Chapter 5

Towards a more silent scanner; an analysis of sound reducing methods

5.1 Introduction

In functional Magnetic Resonance Imaging (fMRI), the demand for high spatial and temporal resolution, together with high signal-to-noise-ratio (SNR), requires Magnetic Resonance (MR) scanners with high magnetic field strengths and fast switching of strong gradients.

The effects of high static magnetic fields on humans have been investigated since the 1960s (e.g., Barnothy [1964]). There is no indication that the static field of an MR scanner is of any harm with the current field strengths (< 4 T) (Schenck [2000]), except for accidents with ferromagnetic objects and pacemakers. Time-varying magnetic fields, such as switching gradient magnetic fields, may cause peripheral nerve stimulation or in extreme cases even cardiac stimulation (Schaefer et al. [2000]). Radio frequency (RF) pulses for selection purposes can interact with implants (Shellock et al. [2004]), and other foreign bodies like tattoos (Tope and Shellock [2002]). Tissue heating by RF pulses is also possible (Shellock [2000]).

The interaction between the strong electric currents carried through the gradient coils and the high magnetic field generate vibrations due to the evoked Lorentz forces acting on the gradient coils (e.g., Mansfield et al. [1995]). Acoustic radiation originating from these induced vibrations easily reaches sound pressure levels (SPL) of 110 dB and higher (Counter et al. [2000]; Hedeen and Edelstein [1997]; McJury et al. [1994]; Moelker et al. [2003b]; Price et al. [2001]; Ravicz et al. [2000]). The auditory system may be damaged when exposed to these high levels (e.g., 29 CFR 1910.95, Occupational Noise Exposure). Moreover, the acoustic noise is a cause
for anxiety (Quirk [1989]), distraction (Barrett [2002]; Elliott et al. [1999]; Mazard et al. [2002]), and spurious activation during functional brain research (Belin et al. [1999]; Hall et al. [2000]; Mathiak et al. [2002]; Talavage et al. [1999]).

Various methods are employed to reduce the sound exposure for subjects, ranging from inserting ear plugs and putting on ear muffs to redesigning the scanner. This paper addresses the technical and practical considerations in MR scanner design regarding sound generation, and discusses the various sound reducing methods, and their effectiveness.

5.2 Sound sources in the scanner room

The walls of the scanner room not only play a role in shielding electric and magnetic fields, they also fulfill a task in sound insulation. In particular, they shield the environment from the scanner noise.

Inside the scanner room, various sound sources can be distinguished (cf. Ravicz et al. [2000]). They are described in the following subsections.

5.2.1 Air-handling system

A modern air-handling system can be relatively silent. The blower inside the scanner bore gives a constant fresh air flow at the head position, but the associated sound level remains low in comparison to other sound sources.

Air-handling of the scanner room also generates a rather constant background noise. Technically it is possible to keep this noise level below 20 dB(A) (e.g., Ramamoorthy et al. [2003]). Even at 30 dB(A), this noise is not a serious problem. Moreover, it always remains possible to switch off the system (temporarily).

5.2.2 Liquid helium pump

A pump is used to keep the helium liquid; the boiled-off helium is recycled by compression and subsequent expansion to liquify the helium again. The pump makes a pounding sound, mainly with low frequencies. SPLs reach up to 80 dB (Ravicz et al. [2000]). Turning the pump off results in helium boil-off; turning the pump off for a prolonged period may result in a quench: the magnetic coils loose their superconductivity due to the temperature increase; the strong currents produce heat in the resistive coils which in turn increases the boil-off rate. Turning the pump off temporarily also results in helium boil-off, this period should not be too long. Relocating the pump, or insulating it, reduces perceived SPLs.
5.2.3 Electric current leads

From our own measurements (unpublished results) we learned that loosely fastened current leads to the gradient coils may produce loud noise at the resonance frequencies of the length of wire. The alternating current through the wires interact with the static magnetic field; the evoked Lorentz forces bang the wires to the cryostat or the shields. These acoustic noise sources can be prevented by tightening the wire connections or by placing stiff current leads some distance from other objects.

5.2.4 Eddy currents

Eddy currents can be induced in the RF coils, the RF shield, and the static field magnet cryostat (Katsunuma et al. [2002]). Changing magnetic fields as the gradient fields in MRI, produce a changing electric field in accordance with Maxwell’s laws. Lenz’s law states that a changing magnetic field induces an opposing magnetic field in a conductor. For this reason, the electric gradient currents have an overshoot in the beginning of a pulse to compensate for this effect. The induced eddy currents are time-variant and the resultant Lorentz forces make the conductor vibrate. The RF coils are located near to a person’s ear, so the vibrations there may be a major sound source. Likewise, eddy currents in the RF shield and the magnet cryostat cause these structures to vibrate with resultant acoustic noise.

5.2.5 Gradient coil noise

The strong electric currents flowing through the gradient coils interact with the static magnetic field. The resultant Lorentz forces are distributed over the gradient coils and try to deflect the gradient coil structure. Actual vibration patterns depend on the force distribution, material properties, frequencies of excitation, and gradient coil dimensions. These vibrations may match eigenfrequencies of the gradient coil assembly resulting in strong vibrations and high sound pressure levels (Tomasi and Ernst [2003]).

