INVITED REVIEW

New fMRI methods for hearing and speech

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Abstract: In this short review, we outline the challenges presented by the hostile environment for auditory fMRI. There are two main elements to this: the high level magnetic fields and the intense sound generated by the scanner operation. We then describe a series of techniques that have been employed to minimize the effects of the scanner noise on the subject and upon the subject's evoked responses to sound stimuli. These include scanner designs, passive attenuation, use of specifically designed scanning sequences, specially designed sound delivery systems and active noise control. Finally, we briefly describe the use of some of these in combination to allow measurement of sound activation of different brain areas non-invasively using fMRI.

Keywords: Magnetic resonance imaging, Auditory, Sound system

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1. INTRODUCTION

Magnetic resonance (MR) scanners have become a routine diagnostic tool providing high-resolution anatomical images of the body. The technique is non-invasive and unlike some imaging methods does not involve exposure to ionizing radiation. The MR scanner generates three different types of electromagnetic fields. For 3 Tesla MR systems, a huge static field is generated by a superconducting magnet. The force is 60,000 times stronger than the earth's magnetic field. Weaker magnetic fields include time-varying gradient magnetic fields and pulsed radiofrequency (RF) fields. MR imaging depends most commonly upon the magnetic properties of hydrogen ions which, when placed in a static magnetic field, can absorb radiowave energy of a specific frequency. The time taken for the ion alignments to return to equilibrium differs according to the surrounding tissue, thus providing the image contrast. Functional brain activation can be assessed by a variety of techniques that rely on two factors (1) the blood supply to active parts of the brain is higher than to inactive parts and (2) the paramagnetic properties of oxy- and deoxy-haemoglobin in the blood differ. The signal detected during functional magnetic resonance imaging (fMRI) is known as the blood-oxygen-level-dependent (BOLD) response. The scanning sequence used to measure the BOLD signal produces very high levels of sound. This background noise is primarily caused by flexing movements of the gradient coils in the MR scanner. During the scanning sequence, large currents (hundreds of amperes) of electricity are passed through three gradient coils of wire to create time-varying magnetic

fields in order to select and encode the 3-dimensional image volume (in the x , y and z planes). Interactions between the gradient coils and the static magnetic field induce pulsed Lorentz forces which act upon the gradient coils to deform them. The gradient coils then act much like a moving coil loudspeaker to produce sound.

The hostile scanner environment creates unique problems for auditory fMRI research. In the first place, the static and time-varying magnetic fields preclude the use of many types of electronic sound presentation equipment. Second, the high sound levels can potentially cause hearing impairment. Third, the scanner sounds mask the presentation of speech and other sorts of acoustic stimuli and fourth, the scanner sounds not only activate parts of the auditory brain, but also interact with the patterns of activity evoked by experimental stimuli e.g. [1]. In this short review we discuss these challenges and some methods to alleviate them. Finally, we review some of the insights into the neural substrate of acoustic analysis and speech comprehension that are emerging from recent fMRI studies.

2. RESTRICTIONS ON EQUIPMENT

No ferromagnetic components can be placed in the scanner bore as they would experience a strong attraction, by the static magnetic field, causing damage. Magnetisable elements also disrupt the magnetic fields locally and induce signal loss or spatial distortion in the image. This is more of a restriction than it may first seem, as many plastic composites (even ear-defenders) contain enough metal to cause visible deterioration in image quality. Image quality can also be affected by electromagnetic interference

generated by electronic equipment and detected by the RF receiver coil. Induced currents in the electronics, caused directly by the time-varying gradient magnetic fields or the RF pulses, are an additional hazard to the electronic devices themselves, while some materials can also absorb the RF energy causing local tissue heating and even burns if the material touches the skin. For these reasons, there are restrictions on scanning people who have cochlear implants fitted and who wear hearing aids.

