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Electric potential probes—new directions in the remote sensing of the human body

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Abstract
In this paper we describe a new approach to the detection of human body electrical activity which has been made possible by recent advances in ultra-low-noise, ultra-high-input-impedance probes. As we demonstrate, these probes, which do not require a real current conducting path in order to operate, can be used non-invasively both on and off body. We present remarkable new data showing the application of these probes to the remote, off-body, sensing of the electrical activity of the heart at distances of up to 1 m from the body and to high-resolution electrocardiograms. We suggest that in the future such probes may form the basis of a radically new technology for measuring the dynamics of the human body as well as in non-contact, imaging systems for pre-emptive and diagnostic medicine.

Keywords: body electrodynamics, human body sensors, electric potential probes, monitoring human-body electrical activity, non-invasive electrocardiogram

1. Introduction

In the last 20 years or so remarkable progress has been made in imaging the human body using a variety of techniques, including x-ray tomography, magnetic resonance imaging and ultrasound. Taking an overview, these techniques are focused on obtaining structural information about the body. This has, of course, been of very great consequence, both scientifically and medically. However, structure is only one aspect of the human body; dynamical function is of comparable importance. Such function can be revealed through the electrical (more accurately, the electromagnetic) activity of the body originating, for example, in the heart (electrocardiograms—ECGs) [1] and the brain (electroencephalograms—EEGs) [2]. In conventional practice electrical signals are detected using voltage probes in contact with the body. These probes, which have input impedances of 10^6 to 10^7 Ω, require real charge current contact to the surface of the body, this invariably being provided by an electrolytic paste. More precisely, silver metal electrodes are applied to the skin with adhesive pads and a silver chloride gel is used to act as an electrical transducer to convert the ionic current flow in the surface of the skin into an electron flow which can then be detected by an electronic amplifier [3]. The relatively low input impedance of these probes, combined with low sensitivity and the need for direct electrical contact, have been the limiting factors in their application to the sensing and imaging of electrical signals circulating in the human body. Furthermore, although in conventional practice input impedances of 10^6 to 10^7 Ω are considered high, these are still low enough to distort strongly the surface electrical potentials created by currents flowing in the body—the source of the electrical activity. These limitations have been recognized in the past and considerable efforts have been made to introduce alternative sensors with more advantageous performance characteristics. In particular, as a substitute for conventional electrical sensors, great efforts have been devoted to using SQUID magnetometer devices in body sensing [4]. SQUID based systems have great sensitivity and can be used non-contact, up to a few centimetres off body. However, they must be operated at cryogenic temperatures, they are expensive and, most importantly, they need extremely costly, magnetically shielded environments to function properly.

It has been clear from the experience gained with cryogenic SQUID magnetometers that off-body (remote with no electrical connections) sensing of electrical activity has great advantages, including the application to current source reconstruction. It has been our goal to achieve exactly this remote detection capability but with room temperature sensors. In this we have been successful using a new class of sensor—the ultra-high impedance electric potential sensor—which we have developed. In this paper we describe the recent progress in remote electric potential sensing as an alternative to both traditional contact electrodes (for ECGs and EEGs) and SQUID magnetometer systems. We show that these electrometer amplifier based sensors combine remarkable sensitivity with extremely high input impedance; sufficient in operation to allow the remote (non-contact) detection of electric potentials generated by currents flowing in the body. We will also demonstrate that these sensors need only modest electrostatic shielding even at the highest levels of sensitivity and are eminently scalable, i.e. can be used in large number arrays for non-invasive imaging of the body. By comparison with traditional (paste) electrodes for electrical sensing, the new sensors draw only a displacement current, not a real charge current, from the body and hence are intrinsically safe. Furthermore, with the input impedances (up to $\approx 10^{15}$ Ω at 1 Hz) and noise levels ($\approx$70 nV Hz$^{-1/2}$ at 1 Hz) achievable with these sensors, we now have the means to access and detect non-invasively any body electrical signal currently of interest, ranging from foetal heart signals through to EEGs of brain function.