5.3 Effects of acoustic noise

In general, continuous exposure to sound with high sound pressure levels has physiologic and psychologic effects. This is also true during MR imaging.

5.3.1 Hearing threshold shift

Exposure to sound pressure levels above 80 dB(A) are considered dangerous. Hearing damage due to acoustic noise cannot be cured, and even a onetime exposure...
to high sound pressure levels (disco, drill, or other noisy environments) can permanently damage the auditory system. The seriousness of damage due to acoustic noise depends on levels and exposure time. Standards, like the Occupational Noise Exposure (29 CFR 1910.95 (R1999)), prescribe a hearing conservation program when workers are subjected to a time-weighted average of 85 dB (measured on the A-scale, slow response) over an 8-hour working day. Permissible noise exposures start from 8 hours at 90 dB(A) (slow response), and are reduced by a factor of 2 for every 5 dB increment of the time-weighted averaged noise, or their combined effect for different levels. All continuous, intermittent, and impulsive sounds with levels from 80 to 130 dB are taken into account for these measurements. Exposure to impulsive or impact noise should not exceed 140 dB peak sound pressure level. Hearing protectors, as mufflers and earplugs, should attenuate the noise to the permissible noise exposure.

When the ear is exposed to high sound pressure levels, muscles in the middle ear can tighten. Through this so-called Stapedius reflex, the transfer of the middle ear can be reduced by about a factor of 6 in amplitude, giving some protection to overload. As the muscles do not relax immediately, a temporary threshold shift occurs (Brummett et al. [1988]) which can last from several minutes to hours, depending on exposure time and level (Elliott and Fraser [1970]).

Still, if noise reduction measures are not sufficient to reduce the exposure to noise below safety levels, permanent threshold shifts may occur.

5.3.2 Discomfort and anxiety

The loud acoustic noise can also cause discomfort or even anxiety. Being inside a relatively narrow bore can cause claustrophobia or panic attacks. Discomfort can come from the examination duration or the temperature inside the scanner bore (Brennan et al. [1988]; Quirk et al. [1989a]). People undergoing MR examination should be provided information on the MR scanner’s (noninvasive) imaging techniques, the small bore that can evoke feelings of confinement, the possible rise in temperature, the acoustic noise, the duration of the examination, and the possible outcomes of the examination (Quirk et al. [1989b]; Harris et al. [2004]). It is good to note that people with a more anxious nature, are more likely to experience anxiety during MR scanning.

While not all feelings of discomfort and anxiety can be taken away, efforts can be made to prevent these feelings. The used techniques can be explained properly, perhaps some time prior to examination. The outcomes of the examination cannot be changed, and the duration of the scanning has a lower limit. A constant stream of fresh air can provide a pleasant ambient temperature. The feelings of confinement can partly be taken away in open scanner systems, or by (optically) shortening the scanner bore (DeMeester et al. [2002]). Finally, efforts can be taken to reduce the acoustic noise accompanying MR imaging.
5.3.3 Confounding stimulus

The acoustic noise does not only have aforementioned effects. During functional brain research, the noise itself is a stimulus with an adverse effect on the outcome of the research.

Interference between task and noise

Brain activation in the auditory cortices as a result of the scanner noise is straightforward (Hall et al. [2000]; Mathiak et al. [2002]; Schmithorst and Holland [2004]; Shah et al. [2000]; Talavage et al. [1999]; Tanaka et al. [2000]). The loud noise immediately hampers functional brain research of the auditory system (Ulmer et al. [1998]). No discrimination can be made between stimulus and noise, unless the noise is not present during the stimulus (Belin et al. [1999]; Di Salle et al. [2001]; Hall et al. [1999]; Schmithorst and Holland [2004]), or the stimulus is the scanner sound itself (Bandettini et al. [1998]; Bilecen et al. [1998]).

Distraction from task

“It’s so noisy, you can’t think straight” (Barrett [2002]). For simple tasks, the visual and motor tasks were not hampered significantly due to the scanner noise, where passive listening resulted in less auditory cortex activation (Elliott et al. [1999]). In a study that required more attention, the motor areas showed an increase in activation under a motor task, and the visual areas showed a decrease in activation under a visual task (Cho et al. [1998a]). A Positron Emission Tomography study by Mazard et al. [2002] showed that tasks became harder when recorded scanner sound was presented. All this does not necessarily imply that fMRI is not suitable for functional brain research, but these effects must be considered when interpreting data.

Background activation increase

The scanner noise gives a continuous activation of the auditory cortices. Effectively, this elevates the baseline of activation in auditory experiments. Differences between activation due to the target stimulus and the baseline decrease, which leads to a lower activation estimate (Bandettini et al. [1998]; Di Salle et al. [2003]; Elliott et al. [1999]; Hall et al. [1999]).

5.4 Sound reducing methods

In this section, we give a thematic overview on noise reducing methods in fMRI. Some published papers treat several of these subjects in one study and connect the different topics. A complete separation is therefore not always possible. Whenever necessary, we comment on the methods and the obtained results.
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5.4.1 Passive methods

Under passive sound reducing methods, we understand every physical measure to attenuate the acoustic noise that does not use electronic components or transducers.

