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Sensor Developments for Electrophysiological Monitoring in Healthcare

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1. Introduction

Recent years have seen a renewal of interest in the development of sensor systems which can be used to monitor electrophysiological signals in a number of different settings. These include clinical, outside of the clinical setting with the subject ambulatory and going about their daily lives, and over long periods. The primary impetus for this is the challenge of providing healthcare for the ageing population based on home health monitoring, telehealth and telemedicine. Another stimulus is the demand for life sign monitoring of critical personnel such as fire fighters and military combatants. A related area of interest which, whilst not in the category of healthcare, utilises many of the same approaches, is that of sports physiology for both professional athletes and for recreation. Clinical diagnosis of conditions in, for example, cardiology and neurology remain based on conventional sensors, using established electrodes and well understood electrode placements. However, the demands of long term health monitoring, rehabilitation support and assistive technology for the disabled and elderly are leading research groups such as ours towards novel sensors, wearable and wireless enabled systems and flexible sensor arrays.

All of these areas could, in principle, benefit from advances in telehealth and telerehabilitation (Winters, 2002). In the case of neurological disorders there is, for example, a recognized need for in-home and unsupervised rehabilitation (Johnson et al., 2007). The problems of taking data from individuals and interfacing with remote data networking systems are being solved by many workers and organizations, using a wide range of wireless communication technologies (see, for example, Dabiri et al., 2009; Ruffini et al. 2007). In addition, some of the interactive gaming technology is beginning to find a place in rehabilitation and as assistive technology for the physically disabled. However, there is still a need for the acquisition of high quality signals at the front-end, using sensors which are easy to attach, comfortable to wear for extended periods, and which provide reliable data under a wide range of operating conditions.

This chapter begins by considering the range of electrophysiological signals and the challenges of acquisition; introduces the application of electrophysiology to healthcare; briefly reviews the standard electrodes and sensors used and their limitations; and then considers in detail two specific areas of development which have particular relevance for the healthcare applications of interest.

2. Classes of electrophysiology signals

The human body generates a wide range of electrical potentials. Most of these arise through the firing of neurons, and associated electrical signal pathways, which respond to stimuli and control the body's physiological responses. These electrical potentials can be detected at the surface of the body and the measurements are classed according to the functional source of the potential. Thus, the electrocardiogram (ECG, or EKG) is a measure of the heart's electrical activity; the electroencephalogram (EEG) arises from the firing of neurons in the brain; the electromyogram (EMG) results from muscular activity; and the electroretinogram (ERG) records the electrical response of cells in the retina. The electrooculogram (EOG) has a rather different source, due to changes in the direction of the standing potential across the eye, and is a useful indicator of eyeball movement. Most of these measurements can be performed subcutaneously and, for clinical diagnosis, such invasive measurements give the most accurate and complete information. For example, in the clinical setting, needle EMG measurements provide the most accurate, spatially resolved measurements. However, their invasive character means that they are unsuitable for extended monitoring, particularly in non-clinical and unsupervised settings such as the home (Hogrel, 2005; Stegeman et al., 2000). In the context of long term monitoring of state of health, it is necessary for measurement techniques to be non-invasive and straightforward for non-specialist personnel to undertake. This dictates, among other things, that the signals be acquired from the surface of the body, or off-body, and this chapter is concerned solely with these types of measurement.

The ease with which surface potentials can be measured depends primarily on their amplitude and the spatial resolution with which they must be acquired. For a typical surface ECG measurement the peak deviation of the QRS complex is around 1-2 mV on the chest (V1-V5 lead positions) and 0.3-0.8 mV on the periphery (I-lead position) (Hampton, 1992). More challenging is the detection of the foetal ECG which is typically an order of magnitude smaller, of order 100 μ V at the surface of the mother's abdomen, and is difficult to deconvolute from the maternal ECG (Martens et al., 2007). The surface EEG also represents a weak signal, typically of order 10-100 μ V (Ruch & Fulton, 1960). The spatial resolution with which an electrophysiology measurement must be made also varies widely. For cardiac function, and with the most sensitive electrodes (see section 5.1.2), a good ECG can be acquired anywhere across the heart, from a pair of electrodes spaced across the chest to a pair located on opposite wrists. On the other hand, meaningful information about muscular activity must be obtained with closely spaced EMG electrodes. In the most advanced high density arrays, there may be multiple electrodes each of diameter, or length, 1-10 mm and spaced by 2-10 mm (Merletti et al., 2001). For neuroelectrophysiology, using arrays of EEG electrodes, the spacing is typically a few cm. However, a high density array may have 256 electrodes located over the surface of the scalp, in which case the accuracy of placement is similar to that for EMG.

All electrophysiology signals are at relatively low frequency, a bandwidth of up to a few hundred Hertz will usually suffice, so high frequency response is not a problem. If clinical quality information is to be obtained from surface electrodes, then the signal bandwidth must be sufficient to capture all of the notable features. Clinical standards have been set such as those prescribed by the American Heart Association for the ECG (Kligfield et al. 2007). In this case, a bandwidth of 50 mHz to 150 Hz is required for high quality adult ECG measurements and this may need to be extended to 250 Hz for infant ECGs. Interference

from other low frequency sources, including mains (line) noise, movement artefacts, and even competing surface potentials, does have to be addressed. This can determine the choice of electrode, as discussed in following sections, and will also dictate the signal conditioning and processing required in a given application.

