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"Silent" MRI With Soft Gradient Pulses

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A method to reduce the acoustic noise generated by gradient systems in magnetic resonance imaging (MRI) is proposed based on the linear response theory. Since the acoustic frequency response function of typical gradient coils is low in the range below 200 Hz, the noise level can be significantly reduced by using gradient pulse sequences whose spectra are limited to this frequency range. Such "soft," i.e., band-limited, pulse shapes can be designed using sinusoidal ramps individually adjusted to available delays. "Silent" versions of three basic MRI sequences [gradient-echo (GE), spin-echo (SE), and rapid acquisition with relaxation enhancement (RARE)] were programmed on 2 and 3 T whole-body scanners. High-quality images could be acquired at noise levels as low as 40 dBA (GE and SE) and 60 dBA (RARE). Magn Reson Med 42:6–10, 1999.

Key words: MRI; acoustic noise; noise reduction

MRI scanners make acoustic noise. Its sources are the pulses of current generated in the gradient coil for the spatial encoding of the NMR signal. Since the gradient coil is placed inside a strong magnetic field, a pulsed Lorentz force is induced, causing vibrations of the coil structure, which in turn generate a compression wave in the air perceived as the ''scanner noise'' (1). In currently used MRI systems, noise levels of 70–110 dB are reported (2–6) depending on the acquisition method. The acoustic noise causes discomfort and anxiety in patients (7,8) and represents a severe obstacle in functional MRI studies of the auditory cortex (9–11). With increasing strength of the magnetic field, one can expect a quadratic increase of the sound pressure generated by the same MRI method. This is because the Lorentz force is proportional to the product of magnetic field and gradient amplitude, and the latter must increase proportionally to the field to keep chemical shift and susceptibility artifacts constant. The reduction of the acoustic noise becomes an important issue when fields as high as 8 T begin to be used for human MRI applications.

In addition to the simple use of sound-attenuating materials, a few methods to reduce the acoustic noise of MRI have been proposed such as active noise cancellation (ANC) using an anti-phase sound (12,13) or the construction of ''quiet gradient coils'' in which the net Lorentz force is compensated between current pathways (14,15). Our work is aimed in a complementary direction, namely, toward optimization of MRI sequences with respect to the acoustical noise. So far, little has been done in this field except for the application of a reduced gradient slew rate in fMRI (16) and a radical attempt to replace gradient pulsing with a mechanically rotated coil (17). The method we propose, recently described in an abstract (18), is based on the design of the gradient waveforms. Unlike the mechanical coil rotation, it may be implemented on standard MRI hardware and is not restricted to the spin-echo sequence. Our method can also be combined with the other approaches—ANC or quiet gradient coils—for a further reduction of the acoustic noise. Using a 3 T whole-body imager, we were able to acquire high-quality spin-echo and gradient-echo images at sound levels lower than the ambient noise of the magnet room.

MATERIALS AND METHODS

Principles

The idea is based on the work of Hedeen and Edelstein (19), who have shown that the acoustic response of an MRI gradient system to current pulses is linear. Consequently, the spectrum of the sound generated by a particular gradient waveform is given by the product of the waveform's Fourier transform (FT) and the frequency response function (FRF) of the gradient system. Our strategy is thus to design gradient current waveforms containing no frequencies for which the amplitude of the FRF is high.

The acoustic FRF of the whole-body gradient system used in ref. 19 was relatively low, in the frequency range below 200 Hz. To confirm this observation on a system from a different manufacturer, we have measured the FRF of a 72 cm inner diameter actively shielded coil (BG72, Bruker Medical) placed in a 3 T magnet and connected to Copley 247 amplifiers. The FRF was calculated as the FT of the acoustic response to a 40-lobe sinc current pulse of 10 msec duration (flat spectrum from 0 to 4 kHz) played in one of the transverse gradient channels. A significant reduction of the response at low frequencies is visible, in particular below 200 Hz (Fig. 1). We can thus expect that frequency components below this threshold will be attenuated in the acoustic spectrum of MRI sequences. The perception of low frequencies is additionally damped by the reduced sensitivity of the human hearing in this spectral range. Therefore, a significant reduction of the total noise effect should be obtained by avoiding high-frequency components in the gradient pulse sequences.

