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Use of Optoelectronics to Measure Biosignals Concurrently During Functional Magnetic Resonance Imaging of the Brain

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1. Introduction

In the rapidly advancing world of biomedical imaging and engineering, magnetic resonance imaging (MRI) has become an indispensable technique to visualize normal and diseased anatomy non-invasively in the human body. The 2003 Nobel prize in Physiology and Medicine to biophysicists Lauterbur and Mansfield is a testament to the significant health care advances that have followed from the early development of MRI technology. In the past three decades, MRI technology has not only matured, but has continued to diversify to encompass new applications in science and medicine. The focus of this chapter is one such application, that involves use of optoelectronic devices to measure electrical biosignals during concurrent mapping of brain activity with a technique called functional MRI (fMRI).
Since the development of early MRI systems, there has been an ongoing need to develop ancillary devices adjacent to or within the magnet bore. For example, clinical MRI systems are now equipped with sensors to measure heart rate, respiratory rate, and electrocardiograph signals that provide essential summaries of the physiological status of the patient during scanning. These and other devices (e.g. wheelchairs, incubators, power injectors to deliver drugs and contrast agents, and interventional devices such as catheters) cover a wide range of functions and electromechanical complexity. However, they are all carefully designed with common consideration of several critical factors. The MRI system uses three electromagnetic fields to produce an image. First, radio-frequency (RF) coils are used to transmit and receive RF fields to and from the patient, at a frequency of approximately 100 MHz. Second, a very strong, static, homogeneous main magnetic field is required primarily for signal-to-noise ratio (SNR) considerations. This field typically has a strength of 1.5 or 3.0 Tesla, or approximately 50 000 – 100 000 times the strength of the Earth's magnetic field, with spatial uniformity to approximately less than 1 part per million over a 20 cm diameter spherical volume. Third, time-varying magnetic gradient fields are produced along orthogonal directions by gradient coils to encode MRI signals spatially, with amplitudes of approximately 10 mT/m and slew rates of approximately 100 T/m/s.

The use of electromagnetic fields and their relation with basic MRI theory have implications for ancillary MRI devices in several different ways. The formation of MR images relies on resonant excitation of magnetization primarily related to hydrogen nuclei on water molecules within biological tissues, and subsequent reception of RF signals after spatial encoding and additional manipulations that introduce signal contrast between different normal tissue types and disease states. For 1.5 T and 3.0 T MRI systems, the resonant frequencies of interest are 64 MHz and 128 MHz, respectively. The use of imaging gradients for spatial encoding leads to a bandwidth of interest of several 100 kHz surrounding these resonance frequencies. Within this bandwidth, it is important that devices introduce negligible RF interference with the MRI system, and *vice versa*. In addition, patient safety must also be considered (Schaefers 2008). Substantial attractive forces can easily occur between the very large static magnet field and ancillary devices, if the device contains ferromagnetic components. Such interactions also distort the static magnetic field uniformity and can introduce unacceptable spatial distortions and signal loss in MR images. There is also the potential for the time-varying electromagnetic fields generated by the MRI system to interact with electrically conductive devices to cause unwanted heating (Lemieux, Allen et al. 1997), given that the RF power amplifiers used in MRI systems are in the range of 1–30 kW. Consequently, much attention has been paid to the establishment of safety standards, through bodies such as the American Society for Testing and Materials (ASTM; http://www.astm.org/), and to evaluation of the extent of compatibility of ancillary devices with MRI systems at different magnetic fields (see http://www.mrisafety.com/). According to the new system of terms recently developed by the ASTM, devices are classified as MR-safe, MR-conditional, or MR-unsafe. MR-safe indicates no known hazards in all MR environments, whereas MR-unsafe indicates the converse. MR-conditional indicates devices can be used under specific MRI conditions, determined by specific device testing.

For obvious reasons, the safety aspects related to MRI ancillary devices are subject to established international standards or guidelines that are applicable in different countries. Although similar standards have been developed to characterize the electromagnetic emissions from such devices, associated with regulatory bodies (e.g. see http://www.fcc.gov/oet/), these standards do not evaluate specifically the potential for electromagnetic interference between the device and the MRI system itself. Electromagnetic interference primarily affects SNR and contrast-to-noise ratio in a manner that has become more stringent with time as the fidelity of MRI systems continues to improve with new technological innovation. Notwithstanding this issue, the static and time-varying electromagnetic fields used in MRI systems usually ensure that conventional electrical design approaches are not feasible, and necessitate dedicated ancillary device development.

In this regard, optoelectronics have played and continue to play a very important role in development of devices that generate biosignals measured concurrently with MRI. Optoelectronics provide several major advantages: 1) the use of non-ferromagnetic optical fibres instead of coaxial cables eliminates a major route for unintended electromagnetic emissions within the critical RF bandwidth of the MRI system, while eliminating possible magnetic field distortions and attractive forces; and 2) optical fibres are not susceptible to interference from external RF fields. For these compelling reasons, optoelectronics have been incorporated into the receiver chains of modern MRI systems, with substantial SNR benefits.