All of the signals described above have been discussed in the context of measurements made on the surface of the body. However, with a sensor capable of detecting electrical potential or electric field, it is possible to detect a signal caused by body movements if the electrode is spaced off from the surface of the body. Although not in the same class of electrophysiology measurement as those discussed previously, this can provide valuable information. For example, the correlation of respiration, detected via chest wall movement, with heart rate variability is of interest for both the care of the elderly and in sports physiology. The detection of restricted limb movements could have application for assistive technology and human-machine interfacing. Finally, the straightforward detection of whole body movement may be of interest for monitoring single occupants of rooms in both a care home (Scanail et al., 2006) and a custody setting (Ebling & Thomas, 2008). The source of such a signal is the change in ambient electric field caused by the movement of a dielectric and conducting object, the human body, (Beardsmore-Rust et al., 2010) and the significance of such a signal as both a problem and an opportunity is discussed further in section 5.1.2.

3. Applications of electrophysiology to healthcare

The areas of healthcare which can benefit from electrophysiological measurements include; the monitoring of long term, chronic conditions; the rehabilitation of patients following trauma; and the facilitation of assistive technology for the disabled. Amongst the most familiar applications of electrophysiology is the monitoring of heart conditions through measurement of the surface ECG. Less familiar, perhaps, is the use of the surface EMG for the assessment of muscle function. However, it provides a suitable set of illustrations of these applications to healthcare and will be used here as an exemplar.

3.1 Monitoring chronic conditions

The role of electrophysiological measurement in this area is well illustrated by considering the assessment of muscle function. There is a significant amount of published research which shows that the measurement of the surface EMG provides useful clinical information for the assessment and evaluation of neuromuscular conditions. It is considered particularly useful for the assessment of motion (gait), multiple muscle groups, response times, tremors, chewing and breathing (Pullman et al., 2000). The type of useful data which may be obtained includes the firing rate and shape of muscle activation signals, changes in muscle fibre conduction velocity (MFCV), muscle responses to electrical or magnetic stimulation, and evidence of reinnervation zones in motor neurone disease. The timing and activation of contraction between different muscles is considered particularly useful for neurological studies (DeLuca, 2008). All of the above parameters can be used to track changes in the progress of disease due to pathology, intervention or ageing (Maathuis et al., 2008), provided a system is available which allows for long term monitoring of the surface EMG. In order to compete with the quality of measurements provided by the needle EMG, it has been necessary to demonstrate that surface EMG systems can provide high spatial resolution, overcome cross-talk from adjacent muscles and include techniques to deconvolute spatial filtering due to volume conduction (Pullman et al., 2000).

3.2 Rehabilitation

In the area of rehabilitation, for example, of neuromuscular conditions such as stroke, there is a great deal of interest in designing techniques which motivate patients to develop and strengthen the correct voluntary muscle movements. The surface EMG is capable of detecting isometric muscle signals, that is, ones which are too weak to result in muscle movement (Reaz et al., 2006; Saponas et al., 2008). In principle, it can be used to provide early indication of voluntary muscle activation and, with visual feedback, promote the correct patterns of activity. However, it is significant that, while surface EMG may be used in a clinical setting to evaluate protocols and assess the use of equipment, many rehabilitation exercises currently use other technologies for data collection, such as position sensors and video (Johnson et al., 2007). This is usually because surface EMG electrodes are considered to be too complicated and awkward to place correctly, attach securely and wear for extended periods of time. In summary, there is a clear need for surface EMG systems to be made more robust, reliable and user-friendly if their capabilities are to be realized for rehabilitation and these requirements apply equally to other applications of electrophysiology to rehabilitation.

3.3 Assistive technology

Assistive technology for the promotion of access for people with severe disabilities aims to utilize whatever limited control an individual possesses. Much effort has been put into exploiting any residual muscle signals for the control of assistive systems. In addition, this area of research crosses over into the two fields of electronically controlled prostheses and of human-computer interaction (HCI) for the gaming and virtual reality industries (Reddy & Gupta, 2007; Saponas, et al., 2008). Often the only muscles under the patient's control are in awkward locations (face, eyes, shoulders, single finger, wrist). Some workers are exploring the use of the EOG signal as an eye tracker for human-computer interaction (Bulling et al., 2009) whilst others are exploring the use of muscle signals for the control of prostheses (Garcia et al., 2007). The same limitations of surface EMG for rehabilitation apply here; the difficulty of attaching existing electrodes securely, of wearing them for extended periods and the limited spatial resolution of most systems. As a result, simpler technologies are often employed, such as active capacitive sensors for the eye and brow (Rantanen et al., 2010), piezoelectric sensors for the brow (Felzer & Nordmann, 2008) and force sensitive resistors for the forearm (Amft et al., 2006). However, new developments in electrode technology for surface EMG, which are beginning to address these issues, are described in section 5.

4. Electrophysiology sensors

The techniques used to detect electrophysiological signals from the surface of the body are well reviewed in the literature (see, for example, Prutchi & Norris, 2005; Searle and Kirkup, 2000). The following brief overview of standard approaches illustrates some of the challenges being addressed by the more recent developments discussed in section 5.

The detection of electrophysiological signals has, for many years, relied on the use of 'wet' silver-silver chloride (Ag/AgCl) transducing electrodes which convert ionic current on the skin surface to electronic currents for amplification and signal conditioning. Such electrodes are cheap and disposable (thus avoiding potential cross-contamination) but require the use of a conducting gel between the electrode and the skin. Changes in the electrochemistry

between the skin and the electrode do cause problems with baseline drift (DeLuca, 2008). Both this, and interference from the larger ECG signal, are a particular problem for surface EMG measurements (Allison et al., 2003). In addition, the use of the conducting gel is associated with problems of drying out, potential skin irritation, discomfort and shorting between adjacent electrodes in an array if not carefully placed. These problems make wet electrode systems unsuitable for use outside of the clinical environment and particularly unsuitable in terms of wearability for long term use.

A further complication is introduced by the development of high density electrode arrays, the most recent examples being for surface EMG mapping. A recent review editorial describes the difficulty of distributing gel to all of the electrodes such that it provides a high conductivity contact, whilst at the same time not shorting adjacent electrodes, and providing good contact stability during movement. It concludes that alternative, non-conducting electrodes could provide a better solution (Merletti, 2010).