Mufflers and earplugs

When vibrations become airborne, the acoustic noise can only be attenuated at the level of the ear. In general, if permissible exposure levels are exceeded, then hearing protection devices (HPD) are mandatory. The HPD should bring the perceived noise levels under the level that is permitted (29 CFR 1910.95 (R1999)).

Conventional HPDs are low-pass filters, attenuation up to 2 kHz can be as much as 30 dB for earplugs. Above 2 kHz, attenuation can increase to 40 dB. Factors of importance are used materials and how well the ear canal is occluded by the inserts (Casali and Berger [1996]). Circumaural HPDs have about the same frequency characteristics, with a somewhat lower attenuation (Berger et al. [1998]). If earplugs are not properly inserted, or the pinnae are not properly covered, then attenuation is not maximal. The combination of earplugs and mufflers give a combined effect, insofar the threshold of bone conduction is not surpassed (Berger and Kerivan [1983]). Custom made HPDs perform better than standard devices, but feasibility of this is low for patients who come for a one-time only examination. However, for health workers whose daily work is in the MR room, personal fit HPDs are recommended.

Further increase of attenuation can be achieved by placing a helmet over a person’s head to prevent acoustic noise from reaching the cochlea via bone conduction (Berger et al. [2003]; Ravicz and Melcher [2001]). Apart from the discomfort caused by the helmet, it takes extra space which in general is not available in a headcoil.

All these measures reduce all incident sound, regardless of being noise or instructions/stimulus. Placing headphones to the pinnae, and have earplugs inserted with a known frequency characteristic and compensate for that, can solve this problem. Another solution is having inserts that block the ear canal from noise, but with a sound guide coming from a loudspeaker. What should be kept in mind is that communication with the person in the scanner should be possible at all times (Moelker et al. [2004]).

What also should be realized is that the given values are maximal values. Non-proper insertion of earplugs leads to a decreased attenuation. The conduction of vibrations via the bed to the skull to the cochlea is not considered. A psychophysical evaluation of this transfer should be conducted, or at least, with an artificial head. The reported sound pressure increase with a person in the bore (Hedeen and Edelstein [1997]; Price et al. [2001]) needs to be investigated further as well.
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Transmission pathway blocking

To prevent the vibrations of internal structures to become airborne, the transmission should be blocked. Acoustic noise reaches the subject in the scanner via different pathways, with all different contributions to the perceived sound pressure levels (Katsunuma et al. [2002]; Moelker et al. [2003a]). The same is true for a health worker near the scanner. To effectively block the acoustic transmission, the different pathways have to be discerned.

The vibrations of the gradient coils are transmitted towards the bore, and the shrouds in the bore transfer the vibrations to the air. In the other direction, the vibrations of the gradient coils are transmitted via the gradient supports to the cryostat, and via the cryostat to the outer shrouds and to the building. Indirect sound coming from reflections and vibrations of the walls contribute little to the sound pressure in the bore, but have a significant contribution to the sound pressure at a health workers location (Moelker et al. [2003a]). Blocking the vibrational pathway from the gradient coils by placing them in a vacuum, reveal other sound sources as eddy current induced vibrations in the cryostat (Edelstein et al. [2002]; Katsunuma et al. [2002]). Placing polyurethane liners in the acoustic pathway is the most straightforward application of passive acoustic screening (Foster et al. [2000]; Mechefske et al. [2002a]; Moelker et al. [2003a]).

In the bore, the highest contribution comes from the direct path of the vibrating gradient coils (Katsunuma et al. [2002]). Most of the work on acoustic liners is done within the bore. Liner materials are known to attenuate more at high frequencies. Foster et al. [2000] give only one number without frequency selectivity, namely 10 dB for foam placed between the gradient coils and the shim coils. Mechefske et al. [2002a] placed a fiberglass cylinder (inner diameter 62 cm, covered on the outside with a noise barrier) within the bore. This cylinder was mechanically decoupled from the MR system. Measured at the iso-center this setup provides a 10 or 17 dB SPL reduction for two different liners at 1000 Hz. Closing an open side of the scanner with an end-cap reduces the SPL with another 9 or 4 dB, respectively. The same values are found for a realistic EPI sequence. Moelker et al. [2003a] covered all surfaces in the scanner room with 7.5 cm thick layers of fiberglass. The results show that covering the outer shrouds and the walls does not significantly reduce the sound pressure levels in the bore for any frequency. Outside the bore, however, covering the combined surfaces lead to a 10–20 dB reduction for frequencies above 500 Hz. The separate coverage of surfaces does not give major results, suggesting that acoustic shielding should be done carefully in order to prevent sound leakage. Attenuation of the sound pressure in the bore can be as high as 29 dB (at 10 kHz), but driving the scanner with real sequences only shows 10–12 dB reduction. Having to trade 15 cm inner bore diameter for this reduction is not an option.

