A fiber-optic system for recording skin conductance in the MRI scanner

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The acquisition of the skin conductance response (SCR) during functional magnetic resonance imaging (fMRI) raises significant safety issues, as well as practical ones, which need to be addressed in order for these experiments to be conducted safely and successfully. Metallic and conductive wires in the presence of time-varying gradient magnetic fields such as those present in fMRI experiments may induce heating, as well as electric fields, in these components and, if in contact with the subject, could produce severe burns and electric shocks. Moreover, these metallic and conductive components can significantly distort the magnetic field, resulting in image artifacts. A system for recording the SCR in humans simultaneously with fMRI is presented. The device is a fiber-optic-based transducer, which records the SCR from two fingers of the same hand, using electrodes containing inline radio frequency (RF) suppression filters and protective resistive loads. The fiber-optic SCR transducer was tested using 1.5 and 3.0 Tesla MRI scanners running EPI sequences. This system was able to safely record SCRs free of RF interference during an fMRI experiment, and the fiber-optic design of the transducer eliminated any artifacts on the MRI scan.

Electrodermal activity (EDA) was first observed in the 19th century (Johnson & Lubin, 1966), although the term *EDA* was introduced as a common term for electrical phenomena of the skin approximately 50 years ago. Since then, EDA has become a widely used scientific method in many areas of research, including social psychology (Porges & Fox, 1986), engineering psychology (Lindholm & Cheatham, 1983; Michaels, 1960, 1962; Preston, 1969), and conventional polygraph tests (Bradley & Warfield, 1984; Furedy, 1986; Furedy, Davis, & Gurevich, 1988; Lykken, 1981). More recently, advances in amplifier instrumentation and a better understanding of the underlying principles of EDA have led to its use in clinical disciplines. However, by far the greatest application of EDA has been in the field of psychophysiology. The majority of this work has focused on EDA concomitants, which appear during stimulation and the processing of information by the brain. Of particular interest is the relationship between EDA and subtle physiological brain processes. Changes in EDA can

occur as a result of a variety of sensory and psychological stimuli. The phasic change in EDA is often termed the *skin conductance response* (SCR), and the stable tonic EDA level the *skin conductance level* (SCL).

Traditionally, EDA has been studied in conjunction with event-related potentials (ERPs), and although these studies provide excellent temporal resolution, the scalp-recorded electrical activity provides limited spatial resolution. One way to overcome this limitation is to conduct studies in which EDA is acquired simultaneously with functional magnetic resonance imaging (fMRI), a methodology dependent on changes in blood oxygenation levels, which enables accurate spatial localization of cortical regions activated in response to sensory and cognitive stimulation in the brain. Thus, simultaneous acquisition of EDA and fMRI can provide greater insights into the spatial localization of brain function in cognitive paradigms.

However, the electromagnetically hostile environment present in the MRI room has limited the number of studies that have attempted simultaneous EDA and fMRI recordings (Buchel, Dolan, Armony, & Friston, 1999; Critchley, Elliott, Mathias, & Dolan, 2000; Williams et al., 2000; Williams et al., 2001). The main problem is the powerful alternating magnetic field environment, which prohibits placing metallic and electrical components in close proximity to the magnetic bore. Radio frequencies (RFs) typically of the magnitude of 600 MHz are present inside the MRI room, meaning that any conductive cable exit-

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ing this room acts as a powerful antenna. This can pose significant problems when one attempts to record information inside the MRI room and transmit these data to an external computer. First, an antenna effect can result in artifacts on the MRI scan, thus severely compromising the quality of the images acquired. This is a significant limitation, since one of the main reasons for employing the fMRI technique is to provide superior image resolution and enhanced anatomical localization. Second, during acquisition, large magnetic gradients are created, and these induce large amounts of electrical interference, which affects the recorded SCR. Moreover, since (like most electrophysiological potentials) the SCR amplitude is smaller than the noise, any interaction between RF interference and the SCR signal will render the acquired SCR data almost useless. Finally, there are safety considerations that need to be observed in order to avoid potential health hazards resulting from currents induced in wires located within the alternating magnetic field. It has been reported that the use of long conductive wires within such an environment can pick up RF energy and act as a secondary antenna (Felblinger et al., 1996). Moreover, in such situations, heat can be generated, resulting in full thickness burns (Bashein & Syroty, 1991; Dempsey & Condon, 2001; Kanal & Shellock, 1990).

