I, Brent Rudd, hereby submit this original work as part of the requirements for the degree of Doctor of Philosophy in Mechanical Engineering.

It is entitled:
Active Tonal and Broadband Noise Control for Magnetic Resonance Imaging Systems

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Active Tonal and Broadband Noise Control for
Magnetic Resonance Imaging Systems

A dissertation submitted to the
Graduate School of the University of Cincinnati
in partial fulfillment of the requirements for the degree of

Doctor of Philosophy

in the Department of Mechanical Engineering
of the College of Engineering

November 2010

by

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ABSTRACT

Magnetic resonance imaging (MRI) is a powerful medical diagnostic tool. It offers advantages over other common medical imaging processes. For example, MRI is capable of producing high contrast images of soft tissue and does not use potentially harmful radiation.

The imaging procedure requires the subject be placed in a high strength static magnetic field. Next, the patient is subjected to magnetic field gradients that are rapidly altered and interspersed with precisely timed radio frequency transmissions and measurements. The magnetic field gradients are created by transmitting electrical current through gradient coils, which unfortunately produces loud noise during the scanning procedure. This noise is a function of the static magnetic field strength and gradient signal, both of which are essential to the procedure. It is loud enough to be unpleasant, and has the potential to be harmful. In specific applications the sound pressure level may be great enough to be a limiting factor in determining imaging protocol.

The unfortunate side effect of noise production has been recognized since the advent of MRI. Engineers and scientists have proposed a variety of approaches to address the problem. Unfortunately, there has been limited success in developing a practical, cost effective solution that is applicable to a wide range of scanners.

This dissertation will address the development of an active noise control system specifically designed to attenuate tonal and broadband noise generated during magnetic resonance imaging. In order to develop an ANC system capable of operating in the magnetic environment, non-ferromagnetic speakers and microphones were tested and implementation
procedures developed that allow for optimal performance. Due to the high cost of operating the MRI scanner, a physical simulation environment including noise and MRI gradient signals was created to allow for testing during the development process. A control system capable of providing significant noise reduction at frequencies up to 5 kHz was developed and tested for multiple common MRI scanning sequences. These scanning sequences exhibit different frequency and temporal characteristics and were selected with the goal of developing a robust control system capable of being tuned for a wide variety of imaging sequences.

Finally, the ANC system was evaluated at the University of Cincinnati’s Center for Imaging Research. The ANC system was interfaced with the equipment controlling the 4 T Varian Unity INOVA Whole Body MRI/MRS system. Live scanning sequences were conducted, and the ANC system’s performance measured and reported. Data from the in-situ test will be presented that shows the ANC system provides substantial overall noise reduction for each of the three scanning sequences conducted. The maximum reduction measured during the in-situ scan for a specific harmonic was 55 dB during a gradient echo multi-slice (GEMS) scan, with 30 dBA overall reduction across virtually all harmonics up to 5 kHz.
ACKNOWLEDGMENTS

First, I thank Professor Teik C. Lim, Ph.D, P.E., and Jing-Huei Lee, Ph.D., who are serving as my academic advisors and committee co-chairs, for their guidance and support during my graduate study. I also thank the members of my dissertation committee, Professor Jay H. Kim, Ph.D., and Professor David Thompson, Ph.D.

In addition to my dissertation committee, I would like to acknowledge Mingfeng Li, Ph.D. (Research Associate in the Department of Mechanical Engineering) for his collaboration on the technical aspects of my research, and Mr. Jeffrey Osterhage (Medical School’s Center for Imaging Research) for providing technical assistance in conducting the in-situ MRI scans utilized for this study.

This research has been supported by the University of Cincinnati’s College of Medicine (Dean’s Discovery Fund), the National Institute of Biomedical Imaging and BioEngineering (EB005042), the National Institute for Occupational Safety and Health (NIOSH) University of Cincinnati Education and Research Center grant #T42/OH008432-05, and Resonance Technology, Inc.

Finally, thanks to my wife Mary for her support throughout the course of my studies.
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<tr>
<th>Acronym</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>ADC</td>
<td>Analog to digital converter</td>
</tr>
<tr>
<td>ANC</td>
<td>Active noise control</td>
</tr>
<tr>
<td>DAC</td>
<td>Digital to analog converter</td>
</tr>
<tr>
<td>EPI</td>
<td>Echo planar imaging</td>
</tr>
<tr>
<td>FIR</td>
<td>Finite impulse response</td>
</tr>
<tr>
<td>FFT</td>
<td>Fast fourier transform</td>
</tr>
<tr>
<td>fMRI</td>
<td>Functional magnetic resonance imaging</td>
</tr>
<tr>
<td>FXLMS</td>
<td>Filtered-x least mean square</td>
</tr>
<tr>
<td>FULMS</td>
<td>Filtered-U least mean square</td>
</tr>
<tr>
<td>GEMS</td>
<td>Gradient echo multi-slice</td>
</tr>
<tr>
<td>LMS</td>
<td>Least mean square</td>
</tr>
<tr>
<td>MDEFT</td>
<td>Modified driven equilibrium fourier transform</td>
</tr>
<tr>
<td>MRI</td>
<td>Magnetic resonance imaging</td>
</tr>
<tr>
<td>NMR</td>
<td>Nuclear magnetic resonance</td>
</tr>
<tr>
<td>RF</td>
<td>Radio frequency</td>
</tr>
<tr>
<td>SPL</td>
<td>Sound pressure level</td>
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</table>
LIST OF SYMBOLS

B  Magnetic field strength
C  degrees Celsius
\(d\)  Disturbance
dB  decibels
dBA  A-weighted decibels
e  Error
f  Frequency
\(h\)  Planck constant
Hz  Hertz
I  electrical current
k  Boltzmann constant
K  degrees Kelvin
KHz  Kilohertz
\(L\)  Length of adaptive filter
MHz  Megahertz
\(M_0\)  Net magnetization
N  Number of protons
Pa  Pascal
\(P_{x'}\)  Power of filtered reference signal
$r$  Reference signal

$S$  Secondary path

$\dot{S}$  Secondary path estimate

$T$  Tesla

$T$  Temperature

$T_1$  Relaxation time constant 1

$T_2$  Relaxation time constant 2

$W$  Adaptive filter

$x$  Reference signal

$x'$  Filtered reference signal

$y$  Controller output signal

$y'$  Control speaker output

$\gamma$  Larmor ratio

$\mu$  Step size

$\pi$  Pi

$\omega$  Larmor frequency
1. INTRODUCTION

Magnetic Resonance Imaging (MRI) is used to generate inner organ images for medical diagnosis. In addition to providing high contrast and soft tissue images not possible with other common medical imaging processes, it doesn’t produce any known potentially harmful side effects that exist with other procedures, such as radiation.

During imaging, the patient is placed in a high strength static magnetic field. Magnetic field gradients are rapidly pulsed and interspersed with precisely timed radio frequency (RF) transmissions and measurements of the MR signals of the water hydrogen atoms. These pulse sequences vary according to the diagnostic procedure being performed. They rely on the characteristic of nuclear magnetic resonance, allowing for different tissue types to be distinguished, and are designed to provide information about hydrogen atoms at specific locations. A brief pulse sequence is capable of providing information about a discrete location, and is then typically repeated with slight variations to provide information for additional adjacent locations. All of the acquired information is then processed to generate images about the subject that was interrogated in a non-invasive manner.

The magnetic field is altered by transmitting electrical current through gradient coils. The gradient coils, which are located in the vicinity of the patient, are subjected to Lorentz forces produced by the conduction of electrical current in the magnetic field. These forces, which vary with the rapidly changing electrical current, subject the coil structure to rapidly changing loads, resulting in deformation. These dynamic loads cause vibration of the gradient coil structure, producing acoustic noise as the coil interacts with the surrounding air,
behaving like a loudspeaker. Sound pressure levels (SPLs) in excess of 115 dB have been measured on a 3 Tesla scanner [5, 10].

The functionality of MRI has been enhanced by increasing static field strengths and implementing faster gradient signals. Unfortunately, a direct consequence of these actions is an increase in SPL. Despite the attention for noise generation MRI has received, it remains an issue. In specific imaging procedures where the speed of acquisition is critical (e.g. brain function activity), it may also be a limiting factor in scanning capability.

A typical step in preparation for imaging is to provide the patient with ear protection to attenuate the sound. This is usually accomplished with a set of headphones that are also used to facilitate communication with the operator during the scanning process. In addition, the headphones may be utilized to provide entertainment (e.g. music or movie) to distract from the unpleasant sound. Due to the magnetic environment, all materials used in these headphones must be non-ferromagnetic.

One potential solution to the noise exposure is the implementation of an active noise control (ANC) system. Ideally, the ANC system produces a sound wave that is of equal magnitude to an offending sound, but 180° out of phase to the original. A perfectly generated correction signal combines with the original sound, resulting in silence through destructive interference. Due to practical limitations, generating an ideal correction signal resulting in complete cancellation is not generally possible. There are various approaches to developing ANC systems. Achieving the best sound reduction requires tailoring the system to the specific sound characteristics and environment for a particular situation.

During previous research activity into the characteristics of MRI noise, More, et al, [10],
found a direct correlation between MRI gradient signals and the sound produced, relating portions of the sound spectrum to specific gradient signals for an EPI scan. Based on these findings, a preliminary investigation was conducted. An EPI scan was conducted multiple times, once for each of three individual gradients and once with all gradients active. Test measurements showed good coherence between the gradient signals and the sound produced. In addition, the individual results were used to analytically reconstruct the sound spectrum produced when all three gradient signals are active. This analytically reconstructed sound spectrum compared favorably to the sound spectrum measured during the live scan, replicating the findings of More, et al.

Kuo and Morgan [52-53] proposed the use of feedforward control systems when no acoustic correlated reference signals are available. Considered in concert with the preliminary testing, the feedforward control system was chosen as the basic algorithm for use in the proposed ANC system tailored for MRI. Since the gradient signal is associated with the production of the undesirable noise, it occurs slightly before the noise, and may allow for more effective reduction than other approaches.

This dissertation addresses the development of an ANC system specifically designed to attenuate noise generated during magnetic resonance imaging.

In chapter 2, patents on the subject of sound reduction technology for MRI is reviewed. Despite numerous patents being granted that are designed to address the noise problem, a practical, commercially viable solution capable of providing substantial noise reduction and reducing exposure has yet to emerge.

The next chapter documents basic MRI operation and demonstrate the requirement for
non-ferromagnetic equipment in the vicinity of the scanner. An initial set of MRI compatible hardware will be evaluated and compared to high quality non MRI compatible equipment to assess the feasibility of the proposed approach. To complete this work, a physical simulation of the MRI environment was developed that includes sound reproduction along with temporally accurate MRI gradient signals which were used as reference signals by the control system. The specific process that allows the MRI compatible speakers to perform at a level that produces a cancellation signal capable of providing substantial noise reduction was also developed.

In chapter 4 a second generation set of hardware is evaluated to determine the potential for the additional functionality of a microphone reference signal. This was found to be necessary after determining the conceptual control system introduced in chapter 3 did not adapt well to all scan types. Introducing additional functionality allowed for better noise reduction of other scanning sequences.

In the following chapter, the complete system developed specifically to address MRI noise under simulated conditions is evaluated during in-situ scanning to assess the performance. This information serves to validate that the physical simulation environment used during development is representative enough to allow for the production of a system that performs well during actual MRI operation. In addition, utilization of multiple reference signals is introduced to enhance the noise reduction when compared to a single reference signal calculation.

In chapter 6, the system is again evaluated during in-situ scanning, but with an updated control system designed to address a broader frequency range. The control system is
enhanced by introducing frequency filters that allow for an individual control algorithm to target a portion of the frequency spectrum for reduction. The control system utilizes multiple parallel algorithms which are individually tuned to provide the optimal noise reduction for the targeted noise, and when combined into a single control signal provide for substantial noise reduction to 5 kHz.

The final chapter provides conclusions as well as a discussion of future work. It is followed by a summary of the publications that have resulted from this research activity to date, a total of 12 publications including 2 journal papers. In addition, multiple submissions have been made to journals for publication consideration.

The result of the research activity documented in this dissertation is that an ANC system capable of providing substantial noise reduction to 5 kHz has been developed and demonstrated specifically for the MRI noise application. If implemented commercially, it has the potential to provide a significant improvement for the patient experience during MRI scanning. In addition, it may reduce noise as a barrier to specific MRI research. The noise level produced during imaging is typically a consideration and sometimes a limiting factor when developing scanning protocols. It may also allow for new directions in research. One such example is brain function. Currently, the patient’s brain is actively processing the loud noise he/she hears during the scanning procedure. By conducting scans with and without the control system active, it may be possible to learn more precisely about brain functionality that is confounded by acoustic noise.
2. CURRENT SOUND REDUCTION TECHNOLOGIES
FOR MRI SCANNERS

Noise has been recognized as a problem since the advent of MRI technology. Unfortunately, the specific procedure that makes the imaging possible is also responsible for the noise generation. The following paper, “Sound Reduction Technologies for MRI Scanners” was published in the journal Recent Patents on Engineering [50]. It has been reformatted for inclusion here. Several patents pertaining specifically to MRI noise reduction are evaluated.
2.1 Summary

Magnetic Resonance Imaging (MRI) is a powerful imaging tool used in clinical diagnosis and biomedical research. One of the undesirable characteristics of the process is the loud noise generated during a scan. A major source of this noise is the result of switching electrical current in gradient coils within the static magnetic field. As the technology has evolved, the quality of images has improved by, among other things, increasing static magnetic fields and gradient switching rates. These improvements have also increased the imaging speed, allowing for observation of dynamic events. For example, functional MRI (fMRI) allows changes in brain function over short time intervals to be observed. However, the downside is a corresponding increase in the level of noise generated. The sound pressure levels (SPLs) the patient is subjected to not only produce discomfort, but are potentially harmful as well. Reducing exposure to these high SPLs would not only result in improved patient comfort, but also allow for more robust imaging procedures, since the magnitude of noise exposure can be a limiting factor in certain types of MRI procedures. In this paper recent patents related to reducing noise exposure with the potential for application to MRI will be examined.

