60-Hz Interference in Electrocardiography

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Abstract—**One of the major problems encountered in recording ECG's is the appearance of unwanted 60-Hz interference in the output. This paper examines the many possible sources of interference, and for each provides a description, an identifying test, and a remedy. Then we look at possible equipment imperfections and guidelines for amplifier design to aid in the evaluation of new developments in the field of electrocardiography. As an illustrative example, we apply the procedure for isolating the source of interference to an exposition display in which each visitor could rapidly view his own ECG.**

INTRODUCTION

INTERFERENCE from 60-Hz ac, sometimes referred to as ac pickup or hum, can be a problem in any biopotential recording situation. The source of this interference is the ac line potential (voltage) that is unavoidably present in any clinical situation, if for no other purpose than to light the room or power the recording equipment. This paper will separate, define, and quantitatively describe the ways by which interference can enter ECG recordings. The methods and calculations outlined here are not restricted to ECG recordings and could be applied to any biopotential recording system.

We begin by noting that ac potentials are always present, and it is not their presence, but rather their effects, that we wish to minimize. We develop simple tests to identify the manner in which interference is entering the system so that it can be eliminated. We hope to impart an understanding of the variables of this problem so that an engineer or technician might knowledgeably evaluate new developments in biomedical instrumentation relating to this problem. A doctor or nurse should benefit from reading the summary, tests, and remedies. Interference in ECG recordings is *not* a necessary evil or recurring nuisance that must be tolerated. By employing an organized approach to the problem we can effectively eliminate the causes of interference and avoid drastic remedies such as changing recording sites or installation of expensive shielding.

SOURCES OF INTERFERENCE AND RECORDING CRITERIA

AC fields that may cause interference can be classified into two independent categories: magnetic and electric. A changing magnetic field B produced by ac can induce in any nearby conductive loop an electromotive force (EMF), which results in an ac potential. A changing electric field E produced by an alternating potential can also produce interference by causing ac currents to flow to ground through the system. These currents flow through tissue and electrode impedances, thus producing ac potentials. Strictly speaking, these currents are conduction currents, which result from displacement currents. However, we will refer to these currents as displacement currents (I_D) to emphasize that they result from capacitive coupling between the fields and the system.

AC fields have many sources including lights, ac wiring and outlets, and other equipment operating nearby. Wolbarsht and Spekreijse [1] list many of these sources. They also discuss interference at high frequencies as from radio stations. Huntsman and Nichols [2] have proposed a solution for this latter type of problem employing a radio-frequency (RF) filter.

Some sources of 60-Hz potential produce electric fields, but not magnetic. For example, equipment that is plugged into an outlet but turned off will still produce an electric field even though no current flows. A 60-Hz potential is still present on one of the wires up to the ON-OFF switch. However, to produce a magnetic field, current must flow in the wires. A common source of magnetic fields is the transformers in the power supplies of most equipment. We can prevent electric fields from entering a circuit by shielding with any highly conducting surface such as copper or aluminum, but for magnetic interference we must use some ferromagnetic material such as mu metal. However, before we begin shielding the room, the ac cords, or even the patient, let us examine how interference could appear in the ECG in the first place.

Fig. 1 shows a schematic of a grounded ECG recording system. All ac quantities and biopotentials in this paper are peak-to-peak magnitudes. The nomenclature is defined as follows:

B	60-Hz magnetic field (magnetic flux density [Wb/m^2]);
S	area of loop enclosed by leads A and B (m^2);
E	60-Hz electric field intensity (V/m);
I_D	60-Hz displacement currents produced by E (A);
Z_1, Z_2	electrode–tissue impedances (Ω);
Z_G	ground electrode impedance (Ω);
Z_I	internal body impedance (Ω);
C_1, C_2, C_B	capacitances coupling into the system (μF);
Z_D	amplifier input differential impedance (Ω);
Z_{in}	amplifier input impedances to ground (Ω).

Initially we assume the amplifier is perfect. First, we must know what value of interference is significant compared to the ECG signal. A typical ECG potential on the body is about 1 mV or 0.001 V. If we reduce the interference to 1 percent of the desired signal, it will not significantly degrade the recording. One percent of 1 mV is 10 μV. In other words, interference may enter the ECG recording, but we won't be able to notice it unless it is greater than 10 μV. In the same manner we could calculate the interference tolerance of any

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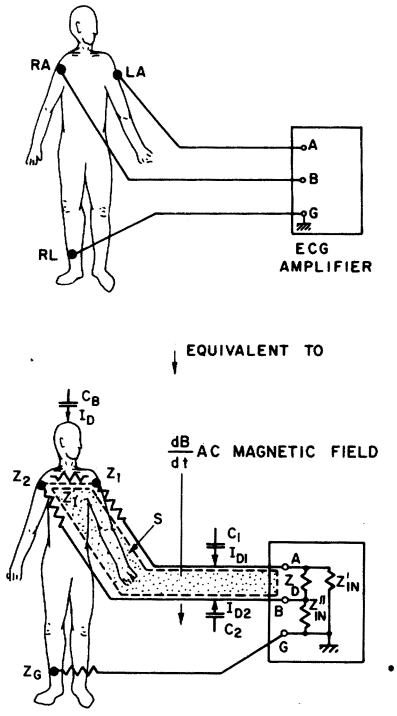

