Review

Invited

Auditory Noise Associated With MR Procedures: A Review

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This review article discusses the various types of acoustic noise produced during the operation of MR systems, describes the characteristics of the acoustic noise, and presents information regarding noise control techniques. In addition, the problems related to acoustic noise for patients and healthcare workers are discussed. J. Magn. Reson. Imaging 2000;12:37–45. © 2000 Wiley-Liss, Inc.

Index terms: magnetic resonance imaging; bioeffects; safety; noise

VARIOUS TYPES of acoustic noise are produced during the operation of MR systems. The problems associated with acoustic noise for patients and healthcare workers include simple annoyance, difficulties in verbal communication, heightened anxiety, temporary hearing loss, and potential permanent hearing impairment (1–8). Acoustic noise may pose a particular hazard to specific patient groups who may be at increased risk.

Patients with psychiatric disorders and elderly and pediatric patients may be confused or suffer from heightened anxiety (2). Sedated patients may experience discomfort due to high noise levels. Certain drugs are known to increase hearing sensitivity (3). Neonates with immature anatomical development may have an increased response to acoustic noise. For example, significant alterations in vital signs of newborns have been reported during MRI examinations, which may be attributed to acoustic noise (4).

Aside from issues of safety, acoustic noise levels also pose problems for the increasing numbers of researchers involved in functional (f)MRI studies of brain activation. For example, MR system-related acoustic noise will interfere with communication of activation task instructions that are typically given during the MRI procedure.

One area of particular interest is the study of auditory and language function. In this work, the response to pure tone stimuli is analyzed (9); any background levels of unwanted or uncontrolled acoustic noise can interfere with the delivery of these sound stimuli and the integrity of the experimental findings.

Acoustic noise levels during echoplanar imaging (EPI) have been reported to increase significantly pure tone hearing thresholds in the optimal frequency hearing range (ie, 0.1–8 kHz) (10). These effects vary across the frequency range. The threshold changes according to the characteristics of the sequence-generated acoustic noise (10). Notably, it may be possible to take into account the MR system-induced auditory activation by using a control series of scans in task paradigms (11). Experimental results have been reported for mapping auditory activation induced by MR system-related acoustic noise (11).

HEARING AND THE IMPACT OF NOISE

The ear is a highly sensitive wide-band receiver. The human ear does not tend to judge sound powers in absolute terms but assesses how much greater one power is than another. Combined with the very wide range of powers involved, the logarithmic decibel scale, dB, is used when dealing with sound power.

The sound level measured will depend not only on the source but also the environment (ie, the proximity of surfaces that may reflect the sound). Thus, the sound level is usually designated in terms of the sound pressure level (SPL), which accounts for the environment of the measurement.

The sensitivity of the ear is also frequency dependent. Peak hearing sensitivity occurs in the region of 4 kHz (9). This is also the region where the potential maximum hearing loss will occur, with damage spreading into neighboring frequencies.

Since the ear is not equally sensitive to all frequencies, data may be weighted using the dB (A) measurement scale, which biases the meter to respond similarly to the human ear. The quality or efficiency of hearing is defined by the audible threshold, that is, the SPL at which one can just begin to detect a sound.

Noise is defined in terms of frequency spectrum (in Hz), intensity (in dB), and duration. Noise may be steady state, intermittent, impulsive, or explosive. Transient hearing loss may occur following loud noise, resulting in a temporary threshold shift (shift in audible

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threshold). Brummett et al (1) have reported temporary shifts in hearing thresholds in 43% patients scanned without ear protection and also in those with improperly fitted protection. Recovery from the effects of noise should be exponential and occur quickly (1). However, if the noise insult is severe, full recovery can take up to several weeks. If the noise is sufficiently injurious, this may result in a permanent threshold shift (ie, permanent hearing loss) at specific frequencies.

Table 1
Permissible Exposure Levels to Acoustic Noise[a]

Noise duration/day (hr)	Sound level (dB)*
8	90
6	92
4	95
3	97
1.5	100
1	102
0.5	105
0.25	115

[a]These U.S. federal guidelines refer to the upper limits for occupational exposures to acoustic noise. No recommendations exist for nonoccupational or medical exposure.
*A-weighted scale.

MRI-RELATED ACOUSTIC NOISE

The gradient magnetic field is the primary source of acoustic noise associated with MR procedures (12–20). This noise occurs during the rapid alterations of currents within the gradient coils. These currents, in the presence of a strong static magnetic field of the MR system, produce significant (Lorentz) forces that act upon the gradient coils.