Ancillary devices are important when it comes to the field of blood oxygenation level-dependent (BOLD) functional MRI (fMRI) neuroscience research (Ogawa, Tank et al. 1992; Hennig, Speck et al. 2003). Functional MRI is a popular imaging tool used in neuroscience

because it measures brain activation associated with cognitive, sensory or motor tasks and behaviours. Compared to other functional neuroimaging techniques like positron emission tomography, near infrared diffuse optical tomography, electroencephalography (EEG), and magnetoencephalography, the BOLD fMRI technique has very attractive characteristics. Neither ionizing radiation nor injectable contrast agents are required. Activity of neurons is inferred through coupled changes in blood oxygenation, volume and flow that are reflected in magnetic resonance signals. Extensive research has addressed the biophysical basis for the fMRI signal, and enough is known about the neurovascular relationship at this point (Logothetis, Pauls et al. 2001; Menon 2001; Attwell and Iadecola 2002) to be confident that fMRI signals are strongly coupled with brain activity in healthy individuals and many patient populations (although there are provisos (Cohen, Ugurbil et al. 2002; Roc, Wang et al. 2006)). The spatial and temporal resolution provided by fMRI (millimeters and seconds, respectively) is not presently achievable throughout the brain volume by any other functional neuroimaging method.

During fMRI, brain activity is often measured for a specific behavioural task. The details of the task are determined by the mental processing under study, be it sensory, motor, or cognitive. With many tasks, there remains a need for an additional, concurrent measurement that serves either to quantify behavioural performance, or to provide an additional probe of mental state. Through appropriate signal processing, these concurrent measurements can be used retrospectively during post-processing to improve the detection of brain activity by fMRI analysis methods (Friston, Holmes et al. 1995).

This chapter focuses on biosignals (i.e. electrical signals indicative of physiological function) that can be measured simultaneously during fMRI. The focus herein will be on example biosignals that complement and augment brain activation maps calculated from fMRI data. In each of the cases that we will explore, optoelectronics have played a critical role in technological advancement. Three biosignals are considered: electroencephalography (EEG), electromyography (EMG) and electrodermal activity (EDA). Each technique has a common methodological approach: electrodes placed on the skin, amplification of the signals of interest, followed by digitization and subsequent transmission of signals from the magnet room to a display or recording device. Early research in this field identified the safety considerations associated with using electrodes in the MRI environment (Lemieux, Allen et al. 1997; Allen, Polizzi et al. 1998). EEG, EMG, and EDA devices are now commercially available for fMRI applications.

2. Electroencephalagraphy and functional MRI

Electroencephalography (EEG) is a non-invasive technique used to measure voltages on the scalp that are produced by the electrical current related to neuronal signaling in the brain. Participants wear a cap covered with numerous electrodes (typically ranging between 32 and 128). Neurons in the brain are electrically charged cells by virtue of the membrane proteins that pump ions in and out of the cell to create a resting potential. Signal propagation along neurons occurs by action potentials, which are rapid changes in the membrane potential caused by the opening of ionic channels. The electrodes detect the sum voltage signals, primarily from post-synaptic potentials, that are conducted to the surface of the head. Pyramidal neurons in the brain's cortex are thought to produce the strongest EEG signals because of their alignment and their firing characteristics. The frequency range of EEG signals is from 4 to 100 Hz and can been deconstructed into different frequency bands.

For example, alpha rhythms are found within 8–12 Hz activity and are present when the participant is awake with their eyes closed (Goldman, Stern et al. 2002).

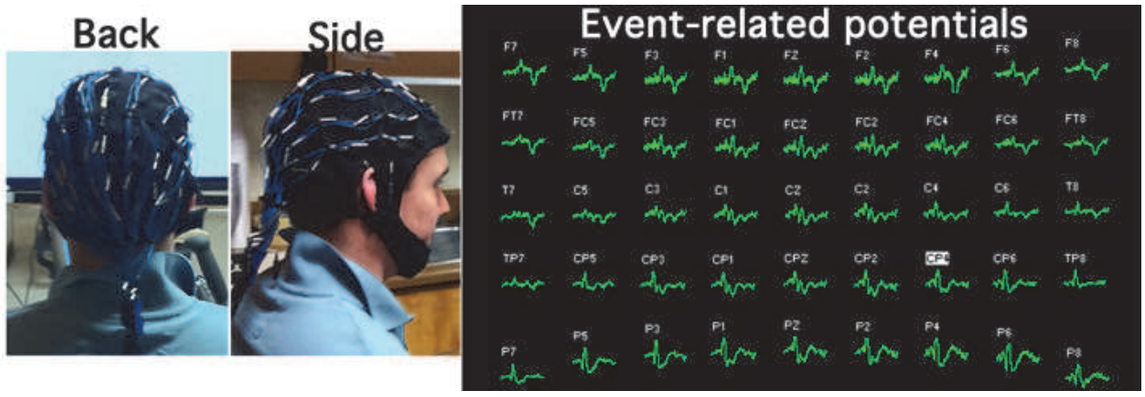

Fig. 1. Left: Back and side views of a participant wearing a 64-channel electroencephalography (EEG) cap. Wires from individual electrodes are collected in a bundle at the back of the head. Right: Event-related potentials (ERP) are shown across each of the different electrode sites, listed according to their position on the head. In this example a motor response (button press) is made in response to a decision task. On inspection, the electrode sites in the region surrounding P1 and P3 have large ERP amplitudes. EEG ERP are typically in the micro-volt range. Data shown in this figure were collected outside of the MRI scanner room and are therefore free of MRI-related artifacts.

