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INTERFERENCE REDUCTION IN ECG RECORDINGS BY USING A DUAL GROUND ELECTRODE

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Abstract – Power distribution lines are a major source of interference in ECG recordings obtained with surface electrodes. Displacement currents coupled to electrodes, their leads and directly into the body, and stray capacitance in power supply transformers, result into differential-mode, common-mode and isolation-mode interference. Shielding electrode leads reduces displacement currents coupled to them. Amplifiers with high common-mode and isolation-mode rejection ratio reduce common-mode and isolation-mode interference, but interference coupled in differential mode cannot be reduced by shielding neither by amplifier design. We propose the use of a grounded electrode close to each recording electrode instead of the customary "right-leg" ground electrode, to reduce differential-mode interference. We provide a qualitative model to explain that interference reduction, and experimental evidence to show its effectiveness even for non-isolated recording systems in the presence of strong power-line interference.

Keywords: ECG recording, power line interference, differential mode interference.

1. INTRODUCTION

Electric power lines have been a major source of interference in surface ECG recordings since electronic amplifiers (i.e. high-impedance devices) were first applied to electrophysiological studies back in the 1930s. Huhta and Webster [1] identified the four basic ways by which interference can enter ECG recordings obtained by an earth-grounded amplifier: a) magnetic induction, b) displacement currents into the electrode leads, c) displacement currents into the body, and d) conversion from common-mode into differential-mode interference at the input of the amplifier. The displacement current into the body flows to ground through the ground (right leg) electrode and "*because the body has finite impedance, the displacement currents entering the body through the arms, legs, and torso will cause different parts of the body to be at slightly different potentials.*" [1]. Therefore, the difference between the 50 Hz (or 60 Hz) potential of the two recording electrodes will be amplified the same as the ECG signal. To reduce that interference component, Huhta and Webster suggested moving the ground electrode to a different location on the body until the potential of the two recording electrodes with respect to ground were equal. This is certainly no practical, but explains, for example, why power-line interference is reduced when the ECG is recorded between one finger from each hand when a close finger is grounded (Fig. 12 in [1]). That interference analysis also explains why two-electrode grounded amplifiers require a higher common-mode rejection ratio (CMRR) and higher common-mode impedance (Z_C) than three-electrode amplifiers to reduce interference to comparable levels [2].

When using three electrodes, amplifiers supplied by batteries or by isolated power supplies (whose signal ground is independent from earth ground), reduce the common mode voltage with respect to signal ground [3]. Two-electrode amplifiers have a larger common-mode voltage with respect to earth ground. However, the rejection of that voltage, properly termed isolation-mode rejection ratio, is usually very high for three- and two-electrode systems [4].

Metting van Rijn *et al.* [5] showed that in isolation amplifiers there is also a displacement current coupled to signal (amplifier) ground through the power supply transformer. Displacement currents coupled to the electrodes also contribute to power line interference [6]. As a result, shielding cables and increasing the CMRR and IMRR is not necessarily enough to obtain high quality ECG recordings. Direct differential-mode interference is always present in biopotential amplifiers no matter how high is the CMRR and how small is the common-mode voltage [7]. Two-electrode methods based on body potential driving [8] [9] do not solve this problem either.

In this work we propose to use a ground electrode close to each recording electrode to reduce differential-mode interference. All these ground electrodes are connected to the amplifier ground (signal ground). This solution does not require any modification in amplifier circuits neither any further signal processing.

2. A MODEL FOR POWER LINE INTERFERENCE

Fig. 1 shows a common circuit model to describe power-line interference coupling in a three-electrode isolated amplifier for biopotential recordings [6]. C_p, C_b, C_c and C_e

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are stray capacitances from power lines to patient, patient to earth ground, power lines to electrode leads, and power lines to electrode, respectively. Z_{t1}, Z_{t2} and Z_{t3} are internal body impedances; Z_{e1}, Z_{e2}, and Z_{e3} are electrode-skin impedances; Z_I is the isolation impedance between amplifier ground (input reference terminal, signal ground) and earth ground; Z_C and Z_D are the common-mode and differential-mode input impedances of the amplifier, respectively.