Fig. 1. Typical ECG recording configuration lead I and the equivalent circuit. RA is right arm; LA is left arm. RL is right leg. The capacitors simulate the entry of displacement currents. Dotted lines indicate the area S of the conductive loop.

recording system. In an EEG system, for example, the desired potential is about 50 μV, so the interference tolerance using our 1-percent criterion would be 0.5 μV. This is the magnitude of the interference potential that must appear differentially between input A and B of the amplifier to be noticeable in the output. The different ways that interference could enter the systems are 1) magnetic induction, 2) displacement currents into the electrode leads, 3) displacement currents into the body, and 4) equipment interconnection and imperfections.

We will examine each variable individually using the following format: 1) description, 2) test, and 3) remedy. Using this procedure we should be able to accomplish a step-by-step elimination of each variable and either find the problem or have good reason to suspect the equipment.

Magnetic Induction

Description: Some authors up until this time have referred to interference as a result of induction into a loop as electromagnetic (EM) interference. We feel that because we are concerned with the radiated fields in the near field, *magnetic* induction would more aptly describe this type of interference. EM interference could then be reserved for the types of high-frequency radiation mentioned earlier where the source is relatively far away (far field).

Any conductive loop in the vicinity of a changing magnetic flux density will have a potential (EMF) induced in it proportional to the area of the loop, its orientation, and the magnitude of the magnetic flux density. The Maxwell-Faraday law of magnetic induction [3] states the following:

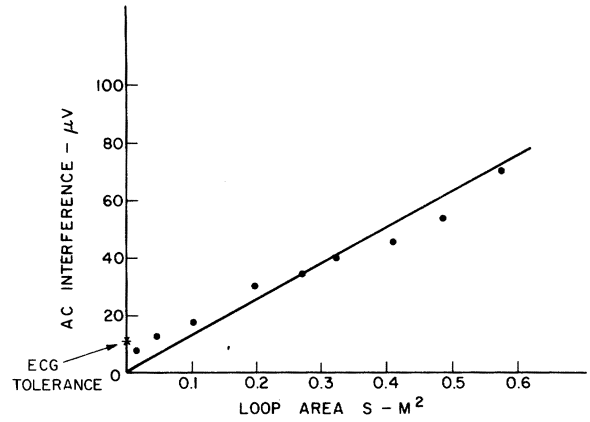

Fig. 2. AC potential at the amplifier input versus loop area for magnetic induction interference.

$$\text{induced potential} = -\frac{d\Phi}{dt} = -\frac{d}{dt}\int \boldsymbol{B}\cdot d\boldsymbol{S} = -\frac{dB}{dt}S \qquad (1)$$

Φ = flux [Wb]

$\boldsymbol{B} = B_M \cos\phi \cos\theta \cos\omega t$ [Wb/m^2]

$\omega = (2\pi)(60)$ [rad/s]

$\cos\phi \cos\theta$ = orientation of loop (θ and ϕ specify the polar coordinates).

Taking the derivative we find the following:

$$\text{induced potential} = \omega S B_M \cos\phi \cos\theta \sin\omega t. \qquad (2)$$

B_M is the peak-to-peak magnitude of the magnetic field. If we assume a constant frequency and loop orientation we have the equation

$$\text{peak induced potential} = KB_M S \qquad (3)$$

where K is a constant. We assume that some 60-Hz flux density B is present in any situation, so the important variable is the *area S* of the loop. See Fig. 1.

The nature of the ac fields depends on the environment. We recorded the largest interference induced into a wire loop connected to an oscilloscope input when the loop was oriented approximately parallel to the floor. This implies that for this particular case, the strongest magnetic field was between the top and bottom of the room. However, any field orientation is possible. We also noticed a 180-Hz harmonic potential when the loop was oriented perpendicular to the floor. All measurements taken by making the subject part of the conductive loop compared very well with measurements in which the loop was only a piece of wire. Fig. 2 shows a graph of interference versus loop area. The linear curve indicates that the B-field was approximately constant throughout the experiment and was calculated by (2) to be about 3.2×10^{-7} Wb/m^2. The slope for this particular curve was about 120 μV/m^2 loop area. Our interference tolerance for ECG recording (10 μV) results from a loop area of less than 0.1 m^2, or about 30 cm by 30 cm. Using shielded wire for the loop gave identical data. In fact, employing a magnetic shield around the wire would give the same result because the flux density B still enters the area S. It would be necessary to magnetically shield this entire area to eliminate this factor. We may ask why the area of the

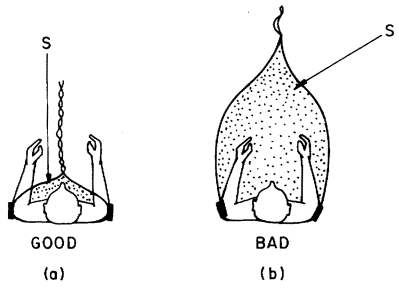

Fig. 3. Electrode lead placement illustrating the magnetic induction loop. (a) Correct lead placement: twisted leads run close to the body yielding small S. (b) Incorrect lead placement: area S as large as 0.2 m^2.