EEG voltages are low amplitude (typically microvolts) and are easily contaminated by environmental noise. As a result, EEG can involve lengthy temporal recording over tens of minutes. Signal averaging over multiple trials is commonly employed in the study of event-related potentials (ERP), the term used to describe the EEG signals that are derived from tasks that are separated into single or multiple brief events to resolve different temporal and spatial aspects of neural activity (Figure 1). Further, intensive signal processing is often used to reduce noise levels. Within an MRI system, EEG signals are additionally degraded to the point that conventional EEG hardware cannot be utilized. First, time-varying changes in the magnetic flux through a loop geometry (which can be created through the EEG electrodes and wires) will produce an electromotive force (EMF) by Faraday's law of induction (Allen, Polizzi et al. 1998). These voltages can be produced in EEG time series data either by movement of an electrode in the static magnetic field, or by a stationary electrode in the presence of a time-varying magnetic field. The former case introduces artifacts arising from head motion, which can be postural in nature, or can arise from the pulsatility of blood flow in the scalp, an effect known as the "ballistocardiogram artifact" (Allen, Polizzi et al. 1998). These artifacts are typically approximately the same order of magnitude as EEG signals. Regarding the latter case, the magnetic field changes produced by rapidly varying imaging gradients can produce very large artifacts, several orders of magnitude larger than EEG signals and the primary MRI-induced artifact in EEG data.

Fortunately, gradient artifacts can be greatly attenuated through optoelectronic technology. Today, EEG systems for use with fMRI typically use twisted-pair electrodes with short lead lengths, local amplification, and conversion to optical signals to reduce EMF artifacts substantially. This strategy ensures that if the MRI system is sufficiently stable in the temporal domain, and EEG data are sampled at sufficiently high frequency, characteristic

artifacts from specific gradient waveforms can be determined by temporal averaging and then simply subtracting from EEG data, effectively suppressing the problem (Allen, Polizzi et al. 1998).

There are also mechanisms by which EEG hardware influences MRI signals. During EEG-fMRI, the EEG cap (Figure 1) can cause spatial distortions in some types of MRI data acquisition, particularly echo planar imaging (Jezzard and Balaban 1995) and spiral imaging (Glover and Lai 1998) that are commonly used in BOLD fMRI. These image distortions arise from spatial non-uniformity imposed on the static magnetic field due to the magnetic susceptibility properties of the cap, electrodes, electrode gels, wires, or pre-amplifiers, and due to the magnetic fields associated with the currents induced in EEG hardware according to the mechanisms summarized above (Bonmassar, Hadjikhani et al. 2001). Empirically, water-based electrode gels have been found superior to oil-based gels within the MRI environment, because the water gels have magnetic susceptibility properties more similar to those of the head (Bonmassar, Hadjikhani et al. 2001). Irrespective of the type of electrode gel, the function of the gel is to reduce the scalp resistance to improve the effective electrode contact and improve EEG signal strength. Unfortunately, the gel has potential to localize RF power deposition during the resonant excitation component of MR image formation. Although this effect is not problematic at 1.5 T and 3.0 T for BOLD fMRI, which involves relatively modest RF excitation, substantial heating is possible when EEG is conducted simultaneously with other MRI acquisitions. The effect is more pronounced at 3.0 T, as power deposition is proportional to the square of the static magnetic field. For this reason, certain MRI applications (for example, "fast spin echo imaging" and perfusion imaging of cerebral blood flow using "arterial spin labeling") may be contraindicated during EEG due to concerns of exceeding heating safety thresholds established by regulatory agencies (e.g. the US Food and Drug Administration).

It is evident that there are numerous challenges involved in conducting simultaneous EEG and fMRI measurements. Addressing these challenges is worthwhile for several different motivations. The major clinical motivation is the investigation and evaluation of epilepsy (Warach, Ives et al. 1996; Seeck, Lazeyras et al. 1998; Schomer, Bonmassar et al. 2000; Benar, Gross et al. 2002; Benar, Aghakhani et al. 2003). Very succinctly, EEG can be used to identify the onset of epileptic discharges and fMRI can be used to help localize where the seizure originated. Although the extent to which this idea will work in clinical applications still needs further investigation, such work has potential in the future to assist in selecting for individual patients the appropriate therapeutic intervention, such as drug treatment, or neurosurgery.

It can be said that "the sum of EEG and fMRI measurements is greater than their individual parts". This is because due to the physical principles underlying each modality, fMRI and EEG signals are highly complementary. EEG reflects electrical activity of populations of neurons in the brain (i.e. pyramidal neurons), over millisecond timescales. However, EEG exhibits limited spatial resolution due to the well known, "inverse problem" of detecting multiple current sources within a volume based on the signals observed in an array of remote detectors (Plonsey 1963). The spatial resolution of EEG is typically on the order of 1 cm, and EEG signals attenuate substantially with depth within the brain. In contrast, BOLD-fMRI measures changes in blood oxygenation with much lower temporal resolution, but markedly improved spatial resolution throughout the entire brain volume. In addition to the complementary nature of fMRI and EEG, the highly variable nature of individual behaviour and brain activity as a function of time often argues for the acquisition of EEG and fMRI

data simultaneously. For example, subtle memory and learning effects can mean that equivalent behavioural performance in repeated tasks, measured sequentially by fMRI and EEG may not be manifested as the same brain activity. In such circumstances, it is essential to acquire EEG and fMRI data simultaneously as brain activity evolves.

In one illustrative basic neuroscience application, fMRI was used to identify the brain regions that were correlated with the EEG alpha band rhythms while participants lay in the MRI scanner, awake with their eyes closed. This experiment is an example of a "resting-state" study, during which the participant is not required to perform specific behavioural tasks. Experiments that probe resting-state brain activity, and the correlations in resting-state activity that occur in specific networks throughout the brain, have been increasingly popular in recent years. EEG-fMRI is well poised to advance this line of research (Musso, Brinkmeyer et al. 2010). Due to lack of task demands, resting state studies are simpler to conduct on patient populations and therefore attractive from a clinical perspective.