Placing the gradient coils inside a vacuum, and mechanically decouple it from the rest of the scanner, results in a decrease of transfer of up to 24 dB (Edelstein
et al. [2002]; Katsunuma et al. [2002]). This depends on the pressure in the vacuum enclosing. Both articles show a dependence on pressure that is conform the theoretical expectations for 2 sources, where the loudest decreases in contribution and the previously softer source becomes the dominant source. The maximal attenuation depends not only on the pressure, but also on the mechanical coupling between the vibrating gradient coil and for instance the cryostat. Mounting the gradient coils independently on the concrete floor leads to better attenuation than mounting the gradient coils on the magnet bore. Only 6-8 dB attenuation is reported for the latter setup.

Eddy current prevention

Vibrations can be caused by eddy currents interacting with the main magnetic field. These eddy currents come from the gradient fields intersecting a conductor, e.g., the cryostat. The changing magnetic field causes a current that creates an opposing magnetic field, according to Lenz’s law. Without prevention of eddy currents, the opposing field has to be compensated for by stronger gradient fields. Not only regarding vibrations, but for gradient magnetic field linearity, eddy currents should be prevented to avoid image degradation.

After blocking the vibrations from the gradient coils, the eddy current induced vibrations become the main source of vibrations. Edelstein et al. [2002] list the order in which the different conductors contribute to the sound pressure in the scanner bore:

1. Eddy current induced vibrations of the conducting cryostat inner bore.
2. Eddy current induced vibrations of the RF coil (body coil or head coil).
3. Eddy current induced vibrations of the cryostat that are mechanically transmitted to the patient bore.
4. Remaining vibrations of unknown source-pathways that are radiating sound to the patient bore.

The general approach to reduce eddy current-induced vibrations is by shielding. Eddy currents have typical dimensions and orientations. By changing the dimensions and/or orientation of the conductors, the eddy currents die out faster. Consequently, they have less influence. Producing a nonconducting cryostat inner bore takes care of eddy currents in that part of the scanner (Edelstein et al. [2002]; Katsunuma et al. [2002]).

Gradient coil design and modeling

Gradient coils are designed to produce adequate gradient magnetic fields. Secondly, gradient coils are designed to dampen the vibrations of the coil assembly and the
induced acoustic noise. The development of a model regarding the vibrations and acoustics of an MRI scanner starts with the choice of the modeling method (Kuijpers [1999]). By choosing a deterministic approach, the choice between analytic and an element method is the next choice to be made. The analytic approach can be used to look at the acoustics of a vibrating duct (Kuijpers [1999]), or a plate (Mansfield et al. [1998]), for an increase in insight. With this knowledge, finite element methods can be developed to numerically analyze the vibrations and sound of the MR scanner.

With increasing computational power, it is feasible to employ computer modeling of an MR scanner. Dedicated toolboxes or software, make it possible to compute the Lorentz force distribution over the gradient coil, the vibrational modes of a gradient coil, the shape of the (gradient) magnetic fields, and the acoustics concerned with the gradient coil vibrations. Without modeling, expensive (∼ 100,000 US$) gradient coils have to be built to test its properties. Finite element analysis (FEA) has proven to be a powerful tool in different research areas, including construction. The material properties of all elements (density, Young's modulus, damping) can be adjusted, as well as their mutual interaction.

Proper arrangement of conductors produces a linear gradient magnetic field when currents are send through the coils. The exact arrangement is important for the produced magnetic fields as well as for the Lorentz force distribution. In MR sequences, the electric currents and the wire length are well known beforehand. This distribution can be calculated with equation 3.1 (Yao et al. [2004]). Applying the calculated (dynamic) forces to the mesh, enables to model the deflection/vibration pattern of the gradient coil assembly (Edelstein et al. [2002]; Mechefske et al. [2002b]).

In modeling, assumptions must be made. The general assumptions comprise the necessary mesh dimensions, the boundary conditions, and material parameters. The mesh dimensions must be smaller than the produced wavelengths \( \lambda (\lambda = c/f) \), where \( c \) is the velocity of sound in air, and \( f \) is the frequency of excitation). The wavelengths of airborne sound are the lowest, as the velocity of sound in solids is much higher than in air. For accurate simulation result, appropriate mesh sizes are oversampling the acoustic wavelength (Edelstein et al. [2002]; Mechefske et al. [2002c]). In this assumption, there is no room for produced overtones with consequently shorter wavelengths. The boundary conditions are the forces acting on the gradient coil, and the suspension to the rest of the scanner. The vibrations of the rest of the scanner, when this starts to vibrate as well, make it computationally harder to predict the vibrations of the gradient coil. Taking the suspension fixed is besides reality but might approximate it. Experimental evaluation is therefore necessary (Mechefske et al. [2002c]; Yao et al. [2004]).

As a gradient coil consists of several elements: the wiring for the separate gradients, the tubing for the cooling, and the resin that holds everything together, it is not feasible to model every detail. Moreover, these details are much smaller than the mesh size. Modeling several layers (wiring, resin) with appropriate parameters pre-
dicts the vibrations in a real gradient coil well. With this vibration data, the sound field and the sound pressure can be computed (Edelstein et al. [2002]; Yao et al. [2004]).