2.2 Introduction

MRI system is one of the most powerful imaging tools available for clinical diagnosis and basic biomedical research. Its continual development to improve image quality,
resolution, and throughput has resulted in steadily increasing static magnetic field strength and gradient coil slew rates. These enhancements unfortunately have resulted in the generation of extremely loud acoustic noise inside and around the scanner. The acoustic noise generated from the MRI system, which has always been an annoyance since the invention of MRI, has become a serious safety concern as well as a technical challenge for MR and acoustic scientists. The MRI acoustic noise is due to the switching of electrical current in gradient coils within the main static field [1-2]. The Lorentz forces induced by the electrical current applied to gradient coils physically excite the structural components, and effectively turn the MRI system into a massive loudspeaker through the structural-acoustic coupling with the surrounding air. The magnitude of structural vibrations responsible for acoustic noise emission depends on the strength of the magnetic field, amplitude of the applied current, scanner structure and geometry, spatial settings, and frequency and waveform of the switching current [3]. Several studies have recently quantified the scanner acoustic noise generated by different static magnetic field strengths [3-7, 12]. They reported that the measured noise levels varied from 82.5 dBA for a 0.23 Tesla (T) system to 118.4 dBA for a 3 T system using conventional acquisition [5-6], and 115 dB for a 1.5 T increasing to 131 dB using EPI [3-7]. Here, dBA is the A-weighted SPL frequently used to gauge the perceived loudness while dB is simply the raw, unweighted SPL in decibels. Recently, Edelstein and his coworkers found that the eddy current induced vibrations of metal structures (e.g. cryostat inner bore and RF body coil, etc.) rather than the self-excited vibrations of the gradient coils may be responsible for the primary noise source [8-9].

There are three fundamental ways to approach noise problems. The first is to reduce the
generation of noise excitation at its source, i.e. the MRI system, or more specifically its gradient coils [13-14]. One downside to this approach is that it would most likely require changes to the basic design of the MR imager. This approach is more viable for new MRI machines that are yet to be built, but is potentially difficult and expensive to retrofit into an existing installation. While the gradient assembly is a major noise source, there are additional components that are also significant contributors [9]. Changes to numerous components of the scanner may be necessary to achieve the desired result.

The second approach is to evaluate the path from the noise source to the patient with the goal of altering it in some ways that produce quieter responses. In the design of MRI machines where the patient is located in the vicinity of the noise source, the primary path can be a combination of both airborne and structure-borne routes. For example, vacuum based bore liners have been proposed for improving isolation of the gradient coil vibration [8-9, 14] in order to eliminate the structure-borne path while minimizing the airborne one.

The third approach is to reduce the level of noise at the point of reception, in this case the patient’s ear. For example, patients may wear a set of protective earmuffs that reduces the transmission of sound waves, similar to the ear protection worn by airport ground crew workers. Even though this form of local control of reducing noise near the patient’s ear can be highly effective, the treatment does not discriminate between multiple sources and/or paths. In other words, if the sound is present, there is potential for reduction. In some testing, this may lead to interference with communication between patient and MRI technicians or other clinicians. On the other hand, a major advantage of this approach is that it would typically require no modification to the MRI scanner and structure, and thus allowing
implementation to various types of installations, including those already in service. The ear protection mentioned is a passive reduction method. Another possibility for the reception reduction approach is to utilize active noise cancellation (ANC) [10-11]. Sound is created by the variation of air pressure about a mean value. This variation alternates between being positive and negative amplitudes relative to the mean value. Ideally, an ANC system measures the varying sound pressure and instantaneously generates the opposite sound pressure, which is additive to the original sound pressure, resulting in the suppression of the original sound. Obviously there are practical limitations to the implementation of ANC systems that impact their effectiveness and prevent them from completely eliminating the undesired noise. The effectiveness of the ANC system used will vary due to the specific characteristics of the noise response in question.

Regardless of the approach used, there is an additional factor to consider: the MRI environment. The magnetic field that is inherent to the procedure limits the use of magnetic components in the vicinity of the scanner. This is for both safety and image quality. An unrestrained magnetic component will become a projectile if brought too close to the magnet of imagers, creating a potentially hazardous situation for the patient and anyone else in the vicinity. Even safely restrained items may disrupt the magnetic field and impact the quality of the image. This limits the choice of materials available for any components that must be deployed in the vicinity of the scanner.

The following sections containing patent reviews will be grouped by the fundamental approach they utilize.
2.3 Existing Patent Review

2.3.1 Noise Source Reduction

Granted in January, 1999, to Dean, et al [15], this patent aimed to reduce acoustic noise and isolate vibration of the magnet cryocooler to increase patient and operator comfort and improve image quality. MR imagers require a large static magnetic field to function. One way to achieve such fields is with a superconducting coil. The coil temperature is reduced to an extremely low value (liquid helium temperature, i.e. c.a. 4 degrees K) to virtually eliminate electrical resistance. An electric current is introduced by attaching a power source to the low temperature coil. After the desired current is realized the power source can be removed and the current will continue to flow as long as the coil has no electrical resistance, which is achieved by keeping the coil at about 4 degrees K with a cryocooler. The continuous flow of electricity through the coil creates the magnetic field.

A cryocooler is analogous to a refrigerator, except that the temperatures involved are much lower. The cryogenic temperature range is generally considered to be below -150°C. Liquid helium is commonly used as the cryogen for MRI superconducting magnets. The operation of the cryocooler motor generates noise that may be transmitted to the magnet through supports and tubing needed to transport the cryogen material.

Dean, et al, introduces two distinct features to address the noise and vibration of the cryocooler [15]. First, the motor is mounted with a vibration isolator tuned below the first resonant frequency of the magnet. In Figure 2.1, the motor (label 13 in Figure 2.1) is attached with elastomeric mounts (Label 30). Second, there is a bellows (Label 28) between the motor
and the magnet. The vibration isolation is achieved by having a relatively small stiffness connection. This stiffness and the mass of the motor combine to create a natural frequency below other resonant frequencies of the system. At excitation frequencies above the isolation frequency, the mass of the motor dictates the motion, which is reduced. At excitation frequencies below the isolation frequency, the stiffness of the isolator connector is the controlling factor of the motion. This stiffness is relatively small, allowing relatively large motion of the motor. In order to support the motor while allowing the axial motion, an appropriate support structure is needed, as shown in Figure 2.2. There is an axle (Label 48) that allows rotation of the support structure and motor.

Figure 2.1: Cryocooler bellows [adapted from ref. 15]
Figure 2.2: Cryocooler support [adapted from ref. 15]
In 2003, Roozen, et al proposed MRI apparatus with a piezo actuator in a non-rigid suspension of the gradient coil carrier to reduce the MRI acoustic noise [16]. Having identified the gradient coil as a major source of undesirable noise, Roozen and his colleagues looked at options for reducing vibration by modifying the support system. Their proposal involves the use of compliant suspension elements coupled with piezo actuators for supporting the gradient coil. This reduces the transmission of vibration, while preventing large deformations that could have an adverse affect on the imaging process. Compliant suspension elements alone have limits to how effective they are. The addition of an actuator increases the suspension’s effectiveness. The piezo actuator is necessitated by the large static magnetic field in which the actuator will operate.

Piezoelectric materials deform when an electrical signal is applied. This avoids the use of magnetic materials in the magnetic field, which would present other difficulties. The piezo actuators are driven by a control system. The control system utilizes a feed forward algorithm that is triggered by the electrical signal supplied to the gradient coil. By assuming that the relationship between the gradient signal and the noise produced are correlated and predictable, the control system “knows” what sound waves will be produced by the supplied gradient electrical signal. This type of control algorithm is commonly referred to as feedforward.

Figure 2.3 is an end view of a cylindrical gradient coil (Label 18). It may have numerous supports (Label 19), two of which are shown. Each support is composed of two major pieces. The first is a resilient component (Label 22), such as a rubber block, with a small stiffness. The stiffness is chosen such that the frequency of vibration of the gradient coil mass on the support stiffness is below the frequency to be isolated, which the authors suggest
to be 700 Hz. The other component is the actuator (Label 21), which is driven by the control system to reduce the displacement of the gradient coil by compressing and expanding the compliant block (Label 22).

Figure 2.3: Gradient coil suspension system [adapted from ref. 16]
In 2005, a design of using an active vibration compensation for MRI gradient coil support to reduce acoustic noise was proposed by William Edelstein [17]. It is similar to the gradient coil suspension proposed by Roozen and his colleagues [16]. However, Edelstein recognizes that the piezoelectric force transducers needed to reduce the motion of the gradient coil are fragile. The gradient coil is massive enough that these transducers may be subject to breaking if they support the gradient coil in the manner proposed by Roozen, et al [16].

The suspension in Edelstein’s patent places the actuator in parallel with the compliant support instead of in series, as in Roozen, et al’s implementation. Figure 2.4 is an end view of Edelstein’s proposed suspension, while Figure 2.5 is a close up of a suspension element (Label 300). The gradient coil (Label 102) is supported by a fixed frame (Label 100) through an elastomeric mount (Label 110). The actuator (Label 140) does not support the weight of the gradient coil, but is configured in a manner that allows the actuator to apply force to the gradient coil.
Figure 2.4: Gradient coil suspension designed by Edelstein [adapted from ref. 17]

Figure 2.5: Detail of gradient coil suspension designed by Edelstein
[adapted from ref. 17]
Roozen, et al designed a noise reduction system specifically for open MRI systems comprised of acoustic resonators in 2006 [18]. Many MRI systems are cylindrical and require the portion of the patient to be imaged to be inside the cylinder. This may be difficult for patients who are claustrophobic. Open MRI systems are designed to reduce this by not requiring the target of the image to be in an enclosed space. Figure 2.6 is a sketch of this type of MRI. The lower and upper housings (Labels 3 and 5) contain the gradient magnets, which are a conical shape.

![Figure 2.6: Open MRI [adapted from ref. 18]](image)

Figure 2.6 shows an overall cross section of the housings. The patent holders recognized that the vibration was greatest at the tip of the cone due to the reduced stiffness in this vicinity. They also recognized that the noise is higher at specific frequencies. This is
described as harmonic or tone, as opposed to broadband noise that has similar strength across a frequency range. Their proposed solution was to design acoustic resonators that were sized to tune out specific wavelengths of sound, and to locate these resonators in the vicinity of the tip of the cone. The length of the resonator is sized to be a factor of the wavelength of the targeted frequency. This factor is \( k/4 \), where \( k \) is an odd number, 1, 3, 5, etc. Figure 2.8 is a close up view of the conical tip with a resonator added.

![Diagram of a conical tip with resonators](image)

**Figure 2.7: Open MRI Cross Section [adapted from ref. 18]**
A novel concept using active acoustic control with flexible joints in gradient coil design for reducing MRI noises was proposed by Peter Mansfield in 2006 [19]. Figure 2.9 shows a typical gradient coil design, made up of a block of material containing a loop of wire that carries a current within the static magnetic field of the imager. The current in the magnetic field generates a force in the plane of the loop of wire, which then vibrates the material and emits sound. Mansfield proposes a coil design that splits the material into two distinct portions connected by a flexible material and adding an additional loop of wire, as shown in Figure 2.10. The inner coil carries current I₂. The design of the coil is such that the distance between the outgoing and returning portions of the inner coil is small so that it will have negligible impact on the magnetic field. The split is designed with flexible joints such that the forces in the plane of the two distinct portions do not interact with each other. Finally, the current I₂ is determined using active control techniques to create a secondary sound wave that can negate the acoustic noise generated by current I₁.
Figure 2.9: Typical Gradient Coil Design [adapted from ref. 19]

Figure 2.10: Proposed Gradient Coil Design [adapted from ref. 19]
2.3.2 Path Modification

Another approach for noise reduction involves the use of acoustic noise path modification to minimize sound propagation across the structural components of the scanner. Sound-absorbing materials can be attached to reflecting surfaces such as the cryostat and the scan room walls, ceiling and floor. These hardware modifications will undoubtedly become part of the next generation of MRI systems. Such modifications will also incorporate different components and materials, improved isolation of the gradients, and/or redresses of the coil shape, while maintaining the wide bore and rapid gradient switching capabilities of our current systems.

One patent has been granted for a MRI system that places gradient coils in a vacuum enclosure [20]. It is known that the acoustic noise is a sound wave that requires media to propagate into the surrounding environment, and air is just such an excellent medium. Therefore, isolating the vibration source with a vacuum chamber is conceptually an ideal way to reduce sound transmission. Figure 2.11 and Figure 2.12, proposed by Pla and his colleagues, illustrate such a MRI scanner that contains two vacuum chambers (Labels 16 and 12). The first vacuum chamber (Label 16), a traditional housing for an MRI superconductive main coil (Label 18), has an ultra-low pressure typically in the range of 10-6 torr. Since superconductive coils only operate at a temperature of about 4-10K degree, the ultra-low pressure is required for heat insulation so as to prevent superconductive coils from quenching (i.e. loss of superconductivity). The second vacuum chamber (Label 12) encloses gradient coils (Label 14) in order to reduce the transmission of sound through the surrounding air.
Figure 2.11: A schematic cross-sectional side-elevational view of an MRI scanner subassembly [adapted from reference 20]
A preferred vacuum pressure of 250 torr was chosen with the following considerations: a pressure below a high pressure limit is needed to achieve a sound reduction, performance maximization by allowing the use of gradient coil assembly materials, and cost minimization. An enclosure with a vacuum of 250 torr would reduce the MRI acoustic noise by half as compared to atmospheric pressure (760 torr). Additionally, a sound-absorption material (Label 28; e.g. loose fiberglass insulation) arranged within the chamber surrounding all surfaces of the gradient coils (Label14) can further attenuate any residual acoustic noise.
2.3.3 Reduction of SPL in Patient Vicinity

For patients, the simplest way to effectively reduce acoustic noise is achieved by utilizing passive acoustic noise control devices such as earplugs and earphones, which is a direct and simple treatment in patient’s vicinity. In fact, most MRI centers require the use of earplugs to protect patients’ hearing from the intense scanner sound. Some centers have further used headphones. However, it has recently become evident that patients’ cochlear function can be affected by MRI acoustic noise even when they are protected with earplugs [21]. Despite this drawback, the use of earplugs is one of the most popular and economical approaches available today.

A relatively sophisticated approach considered for the treatment of MRI noise is active acoustic noise cancellation that works in a manner comparable to current commercially available active headphones [22]. Basically, the scanner acoustic noise is sampled and a control signal is generated nearly simultaneously to produce a secondary excitation with the same amplitude as the original noise source, but phase shifted by 180°, for a given frequency band. The noise suppression is essentially achieved by injecting synthesized out-of-phase signals in the headphone speakers to create a localized null zone. Through destructive interference, the original and secondary sound waves cancel each other out. As the result, the subjects will only perceive a much reduced sound pressure level since its intensity will have been reduced significantly.

Many active noise cancellation systems using the concept of out-of-phase signal have been patented [23-26]. For example, the acoustic noise is attenuated by a generated secondary sound wave to null with the primary wave in a confined space [23]. An apparatus for
improving audibility of a repetitive incident sound using a sensor and an adaptor was
designed by Sound Attenuators Limited, UK [24]. Adaptive filters by Noise Cancellation
Technology, US [25], have the capability to measure the phase delay between the sound wave
detected by the processor and that generated from the environment. Plessey Overseas Limited,
UK proposed a noise reduction system specific to target the low frequency noise [26].
However, none of these are MRI compatible. As noted above, materials selected for MRI
environment must be non-ferromagnetic. A non-magnetic transducer used for MRI active
noise control has been proposed and patented in 1995 [27]. The transducer is made of
materials such as electrets, piezoelectric or condenser type that have no ferromagnetic
material. Below, we review patents related to MRI compatible active noise cancellation
development.