3. Electromyography and functional MRI

Turning attention now to another important biosignal, electromyography (EMG) is the study of electrical signals underlying muscle activity. EMG recordings are often used to evaluate muscle firing patterns, such as during walking (Hof, Elzinga et al. 2002). This is possible because as in the case of EEG, muscle activity is recorded with millisecond temporal resolution. EMG is useful clinically in the study of numerous movement disorders and neurological diseases such as stroke, motor neurone disease and others (Kautz and Brown 1998; Moreland, Thomson et al. 1998).

Briefly summarizing how the EMG signal relates to muscle physiology, the basic cellular unit of skeletal muscle is the muscle fibre. The functional unit within the fibres is called the sarcomere, which contains thick and thin filaments that are arranged alongside a scaffolding of cross-bridges (Gulrajani 1998). A muscle contraction involves the shortening of the muscle fibres, as described by "sliding filament theory". This process involves the sliding movement of the thick and thin filaments within the muscle, over top of and toward each other. As the overlap between the thin and thick filaments increases, the muscle length decreases. Such a muscle contraction is initiated by electrical signals that are like volleys of voltages, known as action potentials, that occur at different fibres within the muscle and originate from the controlling motor neurons in the central nervous system. A single "motor unit" is defined as a single motor neuron and all the muscle fibres that it innervates. The EMG signal is the accumulated voltage from all motor units and muscle fibres as they each act to contract the muscle.

EMG measurements are usually made by placing electrodes on the surface of the skin above the muscle of interest, although subcutaneous placement is also possible. In comparison to EEG signals that although they are noisy they show oscillatory or transient waveform characteristics, EMG signals appear as noise bursts during muscle contractions. The EMG signal is a zero-mean stochastic process and the standard deviation of the EMG signal is proportional to: 1) the number of active motor units and 2) the rate at which the motor units are activated (Clancy, Morin et al. 2002). The EMG signal is typically larger in amplitude compared to the EEG signal, ranging up to approximately 50 mV over a bandwidth of 10 – 200 Hz (Basmajian and De Luca 1985). Signal processing for EMG typically involves rectifying the voltage signal, lowpass filtering, and then evaluating the resulting signal envelope.

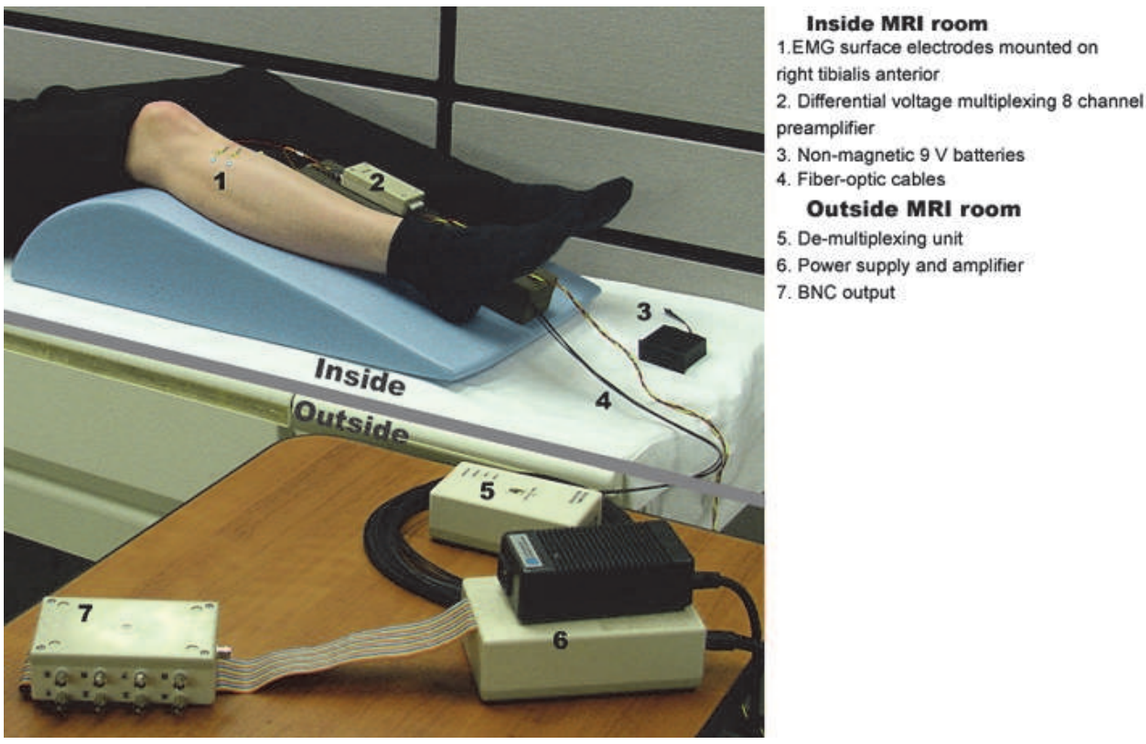

Fig. 2. MRI-compatible electromyography (EMG) system that is capable of measuring muscle biosignals inside the MRI environment and during concurrent fMRI data acquisition. The picture was taken outside of the MRI scanner room. The text "Inside" and "Outside" are used to denote hardware components that are positioned inside and outside of the MRI scanner room during EMG recording sessions. Item 2 is the pre-amplifier device that is capable of measuring 8 EMG channels (i.e. 8 muscles) simultaneously by making use of an optical multiplexer. The pre-amplifier includes optoelectronics to convert voltage signals to fibre optic signals that are sent out of the MRI scanner room via waveguide.