A major aim of our work is to make large number ($\approx$1000 elements) arrays of non-invasive electric potential sensors practicable. With this spatial capability (resolution down to $\approx$0.5 cm), the inherent lack of cross-coupling between adjacent sensors, and the very low noise floors we are now achieving, we expect to be able to use these arrays for imaging electrical activity within, for example, the heart and the brain. By definition, such imaging (and source reconstruction) makes use of intrinsic sources within the body, i.e. it is probing the dynamical activity of the body. However, there is no reason why electric potential sensors (singly or in array format) should not be utilized to great advantage in detecting surface potentials generated by external sources of ac current injected into the body. This technique—electrical impedance tomography [5]—is well known and, to date, has relied on conventional (paste-on) electrodes. Non-contact arrays of electric potential sensors, with greatly enhanced sensitivity, could well facilitate this technique and should certainly be considered as a future area of application. In passing, we note that although we have not termed it electrical impedance tomography, we have been performing equivalent measurements on various metal structures in which ac currents (typically $\approx$ 10 to 100 Hz) have been injected. Thus, we have been able to produce two dimensional scanned images of stress regions in metal samples and corrosion defects inside metal pipes using non-contact electric potential sensors (to be published). Thus, this indicates that electrical impedance tomography of the body will work with these sensors. It is worth emphasizing that in general EIT and non-invasive electric potential sensing will yield different, and quite possibly complementary, information about the body.

2. Electric potential probes

Given that the basic technology associated with the measurement of body electrical signals is almost a century old, it is perhaps not unreasonable to look for innovation in this area, with the possibility of greatly enhanced performance. From the viewpoint of detecting this electrical activity, an ideal sensor would (1) draw no real charge current from the body (and hence be intrinsically safe), (2) have an extremely high input impedance (and thus operate as an almost perfect voltmeter), (3) have a very low noise floor, well below the smallest signal levels generated by the body, (4) be relatively low cost and (5) would appear to be perfectly biocompatible. As regards this last point, since these electric potential probes can either be used remotely, or make contact to the body surface through a completely bioneutral insulating interface, biocompatability is not a problem. This could clearly be of great advantage in many difficult situations (for example, in dealing with burns victims) where body electrical signals need to be monitored. We have now achieved all five of these objectives through our development of feedback enhanced and stabilized electrometer based amplifiers [6, 7]. These operate on displacement, not real, charge current, have input impedances of up to 10^{15} Ω and, with optimal coupling, a noise floor $\approx$70 nV Hz$^{-1/2}$ at 1 Hz. This unparalleled improvement in probe performance would appear to make the non-invasive detection of all known body electrical signals practicable. For example, we demonstrate that it is now possible to obtain electrical signatures from the human heart at up to 1 m from a fully clothed body with no electrical or mechanical contact. Furthermore, without direct electrical contact we can also record the very highest quality ECG at any point on the body surface, even from the fingertips. This is because these probes, with their remarkably high input impedance, present a negligible parallel load to the body. This is the essential point about the requirement (above) for a perfect voltmeter. Since this would not draw any current from the body, it would not, as a measurement device, disturb the intrinsic surface, and above surface, potential distributions due to the electrical sources within the body. This is precisely what should be aimed at in measuring body electrical activity—the sensor should not disturb the signal being detected. As an added advantage, since a perfect voltmeter (with infinite input impedance) does not load down the source, the amplitude of the signal detected is higher. In our experience this constitutes a serious problem when conventional (paste-on) electrodes, with relatively low input impedances ($\approx 10^6$ to 10^7 Ω), are used. Clearly, it is not possible to create a perfect (i.e. infinite input impedance) voltmeter. However, the electrical potential sensors developed by us, and discussed in this paper, are a good approximation for the case of body electrical detection. As such, we consider these devices to be a radical improvement on traditional electrode sensors.