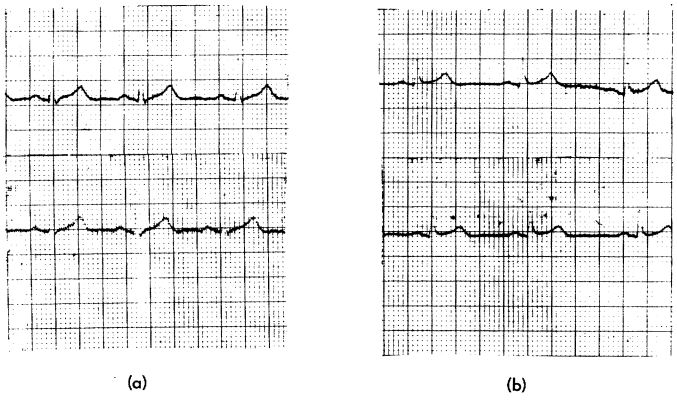

Fig. 4. ECG recordings taken in a typical clinical situation. (a) Lead II electrode placement configuration. (b) Lead I configuration. Notice the decreased interference in the upper traces where we employed the correctly minimized area between leads.

loop from *input B* to ground does not appear in the equation. The answer is that there is an ac potential induced in this loop, but it is common to both inputs A and B. Our ideal amplifier amplifies only the *difference* between the potentials at A and B.

Test: The best way to test for magnetic interference is to vary the size of the loop formed by the two input electrode leads. Spreading the leads apart should increase the interference. It should increase linearly with area and be sensitive to rotation of the plane of the loop.

Remedy: The best way to eliminate magnetic interference is to simply make the loop area as small as possible. This can be accomplished by twisting the leads up to the body and running them close to the body. Fig. 3 shows good and bad lead I electrode lead placement. We can apply the same principles to any lead configuration. Fig. 4 shows ECG recordings taken on a Burdick EK-III ECG machine. These recordings were made immediately adjacent to an air-conditioning unit that was turned on. The only difference between the upper and lower traces for lead I and II is the electrode lead placement. The lower traces show significant interference as a result of increasing the effective loop area.

Displacement Currents into the Leads

Description: Changing the electric field intensity will capacitively couple displacement current into the ECG lead wires.

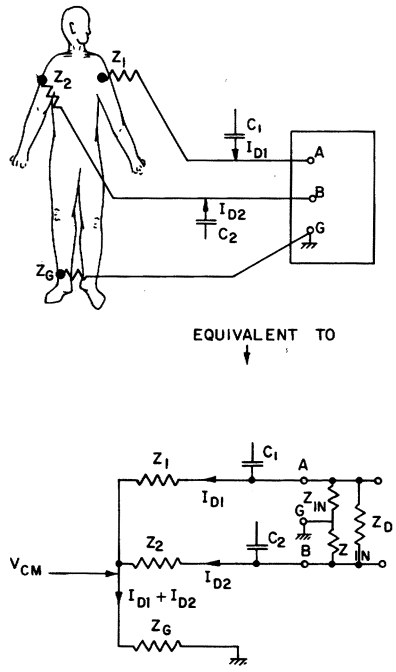

Fig. 5. Simplified model to illustrate ac displacement currents coupling to the unshielded electrode leads. Assume Z'_{in}, Z''_{in}, $Z_D >> Z_1, Z_2, Z_G$. V_{CM} is the CM potential. Internal body impedances (Z_I) are assumed to be negligible.

We assume first that the electrode leads are unshielded and investigate the magnitude of interference that could cause problems in this way. The equivalent model, Fig. 5, shows a recording situation and the electrode impedances Z_1 and Z_2. Internal body resistances have been assumed to be zero. Because an ac electric field is produced by a changing electric potential above ground, any displacement current will flow to ground by the path of least resistance. We assume that Z_{in} and Z_D are very large so that any current coupled into the electrode leads flows through the associated electrode impedance and the ground impedance Z_G. The voltage of importance is that appearing between input A and B, i.e., $V_A - V_B$:

$$V_A = Z_1 I_{D1} + (I_{D1} + I_{D2})\, Z_G \tag{4}$$

$$V_B = Z_2 I_{D2} + (I_{D1} + I_{D2})\, Z_G \tag{5}$$

$$V_A - V_B = Z_1 I_{D1} - Z_2 I_{D2}. \tag{6}$$

If $Z_1 I_{D1} = Z_2 I_{D2}$, the interference from this factor will be zero. Either an electrode impedance unbalance *or* unequal values of displacement current into the leads can cause interference.