Capacitive coupling results into displacement currents into the cables and electrodes (i_L) and into the patient (i_p). Because Z_D, Z_C, and the isolation impedance Z_I are very large, it is usually assumed that most of i_p flows to earth ground via C_b; if the impedance of C_b is Z_b, only a fraction $Z_b/(Z_b + Z_I)$ of i_p will flow along Z_{e3}, thus resulting in a small common-mode voltage. Further, this model assumes that part of i_p will flow along Z_{t1}, which results into differential-mode interference. If the limited CMRR of the amplifier is considered, but otherwise the IMRR is large enough, the equivalent input interference voltage due to capacitive coupling between power lines and the patient will be

$$v_{id} = i_p \left[Z_{t1} + \frac{Z_b}{Z_b + Z_I} \left(\frac{Z_{e2} - Z_{e1} + Z_{t1}}{Z_C / Z_{e3}} + \frac{Z_{e3}}{\text{CMRR}} \right) \right] \quad (1)$$

where v_{id} is the equivalent differential-mode interference at the input of the amplifier, and CMRR is the common-mode rejection ratio of the amplifier, which can be designed to be very large. Eq. (1) describes a worst-case condition where it is assumed that all i_p flows along Z_{t1}.

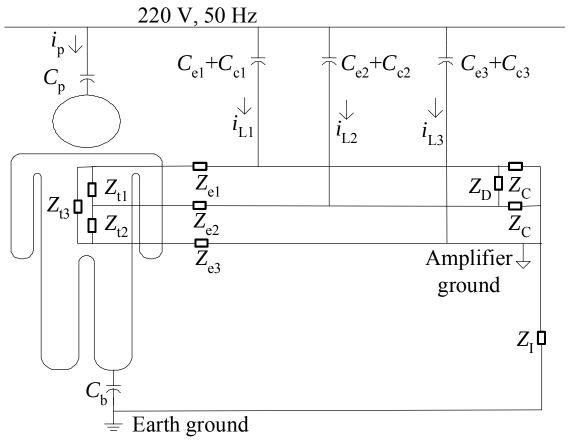

Fig. 1. Common lumped-parameter model to describe power line interference in a three-electrode system to record biopotentials.

The largest differential-mode interference is normally assumed to be contributed by displacement currents coupled into the body and into the electrodes due to both the imbalance between the impedance of the recording electrodes ($Z_{e2} - Z_{e1}$), and the limited common-mode input impedance of the amplifier. Adli *et al.* [10] proposed a system for balancing the skin-electrode impedance in order to reduce power line interference; however, this approach cannot reduce differential-mode interference due to Z_{t1}.

Substituting maximal values for $i_p = 1\ \mu$A and $Z_{t1} = 500\ \Omega$ in (1), we obtain $i_pZ_{t1} = 0{,}5$ mV, which is of the same order of magnitude as the ECG. If we assume a limited amplifier isolation ($Z_I = Z_b$), a large electrode imbalance $Z_{e2} - Z_{e1} = 100$ kΩ, a ground electrode with a poor contact ($Z_{e3} = 100$ kΩ), $Z_C/Z_{e3} = 1000$ and CMRR = 80 dB, which is a moderate value, we would obtain an additional contribution to v_{id} of 0,05 mV, hence much smaller than the differential-mode interference i_pZ_{t1}. For the capacitively coupled current between the primary and secondary of the power supply transformer to yield 0,5 mV across Z_{e3}, it should be 5 μA. A medical-grade isolated power supply would seldom inject that much current to flow to earth ground via Z_3 and Z_b. Anyway, that 0,5 mV drop in voltage across Z_{e3} would be further attenuated by the CMRR. In summary, in a carefully designed system, i_pZ_{t1} can often be the main contribution to power line interference.

It turns out, however, that modelling capacitive coupling between power lines and the body as a single capacitance as shown in Fig. 1 oversimplifies the problem. Displacement current will enter the whole body as shown in Fig. 2. As a result, the actual power line current between the recording points will depend on their 50 Hz (or 60 Hz) potential with respect to earth ground. If they were on the same equipotential line, no power line current would flow between them. Therefore, when using three electrodes, the two recording electrodes should be “symmetrically” placed with respect to the amplifier ground electrode, as suggested in [1] and [7]. An alternative and much easier method to ensure that the recording electrodes are on an equipotential line is to place an amplifier ground electrode very close to each of the recording electrodes (Fig. 2), instead of using a single ground electrode on the right leg, or any other position. If the currents to ground through each electrode are the same, the power line potential at each recording electrode will be about the same too.