An MRI noise control system capable of effectively reducing acoustic noise generated
during MRI scan has been patented in 1995 [28]. This system uses a controller to store the
characteristics of respective acoustic noises being propagated to targeted locations in a space
where sound attenuation is desired; and a supplementary sound wave is driven by the
controller on the basis of each acoustic noise stored in the memory and information on the
pulse sequence such that the total sum of the amplitude of sound wave can be minimized.
Since the characteristic of each respective acoustic noise at a certain location is detected and
recorded in advance, this approach may not be considered as an adaptive “active” (i.e.
adaptive) noise control.

One interesting patent reported by Pla et al is the design of an active noise and vibration
control system for MRI systems to minimize acoustic output by generating a secondary noise
and/or vibration field using vibrational inputs [29]. The inner surface of an MRI scanner is mounted with piezoceramic actuators. With transducers sensing the noise and/or vibrations generating by the scanner and a controller sending a control signal to the actuators in responses to the error signal, the actuators can vibrate and generate a noise (or vibration) field with an out-of-phase signal, which minimizes the total noise emitted from the scanner. This seems to an interesting concept; however, one study demonstrated that the sound pressure levels are spatially dependent inside the MRI bore [10]. A rough estimation of the modal density of the bore acoustic cavity used in that study yields nearly a hundred natural modes in the one-third octave band corresponding to the 1 kHz center frequency. At the 10 kHz center frequency range, approximately 65,000 acoustic cavity modes present. Using an impulse response test [10], it was confirmed qualitatively that the modal density of the bore acoustic cavity is quite high. Based on these findings, one would think that it is extremely difficult to achieve a global reduction in sound pressure levels throughout the bore space due to the dense distribution of the modes that essentially points to the existence of highly intricate mode shapes. Therefore, it is questionable how effective of the system proposed by Pla et al operates. Hence, one may only attempt to achieve noise reduction in the specific regions of interest; that is, mainly at the ear and potentially at the mouth area as well. We believe that noise reduction focusing only on these regions is more controllable and feasible. An example of such a system was demonstrated and patented by Brungart in 2002 [30]. This noise cancellation system employs a pneumatic headset with an error microphone. The error microphone measures an acoustic noise signal at the ear location of a patient and a noise cancellation processor produces an acoustic noise signal with a delayed timing resulting in an
anti-phase of the error signal, hence, the total acoustic noise can be cancelled. A reduction of sound pressure level of 20 dB or more is claimed [30]. However, there are several limitations with the delay based active noise cancellation system. For example, this system is only effective at low frequencies (from 40 Hz to 500 Hz), which is incompatible with most of the MRI noise occurring around 1 kHz. Additionally, the large time delays due to the propagation of the sound signal in pneumatic tubes cannot adequately work with most traditional noise canceling algorithms such as the LMS algorithm. It is also noted that the systems proposed by Pla et al and by Brungart cancel the MRI noise signal at fixed locations inside the MRI, rather than at the locations of the patient's ears. Hence, it would seem more effective to design a headset with an active noise cancellation system comprising non-magnetic microphones located inside the headset to record the error signal and non-magnetic piezoceramic loudspeakers also located inside the headset to generate the noise cancellation signal.


2.4 Current & Future Developments

Advanced whole-body imaging systems have recently improved significantly and further performance enhancements are likely hindered more by human physiology than by hardware limitations. With the recent increase in magnetic field strength (i.e. whole-body systems at 7 T and 9.4 T) and the growing demands on gradient current switching rates, these advanced MRI systems generate acoustic noise at levels that pose significant risks to human subjects even with limited exposure duration. Currently, various passive and active approaches to reduce the SPL in MRI have been proposed. Major approaches include passive acoustic insulation, active acoustic cancellation, gradient coil design, RF pulse sequence modifications, as well as some combinations of the above methods [31]. As noted above, a variety of methods have been patented to reduce acoustic noise emission during MRI scans, but none have proven successful in gaining widespread acceptance. This is because most of these solutions are either too costly, requiring significant modification of the MRI structure, possibly degrading MRI performance and/or image quality, or they are simply a conceptual idea. Perhaps the most obvious approach for the acoustic noise problem is to implement engineering modifications directly at the source of the problem, i.e. the MRI system, or more specifically its gradient coils [32-33]. However, the main drawback for this solution is the high cost, especially for existing systems. Furthermore, it requires more complex gradient driving systems and may compromise gradient performance. To date, the most cost effective approach to reduce scanner noise is to modify RF acquisition sequences by degrading rise time in gradient pulses [34-35] or by avoiding the structure resonance frequency [36]. Of
course, slower ramp time (i.e., smooth gradients) is effective in reducing the amplitude of the noise, but will also slow down data acquisition, which is not desirable. Recent advances in parallel imaging may also permit the use of smooth gradients while retaining the flexibility of existing EPI strategies. Recent work by de Zwart et al demonstrated about 15 dB reductions in SPL using the sensitivity encoding (SENSE) technique with a gradient-recalled echo EPI acquisition [37]. However, most MRI systems are not equipped with parallel imaging capability. Another approach for noise reduction involves the use of passive acoustic insulation to minimize sound propagation across the structural components of the scanner [38-39]. For patients, the simplest way to effectively reduce acoustic noise is achieved by utilizing passive acoustic noise control devices such as earplugs and earphones. In addition, sound-absorbing materials can be attached to reflecting surfaces such as the cryostat and the scan room walls, ceiling and floor. These hardware modifications will undoubtedly become part of the next generation of MRI systems. Such modifications will also incorporate different components and materials, improved isolation of the gradients, and/or redesigns of the coil shape, while maintaining the wide bore and rapid gradient switching capabilities of our current systems [8-9]. The use of vacuum technology to isolate the subject from the source of the noise may be effective; however, it is again very costly and increases the mechanical complexity of the resulting gradient coil design. Besides, it is nearly impossible to implement vacuum insulation in existing systems. A relatively sophisticated approach considered for the treatment of MRI noise is active acoustic noise cancellation that works in a manner comparable to current commercially available active headphones [22, 40]. Though this approach seems promising, few researchers are pursuing this direction due to the difficulty
and complexity of the hardware and software design [22, 40]. The magnetic environment limits material choices in the vicinity of the MRI scanner (for example, conventional speakers utilize magnets), and the control system design is dependent on the characteristics of the scanner acoustic noise. Since this method is frequency and position dependent, poor implementation can lead to mistuning that further increases the acoustic noise amplitude instead of suppressing it. However, with recent advances in controller design and better understanding of the nature of the sound field, novel methodologies may now be developed to make this technology more viable.
3. EVALUATION OF MRI COMPATIBLE HEADPHONES FOR ACTIVE NOISE CANCELLATION

The loud noise produced during MRI is a direct result of the static magnetic field and gradients which are essential to the procedure, which will be reviewed in this chapter. The development of an ANC system for MRI is complicated by the magnetic environment in the vicinity of the scanner. The proposed concept includes having the patient wear a set of headphones that are used to generate the cancellation signal. These headphones must also contain a microphone to measure the sound for use by the control system. A typical loudspeaker contains a magnet, rendering it inappropriate for use in this application.

A set of MRI compatible headphones containing piezoceramic speakers was modified by adding an optical microphone inside the earpiece. Under simulated conditions, the noise reduction performance of this set was compared to a high fidelity set of headphones containing traditional speakers. The results are contained in the following paper, "Evaluation of MRI Compatible Headphones for Active Noise Cancellation". This paper has been submitted to the Noise Control Engineering Journal for consideration, and is based on a conference paper [54] which was published in the Proceedings of the 2008 National Conference on Noise Control Engineering. It has been reformatted for inclusion here.
3.1 Summary

Magnetic Resonance Imaging (MRI) is a powerful medical diagnostic tool. An undesirable side effect is the loud noise produced during scanning. This noise is unpleasant at best, and it may limit imaging potential. Numerous approaches have been studied in the quest to reduce patients’ noise exposure, thus allowing for improved imaging capabilities. One approach considered is to apply active noise control. This is achieved with a set of active headphones that the patient wears, which contain a pair of MRI-compatible speakers to generate the control signal as well as an error microphone to monitor the target response. In order for this approach to reach fruition, any materials used in the vicinity of the MRI scanner must be made of non-ferrous materials due to the strong constant static magnetic field. This limits the choice of speakers and microphones, which may impede the development of the best possible system. To quantify the degree of performance achievable with MRI-compatible hardware, two different headset/microphone combinations will be evaluated. One consists of high fidelity components that are not MRI compatible, while the second utilizes piezoelectric speakers and optical microphones that are MRI compatible. Limitations and salient features of these setups that are critical to the success of applying active control will be discussed.

3.2 Introduction

Medical imaging is an important diagnostic instrument that enhances medical doctors’ ability to diagnose and treat patients. Prior to the discovery of x-rays, it wasn’t possible to “see” inside a body without invasive procedures. During the early 1900s, x-rays evolved from
being viewed as a photographic novelty to a medical necessity [41]. Throughout the century, technological innovation enhanced medical diagnostic imaging capabilities. In the 1980s, magnetic resonance imaging (MRI) was introduced for commercial use, with enhanced development continuing today. The MRI system has the advantage of producing high contrast images including soft tissue with no known side effects, such as the radiation resulting from x-rays.

The operation of MRI is based on the atomic characteristic of nuclear magnetic resonance (NMR) [42]. Nuclear magnetic resonance (NMR) has long been used as a spectroscopy technique for identifying chemical composition. The MRI’s ability to create images based on NMR is made possible by temporally and spatially altering the magnetic field, which allows the researcher to identify types of matter at a specific location. An unfortunate by-product of the magnetic field manipulation is the production of loud noise in the vicinity of the scanner, concentrated where the patient resides [2]. Sound pressure levels (SPLs) in excess of 115 dB have been measured [5, 10], creating an unpleasant and potentially harmful environment for the patient as well as any health care provider in the area. While a variety of approaches have been made to limit noise exposure by introducing some means for passive reduction or altering the scanning sequence, they often conflict with the desire for increased imaging speed and enhanced functionality [43-46].

A common tactic to address the noise issue is for the patient to wear a set of headphones that includes a microphone near the mouth. These headphones serve two purposes. First, they allow for communication between the operator and the patient during the procedure, which would otherwise be difficult if not impossible. Second, they may be used for research in the
form of instruction cues or to provide entertainment such as music (possibly utilizing video if equipped with an optional viewing screen), in order to provide a distraction and alleviate the patient’s discomfort resulting from the noise exposure during the procedure. The headphones are also designed to provide passive noise reduction.

A technique used to address a wide variety of undesirable noise environments is active noise control (ANC). A control system is designed to produce a secondary noise that is generated specifically to reduce the total sound pressure when combined with the environmental sound through destructive interference. Since the noise produced during MRI is still a problem despite the use of communication headphones, the question of whether a suitable ANC system can be developed is raised.

A successful implementation of any ANC system requires accurately reproducing the control signal. In a typical MRI procedure, the headphones needed to introduce the control signal are already in place. However, the magnetic field required for imaging limits the material choices to those that are non-ferromagnetic. This eliminates a common type of speaker consisting of a voice coil and magnet for sound reproduction. One supplier of commercial MRI headset uses speakers composed of piezoceramic materials. Unfortunately, it has been documented that actuators and speakers composed of these materials exhibit nonlinear and hysteretic behavior [47-48]. This raises the question of whether piezoceramic MRI compatible speakers are capable of reproducing the control sound accurately enough to result in a successful ANC implementation.

In order to evaluate the potential for ANC during MRI scanning, two sets of headphone/microphone combinations will be compared. One is a standard high fidelity set
that is not MRI compatible, while the second is a commercially available MRI communication set. An ANC system will be implemented in a physical simulation of the MRI environment (allowing for evaluation of the non MRI compatible set) while measurements are made to quantify their performance. The two sets of headphones/microphones will be compared to evaluate the potential for a successful MRI ANC system.

### 3.3 Magnetic Resonance Imaging Operation

The operation of MRI relies on NMR [42], an atomic characteristic. NMR is exhibited by nuclei containing an odd number of protons or neutrons when placed in a magnetic environment. Hydrogen protons are commonly used for clinical MRI due to their abundance in the human body. When a hydrogen proton is placed in a static magnetic field, it exhibits one of two spin states which have a distinct energy difference. A photon of the appropriate energy, or frequency, will induce a transition between the two states. This frequency is determined by the Larmor equation (3.1) where $\gamma$ is the Larmor ratio and $B_0$ is the strength of the static field. The Larmor ratio for $^1H$ is 42.5759 MHz/Tesla.

$$f = \gamma B_0$$  \hspace{1cm} (3.1)

Each individual nuclear spin state has a magnetic moment that is oriented at an angle to the static magnetic field and precesses about the axis of the field at the Larmor frequency. These individual spin states are aligned either with or against the static field. However, there is a slight bias to the positive spin state, resulting in a net magnetic moment that aligns with the static magnetic field as shown in Figure 3.1. This net moment ($M_0$) for $^1H$ can be
calculated using the equation 3.2,

\[
M_0 = \frac{N\gamma^2 h^2 B_0}{8\pi k T}
\]  

(3.2)

where \( N \) is the number of protons in the sample, the temperature is represented by \( T \), \( k \) is Boltzmann’s constant and \( h \) is Planck’s constant.

Subjecting the sample to an external radiofrequency (RF) pulse at the Larmor frequency will induce a rotation of the net magnetic moment away from its original alignment towards the alternate spin state. For example, given a pulse of the appropriate duration, the result will
be a rotation of the net magnetic moment to a transverse direction, as seen in Figure 3.2.

![Diagram](image.png)

**Figure 3.2: Net magnetic moment of sample at time t after a π/2 radio frequency pulse creating magnetic field B₁**

This duration is commonly described as a π/2 pulse. Following the pulse, the magnetization vector will return to its original alignment, a process known as relaxation and described by a time constant, T₁. This realignment can be measured since the magnetization vector interacts with the static magnetic field. There are additional relaxation time constants associated with the process. For example, immediately after the removal of the RF pulse, the individual spin states result in a net magnetization vector (aligned with the Y axis in Figure 3.2). The individual spin states interact with the original static magnetic field and each other, resulting in a random distribution within the transverse plane. This process is described as T₂ relaxation. These relaxation time constants vary as a function of the composition of the sample and the magnitude of the magnetic field. The RF measurements of the time-varying field allow for computation of the relaxation times and subsequent identification of the sample.