We conducted experiments to measure how much displacement current flows into different sizes and lengths of wire. In a typical experiment, a single 3-m length of number 20 unshielded wire picked up 6 nA (1 nA = 10^{-9} A) of displacement current under poor recording conditions (ac cords and equipment nearby). This value was reduced by 80–90 percent when a grounded object (in this case, a grounded person) was next to the wire. Any grounded equipotential surface will distort the electric field and decrease the displacement current into the leads.

In most recording situations, the lengths of the leads are the same and therefore if the leads run close together, the displacement current into each should be approximately equal. Making the valid assumption that $I_{D1} = I_{D2} = I_D$ (except when an ac source is very near one lead and not the other), we can calculate the interference potential assuming a 5000-Ω electrode impedance unbalance:

$$V_{ac} = V_A - V_B = I_D(Z_1 - Z_2) \tag{7}$$

where

V_{ac} ac interference potential;
$V_A - V_B$ $= (6 \times 10^{-9}$ A) 5000 Ω;
$V_A - V_B$ $= 30\,\mu$V.

This is approximately 3 percent of the ECG potential. Many authors have investigated typical values of electrode-tissue impedance, variations with frequency, and distortion of the ECG as a result of low amplifier input impedance [4]-[9]. These problems will be discussed under equipment design. Electrode impedance at 60 Hz may range from less than 1000 up to 100 000 Ω in some cases [10]. Although the magnitude of the electrode-tissue interface is important, interference enters a system as a result of *unbalance* of the two impedances.

Many commercial electrode leads that are less than 1 m long are unshielded wire. They do not cause excessive interference because the displacement current is too low to cause an appreciable potential drop through the electrode impedance. In another experiment, three unshielded wires were used as electrode leads and although 6×10^{-9} A was again measured in each lead, the ECG waveform was free of noticeable interference.

Some confusion still exists as to whether a larger electrode area will increase or decrease interference. On one hand, the larger area could couple more displacement current into the system. On the other hand, increasing the electrode area will decrease the electrode-tissue impedance [11]. Because of the greatly reduced fields near the body and the fact that increasing the size of both electrodes equally will not increase the differential current, we conclude that larger electrodes will *decrease* the interference and, conversely, making the electrode area very small may increase interference.

Test: We have seen that if electrode leads are unshielded, there is a possibility that ac displacement currents into the leads could cause a problem if the electrode-tissue interfaces have different impedances. To test this factor, twist the electrode leads together to avoid magnetic induction, place the two electrodes immediately adjacent to each other on any part of the body, and observe the output with the right leg ground lead in place. Then remove the three electrodes and place them in a beaker of saline or stick them together with electrode paste. If noticeable interference is present with the electrodes on the body and not with the other configuration, the electrode impedances may be unbalanced.

Remedy: If we determine that there is an impedance unbalance, the cause could be poor skin preparation, dry electrode paste, or any other form of poor electrode to skin contact. In very difficult cases it may be necessary to rub the skin with fine sandpaper to produce very faint erythema or reddening [5].

Interference could also result from electrode lead placement that makes the displacement currents into the leads unequal. In general, lead placement that decreases interference from magnetic induction (as previously discussed) will also reduce this effect. Properly shielded electrode leads should eliminate any interference due to displacement current. However, shielding the electrode leads could introduce other types of interference. We tested 3-m shielded leads for artifacts from flexing. Shaking the leads vigorously produced about 100 μV of noise. This situation would seldom be encountered in a recording situation. Shielded cable capacitance does limit the length that can be used, as we will show later. If the test shows interference is present with the electrodes on the body and also with them shorted together in the saline or paste, suspect an equipment failure.

Displacement Currents into the Body

Description: In the previous section we saw that the displacement currents could enter unshielded leads because of capacitive coupling. The surface of the human body can also act like a capacitor, and the displacement current will then flow through the body if it is grounded. This results in the body being at some potential above ground determined by the displacement current and the ground electrode impedance Z_G. An easy way to estimate the magnitude of the displacement current entering the body is to place a finger on the input terminal of a grounded oscilloscope making sure that the body is not grounded through shoes to the floor or through any other path. Knowing the input impedance of the oscilloscope at 60 Hz we can then calculate the ac displacement current by the formula

$$I_D(A) = \frac{V(\text{ac V measured on oscilloscope})}{Z_0(\text{input impedance of the scope at 60 Hz})}. \tag{8}$$