Most imaging methods use periodic gradient sequences, whose spectra consist of the fundamental peak at the frequency 1/T (T being the sequence period) and a series of harmonics at multiples of 1/T. The deviation from periodicity due to increments of the phase-encoding gradient is ignored in our discussion. Standard MRI sequences have periods of several tens of milliseconds. For example, a spin-echo two-dimensional (2D) FT sequence with a TR of 3 sec and an interleaved acquisition of 30 slices has T = 100 msec. For such methods, the fundamental and the first few harmonics fall well below the noise sensitivity thresh-

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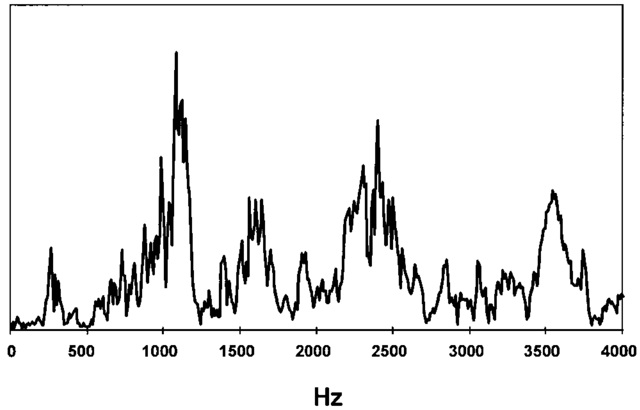

FIG. 1. Acoustic frequency response function of a 72 cm inner diameter gradient coil measured in the range 0 to 4 kHz, plotted in an arbitrary linear scale. The reduced response below 200 Hz is utilized to design the silent MRI sequences.

old discussed above and are practically not audible. The dominating contribution to the overall noise effect comes from the higher harmonics. Amplitudes of harmonics are determined by the FT of a single period of the gradient sequence. Therefore, the reduction of high harmonics could be achieved by low-pass filtering of the gradient waveforms. This would not be practical because some crucial details, such as the limits of the flat sections used for slice selection or readout would be changed, leading to artifacts. Instead, we prefer to build sequences in the traditional way, using rising and falling slopes and plateaux. To ensure that the resulting waveform is maximally band limited, the following rules should be used:

A. Sinusoidal gradient slopes
B. Maximum slope durations
C. Minimum number of slopes

Rule A means that when the gradient changes from G_1 at time 0 to G_2 at time τ, its transient amplitude at time t is given by

$$G(t) = \frac{1}{2}[G_1 + G_2) + (G_1 - G_2)\cos(\pi t/\tau)] \qquad [1]$$

It is easy to show that a pulse sequence with ramps given by Eq. [1] corresponds to a convolution of a hypothetical sequence of rectangular pulses with the function W

$$W(t) = \begin{cases} \cos^2(\pi t/\tau) & \text{when } |t| < \tau/2 \\ 0 & \text{otherwise} \end{cases} \qquad [2]$$

where τ is the duration of the slope, as illustrated in Fig. 2. According to the convolution theorem, the spectrum of such a sinusoidal sequence is a product of the spectrum of the rectangular sequence, with the spectrum of W, which contains essentially only frequencies below $2/\tau$. The power of harmonics above this cut-off limit is attenuated by more than 30 dB.

Rule B makes the cut-off frequency as low as possible, making the band limiting most efficient. The slope duration does not have to be constant within the sequence, nor need it be equal for different gradient channels, as is commonly done. Instead, gradient slopes should be individually adjusted to available delays. Consequently, there should be no plateaux of gradient pulses except for the duration of radiofrequency (RF) pulses and signal sampling. According to Rule C, consecutive gradient pulses on the same channel should be merged whenever possible. In particular, the use of spoiling pulses should be carefully revised. For example, instead of a spoiler on the slice-selection channel, a long rising slope of the slice-selection pulse (taking up to the entire repetition delay) can be programmed. We call such gradient pulses "soft" by a mechanical analogy. Indeed, a similar low-pass filtering of the force acting on the gradient coil could be provided by a layer of soft material placed between the wires and their support.