These new sensors have been used in two electrically isolated configurations: a remote, off-body detection mode and an on-body contact detection mode. In the remote mode we use fixed sensor probe electrodes which form capacitive coupling to the body under measurement. In the contact mode we use single or multiple sensors with coupling of the signal made via appropriately designed (often hand held)

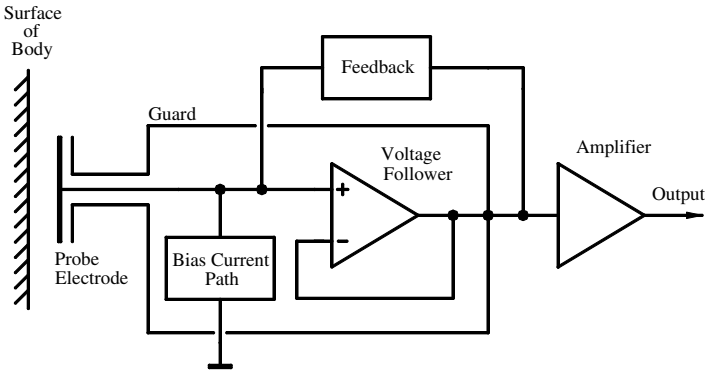

Figure 1. Block diagram of a typical electric potential sensor showing the probe electrode and the feedback, guard and input bias circuits of the electrometer amplifier. The probe electrode is generally a disc of 2 to 20 cm in diameter. This electric potential sensor is shown in the remote off-body mode of operation with no electrical contact to the body.

electrode–body interfaces. The development of the electronics required for these sensors has been discussed in detail in previous publications [6, 7]. The form of the electric potential sensor is shown in the block diagram in figure 1. Here, we provide a schematic of the sensor system, including the guarded sensor probe electrode, which typically has a diameter in the 2 to 20 cm range. To emphasize that no real current path is required for the sensor, we show schematically in figure 1 this electrode standing off body with no electrical contact to the body. As can be seen, we have combined an electrometer based amplifier with an appropriately designed probe to couple to the signal source. In essence, we have applied novel feedback techniques to a low noise electrometer amplifier so as to satisfy the bias requirements, while at the same time increasing the input impedance of the amplifier. This results both in minimum loading of the source by the sensor amplifier and in low-noise operation. Recently, we reported input referred noise levels as low as $\approx 2\ \mu\mathrm{V\ Hz}^{-1/2}$ at 1 Hz, achieved in designs for remote sensing applications [7]. In the latest generation of sensors we have measured input referred noise levels of $\approx 70\ \mathrm{nV\ Hz}^{-1/2}$ at 1 Hz for on-body mode operation. By this we mean physical contact through an insulating interface between the probe electrode and the body but with complete charge current isolation. This noise floor reduces to $\approx 30\ \mathrm{nV\ Hz}^{-1/2}$ at 100 Hz and above. Further discussion on the design and construction of the electric potential sensors described in this paper has been published recently elsewhere [7].

In this paper we are concerned both with the practical application of these probes to detecting electrical signals generated by the human body and with the presentation of the experimental data. Accordingly, the probes have been optimized separately to deal with two distinct situations. First, where the focus is on the remote detection of body electrical signals at relatively large distances from the body and, second, when it is sufficient to make contact (on-body, but electrically isolated) measurements of these signals. Typical voltage gain frequency responses of these two embodiments are shown in figure 2, where curve (a) is for the remote sensing application and curve (b) is for the contact mode. For comparison, we show in figure 3 noise spectral density plots, referred to the input, (a) for the remote case and (b) concomitant with the data of figure 2. Achieving a flat response over the frequency band

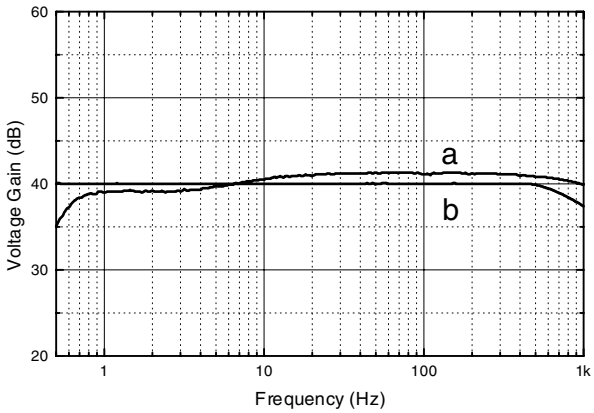

Figure 2. Typical measured frequency response plots for the electric potential probes for (a) remote off-body and (b) contact on-body sensing.