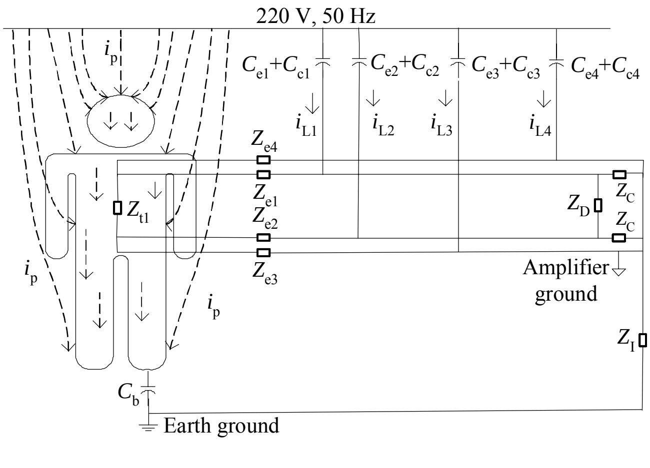

Fig. 2. Model to describe power-line interference in a dual ground electrode system to record biopotentials when displacement current from power lines is considered to couple to the whole body.

3. EXPERIMENTAL DESIGN

We have designed an earth-grounded ECG amplifier ($Z_I = 0$) with a CMRR large enough for the resulting interference from the common-mode voltage to be negligible according to (1). Earth-grounded amplifiers yield larger common-mode interference than isolation amplifiers and, although unsafe for medical use, they are the most challenging situation for interference reduction. Electrodes were connected to the amplifier by coaxial cables whose shield was connected to signal ground, hence earth ground in this case. We successively recorded leads I and II of the ECG, first when using three electrodes (i.e., a single ground electrode in the right leg) and then when using a ground electrode close to each recording electrode in the wrists or ankle (and no right-leg electrode). The use of leads with distant electrodes was intended to show how a dissimilar distance from each recording electrode to power lines affected the efficacy of the use of a ground electrode close to each recording electrode. All electrodes were pre-gelled and disposable. Lead II was also recorded when a power line cord was close to the subject. Finally we recorded lead III using single and dual ground electrodes in normal interference condition an also when a power line cord was close to the subject. The measurements for increased interference condition and lead III were repeated after placing a 100 kΩ resistor in series with one of the recording electrodes to imbalance the impedance. The presence of power line interference in the recorded signal was visually assessed from the time record and from its power spectral density (PSD) obtained by MATLAB version 7.1 and the signal processing toolbox employing Welch's method and a Hamming window.

3.1. ECG amplifier design

Fig. 3 shows the ECG amplifier and subsequent low-pass filter. The circuit was supplied from an earth-grounded power supply, and therefore signal ground was connected to earth ground. The input ac coupling network was designed according to [11] for a corner frequency of 0,05 Hz and fulfils the low-pass frequency response of IEC standards [12]. The gain for the instrumentation amplifier (INA111) was 1000 and the corner frequency of the (second-order) low-pass filter was 100 Hz. The CMRR at 50 Hz was 103 dB for balanced electrodes and 60 dB for a 50 kΩ imbalance between the two recording electrodes. The ECG was recorded using a 12 bit data acquisition module (EAGLE USB μDAQ) at 1 kHz sampling frequency controlled by a program implemented in LabView®.

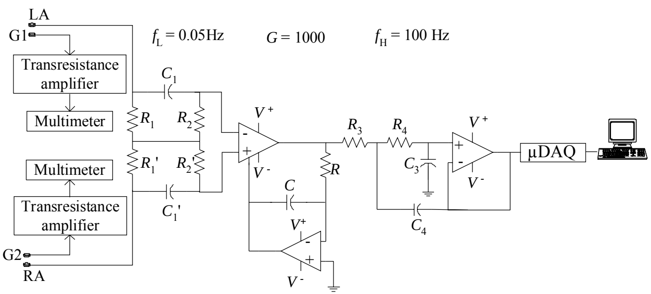

Fig. 3. ECG signal acquisition system that can use one or two electrodes (G1, G2) connected to amplifier ground.

3.2. Displacement current monitoring

Power line current leaving the body was monitored by connecting a transresistance amplifier (Fig. 4) in series with each ground electrode. We selected $R_o = 1$ MΩ, 0,1 % tolerance. Signal bandwidth was limited to 100 Hz by C_o. v_o was simultaneously measured for each ground electrode with two 6 1/2 digit digital multimeters (Keithley 2100 and 2700), whose maximal uncertainty for the measured voltages was ± 2 mV. Therefore, when using two ground electrodes, any difference between the two currents to ground much larger than ±2 mV/1 MΩ = ±2 nA can be attributed to a current imbalance that will result in residual differential-mode interference.