A more user-friendly approach to surface measurements is to use 'dry' conducting electrodes, which make a resistive contact to the skin without the need for gel or paste (Chi et al., 2010; Searle & Kirkup, 2000). The electrode surface metal must be non-irritant, e.g. stainless steel. However, these electrodes still require careful skin preparation such as abrasion, suffer from changes in contact resistance due to sweating or skin creams, tend to be noisier than wet electrodes and can suffer from movement signal artefacts and charge sensitivity (Searle & Kirkup, 2000) if not very securely attached.

A different approach is to dispense with a DC, resistive coupling to the skin and instead couple to the surface potential capacitively, through a thin insulating layer (Clippingdale et al., 1991; Spinelli & Haberman, 2010). Because the signal fidelity does not rely on a good resistive contact, these electrodes do not necessitate skin preparation, or suffer from changes in contact resistance. As with dry electrodes, they can suffer from movement artefact and charge sensitivity (Searle & Kirkup, 2000). However, insulated electrode sensors are considered among the most promising for healthcare applications and their development is discussed in detail in section 5.1.

Much of the problem with noise is due to the impedance mismatch between the high source impedance and the following electronics. This is overcome by the use of active electrodes, commonly used for the detection of very low level EEG signals. These usually comprise an impedance buffer integrated into the wet or dry electrode structure. The disadvantage of active electrodes is that they can be bulky, require power to be supplied and, in most implementations, are not highly miniaturized for good spatial resolution. However, the use of high impedance, active electrodes is recommended for surface EMG (Clancy et al., 2002).

There has been considerable interest, driven by the demands of the applications described in section 3, in developing electrodes which can be deployed through, or embedded in, clothing. A number of workers have attempted to achieve this, using dry or insulating electrodes. For example, Kang et al. have designed both wet and dry, active electrodes fabricated using a nonwoven conductive fabric and flexible thick film circuitry which can be integrated into clothing (Kang et al. 2008). Measurements of the ECG from the torso, at rest and during exercise, indicate that the mounting is secure enough to minimize the movement artefact. However, the system will suffer from the same limitations as other wet and dry electrode structures and its durability remains to be proven. There have been attempts to realize non-contact electrodes by using dry electrodes to acquire electrophysiological signals through clothing; in all cases cotton-based to avoid electrostatic charging effects from man-made fabrics. The problem with this approach is that cotton is not a good, low-leakage

insulator (Chi et al., 2010), so the noise performance of the sensor is compromised. A better solution is to use active, insulated electrodes to couple through clothing and the performance of these types of sensor is reviewed in section 5.1.3.

Whichever type of electrode is chosen, they are usually employed as either bipolar pairs or in linear arrays. For example, in the case of surface EMG systems for neurological disorders, it is recommended that some form of array is used (Merletti et al., 2001, 2003). This allows for; the acquisition of signals from multiple sites over a muscle to aid placement decisions; improved spatial resolution; improved noise elimination, through the comparison of signals from different electrodes; and the deconvolution of cross-talk between interfering muscle fibres. Linear arrays are available commercially but developments of high density two-dimensional arrays are also underway and described in section 5.2.

5. Electrophysiology sensor developments

Recent developments in sensor technology have seen progress in a number of areas, all of which have the potential to allow the healthcare challenges described in the Introduction to be addressed. Two aspects in particular are discussed below; advances in non-invasive electrophysiology measurement offered by a particular type of active, insulated sensor; and advances in the development of high-density and flexible electrode arrays.

5.1 The Electric Potential Sensor

There has long been a recognition that the reliance on wet electrodes does not allow for ease of use and long-term monitoring. As a result, steady progress has been made in the development of dry and insulating electrodes. In particular, the performance of active, insulated electrode sensors depends on the high input impedance at the front end. This performance was given a boost over a decade ago by the introduction of the Electric Potential Sensor (EPS) (Clippingdale et al., 1991, 1994a, 1994b; R.J. Prance et al., 1998). This is a generic electric field measurement technology which has a wide range of applications beyond electrophysiology, including materials testing and characterisation (Gebrial et al., 2006a), imaging of static charge distributions for forensic applications (Watson et al., 2010c), detection of pressure induced voltages in rocks (Aydin et al., 2009) and electric field detection of nuclear magnetic resonance signals (R.J. Prance & Aydin, 2007). As an insulated, active electrode sensor, it requires no resistive contact with the source and relies on the displacement current through a capacitively coupled, thin, dielectric electrode coating.

The EPS is a good example of the cross-fertilisation of ideas from different branches of research. It was originally developed as part of the Sussex group's fundamental physics investigations into superconducting devices for quantum technologies (R.J. Prance et al., 1981; Skinner et al., 2010). There was a need to monitor the charging of a quantum regime weak link capacitor non-perturbatively, that is, without drawing real charge. The extreme experimental requirements precluded the use of a standard laboratory electrometer. The sensor needed to be located in a cryogenic chamber, at a considerable distance from room temperature. This, along with the necessity of non-perturbative monitoring, dictated an active, ultra-high input impedance sensor. In the event, the cryogenic version was not realized at that stage. However, the suitability of the design for room temperature electric potential measurements was quickly recognised and further development undertaken, leading to the first published implementation as a sensor of ECG signals both on-body and up to 5 cm off-body (Clippingdale et al., 1991).