Several studies have involved combined EMG with fMRI (Toma, Honda et al. 1999; Liu, Dai et al. 2000; Dimitrova, Kolb et al. 2003; van Duinen, Zijdewind et al. 2005; MacIntosh, Baker et al. 2007). Similar to the early EEG-fMRI literature, initial EMG-fMRI recordings involved discarding portions of the EMG data collected during fMRI acquisition. This was because the time-varying gradient magnetic fields generated at specific time periods during fMRI saturated EMG preamplifiers. In these early studies, EMG signals were also measured by decreasing the temporal resolution of fMRI to create quiescent periods (Liu, Dai et al. 2000), an approach that was not ideal from an experimental point of view.

Some of the characteristics of a modern fMRI-compatible EMG system are large dynamic range and limited pass frequency bandwidth. With respect to the latter, EMG preamplifiers with 15 to 80 Hz and 33 to 80 Hz bandwidth were found to significantly reduce the amount of variance when EMG data were collected in a static magnetic field (MacIntosh, Baker et al. 2007). EMG signals are converted to optical signals at the preamplifier and transmitted outside the MRI scanner room using fibre optic cables. Figure 2 shows a set-up of an fMRI-compatible EMG system that was developed for use in our laboratory.

The amplitude of fMRI-induced artifacts in EMG signals depends on factors that are analogous to those described above for EEG-fMRI. First, the EMG artifact will depend on

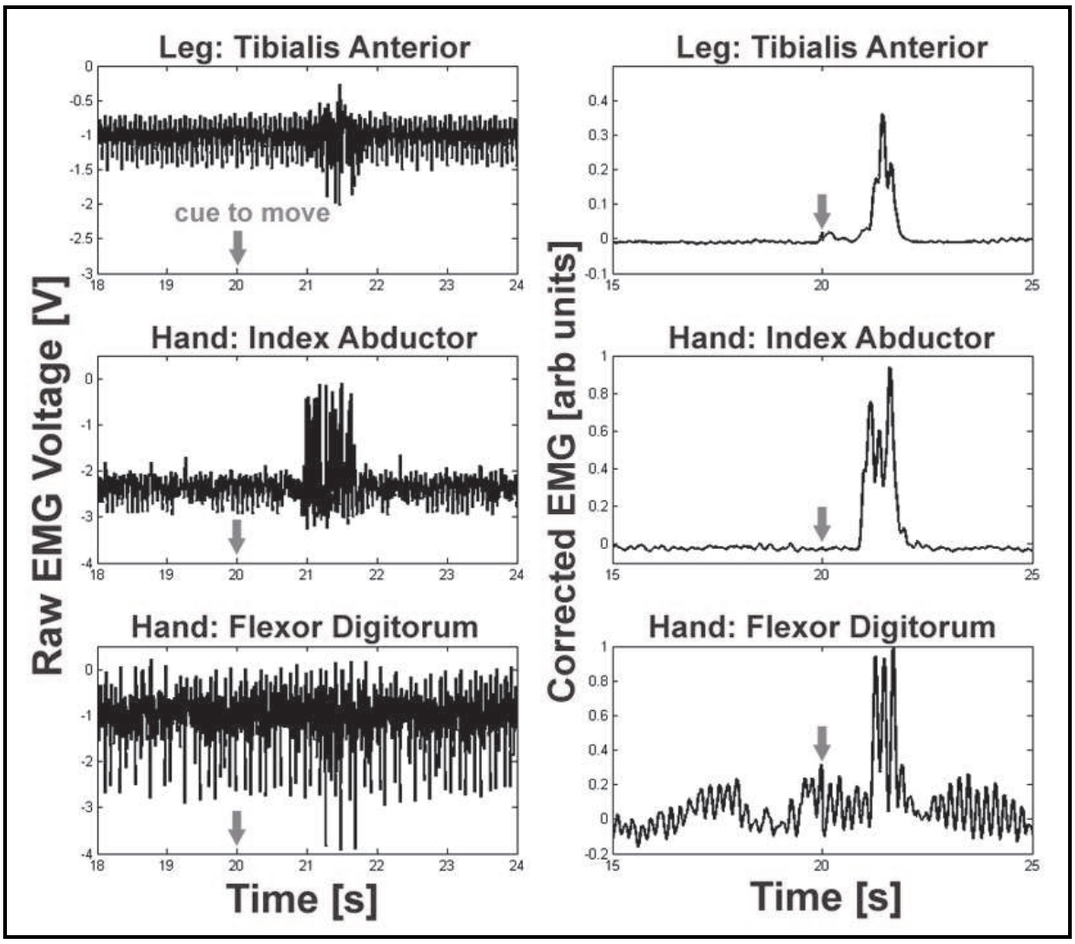

Fig. 3. Electromyography signals from 3 different muscles measured concurrently during an fMRI brain activation scan. On the left the raw amplified EMG signals are influenced by the muscle biosignal of interest as well as the signal artifact associated with the time-varying magnetic fields generated by the imaging gradient hardware active during fMRI. The muscle EMG signal occurs approximately 1 s after the participant is given a cue to move the muscle, which corresponds to t=20 s on the time axes. The three muscles are: 1) tibialis anterior, which is located on the lower leg and used to rotate the ankle; 2) index abductor, the muscle that moves the index finger towards the thumb; 3) flexor digitorum, the muscle on the forearm that flexes the wrist. EMG time series shown on the right have been corrected so as to suppress the artifact contribution. The suppression works well for the tibialis anterior and index abductor, but less so in the hardest case of the flexor digitorum. Details on post-processing approaches are discussed below. These data are based on published work (MacIntosh, Baker et al. 2007).