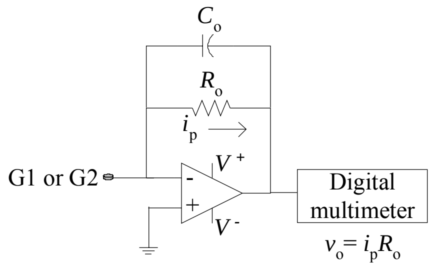

Fig. 4. Transresistance amplifier to measure power-line currents to ground through the subject.

4. RESULTS AND DISCUSSION

Figs. 5 and 6 show the ECG for, respectively, leads I and II recorded under common interference conditions with one and two ground electrodes, as well as their corresponding power spectral density. The intensity of the power line current through the ground electrode(s) in each recording is also shown.

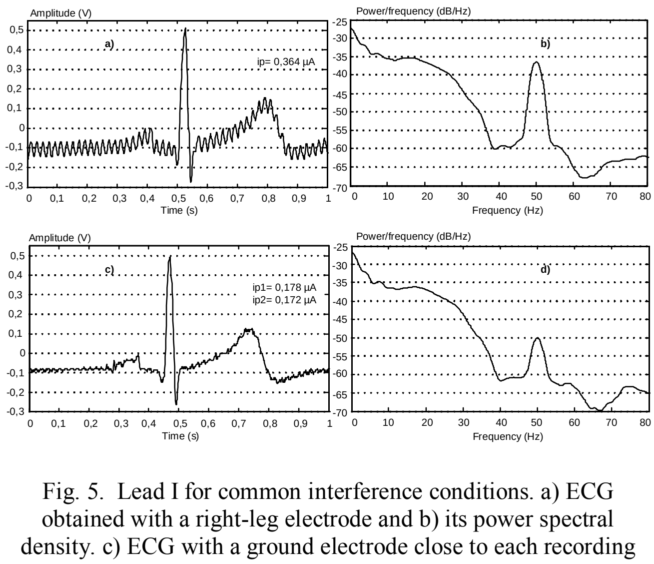

Fig. 5. Lead I for common interference conditions. a) ECG obtained with a right-leg electrode and b) its power spectral density. c) ECG with a ground electrode close to each recording electrode and d) its power spectral density.

For lead I recorded with a single ground electrode (right leg), $i_p = 0{,}364$ μA and Figs. 5a and 5b show a large interference, about 20 % of the ECG peak value, in spite of the large CMRR of the amplifier. When using a ground electrode close to each recording electrode (right and left wrists), and no right-leg electrode, i_p splits into $i_{p1} = 0{,}178$ μA and $i_{p2} = 0{,}172$ μA, the interference decreases by more than 15 dB (Fig. 5d) and is hardly noticeable in the time record (Fig. 5c). Note that $i_p \approx i_{p1} + i_{p2}$ and $|i_{p1} - i_{p2}| = 0{,}006$ µA. The reduced interference when using two ground electrodes cannot be explained from an improved effective CMRR as the ground electrode impedances were small in both cases and the CMRR for the amplifier alone was 103 dB. The balanced current path to ground for power line displacement currents entering the body can explain the improvement because they make both recording electrodes to be close to an equipotential curve.

For lead II and an amplifier with a single ground electrode, i_p increased by about 25 % to 0,450 µA, in spite of the recording being obtained for the same subject and in the same place, day and position. Therefore, for an earth-grounded amplifier, i_p in (1) depends on the lead recorded, as currents coupled to cable shields flow to earth trough ground electrode(s). Simultaneous with the increase in i_p as compared to that for lead I, the interference in the ECG signal almost doubled (Figs. 6a and 6b). This shows again that a lumped parameter model to describe power line current coupled to the subject is a gross model, because the observed interference is not proportional to i_p. Differential-mode interference depends on the path followed by the displacement current coupled to the body, not only on its amplitude. When using a ground electrode close to each recording electrode (right wrist and left leg), and no right-leg electrode, $i_{p1} = 0{,}24$ µA and $i_{p2} = 0{,}25$ µA. The interference decreased (Figs. 6c and 6d), but the remaining 50 Hz voltage was a bit larger than in lead I, as can be expected from the larger imbalance between ground currents for lead II.