3. PROPERTIES OF THE SCANNER SOUND

The acoustic properties of the scanner noise (e.g., level, bandwidth, fundamental frequency, spectral envelope) depend on the mechanical resonances of the coil assemblies, on the type of imaging sequence used, and on its switching frequency. Sequences commonly used in neuroscience research exceed 100 dB SPL. The most common technique for imaging brain function is Echo Planar Imaging (EPI). EPI scans capture a brain scan in a very short acquisition time. The acoustic spectrum of EPI sequences is tonal in nature with energy generally concentrated on a single frequency (normally between 500 Hz and 2,000 Hz) with odd and even harmonics of that frequency also present. An example of the EPI acoustic waveform and frequency spectrum is shown in Fig. 1. In this example, the waveform has a large amplitude for each of the 16 brain slices that comprise the scan and the spectrum has a peak component at 650 Hz with several other prominent harmonics of this frequency. The sound level at the peak (fundamental) frequency can be as loud as 130 dB SPL [2,3]. Clearly, exposure to such high sound levels is damaging.

Fig. 1 Acoustic spectrum and (inset) time waveform of a 16 slice EPI scan with a fundamental frequency of 650 Hz.

4. SOLUTIONS FOR PRESENTING AND MONITORING SOUNDS

In this section, we consider ways of delivering and monitoring sounds in the MR scanner. For fMRI studies of auditory brain activation, a system is required to produce a range of sound levels (up to 100 dB SPL), with low distortion, a flat frequency response and a smooth and predictable phase response. The phase response is important for some binaural experiments where inter-aural timing or phase differences are to be studied e.g. [4]. The requirements for sound monitoring are a small microphone with a flat frequency response and a dynamic range sufficient to work linearly in the very high-level local sound field. It is possible for the microphone to pick-up voltages caused directly by the time-varying gradient magnetic fields or the RF pulse. In our experience, electrical pick-up can be a particular problem if the microphone is placed too close to the RF coil. A microphone for subject communication can be placed further away from the RF coil than a microphone at the ear for measuring the scanner sound.

4.1. Sound Delivery

The simplest form of sound delivery system that meets many of the requirements for fMRI is the tube phone. This system consists of a set of ear defenders with a plastic tube that is connected through the sound-attenuating body of the defenders into the ear cavity. Sound is generated by a loudspeaker at the remote end of the tube and travels through the tube to the subject's ear. Clearly only plastic tubes need be within the proximity of the scanner and hence problems with the magnetic field, the RF interference or the image quality are avoided. The ear defender provides attenuation of the scanner sound reaching the subjects' ears. This system is commercially available and quite widely used. However, performance is compromised in three ways. First, the acoustic spectrum at the ear is not flat, but has peaks and troughs and there is a general reduction in sound amplitude at the higher frequencies. This distortion is caused by the effect of the tubes on the sound propagation. Second, the phase response of the system is also compromised and this is important for when considering the use of the headphones for binaural presentation and also for Active Noise Control. The third problem with the tube phone system is the leak of the scanner sound through the pipe walls to the pipe inner and hence the ear. The placement of the pipe inside the scanner is important in this respect.

Despite the restriction on the materials that can be used in a scanner, there are active headphone driving units that can be made to operate in the MR environment without disrupting the imaging process. An ingenious system has been developed (and subsequently marketed) by the

Fig. 2 Diagram showing the construction of the Active Noise Control Headphones used for the presentation of sound to subjects in an MR scanner.

imaging group in Magdeburg, Germany [5]. This system uses the scanner's static magnetic field in place of the permanent magnets that are found in conventional headphones and loudspeakers. This system is relatively inexpensive and represents an almost turn-key solution producing high-level, high-quality sound.

Piezo-electric elements have also been designed to deliver sounds inside a scanner. These use the Piezo electric effect to turn an applied electric voltage into movement. Their use is not widespread because the audio performance is not generally as good other solutions. In a sound system developed specifically for auditory research, we use electrostatic headphone drivers [6]. The electrostatic headphones have two or three plates of very thin insulating material covered with a conducting film only microns thick. High voltages (in the order of 600 V) are then applied to these plates which causes the plates to move due to the induced electrostatic forces between them. Headphones based on this principle are marketed for the consumer audiophile and although expensive are an almost ready-made solution to the challenges of auditory fMRI. Although the electrostatic drivers work on dangerously high voltages, the level of current used is very low (in the order of microamps). Thus it is possible to use resistive carbon fibre leads to connect the electrostatic drivers to the amplifier. These leads are relatively transparent to the magnetic fields in the scanner and so do not affect image quality.

