Interventional MR Imaging at 1.5 T: Quantification of Sound Exposure[1]

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Abbreviations:
GRE = gradient-recalled echo
$L(A)_{eq}$ = equivalent-continuous A-weighted SPL
$L(A)_{eq8h}$ = equivalent-continuous A-weighted daily (8-hour) noise exposure
$L(L)_{eq}$ = equivalent-continuous linear SPL
SPL = sound pressure level

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Sound pressure levels (SPLs) during interventional magnetic resonance (MR) imaging may create an occupational hazard for the interventional radiologist (ie, the potential risk of hearing impairment). Therefore, A-weighted and linear continuous-equivalent SPLs were measured at the entrance of a 1.5-T MR imager during cardiovascular and real-time pulse sequences. The SPLs ranged from 81.5 to 99.3 dB (A-weighted scale), and frequencies were from 1 to 3 kHz. SPLs for the interventional radiologist exceeded a safe SPL of 80 dB (A-weighted scale) for all sequences; therefore, hearing protection is recommended.

Acoustic noise has long been recognized as an important issue in magnetic resonance (MR) imaging because of the potential risk of induced hearing impairment (1–11). The recent advent of interventional MR imaging has potentially created an occupational hazard for radiologists, that is, the acoustic burden on the interventional radiologist who works near the MR magnet bore. The risk for the interventional radiologist is in the accumulation of noise exposure during long and repetitive interventional MR procedures. The risk that hearing loss will develop slowly and imperceptibly over several years as a result of chronic noise exposure at levels less than pain is well documented in the literature (12,13).

Previous emphasis in discussions about the acoustic noise of MR imaging has been on patient exposure during MR imaging. Because of this emphasis, most published noise measurements represent the acoustic noise levels in the isocenter of the magnet bore. Only a few investigators have reported the sound pressure levels (SPLs) at the entrance of the magnet bore (4,8), which are more relevant to assess the acoustic burden on the interventional radiologist. Furthermore, no data are available for the acoustic noise of newer real-time sequences used in interventional MR imaging. The high-performance gradients and fast gradient switching necessary for real-time MR imaging are likely to cause greater acoustic noise levels (3,6,10,14). In addition, there is a growing tendency for use of the higher magnetic field strength of 1.5 T (15,16), with correspondingly increased acoustic noise levels.

The purposes of our study were *(a)* to quantify the SPLs of the imaging sequences that are relevant for interventional MR imaging at 1.5 T for the interventional radiologist and *(b)* to determine the patient's acoustic exposure in interventional MR procedures.

Materials and Methods

Data were obtained with a 1.5-T cardiovascular MR imager (Signa CV/i, with LX 8.4 software; GE Medical Systems, Milwaukee, Wis) with gradient of 40 mT · m^{-1}, slew rate of 150 T · m^{-1} · sec^{-1}, and rise time of 268 μsec, with use of an integrated quadrature-driven transceiver and a radio-frequency body coil. This cardiovascular MR system allows fast imaging with high signal-to-noise ratios suitable for real-time imaging. Fast pulse sequences with rapid data collection and calculation are required to achieve adequate image refresh rates that allow visualization of the anatomy depicted and devices used during an interventional MR procedure; the sequences result in an on-the-fly adaptation of the image formation.

Noise measurements were made with a ½-inch (1.27-cm) prepolarized free-field condenser microphone (type 4189; Brüel & Kjær, Nærum, Denmark) mounted on a tripod, with a 10-m extension cable (AO-0442; Brüel & Kjær) connected to a type 1 digital sound level analyzer (Investigator 2260; Brüel & Kjær) with oscilloscope (PM-3218; Philips) located in the adjacent MR control room. All SPLs were measured during 50 seconds and recorded on both linear and A-weighted scales. The linear SPL, which is expressed in decibels, is the logarithm of the ratio of P_1 (measured in micropascals) to the international standardized reference sound pressure (P_0) of 20 μPa:

$$\mathrm{SPL} = 10 \times \log \left(\frac{P_1}{P_0}\right)^2. \qquad (1)$$

The human ear is not uniformly sensitive to the audible frequencies (decrease of sensitivity to less than 1 and more than 6 kHz). This decreased sensitivity is accounted for by adding an A-weighted filter to the linear (unfiltered) SPL; the filter adjusts for the acoustic response of human hearing. In addition, the peak SPLs (the highest instantaneous sound pressure level in less than 50 μsec, $L(L)_p$) and the frequency distributions on ⅓-octave bands (ie, doublings of 16, 20, and 25 Hz) to 12 kHz were recorded on a linear scale. The time weighting (ie, the time to average the instantaneous fluctuations in sound pressure) was 125 msec (3,10). To estimate hearing damage due to occupational sound exposure, the equivalent-continuous A-weighted SPL, $L(A)_{eq}$, is the preferred measure; it reflects the overall (time-averaged) SPL during the 50-second measurement period. The sound profiles were monitored with an oscilloscope for impulse noise (which is characterized by a sharp increase and rapid decay of SPL in less than 1 second that is more than 10 dB above background SPL in less than 250 msec) (17).

In a pilot experiment, tests were conducted to optimize the experimental set-up for the acoustic measurements (18). The initial SPL measurements showed that the MR imaging suite, with its flat and hard surfaces without noise damping materials, was (in acoustic terms) a diffuse field (ie, the SPL increased less than 3 dB when the distance to the sound source was halved). This finding led us to position the microphone vertically in the experimental set-up (in compliance with American National Standard protocol S1.13-1995 of the Acoustical Society of America) (3,4,17). Because the frequency distribution was well below 20 kHz in our pilot experiment, sound wave diffraction around the microphone was negligible. Therefore, a free-field microphone, which corrects for diffraction in free-field measurements, could be used in the diffuse-field MR imaging suite.

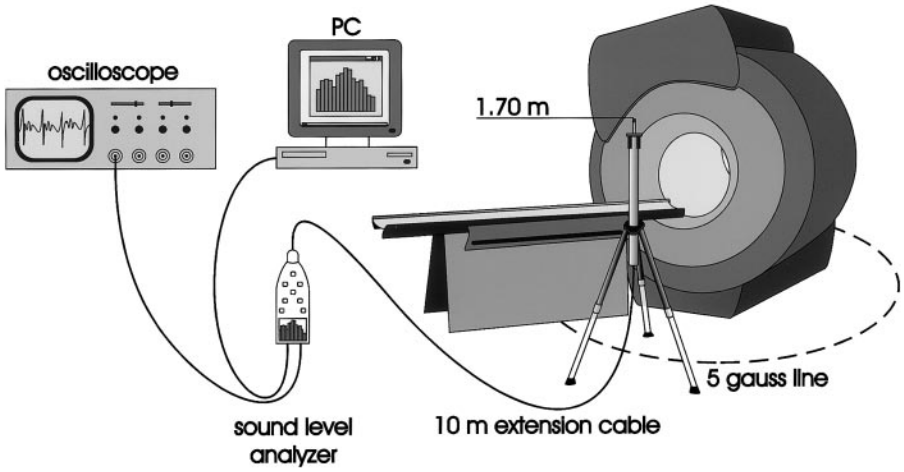

Figure 1. Experimental set-up of noise measurements. *PC* = personal computer.

Findings in previous reports have shown that, despite the presence of some amount of ferromagnetic material (mostly nickel), the accuracy of the microphone is not influenced by gradients and the radio-frequency system (3,4,6,10,19,20). We also ruled out possible interference of the static 1.5-T magnetic field by the sensitivity of the microphone by coupling a reference sound to the microphone with a fixed-wave propagation path (2-m-long plastic tube). Introduction of the microphone into the magnet isocenter resulted in SPLs equal to those measured in the MR control room. The sound level meter calibration was checked, with 94 dB at 1 kHz, at regular intervals throughout the experiments.

To quantify the operator exposure, the microphone was placed 0.8 m from the MR imager (at the 5-G line from the magnet bore) at a height of 1.70 m, which is a plausible location for the ear of the interventional radiologist (Fig 1). These measurements were performed without a person in the magnet bore. To measure patient exposure to noise, the microphone was positioned inside the magnet bore at the isocenter.