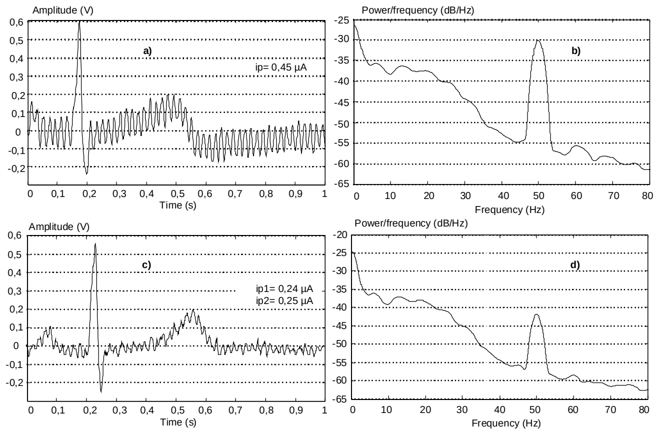

Fig. 6. Lead II for common interference conditions. a) ECG with a single ground electrode and b) its power spectral density. c) ECG with dual ground electrodes and d) its power spectral density.

When a power line cord was placed on the back of the subject, i_p for the single-ground electrode amplifier increased to 3,7 µA, about 8 times that in normal interference condition. The interference observed in lead II also increased (Figs. 7a and 7b), but was only about twice the previous value, thus confirming that the current path can be more significant than the value for i_p. When using two ground electrodes $i_{p1} = 1{,}80$ µA and $i_{p2} = 1{,}72$ µA, the resulting interference was negligible (Figs. 7c and 7d), even smaller than in normal interference condition (power cords far from the body, Figs. 5c, 5d, 6c, and 6d). This can be attributed again to the path to ground followed by displacement currents entering the body, which have a different effect on each lead. The increased value for i_p yielded a larger common-mode voltage but the CMRR was large enough to minimize its effect. Differential mode interference reduced because of the balance between the paths to ground from each recording electrode.

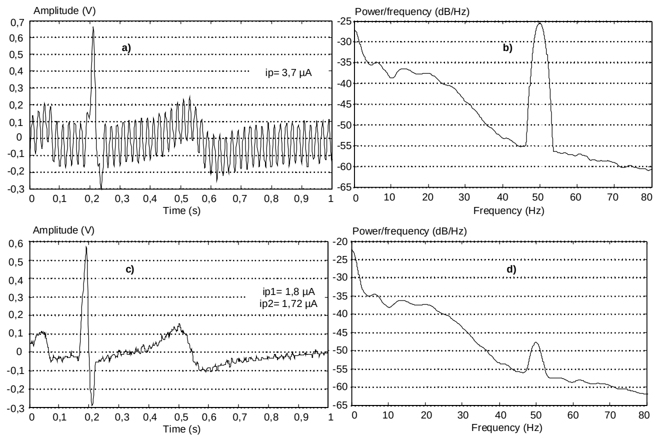

Fig. 7. Lead II for increased interference conditions (power cord on the back of the subject). a) ECG with a single ground electrode and b) its power spectral density. c) ECG with a ground electrode close to each recording electrode and d) its power spectral density

For lead III and the power line cord on the back of the subject, i_p was 1,4 µA, hence less than half that for lead II in the same conditions, and the interference was smaller (Figs. 8a and 8b). When two ground electrodes were used, they carried not so similar currents ($i_{p1} = 0{,}74$ µA, $i_{p2} = 0{,}56$ µA), but the interference (Figs. 8c and 8d) still underwent a significant reduction as compared to that when using a single ground electrode. However, it was larger than that for lead II in the same interference conditions (Figs. 7c and 7d), probably because $i_{p1} \approx i_{p2}$ for lead II.