This value will rarely exceed 1 μA even when holding onto an ac line cord and will more likely be about 0.1 μA. If we don't know the input impedance of the oscilloscope we can quickly determine the approximate value with this simple test. Record the value of the ac potential with a finger on the input. Call this value V_1. Remove the finger and holding on to one lead of a 1-MΩ resistor, touch the other lead to the oscilloscope input. Call this value V_2. Then calculate Z_0, the input impedance, by the formula

$$Z_0 = \frac{V_2(1\ \text{M}\Omega)}{V_1 - V_2}. \tag{9}$$

The value of I_D is the current that will flow through the body and ground through the ground electrode. Negligible current returns to ground through the input impedance of the amplifier. If we neglect any internal body resistance, we can calculate the potential of the body, sometimes called the *common mode* (CM) *potential* (V_{CM}). We neglect currents into the leads:

$$V_{CM} = I_D Z_G. \tag{10}$$

This potential is common mode because it appears at our ideal differential amplifier inputs A and B simultaneously and should not appear on the ECG waveform at the output. If we assume a maximum I_D of 1 μA and a very high ground electrode impedance Z_G of 100 kΩ, the maximum value of the CM potential that will appear on the body will be $(1 \times 10^{-6}) \cdot (1 \times 10^{5}) = 0.1$ V. This is the same value mentioned by the American Heart Association (AHA) [12]. Values between 1 and 10 mV would more likely be found in the typical situation.

But so far, how could this CM potential cause a problem, assuming that our amplifier meets specifications (to be discussed later)? In describing the displacement currents in the body, we originally neglected the internal impedances of the body. Because they are finite, any current flowing through the body will flow through the body's impedance, causing a potential drop. The human body is often represented as a stickman with resistors in his arms, legs, and torso. See Fig. 6. The magnitudes of these resistors (impedances) are small compared to the electrode-to-tissue impedance. They are usually about 20 Ω through the torso and can be as high as 400 Ω from shoulder to finger. This impedance is sometimes called the subcutaneous impedance. 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. If we now attach our electrodes, the difference between the potential at one point and that at another will be amplified with the ECG. We can represent this potential difference by the displacement current flow through an internal impedance Z_I between the points of electrode attachment. See Fig. 7. The maximum possible interference would be given by

$$V_{ac} = Z_I I_D \text{ (in the body).} \tag{11}$$

For V_{ac} to be greater than our 1-percent criterion (10 μV), assuming I_D only one-tenth the value we assumed before, Z_I must be only

$$Z_I = \frac{V_{ac}}{I_D} = \frac{10 \times 10^{-6} \text{ V}}{0.1 \times 10^{-6} \text{ A}} = 100\ \Omega. \tag{12}$$

This value could be easily exceeded when the electrodes are placed on any of the appendages. However, in general, this interference will be less than the maximum value shown above, since I_D does not flow directly through Z_I. Thus only a projected component of I_D results in interference. Placing the electrodes close together should decrease the impedance between the electrodes and eliminate the interference from this source. This is the situation when the three leads are placed on the sternum or in EEG applications where the electrodes are placed close together. But even if the electrodes are placed far apart, the potentials at the two points *can* be the same with respect to ground so that no interference will appear differentially. In other words, the position of the ground electrode dictates by what path the displacement current will flow to ground and what potentials will appear at each electrode. This should allow us to also eliminate this factor by changing the ground electrode position.

Test: To determine if nonsymmetry of the potential on the patient is a problem, simply move the ground electrode (usually the right-leg electrode) to a different location. Decreased interference illustrates isolation of at least part of the problem.

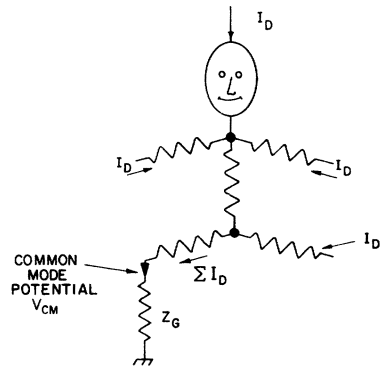

Fig. 6. Model illustrating the body's internal resistance and displacement currents flowing to ground with only the ground electrode attached.

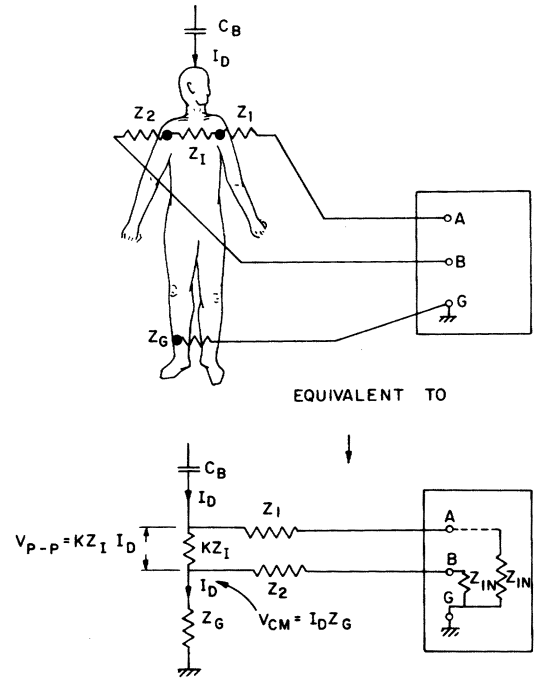

Fig. 7. Simplified model of displacement currents into the body. K is a fraction between 0 and 1.0. I_D is the sum of all displacement currents entering the body.