Acoustic noise, manifested as loud tapping, knocking, or chirping sounds, is produced when the forces cause motion or vibration of the gradient coils as they impact against their mountings which, in turn, also flex and vibrate.

Alteration of the gradient output (rise time or amplitude) caused by modifying the MR imaging parameters will cause the level of gradient-induced acoustic noise to vary. This noise is enhanced by decreases in section thickness, field of view, repetition time, and echo time. The physical features of the MR system, especially whether or not it has special sound insulation, and the material and construction of coils and support structures also affect the transmission of the acoustic noise and its subsequent perception by the patient and MR system operator.

Characteristics

Gradient magnetic field-induced noise levels have been measured during a variety of pulse sequences for clinical MR systems with static magnetic field strengths ranging from 0.35 to 1.5 T (12–21) and one 2.0 T research system (19). Hurwitz et al (13) reported that the sound levels varied from 82 to 93 dB on the A-weighted scale and from 84 to 103 dB on the linear scale (13). The report concluded that gradient magnetic field-induced noise was an annoyance but well within recognized safety guidelines (13).

Later studies performed using other MR parameters including "worst-case" pulse sequences showed that, unsurprisingly, fast gradient-echo pulse sequences produced the greatest noise during MR imaging (14–17); 3D sequences (eg, MP-RAGE), in which multiple gradients are applied simultaneously, are among the loudest sequences. Acoustic noise levels in these investigations did not exceed a range of 103–113 dB (peak) on the A-weighted scale (14–17).

More recent studies have been performed to include measurement of acoustic noise generated by echoplanar and fast spin-echo sequences (19–21). Echoplanar sequences, in collecting a complete image in one radiofrequency (RF) excitation of the spin system, tend to have extremely fast gradient switching times and high gradient amplitudes. They can generate potentially high levels of acoustic noise, although the duration of the sequence (and thus patient exposure) is shorter than conventional sequences.

Shellock et al (20) reported comparable high levels of noise of 114–115 dBA on two different high-field-strength MR systems tested when running EPI sequences with parameters chosen to represent a worst-case protocol. Although high, the recorded acoustic noise levels were within current permissible limits (Table 1).

Increased interest in the use of diffusion-weighted and fMRI techniques has meant an increased utilization of ultra-high-field-strength MR systems (eg, MR systems with ≥3 T static magnetic fields) with fast gradient capabilities (25–30 mT/m switching rates and high amplitudes) to acquire high-quality, multislice EPI images. At this point, no comprehensive data are available from MR systems with static fields above 2.0 T or gradient strengths greater than 25 mT/m.

Notably, measurements of sound pressure levels offer a limited amount of information with regard to the quality of the noise and its impact on hearing. In addition to measurements of noise level, several authors have recorded and analyzed the acoustic noise (12,16–19,21). Similar noise levels and characteristics are found when comparing different clinical MR systems (17). Frequency analysis of the noise shows that noise is pseudo-periodic, with variation in the degree of periodicity depending on the imaging parameters used (17). Peak noise levels are found at the low-frequency region of the spectra.

Figure 1 shows an example of the time and frequency acoustic noise spectra for a fast spin-echo (FSE) pulse sequence. Spectral peaks are found in the range of 0.2–1.5 kHz (17). Characteristics of the frequency spectra depend on the MR system hardware and protocol parameters. The presence of acoustic dampers (eg, encased acoustic absorbent foam surrounding the magnet) reduces noise levels by 3 dB (peak on A-scale) or 9 dB (RMS A-scale) (17). Notably, Cho et al (19) found that prescan noise generated high levels (100 dB-C) across a wide spectral range up to 4 kHz with peaks around 2.4 kHz.