The MR imaging sequences to be tested for acoustic noise were chosen on the basis of their relevance to cardiovascular interventional MR imaging: single-shot fast spin echo, fast spoiled gradient-recalled echo (GRE), time-of-flight fast spoiled GRE, fast GRE echo train (a hybrid echo-planar fast spoiled GRE sequence [21]), and spiral trajectory k-space sampling.

Of these sequences, the fast spoiled GRE and fast GRE echo train sequences seem especially suitable for real-time interventional MR imaging. Relevant imaging parameters, including repetition time (n = 74 measurements), echo time (n = 32 measurements), flip angle (n = 84 measurements), field of view (n = 83 measurements), section thickness (n = 81 measurements), matrix size (n = 50 measurements), and plane of imaging (n = 41 measurements), were varied over a wide range for each of the sequences tested. The influence of imaging parameters recorded both inside and outside the magnet bore was evaluated with median values and quartiles of the differences between the recorded SPL and the mean SPL for each sequence. The influence of a person inside the magnet bore on the noise level, with respect to the operator, was assessed at the 5-G line by using different sequences (n = 42 measurements), mainly GRE, both with and without a person in the magnet bore.

The recordings with a volunteer in the bore were not performed during routine procedures. We obtained informed consent from each volunteer. We consulted the chairman of our institutional review board, and he concluded that board approval was not required for the volunteer study. In addition, baseline noise levels in the MR suite were recorded that represented the sound level of the in-room air-conditioning and ventilation systems and the MR cooling cryogen (ambient noise). Ambient noise was negligible during the measurements of the imaging sequences because the resulting SPLs were then much higher than 10 dB (17).

Results

Table 1 lists the equivalent-continuous and peak SPLs on linear and A-weighted

TABLE 1
Typical Imaging Protocols with SPLs Measured at the 5-G Border and Magnet Isocenter

	Imaging Parameters							5-G Border			Isocenter		
Sequence	Repetition Time (msec)	Echo Time (msec)	Echo Train Length	Field of View	Matrix	Section (mm)	No. of Sections per Second	$L(A)_{eq}$	$L(L)_{eq}$	$L(L)^{*}_{p}$	$L(A)_{eq}$	$L(L)_{eq}$	$L(L)_{p}$
Rest	NA	NA	NA	NA	NA	NA	NA	52.3	69.1	82.0	63.8	84.5	96.7
Fast spin echo	200	14	8	25	256 × 256	10	0.15	88.4	88.1	101.8	98.8	98.1	117.5
	1,040	14	8	25	256 × 256	10	0.03	81.5	81.3	101.7	92.1	92.3	115.4
	200	14	8	40	256 × 256	10	0.15	88.4	87.9	101.9	100.1	99.5	118.1
	560	14	20	25	256 × 256	10	0.12	87.9	87.5	102.5	98.7	98.3	114.2
	200	14	8	25	256 × 128	10	0.31	88.2	87.8	102.3	100.5	99.8	118.0
Single-shot fast spin echo		42	NA	25	256 × 256	10	0.31	86.6	86.1	102.6	97.4	97.0	116.3
Fast spoiled GRE	4.4	1.3	NA	25	256 × 96	4	2.41	96.6	96.3	108.4	109.2	108.1	118.9
	4.4	1.3	NA	40	256 × 96	4	2.41	90.5	90.4	101.8	104.4	104.4	118.6
	4.6	1.3	NA	25	256 × 192	4	1.19	98.1	97.8	109.3	110.4	109.4	121.9
	4.4	1.3	NA	25	256 × 96	8	2.41	96.8	96.5	106.6	108.5	107.4	118.7
Three-dimensional time of flight	5.3	1.0	NA	25	256 × 192	4	0.97	96.2	96.2	107.9	106.2	105.3	118.9
Fast GRE echo train	10.2	1.9	4	25	256 × 256	8	1.10	99.3	99.0	110.0	109.7	108.6	121.1
	9.6	1.7	4	40	256 × 256	8	1.16	93.4	92.9	106.3	105.9	105.1	120.3
	12.0	3.1	4	30	256 × 256	2	0.98	96.1	96.4	107.1	105.2	104.4	118.7
	17.0	1.7	8	30	256 × 256	8	1.25	95.1	95.1	107.3	104.2	103.5	117.6
	40.0	1.8	4	30	256 × 256	8	0.36	90.0	90.1	102.9	99.3	98.7	116.6
Spiral trajectory k-space sampling	22	2.3	NA	25	2,048/20*	6	2.22	94.9	95.2	107.3	103.1	102.4	121.0
	20	2.2	NA	25	2,048/20*	8	2.21	94.9	95.2	107.4	103.0	102.4	121.0
	22	2.5	NA	25	2,048/10*	6	3.60	94.3	94.4	105.4	103.7	103.2	120.4