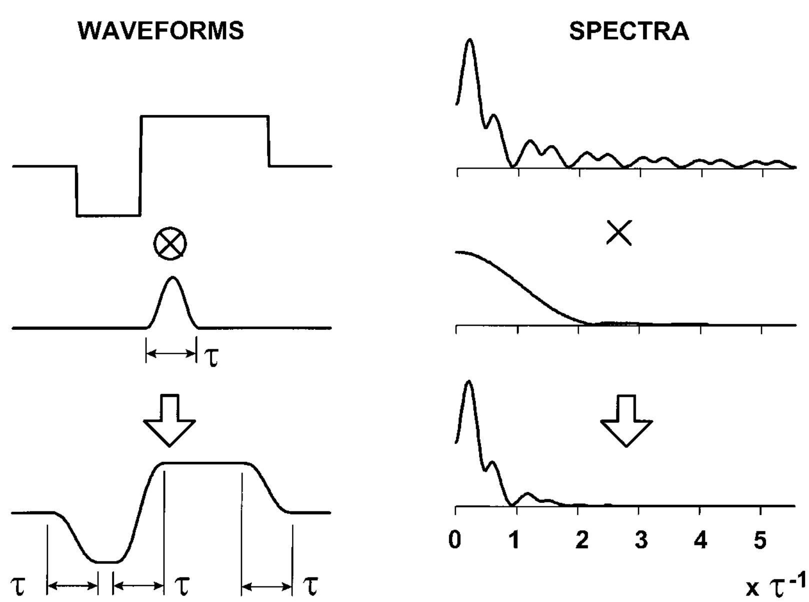

FIG. 2. Gradient waveforms and their magnitude spectra. The soft waveform with sinusoidal ramps of length τ (bottom left) can be obtained by convolution of a hard (rectangular) waveform with a cosine window of length τ. Consequently, its spectrum (bottom right) is a product of the hard spectrum with the FT of the window and is therefore practically band-limited to $2/\tau$.

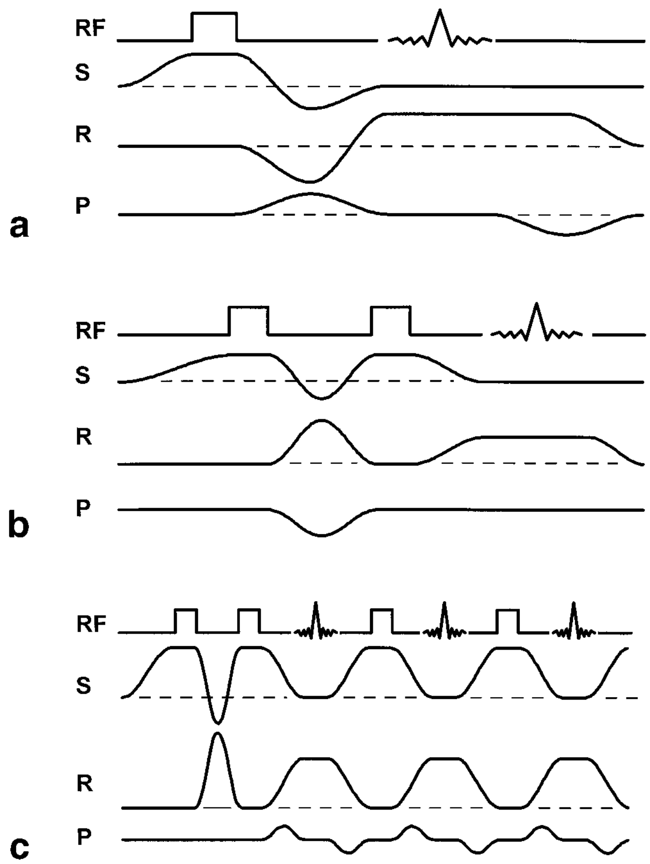

FIG. 3. Soft gradient waveforms (S, slice selection; R, read; P, phase encoding) for silent gradient-echo (**a**), spin-echo (**b**), and multiple spin-echo or RARE (**c**) imaging. The amplitudes of phase-encoding pulses for RARE vary from one echo to another.

Implementation

The rules of soft pulsing were used to design quiet versions of two standard MRI methods—gradient-echo (GE) and spin-echo (SE) imaging. Examples of gradient waveforms for these methods are plotted in Fig. 3a and b, respectively. The actual gradient pulse shapes depend on the echo time and other parameters of the sequence. A soft-pulsed multi-echo imaging sequence was also programmed (Fig. 3c), which, depending on the order of phase encoding, can also be used for rapid acquisition with relaxation enhancement (RARE) (20). The methods were implemented on Medspec Avance MRI systems (Bruker Medical, Germany) with 3 and 2 T whole-body magnets. The gradient controller of these systems allows specification of arbitrary gradient waveforms with 10,000 16-bit words on all channels, and has a minimum digital-to-analog conversion time of 3 μsec. In practice, up to 3000 words per channel were used with a 50 μsec conversion time. Acquisition software was programmed to generate gradient waveforms automatically based on imaging parameters such as the echo time, repetition time, RF pulse lengths, bandwidth, matrix size, etc. The methods were used with the whole-body gradient coil mentioned above.