5.1.1 Design of the Electric Potential Sensor

In order to produce a sensor with ultra-high input impedance, the design of the active front-end should be based on a high impedance amplifier configuration. In addition, a technique must be found to provide the DC bias required by the device, without compromising this input impedance. There are a number of methods used to achieve this, including the use of a low-leakage diode as the biasing component (R.J. Prance et al., 1998) or careful circuit board design which, through on-chip guarding and control of parasitics, precludes the need for a specified DC bias path (R.J. Prance et al., 2000). The resistance of the DC bias component has a significant effect on the output noise performance of the sensor, as illustrated in figure 1 which shows a continuous reduction in noise voltage measured up to resistances of order $10^{12} \Omega$. Well understood bootstrap and guarding circuit techniques (Graeme, 1973; Yeager & Hrusch-Tupta, n.d.) are used to maintain the effective input impedance as close as possible to the intrinsic device specifications. Whilst these techniques do not enhance the signal to noise, neither do they degrade it and they are crucial to retaining good low frequency operation, particularly in weakly coupled applications, as explained in due course.

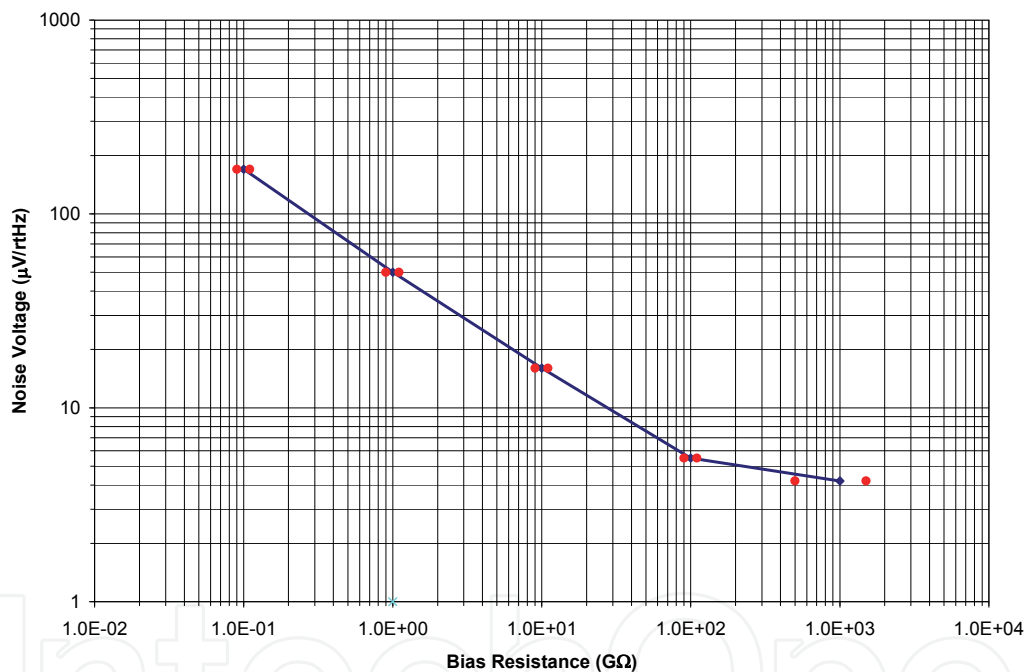


Fig. 1. Plot of input noise voltage at 1 Hz as a function of bias resistance value. Resistor tolerances, shown in red, are $\pm 10\%$ up to 100 G Ω and $\pm 50\%$ for 1 T Ω .

A generic block diagram of the EPS is shown in figure 2. It has been configured with a range of implementations and the details of the techniques used are covered by a suite of University of Sussex patents.

The combination of such a high input impedance, active front-end with an insulating electrode means that the EPS draws no real current from the source and can be thought of as approaching an 'ideal voltmeter'. Its performance as a non-perturbative detector of local electric fields has recently been verified by comparing measurements of the electric field between two large capacitor plates with finite-element simulations of the field (Aydin et al., 2010). This work demonstrates that the sensor causes no significant perturbation of the field and is capable of measuring fields as low as 2.6 $\mu\text{V}/\text{m}$ with 2% accuracy.

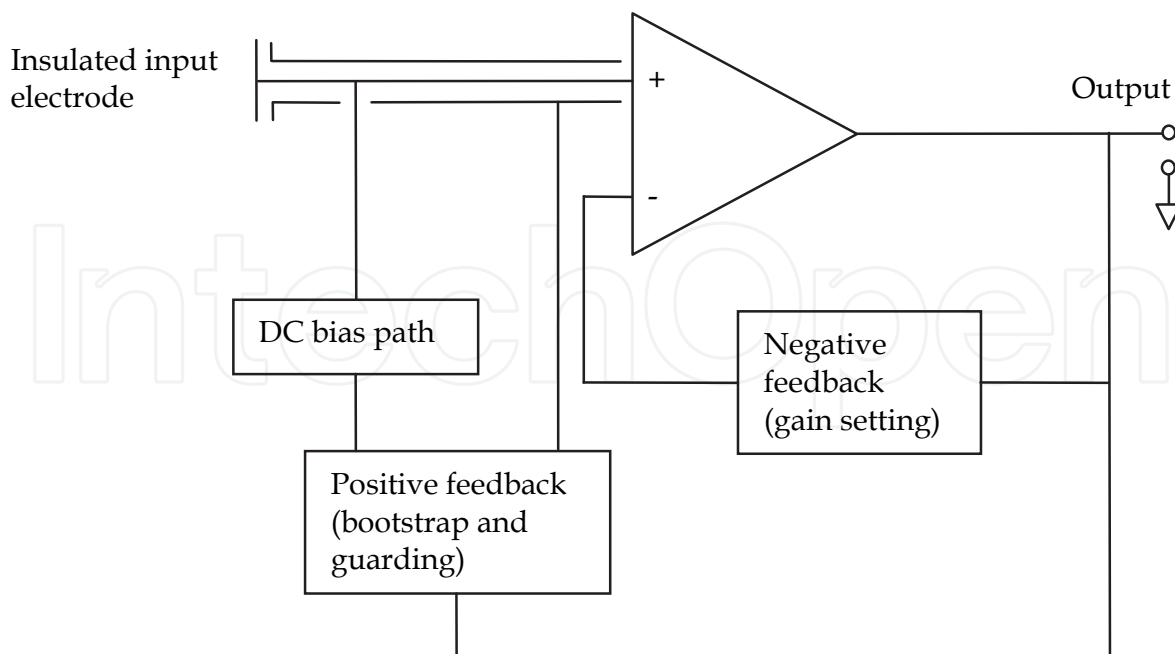


Fig. 2. Generic block diagram of the Electric Potential Sensor (EPS).