The audio quality of these latter solutions is much better than tube phones. Attenuation of the external sound

by 40 dB (in the range 2–10 kHz) is achieved by mounting the headphones in commercially-available ear defenders (see Fig. 2 for construction details). This construction results in a good frequency response and a predictable phase response since the transmission path of the sound to the ear is only about 1–2 centimeters. These systems can produce a level of sound in the region of 110 dB SPL which is sufficient for most experimental situations.

4.2. Sound Measurement, Monitoring and Subject Communication

It is important to quantify the sound level at subjects' ears because there are legal limits on noise exposure (see [2]). A microphone placed under the ear defender can be used to ensure that the sound attenuation measures are working correctly and that the noise exposure limits are not exceeded [2]. Close proximity to the ear also allows for Active Noise Control (ANC) (see Sect. 5.4). Tiny electret hearing-aid microphones are available with good sensitivity and frequency response which, at first, seem ideal for use in an MR scanner. However, these are typically steel-cased, low-voltage devices that are connected to the monitoring electronics via copper cables, thus making them non-MR-compatible. The best microphone solution for fMRI at present is the optical microphone. This device works by bouncing light off a movable reflective diaphragm. Light is supplied by one optic fibre and the modulated output is returned by another (see [7] for a detailed description). This solution contains no metal and

Fig. 3 Acoustic waveform for a male speaking an normal conversation level in the scanner in quiet and then repeating the same sentence during EPI scanning. Note that the speech signal is obscured by the time-varying bursts of scanner sound. Without filtering, speech intelligibility is severely degraded.

causes no image artifacts. The microphone is much bigger than the hearing-aid microphones, but it is still possible to build them into the sound-presentation headphones.

For the purpose of subject-to-experimenter communication, the microphone may be located at a greater distance from the subject. Although distance somewhat reduces the audio quality, it ensures that the microphone does not interfere with the MR imaging.

Understanding the subject's speech whilst the scanner is in operation may require additional processing of the microphone signal (see Fig. 3). The simplest method is to use a notch filter to remove the fundamental and harmonics of the scanner sound. However, the spectrum of the EPI sound overlaps the human speech spectrum, so simple filtration is ineffective in removing the EPI sound without also degrading the speech signal. An adaptive spectral subtraction algorithm has been developed to overcome this problem. This relies on two principles, first that the frequency spectrum of the recorded signal is the sum of a vocal response spectrum and a noise spectrum, and second that the frequency spectrum of the noise is constant, while only its amplitude varies. The entire spectrum of the noise can be removed, while leaving the speech signal essentially intact [8]. Adaptive filter algorithms (e.g. ANC, see Sect. 5.4) can also reduce the level of scanner sound on the microphone signal, although none have yet been implemented in this context.

5. METHODS FOR MINIMIZING THE EFFECTS OF SCANNER SOUND

A number of ways to reduce the effect of the intense background noise have been implemented. These methods fall into a number of different classes: i) reducing the sound at source, ii) passive sound attenuation, iii) software modifications to the pulse sequence and iv) new methods for stimulus delivery.

5.1. Reducing the Sound at Source

Manufacturers have attempted to minimize scanner sound levels by modifying scanner design. Novel designs

for the MR system hardware include the construction of "quiet" gradient coils in which the net Lorentz force is compensated between current pathways [9,10] and a vacuum-based acoustic noise reduction system in which the gradient coils are encased in an evacuated chamber and mounted directly on the floor. While the latter system can significantly attenuate the scanner sound (by 20 dB) [3], it has not been a commercial success due to gradient coil heating. A simple method used to reduce scanner sound is to line the bore with acoustic foam. This reduces echoes directed back toward the subject from the inner circular section of the scanner bore. We have demonstrated that foam reduced the sound levels measured at the position of the subject's head by about 10 dB [2]. However, it is not generally practicable to line the bore of commercial MR systems.