In order to generate an image of a heterogenous sample such as a portion of a living
subject, information characterizing the matter present at a specific location is needed. This is made possible by introducing spatially varying magnetic field gradients to create a relationship between location and frequency. Figure 3.3 depicts a cylinder located in a z oriented and varying static magnetic field. The plot in the lower portion of the figure shows the Larmor frequency as a function of position. Under these conditions, an RF pulse of a specific frequency will affect the net magnetic moment only at the slice where the magnetic field results in a Larmor frequency corresponding to the frequency of the pulse.

![Diagram of a cylinder in a varying magnetic field](image)

**Figure 3.3: Magnetic field gradient and spatially varying Larmor frequency**

Generation of three dimensional images is achieved by turning the three directional magnetic field variations on and off in combination with bursts of RF pulses and measurements. These very precise and rapid combinations are known as pulse sequences. A
variety of pulse sequences have been designed for specific imaging procedures. Figure 3.4 contains the three magnetic gradient pulses in X, Y, and Z directions for a gradient echo multi-slice (GEMS) sequence in the time domain.

![GEMS Pulse Sequence](image)

**Figure 3.4: Magnetic field gradient signals for time slice of GEMS sequence**

The magnetic field gradients are generated by introducing an electrical current to a coil, known as the gradient coil. Recall that this is occurring in a static magnetic field, which results in an electromotive force [49]. The current flow in the gradient coils within the magnetic field produces Lorentz forces that physically excite the structure and result in structural vibration that produces noise as the structure excites the surrounding air [1-2]. Noise levels in excess of 115 dB have been measured on a 3 tesla (T) imager [5]. Improvements to image quality and scanning time have been achieved by increasing the static
field strength and gradient switching rates, which unfortunately also increases the magnitude of acoustic noise emission [3, 11]. In addition to being unpleasant for the patient and any health care worker in the vicinity, the magnitude of the noise is great enough to be potentially damaging as well.

The resulting noise during imaging has been recognized as a problem throughout the evolution of MRI, and numerous approaches have been suggested to address the problem [40, 50-51]. Unfortunately, some approaches require fundamental changes to the scanner rendering them infeasible for existing MRI systems [13]. An ideal solution would be one that is independent of the MRI system, allowing it to be used with existing machines. Another approach which requires modification of the gradient sequences compromises the imaging speed as well as quality of the image [35]. In order to successfully implement the use of active noise control for MRI, a robust controller and control signal delivery means is needed.

3.4 MRI Compatible Headphones with Piezoceramic Speakers

The magnetic environment that is crucial to the operation of the MRI requires that equipment used in the vicinity of the scanner be non-ferromagnetic, as noted above. In particular, this disallows the use of headphones constructed with common speakers consisting of a magnet and voice coil for sound reproduction. Commercial headphones used for entertainment purposes for MRI patients commonly utilize piezoceramic speakers which are impervious to the magnetic field.
Piezoceramic materials are used in a variety of transducers and actuators. However, it has been shown that they exhibit hysteresis under quasi-static excitation as well as nonlinear response as a function of both the level and frequency of excitation when subjected to dynamic excitation [47-48].

This behavior raises the question of whether the piezoceramic speakers are adequate to achieve good results for an ANC application intended for MRI noise, and whether this behavior must be considered in developing the control system and operational procedures.

3.5 MRI Physical Environment Simulation

The non-MRI compatible headphone/microphone combination could not be tested during a live scan due to the magnetic environment. In addition, MRI operating costs are on the order of several hundred dollars an hour. Performing extensive development work requiring significant scanning would be cost prohibitive; therefore, an alternative is needed. A physical acoustic simulation of the MRI environment that also provides real time access to specific MRI operational information is needed.

First, a scan was recorded for use in the physical simulation. There are a variety of different types of MRI scans. A commonly used sequence for functional MRI (fMRI) research was selected for the study, an echo planar image (EPI) that is also notorious for its loud noise. A dummy was placed on the patient table of the scanner and introduced to a typical scanning position in the bore of the 4T MRI scanner. Microphones were located on the side of the head close to the ear position. During scanning, the microphones measured the
SPLs at the patients’ ear locations. The control system to be evaluated would also require the simultaneous playback of the gradient signals controlling the scan. These were recorded on the multi-channel recorder along with the microphone signals while the scanning was in progress. By simultaneously recording these signals on a single multi-channel recorder, it would be possible to replay all channels at once. The sound and gradient signals would have the same temporal relationship during replay that they had during the initial scanning procedure.

A sound quality chamber was utilized for the physical simulation. This room, while not as quiet as an anechoic chamber, is a controlled environment with acoustic insulation. The magnitude of SPL to be reproduced is great enough that the sound quality chamber was adequate for the purposes of this study.

A single sound channel of the recorder was connected to the stereo system in the chamber for replay. A binaural head containing microphones at the ear locations was also positioned in the chamber. It would “wear” the headphones and provide the means to measure the results during testing. The stereo settings and speaker location were adjusted to approximate the SPL measured during the original scan.

The data acquisition/control system (dSPACE GmbH, Technologiepark 25, 33100 Paderborn, Germany) is also located in the sound chamber. In addition to acquiring the microphone signal from the ear to be used as the error signal, the data channels from the digital recorder corresponding to the gradient signals were connected to the control system for use as reference signals. When the control algorithm is operating during the simulated scanning process, a secondary signal (the control signal) is output from the control system.
and played through the headphones “worn” by the dummy.

Figure 3.5 is a graphical depiction of the simulated MRI environment. This setup allows for the MRI scan to be run repeatedly during development and testing without the cost and inconvenience of working at the actual scanner.

![Diagram of MRI simulation setup]

Figure 3.5: Physical simulation of MRI environment for testing (note: filters and amps not shown)

### 3.6 Active Noise Control System

During earlier investigations into MRI noise, it was observed that there is a correlation between gradient signals and MRI noise [10]. For the EPI scanning sequence that was selected for axial slicing, the Y-gradient “current” signal and the first harmonic of the recorded sound both have a peak at the same frequency, which was targeted for this study. Figure 3.6 is a plot comparing the frequency spectrum of the Y-gradient signal and the measured noise (note that the Y-axis scale is removed since this is for subjective comparison
only) that shows a strong correlation between the Y-gradient signal and the generated MRI noise at that frequency. The target frequency for reducing response is approximately 850 Hz.

![Comparison of Y Gradient Spectrum to MRI Noise](image)

**Figure 3.6:** Frequency spectrum comparing MRI noise SPL to Y-gradient current signal

The control system algorithm implemented on the control system is a filtered-x least mean squares (FXLMS) feedforward system [52]. A diagram of the proposed control system is shown in Figure 3.7. The sound being reproduced by the stereo is designated as d(n), while the Y-gradient signal is the reference signal r(n). The ear microphone measurement which is targeted for reduction is the error signal, e(n).
The ANC algorithm running on the control system outputs a control signal that is played through the headphones. Unfortunately, it is not possible to perfectly reproduce the control signal generated by the controller because the headphone speaker, microphone, and any amplifier and filters utilized have their own system dynamics. The relationship between the controller output and the resulting error sensor measurement is known as the secondary path, $S(z)$. Hence, to maximize the potential of the algorithm, the dynamics of the secondary path must be measured and accounted for in the algorithm. This is specifically why the FXLMS algorithm was developed [53]. It compensates for the system dynamics involved in reproducing the control sound by updating the weight values of the adaptive filter.

When the ANC algorithm is operating, the ear microphone is measuring the combination of the original MRI noise and control signal output by the headphones. This error signal, $e(n)$, is what the patient hears and is targeted for minimization by the control algorithm.
3.7 Experimental Test

The high fidelity headphone used for the testing was a Sennheiser HD580 Precision model. The microphone used for measurements was a calibrated Bruel&Kjaer type 4189-L-001. Figure 3.5 shows the test setup, with the ear microphone contained in the binaural head. The MRI compatible headphone tested is a Resonance Technology, Inc. (Northridge, CA, USA) headset containing piezoelectric speakers that is typically used for fMRI study, patient communication and entertainment during scanning. Instead of using the binaural head microphone, a calibrated Optoacoustics optical microphone was added to the headset to allow for noise measurements at the ear. This combination resulted in a wearable headphone/microphone combination that is MRI compatible and could subsequently be used to acquire data during an actual scan at the MRI facility in the future.

Figure 3.8 shows the MRI compatible headset, with the right photograph showing a close up of the optical microphone installed in the earpiece.

![Image of MRI compatible headset with microphone inset]

**Figure 3.8: Proposed MRI compatible headphone/microphone set**
The data acquisition portion of the control system includes an analog to digital converter (ADC) for input signals and a digital to analog converter (DAC) for output signals. In this study, the digital sampling rate was 6 kHz. To avoid aliasing problems due to the conversions, the microphone and control signals were passed through a 2.5 kHz analog low-pass filter.

Each set was tested by following the same procedure. The first step was to obtain the secondary path, referred to as system identification. The controller outputs a white noise signal to the headphone and captures the microphone response measurement. Utilizing an adaptive finite impulse response (FIR) filter and the least means square (LMS) algorithm, the model of the secondary path in FIR form is obtained. Next, the digital recording of the EPI scan was replayed without the control algorithm activated, but the headphones in position allowing for passive reduction. The resulting measurement represents what a patient would hear under normal circumstances. Finally, the replay of the scan was repeated with the controller activated. The two transient measurements, with and without the controller active, were processed with the fast Fourier Transform (FFT) and compared in the frequency domain.

The tests were repeated for each set multiple times to ensure the SPL reduction was maximized and that the results were repeatable. During the course of testing, it was observed that the performance of the MRI compatible set was sensitive to the system identification procedure, while the high fidelity set was not. This is assumed to be a result of the nonlinear behavior of the piezoceramic material. The estimated model of the secondary path is a linear model at the level of stimulus present during the identification procedure. To best match the
dynamics of the estimated secondary path model and real plant, it is preferable to adjust the power of the white noise to an output level comparable to that present during the replay of the MRI noise.

### 3.8 Test Results

Figure 3.9 contains a plot comparing the SPL of the two measurements for the high fidelity set. The solid black line represents the uncontrolled sound, while the dashed blue line shows the result with the controller active. Note the reduction in SPL in the vicinity of 850 Hz, which is the targeted frequency based on the Y-gradient reference signal.

![High Fidelity Headphone and Microphone](image)

**Figure 3.9: High fidelity headphone/microphone noise reduction results**
Figure 3.10 has a corresponding plot for the MRI compatible set. Note the lower magnitude response for the uncontrolled sound compared to the high fidelity results in Figure 3.9. This is due to the different headphone employed. The high fidelity set is designed for high quality reproduction with no consideration for passive attenuation. The MRI compatible set is designed with a goal of achieving passive reduction of sound for the patient during scanning. As for the SPL reduction of the MRI compatible set, once again, there is a significant reduction at the targeted frequency.

Figure 3.10: MRI compatible headphone/microphone noise reduction results
Table 3.1 contains the SPL measurements for the two different headphone/microphone combinations. The SPL at the targeted frequency of 850 Hz is shown along with the overall SPL over the frequency range from 800 to 900 Hz. At the targeted frequency of 850 Hz, the reduction with the high fidelity set is 34.1 dB, while the MRI compatible set is 29.3 dB. However, the additional passive reduction characteristic of the MRI compatible headphone results in a 10.3 dB lower SPL for the patient. Over the frequency range of 800 to 900 Hz, the SPL is reduced by 25.3 dB and 22.3 dB, respectively for the two sets, with the high fidelity set once again achieving the greater reduction. However, the magnitude of SPL is 12.2 dB less with the MRI compatible set. While the MRI compatible set does not reduce the SPL as much as the high fidelity set when the ANC is active, the patient would be exposed to lower SPL due to the additional passive reduction with the MRI compatible set when compared to the high fidelity set.

<table>
<thead>
<tr>
<th></th>
<th>High fidelity set</th>
<th>MRI compatible set</th>
</tr>
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<tbody>
<tr>
<td>Frequency (Hz.)</td>
<td>850</td>
<td>800-900</td>
</tr>
<tr>
<td></td>
<td>850</td>
<td>800-900</td>
</tr>
<tr>
<td>SPL with ANC off (dB)</td>
<td>106.8</td>
<td>110.8</td>
</tr>
<tr>
<td>SPL with ANC on (dB)</td>
<td>72.7</td>
<td>85.5</td>
</tr>
<tr>
<td>Reduction (dB)</td>
<td>34.1</td>
<td>25.3</td>
</tr>
</tbody>
</table>

Table 3.1: Measured sound pressure levels (dB) of high fidelity and MRI compatible headsets
3.9 Conclusions

The physical environment of an MRI was simulated in a sound quality chamber to aid in the development of an ANC system that could eventually be utilized with an actual MRI scanner. Two sets of headphone/microphone combinations were compared, one for the purpose of achieving the best possible sound reproduction and measurement to show the potential of the control system, the other to compare the MRI compatible set to check the potential for noise reduction on an actual scanner. The MRI compatible headphone/microphone combination is capable of achieving significant SPL reduction in the sound chamber configured to simulate the physical environment and operation of the 4T MRI scanner. The MRI compatible set did show sensitivity to the SPL during system identification. In order to maximize the performance of the MRI compatible set, the white noise used for system identification needed to produce a noise level comparable to the SPL experienced by the patient during scanning. The high fidelity set was not as sensitive. While the reduction with ANC is not as great as the high fidelity set, the end result of lower SPL for patient exposure is achieved.
4. FEEDFORWARD ACTIVE NOISE CANCELLATION FOR MRI UTILIZING REFERENCE MICROPHONE

Initial results for noise cancellation covered in chapter 3 focused on a common scanning sequence, EPI. After achieving good results for EPI using the three MRI gradient signals as references, an additional scan type, GEMS, was evaluated. Only one gradient signal could be successfully processed at once with the GEMS scan, limiting the amount of reduction that could be achieved. In order to improve the reduction for this scan type, the approach was modified. Rather than rely solely on the gradient signals as references, a microphone located outside the earpiece but close to the headset was evaluated for use as an additional reference signal. A second set of MRI compatible headphones was modified by adding condenser microphones inside the earpiece for use as the error signal, and outside the earpiece for use as a reference signal.

The following paper documenting the results of this evaluation, “Feedforward Active Noise Cancellation for MRI Utilizing Reference Microphone”, was published in the proceedings of ACTIVE 2009 [55]. It has been reformatted for inclusion here.
4.1 Summary

Magnetic Resonance Imaging (MRI) is a powerful medical diagnostic tool. Unfortunately, the loud noise produced during scanning is unpleasant and potentially harmful to the patient and may limit imaging protocol. Numerous approaches have been proposed to reduce noise exposure. In this study, an active noise control system that generates a secondary sound signal fed into a set of headphones worn by the patient is proposed to suppress the noise coming from the MRI system. Our specific effort is to examine the feasibility of feedforward versus feedback algorithms. Even though feedforward ones have been shown to have more potential for noise reduction, they require a coherent reference signal. While pulse gradient signals showed promise as a reference signal, the amount of cancellation varies for different scanning sequences. Alternatively, a microphone attached to the outside of the headphones was evaluated as a reference signal. To quantify the performance of these two reference signals, noise due to two common scanning sequences were simulated in a sound quality chamber containing a binaural head wearing a pair of MRI compatible headphones. The noises at the patients’ ear were measured with and without active control operational, and their performance compared.