The standard limitation of high input impedance laboratory electrometers is that, with a true DC response, any charge present causes the input to drift to the power supply rail. Such electrometers therefore have to be reset repeatedly and are unsuitable for use in open, noisy environments. In addition, with an ambient atmospheric electric field of the order of 100 V/m present over the surface of the earth (Feynman et al., 1964), any movement of a sensor with a DC response will generate an associated AC movement artefact. The EPS is designed for both stability and usability. It overcomes the limitation of standard electrometers by dispensing with a true DC response and, in so doing, achieves excellent stability in standard working environments. Some commentators have perceived the lack of a true DC response as a limitation of insulated electrode sensors (Chi et al., 2010). However, it is worth noting that, in practice, almost every measurement made of voltage (potential) does not require knowledge of the true DC level. What is usually of interest at very low frequency is the variation, i.e. trend, in the signal. The EPS can be designed to provide any frequency response necessary for a given application, from the order of 30 mHz (R.J. Prance et al., 2000) to over 100 MHz (Gebrial et al., 2002), with either a broadband (R.J. Prance et al., 1998, Watson et al., 2010b) or tuned response (Clippingdale et al., 1994b), as required. The ability to set both upper and lower corner frequencies by design has a number of advantages. As would be expected, sensitivity to out of band noise is reduced by restricting the upper cut-off frequency. Restriction of the lower cut-off frequency minimises the DC drift problems as described above. In addition, the ability to choose the lower cut-off, by careful design of the bootstrap and guarding circuitry, is a benefit which has not been widely recognized. In particular, for weak coupled applications such as through clothing, it is essential to ensure that the appropriate standard operating bandwidth for a given electrophysiology application is retained. The design of the EPS manages to achieve this with no offset adjustments required at construction, set-up or in use. For example, for clinical quality adult ECGs, the operating bandwidth is typically 50 mHz to 150 Hz.

The physical embodiment of the EPS has developed over the years and been optimized for different applications. Figure 3 shows a typical pair of sensors and differential amplifier electronics used for on-body or through clothing electrophysiology measurements. The electrodes range from 1-5 cm in diameter and contain the active front-end electronics. The output is buffered to drive a long length of screened cable without sensitivity to mains (line) interference or cable movement. The metal electrode surface is coated with a thin, very low leakage dielectric and is guarded by a conducting ring. As an active sensor, the EPS electrode is not disposable by design. However, the electrodes can be completely potted in an inert compound which is inherently electrically safe and allows for sterilisation. They benefit from all of the advantages of insulated electrodes, requiring no skin preparation and being immune to changes in skin conductivity caused by sweating. For many electrophysiology applications, a differential measurement from a single pair of sensors is sufficient. The differential amplifier on the output is configured with minimal filtering, just a notch filter at the line frequency and a switchable low pass filter, as well as switchable gain. In practice the notch filter is not usually required due to the high quality differential measurement.



Fig. 3. Photograph of one embodiment of the EPS electrophysiology system, comprising a pair of insulating electrodes and a differential amplifier with analogue output.

5.1.2 Performance of the Electric Potential Sensor

Typical raw data, with no averaging or additional post-processing is shown in figure 4, which shows the electrocardiogram acquired from the wrists in an open, noisy environment. The performance of the sensor is such that comparable quality ECG signals can also be acquired through clothing, provided the material does not introduce electrostatic charge interference. Other workers have followed the EPS developments and implemented their own versions of active, insulated electrodes for through clothing electrophysiology. Sullivan et al., for example, use a circuit which incorporates bipolar transistor reset switches to combat the effect of the preamplifier input bias current drift (Sullivan et al., 2007). However, the insulation and reset circuitry are not low leakage and both the signal to noise and bandwidth of the measurement are compromised in this implementation. The performance of this, and other, competing electrode technologies are compared in section

5.1.3. Spinelli & Haberman followed the EPS developments very closely and recently independently verified that ECG signals comparable to those from contact wet electrodes could be acquired through clothing using high input impedance active, insulating electrodes (Spinelli & Haberman, 2010).

For insulated electrodes in contact with the skin, the EPS technology has been proven to acquire clinical quality signals and even the His bundle feature, usually only observed with catheter ECG electrodes (Harland et al., 2002b). For example, the ECG signal measured in a 10 mHz to 100 Hz bandwidth with a small, 4-element array of EP sensors placed on the chest has been shown to provide equivalent information to the conventional 7-lead ECG (Harland et al., 2005).