5.2. Passive Sound Attenuation

It is common practice for fMRI subjects to wear some form of ear protection. Ear plugs can compromise the acoustic quality of the experimental sounds delivered to the subject and so ear defenders are preferable. Several considerations concern the limited space available inside the RF coil for the wearer and the MR compatibility of the materials from which the ear defenders are made. Good quality ear defenders produce up to about 40 dB of attenuation of airborne sound at frequencies > 1 kHz, but are less efficient at lower frequencies.

5.3. Software Modifications to the Pulse Sequence

One software modification reduces acoustic noise by slowing down the gradient switching in the scanning sequence. This approach is based on the premise that the spectrum of the acoustic noise is determined by the product of the frequency spectrum of the gradient waveforms and the frequency response function of the gradient system [11]. The frequency response function is generally substantially reduced at low frequencies (below 200 Hz) and so the sound level can be reduced by using gradient pulse sequences whose spectra are band-limited to this low-frequency range using ramped pulse shapes [12]. Using a low-noise FLASH sequence, a peak scanner sound level of 58 dB SPL, at the position of the ear, has been used for neuroimaging of central auditory function [13,14]. However, the low noise is achieved at the expense of slower gradient switching, extending the acquisition time. Low-noise sequences are not suitable for EPI in which the fundamental of the gradient waveform is greater than 200 Hz.

Another sequence method exploits the delay of the BOLD signal [15] to minimise the effect of the scanner sound on the measured patterns of auditory cortical activation during EPI. This class of techniques is generally

Sparse acquisition

Continuous acquisition

Fig. 4 Schematic diagram of sparse and continuous acquisition protocols, for a simple fMRI experiment in which stimuli are presented during the 9 s periods (shown in grey). Scan onsets are denoted by the arrows. In sparse acquisition, an EPI scan is acquired once every 9 s at the end of the stimulation period and the slices within the scan are acquired as close together in time as possible (clustered acquisition). In continuous acquisition, several EPI scans are acquired during the stimulation period (every 3 s in this example). The estimated auditory BOLD response to the experimental auditory stimulation is shown by the dashed lines and the estimated response to the scanner sound is shown by the dotted lines. For sparse acquisition, scan onset occurs at the maximal point of the experimental BOLD signal and the minimal point of the noise BOLD signals.

known as clustered-volume or sparse acquisition [16]. The essence is to present an auditory signal (producing auditory activation) and after at least 6 s to acquire a brain scan as rapidly as possible (acquisition time < 2 s) (Fig. 4). Thus, the activation associated with the experimental sound can be separated from the activation associated with the scanner sound. In addition, since the scanner sound is temporally offset, it does not produce simultaneous masking and does not distract the listener. One drawback is that the number of brain images that can be captured per unit time is reduced compared to continuous acquisition protocols, but see Schwarzbauer [17] for an alternative method which acquires a set of images following each period of quiet.

5.4. New Methods for Stimulus Delivery (Active Noise Control)

ANC is a method for canceling the airborne sound of the scanner at the subjects' ears, thus attenuating its level for the listener. ANC relies on the precise superposition of sound waves. If two sound waves of exactly the same frequency and amplitude arrive at the same point in space 180 degrees out of phase, then they will destructively interfere producing no sound at all at that point. However, any amplitude or phase mismatch will cause a residual

sound. Simple calculations show that a 3 degree mismatch in phase between two equal amplitude sine waves will leave a residual sound only 25.6 dB smaller than the original amplitude. Similarly, an amplitude mismatch of only 0.1 dB will leave a residual amplitude 38.7 dB below the original amplitude. These examples show how critical is the matching of the control to the noise signal: the precise amplitude and phase relationships must be maintained at all frequencies at which noise control is to take place, typically from 100 Hz to over 3,500 Hz. It is fortuitous, in this context, that the MR scanner sound is precisely replicable on each EPI burst (see Fig. 1), which means that one burst can be used as the control signal to cancel a subsequent burst.