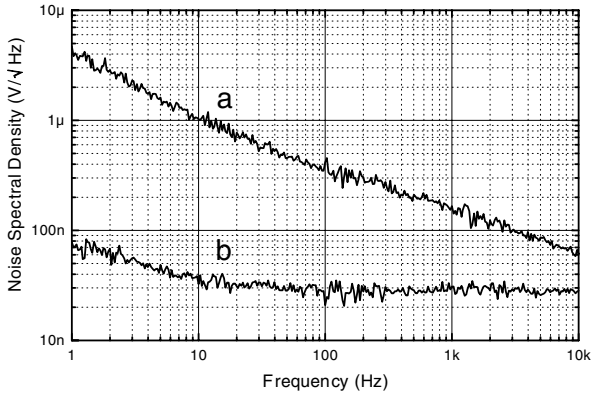

Figure 3. Typical measured input referred noise spectral density plots for the electric potential probes for (a) remote off-body and (b) contact on-body sensing.

of interest ($\approx$0.5 to $\approx$100 Hz) is relatively easy to maintain for contact probes (figure 2(b)) located on any part of the body. For example, direct comparison can be made with the different signals associated with a standard, conventional 12-lead ECG measurement system by simply moving a hand-held contact over the body. However, for the remote, off-body, mode, where there is very weak coupling, the design of the

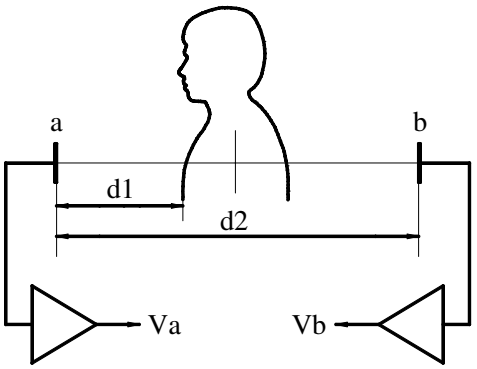

Figure 4. A diagram showing the relative positioning of the human body and the electric potential probe electrodes when used for remote off-body measurements. A single probe (a) is used for recording of a single-ended signal (V_a) with an electrode to body distance d_1. Two electrodes (a and b) are used for the recording of differential signals ($V_a - V_b$) with an electrode separation of distance d_2. No electrical (or earth) connections are made to the body.

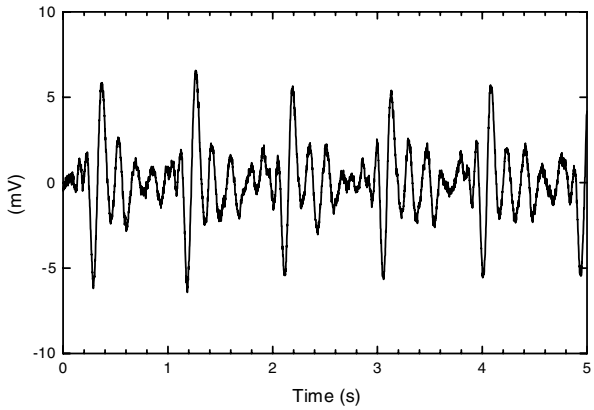

Figure 5. An example of the remote off-body detection of the human heartbeat. A differential measurement is shown where the body was positioned symmetrically between two electric potential probe electrodes placed 1 m apart (d_2 in figure 4) with no electrical connection made to the body (1 to 30 Hz bandwidth).

feedback and biasing circuits must be designed specifically to suit the signal source and body position. Furthermore, the different couplings associated with the two modes of operation also results in a different noise performance, as evidenced by the data of figure 3. As can be seen from figure 3(a), the typical noise floor of one of these new sensors, designed for a practical, remote off-body application, is $\approx 4\ \mu\text{V Hz}^{-1/2}$ at 1 Hz. This should be compared with the previously quoted noise level of $\approx 2\ \mu\text{V Hz}^{-1/2}$ at 1 Hz for a remote sensor under ideal conditions [7]. In the contact, but electrically isolated, mode typical noise floors are $\approx 70\ \text{nV Hz}^{-1/2}$ at 1 Hz reducing to $\approx 30\ \text{nV Hz}^{-1/2}$ at 100 Hz (figure 3(b)). Clearly, such a noise performance could be of great importance in, for example, the detection of the foetal heartbeat at the maternal body surface ($\approx 10\ \mu$V peak at a few Hz [1]) and in the sensing of the human body EEG (≈ 10 to 40 μV peak at a few Hz [2]).