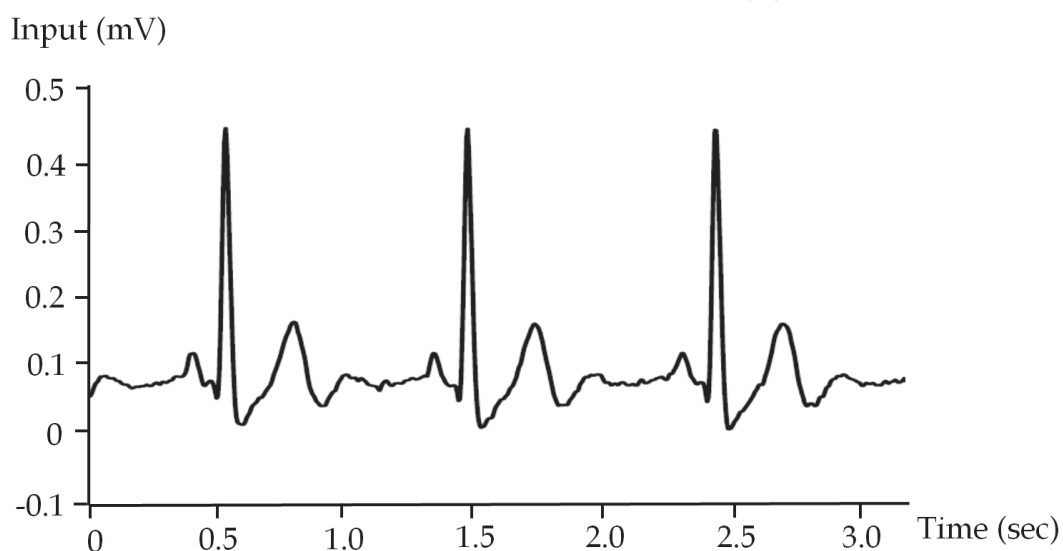


Fig. 4. I-lead ECG acquired from a differential pair of EPS electrodes positioned on the wrists. Signal acquired in real time, with no averaging, in a 0.5 - 30 Hz bandwidth.

The front-end electronics is low power and has been implemented with a wireless link, to monitor the ambulatory ECG, as a proof-of-principle (Harland et al., 2003a). The same demonstration shows that, provided the sensors are securely attached to the body, in this case by mounting each of the pair of electrodes in a wristwatch type fixture, the signal is not disrupted by strenuous movement.

The EPS has also been shown to have the sensitivity required for the weaker electrophysiology signals described in section 2. The alpha-blocking phenomenon in the EEG has been acquired through hair, and up to 3 mm off-body in a screened environment (Harland et al., 2002a). In addition EMG signals acquired from miniature sensors placed on the forearm show early indications of single motor unit action potential activity (H. Prance et al., 2009). In this case, three EPS electrodes; a closely spaced differential pair plus one mounted on the wrist; were used. By contrast, the EOG due to eyeball movement and blinking is of larger amplitude and can be acquired straightforwardly using a single pair of electrodes, with no reference, placed on the forehead (Harland et al., 2003b).

In the applications described above, the active electrodes are screened by physical contact with, or close proximity to, the surface of the body. They are therefore relatively immune to interference from external noise, including from the mains (line) supply, and what interference is picked up can be easily removed by the differential output amplifier.

However, if the electrodes are spaced off from the body in an unscreened environment, then the problems of interference must be addressed. It is not sufficient to limit the output bandwidth with a low pass filter, even if the application allows this. This is because large interference fluctuations at the input of the sensor may cause saturation. As a result, the signal of interest cannot be extracted satisfactorily by post-processing in either hardware or software. The EPS design has addressed this problem by enhancing the dynamic range of the sensor. One approach is to use an analogue comb filter to suppress the mains (line) frequency and its harmonics (H. Prance et al., 2007). An alternative solution uses high selectivity notch filters, digitally tuned in a smart sensor configuration to reject unwanted frequencies by up to 95 dB, thereby taking the level to below the intrinsic noise floor of the sensor (Beardsmore-Rust et al., 2009a; R.J. Prance et al., 2007, 2008). In both cases, the filters are incorporated into the feedback loop of the sensor, to reduce the front-end sensitivity at the unwanted frequencies and so enhance its dynamic range.

Whereas a previous design of EPS allowed the remote detection of electrophysiology signals in a screened room (Harland et al., 2002a, 2008); as a result of the developments described above, it has been possible to demonstrate the acquisition of both cardiac and respiration signals as a distance of up to 40 cm off-body, in an open, noisy laboratory environment (Beardsmore-Rust et al., 2009a; R.J. Prance et al., 2008). It is important to recognize that the ECG represents a body potential signal and, as such, cannot be measured by electrodes spaced by an air-gap from the body. This is demonstrated by comparing the phase of a cardiac electrical signal measured at the surface of the body, and at increasing distances off-body, with the phase of the arterial pulse, as measured by a pulse oximeter (Harland et al. 2002b). The signal at the surface of the body is, as expected, out of phase with the arterial pulse but it gradually moves into phase as the separation from the body increases. The cardiac signal acquired off-body will comprise a combination of electrical potential variation due to the electric field of the heart and a movement signal caused by the arterial pulse moving the chest wall. The movement signal will predominate at separations of over a few centimeters. The effects of body movement, and the phase dependence of the cardiac signals, are also observed, although the latter is not commented on, by Luna-Lozano & Pallas-Areny (Luna-Lozano & Pallas-Areny, 2010).

At significant sensor to body separations, relative movement between the two will result in an additional signal, as noted in section 2 and above, which is typically much larger than bioelectric signals. This can be a problem and secure attachment of electrodes is required to eliminate this source of interference. However, as explained in section 2, the ability to detect, and potentially track, the movement of persons does have application in home healthcare as well as security. An EPS sensor has been used to detect the movement of an individual through a wall (Beardsmore-Rust et al., 2009b; Harland et al., 2008) and a 4-element array of sensors was also used to track the movement of an individual within a 6 m x 6 m area (Beardsmore-Rust et al., 2011). The advantage of this, over competing techniques based on video or infra-red cameras or radar surveillance, is that the method is low bandwidth, low power and inherently passive, using the ambient electric field of the earth as the signal source.

5.1.3 Characterisation of the Electric Potential Sensor

We have termed the two extreme modes of application, one in physical, but not resistive, contact with the source and the other spaced off from the source; contact and remote modes respectively. The challenge for characterising sensor performance is that of

accurately defining a coupling capacitance which reflects that expected for each mode of operation. In the case of contact mode, coupling capacitances are typically from 1 nF down to around 1 pF, and these can be replicated using good quality lumped components. However, for remote mode, the coupling is significantly weaker because the self capacitance of the electrode structure, which couples to the local electric field, is typically of the order of 100 fF. Here, suitable lumped components are not available and instead custom-made, well characterized, guarded capacitors have been used to couple to the EPS in order to characterize its performance for remote and weakly coupled operation, including for microscopic imaging applications (Watson et al., 2010b). Figure 5 shows typical frequency responses and noise spectra for two extreme coupling scenarios (Harland et al., 2002b).