In addition to dependence on imaging parameters,

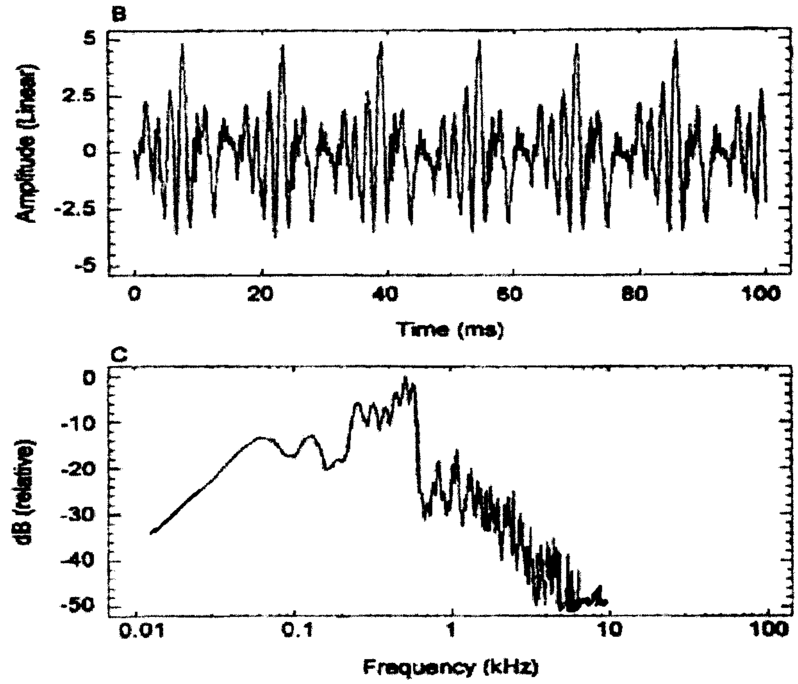

Figure 1. Time (top) and frequency (bottom) spectra for scanner-induced acoustic noise generated by an FSE sequence. (Reprinted with permission.)

acoustic noise is dependent on the MR system hardware, construction, and the surrounding environment. Noise characteristics also have a spatial dependence. For example, noise levels have been found to vary by 10 dB as a function of patient position along the magnet bore (18). The presence and size of a patient may also affect the level of acoustic noise. An increase of from 1 to 3 dB has been measured with a patient present in the bore of the MR system (18), which may be due to pressure doubling (ie, an increase in sound pressure) close to a solid object, as sound waves reflect and undergo an in-phase enhancement.

Hedeen and Edelstein (18) reported the similarity between the gradient pulse spectrum and the acquired noise spectrum, which is affected by additional system acoustic resonances (Fig. 2). These investigators derived an acoustic transfer function, which is independent of input (18). Once it is determined, it may be applied to any input impulse function to predict the generated acoustic noise (18). This model has been applied to an FSE sequence and achieved an agreement between measured and predicted noise level to within 0.4 dB. Figure 3 shows the (a) predicted and (b) measured noise spectra for the FSE sequence.

Permissible Limits

In general, the acoustic noise levels recorded by various researchers in the MR environment have been below the maximum limit permissible by the Occupational Safety and Health Administration of the United States. This is particularly the case when one considers that the duration of exposure is one of the most important physical factors that determines the effect of noise on hearing (22–24).

Other physical factors involved in hearing loss include the sound frequency, temporal pattern, and intensity of noise (8,22–25). High root-mean-square (RMS) and peak noise levels at low frequency can induce strong vibration of the entire cochlea, which may lead to a temporary shift in hearing threshold or potentially permanent shift in noise sensitive patients.

Table 1 shows the relationship between the noise

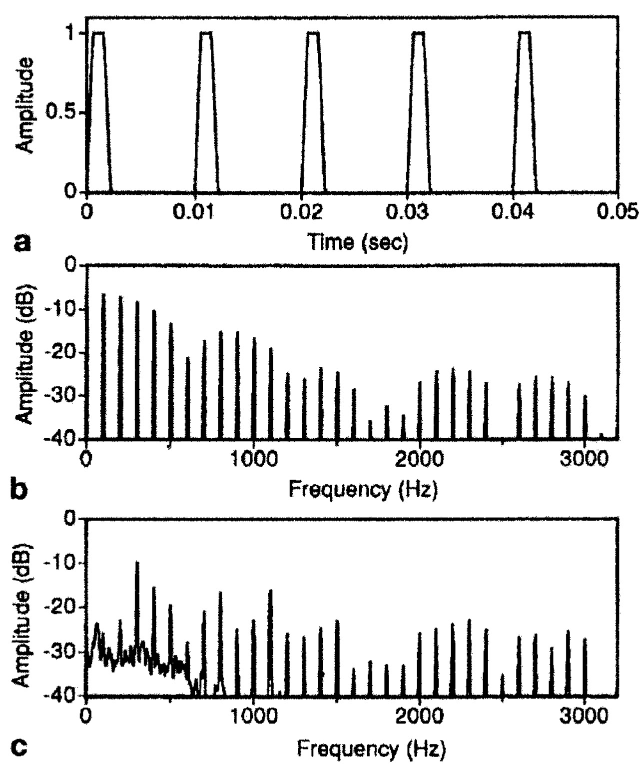

Figure 2. Gradient trapezoidal current excitation and acoustic response. **a:** Trapezoidal current waveform time series. **b:** Fourier transform of trapezoidal time series. **c:** Measured acoustic response. (Reprinted with permission.)