4.2 Introduction

MRI is an imaging procedure for medical diagnosis that has two distinct advantages over other methods. First, the resulting image is capable of providing more soft tissue
contrast than other common imaging procedures including x-ray and computer aided tomography (cat) scans. Second, there is no potentially harmful side effect with MRI; x-rays, for example, rely on radiation which can be hazardous with too much exposure [41].

The imaging procedure requires placing the patient in a high strength static magnetic field and altering the field rapidly at localized points through the use of gradient coils and amplifiers. To achieve this, an electrical current is transmitted through the gradient coils, which results in Lorentz forces due to the static magnetic field. The magnitude and speed of the changing Lorentz forces causes deformation of the structure. The structure behaves like a loudspeaker, producing noise in the surrounding air [2]. The sound frequency spectrum and sound pressure level (SPL) are a function of the static field strength and scanning parameters. The SPL is great enough to be annoying and potentially hazardous to the patient. SPLs in excess of 115 dB have been measured on a 3 Tesla scanner [5].

As the technology of MRI evolved, improved imaging potential has been achieved at the expense of increasing SPL to the point that sound generation is a consideration in the selection of scanning parameters. Numerous approaches that have a goal of reducing the patient’s exposure to noise have been granted patents. Many of these approaches require changes to the underlying scanner/coil design which may not be economically feasible to implement on existing scanners [50].

Active Noise Control (ANC) has been demonstrated as a potential solution for noise reduction under selected scanning conditions [57]. However, MRI technicians have a wide variety of scanning parameters available depending on the diagnostic procedure. These may result in vastly different sound frequency spectrum characteristics. A control system that
works well for one type of scan may not achieve the same results for an alternate scan type.

First, a previous demonstration of ANC applied to sound from an echo planar image (EPI) scan that utilizes the gradient signal as a reference is reviewed. Next, this implementation is used with another common scanning sequence, gradient echo multi-slice (GEMS). Finally, a microphone is substituted as the reference signal and the results compared to those obtained with the gradient signal reference.

4.3 Experimental Setup

The hardware utilized for running the noise control system to obtain experimental data is a dSPACE system (dSPACE GmbH, Technologiepark 25, 33100 Paderborn, Germany). It consists of two primary components. One is a processor that is controlled by a laptop computer. The other is an I/O unit that contains an analog to digital converter (ADC) for input signals and a DAC for output channels. The control system was developed in Matlab Simulink (The MathWorks, Inc., 3 Apple Hill Drive, Natick, MA, USA). The algorithm implemented for these experiments is a feedforward system. Kuo and Moran have proposed using this type of system when a reference signal is available that has a high level of coherence to the sound being targeted [52]. In the classic duct example shown in Figure 4.1, a microphone measures the offending sound before it reaches the cancellation location, and uses this measurement as the reference signal.
However, Kuo and Morgen also suggest that alternate non-acoustic reference signals may be used. In the case of MRI scanning, it has been shown that there is a relationship between the gradient signals and the resulting noise, making them a potential candidate for use as a reference signal [10]. There are two other important components necessary for an effective feedforward filtered-X Least Mean Square (FXLMS) system that is an extended version of a standard LMS algorithm. First, the addition of an adaptive filter to generate the cancellation signal. Second, the secondary path effect must be accounted for. The secondary path is made up of all the components needed to reproduce the cancellation signal, including the headphones and microphones as well as amplifiers and filters. The basic version of the adaptive LMS algorithm will potentially diverge due to the system dynamics of these various components, requiring compensation for its effect. This is achieved by applying a model of the secondary effect to the reference signal before input to the least means square calculation. The resulting control system, known as the FXLMS algorithm, is shown in Figure 4.2.
MRI systems are expensive, and conducting numerous experiments during live scans would lead to significant cost. In order to reduce the expense during development, an alternative was needed. A physical simulation of the MRI environment including noise and gradient signals was prepared. During a live scan on a 4 T scanner, microphones were placed at the ear locations of a dummy that was inserted into the imager. The acoustic signals were recorded on a digital recorder simultaneously with the gradient signals.

Figure 4.3 shows the dummy at the MRI scanner ready for data acquisition.
Figure 4.3: Dummy with microphones for live MRI scan data acquisition

In a sound quality chamber, the recording is played back with the measured sound being amplified by a stereo system to replicate the sound environment, while the gradient signals are fed directly to the dSPACE system. All recorded channels are replayed simultaneously with a goal of representing the actual MRI scanning environment as accurately as possible. Figure 4.4 contains a schematic of the experimental setup used in the sound quality chamber for the physical simulation performed to measure the SPLs present.
Figure 4.4: MRI physical simulation schematic
4.4 ANC Utilizing Gradient Reference For EPI Scan

In a previous study [54], a set of MRI compatible headphones/microphones were subjected to EPI scanning noise in the sound chamber. Since it was recognized that the Y gradient was coherent with the first significant response peak at approximately 850 Hz., the Y gradient was used as the reference signal.

Figure 4.5 shows a comparison of the SPL measured with and without the ANC system active. Note the reduction in the vicinity of 850 Hz.

![MRI Compatible Headphone and Microphone](image)

**Figure 4.5: SPL at ear during EPI scan simulation**
4.5 ANC For GEMS Scan

In order to further evaluate the potential for MRI noise reduction with gradient reference signals, another common scanning sequence (GEMS) with a different frequency spectrum was evaluated. The GEMS spectrum contains many more harmonic peaks over the same frequency range when compared to the EPI scan. The sound frequency spectra of all three gradient signals contain corresponding peaks of varying magnitudes at each of these harmonics. This differs from the EPI scan which had noticeably different frequency spectra for the different gradients. The SPL at any given peak for the GEMS scan was not dominated by a single gradient, which was the case for the EPI scan. The noise reduction for the EPI scan was optimized by adjusting the step size of the FXLMS algorithm. In the case of the GEMS scan, adjusting the step size for reduction at a specific harmonic did not necessarily provide reduction at other harmonics and in some cases increased the sound due to divergence at the frequency was in question. This characteristic is generally known as a frequency dependent step size.

Figure 4.6 shows an example of the noise reduction achieved using the X gradient signal for reference. Note the sound reduction at approximately 750 Hz achieved with the control signal on, while at 500 Hz the SPL increased.
Figure 4.6: SPL at ear during GEMS scan simulation with gradient reference

In the quest for better results, an alternative headset containing an outside microphone to use as the reference signal in place of the gradient was substituted.

Figure 4.7 shows a picture of the new headset containing the outside microphone for reference.

Figure 4.7: MRI compatible headphones with outside reference microphone
The advantage to this approach is that the reference signal is more coherent to the noise at the ear, the targeted reduction location. However, there are potential disadvantages to this approach. Since the reference microphone is close to the target reduction location, the time for the algorithm to generate the cancellation signal is limited. It also creates the potential for feedback if the reference microphone picks up the cancellation signal. During physical simulation of the GEMS scan, the SPL reduction was dramatically improved compared to that achieved with the X gradient reference.

Figure 4.8 shows a comparison of the SPLs measured utilizing the outside microphone as the reference signal for the feedforward control system.

![GEMS Physical Simulation Outside Microphone](image)

**Figure 4.8: ANC results for GEMS scan with outside microphone for reference**
4.6 Conclusions

An ANC system has been evaluated for the application of sound pressure level (SPL) reduction for MRI patients. The specific implementation that utilized one of the gradient signals as a reference signal performed well on sound from an EPI scan. However, the SPL reduction achieved when using the same implementation with another common scan, GEMS, which generates a very different frequency spectrum, did not perform as well. An alternate implementation that utilizes a microphone mounted onto the outside of the headset as the reference was created. With the alternate reference, the SPL reduction for the GEMS scan was seen to improve dramatically.

The feedforward control system developed for use in applying ANC to MRI acoustic noise shows the potential for providing substantial SPL reduction at the patient ear location. Maximizing its potential will require adjusting the control system by using different reference signals for various scanning sequences. This effort is still on-going.
5. IN-SITU ACTIVE NOISE CANCELLATION APPLIED TO MAGNETIC RESONANCE IMAGING

Due to the high cost of operating the MRI scanner, much of the ANC system development was conducted in a sound chamber, where a physical simulation of the MRI environment was created. This was achieved by recording noise and gradient signals on a multi-channel digital recorder during live scanning, and then replaying them in the sound quality chamber. The ability of the physical simulation to adequately emulate the MRI operation and environment was unknown.

Once the system was developed to the point where it performed consistently with good reduction in the simulation chamber, the ANC system was taken to the College of Medicine’s Center for Imaging Research, where its performance could be evaluated during in-situ operation of the MRI. The following paper documenting the results of an initial evaluation to validate the similarities between the simulated and actual environments, “In-situ Active Noise Cancellation Applied to Magnetic Resonance Imaging”, has been submitted to the ASME Journal of Vibration and Acoustics for publication consideration. It is based on a conference paper of the same name that was accepted for publication at the 2009 ASME International Mechanical Engineering Conference and Exposition [57]. It has been reformatted for inclusion here.
5.1 Summary

Magnetic Resonance Imaging (MRI) is a powerful medical diagnostic tool. Unfortunately, the loud sound produced during scanning is unpleasant, potentially harmful to patients and may limit imaging protocol. Previously, a variety of approaches have been proposed to reduce noise exposure with limited success. This work is directed at the application of an active noise control system. To this end, prior studies have been conducted in a sound quality chamber to aid in the development and implementation of the hardware, algorithms and procedures, which resulted in an active noise cancellation system tailored to conditions present during MRI. The active noise control system generates a secondary sound signal fed into a set of headphones worn by the patient. This system performs well during physical simulation of scanning conditions. In this study, the headphones are worn by a dummy during in-situ MRI scanning. Our specific effort is to take a selected set of successful experiments under simulated conditions and repeat it during live scanning to evaluate the real time performance of the system conducted in-situ. Evaluation of a common scanning sequence was conducted and the procedure adjusted to maximize the performance of the system. The sound pressure levels at the patient’s ear were measured with and without active control operational, and the results were compared to evaluate the active noise cancellation system’s performance during live scans.
5.2 Introduction

Magnetic Resonance Imaging (MRI) is an imaging procedure for medical diagnosis that is capable of providing enhanced soft tissue contrast in comparison to other common imaging procedures such as x-ray. Unlike x-rays, which rely on radiation, MRI has no known potential harmful side effects due to extended exposure [41]. Therefore, for many medical imaging procedures, MRI is highly desirable.

MRI relies on the characteristic of nuclear magnetic resonance [42]. This technique has long been used to aid in identifying chemical composition through spectroscopy. In order to produce images of an item composed of varying types of matter, such as the human body, MRI introduces magnetic field gradients to allow for spatial identification as well as chemical composition. During imaging, patients are placed in a high strength static magnetic field. Three different gradient signals (X, Y, Z) are available to introduce a variation in the magnetic field in different directions. The gradient signals are rapidly altered by transmitting electrical current through a set of gradient coils. The gradient signals are combined with precisely timed radiofrequency transmissions to create a scanning pulse sequence that manipulates the MR magnetization needed to compose the desired image(s). A variety of scanning pulse sequences have been developed for different imaging protocols.

The electrical current passing through the gradient coils produces Lorentz forces due to its interaction with the static magnetic field. These forces vary with time as the current changes, subjecting the structure to rapidly changing dynamic loads and deformation. The resulting vibration of the structure causes it to behave like a loudspeaker, producing acoustic
noise in the surrounding air [1-2]. The resulting sound pressure level (SPL) is a function of the static field strength and scanning parameters. The SPL is both annoying and potentially hazardous to the patient and anyone else in close proximity to the scanner. Sound pressure levels in excess of 115 dB have been measured on MRI with a magnetic field greater than 3 Tesla (T) [5, 10].

As MRI has evolved, improvements have been achieved by introducing higher static magnetic field strengths and new scanning sequences. Unfortunately, this has typically been at the expense of increasing SPL. Even when sound generation is specifically considered as a design goal when developing the MRI system, the result has been to compromise the imaging speed and/or functionality, neither of which is desirable [43-47]. The problem of sound generation during MRI scanning is well recognized. In addition to altering the scanning pulse sequence, others have approached the problem by additional means, as evidenced by a number of patents that have been granted for devices and new designs that specifically target the reduction of the patient’s noise exposure. Unfortunately, many of these approaches require changes to the underlying scanner/coil design which may not be economically feasible or possible to implement for use with existing scanners [50].

Another approach that is used for noise problems is Active Noise Control (ANC). This approach has also been suggested for use with MRI [40, 51]. While these earlier approaches did not result in a commercially successful product, the continued evolution of faster computer hardware and new software algorithms increases the potential for such a system. An ANC system composed of components that are compatible with the magnetic environment has been demonstrated as a potentially viable solution for noise reduction during simulated
scanning conditions in a controlled environment [54]. This system was also successfully demonstrated during a live scan for a single dominant frequency component of the MRI noise [56]. While that test resulted in a large reduction at the major harmonic and was encouraging, the overall reduction across the audible frequency range was limited due to the narrow frequency range targeted by the controller for that study. Enhancing the control system to be effective across an expanded frequency range with the goal of producing a noticeable reduction in noise levels across the audible range is the next logical step in developing a robust control system. In order to further enhance this system and improve its potential for patient use, the system was refined and experiments were conducted during scanning at an MRI facility in-situ, which is the primary focus of the present study.

5.3 Active Noise Control System

The ANC system developed specifically for MRI use is comprised of the dSPACE DS1103 PowerPC controller board (dSPACE GmbH, Technologiepark 25, 33100 Paderborn, Germany) running a feedforward control algorithm developed in the Matlab Simulink software package (The MathWorks, Inc., 3 Apple Hill Drive, Natick, MA, USA). The Simulink control algorithm is compiled on a computer laptop and loaded on the dSPACE processing board. This processor has a high speed connection to the dSPACE input/output (I/O) component that also converts the analog signals to digital and vice versa (ADC and DAC). This hardware configuration allows for fast calculation and communication of the control signal to the MRI compatible headset. This headset, provided by Resonance
Technology (Northridge, CA), is a commercially available set originally used for entertainment purposes that has been modified with the addition of microphones for communication purposes. The dSPACE controller hardware employed in this study is illustrated in Figure 5.1.