Despite the problems identified above, Shastri et al. (2001) have described a system able to record SCR in an MRI environment by using a non-fiber-optic configuration. Their system differs significantly from the fiber-optic system described in our study. The most obvious difference is that the SCR signal of Shastri and colleagues is transmitted outside the scanner room, using a shielded cable via a specially designed set of filters installed in the filter panel of the MRI suite. The resultant signal is then filtered again before acquisition, so as to remove artifacts. The results of this study indicated that postprocessing of the signal was necessary to achieve a noise-free SCR signal. Postprocessing of the SCR signal (depending on the filter characteristics) can affect the overall tonic skin conductance level, which is a useful index of arousal. Moreover, it may alter aspects of the SCR signal itself, such as the ramp and tail sections of the SCR, as well as affecting the amplitude, all of which are important when one is trying to decompose overlapping SCR signals into discrete components (Lim et al., 1997). The installation of custom filters into the filter panel may raise problems in some MRI sites. In some instances, there may be limited or no available space for additional filters, and in other instances, scanner manufacturers may be less than receptive to the addition of these "nonstandard" filters. The Shastri system also requires the use of a Perspex stabilizing splint on the hand that is used to acquire the SCR signal, and again, this represents a significant design difference from our system. The problem with a splint is that it makes it difficult to acquire reaction time measures with the affected hand, hence requiring a one-handed reaction time buttonpress, which is not ideal, since it introduces lateralized effect into the fMRI data.

This article outlines the design of a fiber-optic-based skin conductance transducer compatible with an MRI environment. By using this system, artifact-free EDA can be safely recorded (without the use of any hand-restraining device or the installation of additional filters in the filter panel) simultaneously with fMRI. The resultant SCR signal requires no further postprocessing.

METHOD

The design of the fiber-optic SCR transducer can be divided into several distinct stages, as is illustrated by Figure 1. The first stage of the design is the electrodes and RF suppression filters. These connect the subject to the fiber-optic SCR transducer. Skin conductance response is measured by imposing a constant DC voltage across two Ag/AgCl electrodes, which are placed on the volar surfaces of two fingers of the subject's nondominant hand. The electrode pair forms part of the input circuit and is supplied by a constant voltage (0.5 V). Resultant fluctuations in the transdermal conductance are detected as changes in the flow of current. The electrodes come into contact with the skin, using a conductive paste, and subsequent opening of the sweat glands, as a result of autonomic nervous system activity, increases the conductivity and leads to an SCR. The leads connected to the electrodes are twisted together and connected to individual inline RF suppression filters (Figure 2, insert) before connecting to the main recording unit (Figure 2). The twisting of the electrode leads minimizes the effective loop area and, hence, minimizes induced voltages. This limits any heating effects that may result from the magnetic field (Lemieux, Allen, Franconi, Symms, & Fish, 1997). The inline filters act to exclude the RF interference. These consist of active second-order low-pass Bessel filters and are placed approximately 3 in. from the electrode surface. The length of conductive wire connected to the subject is kept to a minimum, which enhances subject safety. The output side of the inline filters is attached to the main recording unit via gold-foil-shielded twisted copper wires. The main recording unit is located outside the bore of the scanner and is housed in an RF-shielded case, protecting it from electromagnetic interactions and the effects of heating. The casing itself is grounded to the Faraday cage surrounding the MRI scanner room. The main unit is battery powered, and its main intention is to be used as a signal conditioner at the front end of an isolated acquisition system. The SCR signal is directed into an instrumentation amplifier with a feedback system in order to increase the common mode rejection. The amplifier outputs the difference between the conductance signal and the null signal. It has a gain of +2 for the conductance signal and -1 for the null signal, and its overall sensitivity is -100 mV/uS. The output stage drives the null meter, and after amplification, the signal is pulse width modulated (PWM), where the electrical signal is converted into a series of light pulses and the resultant optical signal is transmitted outside the scanner room, using a 10-m length polyethylene-coated optical