how much the electrode moves in the main magnetic field as a result of muscle contractions and relaxations. This can be problematic if the task is intended to create large movements. Second, the EMG artifact will depend on the locations of electrodes on the patient, in relation to the time-varying magnetic fields produced by the imaging gradients during fMRI. For example, the imaging gradient along the longitudinal axis of the magnet bore produces small time-varying fields at isocentre (where the head is located for fMRI). These

fields increase linearly with distance from isocentre to a maximum approximately at the entrance to the magnet bore, and then reduce substantially in amplitude with increasing distance outside the magnet, toward the end of the patient table. Regarding this point, Figure 3 shows that the fMRI-induced EMG artifact is more substantial for recordings of hand muscles, compared to those of the lower leg, and that fMRI-induced artifact at the forearm is even larger, as observed in EMG signals from a muscle that moves the wrist (MacIntosh, Baker et al. 2007). These effects will depend on the specific MR imaging sequence used, as well as the specific characteristics of the EMG preamplifiers.

EMG-fMRI studies are useful for improving understanding of how the brain controls muscle movements, as reflected in recent literature (van Duinen, Zijdewind et al. 2005). We have shown that the use of EMG to estimate the onset, offset, and duration of brief muscle contractions, and subsequent use of this information in fMRI data processing, leads to improved visualization of motor networks (MacIntosh, Baker et al. 2007). EMG-fMRI has also been used to show differences in brain activation associated with active, passive and electrically stimulated muscle contractions (Francis, Lin et al. 2009).

4. Electrodermal activity and functional MRI

The skin plays an important role in many biological processes such as sensory and motor exploration, immunity, and thermoregulation. In the latter process, eccrine sweat glands in the skin secrete water and salts to regulate body temperature. They are innervated by a structure in the brain known as the hypothalamus. Sympathetic nerves make the connection between the hypothalamus and the sweat gland, as part of the autonomic nervous system (sometimes referred to as the visceral nervous system). Secretion of sweat in skin is therefore a window into the human body's visceral controls in the sense that they occur without being conscious. An example of the autonomic nervous system acting with little willful control is the fight or flight response that occurs when we are in a stressful situation.

The voltages related to electrodermal activity (EDA) are very weak and poorly localized. Therefore, skin conductance is the preferred physiological recording parameter to characterize this aspect of the autonomic nervous system. Changes in EDA are described on two time scales of changing skin conductivity: skin conductance level (SCL), reflecting low frequency changes (i.e. approximate frequencies are < 0.1 Hz); and skin conductance responses (SCRs), reflecting more rapid changes (i.e. approximate frequencies are > 0.5 Hz). Both types of fluctuations obviously occur at much lower frequencies than EEG and EMG measures.

EDA is also a useful surrogate measure of cognitive and emotional processes and therefore has appeal in psychology and neuroscience research. One important aspect of cognition is arousal, the psychological state of heightened activity or attention. Being alert by directing our attention to a specific task allows us to carry out certain actions or behaviours. Arousal plays an important role when our world is altered quickly, such as due to a postural instability (Sibley, Lakhani et al. 2010; Sibley, Mochizuki et al. 2010). In this case, autonomic and cortical responses help to avoid a fall or injury. EDA responses relate to arousal as they are part of the behavioural inhibition system (with sensitivity to punishment and avoidance motivation), whereas heart rate is part of the behavioural activation system (sensitivity to reward and approach motivation) (Fowles 1980). Interestingly, EDA and heart rate are two biosignals that are used in a lie-detector test.

The first EDA-fMRI studies investigated the effects of attention on cognitive tasks (Williams, Brammer et al. 2000). Subsequently, other researchers used EDA and fMRI to monitor brain responses associated with gambling (Patterson, Ungerleider et al. 2002), biofeedback (Critchley, Melmed et al. 2001), motor tasks (MacIntosh, Baker et al. 2007), motor recovery after stroke (MacIntosh, McIlroy et al. 2008) and sense of effort (Mochizuki, Hoque et al. 2009). A simple circuit diagram for measuring skin conductance as part of an fMRI-compatible EDA system is shown in Figure 4. The optoelectronic components are listed as IF-E96 and IF-D91. IF-E96 is a plastic fibre optic red light emitting diode (commercially available at: http://i-fiberoptics.com/). Optical cable can easily be mounted to this component to transmit EDA signals outside of the MRI scanner room via a waveguide and converted back to a voltage at the IF-D91 photodiode detector. The fMRI-compatible EDA system shown in Figure 4 is MRI-safe for use at 1.5 and 3 T scanners. Also shown is the battery box that contains a non-magnetic 9V lithium battery.

The physiological frequency content in EDA is narrow and low frequency, so sharp lowpass filters are advisable. One hardware implementation that is therefore advisable in EDA design is a slew-rate limited preamplifier (Ives, Warach et al. 1993), which yields more effective lowpass filtering. The instrumentation amplifier is shown as INA-120 in Figure 4.

