with very little effect on $T_1$. $T_2$ was cut at least in half relative to pure saline in the concentrations that we explored. This has a dramatic effect on contrast, pushing the $T_2/T_1$ ratios of the fluid closer to that of cartilage, and reducing the ability to delineate cartilage from fluid in the joint capsule. This has a profound impact on the ability to produce an accurate 3D model of the joint when saline mixed with either isovue or lidocaine is present in the joint capsule.

5.3 Discussion

As outlined, the results of the experiment showed that adding isovue and lidocaine to saline does indeed change the values of both $T_1$ and $T_2$ in the solution, but the relative change of $T_1$ is very small, while the relative change of $T_2$ is very large. These kinds of mixtures were used to image the hip because (1) the isovue provides good delineation of the fluid under x-ray fluoroscopy (for guiding the needle in the injection of the hip) and under CT (which was previously used to acquire the 3D images), and (2) because the lidocaine was helpful in making the injection and subsequent traction placed on the hip less painful for the subject. When we moved to MRI for trying to do these high resolution images, the injection still needed to be guided by x-ray fluoroscopy, and so the isovue was still used. The lidocaine was still needed for pain.

However, under these circumstances, our preliminary scans showed very little difference in the signal levels between fluid in the joint and the cartilage, making segmentation and modeling of the cartilage very difficult from the scans. Our relaxometry results (measurement of $T_1$ and $T_2$ in the different fluid mixtures) explain exactly why this was happening.

As a direct result of the work presented here, the orthopedists are exploring alternate approaches for the introduction of fluid into the joint capsule during MRI imaging. One option that has been tested on cadavers is the injection of pure saline, where ultrasound is used to guide the injection into the joint capsule rather than x-ray fluoroscopy. While more difficult for the orthopedist, this doesn’t require a mixture of isovue with the saline. The lidocaine is also being left out of the injection. While this may reduce subject comfort during
these exams, it is felt that most subjects should be able to tolerate the discomfort with a saline-only injection.
Chapter 6

Conclusion

6.1 Summary

6.1.1 Biomechanical Modeling of the Larynx

In summary, the motivation for this project came from collaboration with Dr. Thomson’s group to optimize the high-resolution 3D image of a human larynx. To produce a high resolution 3D image of the larynx a surface coil and a 2-channel phased array was constructed. The design of these custom MRI coils were determined by the Q-ratio of eight different designs. Of the eight, the design with the highest flexibility and Q-ratio was chosen. This design consisted of a coil wound with 18-gauge wire and two tuning capacitors.

The surface coil and 2-channel phased array were compared to an off-the-shelf Siemen’s neck coil. The comparison consisted of using a uniform phantom and subjecting it to the same pulse sequence. The voxels’ SNR that was 1.5 cm to 3.2 cm into the phantom was used for the comparison. The results showed that the surface coil and 2-channel phased array had higher SNR in the region of interest than the Siemen’s coil. Of the two custom coils, the 2-channel phased array had a more uniform voxel SNR than the surface coil.

The 2-channel phased array was used to image an excised pig larynx. The high-resolution larynx image that was produced in this thesis had a voxel size of 0.2 x 0.2 x 0.9 mm. The high-resolution image allowed MR relaxometry measurements on larynx tissue that had not been previously recorded.

6.1.2 Biomechanical Modeling of the Hip

Additional work was also done with the orthopedists at the University of Utah. The orthopedists wanted to produce a biomechanical model of the hip. The initial experimental
MRI images acquired to explore building 3D models of the hip yielded a signal from the fluids in the hip joint that was almost the same intensity as the signal from cartilage, making it very difficult to distinguish cartilage from fluid. To determine the extent of the problem, relaxometry values of several different solutions consisting of saline, isovue, and lidocaine were measured. The results showed that adding isovue or lidocaine to saline did decrease the $T_1$ and $T_2$ of the solution. These findings helped shows that the solutions used in the initial MRI scan were not an effective contrast agent.

6.2 Future Work

Future work for the larynx project will include exploring a 3-channel phased array to improve the image homogeneity across the larynx. In addition, using the relaxometry values of the larynx to design an optimized pulse sequence that provides better contrast between tissues.