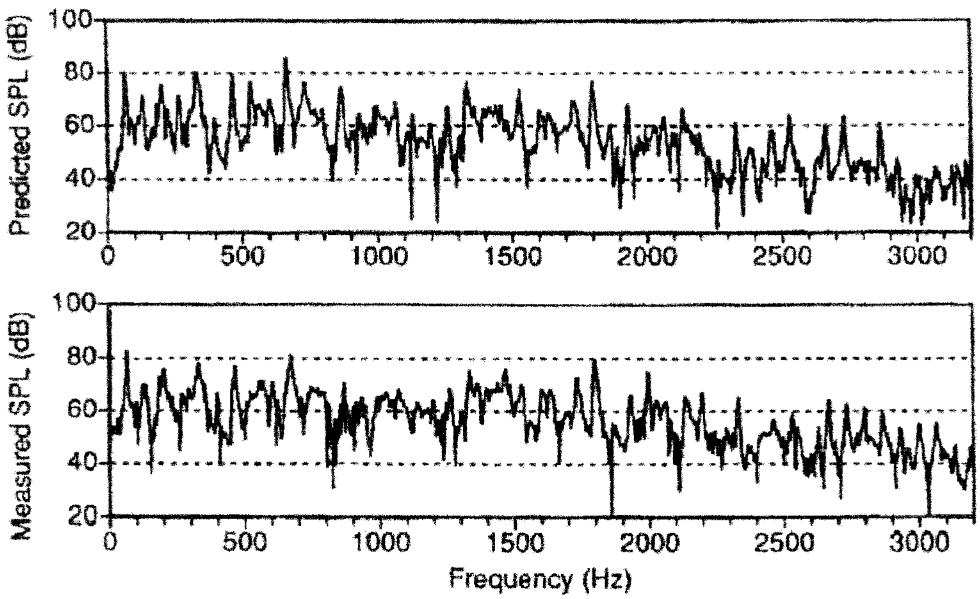

Figure 3. Predicted and measured SPL for an FSE sequence. Overall predicted and measured levels are 93.1 and 92.7 dB, respectively.

duration and recommended permissible sound levels for occupational exposures. The U.S. Food and Drug Administration indicates that the acoustic noise levels associated with the operation of MR systems must be shown to be below the level of concern established by pertinent federal regulatory or other recognized standards-setting organizations (26). If the acoustic noise is not below the level of concern, the sponsor (ie, the manufacturer of the MR system) must recommend steps to reduce or alleviate the noise perceived by the patient.

These recommended limits for acoustic noise produced during MR procedures are based on recommendations for occupational exposures that are inherently chronic exposures with respect to the time duration (26). Of note is that comparable recommendations do not exist for nonoccupational exposure to relatively short-term noise produced by medical devices.

Guidelines in the United Kingdom have been relaxed since the initial measurements of MR-related acoustic noise. Current for the safe use of medical equipment state that exposure to acoustic noise must be restricted to levels below 99 dBA (27). This would suggest that hearing protection should always be worn by patients undergoing MR procedures. The document includes no recommendations on the duration of exposure.

The exposure of staff and other health workers in the MR system environment is also a concern (eg, those involved in interventional MR procedures or who remain in the room for patient management reasons). Shellock et al (20) reported levels of noise ranging from 108 to 111 dB at the entrances and exits of the magnet bores of MR systems while EPI sequences are running. The acceptable duration for exposure to these noise levels is 15–30 minutes (Table 1). This suggests that staff should wear ear protection if they remain in the MR system room for longer periods of time. In the United Kingdom, guidelines issued by the Department of Health recommend that hearing protection be worn by staff exposed to an average of 85 dB over an 8 hour day (28).

While acoustic noise levels suggested for patients exposed during MR procedures on an infrequent and short-term temporal basis are considered to be highly conservative, they may not be appropriate for individuals with underlying health problems, who may have problems with noise at certain levels or at particular frequencies. The acoustic noise produced during MR procedures represents a potential risk to patients.

As mentioned previously, the possibility exists that significant gradient magnetic field-induced noise may produce substantial hearing problems in patients who are susceptible to the damaging effects of loud noises. In fact, there have been unconfirmed claims of permanent hearing loss associated with MR examinations (8).

NOISE CONTROL TECHNIQUES

Passive Noise Control

The simplest and least expensive means of preventing problems associated with acoustic noise during MR procedures is to encourage the routine use of earplugs or headphones (1,6–8). Ear plugs, when properly used, can abate noise by 10–30 dB, which is usually an adequate amount of sound attenuation for the MR environment. The use of disposable earplugs has been shown to provide a sufficient decrease in acoustic noise that, in turn, would be capable of preventing the potential temporary hearing loss associated with MR procedures (1).