3. Experimental results

3.1. Remote off-body sensing

As an illustration of remote off-body sensing of human electrical activity using these new electric potential probes we have made use of the electrode scheme depicted schematically in figure 4. Here, we show the relative positioning of the body and the electric potential probe electrodes when used for remote off-body measurements. A single probe (a) is used for the recording of a single-ended signal (V_a) with a electrode to body distance d_1. Two electrodes (a and b) are used for the recording of differential signals ($V_a - V_b$) with an electrode separation of d_2. We stress that no electrical (or earth) connections are made to the body. In figure 5 we show a differential measurement of the electrical potential signal generated by current flow in the heart where the electrode separation (d_2 in figure 4) is 1 m. This measurement, and the others discussed in the paper, were made in a simple, electrostatically shielded, room. The periodic pattern displayed does not exactly mimic the conventional on-body ECG because of the multi-polar nature of the heart as a current generator. This will lead to the multi-polar contributions to the observed electrical potential signal varying with distance from the body. However, the principal, and periodic, contribution in figure 5 is obvious. This is the major pulse which corresponds to the QRS complex of the ECG waveform, often termed the arterial pulse [8]. We emphasize that the data of figure 5 originates in the electrical activity of the heart and is not due to body motion. We also note that these data were recorded in real time (i.e. unprocessed) in a bandwidth of 1 to 30 Hz. In figure 6 we show the present limit of this remote sensing capability. In recording this data the sensor probe electrode was configured for a single-ended measurement (figure 4) and was positioned at a distance (d_1 in figure 4) of 1 m from the chest. Again, there is no electrical connection to the body and there is no earthing point on the body. The unprocessed waveform in figure 6(a) was made using a single-ended measurement in a 1 to 30 Hz bandwidth. The same signal is presented in figure 6(b) but now in a reduced bandwidth of 5 to 15 Hz so as to emphasize the main peak (QRS—arterial pulse) of the heartbeat. We note that the heartbeat of figure 6(b) is delayed in time compared with that of figure 6(a) due to the inherent phase change originating in the low frequency bandpass filter. For comparison, in figure 6(c) we show an $Sa\text{O}_2$ time reference signal for the waveforms of figures 6(a) and (b) derived from a commercial pulse oximeter [9]. As is standard practice, the pulse oximeter detector was clamped on a forefinger. Pulse oximetry was used as a time referencing device because it does not require electrical connection to the body in keeping with our off-body measurement techniques.

Signal to noise improves dramatically at much smaller distances from the body. This is illustrated in the waveforms of figure 7 which were all recorded using a single-ended measurement in a 1 to 30 Hz bandwidth and with the same electronic filter. For example, the waveform of figure 7(a) shows the same remote arterial pulse (heartbeat) as displayed in figure 6(a) (at 1 m) but now detected at a distance of 30 cm (see figure 4) and through two layers of clothing. Again, as for figure 6, an $Sa\text{O}_2$ time reference signal derived from a commercial pulse oximeter is shown in figure 7(b). It is apparent that the arterial pulse (related to the QRS complex of the ECG) in figure 7(a), and the $Sa\text{O}_2$ peaks of figure 7(b), are synchronized in time.

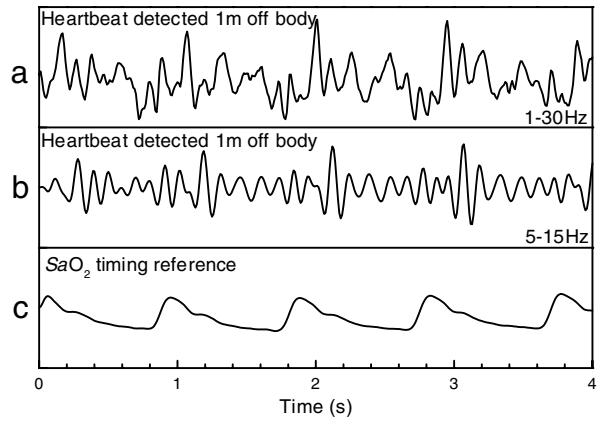

Figure 6. An example of the remote off-body detection of the human heartbeat using a single-ended mode measurement with a probe electrode to body separation (d_1 in figure 4) of 1 m. (a) The unprocessed human heartbeat waveform (1 to 30 Hz bandwidth), (b) the same signal as displayed in (a) but with a reduced bandwidth (5–15 Hz) to emphasize the main pulse, (c) an SaO_2 signal shown as a timing reference. See the text for a discussion.