The prevention of eddy currents is also investigated by numerical methods. This shielding is about the confinement of (gradient) magnetic fields, which is especially challenging in shorter magnets (Brown et al. [2002]; Shvartsman et al. [2001]). Perfect shielding is impossible, the deviation from perfect shielding, the “shielding error”, should be as low as possible. Numerical methods have been developed to assist the development of better shielding when making gradient coils.

Cho et al. [1998b] proposed a rotating DC gradient. The switching of the read-and phase gradients produces the loudest noise during MRI scanning. Omitting these sources leaves the slice selection gradient which produces less acoustic noise. A gradient coil, with a constant gradient field perpendicular to the selected slice, is rotated with suitable angular increment. At each angle a recording is made. Images are obtained by projection-reconstruction, like in Computed Tomography (CT) scanning. A 20.7 dB reduction in acoustic output is reported without mentioning the time needed for image acquisition.

### 5.4.2 Active methods

Active methods to reduce the vibrations of the scanner or the acoustic noise include counterbalancing Lorentz forces, anti-noise, and active structural acoustic control.

**Vibration cancellation**

The surfaces that contribute most to the sound pressure within the scanner bore, are the shrouds of the bore. If these do not vibrate, and the outer housing contribution is negligible, then no acoustic noise is radiated. Piezoelectric actuators or magnetostrictive actuators can replace the shrouds, and they can be electronically controlled to vibrate with the exact amplitude and anti-phase as the shrouds would normally vibrate. In this way, the inner surface of the bore does not vibrate and consequently does not radiate acoustic noise. This has already been investigated in the aviation industry for turbo-prop aircraft cabins (Wu et al. [2003]; Aurilio et al. [2003]). Plans exist to investigate this in MRI (Moelker et al. [2003a]), but to our knowledge, so far, no results have been published. The magnetostrictive method does not work in MRI, as extra magnetic fields are needed for the shrinking or enlargement of the material. These extra fields influence magnetic field homogeneity. If the changing electric fields from the piezoelectric actuators also influence the magnetic field homogeneity, or otherwise distort image acquisition, then this technique will also not be feasible in MRI.
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Active noise control

“Can’t you use anti-sound?” is one of the most frequently asked questions by the audience concerning acoustic noise in MRI. Recording the acoustic noise and playing it back in anti-phase seems an easy and feasible solution. The general approach is to have the subject wear earphones with microphones near the ear, if possible, close to the eardrum. Noise coming from outside the headset is measured by the microphones and sent to a Digital Signal Processing unit. An adaptive filter produces the anti-noise signal that is presented to the ear via the earphones. Through constant adaptation, the noise at the ear drum should be kept as low as possible (Casali and Berger [1996]). The damping effect of this method is limited to about 40 dB. After cancellation of all sound in the ear canal, the sound that reaches the cochlea is dominated by bone conduction. To obtain a 40 dB reduction for pure sinusoids, the amplitudes of the sound and the anti-sound must not differ more than 1 % with perfect phase matching. With perfect amplitude matching, the phases need an accuracy of 0.01 radians, or 0.6 degrees (Chambers et al. [2001], see also appendix B.2). Inside the scanner, sub-millimeter displacements are typical for subjects. Such displacements can destroy the phase matching if it is not compensated for displacements accurately. Health workers near the scanner are moving more, and faster. Depending on the speed of the DSP and the accuracy of the feedback system, the efficacy of anti-noise for health workers is limited to low frequencies. The combination of active noise reduction with passive hearing protection, which is limited to higher frequencies, can give a 20 dB reduction above 125 Hz (Casali and Berger [1996]).

McJury et al. [1997] recorded MR scanner noise and replayed the sound in an audio laboratory via loudspeakers. A reference microphone, connected to a DSP unit, controls a secondary source to emit the anti-noise. An error microphone in the desired zone of quiet is also connected to the DSP to optimize the anti-noise. Results in this test were satisfying below 350 Hz, with an average reduction of 10–15 dB up to 30 dB for single frequencies.

Chambers et al. [2001] performed three anti-sound experiments. The first was, in a standard setup, reducing white noise and pure-tone target detection in a sound-laboratory. The second experiment used recorded scanner sound in a feed forward fashion, and was conducted within a replica scanner. Using an error reference microphone too close to the ear, gives too little time to the DSP to generate the anti-sound. Subjects were asked to report the subjective reduction in SPL. For two different EPI recordings at fundamental frequencies of 600 and 1900 Hz, the subjective attenuations were 12 and 5 dB on average, respectively. Further, filtered noise experiments (to simulate scanner noise) were conducted in the replica scanner to obtain difference spectra at 8 frequencies between 500 and 3000 Hz. At these frequencies, the attenuation was approximately 40 dB. The third experiment was conducted in a real scanner with two EPI sequences (fundamental frequencies 750 and 1500 Hz), to
find 40 dB attenuation at the fundamental frequency. No subjective data have been reported. The fundamental frequency of an EPI sequence is most prominent, and therefore most easily be detected and possibly attenuated most. The published spectra show that the rest of the spectrum is hardly affected. The removal of the most prominent peak reduces the total sound pressure to the integral of the rest of the pressure spectrum. This leads to about 20 dB reduction in total SPL. This makes it still worthwhile to use active noise control during MRI.