Figure 1. Schematic illustration of the fiber-optic skin conductance response (SCR) device. The design is depicted in various stages. The electrodes (A) are composed of Ag/AgCl and are attached to the second and third digits of the left hand. The electrodes are connected to the main recording unit (C), through individual radio frequency (RF) filters (B), where the signal is initially received and amplified. The signal is then converted into an optical signal (D) using pulse width modulation (PWM) and is transmitted through a fiber-optic cable outside the scanner room, where it is demodulated (E), passed through a low-pass filter (F), and acquired (G).

fiber (bandwidth, 400 MHz/km; 3 dB/km attenuation). The output prior to PWM has three gain settings via S2 to provide output sensitivities of 100, 500 and 1,000 mV/uS. The output can also be optionally nulled to establish a reference baseline, so that subsequent small conductance changes can be further amplified. The establishment of a reference baseline is essential when the tonic skin conductance level is acquired across subjects and is important when arousal levels are assessed. Nulling is achieved by enabling a switch, a variable adjust control, and a center zero meter. Maximum sensitivity to these small conductance changes is achieved by shifting the baseline signal to the left of the meter scale with the null control. This makes the full negative-to-positive range of the amplifier available.

From outside the scanner room, the optical SCR signal is demodulated back to an electrical signal and acquired in real time, using a laptop computer fitted with a PCMCIA data acquisition card and custom LabVIEW software.

PERFORMANCE EVALUATION

The first stage in the performance evaluation of the fiber-optic skin conductance device was conducted on a 1.5 Tesla GE LX Signa MR scanner on a subject who was participating in an existing auditory fMRI project.

Functional MRI volumes were acquired using the following gradient EPI sequence: echo time (TE) = 40 msec; repetition time (TR) = 2,000 msec; field of view = 400; matrix = 64×64 . The volunteer was fitted with head-

phones, and auditory tone stimuli were presented at 90 dB binaurally, using an MR-compatible auditory system, at an interstimulus interval of 8,000 msec. Eighty-five percent of the tones were 1000 Hz (background tones), and the remaining 15% were presented at 1500 Hz (target tones). The target tones commonly evoke a robust SCR. The SCR signal was time locked to each functional volume that was acquired via a TTL pulse generated from the scanner at the beginning of each volume acquisition. This allowed the identification of functional images in which an SCR had been generated in response to target tones. Electrodes were taped onto the volar surfaces of the subject's left hand, using a 0.05-M NaCl conductive paste. The volunteer was then positioned into the MRI scanner, and the electrode leads were plugged into the main recording unit (located outside the bore of the magnet). The fiber-optic cable was then connected to the main recording unit, directed outside the scanning room, via the waveguide, and connected to the data acquisition module. Radio communication was maintained with the subject in the scanner at all times.

The beginning of the EPI sequence was timed to coincide with the beginning of the presentation of the auditory tones, and the sequence ran for 7 min 30 sec. In that time, 56 total tones were presented. Figure 3 depicts the noise-free SCR that was recorded. The subject reported no heating or electromagnetic sensations from the electrodes throughout the entire scan.

In order to highlight the significance of the RF noise in the MRI environment on the SCR signal, the fiber-optic

Figure 2. The skin conductance circuit. The 0.5-V source is generated by U1. This device has an internal reference of 200 mV, which is amplified by the combination of R18 and R19. S1 selects the mode of operation simply by connecting the desired current path between the voltage source and the current amplifier (U2). The selected current is fed directly into U2, which converts to an inverted proportional voltage. R7 determines the sensitivity at -50 mV/uS. C1/R7 and C2/R9 act as low-pass filters. Insert: inline RF suppression filter used along each electrode lead.