Note.—NA = not applicable. $L(L)_p$ = highest instantaneous SPL in less than 50 μsec.
* Number of data collection points per spiral per number of spirals in *k* space.

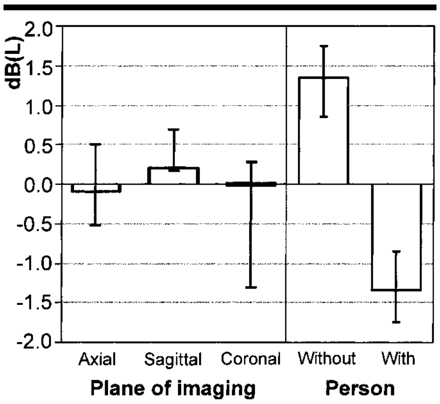

Figure 2. Bar graphs of the plane of imaging (n = 41 measurements) and the presence of a person (n = 44 measurements) depict median (height of box) and 25th and 75th quartiles of the differences between measured SPLs and their mean SPL for each sequence. Different planes of imaging had identical SPLs. The presence of a person inside the magnet bore caused a decrease in SPL of about 3 dB.

scales. Of these measures, $L(A)_{eq}$ is considered a reliable predictor of noise-induced hearing loss for all types of noise (ie, continuous and impulse noise) (22).

The ambient noise in the MR imaging suite at rest (measured at the 5-G line) was remarkably high, with equivalent-continuous linear SPL, $L(L)_{eq}$, of 69.1 dB. Because the main frequency of the ventilation system was at about 100 Hz, however, the more relevant $L(A)_{eq}$ was considerably lower (52.3 dB). These results are similar to previously reported values (3,5,23).

The $L(A)_{eq}$ values depended on the sequence (Table 1) and ranged from 81.5 to 99.3 dB (A-weighted scale) at the 5-gauss line. The fast sequences, fast spoiled GRE and fast GRE echo train, had $L(A)_{eq}$ values as high as 98.1 and 99.3 dB (A-weighted scale), respectively. The peak SPLs were greater than 100 dB for all sequences (range, 101.7–110.0 dB [linear scale]). On average, the noise exposure to the patient was 11 dB greater than that to the interventional radiologist. The presence of a person inside the magnet caused a noticeable decrease in SPL for the operator of 2.7 dB (Fig 2). In audiophysics, 3 dB is generally taken as the transition between irrelevant and relevant differences.

On the basis of our analysis, it appears that in all sequences tested, the main parameter influencing the SPL was repetition time. With all other variables unchanged, a doubling or quadrupling of repetition time resulted in a decrease in SPL of about 3 and 6 dB, respectively (Fig 3). The fast spin-echo sequence (200/14.3 [repetition time msec/echo time msec]) generated 88.1 dB (linear scale), while the same sequence with 1,040/14.3 produced 81.3 dB (Table 1). On the other hand, changing the echo time (Fig 4), flip angle (Fig 3), section thickness (Fig 5), and matrix (Fig 4) had no noticeable effect on SPL. On average, an increase in the field of view resulted in a small reduction in SPL of less than 1 dB (Fig 5). For sequences with very short repetition time (<15 msec), as in fast spoiled GRE and fast GRE echo train, the SPL decreased remarkably when the field of view was enlarged (Table 1); an increase in repetition time resulted in a reduced influence of the field of view on the measured SPL. Variation of the orientation and positioning of the imaging plane seemed not to have an influence on SPL: axial, sagittal, and coronal planes had identical SPLs (Fig 2), as did imaging planes that were