![Image of dSpace hardware](image_url)

**Figure 5.1: Photo of the dSpace hardware comprising of I/O controller with ADC/DAC (above) and processor (below)**
The algorithm implemented in the control system for the proposed ANC is a feedforward filtered-X Least Mean Square (FXLMS) system [52], as shown in Figure 5.2.

![Diagram](image)

**Figure 5.2: Applied FXLMS active noise control system diagram**

A feedforward control system is typically utilized in applications where a coherent and advanced reference signal is available. The performance of the control system is dependent on the coherence between the reference signal and the unwanted acoustic noise. For the broadband noise, to satisfy the causality, the reference signal should be advanced in time in comparison to the unwanted acoustic noise. Just like the classic duct example, the reference microphone should be placed upstream of the canceling speaker to ensure that the reference sound signal is sensed before it reaches the cancellation location.

Another important aspect of the FXLMS control system is accounting for the imperfect sound reproduction when generating the control signal. After the control signal is calculated by the controller, it is reproduced using several components including filters, amplifiers,
speakers, etc., which each has its own system dynamics. This combination of components is collectively referred to as the secondary path, \( S(z) \). The FXLMS algorithm has been derived to compensate for the effect of secondary path dynamics [53]. Compared to the conventional Least Mean Square (LMS) algorithm, a model of the secondary path is added and applied to the reference signal prior to the LMS calculation. The calculated control signal will then account for the secondary path effects resulting in the desired cancellation signal being output by the speaker.

In some applications, a non-acoustic reference sensor is suggested, because the non-acoustic references have a potential advantage due to the lack of feedback from the control speaker to the reference microphone [53]. However, good coherence between the reference and resulting sound is still necessary for noise reduction. This is typically more achievable using an acoustic noise reference sensor or a combination of acoustic and non-acoustic type.

Lee and his co-workers have shown that there is a relationship between the MRI gradient signals and the resulting noise [10]. They utilized the individual gradient impulse measurements for an echo planar imaging (EPI) scan to develop transfer functions between the gradient signal and resulting sound. The response is repeatable and highly linear over a range of impulse magnitudes. These transfer functions were then utilized to synthesize the overall acoustic response for the scanning sequence. The predicted response and the response measured during the actual scan were comparable.

In addition, for the EPI scan, they showed that each of the individual gradients produced a particular portion of the resulting sound spectrum. They reported the X gradient
was generally responsible for even harmonics, while the Y gradient produced odd harmonics. The contribution due to the Z gradient was more broadband in nature. This leads to the expectation that if a specific gradient is used as a reference it will reduce the sound only over select portions of the frequency spectrum, which correspond to the sound that reference is responsible for producing. It also raises the possibility that processing multiple references in parallel may result in reduction across a wider frequency range compared to just a single reference.

Figure 5.3 shows the MRI compatible headphones used for the in-situ test.

![MRI compatible headphones with outside reference microphone](image)

**Figure 5.3: MRI compatible headphones with outside reference microphone**
Any materials introduced to the vicinity of the MRI must be non-ferromagnetic. Magnetic materials could become projectiles in the strong magnetic field, and may also produce interference in the image. These headphones were modified from a commercially available set specifically designed for use in MRI patient communication and entertainment. In addition to being composed largely of plastic, the headphones contain piezoceramic speakers and aluminum wiring. They were modified by adding condenser microphones. The microphone mounted to the outside of the earpiece will be used as a reference signal in addition to the gradient signals. Not visible in the figure is an error microphone mounted on the inside of the earpiece, which will be targeted for reduction.

Each of the four references is processed independently, and their individual control signals combined to generate the total cancellation signal. This improves the control system performance for two reasons. First, as mentioned previously, the three gradient signals are each responsible for different portions of the resulting audio spectrum. Second, it is possible to tune each individual algorithm differently to optimize its performance.

Optimizing the performance is advantageous for two reasons. When the algorithm has converged and is in steady state, the maximum reduction is desired. In addition, the rate at which the converged state achieved is influenced. This is particularly important since some scanning sequences are not temporally constant. In the case of the EPI scan used for this test, it consists of approximately 1.8 seconds of sound followed by 0.2 seconds of silence. Figure 5.4 shows a plot of the transient sound measurement of an EPI scan for an uncontrolled signal.
Figure 5.4: Measured transient in-situ acoustic noise response of the EPI scan

For other scan types, the length of the silent interlude may exceed the time of loud sound. If the convergence rate is too slow, good reduction may not be achieved before the silent period occurs, causing the algorithm to deviate and then begin to converge again when the next loud period commences.

The FXLMS algorithm utilizes an adaptive filter that is updated at each time step. In the equation 5.1 given below, the adaptive filter $w$ is updated with the product of the step size $\mu$, the filtered reference signal $x'$, and the error measurement $e$,

$$w_i(n+1) = w_i(n) + \mu x'(n-l)e(n)$$ (5.1)

The step size can be increased to improve the rate of convergence, which enhances the performance of the algorithm. If the step size is too small, there may not be enough time for
the adaptive filter to converge to an optimal state during a single burst of sound during the transiently varying operating noise. Increasing the step size for rapid convergence may allow this to happen. However, there is a limit to the amount the step size can be increased before causing instability in the algorithm. The range of stable step size values is given by [53],

$$0 < \mu < \frac{2}{LP_{x'}}$$

(5.2)

where $L$ is the length of the adaptive filter and $P_{x'}$ is the power of the filtered reference signal $x'$. Since the power of each of the reference signals may differ, the value of $\mu$ that provides the best performance while maintaining stability will also vary. Implementing multiple algorithms with different reference signals in parallel allows for each adaptive control filter to be independently optimized for a particular frequency range to maximize overall performance of the system. Figure 5.5 shows the flowchart of the complete control system applied in the in-situ testing effort.

![Flowchart of the complete control system](image)

**Figure 5.5: Multiple-reference FXLMS active noise control system diagram**
Each FXLMS block represents the details of the active control algorithm including system identification model and adaptive filter from Figure 5.2, which has been coded into the dSPACE processor.

### 5.4 Experimental Results

In-situ tests were conducted on a 4T MRI (Varian, Palo Alto, CA). The photograph in Figure 5.6 shows a dummy with headphones ready to be inserted into the bore for scanning.

![Image](image.png)

*Figure 5.6: Dummy wearing headphones inserted in the 4T MRI scanner*
The rest of the equipment was set up in the adjacent control room that is outside the MRI magnetic field. In addition to the controller hardware, an anti-aliasing filter and amplifier were utilized for signal processing. Inputs to the controller includes the three gradient signals output by the MRI controller, the outside microphone reference signal, and the inside microphone error signal. The output was the control signal that was played over the headphone speaker. In order for the control system to run efficiently, the sampling rate was limited to 10 kHz for the ANC system. A digital recorder simultaneously acquired the microphone signals at 48 kHz. This data was used for post-processing of results and for performing overall SPL calculations that encompassed the full audible spectrum of 20 Hz to 20 kHz.

An EPI scan was conducted on the 4T MRI. Figure 5.7 shows the frequency spectrum up to 10 kHz, which was recorded without the ANC system.
The response at higher frequencies in the audible spectrum continued to decline and is not considered to be a significant contributor to the noise. The overall SPL near the patient ears but outside the headphones was 117 dBA, while the SPL inside the headphones at the ear location was 99 dBA.

The results in Figure 5.7 demonstrate the passive reduction achieved, which is more substantial at higher frequencies. However, the highest SPLs occur at lower frequencies where the passive reduction of the headphones is not as effective. Fortunately, the ANC system is more effective in the lower frequency range, which complements the passive reduction at higher frequencies, allowing for reduction in SPL exposure for the patient across a broad audible frequency range.
First, a set of tests were performed using just a single reference at a time. Each of the gradients (X, Y, Z) as well as the outside microphone was used as a reference, which gives a total of four sets of measurements. All of the following results are for the microphone inside the headphone near the patient’s ear, which were processed from the recorder that was sampling at 48 kHz. Figure 5.8 through Figure 5.11 each contain two plots. The upper plot is the normalized power spectrum for the specific reference used for the ANC operation. The lower plot shows the SPL of the ANC results for each of these compared to the SPL without the active control system operating. The display of the reference power and SPL in tandem allows for the comparison of the frequencies where the ANC system is effective for the specific reference signal. Based on the observations by Lee, et al. [10], and the FXLMS algorithm discussion, the effective reduction frequencies should correspond to those with higher power.

Figure 5.8 shows the results obtained using the X gradient as the sole reference. It was most effective at three frequencies, specifically 850, 1700, and 3400 Hz. The reduction of each of these approached 20 dBA at the peak spectral line. Due to the limited frequency range of reduction the overall A-weighted SPL reduction in the audible spectrum was limited to 3 dBA only.

The results for the sole Y gradient reference are in Figure 5.9. The most significant reduction using the Y gradient as the only reference was 30 dBA at 850 Hz. There is almost 20 dBA reduction at 2500 Hz as well. Once again, the limited frequency range of reduction limits the overall A-weighted SPL reduction to slightly less than 3 dBA.
Figure 5.8: Results of (a) EPI scan X gradient power and (b) in-situ test results
Figure 5.9: Results of (a) EPI scan Y gradient power and (b) in-situ test results
While the X and Y gradient SPL reductions were significant over small frequency ranges, utilizing the Z gradient as reference yields more of a broadband reduction up to 1500 Hz, as seen in Figure 5.10. At numerous spectral lines, about 10 dBA reduction was achieved, with an overall A-weighted SPL reduction over the audible spectrum of a little more than 1 dBA.

The final test using an individual reference utilized the outside microphone. These results are shown in Figure 5.11. The ANC system performance for the microphone is affected by the reduced delay between the reference and cancellation signals compared to the gradient references. Unlike the gradient signals which contain only certain frequency spectra, the microphone reference signal has a similar frequency response compared to the error microphone. The reduction of SPL using the microphone reference was most noticeable around the frequencies of 850 and 1300 Hz, with modest reductions at 350 and 2500 Hz. The audible spectrum overall A-weighted SPL reduction for the microphone reference was about 4 dBA.
Figure 5.10: Results of (a) EPI scan Z gradient power and (b) in-situ test results
(a) Control off; Control on

(b) Control off; Control on

Figure 5.11: Results of (a) EPI scan microphone power and (b) in-situ test results
Each of these individual reference tests demonstrates substantial reduction over a select frequency range, with a modest reduction when considered across the entire audible spectrum. With the exception of the 850 Hz harmonic, there is limited overlap in the frequency range of effective reduction for an individual reference when the four distinct tests are compared. The final measurement was made running the four distinct algorithms simultaneously, each of which utilized one of the references from the individual reference tests. The step size for each of the individual algorithms was initially set to the optimized value determined during the individual reference tests. However, this resulted in instability at the 850 Hz harmonic where there was an overlap in effective reduction from the individual reference tests. Step size adjustments were then made to avoid this instability while simultaneously achieving the best overall reduction. The frequency spectrum for the multiple reference ANC test is shown in Figure 12. The plots show fairly substantial reduction over a broad band up to 1700 Hz as well as the peaks that occur at 2500 and 3400 Hz. The overall A-weighted SPL reduction was 11 dBA.
Figure 5.12: EPI in-situ test results for combined response of the microphone and X, Y, Z gradients as references (— Control Off; ………… Control On)
Table 5.1 contains a summary of the measurements made with the inside microphone located near the patient’s ear. Column 1 contains the overall A-weighted sound pressure levels for the audible spectrum (20 Hz - 20 kHz) for each of the tests, while column 2 lists the reduction in dBA achieved by the ANC system.

<table>
<thead>
<tr>
<th>References</th>
<th>SPL (dBA)</th>
<th>Reduction (dBA)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Control Off</td>
<td>99</td>
<td>-</td>
</tr>
<tr>
<td>X Gradient</td>
<td>95</td>
<td>3</td>
</tr>
<tr>
<td>Y Gradient</td>
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</tr>
<tr>
<td>Z Gradient</td>
<td>97</td>
<td>1</td>
</tr>
<tr>
<td>Microphone</td>
<td>95</td>
<td>4</td>
</tr>
<tr>
<td>XYZ Grad &amp; Mic</td>
<td>88</td>
<td>11</td>
</tr>
</tbody>
</table>

Table 5.1: Sound pressure measurements at patient's ear location
5.5 Conclusions

An active noise control system specifically designed for MRI patients has been developed. It consists of a feedforward control algorithm running on a high-speed processing board that is interfaced with the imager electronic unit to receive gradient signals for use as reference in the active controller. In-situ tests were conducted during a live EPI scan on a 4T imager. The best performing test configuration utilized all three gradient signals as well as an outside microphone simultaneously as references. Overall A-weighted SPL reduction of 11 dBA was achieved within the audible spectrum with the reduction at select frequency spectral lines approaching 40 dBA. Test results also show that the current system is not yet optimized for some frequency ranges. To improve the performance of the system further, effort is ongoing to refine the performance through either optimizing additional specific frequency ranges or using frequency domain algorithms that can tune and optimize the system within the different frequency ranges.
6. IN SITU ACTIVE CONTROL OF 4 TESLA MRI SCANNER NOISE

In the previous chapter, the ability of the physical simulation to adequately emulate the MRI operation and environment was demonstrated by measuring the results during live MRI scanning. The noise reduction was very good up to around 2 KHz, with limited effect above that frequency where substantial noise levels were still present, raising the question of how to improve the performance at higher frequencies. The controller was updated with the goal of achieving noise reduction at higher frequencies. In this chapter, the in-situ results of the upgraded controller are presented.

Following is a complete late stage draft of a paper that has been prepared for submission to a yet to be determined journal for publication consideration. It has been reformatted for inclusion here.
6.1 Summary

The acoustic noise generated by an MRI scanner during operation is extremely annoying and potentially hazardous to patients and healthcare workers. Hence, an effective noise control approach for MRI noise is highly desirable. The purpose of this study is to evaluate the effectiveness of the proposed active noise control system for the reduction of the acoustic noise emission generated by a 4 Tesla MRI scanner during operation and to assess the feasibility of developing a noise control device that could be deployed in situ. Three typical scanning sequences, namely the echo planer imaging, the gradient echo multi-slice and the MDEFT (Modified Driven Equilibrium Fourier Transform), were used for evaluating the performance of the active control system. The result from the in situ test is highly encouraging because it shows great potential for treating MRI noise with an active noise control application during real time scanning.
6.2 Introduction

Although MRI has evolved into an important imaging tool in medical diagnosis and research, the noise emitted by the scanner during operation remains a concern because it interferes with communication and could be hazardous to hearing. Safety issues related to the use of MRI, such as the physiological effects of loud noise, radio frequency (RF) and magnetic field, has been documented [58]. Numerous previous studies have shown that the exposure to the high levels of MRI sound could cause psychological problems including temporary hearing threshold shifts, anxiety, stress, annoyance, mental fatigue and fear [59-63]. Hence, it is mandatory for all patients and healthcare workers in or near the scanner chamber to wear ear protection [38, 64-65].