SCR transducer was swapped for a standard SCR transducer. The electrodes, containing the inline filters, connected to the subject's hand, however, were left in place. These were connected to the standard SCR transducer, and the signal from this new transducer was conveyed outside the scanning room, using a coaxial cable via the scan room waveguide. An identical EPI sequence was then run. Once again, the subject reported no heating or electromagnetic sensations for the duration of the scans, highlighting the importance of the safety measures employed in this design, which include minimizing the length of conductive wires located inside the bore of the scanner, twisting pairs of wires together, and the use of individual inline filters, which not only act to filter out noise, but also afford pro-

tection from heating effects by way of their resistive load. Figure 4 depicts the recorded SCR signal using a standard SCR transducer. As is indicated, the RF interference associated with the switching of the scanner gradients significantly contaminates the resultant SCR signal and almost renders the signal useless.

The performance of the fiber-optic skin conductance device was further assessed on a 3.0 Tesla MRI scanner. A second volunteer was asked to complete a face recognition task, which involved the presentation of a series of unfamiliar faces, as well as nature scenes, to the subject. The subject viewed a computer screen (via a mirror mounted on the head coil) and was instructed that he should passively view the images and that he would be asked ques-

Figure 3. An acquired skin conductance response (SCR; using the fiber-optic SCR transducer) free from any gradient-induced radio frequency interference on the 1.5-T MRI scanner.

tions about the images at the completion of the scanning session. The functional experiment was a two-block recurring design where the experimental condition alternated every 30 sec. Each of ten face or nature scene stimuli was presented for 3 sec, separately, in each block in a counter-balanced fashion.

All imaging was performed at the Mayne Clinical Research Imaging Centre in Sydney, Australia, using a 3T

Philips Intera MRI scanner. Images were acquired using T2*-weighted gradient EPI in the axial plane with a TR of 3,000 msec, an effective TE of 40 msec, and a flip angle of 90°. Twenty-eight 4-mm functional slices were collected at a 128×128 matrix, and a total of 210 volumes were collected. Three identical functional EPI sequences were run under three different test conditions. In the first test, functional data were acquired without the device being

Figure 4. The acquired skin conductance response signal on a 1.5T MRI scanner when it was not converted to an optical signal (pulse width modulated) and, instead, was transmitted using a coaxial cable. The coaxial cable was directed outside the MRI scanner room through a waveguide.

Figure 5. A graphical representation of the imaging results from the performance evaluation for the three test conditions: no skin conductance response (SCR), fiber SCR, and cable SCR. Panel A shows the mean image (a snapshot of 6 functional images from a total of 28) constructed from the entire functional run consisting of 210 volumes. Panel B represents a signal-to-noise map where the mean image is divided by the variance image. Panel C represents the histogram of the signal-to-noise map and illustrates the frequency of occurrence of the different intensity levels in the image. The x-axis shows the intensity levels of the image, and the y-axis represents the number of pixels with certain intensity. Finally, panel D is an actual SCR recording. The time base illustrated is 10 sec.

connected to the subject (no-SCR). In the second test, functional data were acquired with the device attached and collecting SCR data via the fiber-optic cable (fiber-SCR), and in the final test, functional data were acquired while SCR data was being collected using a traditional cable transfer method (cable-SCR).

The functional data were analyzed in two ways. First, they were quantitatively analyzed to assess whether the presence of the device significantly increased the noise level of the MR signal. In order to do this, the *z*-statistic method (Devore, 1982), was chosen, in which changes in the variance of the data with and without the device present were evaluated. The *z*-statistic method can be expressed as the following:

$$z = \frac{s1 - s2}{s2/\sqrt{2n}},$$

where *s1* and *s2* are the standard deviations with and without the device attached, respectively, and the variable *n* refers to the number of samples, which in this case is 210 and corresponds to the total number of volumes. The *z*-statistic test provides a quantitative assessment of signal noise and not simply a measure of interference. Since it

was envisaged that the fiber-optic skin conductance device would have utility in a range of cognitive paradigms (thus, activation foci could spatially differ considerably, depending on the cognitive processes at work), a conservative whole-brain analysis was adopted, rather than a region-of-interest approach, which may not necessarily be representative of whole-brain activity, especially in the case of ventral regions. An alpha level of .05 was chosen to control for Type I error. This level corresponded to a *z*-score of 5.30 corrected for a 9,000-voxel (whole-brain) comparison. Thus, an absolute *z*-score of greater than 5.30 would indicate that the addition of the SCR device during the acquisition of the functional images had significantly affected the data. Figure 5 is a graphical representation of the results from the three test conditions, and Table 1 is the tabulation of the associated *z*-scores for the various comparisons.