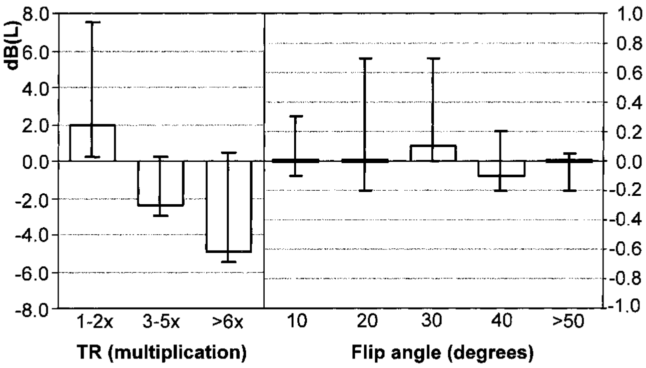

Figure 3. Bar graphs of multiplications of repetition time (*TR* multiplications) (n = 74 measurements) and flip angle (n = 84 measurements) depict median (height of box) and 25th and 75th quartiles of the differences between measured linear SPLs and their mean SPL for each sequence. SPL decreased with increasing repetition time. The flip angle did not influence the SPL.

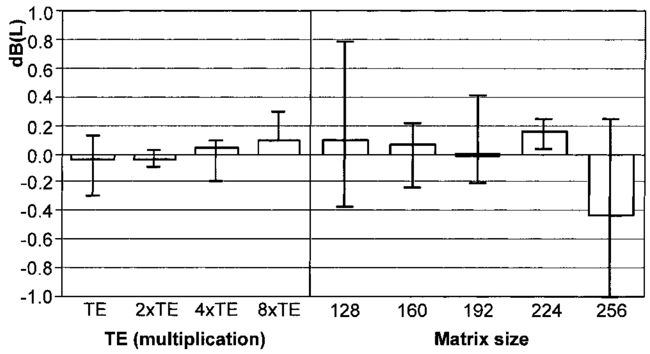

Figure 4. Bar graphs of multiplications of echo time (*TE* multiplications) (n = 32 measurements) and matrix size (in pixels) (n = 50 measurements) depict median (height of box) and 25th and 75th quartiles of the differences between measured SPLs and their mean SPL for each sequence. Neither echo time nor matrix size influence SPL.

translated in cranial or caudal direction along the z axis (not shown).

All but one sequence had a frequency distribution ranging from 1 to 3 kHz, with a distinctive peak around the 2-kHz octave band when measured inside the MR imager. The exception was the fast GRE echo train sequence, which had frequencies ranging from 2 to 5 kHz (Fig 6). The higher frequencies in the fast GRE echo train sequence are probably caused by the increased slew rates of the gradients (14). A comparison of measurements inside the magnet bore and at the 5-gauss line showed attenuation of frequencies higher than 2 kHz. Therefore, in the frequency distribution, maximum SPLs were between 800 and 1,600 Hz at the 5-gauss line. Because these frequencies are precisely within the frequency range that is important for speech (0.5–2.0 kHz) (9), hearing loss due to gradient noise exposure will primarily affect the frequencies that are used in speech, followed by dissemination into neighbouring frequencies. Moreover, speech-to-noise ratios will drastically decrease during the interventional procedure and reduce the intelligibility of speech.

Analysis of the noise profile with an oscilloscope did not reveal impulse features but rather a quite complex disordered profile (not shown). This profile probably reflects the higher harmonics of the noise generated in the coil supports (23).