It is well known that the acoustic noise emitted during MRI scanning is a result of the rapidly switching electric currents that drive the pulse gradient magnetic fields. The Lorentz forces caused by the electrical current applied to the gradient coils in the main static field excite the structural components of the MRI scanner. The structural excitation effectively turns the MRI system into a loudspeaker [1-3]. The magnitude of structural vibration that causes the acoustic radiation is dependent on magnetic field strength, gradient strength, scanner structure and geometry, spatial settings, and the frequency and waveform of the switching current [3, 50].

Several remedies have been proposed to treat MRI sound emissions [50]. Designing “silent” MRI pulse sequences is one option to reduce the generated MRI noise [35]. However, this approach may limit the imaging applications for some cases. A traditional way to deal
with the high levels of MRI noise is to use passive control approach. The high level noise is reduced by using absorbing materials or modifying the system dynamics to dissipate or redistribute acoustic emissions. A common mode of implementation is having the patient wear ear plugs and/or a headset. This approach usually works well for high frequency noise stimuli but is less effective at lower frequencies. Alternatively, active noise control (ANC), which introduces an out-of-phase secondary acoustic wave form to cancel the sound source response, is a more effective approach for controlling low frequency noise.

In recent years, using active control for MRI noise reduction has received attention from many researchers [11, 22, 40, 51, 55, 57, 66, 69-71]. To apply the ANC technique to MRI noise suppression, two issues need to be addressed. First, one needs to identify an appropriate actuation approach. Because of the unique magnetic environment, an MRI compatible speaker is required to deliver the control acoustic wave to the desired space. Second, an optimal controller must be developed for effectively reducing the sound pressure level (SPL). The earliest application of ANC on MRI noise was conducted by Goldman et al. [22]. They implemented an active MRI noise control system by injecting a synthesized anti-phase signal, which is generated by inverting the phase of the major frequencies components of a recorded MRI signal. This synthesized signal was synchronized with the scanner sound by utilizing a trigger generated by the scanner computer to align with the pulse sequence. However, this method is lacking flexibility in tracking the changes of the system properties and MRI response signals. Furthermore, in their system, a pneumatic tube is used to deliver the anti-phase sound, which is generated outside the magnetic environment, to the patient. This introduces additional time delay to the system and consequently degrades
In 1995, Pla et al. [67] used a pair of piezoelectric speakers placed close to the subject’s ears and an adaptive controller with a multi-channel filtered-x least mean squares (FXLMS) algorithm as the ANC system. Up to 25 dB of noise reduction at frequencies up to 1.2 kHz was reported for their application. However, it is noted that the MRI noise treated was primarily the first few harmonics.

In 1997, McJury et al. [40] tested a headset system equipped with a feedforward controller adapted by the filtered-U LMS (FULMS) algorithm in the laboratory using pre-recorded MRI scanner noise presented through a loudspeaker. In that study, a 10-15 dB reduction in the frequencies below 350Hz was obtained. However, this was not an *in situ* study.

In 1999, Chen et al. [51] proposed a feedback controller system with cascading neural-network architecture to achieve the reduction of MRI noise. This system was also tested in the laboratory (i.e. a non *in situ* study) while playing back pre-recorded scanner noise using a loudspeaker. Using this approach, they achieved an average sound power reduction of approximately 19 dB.

Mechefske and Geris [67] investigated two feedforward ANC systems; a headset-based system and a pneumatic tube-based system. The former was equipped with non-magnetic components (speakers, microphones, and preamps) inside an ear defender, while the latter used a tube to transmit noise cancellation signals to the ear defender. Their results revealed that the headset-based system provided satisfactory overall noise attenuation when using EPI sequences, however, it proved to be less effective when tested inside the MRI scanner as
opposed to the laboratory set-up. In addition, the tube-based system was less effective than the headset-based system due to the time delay created by the length of tube used.

More recently, Kahana et al. [68] implemented a feedforward ANC system in an MRI communication system by utilizing an optoacoustical (i.e. a piezoelectric speaker driven by optical signals) ear defender. Their results showed 35-50 dB attenuation at the fundamental frequency component. This value was in addition to 15 dB of passive attenuation due to the use the slim ear defender headset. Furthermore, Chambers and his colleagues implemented an ANC system on an electrostatic headphone for MRI noise and evaluated the system performance using acoustic, psychophysical and neuroimaging measurements [69-71]. In the most intense component of the scanner noises, an objective reduction of 30-40 dB was obtained for the frequency between 0.5 kHz and 3.5 kHz.

Our research team has also conducted pioneering studies on MRI active noise control. In 2008, we used hybrid control which combined a simple feedforward controller and an H-infinity feedback controller to treat the MRI acoustic response, generated by an Echo Planer Imaging (EPI) sequence, which is dominated by a principal harmonic and its sideband [11]. Later, a feedforward system with multiple reference signals was proposed for various typical sequences including gradient echo multi-slice (GEMS) [55] and EPI [57]. Significant reduction was obtained for various scanning sequences during both physical simulation and in-situ tests [55, 57]. However, the reduction observed during in-situ tests has been limited to frequencies up to 2 kHz which is the effective bandwidth of the speaker system. The physical simulation tests have performed well to slightly higher frequencies, albeit in a more controlled environment.
In this study, the ANC system was enhanced with the goal of achieving improved performance at higher frequencies. It was tested during live imaging scan on a 4T scanner at the Center for Imaging Research at the University of Cincinnati. The results show that the effective frequency range of the ANC system has been increased to 5 kHz which covers most dominant frequency components of typical MRI scanner noise.
6.3 Materials and Methods

6.3.1 Experimental Setup

A 4-Tesla Varian Unity INOVA whole-body MRI scanner (Palo Alto, CA) located in the Center for Imaging Research at the University of Cincinnati was used to obtain all test results presented here. The acoustic noise measurement targets the vicinity of the scanner bore isocenter where the patient’s head is normally located during the scanning. A humanoid dummy (Model TP-1500; Dummies Unlimited, Pomona, CA) was used in all measurements to simulate typical imaging conditions.

The secondary acoustic waveform generator is a magnetic compatible headset with PZT (piezoelectric transducer) speakers (Resonance Technology, CA). In addition, four magnetic compatible microphones are also installed. Two are located inside the headset near the wearer’s ear and measures residual acoustic response, which serves as the error signal for the adaptive controller. Another two are located outside the headset and used as a reference sensor for the control system.

All other electronic equipment is located in the adjacent control room to avoid interference with the magnetic field. The ANC system developed specifically for MRI use is comprised of a dSPACE system (dSPACE GmbH, Technologiepark 25, 33100 Paderborn, Germany) running a feedforward control system developed in Matlab Simulink (The MathWorks, Inc., Natick, MA, USA). The Simulink control system is compiled on a laptop and loaded on the dSPACE processor. This processor has a high speed connection to the dSPACE input/output component, which also converts the analog signals to digital and vice
versa. This hardware configuration allows for fast calculation and communication of the control signal to the headset. Since the ANC system was implemented with a digital sampling rate of 10 kHz to allow for adequate processing time, a high-speed digital recorder (Teac LX-10) was used to simultaneously record the data at 48 kHz in order to obtain the entire audible frequency range of the acoustic signals for subsequent processing. The digitized signals were then stored on the PC computer (Intel Pentium M, 1.86 GHz, 1 GB RAM) that controlled the data acquisition.

In the live scan study, three typical scanning sequences were considered for testing the performance of the developed ANC system. These three MRI scanning sequences are echo planer imaging (EPI), gradient echo multi-slice (GEMS) and Modified Driven Equilibrium Fourier Transform (MDEFT).

6.3.2 Controller Design

The controller plays an important role in the ANC system. The implemented controller is a multiple reference feedforward filtered-X least mean square (FXLMS) control system. In order to preserve stability of the feedforward system, the effective frequency range is limited by the spectral characteristics of the reference signals. In our previous study, multiple reference signals were also utilized; however, the effective frequency range of the control system was limited to 2 kHz. In this experiment, a refinement is made to the previous version of the control system to extend the overall effectiveness to higher frequencies. Multiple copies of the algorithm are implemented in parallel and each one is tuned to maximize reduction of a different portion of the frequency spectrum by adding high pass frequency
filters to the reference signals. Figure 6.1 shows the diagram of the ANC system used in this study.

![Diagram of the ANC system](image)

**Figure 6.1: Refined multiple reference FXLMS algorithm used in the control system**

It utilizes four different reference signals; the Z gradient, the unfiltered microphone, and two different high pass filtered microphone signals. The two high pass filtered microphone reference signals are used to enhance the reduction for specific frequency ranges. The resulting individual algorithm control signals are combined to create the total control signal fed to the headset speaker which produces the cancellation sound. The error signal, measured by the microphone located inside the earpiece near the “patient’s” ear, is the net response of the original MRI noise and the canceling sound signal. This error signal is to be minimized by the control algorithm.

As mentioned previously, the FXLMS algorithm was used in the control system. The diagram of the typical ANC system using a standard FXLMS algorithm [53] is shown in Figure 6.2.
Figure 6.2: Diagram of active control system with standard FXLMS algorithm

The algorithm can be expressed by the following equations (6.1-6.3):

\[
W(n+1) = W(n) - \mu X'(n)e(n)
\]  \hspace{1cm} (6.1)

\[
e(n) = d(n) + y'(n)
\]  \hspace{1cm} (6.2)

\[
x'(n) = \hat{s}(n) * x(n) = \hat{s}^T(n) \cdot X(n) = \sum_{i=0}^{L-1} \hat{s}_i(n) x(n-i)
\]  \hspace{1cm} (6.3)

where \( d(n) \) represents the original MRI response, \( y'(n) \) is the canceling sound from the control speaker, \( e(n) \) is the error signal measured by the inside microphone, and \( x(n) \) is the reference signal (which varies for each of the algorithms in the multi-reference parallel control system). The weighted values of the adaptive filter \( W(z) \) can be expressed by vector \( W(n) = [w_0(n) \ w_1(n) \ \cdots \ w_{L-1}(n)]^T \). Vector \( X(n) \) is formed by the current reference signal data and its previous data as \( X(n) = [x(n) \ x(n-1) \ \cdots \ x(n-L+1)]^T \). \( \hat{s}(n) \) is the estimated impulse response of the secondary path \( S(z) \) which is from the input signal for the speaker to the output signal of the error sensor. The vector \( \hat{s}(n) \) is formed by the
sampled data of the impulse response of the secondary path 
\( \hat{s}(n) = [\hat{s}_0(n) \, \hat{s}_1(n) \, \cdots \, \hat{s}_{L-1}(n)]^T \). The parameter \( \mu \) is the step size which affects the convergence speed and stability of the algorithm.

### 6.3.3 In-Situ Testing

Three typical MRI scanning sequences were used for testing. The baseline response (i.e. when ANC is off) for each of the three scanning sequences is recorded first. Then, the control system was tuned for the different scanning sequences by adjusting the step size of the FXLMS algorithm. Once the ANC system converged, the residual noise stayed at that level. Consequently, no further reduction was observed.

### 6.3.4 Data Processing

The recorded data was post processed and analyzed using an advanced sound data analysis software package (B&K Sound Quality) and Matlab. Using these signal analysis software packages, the measured data was analyzed in the time and frequency domains using a variety of computational tools including SPL calculation in linear-weighted decibel (dB) and A-weighted decibel (dBA), Fourier Transforms and other time-frequency processors.
6.4 Results

6.4.1 Time-Frequency Analysis of MRI Noise

In this study, three different scanning sequences were used, namely, GEMS, 3D MDEFT and multi-slice EPI. These three scanning sequences are commonly used in imaging and have very different signatures in time and frequency domains. The time histories and spectra of the noise for the three imaging protocols are shown in Figure 6.3 (a) and (b), respectively.
Figure 6.3: Time histories (a) and spectra of baseline responses (b) of the MRI noise for three tested scanning sequences (GEMS, MDEFT, and EPI, from top to bottom)

The upper plots in Figure 6.3 (a) and (b) are the recorded noises for the GEMS scanning sequence. It is clearly seen that the MRI noise of the GEMS sequence is dominated by a cluster of harmonics in the frequency domain. In the time domain the generated noise is continuous and steady state. The middle plots are for the noise generated by the MDEFT scanning sequence. Similar to the GEMS frequency domain result, the spectrum of the MDEFT sequence is also dominated by many harmonics. Obviously, the energy distribution
in the frequency domain for the MDEFT scan is different from that for the GEMS. In general, the response of higher frequency components is less than lower frequency components. Furthermore, in the time domain, the MDEFT response shows variation. One can see that there is a silent period between two segments of sound. The bottom plots are for the generated MRI noise of the EPI sequence. As seen from the results, EPI noise has not only harmonic components but also a strong broadband component. In the time domain, similar to the MDEFT result, the EPI noise exhibits transient properties. A short period of silence occurs between two sound segments.

Measuring the performance of the developed ANC system on these three examples is a good evaluation of overall system performance since these three examples are commonly used for MRI scanning and cover varying acoustic signatures such as tonal and broadband noise, as well as steady state and transient sound.

6.4.2 Control Result of GEMS Sequence

The first test case is to control the MRI noise generated by the GEMS scanning sequence. The results are shown in Figure 6.4. The dotted line is the baseline response of the sound measured by the microphone inside the earpiece of headset. It is clearly seen that the spectrum of MRI noise during the GEMS scan is dominated by clusters of harmonics. The solid line in Figure 6.4 is the resulting noise measurement inside the earpiece with the ANC system turned on.
Figure 6.4: Control results of GEMS scanning sequence in frequency domain
(·············: baseline response; ————: controlled response)

Compared to our previous study, the refined system extended the effective frequency range of control system to 5 kHz, a few thousand Hertz higher than that of the previous system. The reduced harmonic with the highest frequency is above 4.5 kHz. The maximum reduction, about 55 dB, occurs at a frequency of 1.3 kHz.

Furthermore, there are two frequency ranges where the reduction is not significant in comparison to other harmonics. The maximum reduction for these harmonics is less than 20 dB. The first frequency range is roughly from 2.3 kHz to 2.6 kHz. The other frequency range is higher than 3.9 kHz. For the harmonics in other frequency ranges, 30 dB reductions and above can be achieved. A possible reason for less reduction occurring in these two frequency
ranges is that the original response level of the harmonics within these two frequency ranges is lower than other harmonics (see more below). A second possibility may be related to the dynamics of the speaker-microphone pair, which is generally referred to as the secondary path. In fact, as shown in Figure 6.5, these two frequency ranges do correspond to the weak magnitude response measured for the speaker-microphone pair.