Therefore, it behooves all MR facilities to recommend that patients undergoing MR examinations wear these protective devices. MR-safe headphones that substantially muffle acoustic noise are also commercially available.

Unfortunately, the passive noise control methods suffer from a number of limitations. For example, these devices hamper verbal communication with patients during the operation of the MR system. In certain circumstances, they can cause discomfort or inhibit the immobilization of the patient's head when optimal immobilization is required for certain studies that are sensitive to patient movement (ie, diffusion-weighted and fMRI studies). Additionally, standard ear plugs are often too large for the ear canal of young infants.

Importantly, passive noise control devices offer nonuniform noise attenuation over the hearing range. While high-frequency sound may be well attenuated, attenuation is often poor at low frequency. This is unfortunate because the low-frequency range also occurs where the peak MR-related acoustic noise is generated.

Passive noise control techniques provide poor attenuation of noise transmitted to the patient through bone conduction (29). Notably, the presence of an insulating foam mattress on the patient couch has been found to reduce vibrational coupling to the patient and noise levels by around 10 dB (21).

"Quiet" MRI Sequences

Since the dominant effect on acoustic noise levels lies with the signal details of a particular scan protocol rather than the structure of the MR system (18), it follows that it should be possible to reduce the noise level by optimizing the choice of the imaging parameters. Simply using a spin-echo (SE) sequence rather than a gradient-echo (GE) sequence and running the sequence with reduced gradient parameters (rise time and amplitude) can significantly reduce the levels of acoustic noise.

Skare et al (30) have designed "quiet" sequences in this manner. They also reported the so-called "quietness factor" (QF) as being the ratio of rise time of the modified sequence and the original/default rise time. Thus, the "quietness factor" (QF) is:

$$QF = RT_m / RT_s$$

where RT_m is the rise time of the modified sequence and RT_s is the original/default rise time. For example, on a 1.5 T Signa MR System (General Electric Medical Systems, Milwaukee, WI) a QF of 6 resulted in a noise attenuation of 20 dB. This procedure, however, lengthens the TE, reduces the number of acquisition slices, and results in a longer examination time.

A reduction in acoustic noise levels may also be achieved by decreasing the level of gradient pulsing in an imaging sequence. For example, stimulated echo acquisition mode (STEAM)-burst is a rapid single-shot technique based on stimulated echoes without the rapid gradient switching necessary in other single-shot techniques such as EPI (31). STEAM-burst uses a combination of STEAM along with the burst technique (32), involving the application of multiple RF pulses under a constant gradient and subsequent refocusing of the resultant set of echoes (33).

Limited data on acoustic noise measurements show peak noise attenuation of 15 dB compared with a similar EPI sequence (31). This sequence offers the potential for rapid acquisition of images, is insensitive to artifacts due to B_0 inhomogeneities, and generates reduced levels of acoustic noise. However, this sequence suffers from a low signal-to-noise ratio (SNR) and reduced image resolution.

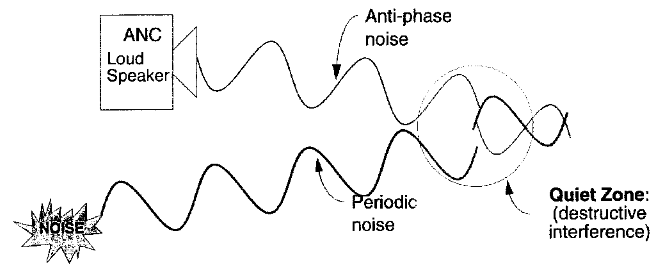

Figure 4. Principle of sound attenuation using antiphase noise to create a "zone of quiet."

Cho et al (19,34) developed a "quiet" MRI technique based on a variation of the projection reconstruction method, which also minimizes gradient pulsing. In conventional projection reconstruction, the frequency and phase-encoding gradients remain simultaneously at a low level throughout the acquisition of a line of image data (35). Less gradient pulsing is done, and gradient amplitudes are generally lower than in conventional 2D Fourier transform imaging.

Cho et al (19,34) reduced gradient pulsing with a projection reconstruction type of sequence and replaced the two gradient pulsings, with a single mechanically rotating DC gradient coil (Fig. 4). In addition, the DC gradient remains on for the entirety of the image acquisition. (In conventional projection reconstruction, the gradients are pulsed on for the acquisition of each line of image data.)

Use of the "quiet" MRI technique results in a 20.7 dB attenuation in sound level (34). This technique suffers from two important limitations: the slices may be selected in the *z*-axis only (due to the readout gradient rotating around the *z*-axis), and there is a loss of slice volume at each angle of rotation (due to a tilting of the selected slice caused during gradient rotation).