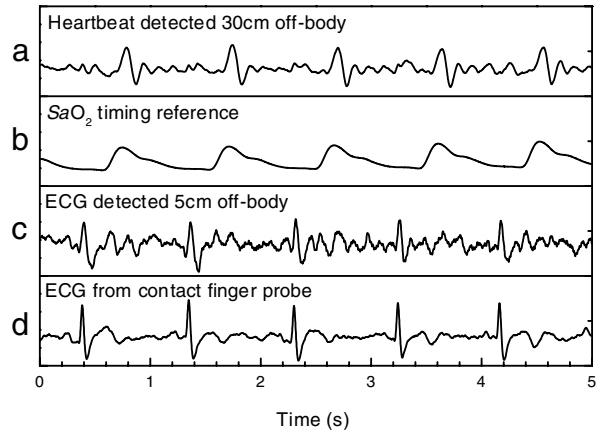

Figure 7. Examples of electrical activity detected from the human body (1 to 30 Hz bandwidth). (a) A heartbeat signal corresponding to the arterial pulse detected off body at a distance of 30 cm from the chest through clothing, (b) an SaO_2 signal shown as a timing reference, (c) an ECG detected off body at a distance of 5 cm from the chest, through clothing, (d) an ECG detected at the tip of the right-hand forefinger using a single, electrically isolated, contact probe. See the text for a discussion.

At small detector distances off-body (a few centimetres) the ECG can be detected. An example of an ECG monitored at 5 cm from the chest wall through two layers of clothing is provided in figure 7(c). For comparison we show in figure 7(d) the corresponding ECG detected using a single, hand-held, non-invasive finger-contact probe electrically isolated from the skin (see section 3.2).

When the ECG waveforms in figures 7(c) and (d) are compared in the time domain with the arterial pulse waveform in figure 7(a) a time delay is observed between the ECG and the arterial pulse. We suspect that this time difference is of considerable significance and may prove to have diagnostic potential in future measurement systems. Thus, the near (a few centimetres) and on-body waveforms (figures 7(c) and (d)) follow the voltage at the surface of the body. However, at larger

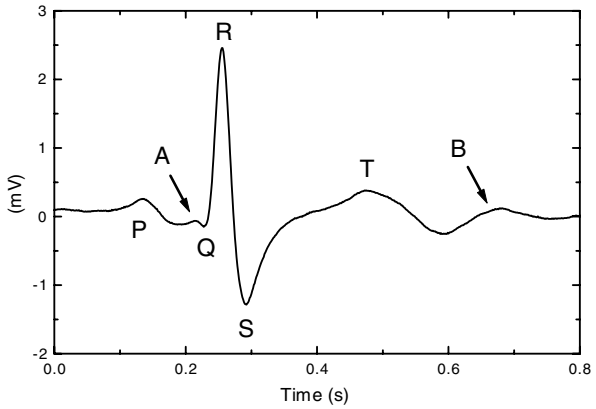

Figure 8. An example of a first-order, differential, high-resolution electrocardiogram recorded at the tips of the forefingers using a pair of electrically isolated contact probes (1 to 10 Hz bandwidth). In addition to showing clearly the P wave, QRS complex and T wave, the arrows A and B correspond in timing to the His–Perkinje discharge and the U wave, respectively.

distances (figure 7(a)) we are detecting changes in the electric potential which may originate in the multi-polar dynamical fields generated by the cardiac system. If this proves to be the case, it may be that these electric potential sensors can be used to map quite subtle aspects of body electrical activity.