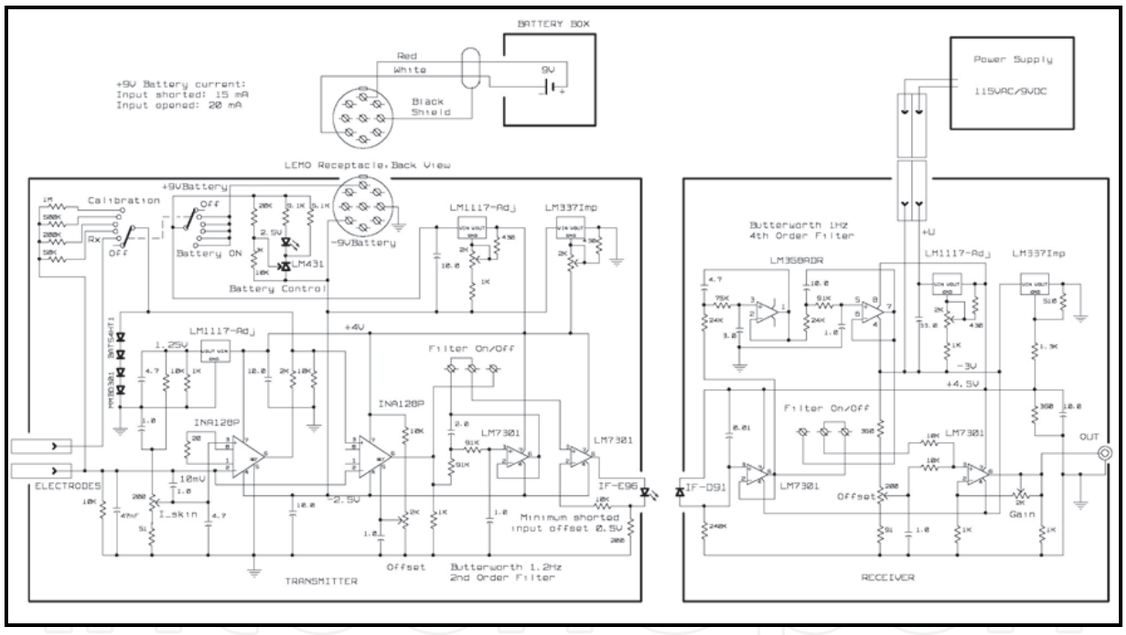

Fig. 4. The circuit diagram for a low-cost MRI-compatible electrodermal activity (EDA) system that was prototyped at Sunnybrook Research Institute. The circuit design is based on the Wheatstone bridge preamplifier with a gain of 1000 and is based on early work described by others (Shastri, Lomarev et al. 2001). Additional details associated with this system can be found elsewhere (MacIntosh, Mraz et al. 2007).

EDA-fMRI can be used to look at how behaviour changes over the course of an experiment. If we use the EDA data shown in Figure 5 as a model in the fMRI analysis then we can identify the brain regions that are correlated with the EDA trial-by-trial trends. Figure 6 shows the EDA trial averages that were used as model predictors in the fMRI analysis.

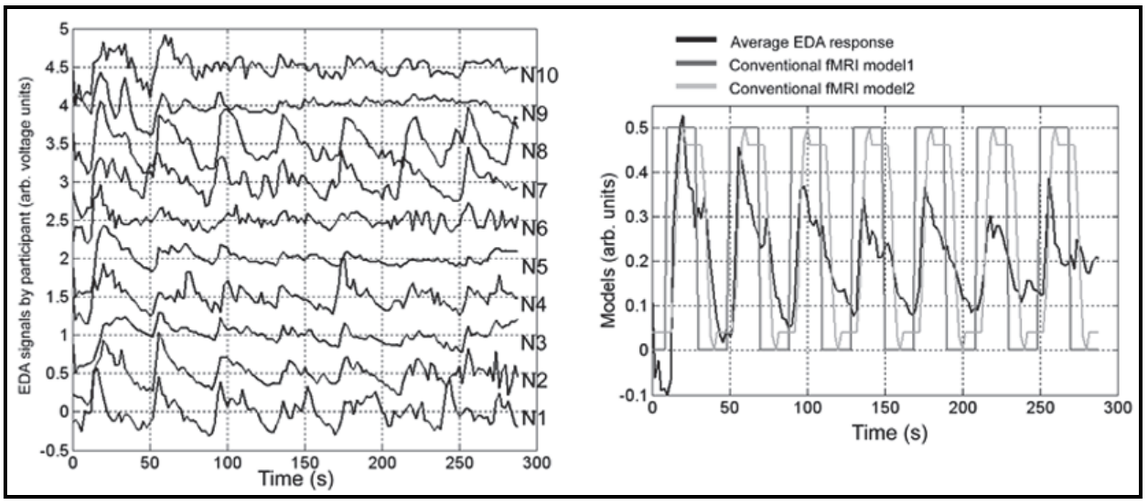

Fig. 5. Left: Electrodermal activity time series data from healthy adult participants who performed a hand motor task that required concentration and arousal to perform effectively. Each trace corresponds to a single participant (i.e. N1 corresponds to participant 1, N2 corresponds to participant 2, for a total of N=10 participants). The task, affectionately known as the "Spock task", involved separating the middle and ring fingers, requiring some concentration and dexterity. Participants alternated between 20 seconds of performing the task and 20 second of rest. Fluctuations in EDA signals across the seven task trials are more evident in some participants compared to others. Right: The average EDA time series is shown (black) from all participants. Two typical fMRI models are shown in gray for comparison and it is clear that the EDA time series data has unique temporal features that are not captured by the fMRI models (which assume constant neural activity). For example, the amplitude of the EDA signals at later trials is decreased compared to early trials. These data are based on published work (MacIntosh, Mraz et al. 2007).