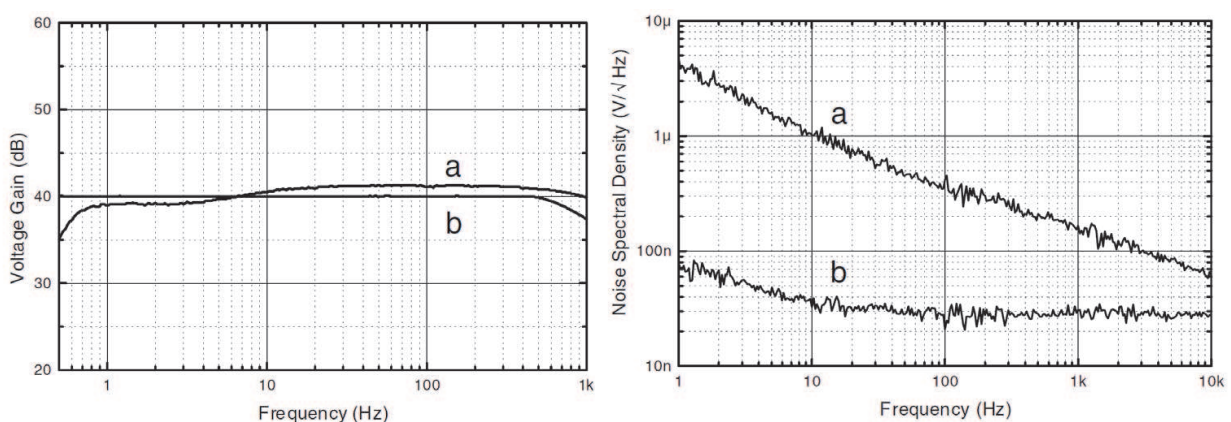


Fig. 5. (Reproduced with permission of IOP Publishing, from Harland et al., 2002b, figs 2 and 3). Frequency response (left) and noise spectra of an EPS sensor (right) for a) remote coupling through less than 1 pF and b) contact coupling through of order of 1 nF.

In the case of coupling through very small air-gaps, or through hair or clothing, the situation is much less well-defined. However, the sensor performance in these cases will lie between the two extremes depicted in Figure 5. Table 1 shows typical specifications for an EPS sensor for the three cases of; contact mode, for on-body electrophysiology measurements in contact with the skin; the weaker, through clothing, mode; and remote mode, for off-body applications such as the remote detection of life signs. These specifications are compared, where possible, to those presented by a number of authors who have reported active electrode systems similar to, or in competition with, the EPS.

5.2 High density and flexible electrode arrays

The use of arrays of electrophysiology sensors, for example for cardiac imaging, accurate EMG detection and EEG investigations, allows for the real time mapping of surface potentials. The increasing emphasis on non-invasive techniques for diagnosis and health monitoring has extended to this area and the use of large numbers of electrodes has also been aided by advances in data acquisition and processing. The advantages of body surface mapping in electrocardiography, for example, are that high spatial resolution images of surface potentials improve the early detection of abnormal activity (Lefebvre & Hoekstra, 2007). Similarly, high density arrays for surface EMG offer the capability of identifying and tracking single motor unit action potentials.

	EPS	Chi et al. 2010 dry	Chi et al. 2010 wet	Spinelli 2010	Sullivan et al. 2007	Oehler et al. 2008
Contact mode						
Input impedance	$10^{15} \Omega$, 1 pF	1.3 M Ω , 12 nF	350 k Ω , 25 nF			
Coupling capacitance	1 nF			100 pF	(0.1 mm spacing)	
Frequency response	10 mHz – 200 MHz			< 10 mHz – 500 kHz	3 – 70 Hz	200 mHz – 80 Hz
Input noise per root Hz at 1 Hz	70 nV	1.5 μ V	2.5 μ V	1 μ V	0.3 μ V (at 3 Hz)	20 μ V
Weaker mode, e.g. through clothing						
Input impedance	$10^{15} \Omega$, 10^{-13} F	305 M Ω , 34 pF				
Coupling capacitance	Between extremes of contact and remote modes	1 pF		10 pF	(3.2 mm spacing)	
Frequency response				~ 10 mHz – 200 kHz	4 – 60 Hz	200 mHz – 70 Hz
Input noise per root Hz at 1 Hz		2.7 μ V		10 μ V	2 μ V (at 3 Hz)	
Remote mode						
Input impedance	$10^{18} \Omega$, 10^{-16} F					
Coupling capacitance	0.4 pF					
Frequency response	10 mHz – 200 MHz					
Input noise per root Hz at 1 Hz	4 μ V					

Table 1. Comparison of selected electrophysiology sensor specifications under three different source-electrode coupling schemes. All electrodes, including the EPS, are active and insulating, except where stated.

The EPS has been used in array format since its earliest implementations. A 25-element array of spring coupled electrodes; mounted in a bench and designed to conform to the surface of the chest when laid upon; was used to image the electrical activity of the heart across the chest in real time (Clippingdale et al., 1994a). In a further development, a 7 cm diameter, 4-element array of EPS electrodes placed on the chest was used to recreate the conventional 7-lead ECG vectors (Harland et al., 2005). A similar approach was taken later by Oehler et al., who used a 15-element array of flexibly mounted, active, insulated electrodes with integral tablet PC to image chest cardiac activity in real time (Oehler et al., 2008). In this case, the diameter of the array was 185 mm and, with an input noise

performance of $20 \mu\text{V}/\text{rtHz}$ at 1 Hz, the patient had to be grounded to improve signal to noise. However, the detection of 15 ECG channels through cotton clothing was also demonstrated.