The results from the hip project showed that isovue and lidocaine have limitation as an effective contrast agent in distinguishing cartilage in the hip. The orthopedics at the University of Utah are exploring a different approach on using a pure saline solution as a contrast agent. To inject into the proper location in the hip, the orthopedics are planning to use ultrasound, instead of CT, to guide the injection. If this approach is proven successful, then the orthopedics may be able to use MRI imaging to produce a hip model that can be used surgeries.
Appendix A

High Resolution Images of a Pig Larynx

![Figure A.1: High Resolution Image of a Pig Larynx Slice 1-4](image)
Figure A.2: High Resolution Image of a Pig Larynx Slice 5-10
Figure A.3: High Resolution Image of a Pig Larynx Slice 11-16
Figure A.4: High Resolution Image of a Pig Larynx Slice 17-22
Appendix B

Larynx Coil Hardware

This section of the thesis will cover the hardware design of the larynx coil. The design of the support structure that house the larynx coil is shown in this section. To connect the larynx coil to the MRI scanner, a preamp board was built. The basic concept of each component inside the preamp board is will be reviewed.

B.1 Support Structure for the Larynx Coil

The MRI coil’s resonance frequency is sensitive to the sample that loads the coil and any movement from the coil. To build a robust MRI coil, a support structure needs to be designed to give the MRI coil proper structural support and protection to the patient. While providing these properties, the distance between the patient and the coil should be kept to a minimal for the purpose of maximizing the field of view of the coil.

This was taken into account in designing the support structure for the larynx coil. The support structure was curved to be able to lie on top of the patient neck. The top of the structure was covered to protect the patient from any contact with the electronics. A hole was placed on the top structure to allow moving and tuning the coil. To reduce the distance between the patient and the larynx coil, an indentation was placed at the bottom of the structure to allow for the adam’s apple to rest. This is seen in Figure B.1.

B.2 Preamp Board

For any MRI coil, supportive circuitry is required to connect the coil to the MRI scanner. For a transmit coil, a TR switch is required to connect the coil to the scanner. As for a receive coil, such as the larynx coil, a preamp board is required. The preamp board
that was used for this thesis consist of three components; cable trap, phase shifter and a preamplifier. The subsections below will cover the concepts for each of these components.

B.2.1 Cable Trap

Cable traps protect the preamp from stray current caused by the transmitting body coil. The cable trap applies the principles of resonance to prevent stray currents from reaching the preamp. A cable trap works through a capacitor and inductor in parallel (see Figure B.1). Applying Kirchhoff’s voltage law to Figure B.1, the impedance of the system is thus defined as:

\[ Z_{\text{system}} = \frac{Z_c Z_L}{Z_c + Z_L} = \frac{L}{\frac{1}{jwC} + jwL}. \]  
(B.1)

Resonance for this system occurs when \( w = \frac{1}{\sqrt{LC}} \). At the resonance frequency, the system impedances denominator will approach zero. This will stop the common-mode current from passing through this system through heat dissipation. The capacitor and inductor that are used in this system are designed to withstand high voltages and currents to prevent breakdown. The capacitor voltage rating should be at least 1000 volts. The inductor should be made of semi rigid coaxial cable. An RF shield is placed around the cable trap to prevent RF leaking to the MRI coils.
B.2.2 Phase Shifters

The wavelength for hydrogen in a 3 tesla MRI is 2.4 meters. The cables that are used to connect the surface coil to the supportive circuitry have to be about 25 cm in length due to a research agreement with Siemens. The cable length is long enough that transmission line effects the NMR signal before being received by the preamp. A phase shifter is used to reduce the transmission line effects by making the distance from the coil to the preamp half wave distance. The circuit diagram of the phase shifter can be seen in Figure B.2

![Cable Trap](image)

Figure B.2: Cable Trap

![Phase Shifter Schematic](image)

Figure B.3: Phase Shifter Schematic

B.2.3 Preamplifier

The preamp’s purpose is to amplify NMR signals to a reasonable level that can be sampled by the MRI scanner. The preamp is a GaAs-FET low noise preamp nonmagnetic
type. The preamp has to be a high reflection input type to help with decoupling between surface coils in the phased array.
Bibliography