**Lorentz force balancing**

Sir Peter Mansfield and coworkers have been active in tackling the vibration problem at the source: the Lorentz forces acting on the gradient coils. Closed arc loops are fabricated in which the Lorentz forces acting on the separate wire segments are opposite, and therefore cancel. The arcs are casted in polystyrene resin mould to mechanically couple the wire segments of one arc. The segments have some space between them, else the currents through the segments would also cancel the magnetic fields produced by the currents flowing through the separate arc segments. With an optimized distribution of such coils, gradient linearity can be obtained within the imaging volume (Chapman and Mansfield [1995]). However, the resin deforms under alternating forces, and radiates an acoustic wave perpendicular to the resin surface (Mansfield et al. [1995]). The problem of the compressive wave in the resin has been further investigated and described theoretically in succeeding articles (Mansfield and Haywood [2000]; Mansfield et al. [2001]). To fully understand the problem, arcuate coils are replaced by a rectangular loop, embedded in a rectangular plate of resin. When driven with an alternating current inside a magnetic field, the rectangular plate radiates sound perpendicular to the plate. With the material parameters (size, density, Young’s modulus), the acoustic emission (SPL, angular distribution) can be predicted. This is also validated experimentally. To enhance the acoustic radiation, an extra loop is added in the center of two halve plates, separated by a small air gap. Driving that loop with a current of appropriate amplitude.
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Figure 5.2: Top view of a plate containing a current loop. Lorentz forces acting on the wires causes stress. The direction of the stress causes compression (solid lines), or tension (dotted lines). With alternating forces, acoustic waves are produced, perpendicular to the plate surface.

Figure 5.3: Front view of a plate containing a current loop. A: The Lorentz forces exert tension on the plate. B: The solution as proposed by Mansfield. An additional current loop, in combination with a segmented plate, prevents a compressional wave. The plate segments, however, experience a force by both wire segments in the same direction, and hence, these segments displace. The original goal of Lorentz force balancing was to prevent this displacement.
and phase-lag, the compressive wave is not building up, but more or less damped by impedance matching. The optimum phase-lag depends on the compressional wave velocity and the dimensions of the plate. Results are reported for different plate materials, which can be as high as 50 dB attenuation at specific frequencies and microphone positions. The average reduction is about 30 dB. It is shown, that with a proper placement of thus constructed coils, gradient fields can be produced.

Finally, it is shown how these coils can be used for imaging (Chapman et al. [2003]). The coils are driven with a sinusoidal electric current, matching the plate’s compressional wave resonance frequency. Instead of just switching the currents on and off (multiplying with a tophat signal, with a sinc as Fourier transform), the currents have an envelope that is generate with the Fourier transform of a truncated sinc. This process, which is called optimized gradient pulse, gives an additional reduction in the produced SPL. A 50 dB reduction is reported. It should be noted that this is a 50 dB reduction at only the plate resonance frequency. Taking all the harmonics into account, the top hat gradient pulse does give a reduction of about 37 dB. However, the optimized gradient pulse gives a reduction of 41 dB in total SPL, instead of the reported 50 dB. When the contribution of the fundamental frequency is decreased, the influence of the harmonics cannot be neglected anymore.

It is not clear what the efficacy of Lorentz force balancing eventually is. The reduction is limited to about 10 dB at low frequencies (Mansfield et al. [1995]). Is this an intrinsic limit or does the compressional wave noise becomes the loudest source? No explanation is given on this topic. Further, the idea of Lorentz force balancing leans on mechanical coupling of two loop segments. The additional current loop to overcome the compressional wave problem only makes sense when there is a gap between two plate halves. In our view, the gap introduces a decoupling between the two wire segments that, for Lorentz force balancing, should be mechanically coupled. The solution to the compressional wave problem is the end of the solution to the initial problem. Without the use of active acoustic control, acoustic absorbent material can reduce the compressional wave sound, while Lorentz force balancing reduces the gradient coil sound.

### 5.4.3 Silent sequences

The greater part of what is described before, can be implemented in newly built scanners. For existing systems, substantial hardware modifications would be required. There are intelligent ways to deal with the scanner noise to avoid influence of scanner noise in functional imaging without these (expensive) hardware modifications (Amaro et al. [2002]).
5.4. SOUND REDUCING METHODS

Scanning techniques

Functional brain mapping uses the blood oxygenation level-dependent (BOLD) effect. After a stimulus has been presented, there is an increase in blood flow to the activated brain region to supply oxygen for the aerobic glucose metabolism. The MR signal depends on the oxygenation-level of the blood, therefore an increase in oxygenized blood flow results in an increase of signal. This hemodynamic response has a delay compared to the stimulus. Maximal MR signal is obtained 4–7 seconds after the stimulus, and the hemodynamic response is back to baseline after 12–15 seconds after stimulus onset. These numbers may depend on stimulus and the active brain area.