Remedy: If movement of the ground electrode does decrease the interference, consider taking the patient's ECG with the ground in its new position. This will in no way change the magnitude or shape of the ECG to be taken. Our experiments have shown that if the right-leg ground is causing some interference problems, the best alternative grounding location is the stomach or chest. The alternate ground position depends on the configuration and the path through which the interference enters the body. For instance, Fig. 8 shows lead II ECG waveforms in which the displacement current was increased by placing the right hand near an ac line cord. The upper trace is normal lead II with the right leg grounded and the lower trace shows the lower interference when the right-leg electrode was moved to the right chest.

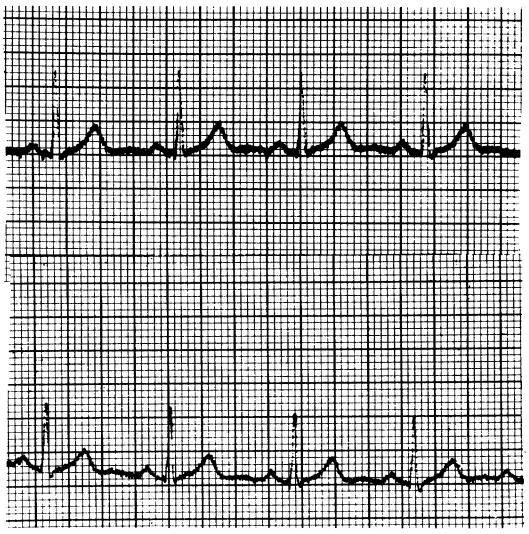

Fig. 8. Lead II ECG recordings. Upper trace–right-leg ground, lower trace–chest ground. Notice that there is no change in the shape of the ECG waveform.

Another possible remedy, suggested by Schmitt [15], is to cover the patient with a grounded conductive blanket to shield against displacement current entering the body.

Equipment

We will first look at how the interconnection of equipment and the patient could cause interference. Then we will turn to the amplifier specifications and our recommendations for eliminating ac interference. Finally, we will look at new developments in biopotential recording instrumentation and how they relate to patient safety and the interference factors discussed thus far.

Interconnection of Equipment

With the increased use of internal conductive catheters in surgical and coronary care monitoring, small leakage currents could cause ventricular fibrillation. Ten μA is considered a safe maximum for leakage current. In some cases increased interference may be the only indication of a hazardous situation. One way in which this can result is if the third-wire ground of an instrument breaks or is not connected within the socket. Leakage current may cause the instrument enclosure to be at a high potential.

If there is a potential difference between two instruments it may appear on the ECG trace. This situation could occur if the patient is grounded to two instruments that are plugged into different power outlets. A less-hazardous interference problem could develop if the ECG is being monitored remotely and the remote instrument is at a different ground potential than the main unit. An example of this situation is a remote monitor oscilloscope connected to the recording amplifier, but plugged into a different power outlet. A fault anywhere in the electrical circuit causes leakage current to flow in the ground return. This current flowing through the resistance of the connecting ground wire between the instruments produces a potential difference that appears on the ECG.

Grounding more than one instrument to the patient can also cause a ground loop. AC magnetic fields can induce a potential in the loop that may appear on the patient and the ECG.

These defects can be remedied by the following procedures.

1) Periodic checks of ground-wire continuity of all equipment.

2) Ground all equipment connected to the patient at only one point on the body. Check the ground potentials of all equipment and ground returns with respect to one another with an ac voltmeter capable of detecting a 10-mV difference.

3) Provide a common ground panel for all power outlets to be used in any recording situation.

As with any instrument, ac interference in the output could be a result of a power-supply failure or inadequate filtering. This interference would be present whenever the instrument is operating.

Specifications

In 1967, the Committee on Electrocardiography of the AHA revised their ECG equipment recommendations [12]. Some of the specifications relevant to this discussion are given below and apply to direct-writing ECG's.

Input impedance: 500 000 Ω between any single patient electrode and ground.

Common-mode rejection ratio (CMRR): 1000 to 1 between 45 and 65 Hz and with a 5000-Ω lead unbalance.

Frequency response: $\pm$0.5 dB from 0.14 to 50 Hz; $\pm$3 dB from 0.05 to 100 Hz.

Input Impedance

The input impedance has been represented here as the CM input impedance (Z_{CM}) and the differential input impedance (Z_D) at 60 Hz. The CM impedance is the impedance measured between ground and the two inputs A and B when they are connected together. This value will be equal to $Z_{in}/2$ for the configuration we are using. The differential input impedance Z_D will be the value measured between inputs A and B. We want Z_D to be large enough so that there is no loading effect on the differential potential.

When the ECG was first being recorded, the electrodes consisted of buckets of saline. This large electrode-to-skin area was needed to lower the value of the electrode–tissue impedance so that it was small compared to the low value of input impedance of the string galvanometers used at that time. The electrode impedance must be kept small so that the cardiac potential is not reduced or differentiated causing distorted waveforms [8]. As ECG instrumentation improved from vacuum tubes to transistors and operational amplifiers, the input impedance that could be attained was much higher than the electrode impedance, even using a small electrode.