The acoustic measurements were done using an RO-1350 sound level meter (ROLINE) with an electret condenser microphone placed in the center of the gradient coil and connected to the device via a shielded cable. A sound attenuation test as described in ref. 19 was performed to verify whether the measured signal was not electromagnetically induced by gradient current switching. For spectral analysis of the acoustic response, the output of the amplifier of the sound level meter was connected to one of the ADCs of the MRI system. A pulse program was specially written to digitize the microphone signal during and after the application of a single gradient waveform. All measurements were performed with a 10 kHz sampling frequency and 1024 points, covering an acoustic spectral range of 0–5 kHz with 9.7 Hz resolution.

RESULTS AND DISCUSSION

To demonstrate the influence of the shape and duration of gradient ramps on the acoustic noise we have measured the acoustic responses to the readout gradient of the GE sequence with two types of ramps: linear with 0.5 msec duration and sinusoidal with 4 msec duration. Other parameters of the sequence were identical: TE 20 msec, field of view (FOV) 15 cm, 128 points, 10 kHz bandwidth. The experiment was done in a 3 T field. The sequence with short ramps generates a broad acoustic spectrum extending beyond 1 kHz (Fig. 4). With the 4 msec sinusoidal ramps, components above 500 Hz are significantly suppressed, as expected. The overall noise level, given by the integral of the power spectrum is therefore reduced.

Noise levels of the soft-pulsed spin-echo and gradient-echo sequences were measured as a function of the echo time in the 3 T scanner. Parameters of the GE sequence were chosen as for a typical fMRI experiment: matrix size 128×128, FOV 25×25 cm, 10 kHz signal bandwidth, 4 msec gaussian RF pulse (450 Hz bandwidth), and 4 mm slice thickness. The SE sequence was used with a higher resolution (256×256) and a bandwidth of 20 kHz, corresponding to a standard application of this method—an anatomic T2-weighted scan. The repetition time of the gradient sequence was 100 msec in both cases. The phase-encoding gradient pulse was fixed at the maximum

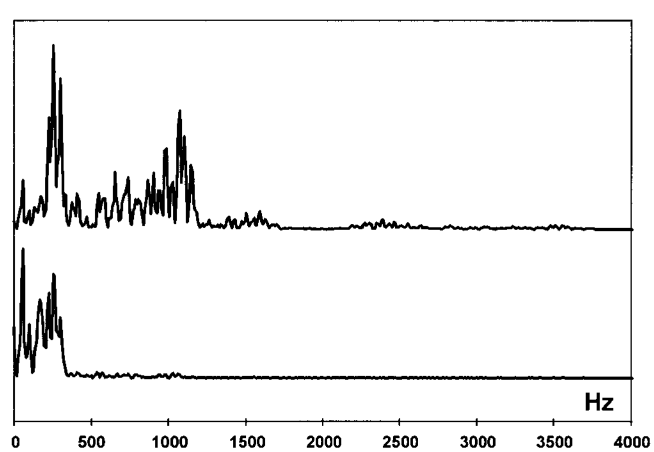

FIG. 4. Magnitude spectra (in an arbitrary linear scale) of the sound generated by the readout gradient of the FLASH sequence with different ramp types: linear, 0.5 msec duration (upper curve), and sinusoidal, 4 msec duration. The sinusoidal ramps remove high-frequency components.