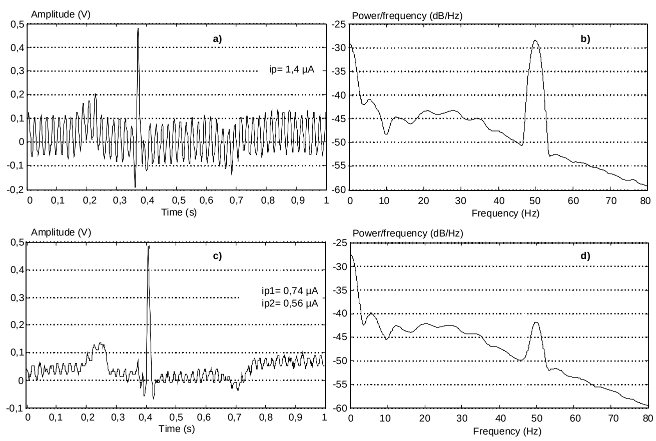

Fig. 8. Lead III for increased interference conditions. a) ECG with single ground electrode and b) its power spectral density. c) ECG with dual ground electrodes and d) its power spectral density.

Next, a 100 kΩ resistor was connected in series with one of the acquisition electrodes to simulate a skin-electrode impedance imbalance ($\Delta Z_e = Z_{e2} - Z_{e1}$), and lead III was recorded again. The displacement current into the patient remained the same ($i_p = 1{,}4$ µA), as expected, yet the interference for a single-ground electrode was a bit larger (Figs. 9a and 9b). This means that perhaps Z_C/Z_{e3} in (1) was large enough for the direct differential mode interference ($i_p Z_{t1}$) to still predominate over the component due to electrode imbalance. When using two ground electrodes (Figs. 9c and 9d), the interference was smaller than that obtained for balanced electrodes (Figs. 8c and 8d). Currents i_{p1} and i_{p2} were somewhat closer now ($i_{p1} = 0{,}76$ µA, $i_{p2} = 0{,}59$ µA), and this could explain the good interference reduction. It also confirms that direct differential-mode interference predominates over differential-mode interference due to electrode imbalance. Nevertheless, if Figs. 9c and 9d are compared to Figs. 6c and 6d (for lead II), where $i_{p1} \approx i_{p2}$, we conclude that closer values for i_{p1} and i_{p2} imply a reduced interference for a given lead but interference in a different lead can be smaller even if currents to ground are less balanced. The explanation could be that for two recording electrodes to be on the same equipotential line, we need that, in addition to $i_{p1} = i_{p2}$, a similar impedance between each recording electrode and the corresponding ground electrode. This condition may be difficult to achieve because, even if the distance between the two electrodes of each pair is the same, the skin-electrode impedance is known to change for different body sites [13].

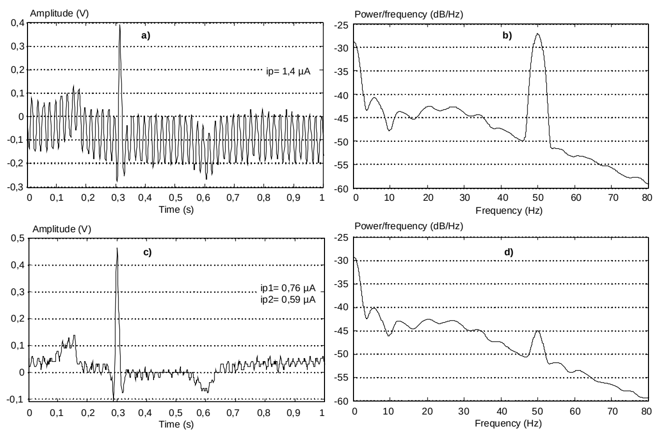

Fig. 9. Lead III for increased interference conditions and 100 kΩ imbalance between recording electrodes. a) ECG with single ground electrode and b) its power spectral density. c) ECG with dual ground electrodes and d) its power spectral density.

For an isolation amplifier, the common-mode voltage would be much smaller because the isolation-mode impedance Z_I is much larger than the common-mode impedance and most of the power line voltage would drop across Z_I. However, the power line current entering the body would probably follow a different path with respect to an earthed amplifier, because in Fig. 2 the impedance from signal ground to earth (isolation impedance) can be higher than that of C_b, and therefore a fraction of the power line current entering the body will flow to earth though C_b. Nevertheless, an electrode close to each recording electrode and connected to signal ground will place each recording electrode on an equipotential 50 Hz (or 60 Hz) line, regardless of how much power line current flows to earth trough each of these two ground electrodes (and Z_I), provided they are similar. Also, power line current coupled from the power supply will depend on the stray capacitance of the isolation transformer [5]. Part of this current can also flow to earth through C_b, but the remaining current will flow through the electrode(s) connected to signal ground (and Z_I). Therefore, the differential-mode interference from the isolation transformer should be smaller when using a ground electrode close to each recording electrode.