The CM input impedance must be large compared to the electrode impedances to minimize the effects of an electrode impedance unbalance. We recall that the body has some CM potential. A severe unbalance in the electrode impedances or the input impedances will cause this potential to be higher at one input than the other. We will call this phenomenon the

potential divider effect [13]. We can calculate what value of Z_{in} is necessary so that a 5000-Ω unbalance does not produce interference above 1 percent. The differential ac potential is the CM potential (V_{CM}) times the difference of the potential divider ratios (see Fig. 5):

$$V_A = V_{CM}\left[\frac{Z_{in}}{Z_{in} + Z_1}\right]$$

and

$$V_B = V_{CM}\left[\frac{Z_{in}}{Z_{in} + Z_2}\right] \tag{13}$$

$$Z_{in} = Z'_{in} = Z''_{in}$$

$$V_A - V_B = V_{CM} Z_{in}\left[\frac{1}{Z_1 + Z_{in}} - \frac{1}{Z_2 + Z_{in}}\right]$$

$$= V_{CM} Z_{in}\left[\frac{Z_2 - Z_1}{Z_1 Z_2 + Z_{in}(Z_1 + Z_2) + Z_{in}^2}\right]. \tag{14}$$

Assuming Z_1 and Z_2 much less than Z_{in},

$$V_A - V_B = V_{CM}\left[\frac{Z_2 - Z_1}{Z_{in}}\right]. \tag{15}$$

Solving for Z_{in} we find

$$Z_{in} = \frac{V_{CM}}{V_A - V_B}\,[Z_2 - Z_1]. \tag{16}$$

We know that $V_A - V_B$ must be less than 10 μV and we can use the average value of the CM potential that we used before (10 mV).

Then the minimum Z_{in} to prevent interference is

$$Z_{in} = \frac{10 \times 10^{-3}}{10 \times 10^{-6}}(5 \times 10^3) = 5\ \text{MΩ at 60 Hz}$$

$$Z_{CM} = Z_{in}/2 = 2.5\ \text{MΩ}. \tag{17}$$

A Z_{in} of 50 MΩ would be adequate for CM potentials up to 100 mV or impedance unbalances of 50 000 Ω, which are both unrealistic values, except with situations of high ac potentials such as when using electrocautery. The two input impedances Z_{in} were assumed to be equal in the previous discussion. If the two impedances Z'_{in} and Z''_{in} were different, the potential divider ratios would again be unequal. The more general equation is

$$V_A - V_B = V_{CM}\left[\frac{Z'_{in}}{Z'_{in} + Z_1} - \frac{Z''_{in}}{Z''_{in} + Z_2}\right]. \tag{18}$$

We found earlier that the ratio of $(V_A - V_B/V_{CM})$ had to be less than 0.001 to limit the interference to 1 percent. Even if $Z_1 = Z_2 = 10$ kΩ (electrode impedances balanced) we can have problems. If $Z'_{in} = 5$ MΩ and $Z''_{in} = \infty$ (input impedances unbalanced), we can calculate how much interference will result with a V_{CM} of 10 mV:

$$V_A - V_B = V_{CM}\left[\frac{5}{5.01} - 1\right] = V_{CM}\ 0.002. \tag{19}$$

For this case, with input impedances greater than 5 MΩ, we obtain a 2-percent ac interference. Larger values of Z_1 would produce larger interference.

We can also calculate a value for the differential input impedance Z_D so that the signal magnitude is not reduced more than 5 percent. This value would also present minor waveform distortion as a result of low input impedance. For this criterion, Z_D must be 20 times the sum of the electrode impedances. Assuming Z_1 and Z_2 to be 10 000 Ω each, Z_D must then be at least 400 000 Ω. Strong [14] stated that a low value of Z_D is desirable to equalize the electrode dc potentials. Our measurements have shown that except for very small electrodes, this is a questionable consideration because it would take days to discharge these potentials through a 2-MΩ resistor as suggested. Most amplifiers are built to handle 100 mV dc without saturating, so these potentials are not a problem.

We must also remember that the input impedance Z_{in} is specified at 60 Hz, and therefore any capacitance to ground at the input will lower its value. For instance, if the input resistance is 7 MΩ and we place a 380-pF (1 pF = 10^{-6} μF) capacitor, which has an impedance of about 7 MΩ at 60 Hz, in parallel with each input, the effective Z_{in} will be reduced to 5 MΩ. If shielded cables are used for electrode leads, the maximum length that can be used will be limited by the cable's distributed capacitance. This value is usually about 30 pF/ft, so that its capacitance shouldn't normally degrade the input impedance excessively, providing cables are shorter than 12 ft. Fixed capacitances in the range of 470 pF are prevalent at the inputs of most equipment on the market today. These capacitors are designed to eliminate RF interference, but could also introduce interference problems by degrading the input impedance. The RF filter previously mentioned [2] employs a total of 1000 pF from each input to ground. The impedance of this capacitance at 60 Hz is less than 3 MΩ, and thus lowers the input impedance of the amplifier.