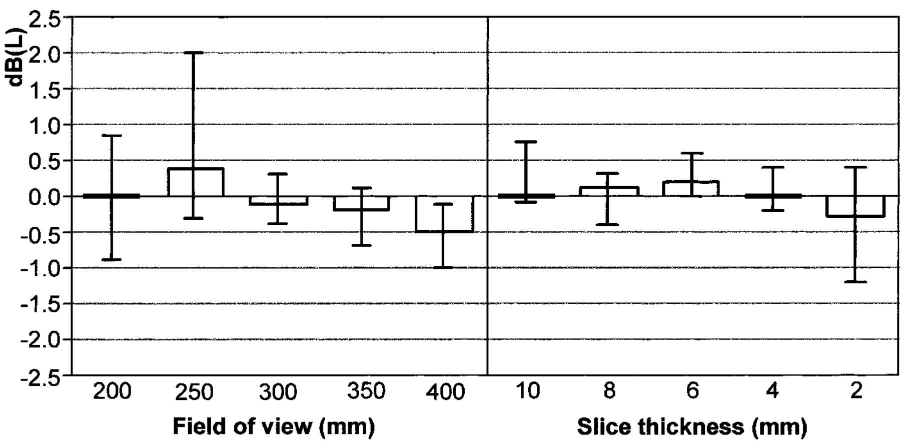

Figure 5. Bar graphs of field of view (n = 83 measurements) and section thickness (n = 81 measurements) depict median (height of box) and 25th and 75th quartiles of the differences between measured linear SPLs and their mean SPL for each sequence. The field of view and section thickness did not influence the SPL, although a decreasing trend in SPL with increasing field of view was seen.

Discussion

Our measurements show that the acoustic burden on the interventional radiologist is of great magnitude. $L(A)_{eq}$ values as high as 99 dB (A-weighted scale) were common with the MR imaging sequences likely to be used for real-time imaging during interventions (fast GRE echo train and fast spoiled GRE). These values were measured at approximately 80 cm from the magnet bore entrance, but they will be higher whenever the operator works more closely to the patient in the magnet during the actual intervention. The results also show a small effect of a 2.7-dB reduction in SPL when a person is lying inside the magnet; this reduction is probably caused by attenuation of frequencies above 2 kHz (not shown). In previous experiments, authors found that, when measured inside the magnet bore, the SPLs are about 3 dB higher whenever a person was inside the magnet; a tentative explanation given by these authors is that in-phase reflections inside the magnet bore could cause a doubling of pressure and a subsequent increase in SPL (19).

In additional analysis, the effects of various MR parameters on the acoustic noise were assessed. It appeared that repetition time was most relevant in this respect. The image acquisition time is covered by a series of repetition times, each with an equal number of gradient pulses. Therefore, as is expected, a doubling or quadrupling of the number of encoding steps per unit time (ie, the amount of acoustic energy per unit time) caused by shortening repetition time, resulted in about a 3- and 6-dB increase in SPL, respectively. The field of view

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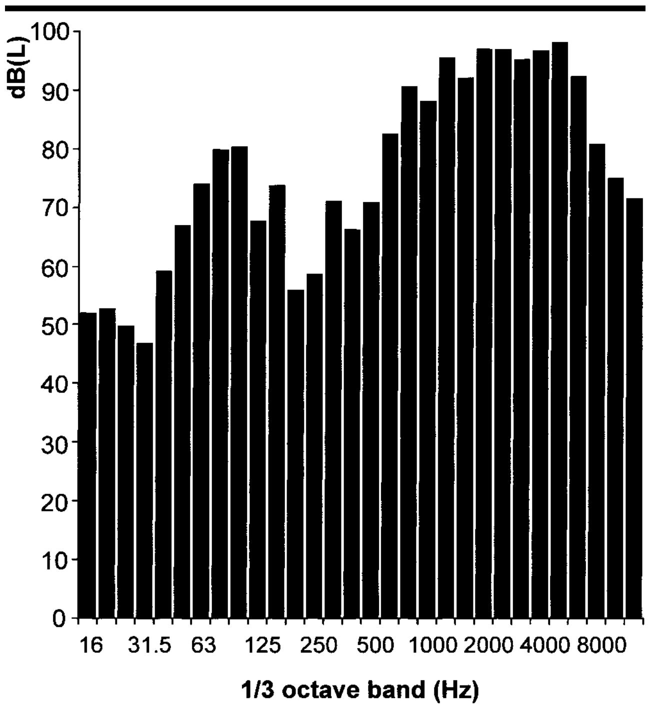

Figure 6. Frequency spectrum of the ⅓–octave band of a fast GRE echo train sequence (isocenter). High SPLs with frequencies between 1,000 and 5,000 Hz exceed ambient noise levels (y axis) with frequencies around 100 Hz.