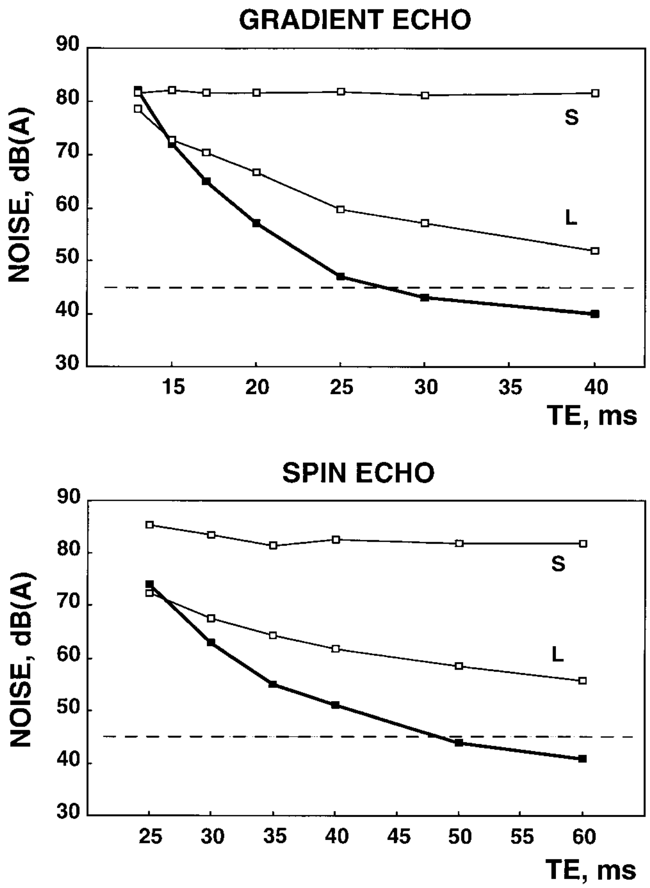

FIG. 5. Acoustic noise level of gradient-echo and spin-echo sequences measured as a function of TE at 3 T (see text for details). Thick lines, sequences using soft gradient pulses (sinusoidal ramps of maximal duration); thin lines, sequences with linear ramps of maximal duration (L), and standard sequences installed by the scanner manufacturer (S). The dashed lines show the level of noise generated by the air conditioning in the magnet room during patient examinations (turned off for the acoustic measurements).

value. Average noise levels (integrated over 1 sec) were measured with the application of A-weighting, which takes into account the frequency-dependent sensitivity of human hearing, and are expressed in dB(A). The results are presented in Fig. 5 (thick lines). The noise levels of both sequences rapidly increase with a reduction of the echo time. This is because the delay available for gradient refocusing becomes shorter, and the amplitudes of the refocusing and phase-encoding pulses have to be correspondingly higher. This leads to an increased contribution of the high-frequency components in the acoustic spectrum and increases the total acoustic noise level. However, at longer but still practically useful echo times, such as 60 msec for the SE sequence and 40 msec for GE, the noise level falls to 40 dB(A). This is well below the noise of 45 dB(A) produced by the air circulation system in our magnet room during patient examinations. The acoustic measurements were done with the air circulation turned off to avoid masking the sound of the silent MRI sequences. The level of the residual background noise was 33 dB(A).

For comparison, noise level measurements were repeated for similar GE and SE sequences with the only difference that the gradient ramps were linear instead of sinusoidal (thin lines marked "L" in Fig. 5). Such sequences resemble the method proposed in ref. 16. The advantage of sinusoidal ramps becomes apparent with increasing echo time. At a TE of 40 msec, a reduction of 10 dB(A) is achieved with respect to the linear ramp shape. Noise produced by standard GE and SE sequences installed on the scanner, which use a higher number of linear ramps of a fixed 0.5 msec duration, was also measured (marked "S" in Fig. 5). It is up to 40 dB(A) higher than with our methods based on soft gradient pulses.

The multi-echo sequence is usually run at high magnetic fields with a longer repetition time to avoid RF heating. Therefore, measurements of the noise level integrated over 1 sec would not be adequate for this method. We have therefore measured the peak noise value for the multi-echo sequence with soft gradient pulses, which was 60 dB(A) for the following parameters: 8 echoes, matrix size 256×256, FOV 25×25 cm, echo spacing 30 msec, bandwidth 20 kHz. Of course, with this noise level it is somewhat exaggerated to call this method silent. However, in comparison with the standard RARE sequence available on this scanner, which uses linear 0.5 msec ramps, a noise reduction by 22 dB was obtained.