Scanner noise itself is a stimulus (see above), which also gives a BOLD response. The BOLD response of the scanner noise however, does not coincide with the BOLD response of a stimulus as acquisition is best done when the signal is highest. Continuous acquisition would lead to a baseline shift (in auditory fMRI), giving an apparent reduced response to stimuli. Hall et al. [1999] reported a sparse imaging technique, to map the auditory brain areas without scanner noise interference (figure 5.4B). An auditory stimulus is presented and the MR signal is recorded several seconds later. After the hemodynamic response of the scanner sound has vanished, the next stimulus can be presented. The repetition time can therefore be as large as 20 seconds. With more conditions of different stimuli, this can lead to undesirable long scanning sessions. As the hemodynamic responses might differ between subjects, there is a chance to miss the maximal MR signal response. Presenting stimuli of the same type repeatedly within one epoch elongate the hemodynamic response (figure 5.4C). A possible habituation effect is then not considered (Talavage et al. [1999]; Tanaka et al. [2000]).

A variation to this theme is presented by Yang et al. [2000]. Within a repetition time of 40 seconds between two auditory stimuli, both recordings of the BOLD response to the stimulus and of the rest condition are acquired. During the hemodynamic response due to the scanner noise, no acquisitions are made.

Assuming that the scanner noise only produces a baseline shift in the hemodynamic response, gives room for the clustered volume acquisition (Edmister et al. [1999]) or behavior interleaved gradient technique (Eden et al. [1999]). During the silent period, the auditory stimulus is presented so no scanner noise interferes with the stimulus (figure 5.4A). However, the observable dynamic range of the hemodynamic response in the auditory cortex may be affected. Resultantly, small task-related signal changes may not be observed due to scanner noise (Yang et al. [2000]).

In the case that continuous scanning is necessary or preferred, the scanning parameters can be adjusted to minimize the acoustic noise. The general shape of the acoustic transfer function of a gradient coils is a high-pass filter (cut-off at 600–800 Hz) with some resonance peaks (e.g., from the banana-shape and the cone-shape
Figure 5.4: While continuously scanning, the BOLD response due to the scanner noise becomes stationary. Presenting an audio task, should lead to detectable variation of the BOLD response. A: The detectability of an audio task is enhanced by stopping the scanning while presenting the task. The variation of the BOLD response can also be too small to detect. B: Sparse sampling does not give interference between BOLD responses due to an audio task and scanner noise. C: Continuous presentation of one task condition gives a fairly constant BOLD response. The acquisition of images produces a BOLD response, but not at times of acquisition.
5.4. SOUND REDUCING METHODS

The peaks in the spectra of the read-out gradient train pulses in EPI, and of any trapezoidal pulse, should not coincide with the peaks in the acoustic transfer function (Tomasi and Ernst [2003]; Wu et al. [2000]). Differences of up to 10–12 dB between read-out frequency and pulse-width settings are obtained. Making the ramps of the gradient pulses longer, automatically shifts the fundamental frequency to low frequencies, where the acoustic transfer is low, and the human ear is less sensitive (Brechmann et al. [2002]). This does, however, increase the acquisition time.

**Sinusoidal gradients**

Maximizing the slope duration is one of the guide rules for silent imaging, as set out by Hennel et al. [1999]. The gradient waveform should: (a) have sinusoidal gradient slopes, (b) have maximum slope duration, and (c) use a minimum number of slopes. Fast switching of gradients, introduces higher frequencies in the acoustic spectrum. Introducing soft transitions from one state to another, minimizes the harmonics that would be in the range of higher acoustic transfer. Thus limiting the bandwidth of the gradient pulses to < 200 Hz, the bandwidth of the acoustic noise is also limited as the response to gradient pulses is presumably linear (Hedeen and Edelstein [1997]). In standard EPI, the time between consecutive pulses is too short to apply this technique. Applying this technique increases the acquisition time, initially limiting its use to anatomical images (Girard et al. [2000]) that are recorded with an SPL of 59 dB(A), just above background noise of 55 dB(A).

Later, it has been shown that this method could be improved by removing all plateaus in the gradient waveform, and use pure sinusoidal gradient waveforms (Hennel [2001]). This results in an additional 10–20 dB reduction in acoustic noise, in comparison to the use of only sinusoidal ramps. Image reconstruction first needs regridding as \(k\)-space trajectories are curved. This procedure is also necessary in spiral trajectory imaging (Oesterle et al. [2001]). Comparing image quality with standard pulse sequences yields only minor differences, while the acoustic noise is reduced significantly. Functional imaging with this silent technique is possible (Loenneker et al. [2001]; Marcar et al. [2002]). The BOLD response can be measured but needs a long imaging time and is therefore more sensitive to scanner drift. EPI is not replaced by this technique, but can be a better alternative when acoustic noise is not desirable. Examples are studying the influence of scanner noise, or when patients are unsettled, like pediatric or psychiatric patients.

**SENSE**

Undersampling \(k\)-space trajectories improves temporal resolution in MR imaging, or, with additional interpolation, it provides a better spatial resolution without changing acquisition time (Jesmanowicz et al. [1998]; Tsai and Nishimura [2000]).
tifacts and SNR changes due to this technique are discussed. One way to improve these limiting factors, is with Sensitivity Encoding (SENSE). Using a multi-coil array, images are obtained in less time, depending on the number of coils used (Pruessmann et al. [1999]). The sensitivities of the separate coils to the MR signal from one position are recorded, and these provide spatial encoding possibilities. fMRI is also possible with this technique (Golay et al. [2000]; Preibisch et al. [2003]).