TABLE 2
The 3-dB Trading Rule for Maximum $L(A)_{eq8h}$

SPL	Exposure Time	Common Sound with Comparable SPL
90	8 h	Motorcycle at 10 m
93	4 h	
96	2 h	Subway (inside)
99	1 h	
102	30 min	Diesel truck at 10 m
105	15 min	
108	7 min	Power mower at 1 m

proved to be less important in this respect. The field of view is inversely proportional to the gradient strength of either the readout or phase-encoding gradient. Therefore, the relationship between field of view and SPL resulted in a small audiologically irrelevant (<1-dB) decrease in SPL, especially with the fast GRE echo train and fast spoiled GRE sequences.

In contrast to findings in previous reports (3), our results did not show an effect of changing echo time. An inverse influence of echo time on SPL has been suggested, but this suggestion was based on an observation of combined simultaneous lengthening of both repetition time and echo time (3). However, the echo time changes the timing of the gradient pulses within repetition time and should theoretically not influence the SPL. In terms of sound production, the section-select gradient is less important than are the readout and phase-encoding gradients; therefore, section thickness plays a minor role in the total loudness of generated noise. Authors of previous reports, however, suggest that section thickness has more effect on SPL (3,5,6), probably owing to the larger section-select gradient amplitude relative to readout and phase-encoding gradient amplitudes. Our results did not show an effect of changing the flip angle, which could be expected because radio-frequency pulses are short in contrast to the length of encoding gradients. In pulse sequences with multiple radio-frequency pulses (eg, burst imaging), the flip angle may influence SPL to a greater extent.

Permanent hearing loss may occur as a result of chronic exposure to noise at levels greater than 80 dB (A-weighted scale) (22). Safety guidelines have been established for industry workers to limit the maximum (daily) noise exposure, on the basis of an 8-hour working day for 5 days a week (Appendix). The main rationale for these safety guidelines is to preserve hearing for speech discrimination (12,13, 24).

According to the European Community guidelines, the maximum equivalent-continuous A-weighted daily (8-hour) noise exposure, $L(A)_{eq8h}$, should not exceed 90 dB without hearing protection (25). The SPLs for all but one sequence were well above this permissible noise pressure level. A so-called 3-dB trading rule (or equal-energy rule) applies: An increase in SPL of 3 dB will halve the permitted exposure time (Table 2). Thus, noise exposure at 102 dB (A-weighted scale) during an interventional MR procedure is permitted for only 30 minutes a day. In the United States, the Occupational Safety and Health Administration, or OSHA, has recommended an $L(A)_{eq8h}$ value of 90 dB with a 5-dB exchange rate (24), although the National Institute for Occupational Safety and Health, or NIOSH, advises maximum $L(A)_{eq8h}$ of 85 dB (A-weighted scale) with a 3-dB exchange rate (22). Many different nongovernmental U.S. employment sectors have adopted the more prudent NIOSH guidelines.

In European Community countries and the United States, hearing-protection equipment should be provided to employees exposed to $L(A)_{eq8h}$ of greater than 85 dB (A-weighted scale). (In several European countries, $L(A)_{eq8h}$ is set even lower, at 80 dB.) The use of such equipment is mandatory for $L(A)_{eq8h}$ of greater than 90 dB (A-weighted scale) (1,11,24). Good hearing protection for SPLs as high as 110 dB (A-weighted scale) can be achieved by using universal passive earplugs, custom molded earplugs, or earmuffs. Universal passive earplugs have increased attenuation for frequencies above 1 kHz (5,7), with a 35-dB decrease in air- conducted SPLs at a relevant frequency component of around 2 kHz for interventional MR imaging (10). The combination of passive earplugs and earmuffs could achieve greater than 40-dB sound attenuation for frequencies below 2 kHz (26).

An issue that has been raised with passive hearing-protection aids is possible interference with communication (1,3, 11). As has been recently shown, however, passive hearing protection actually improves speech intelligibility for people with normal hearing in acoustic environmental noise (26). In contrast, passive hearing protection has a negative effect on speech intelligibility for hearing-impaired listeners (27). Hearing-protection aids, which encompass nonlinear acoustic filters or built-in noise reduction systems, may allow better communication while still providing adequate protection from acoustic noise. However, selective filtering or suppression may be complicated by similar frequency distributions of both speech and gradient noise. A 10–30-dB reduction in SPLs has been achieved with active noise reduction systems in MR imaging (2,7,28).