Figure 6 shows human head images obtained with the silent sequences. The GE and SE images do not differ from the ones obtained using standard methods with identical parameters and do not suffer from any particular artifacts. A slight ghosting of the cerebrospinal fluid is present only in the quiet RARE image. This may be attributed to an increased sensitivity of the soft-pulsed sequences to flow. The application of long sinusoidal ramps increases the first moment of the readout and slice-selection gradient waveforms compared with short linear ramps, and therefore leads to higher velocity-related phase errors.

It should be mentioned that, unfortunately, the method of soft gradient pulses is not suitable for ultra-fast sequences like snapshot fast low-angle shot (FLASH) (21) or echoplanar imaging (EPI) (22), in which the fundamental of the gradient waveform is in the range of high acoustic response, i.e., above 200–500 Hz. In such a case the reduction of harmonics will have little influence on total noise level and can be more than compensated for by an enhancement of the fundamental. The increase of voltage by a factor of $\pi/2$ needed to perform a sinusoidal gradient ramp instead of a linear one can be another argument against soft pulsing in ultra-fast MRI.

CONCLUSIONS

A significant reduction in the sound generated by standard imaging techniques such as spin-echo or gradient-echo MRI can be achieved by low-pass filtering of the gradient waveforms. This is because the acoustic response function of typical gradient coils is low at low frequencies. We have described a simple method for designing band-limited, soft gradient waveforms based on the use of sinusoidal ramps

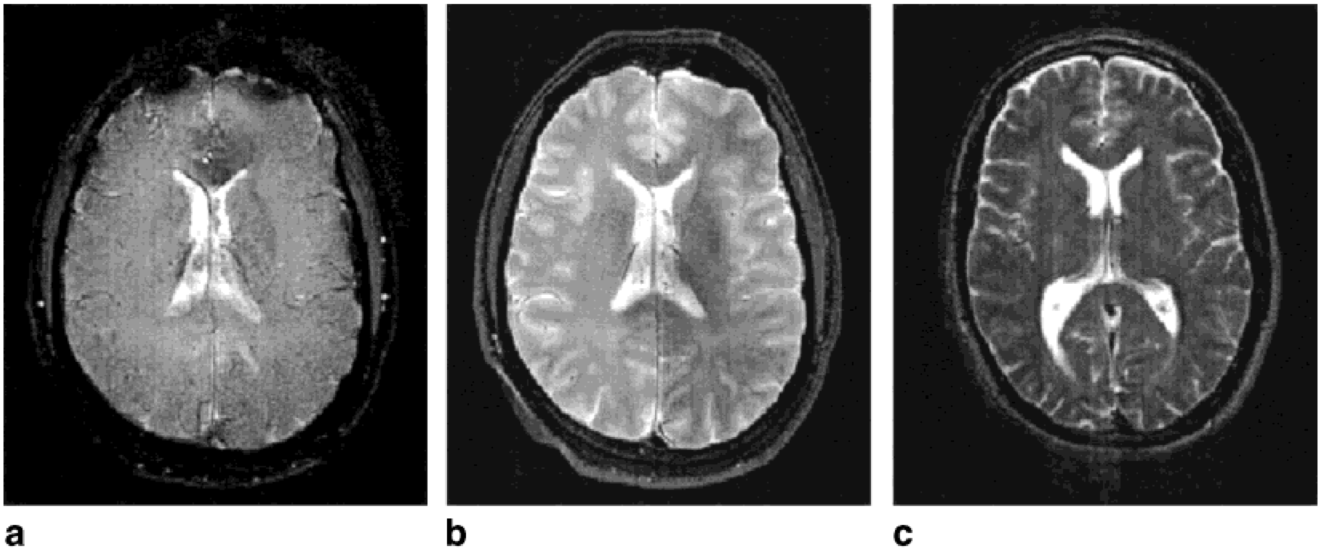

FIG. 6. Single accumulation, 1 mm resolution images of human heads obtained with silent MRI methods. **a:** GE, TR/TE 100/25 msec, $B_0 = 3$ T, noise level 48 dBA. **b:** SE, TR/TE 3000/50 msec, $B_0 = 3$ T, noise level 45 dB(A). **c:** RARE, 8 echoes, TR/TE 3000/120 msec, $B_0 = 2$ T, peak noise level below 60 dB(A).

of maximal duration. A further acoustic optimization of gradient waveforms is certainly possible, e.g., by selective suppression of peak response frequencies. However, even with our simple design, noise levels as low as 40 dB(A) could be achieved with useful imaging parameters at the relatively high magnetic field strength of 3 T. The residual noise at such a level can easily be masked by the background (air conditioning, helium pump), making our sequences practically silent. This opens a way to a variety of applications such as functional imaging of low auditory stimuli or scanning during sleep.