The basic method for ANC, the Least Mean Squares (LMS) adaptive filter, was devised by Windrow [18] (see also [19] for a complete and mathematically rigorous description). This method adjusts the characteristics of a digital filter in order to minimize the component of the error signal that is correlated with its input. Figure 5 shows a basic LMS system for the removal of scanner sound from a subject microphone signal. The system input $p(n)$ (in this case the sound from a microphone with just the MR scanner sound) is input to the adaptive filter $W(z)$. The output of the filter $y(n)$ is mixed with another input $m(n)$ derived from a microphone near the subject. The difference between the two signals appears at the output of the mixer as an error signal $e(n)$. This signal is used to adjust the characteristics of $W(z)$ such that the amplitude of the error signal $e(n)$ is minimised. The filter $W(z)$ will adjust the amplitude and phase of each component of $p(n)$ so that it cancels with the corresponding component of $m(n)$. The repeatability of the MR scanner sound enables a recording of the sound in one EPI burst of scanning to be used as $m(n)$, the reference for the adaptive system. Alternatively $m(n)$ can be provided by a directional microphone placed

Fig. 5 The block diagram of a LMS adaptive filter system for the attenuation of scanner noise from a subject microphone.

Fig. 6 The block diagram of the LMS adaptive filter system used to implement active noise control.

orthogonally to the subject microphone.

Figure 6 shows how the adaptive filter can be reconfigured for use as an ANC system for the reduction of sound inside an ear defender (for a full description see [20]). $M(s)$ represents the transfer function from the scanner coils, which are the source of the sound, to the error microphone positioned under the headphones. The system records the sound received by the error microphone during one EPI brain scan, this reference sound has already passed from the scanner coils to the error microphone and so has been filtered by $M(s)$. The reference can be played in advance of the next EPI brain scan as indicated by the negative delay $z + k$. The recorded reference is filtered by the adaptive filter $W(z)$ to produce an output $y(n)$. This control sound wave is delivered via the headphones and propagates to the error microphone where the superposition of the scanner sound with this control sound produces a residual error sound $e(n)$. The transfer function $H(z)$ from the adaptive filter output to the error microphone is termed the “plant.” This needs to be included in the input to the update algorithm. The transfer function $H(z)$ was obtained by using a Maximum Length Sequence (MLS) technique [21] to quickly give $P(z)$ an accurate estimate of the transfer function $H(z)$. Inclusion of $P(z)$, means that the adaptive filter has the same transfer function on both reference and error inputs effectively compensating for the plant transfer function. This arrangement is called a single-channel feed-forward filtered-x adaptive controller (e.g. [19,22]).

To evaluate the effectiveness of ANC in attenuating MR scanner sound, we have synthesized a range of scanner sounds based upon different fundamental frequencies. These sounds were played out to a subject in a scanner simulation with loudspeakers delivering the sound close to the subject’s head. Objective measurements of the sound level, with and without ANC, were made at the ear using an optical microphone. The results are plotted as a function of the fundamental frequency in Fig. 7. We also made two similar measurements in a real MR scanner. The ANC system clearly generates high levels (32–45 dB) of objec-

Fig. 7 The level of objective cancellation the noise control system produced on simulated (black circles) and real (white circles) scanner sounds.

tive attenuation of the airborne sound at the subject’s ear for all fundamentals (see [20], for more details).

These apparent benefits are not fully realized in terms of subjective loudness reduction. Further tests involved matching the loudness of a sound played in the scanner simulation with that of the same sound played through the headphones, with and without ANC. Results varied between listeners, but the average subjective benefit for an EPI burst containing peak energy at 1,900 Hz was only 5 dB and at 650 Hz was only 12 dB. At 1,900 Hz the most subjective attenuation seen was 8 dB, whilst the least was only 3 dB. At 650 Hz the most subjective attenuation seen was 17 dB whilst the least was 7 dB. The considerable difference between the objective and subjective measurements of attenuation using ANC is likely to be a result of bone-conducted sound reaching the cochlea [23,24]. Bone-conducted sound at 500 Hz is 55 dB lower in level than the airborne sound, but gradually increases to a maximum that is only 40 dB below the airborne sound at about 1.5–2.0 kHz. The passive attenuation of the ear defenders is in the region of 40 dB at frequencies above 1 kHz. Thus, once the attenuation of the external sound reaching the ear is greater than the relative attenuation of the bone conduction path, the bone conducted sound energy will dominate perception and further attenuation of the airborne sound will provide no benefit.