No significant increase in *z*-score was noted when the no-SCR and fiber-SCR trials were compared, indicating that the addition of the fiber-optic device did not significantly contribute noise to the MR signal. On the other hand, when the no-SCR condition was compared with the cable-SCR condition, a significant increase in the *z*-score was recorded. Similarly, when the fiber-SCR and the cable-SCR conditions were compared, a signifi-

Table 1
Tabulation of the Standard Deviation (SD) and z-Score Values
for the Tests in Which the Noise Contributions From Three
Configurations Were Evaluated With the z-Statistic

	No SCR	Fiber SCR	Cable SCR
<i>SD</i>	37.73	41.22	68.21
Score ¹	—	1.89	16.56
Score ²	1.73	—	13.43
Score ³	9.15	8.11	—

Note—Score¹ corresponds to the z-score comparisons in which the SD of the no skin conductance response (SCR) was used as the reference value, *s*₂ (see the Performance Evaluation section for the relevant equation). Score² and Score³ correspond to the z-score comparisons in which the SDs of the fiber-SCR and cable-SCR configurations, respectively, were used as the reference value *s*₂. An alpha level of .05 was chosen to control for Type I error, which corresponded to a z-score of 5.30. An absolute z-score greater than 5.30 for any comparison indicates a significant effect on the temporal data.

cant increase in the z-score was also recorded. The latter two results indicate that the cable-SCR trials contributed significant noise to the MR images.

It is worth mentioning that all of the functional images acquired for the z-statistic tests were at a 128 × 128 matrix, and thus it was possible to visually detect a noise artifact when one was present (see Figure 5). However, a large proportion of functional images are acquired at a lower (64 × 64) resolution, and thus, it is more difficult to de-

tect additional noise artifacts. In cases in which traditional cables methods are used to transmit the SCR signal outside the scanner room, it would be prudent to investigate the whole-brain noise levels of MR images, especially in situations in which the SCR signal is sufficiently noisy and requires postprocessing to remove artifacts.

The data were also analyzed using Brainvoyager QX (Brain Innovations, 2001) to identify regions of significant blood oxygenated level dependent (BOLD) activity, and any subsequent activation foci were compared with findings in the current literature with respect to the regions activated. The functional data for all three test conditions were preprocessed using the following protocol. First, all the data were motion corrected and were corrected for slice timing by 107 msec. The data were then smoothed using a Gaussian 4-mm FWHM (3 × 3 kernel) smoothing filter, and finally, a voxelwise detrending algorithm was applied in order to remove any linear trends in the MR signal due to scanner drift. All three test conditions produced statistically significant fusiform gyrus activations as a result of the face stimuli and parahippocampal gyrus activations to the nature scene stimuli, in accord with previous studies in which a similar paradigm was used (Epstein, Graham, & Downing, 2003; Grill-Spector, Knouf, & Kanwisher, 2004; Horowitz, Rossion, Skudlarski, & Gore, 2004). Figure 6 illustrates these activations, together with the temporal display of the BOLD signal across the functional para-

Figure 6. Brain activation maps and accompanying blood oxygenated level dependent (BOLD) temporal display curves. The green/blue activations represent those correlating to the presentation of nature scene stimuli, and the yellow/orange activations correlate to those of the face stimuli. The BOLD temporal display curves were all taken from an identical region of interest in the parahippocampal gyrus across the three different test conditions. SCR, skin conductance response.

digm. Although significant activation foci were identified for all three conditions, the cable-SCR trial was associated with a smaller activation cluster for the parahippocampal gyrus. Moreover, the BOLD signal temporal display in the cable-SCR condition visibly contained more noise than did those in the no-SCR and fiber-SCR trials.

CONCLUSION

The system described in this article, for measuring skin conductance, using a fiber-optic-based SCR transducer specifically designed for use in an MR environment, offers a convenient solution to the technical problems associated with biological signal recording in an MRI environment. The system permits the acquisition of continuous noise-free SCR but can also be modified to record most biological signals safely and free from artifacts.

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