3.2. Contact on-body sensing

The ECG of figure 7(d) was obtained by placing a single, hand-held probe onto the tip of one forefinger. However, the fidelity of a single probe ECG will inevitably be limited by electromyographic and other body noise sources [1]. These can be largely eliminated by using a differential signal from two probes, as evidenced by the high-resolution ECG presented in figure 8, which was obtained using two contact probes. For this we simply positioned the tips of the forefingers of each hand on two separate sensors and recorded the differential signal (left − right). The ECG shown in figure 8 is equivalent, in cardiology terminology, to the I lead, where the signal is derived from the difference between the two arm leads (LA − RA). In this case the I lead is derived from the difference between the left and right finger signals. The waveform displayed in figure 8 possesses the usual characteristics of a high-resolution ECG [1], namely the P wave, the QRS complex and the T wave. However, this pattern also contains features which are not usually seen in a conventional 12-lead ECG. Thus, in figure 8 we show (arrows A and B) events which correspond in timing to the positions of the His–Perkinje discharge (H) and the U wave. The time relation between the H peak and the atrial and ventricular depolarizations in the heart is an important diagnostic signature. In current cardiology practice the H peak is monitored using intracardiac techniques where the sensor is inserted into an artery via a cardiac catheter [1]. The His–Perkinje discharge is one of the low-level electrocardiographic potentials which is currently too small to be detected by routine measurement techniques and there may be great diagnostic value in its detection by surface sensors [10]. As might be expected, ECGs of comparable quality to that displayed in figure 8 can be obtained by making contact (but electrical

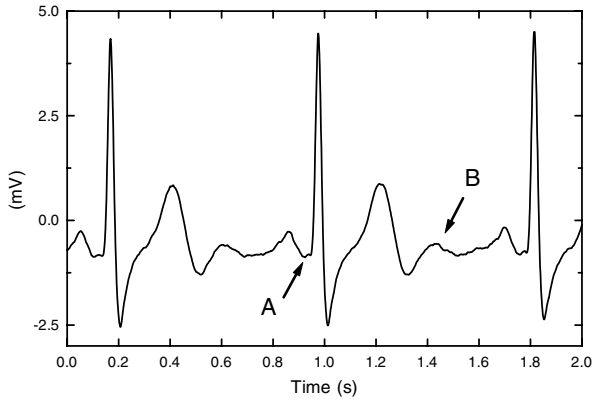

Figure 9. An example of a high resolution electrocardiogram (V4 lead) recorded from the chest (1 to 10 Hz bandwidth). In addition to showing clearly the P wave, QRS complex and T wave, the arrows A and B correspond in timing to the His–Perkinje discharge and the U wave, respectively.

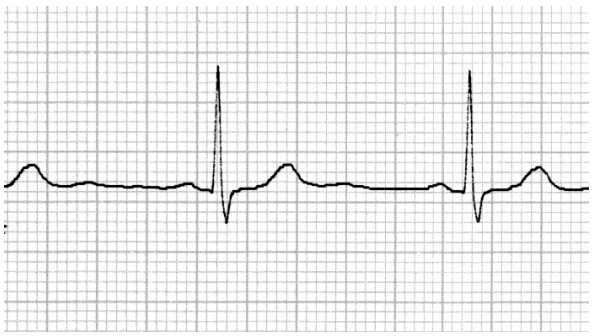

Figure 10. An example of a typical V4-lead ECG, as produced in a hospital cardiac unit, using standard paste-on electrodes and post-detection signal processing The waveform shows approximately 2.4 s of data.

isolated) measurements at the chest surface. In figure 9 we show a typical result where a sensor has been positioned on the chest approximately in the V4 position [1]. A second sensor has been positioned on the right-hand forefinger to provide a reference. The resultant differential signal (V4−RA) is shown. Again, as in figure 8, the structures A and B, corresponding in timing to the H and U features, can be seen. In common with all the data presented here, figures 8 and 9 show raw data without averaging or any other signal processing techniques having been applied (except in some cases limiting of the bandwidth). For comparison with the ECG waveform of figure 9, we show in figure 10 an example of a conventional V4-lead ECG, as produced in a hospital cardiac unit, using standard paste-on electrodes and post-detection signal processing.