Ultimately, noise reduction should be achieved at the source (ie, the design of the MR gradient system and supports) (7,10,20). Recently, more quiet MR systems have become commercially available with vacuum-enclosed gradient systems in addition to insulators (Excelart, Toshiba, Tochigi, Japan; Signa Twinspeed, GE Medical Systems). Vacuum enclosures provide greater than 15-dB noise reduction (for the Excelart system) (29). Additional noise reduction by means of vacuum enclosure may be restricted, however, by the requirement for an adequate gradient cooling system. The application of passive noise-reducing materials is also limited because it counterbalances the dimensions of the magnet bore.

There is a growing trend for use of MR systems with high field strength for interventional MR procedures. Such systems provide better homogeneity and stability of the main magnetic field, higher signal-to-noise ratios and resolution, and faster imaging (15,16). However, many interventional procedures are currently preformed with less than 1.0-T MR imagers. Therefore, our results may not be directly applied to interventional MR imaging at lower field strengths. Some conclusions, however, can be derived with cautious extrapolation of our results because there is a linear relationship between field strength and SPL (29). Similar sequences at 1.5 and 0.7 T, for example, will generally differ by only 3 dB (less noise at 0.7 T).

In conclusion, interventional MR imaging at 1.5 T is noisy and may be a likely cause of hearing loss for the interventional radiologist if no hearing protection is used during procedures. SPLs outside the magnet bore exceed the safety limits for chronic noise exposure during interventional MR imaging, from both the scientific (12, 13) and judicial perspectives that are valid in the European Community (25) and United States (24). Interventional radiologists should be aware of this occupational hazard. They should use adequate hearing protection such as earplugs and earmuffs, because noise-induced hearing loss is virtually totally preventable by avoiding excessive SPLs. Likewise, hearing-protection equipment should be provided to the patient undergoing the interventional MR procedure. As with industrial workers, we believe interventional radiologists who are to perform interventional MR procedures on a regular basis should undergo baseline audiography. Active audiologic screening of interventional radiologists who perform interventional MR imaging at regular recurrent intervals may be considered.

APPENDIX

This appendix provides details about the concept of cumulative operator exposure during an interventional MR procedure. To estimate hearing impairment and risk of hearing handicap as a result of exposure to noise, the noise exposure level is normalized to a nominal 8-hour working day, $L(A)_{eq8h}$, which can be calculated from SPL measurements and exposure time. Calculations of the daily cumulative exposure are possible for interruptions and changes in SPL with use of the following functions (10,17,30).

$$L(A)_{eq8h} = L(A)_m - 10 \log \left(\frac{\mathrm{T_r}}{\mathrm{T_m}}\right) \quad (2)$$

is used to calculate daily noise exposure for one equivalent-continuous sound level $L(A)_m$, where T_r is 28,800 seconds (8 hours) and T_m is the duration of noise exposure in seconds.

$$L(A)_c = 10 \log \left(\sum_{i=1}^{n} 10^{0.1 \cdot L(A)_{eq8h},i}\right) \quad (3)$$

is used to obtain a combined daily noise exposure $L(A)_c$, in which $L(A)_{eq8h,i}$ equals n number of equivalent-continuous daily sound exposures, $L(A)_{eq8h}$. $L(A)_c$ is used to estimate the risk of hearing loss (Table A1) and should not exceed 90 dB (A-weighted scale). The equal-energy rule can be deduced from reciprocal use of Equation (2): The halving of exposure time T_m results in a 3-dB (10log2) decrease in SPL.

TABLE A1
Median Permanent Noise-induced Threshold Shifts in Hearing Levels (in decibels) across Averaged Audiometric Test Frequencies of 1–4 kHz

$L(A)_{eq8h}$ (dB)	Exposure Time (y) 10	20	30	40
85	2	3	3	4
90	5	7	8	8
95	11	14	16	17
100	18	23	26	29

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