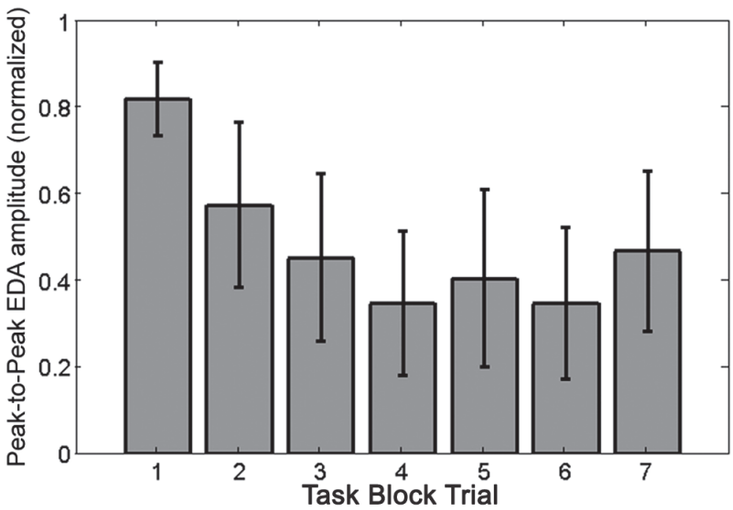

Fig. 6. Normalized peak-to-peak EDA amplitude plotted as a function of the task block trials. After the second trial the amplitude is consistently reduced. Error bars denote standard deviation.

5. Other examples of optoelectronics used in fMRI

A recent example of optoelectronics used in fMRI is a system that is able to measure diastolic blood pressure through an fMRI experiment (Myllyla, Elseoud et al. 2011). This device uses a fibre optic sensor to measure skin movements associated with blood pressure changes in the underlying blood vessels. The system works in the MRI scanner room and is useful for collecting blood pressure data because it is non-invasive (i.e. no intra-arterial line) and it does not involve an inflatable cuff. Fibre optic technology has also been used to monitor movements, or kinematics, during fMRI. This has been done to measure amplitude and velocity in ankle movements, using optical fibres specially configured as bend sensors (MacIntosh, Mraz et al. 2004).

6. Software solutions for reducing fMRI-induced artifacts in EEG, EMG and/or EDA

Having discussed some biosignals that are of interest during fMRI, it is important to now consider post-processing software approaches to minimize or correct for some residual or inevitable fMRI-induced artifacts. For software approaches to be effective, the measured voltages must fall within the dynamic range of the amplifier system. There is little that can be done when an amplifier saturates. Post-processing software approaches that attempt to reduce the fMRI-induced artifacts include: adaptive noise canceling, direct artifact subtraction (Allen, Polizzi et al. 1998; Allen, Josephs et al. 2000), filtering (Hoffmann, Jager et al. 2000), and principal component analysis (PCA) and independent component analysis (ICA) (McKeown, Makeig et al. 1998; McKeown and Radtke 2001). In the direct artifact subtraction approach, the average artifact signal is calculated over a short time-window and then retrospectively subtracted from the original biosignal (Allen, Josephs et al. 2000). The timing of the artifact with respect to the MRI acquisition is critical, therefore high temporal resolution sampling (i.e. 10 microsec) of the biosignal is required.

PCA and ICA are data-driven feature extraction and data reduction techniques that have been used to reduce fMRI-induced signal contamination in EEG data (Negishi, Abildgaard et al. 2004) and EMG (MacIntosh, Baker et al. 2007; van der Meer, Tijssen et al. 2010). In PCA, biosignal data are decomposed into orthogonal principal components that are ranked according to the amount they explain the total variance. In ICA, biosignal data are represented as linearly independent components (ICs), in what is sometimes referred to as blind source separation. Both PCA and ICA rely on redundancy across trials or across sampling sources to separate signals. Although useful, it can sometimes be challenging with these techniques to separate the components of interest (those containing biosignal) from those that contain artifact and noise.

7. Conclusion

Several components of an MRI system can affect the operation of ancillary devices for use during imaging. The RF transmit and receive fields, large static magnetic field, and time-varying gradient magnetic fields must all be considered in device design. Measuring electrical biosignals in an MRI environment poses a significant challenge as a consequence, but over the past decades numerous solutions have been proposed and successfully implemented. The scope of human brain mapping continues to grow with assistance from

an increasing number of peripheral devices that can be used during fMRI to make concurrent measurements. Viable products are now available that enable high quality, simultaneous fMRI, EEG, EMG, and EDA recordings.

This chapter illustrates that optoelectronics are a recurring engineering theme when it comes to measuring biosignals in the MRI environment. Conversion to optical signals, which can then be passed outside the magnet room via fibre optic cable and subsequently converted back to electrical signals in the console area with minimal loss or signal degradation, is of substantial benefit. Optoelectronics will, therefore, continue to play a key role in MRI-compatible hardware design.

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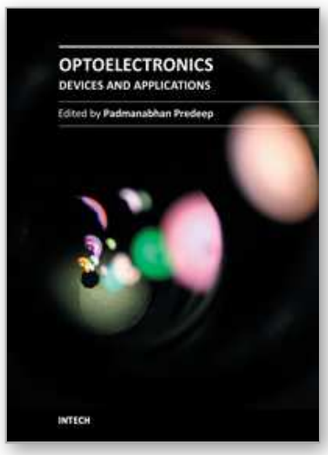

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