5. CONCLUSIONS

Circuit models to describe power line interference in biopotential acquisition systems commonly use lumped parameters to simplify the analysis. Interference reduction techniques based on these models usually focus on the effective CMRR as a major factor contributing to a cleaner signal. However, ECG recordings obtained with high-performance amplifiers still show power line interference, regardless of whether they use three- or two-electrode voltage amplifiers or two-electrode amplifiers based on body potential driving [8][9]. Huhta and Webster [1] pointed to interference directly coupled in differential mode as an interference source that cannot be reduced by improving the CMRR but can be reduced if the recording electrodes are on equipotential lines of the electric field produced by power line currents entering the body.

We have analyzed interference coupling by assuming a distributed-parameter model that considers capacitive coupling from power lines to any point of the body. It follows that a method for the recording electrodes to be on an equipotential line is to have a ground electrode very close to each of them. If the current through each electrode is the same, nearby points will have the same 50 Hz or 60 Hz potential and no power line current will flow between them.

We have tested our method by building an earth-grounded amplifier whose CMRR is large enough for the output interference to be very small, and recorded the ECG (leads I and II) under common power line interference. Lead II was also recorded when a power line cord was in contact with the back of the subject. The ECG (lead III) was also recorded with a power line in contact with the back of the subject and when a 100 kΩ resistor was connected in series with one of the recording electrodes, in order to increase the imbalance between them (ΔZ_e). We selected leads I, II and III because they use distant electrodes, so are more prone to differential-mode interference. We used an earth-grounded amplifier because it yields the largest common-mode voltage. The output power line interference was visually assessed from the time record and from the power spectral density of the recorded signal.

The experimental results show that using a ground electrode close to each recording electrode instead of a single ground electrode (on the right leg) reduces interference for all leads and situations: common interference, power line cord on the back of the subject ("increased interference condition"), balanced electrodes and unbalanced electrodes. For a given lead and a single ground electrode, a larger displacement current i_p coupled to the body results in an increased interference but not necessarily proportional to i_p (Figs. 6a and 6b compared to Figs. 7a and 7b). For different leads and a single ground electrode, interference does not necessarily increase with i_p (it is smaller in Figs. 7a and 7b than in Figs. 8a and 8b). This shows that the fraction of i_p that flows between the two recording electrodes strongly depends on the position of the electrodes, and therefore i_pZ_t in (1) should better be replaced by αi_pZ_t ($0 \leq \alpha \leq 1$) to indicated that only a fraction of i_p flows between the recoding electrodes, as stated in [1].

When using a ground electrode close to each recording electrode (and no right-leg electrode), interference reduction depends not only on the matching between the two ground currents i_{p1} and i_{p2}, but also on the lead, and the closeness to power lines. Thus, relatively dissimilar ground currents for one lead can yield interference similar to more balanced currents for a different lead (Figs. 8c and 8d compared to Figs. 6c and 6d). This can be attributed to different impedance between each recording electrode and the corresponding ground electrode, as skin-electrode impedance depends on the body site [13]. Also, for a given lead, interference can be smaller even if the difference between i_{p1} and i_{p2} is larger, depending on the distance to the power lines (Figs. 7c and 7d compared to Figs. 6a and 6b).

Electrode imbalance has no significant effect on the effectiveness of the dual ground electrodes method. If the ratio between the common-mode input impedance of the amplifier and electrode impedance (Z_C/Z_{e3}) is not large enough, electrode imbalance may reduce the effective CMRR and the output interference increase, as observed in Figs. 9a and 9b as compared to Figs. 8a and 8b. However, when using a ground electrode close to each recording electrode, for given lead and distance to power lines, the interference decreases whenever $i_{p1} \approx i_{p2}$, as observed in Figs. 9c and 9d compared to Figs 8c and 8d.

In summary, no single instance has been observed where, for given lead and position of the subject relative to power lines, the use of a ground electrode close to each recording electrode did not reduce interference compared to the use of a single ground electrode on the right leg, for an earthed amplifier. This method to reduce differential mode interference should also work for isolation amplifiers because its effectiveness does not depend on how much power line current flows to signal ground but on the balanced between currents flowing through from each signal ground electrode. Finally, this method can be applied to common recording equipment as it does not imply any circuit modification or special connections.

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