The examples in this section have been provided to demonstrate the ease with which ultra-high-input impedance electric potential sensors can be applied to the monitoring of familiar electrical signals from the body. Furthermore, with no need for electrical contact, and therefore with negligible loading by the sensor, there is essentially no reduction in the signal detected. This is evidenced by the data of figures 8 and 9, where it appears that the resolution of the waveform is high enough to reveal detailed features of the ECG, i.e. structures which may be associated with the His–Perkinje discharge and the U wave. The fact that these sensors can be positioned anywhere on the body within seconds, with absolutely no preparation required, and pick up a whole variety of body electrical signals at signal to noise ratios exceeding conventional surface mount (paste-on) detectors, must clearly be considered advantageous. Indeed, these signals can be picked up through clothing (or hair) and remotely from the body. We emphasize that these developments have come about because we have taken a different approach to the detection of electrical signals generated by the body. For these signals we have tried to create a good approximation to the perfect voltmeter (infinite input impedance). This has freed us from the difficulties and constraints imposed when conventional electrode detectors are used, including the problem of making good electrical contact to the body, the loading down of the body leading to a reduction in signal amplitude and the distortion of surface body potentials.

4. Conclusions

The sensitivity of these new electric potential probes points to two immediate areas of application. First, since we can now sense the human arterial pulse remotely, it should now be possible to perform cardiotachometry (the measurement of the heart rate) non-invasively at distances up to 1 m from a clothed body. Second, using the hand-held contact probe fast electrocardiography can be performed with the convenience of a movable sensor which can be moved and positioned at any point on the body without the need for sticky pads and gels. In conventional practice the use of paste-on electrodes has several disadvantages, each of which can be eliminated by substituting ultra-high-impedance electric potential sensors. The most important compared with conventional electrodes are the input impedance and the very low noise floor. The impedance level (up to 10^{15} Ω at 1 Hz) means that the sources of body electrical signals are not loaded down by the sensor. In turn, this means that there is negligible attenuation of these signals. This is not the case with conventional paste-on sensors, where input impedances are in the 10^6 to 10^7 Ω range, resulting in a substantial fraction of the signal being lost. Again, the very low noise floor ($\approx$70 nV Hz$^{-1/2}$ at 1 Hz with optimal coupling) allows for the detection of a wide range of body electrical signals at signal to noise ratios higher than can be obtained with conventional detectors. In addition, the simple convenience of not having to resort to electrolyte paste-on methods, which are time consuming (particularly in array formats), and have biocompatability and cross-talk problems, should prove to be a considerable plus in clinical (and research) practice. In our opinion, even these limitations of conventional ECG/EEG electrodes would appear to point to the widespread use of electric potential sensors in the relatively near future.

As is very well known the human body contains many different sources of electrical activity. Since these sources are in general multipolar in nature, they will generate spatially complex field distributions (both electric and magnetic) around the body. Conversely, mapping these distributions as a function of distance off body provides information concerning the sources themselves, sometimes leading to partial or full reconstruction of particular sources (the inverse problem). Clearly, in order to map such distributions the probes used

must be sensitive enough to detect these fields at a distance. To date this has been achieved to a limited degree using cryogenic SQUID magnetometers. As is apparent from the data presented in figures 5 and 6, this mapping can now be accomplished at room temperature, and over a larger range of distances (to date, up to 1 m), with the new electrometer based probes described in this paper. There seems little doubt that in the future probe sensitivity will be improved sufficiently for such measurements to be made even further off body, if required. We note that, as with conventional 12-lead cardiology data, the details of the ECG waveforms taken close to (or at) the body using electric potential sensors change with position. However, we have also observed that changes are also seen with variation in distance from the body, as is to be expected if the body electrical sources being probed are dipolar or multipolar in nature.

It seems clear that the developments described here have opened up a new approach to the exploration of the dynamical behaviour of the human body, which we term body electrodynamics (BE). These developments have their origin in a new class of electric field sensor of remarkable sensitivity. This, together with ease of use, relatively low cost compared with magnetic sensors and intrinsic lack of cross-coupling, would appear to make high-resolution (i.e. very large number of elements) imaging arrays for heart [11] and brain scanning a very distinct possibility. Anticipated reductions in the noise floor of these systems over the next few years, quite probably to $\approx$10 nV Hz$^{-1/2}$ at 1 Hz, will further enhance performance. Given these capabilities, and the low cost of any (electrostatic) environmental shielding that might be required, it seems reasonable to assume that the complete spectrum of human body electrical activity will become accessible over the next few years.

Acknowledgments

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