The potential of the EPS for use in high density arrays, for example for EMG detection, is illustrated by work done for materials and semiconductor sensing applications. This demonstrates that the sensors are highly scaleable, with spatial resolutions demonstrated down to $6 \mu\text{m}$ (Watson et al., 2010a, 2010c). In addition, the high input impedance and lack of a conducting interface to the source ensures that adjacent sensors in an array do not cross-couple to each other and can therefore be closely packed. Finally, the requirement to match sensor elements in an array, for straightforward data processing, is simply a matter of incorporating a small numerical correction factor in software to normalize the outputs of the array. In this way seamless data acquisition has been achieved without hardware adjustment or component selection (Gebrial et al., 2006b).

Large element arrays for electrocardiographic imaging, for example over 200 electrodes, have been developed (Ramanathan et al., 2004; Rosik et al., 2007). These are based on dry electrodes, for ease of use. Similarly, dry electrodes have been used in an attempt to make the imaging of brain activity, in this case for human-computer interfacing, simpler and more convenient than with the traditional, wet electrode, EEG cap array (Popescu et al., 2007).

Whilst cardiac imaging arrays can be relatively easily incorporated into a vest for chest mounting, the need to detect muscle activity in locations such as the jaw, for the assessment of swallowing (McKeown et al., 2002), or on curved limb surfaces is more challenging. A high density two-dimensional surface EMG array has been implemented using dry electrodes but in a very bulky system with uncomfortable, and somewhat painful, pointed electrodes (Blok et al., 2002, 2006; Huppertz et al., 1997; Rau & Disselhorst-Klug, 1997). Ideally, the array should be flexible and easily positioned and recent work has shown great progress in this direction. A flexible, high density two-dimensional array format based on wet electrodes has been developed for use on the face (Lapatki et al., 2003, 2004, 2010; Maathuis et al., 2008). In other work, the performance of flexible, washable dry electrodes for surface EMG was compared with conventional electrodes (Laferriere et al., 2010). This work concluded that, while the wet electrodes were lower noise, the dry electrodes used had sufficient sensitivity for mobility monitoring applications.

As yet, no-one has implemented an insulating electrode array for EMG. However, a pair of contactless electrodes, based on active, insulated sensors, has been used to acquire an EMG signal from the biceps through cotton (Gourmelon & Langereis, 2006). The noise performance is not optimum, at $10 \mu\text{V}/\text{rtHz}$ at 1 Hz, and the subject had to be grounded for through clothing measurements. The high spatial resolution acquisition of an EMG signal using three EPS electrodes, referred to in section 5.1, shows more promise and work is underway to implement a multi-element high density array for surface EMG investigations.

6. Conclusion

The development of novel electrophysiological sensors, based on active, insulated or dry electrodes, and of high density sensor arrays are both highly active research areas. This is driven by the needs of healthcare monitoring, including the demands of home health monitoring, telehealth and sports physiology, as well as the related fields of

human-computer interfacing and security. The acquisition of surface EMG signals, for the assessment of muscle activity, has been offered here as an exemplar to illustrate how electrophysiology measurements can be used to monitor chronic conditions, to support rehabilitation and to provide assistive technology for the elderly and disabled.

For these types of application, it has been noted that active, insulated electrode systems out-perform both wet and dry electrode systems in terms of signal to noise, immunity to interference and ease of use. They are able to offer all of the advantages of competing dry electrodes, active electrodes and electrode arrays but without the problems of skin preparation, skin irritation, motion artefact and cross-coupling between adjacent electrodes.

Having reviewed developments in active, insulated electrode sensor technology, it is clear that, by careful design, it is possible to retain the low noise performance and stable operation even in the weakly coupled limit of signal acquisition through clothing or an air gap. In addition, even in this coupling limit, it is possible to achieve the correct low, as well as high, frequency performance required for high quality electrophysiology measurements, as demonstrated by both the Prance group, with their EPS, and later by Spinelli et al.

Significant progress has been made in the development of high density, two-dimensional electrode arrays, particularly in the relatively new field of surface EMG imaging. However, whilst they are able to provide for a flexible array, wet electrodes are far from ideal for this application due to their inferior noise performance and susceptibility to cross-talk. Dry electrode high density arrays also have limitations in terms of the measures that are required to ensure good contact with the skin and it is likely that insulated electrodes will offer the best performance with the highest isolation between array elements and hence the highest spatial resolution. The ultimate goal is to develop flexible arrays of miniaturized, active, insulated electrodes, capable of high spatial resolution surface potential measurements. Such an array would find application in all of the areas of healthcare reviewed here, and particularly for both electrocardiographic and electromyographic imaging.

7. Acknowledgment

Support for research into the applications of the Electric Potential Sensor has been provided by the Engineering and Physical Sciences Research Council, UK (grant numbers GR/R87550/01 and EP/E042864/1). I am grateful for useful discussions with my colleagues Professor Robert Prance, Dr Christopher Harland and Philip Watson.

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Applied Biomedical Engineering

Edited by Dr. Gaetano Gargiulo

ISBN 978-953-307-256-2

Hard cover, 500 pages

Publisher InTech

Published online 23, August, 2011

Published in print edition August, 2011

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How to reference

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Helen Prance (2011). Sensor Developments for Electrophysiological Monitoring in Healthcare, Applied Biomedical Engineering, Dr. Gaetano Gargiulo (Ed.), ISBN: 978-953-307-256-2, InTech, Available from: <http://www.intechopen.com/books/applied-biomedical-engineering/sensor-developments-for-electrophysiological-monitoring-in-healthcare>

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