REFERENCES

1. Mansfield P, Glover PM, Beaumont J. Sound generation in coil structures for MRI. Magn Reson Med 1998;39:539–550.
2. Hurwitz R, Lane SR, Bell RA, Brant-Zawadzki MN. Acoustic analysis of gradient-coil noise in MRI. Radiology 1989;173:545–548.
3. Shellock FG, Morisoli SM, Ziarati M. Measurement of acoustic noise during MR imaging: evaluation of six worse-case pulse sequences. Radiology 1994;191:91.
4. McJury M, Blug A, Joerger C, Condon B, Wyper D. Acoustic noise levels during MRI scanning at 1.5 T. Br J Radiol 1994;67:413–415.
5. McJury MJ. Acoustic noise levels generated during high field MRI. Clin Radiol 1995;50:331–334.
6. Shellock FG, Ziarati M, Atkinson D, Chen DY. Determination of gradient magnetic field-induced acoustic noise associated with the use of echo planar and three-dimensional fast spin-echo techniques. J Magn Reson Imaging 1998;8:1154–1157.
7. Quirk ME, Letendre AJ, Ciottone RA, Lingley JF. Anxiety in patients undergoing MR imaging. Radiology 1989;170:463–466.
8. Brunnet RE, Talbot JM, Charuhas P. Potential hearing loss resulting from MR imaging. Radiology 1989;169:539–540.
9. Shah NJ, Jäncke L, Grosse Ruycken ML, Posse S, Müller-Gärtner HW. On the influence of acoustic masking noise in fMRI on the auditory cortex during phonetic discrimination. In: Proceedings of the ISMRM Fifth Meeting, Vancouver, 1997. p 340.
10. Cho ZH, Chung SC, Lim DW, Wong EK. Effects of the acoustic noise of the gradient system on FMRI: a study of auditory, motor, and visual cortices. Magn Reson Med 1998;39:331–335.
11. Bandettini PA, Jesmanowicz A,Van Kylen J, Birn RM, Hyde JS. Functional MRI of brain activation induced by scanner acoustic noise. Magn Reson Med 1998;39:410–416.
12. Golman AM, Grossman WE, Friedlander PC. Reduction of sound levels with anti-noise in MR imaging. Radiology 1989;173:549–550.
13. McJury M, Stewart RW, Crawford D, Toma E. The use of active noise control (ANC) to reduce acoustic noise generated during MRI scanning: some initial results. Magn Reson Imaging 1997;115:319–322.
14. Mansfield P, Chapman BLW, Bowtell R, Glover P, Coxon R, Harvey PR. Active acoustic screening: reduction of noise in gradient coils by Lorentz force balancing. Magn Reson Med 1995;33:276–281.
15. Bowtell R, Mansfield P. Quiet transverse gradient coils: Lorentz force balanced designs using geometrical similitude. Magn Reson Med 1995;34:494–497.
16. Baumgart F, Gaschler B, Tempelmann C, Tegeler C, Stiller D, Heinze HJ, Scheich H. Masking of acoustic stimuli by the gradient noise alters the fMRI detectable activation in a particular human auditory cortex. MAGMA 1996;4(suppl):185.
17. Cho ZH, Chung ST, Chung JY, Park SH, Kim JS, Moon CH, Kong IK. A new silent magnetic resonance imaging using a rotating dc gradient. Magn Reson Med 1998;39:317–321.
18. Hennel F, Girard F, Loenneker T, Alonso A, Macher JP. Silent MRI with soft gradient pulses. MAGMA 1998;6(suppl):32.
19. Hedeen RA, Edelstein WA. Characterization and prediction of gradient acoustic noise in MR imagers. Magn Reson Med 1997;37:7–10.
20. Hennig J, Nauerth A, Friedburg H. RARE imaging: a fast imaging method for clinical MR. Magn Reson Med 1986;3:823–833.
21. Haase A. Snapshot FLASH MRI. Applications to T1, T2, and chemical-shift imaging. Magn Reson Med 1990;13:77–89.
22. Mansfield P. Multi-planar image formation using NMR spin-echoes. J Phys C 1977;10:L55–58.