![Figure 6.5: Measured dynamic response of speaker-microphone pair](image)

**6.4.3 Control result of MDEFT sequence**

Unlike the GEMS sequence which generated noise continuously throughout the measurement period, the MDEFT sequence exhibits transient characteristics consisting of a brief period of sound followed by an interlude without scanning noise. Since the length of the
quiet portion exceeds the loud one, the calculated results presented are taken from a time
block of data that consists of only the loud scanning period without the inclusion of the
quieter interlude. The frequency domain results for this time block for the MDEFT sequence
are shown in Figure 6.6.

![Frequency domain results](image)

**Figure 6.6: Control results of MDEFT scanning sequence in frequency domain**

(----- : baseline response; --- : controlled response)

The dotted line is the baseline response of MRI noise and the solid line is the controlled
response. Similar to the GEMS results, more reduction is obtained for the harmonics below 2
kHz. For the two frequency ranges in the GEMS test that showed less reduction with control,
no substantial reduction is achieved for the MDEFT test either. Aside from these two
frequency ranges, less than 10 dB reduction is obtained for the harmonics between 3.2 kHz
and 3.8 kHz. This might be due to the fact that these harmonics are much weaker in
comparison to the GEMS case. The most reduction (approximately 45 dB) is observed to be at 2.7 kHz for the MDEFT test as opposed to 1.3 kHz for the GEMS test. This is because the harmonic at 1.3 kHz is not the dominant component for MDEFT. In general, less reduction is obtained for most harmonics in the MDEFT sequence when compared with the GEMS sequence. The reason may be that the MDEFT response level is lower than GEMS case.

Since the sound generated during the MDEFT scan is not continuous, it is also of interest to observe the transient results, shown in Figure 6.7. The upper plot, Figure 6.7(a), is the baseline response. The bottom plot, Figure 6.7(b), is the controlled response.

Figure 6.7: Control results of MDEFT scanning sequence in time domain
(a: baseline response; b: controlled response)
Obviously the response level of the controlled measurement is much less than baseline measurement. Within the silent period, the controlled response is of the same level as the baseline response, which is background noise. At the beginning and end of the sound period, a burst can be identified in the controlled response, though it is still much lower than the baseline response. This burst may be related to the controller’s characteristics changing during the relatively long silent period. Once the loud scanning noise commences, it takes a short duration for the controller to converge to the optimal solution again. In the light of the fact that the MRI noise is repetitive, the optimal solution of the controller is assumed to be similar for different loud sound periods. It may be possible to set a smaller step size during the silent period to prevent the controller from shifting away from the optimal solution. This could reduce the time to re-converge to the solution when the subsequent loud period starts, improving overall performance. Further study is under way to refine the system performance with this type of transient signal.

6.4.4 Control results of EPI sequence

The EPI sequence may be the most difficult case considered since it contains a strong broadband component in addition to strong harmonics. The existence of phase delay in the secondary path (control speaker to inside microphone error sensor), renders the broadband component more difficult to address. Since the noise generated by the EPI scan is not continuous, a loud time block is used for the frequency domain calculations, the same as for the MDEFT calculations. Similar to the previous plots, as shown in Figure 8, the dotted line represents the baseline response of EPI noise and the solid line represents the controlled
response. The dominant harmonics, at about 900 Hz and 2.8 kHz, are reduced substantially. Approximately 30 dB reduction is obtained at the harmonic near 900 Hz, while more than 40 dB reduction can be seen at the harmonic around 2.8 kHz. In addition, more than 10 dB of reduction is achieved for frequencies adjacent to the 900 Hz harmonic. In our previous study [10], these frequency components were found to arise from the broadband response of EPI noise and generated by the Z-gradient excitation (slice selection gradient).

\[\text{Figure 6.8: In situ Control results of EPI scanning sequence in frequency domain} \]
\[\text{(...--... : baseline response; - - - - - : controlled response)} \]

As discussed previously, the EPI sequence exhibits a transient characteristic to its signature. The time history of the response before and after control for the EPI scan is shown in Figure 6.9. Similar to the MDEFT results, Figure 6.9 (a) and (b) are the baseline response and the controlled response, respectively. Unlike the MDEFT case, the EPI has a much shorter silent period. In addition, there is no burst at the beginning and end of the sound
period. This may be because the silent period is too short to change the characteristics of the controller dramatically. The controller is still nearly optimal when the next loud period begins.

Figure 6.9: In situ control results of EPI scanning sequence in time domain
(a: baseline response; b: controlled response)
6.5 Discussion

It is well known that the MRI acoustic noise generated during imaging is very loud. In 1989, Hurwitz et. al. [12] measured the sound pressures at the magnet isocenter and reported levels between 82 and 93 dBA for static magnetic field strengths of 0.35 to 1.5 Tesla. In a study by Ravicz et. al. [4], the measured SPLs of two scanners (1.5 and 3-Tesla) operated with EPI sequences were found to vary from 123 to 138 dB. In our study [10], SPLs as high as 130 dB were recorded from a 4-Tesla MRI machine running an EPI sequence. It is generally agreed that SPLs are greater at higher magnetic fields. The high level of acoustic noise is not only annoyance, but could also result in hearing loss in humans. Hence, it is desirable to reduce the noise exposure experienced by the patient and anyone in the immediate vicinity of the scanner.

In this study, we developed an active headset system and tested the performance in situ. Experimental work was conducted on three common MRI scanning sequences, namely GEMS, MDEFT and EPI. Substantial reductions in noise levels at the patient’s ear were obtained in all tests. The greatest reduction, about 55 dB, was found at the harmonic at a frequency of 1.3 kHz in the GEMS case. Approximately 21 dB and 30 dBA overall reduction was achieved for GEMS noise across the entire audible frequency range. For the MDEFT sequence, the control system achieved 14 dB and 14 dBA overall reduction over the audible frequency range, while 13 dB and 14 dBA reduction was obtained for the EPI case. These linear-weighted and A-weighted sound pressure levels are calculated using the B&K Sound Quality software package. Only the time blocks representing the loud sound periods for the
MDEFT and EPI cases were used in the calculation of frequency domain response and SPLs. Also, note that these results represent the reduction due to the active control only. Since the measurements were obtained inside the earpiece, the contribution of passive reduction of the headset has already taken place for all tests, with and without the controller active. The total noise reduction a patient would observe compared to the ambient level during scanning would be a combination of the headset passive reduction and the active control reduction. By comparing the outside and inside microphone responses, the passive reduction attributed to the headset can be evaluated. The passive device provided addition reductions to ANC system with a similar overall reduction for the three different scanning sequences. For GEMS noise, about 16 dB and 20 dBA reductions can be obtained by passive means. 15 dB and 19 dBA reductions were achieved for MDEFT case. And, 18 dB and 18 dBA reductions were measured for EPI sequence. The detailed measured attenuations for these three scanning sequences by active and passive means are summarized in the Table 6.1.

<table>
<thead>
<tr>
<th>Sequence</th>
<th>Active Means</th>
<th>Passive Means</th>
<th>Total</th>
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<tr>
<td></td>
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<td>dBA</td>
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</tr>
<tr>
<td>MDEFT</td>
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<td>15</td>
</tr>
<tr>
<td>EPI</td>
<td>13</td>
<td>14</td>
<td>18</td>
</tr>
</tbody>
</table>

**Table 6.1: Measured attenuations by active and passive means for three different sequences**
In conclusion, the proposed ANC system works well on noise generated during a variety of common MRI scanning sequences. In-situ testing was conducted during live scans on a 4-T MRI for three scanning sequences, namely GEMS, MDEFT and EPI. Very promising results were obtained. These encouraging results indicate potential for the development of a commercial product that could provide a substantial level of noise reduction for the patient, independent of the type of scanner in use. While integrated with the imaging computer to monitor the scanning sequence gradients, it does not require any retrofitting of existing scanners.
7. CONCLUSIONS AND FUTURE WORK

The advent of clinical MRI in the 1980s provided a significant advancement in medical diagnostic imaging. The loud noise produced during scanning, which is a direct result of the fundamental MRI operation, is not only uncomfortable and potentially harmful to patients, but it is also an impediment to realizing further benefits from MR scanning. Although the noise has been recognized as a factor limiting the evolution of MRI, as demonstrated by the number of patents reviewed in chapter 2, a practical and effective solution remains elusive. This dissertation has focused on the research and development of an ANC system as a potential solution to the problem. The approach proposed considered prior investigations into the relationship between the MRI gradient signals and noise produced. When considered along with published findings by recognized ANC experts, the feedforward algorithm was chosen as the basic approach to be implemented in the MRI specific ANC system.

In chapter 3, MRI compatible headphones containing piezoceramic speakers were evaluated and demonstrated to perform at a level that would allow them to be effective for this application when implemented appropriately with the ANC system. This was accomplished with the feedforward control system using the MRI gradient signal as a reference.

After considering additional scan types, it was determined that the gradient signals alone would not always provide the level of noise reduction desired. In chapter 4, a set of headphones with an additional microphone outside the earpiece that could be used as a reference was evaluated. It was demonstrated that this approach had merit in addition to
utilizing the gradient signals for reference.

Due to the cost of operating the MRI scanner, it was not feasible to perform live testing on a regular basis. A physical simulation environment was needed for routine use. While this simulation was accurate enough for development purposes, periodically the system was evaluated during in-situ during live MRI scanning to assess its performance. Chapter 5 documents the results of an interim control system capable of achieving significant noise reduction up to 2 kHz.

In spite of the successful in-situ scan demonstrated in chapter 5, a substantial amount of noise still existed with that version of the control system design operating to its full potential. The control system was enhanced by introducing multiple parallel algorithms that utilized frequency filtered reference signals which could be individually optimized to achieve additional noise reduction to higher frequencies. The results obtained during in-situ scanning for this control system were documented in chapter 6. This version of the control system was demonstrated to have substantial noise reduction up to 5 kHz for multiple scan types. In the best case, a GEMS scan, the control system alone provided overall reduction of 30 dBA, including 55 dB reduction at the dominant harmonic frequency. The headphones include insulation designed to provide passive reduction, which amounted to 20 dBA for the GEMS scan without the control system in use. The combination of the passive and active means of noise reduction, which compares the ambient levels adjacent to the patient with that measured inside the headphones, produced a total reduction in SPL of 50 dBA. The control system design utilizes individually optimized algorithms that produce control signals that are combined for the single overall control signal that is played through the headphones. It is
expected that this modular design approach will allow for it to be adapted to and optimized for additional scan types that produce noise with differing time and frequency spectra. A “tool box” with multiple reference signals and frequency filters could be used in different combinations and tuned to the specifics of a particular scan.

The level of noise reduction achieved has the potential to provide the patient with relief while undergoing scanning. It could also open the door to additional research areas by reducing the role noise currently plays as a barrier. Scanning protocols currently conceived but not implemented due to the level of noise production might become available for use. Research into brain function that is affected by noise could also be conducted. For example, conducting a scan with/without the control system active would allow for functional brain imaging with different levels of noise exposure not currently possible.

Despite the progress that has been made during the course of this research, there is additional work to be completed. The ANC system as currently implemented is designed to reduce the noise at the reference microphone located inside the headset. Unfortunately, this location is not coincident with the patient’s tympanic membrane where sound is sensed by the patient. At lower frequencies where the acoustic wavelength is large relative to this offset, this distance is not expected to influence the results. As the controlled frequency increases and the acoustic wavelength decreases, the effective control volume decreases. Further testing should be conducted to determine whether the 5 kHz control frequency is effective with the current error sensor location from the patient’s perspective.

In addition, the current ANC system has only been tested during in-situ scanning with a single MRI scanner. It should be tested at an alternate site with a different scanner to validate
whether the control system is capable of performing at a high level independent of the specific scanner and its installation environment.

The ultimate goal of the MRI ANC system is to allow for use by people. There is potential for other noise paths such as bone conduction that could prevent the ANC system from performing as well from a patient’s subjective perspective as the objective measurements indicate. While it may be possible to create a physical test environment that would allow some level of investigation, the most reliable way to determine whether this is an issue is to perform tests on human subjects. In order to do this, it will be necessary to implement safety features that protect the patient from being subjected to undesirable increases in SPL which have the potential of damaging hearing. Once such a system is available, it will be possible to conduct patient testing which will include subjective evaluation of the ANC system’s performance.
PUBLICATIONS

The research activity of the active noise control system for MRI covered in this dissertation has resulted in the following technical publications [50, 54-57, 72-78]:


BIBLIOGRAPHY


[35] F. Hennel, F. Girard and T. Loenneker, “‘Silent’ MRI with soft gradient pulses”,

[36] D.G. Tomasi and T. Ernst, “Echo-planer imaging at 4 Tesla with minimum acoustic


[38] C.K. Mechefske, Y. Wu and B.K. Turr, “Acoustic noise reduction in a 4T whole


[41] B.H. Kevles, “Naked to the bone: medical imaging in the twentieth century”, New


APPENDIX A – SUPPLEMENTAL FIGURES

Figure A.1 contains an example of a Matlab Simulink Control System. Use of this tool for developing the control systems is referenced throughout the dissertation.

Figure A.1: Matlab Simulink Control System Example
Chapter 3 covers the evaluation of the MRI compatible headphones. The ANC system performs best with linear response. In Figure A.2 the performance of the MRI compatible headphones is compared with the high fidelity set to demonstrate the degree to which each varies from linear response.

![Headphone SPL Linear Performance](image1)

![Headphone Phase Performance](image2)

**Figure A.2: Headphone performance comparison**
In chapter 6, the ANC system is enhanced to work at higher frequencies by introducing high pass filtered reference signals that allow for optimal tuning of each algorithm. The step size of the algorithm using the filtered signal can be increased for improved convergence without causing instability. A plot of the power of the filtered and unfiltered microphone reference signal is shown in Figure A.3.

![Plot showing power of microphone reference signal original and high pass filtered](image)

**Figure A.3: Power of microphone reference signal original and high pass filtered**
To validate the proposal of using frequency filtered reference signals, a version of the ANC system using a single algorithm with a high pass filtered reference signal was tested. The result is shown in Figure A.4.

![GEMS Physical Simulation](image)

**Figure A.4:** ANC system response with single high pass filtered reference signal
APPENDIX B – SOUND SAMPLES

The following sound samples were acquired during in-situ testing of the version of the active noise control system covered in chapter 6. Results for three different scanning sequences are plotted in Figure B.1 through Figure B.3. First, a transient plot comparing the normalized response of the error microphone with and without the controller active is presented for the scan. Below the plot are two text boxes labeled “Control Off” and “Control On”. When viewing as a PDF in “hand mode”, clicking on the box will active the appropriate sound file.

Figure B.1: Transient response of controlled/uncontrolled EPI noise
Figure B.2: Transient response of controlled/uncontrolled MDEFT noise
Figure B.3: Transient response of controlled/uncontrolled GEMS noise