Active Noise Control

A significant reduction in the level of acoustic noise caused by MR procedures has been accomplished by implementing the use of an active noise cancellation (or "antinoise") technique with the existing audio system (12,36,37). Controlling the noise from a particular source, by introducing antiphase noise to interfere destructively with the noise source, is not a new idea (38,39). In fact, initial results were disappointing.

Goldman et al (12) used combined passive and active noise control system (ie, an active system built into a headphone), achieving an average noise reduction of around 14 dB. This performance is similar to that of a standard passive headphone.

Recent advances in digital signal processing (DSP) technology allow efficient active noise control (ANC) systems to be realized at a moderate cost (40). The antinoise system involves a continuous feedback loop with continuous sampling of the sounds in the noise environment so that the gradient magnetic field-induced noise is attenuated. It is possible to attenuate the pseudo-periodic scanner noise while allowing the transmission of vocal communication or music to be maintained. Some commercial manufacturers are cur-

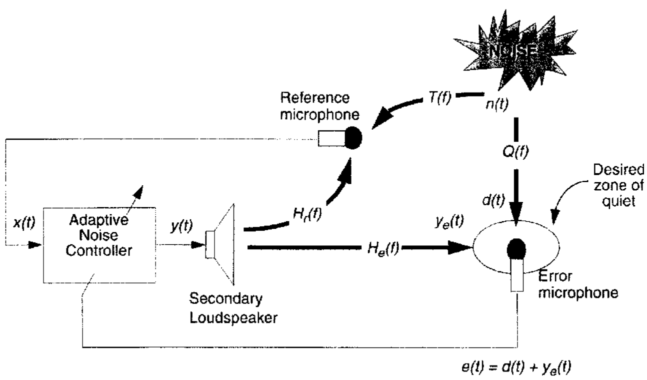

Figure 5. A schematic of the adaptive ANC system. The acoustic transfer path from the secondary source to the error microphone is denoted as $H_e(f)$, and the feedback from the secondary source to the reference microphone is denoted $H_r(f)$.

rently offering ANC systems, based on delivering antinoise to headphones in a manner similar to airline music systems.

McJury et al (36) reported the sound attenuation of MR system-generated acoustic noise with a real-time adaptive ANC system. In the adaptive ANC controller, the system attempts to minimize the error signal power using a feedback control algorithm (9). If it is successful, a "zone of quiet" appears around the error microphone (Fig. 5). A number of general points must be addressed by the architecture, namely:

1. The use of a feedback algorithm requires that careful attention be paid to algorithm stability and numerical integrity.
2. If a reference microphone is to be used, there is a feedback path through the acoustic transfer path that may cause stability problems.
3. A larger "zone of quiet" than achievable with one microphone is likely to be desirable.
4. Depending on the physical location of the reference microphone and error microphones, the noise controller may not be able to cancel random noise due to the noncausality of the architecture.

To address these points, a multichannel, filtered U-least mean square (LMS) algorithm was implemented, whereby more than one secondary source and error microphones were used (16). Scanner noise was measured with octave band sampling and found to peak at approximately 250 Hz (16). Thus, the ANC controller was optimized to control noise below 500 Hz by using a sampling rate of 2 kHz and antialias filter cutoff at 700 Hz.

Acoustic noise was recorded digitally from a series of typical clinical MR protocols that included SE and GE sequences performed on a 1.0 T MR system. This noise was then replayed through the bench-top adaptive real-time DSP ANC system (Motorola DSP 56001). A typical peak sound attenuation of approximately 30 dB was achieved over the frequency range 0–700 Hz (Fig. 6).

Chen et al (37) used a similar bench-top adaptive technique, achieving an average noise attenuation of 18.8 dB with a cutoff of 4 kHz. The results clearly showed that antinoise could efficiently cancel MR system-related noise, while leaving speech and music pneumatically piped into the MR system relatively unaffected (37).

"Quiet" MR System Gradient Coils

Obviously, the best solution for eliminating noise is to address the source by designing a "quiet" gradient coil. Gradient coil windings can be designed such that all Lorentz forces generated by the pulsing of current are balanced (ie, each force is effectively cancelled by one of equal magnitude at a conjugate position relative to the coil center) (41–44). If the coil is considered as a harmonic oscillator and coupled to another, back-to-back (Fig. 7), and if the masses and spring constants are equal, the center of mass of the system will be constant without the need for a heavy mounting apparatus (41). This is simplistically the principle of active force balancing, which may be applied to gradient coil design.