SENSE can be used to acquire more lines in the same time as with full k-space imaging, or for faster imaging. By applying SENSE, but keeping the acquisition time constant, de Zwart et al. [2002] decreased the slew rates and the maximum gradient strength. This effectively lowers the harmonic frequencies of the gradient pulses (a change in pitch is reported), and lowers the Lorentz forces on the gradient coil structure. In this way, a reduction of the acoustic noise is achieved. Theoretically, with this method applied to rate-2 and rate-3 SENSE, a reduction of respectively 12 and 19 dB (linear) can be achieved. Averaged over three different scanners, reductions of 11.3 dB(A) and 16.5 dB(A) are found. When the read-out gradient was turned off, a reduction of 16.4 dB(A) was found. Other differences can be attributed to the shift of frequencies and scanner frequency response functions. Important to notice is that this application of SENSE does not significantly alter the sensitivity to detect brain activation.

### 5.5 General issues

This section presents some general considerations.

First, there are substantial differences between the measurement protocols that have been used in the sound pressure measurements reported so far. This complicates the evaluation.

Further, some sound recording devices may be affected by strong and changing magnetic fields and the RF pulses.

Sound pressure measurements taken some distance from the bore do not necessarily reflect the SPLs inside the bore. The levels outside the bore contain strong components generated by the scanner’s outer shrouds, and also from multiple reflections from the walls. The same complexity makes the evaluation of reported attenuation more difficult. For the proper acoustical assessment, both room characteristics and precise microphone position and properties are required, and both attenuation outside the scanner and inside the bore should be reported.

In general, attenuation of acoustic waves is frequency dependent. It appears to be common practice, however, to present a single value. This is usually the attenuation value at the frequency for which maximum attenuation is achieved, which overestimates the broadband result. This can lead to the situation where frequency bands which at first sight looked harmless, are going to play an important role in the result. Formally, the sound pressure level is an rms measure, which averages across
a certain time window. It contains contributions from the entire spectrum.

Shifting frequencies of signal components, or the removal of higher frequencies in signals, changes the SPL. Therefore, when results are presented, the end value after sound reducing measures only has meaning when the frequency content of the pulse sequence is known. In that respect the comparison of scanners driven with ‘worst case’ sequences is of little value. Driving the scanners with the same gradient strengths, and same slew rates gives results that can be compared. The acoustic noise should in such a comparison be one of many parameters. In such a way it is like comparing a sports car and a family car driving at top speed, and only looking at fuel economy.

Displaying measured waveforms is of little use, just like presenting reductions in percentages. Under a factor 10 in amplitude reduction (\(\approx -20 \text{ dB} \)), a visual display is not able to give a precise impression of the achieved attenuation. Linear axes when displaying spectra, or frequency selective attenuation, does not lead to a correct idea of what is presented or achieved. In the end, the perception of sound pressure levels is about logarithmic scales. Presenting attenuations in percentages should therefore be discouraged. Further, it should be made clear whether the reduction comprises energy or waveform amplitude. Attenuations of 99 % in energy correspond to 20 dB reduction, while an attenuation of 99 % in waveform amplitude corresponds to a 40 dB reduction. Further, the just noticeable difference between sound pressure levels is about 1 dB. Presenting numbers in tenths of dBs does not add much value to the results that are found.

Finally, the general idea about linearity of gradient systems (Girard et al. [2000]; Hedeen and Edelstein [1997]) should be reconsidered. This is true for the occurrence of harmonic distortion and for a nonlinear amplitude response as well. When determining the acoustic transfer function, often only one current intensity is used. It has been shown that the sound pressure does not linearly increases with the Lorentz forces (Moelker et al. [2003b]).

5.6 Discussion and conclusion

A number of approaches, both passive and active, aiming to reduce MR-related acoustic noise, has been analyzed. In general, the attenuation results of more than one measure are not simply additive in dBs (\(=\) multiplicative in power). Moreover, it is necessary to consider the precise frequency characteristics of the single steps. MRI scanning at a lower rate changes the dB(A) level, because on the one hand, the fundamental frequency is reduced in strength under 500 Hz, on the other hand, more overtones are present in the frequency range from 500 to 4000 Hz.

Passive attenuation with earplugs and mufflers (the most widely used attenuators) gives less attenuation at the lower frequencies (Moelker and Pattynama [2003]). A combination with active noise reduction that performs well at low fre-
quencies, should give appreciable results. The combination of SENSE and a gradient coil in a vacuum chamber should also give good results, only until the eddy current evoked vibrations start to become the main contributor to the sound pressure.

The conclusion that acoustic noise need not be considered an annoying but unavoidable feature of MR imaging examinations (Girard et al. [2000]) is wrong. Effort should be put in to the research of sound reducing methods in (functional) MR imaging. The combination of various approaches should be tested to the limit to have the highest possible sound reduction.