6. APPLICATIONS TO AUDITORY FMRI

Many of the fMRI techniques discussed so far have been used to map out responses within the human auditory cortex to a range of acoustic features including sound level, frequency, pitch, bandwidth, modulation, interaural correlation and motion see [25,26] for reviews. Using sparse image acquisition, we typically observe that no matter what sort of complex sound pattern is presented, activation generally engages Heschl’s gyrus and adjacent parts of the planum temporale. For example, sounds with a broad

Fig. 8 Activation for three listeners for each sound level of a 300-Hz tone (measured in dB SPL). Activation is shown overlaid onto a horizontal slice of each individual’s normalized brain scan. Activation is primarily observed in primary and non-primary auditory regions, as indicated in the image at the bottom right.

frequency spectrum activate surrounding auditory regions that extend from Heschl’s gyrus across the planum polare and planum temporale [27]. The planum polare has been associated with pitch sensitivity [4], while the planum temporale is linked to the processing of many different sounds that have complex spectrotemporal patterns [4,25,28].

The primary auditory cortex, located on medial Heschl’s gyrus, is chiefly involved in processing basic sound properties, such as frequency [29] and sound level [30]. For example, both the extent and magnitude of tone activation in Heschl’s gyrus increased with the tone level from 42 to 96 dB SPL (Fig. 8). At 96 dB SPL, about 30% of this region was significantly activated. This result neatly illustrates how the pattern of auditory activation can so easily be modified by background scanner sounds that fall within this range of sound levels. The auditory BOLD signal is also modulated by the intensity of speech sounds, as well as tones and noise [31]. In this early study, the intensity of spoken words was individually determined to cover a range of speech intelligibilities (low intelligibility, 65–70 dB C-weighting; perfect intelligibility, 100 to 110 dB). Although auditory activation increased across the first four speech intensities, it then decreased slightly within the most intense speech condition. This result probably reflects a combined contribution from the responses to sound level and speech intelligibility. Subsequent brain measures of speech intelligibility have been required to find other ways to manipulate speech intelligibility, while keeping the sound level constant. ‘Intelligibility’ refers to listeners ability to comprehend the speech signal and includes several properties of language such as the phonetic features for word recognition, syntax and semantics. Since intelligibility is not determined by any single cue, it is difficult to design sound stimuli that are as acoustically complex as speech, but lack the potential for intelligibility.

Even signal-correlated noise which controls for many of the temporal properties of speech is less acoustically complex because it has a flat spectrum and does not match the distinctive pitch and harmonic structure found in normal speech. Some researchers have favoured a spectral inversion of the speech which preserves most of the spectral and temporal complexity of the speech, including some phonetic features and intonation, but destroys its intelligibility, at least to the untrained listener. Spectral inversion has been used successfully to investigate the neural substrate of speech intelligibility [32,33]. These studies have identified the speech-specific effect of intelligibility by contrasting speech and inverted speech, and have revealed activity in distinct anterior and posterior regions of the auditory cortex, with a strong left hemisphere dominance. Recently, Davis and Johnsrude [34] have systematically reduced speech intelligibility using not just one, but three different types of acoustic alteration: (i) replacing segments of the speech with signal-correlated noise, (ii) noise-vocoded speech, and (iii) speech in a noise background. For all three type of alteration, the three levels of speech intelligibility significantly correlates with BOLD activity in the left superior and middle temporal gyri.

7. CONCLUDING REMARKS

The combined use of passive attenuation by ear defenders and of sparse acquisition scanning techniques has so far proved to be the most popular way of conducting auditory fMRI research and has shed important insights into the functional organization of auditory regions on the human superior temporal gyrus. While sparse acquisition is optimal for mapping response regions, its main limitation is its lack of temporal sensitivity, since only one scan can be acquired every 9 seconds or so. Commercial solutions for reducing the scanner noise at source do not seem imminent

and so the application of ANC for auditory fMRI holds the most exciting promise. Our further work to evaluate the benefits afforded by ANC in the real scanning environment will seek to quantify these benefits. Our ultimate goal must be to reduce the scanner sound sufficiently to be able to run continuous acquisition protocols without introducing auditory masking or interference with activation patterns and without impairing speech intelligibility.

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