Lorentz force balancing is only effective in those parts of a gradient coil that lie in the homogeneous magnetic field. However, most gradient coil formers lie in the part of the magnet where the magnetic field is inhomogeneous.

Figure 7 shows that for a rectangular coil oriented perpendicular to the magnetic field direction, all forces F and F' are balanced. All solids have viscoelastic properties that result in residual movement of the conduc-

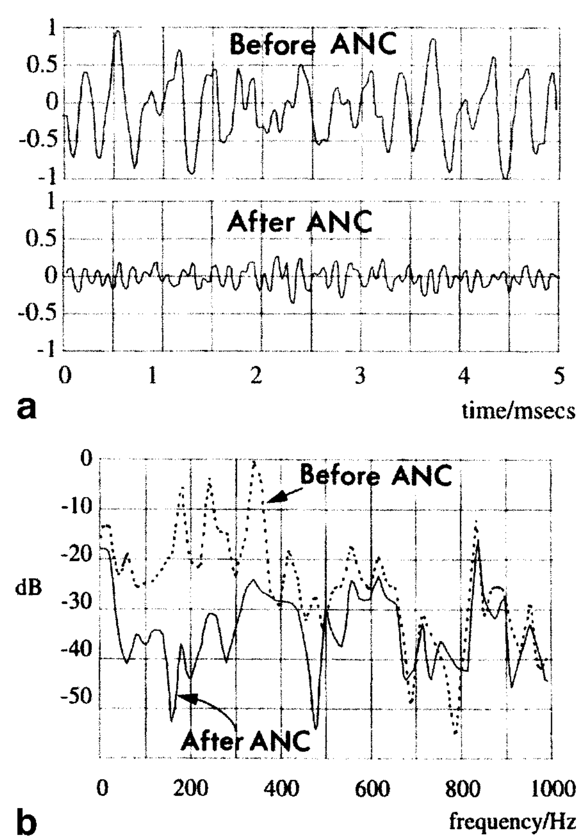

Figure 6. Results of noise cancellation for a typical clinical SE sequence. Noise level spectra before (dotted line) and after (solid line) cancellation are shown for time (**a**) and frequency (**b**) domain spectra. (Reprinted with permission.)

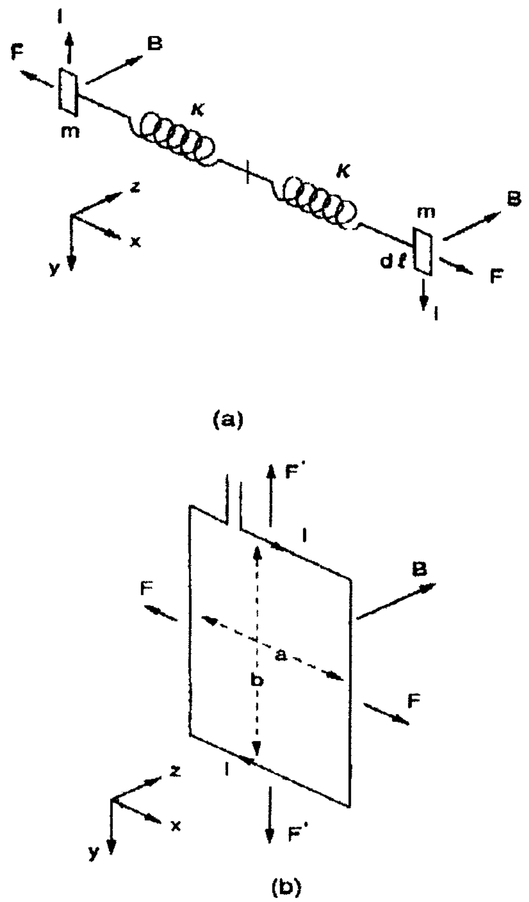

Figure 7. a: Diagram representing two coupled line elements of conductor (dl) of equal mass (m) carrying equal and opposite currents. The center of mass of the system remains fixed if the spring constants (κ), are equal. The system is placed in a magnetic field (B), which gives rise to the forces (F) causing displacements. b: Rectangular conductor loop carrying a current (I), placed in a magnetic field (B), such that the loop plane is normal B. All forces (F and F′) are balanced. (Reprinted with permission.)

tors, limiting the ideal noise cancellation. These movements produce compression waves propagating through the material with velocity,

$$v = (E/\rho)^{1/2}$$

where E is Young's modulus and ρ is the density of the material. Knowing that wave velocity and frequency are related by $v = f\lambda$, a slow wave velocity, will result in low wave frequency, above which progressive phase effects will be expected that will interfere with noise cancellation.

For optimal acoustic noise cancellation, strut material should have a large value for E and small value for ρ, resulting in a high compressional wave velocity. Thus, a light coupling structure of high strength may perform as well as a heavier structure. Composite materials (eg, glass-filled nylon) have been tested and found to perform well with a frequency response up to 20 kHz (ie, for a single loop coil of dimensions 30×20 cm) (41).

Using a bench-top prototype two-coil system (a square coil design measuring 40 cm along each dimension, powdered by a 10A sinusoidal current), the noise attenuation, when powered in balanced mode, was approximately 40 dB at 100 Hz, dropping to 0 dB at 3.5 Hz. Notably, these results agree reasonably well with the theoretical prediction (41).

Designs have also been extended and tailored to include the capability of current balancing, acoustic screening, and magnetic field screening in one coil (44). A head-gradient coil has been designed analytically using coaxial return paths. When performing EPI (eg, 3 mm slice thickness, gradient switching frequency 830 Hz, current 207 A) in a 3.0 T whole-body MR system, noise levels were calculated to be 102 dB (RMS) (44).

The implications for gradient coil characteristics of the acoustic screening design mean a loss of gradient strength and increase in coil inductance and, therefore, a decrease in the high-speed performance of the MR system. Estimates of the increase in inductance for a three-cylinder coil system (ie, fully force-shielded and acoustically and magnetically shielded) over a coil with only active magnetic shielding give a factor of approximately 8. To counteract this in keeping performance constant, an increase in driver current of $\sqrt{8}$ is required which will, in turn, increase the acoustic noise output of the coil.

Throughout the data on sound generation in relatively simple gradient structures (43), the highest sound pressure levels were noted to come from spurious resonances, thought to be due to bending and buckling of the coil structures (ie, Chladni resonances). Designs to completely cancel acoustic noise, including contributions from these sources, is a considerable challenge.

Passive Coil Methods

As suggested above, greater coil stiffness should reduce mechanical vibration and associated noise. Stiffness is dependent on material properties and geometric factors. Altering coil dimensions or altering materials to increase Young's modulus will help reduce acoustic noise due to vibration.

The mechanical vibration may also be attenuated by damping the coil. This can be achieved by using particular materials for construction or by mounting the coil in such a way that it is surrounded by an absorber. Damping will be most efficient at resonance and will reduce overall stiffness of the coil structure. Several commercial manufacturers use damping in their gradient coil systems (17). A reduction of around 3 dB (A) due to the use of acoustically damped commercial gradient systems has been reported (17).

OTHER SOURCES OF MR SYSTEM-RELATED ACOUSTIC NOISE

RF Hearing

When the human head is subjected to pulsed RF radiation at certain frequencies, an audible sound (perceived as a click, buzz, chirp, or knocking noise) may be heard (45–47). This acoustic phenomenon is referred to as "RF hearing," "RF sound," or "microwave hearing" (45–47).

Thermoelastic expansion is believed to be responsible for the production of RF hearing, whereby there is absorption of RF energy that produces a minute temperature elevation (ie, approximately 1×10^{-6}°C) over a brief period (ie, approximately 10 μsec) in the tissue of the head (45–47). Subsequently, a pressure wave is induced that is sensed by the hair cells of the cochlea via bone conduction. In this manner, the pulse of RF energy is transferred into an acoustic wave within the human head and is sensed by the hearing organs.

With specific reference to the operation of MR scanners, RF hearing has been found to be associated with frequencies ranging from 2.4 to 170 MHz (45). The gradient magnetic field-induced acoustic noise that occurs during MR procedures is significantly louder than the sounds associated with RF hearing. Therefore, noises produced by the RF auditory phenomenon are effectively masked and are not perceived by patients or MR operators (45).

Currently, there is no evidence of any detrimental health effects related to the presence of RF hearing (45). However, Roschmann (45) recommends an upper level limit of 30 kW applied peak pulse power of RF energy for head coils and 6 kW for surface coils used during MR imaging or spectroscopy to avoid RF-evoked sound pressure levels in the head increasing above the discomfort threshold of 110 dB.

Noise From Subsidiary Systems

Fans for patient comfort and cryogen reclamation systems associated with superconducting magnets of MR systems are the main sources of ambient acoustic noise found in the MR environment. Cryogen reclaimation systems are devices that are effectively used to minimize the loss of cryogens and function on a continuous basis. Acoustic noise produced by these subsidiary systems is considerably less than that caused by the activation of gradient magnetic fields during MR procedures. Therefore, this acoustic noise, at the very least, may only be an annoyance to patients or MR system operators.

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