The considerations for amplifier specifications are different when averaging (weighting) networks are used such as the aVR, aVL, aVF, and V leads on most conventional recorders. The Wilson central terminal (V lead) configuration requires averaging resistors of at least 333 kΩ each in each of the three averaging leads to meet the AHA specification of 500 kΩ to ground on each electrode with the other grounded. This resistor in each lead prevents degradation of the cardiac potential that is being averaged. A compensating resistor equal to the parallel combination of the averaging resistors is needed in the other lead to balance the input (in this case, 111 kΩ). This high impedance in series with the inputs greatly increases the requirements for Z'_{in} and Z''_{in}, if they are unequal. Equation (18) now becomes (worst case)

$$V_A - V_B = V_{CM}\left[\frac{Z'_{in}}{Z'_{in} + 111\ \text{kΩ}} - 1\right] = 0.001\ V_{CM}. \tag{20}$$

Solving for Z'_{in} we find that Z'_{in} must be greater than 111 MΩ. Notice that Z_1 and Z_2 now represent any impedance in series with the input leads. For this case, Z'_{in} and

Z_{in}'' must be above 111 MΩ to limit the potential divider effect interference to 1 percent. We can approach the ECG amplifier design problem in several ways. We know that an input impedance greater than 5 MΩ will allow us to record I, II, and III lead configurations without ac interference caused by the potential divider effect. The aVR, aVL, aVF (unipolar limb leads), and V lead (precordial lead) configurations require either 1) very high Z_{in} by careful amplifier design [14], 2) some impedance-matching device such as a buffer amplifier before the averaging resistors that will allow us to use smaller averaging resistor values [15], or, 3) some method like driven right leg (to be discussed) that effectively cancels the CM voltage (V_{CM}), which is causing the problem in the first place.

Common-Mode Rejection Ratio

The *differential* gain of a biopotential amplifier is the ratio of the output to the input potential at a specified frequency with one input grounded:

$$A_D = \frac{V_{out}}{V_{in}}. \tag{21}$$

This amplification factor is usually adjustable between 200 and 2000 for ECG recording. The *CM gain* is the ratio of the output to the input potential when both inputs are ungrounded and connected together. For an ideal differential amplifier, this gain should be zero because we only wish to amplify the difference between the inputs:

$$A_{CM} = \frac{V_{out}}{V_{in}}, \quad \text{inputs connected together.} \tag{22}$$

The ratio of these two amplification factors is called the *CMRR* for an amplifier:

$$\text{CMRR} = \frac{A_D}{A_{CM}} = X. \tag{23}$$

It is defined as the ratio of the amplitude of the CM potential to the amplitude of an equivalent differential signal that would produce the same output [14]. Ideally, the CMRR should be infinite. It is usually expressed in decibels:

$$\text{CMRR}_{dB} = 20 \log X. \tag{24}$$

If we again assume a CM potential of 10 mV on the body and no electrode impedance unbalance, we can calculate the minimum CMRR for our interference tolerance:

$$\text{CMRR}_{min} = \frac{10 \times 10^{-3}}{10 \times 10^{-6}} = 1000, \quad \text{at 60 Hz or 60 dB.} \tag{25}$$

It should be obvious that we must specify the input impedance *and* the CMRR of an amplifier. Any unbalance in the electrode impedances will reduce the "effective" CMRR because of the potential divider effect, but we will treat these variables independently. Their contributions are

$$V_{ac} = Z_G I_D \left[\frac{1}{X} + \frac{Z_2 - Z_1}{Z_{in}} \right] \tag{26}$$

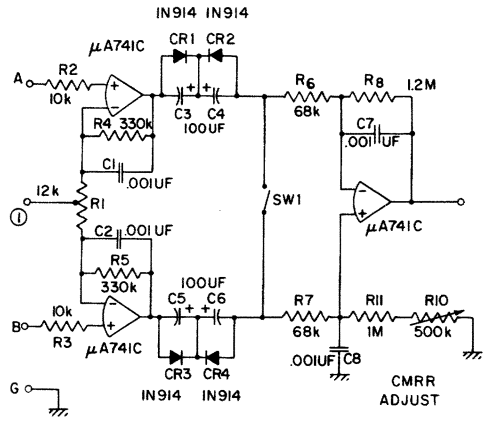

Fig. 9. Circuit of the ECG amplifier used in these measurements. Point 1 is where the CM potential is sampled for guarding or driven right-leg circuitry.

where $Z_G I_D$ is the CM potential previously defined. The frequency at which the CMRR is measured must be specified because it is a function of frequency. The minimum specifications for interference without averaging networks are, therefore, as follows:

input impedances:
differential 400 000 Ω;
CM 2.5 MΩ;
CMRR 60 dB at 60 Hz with a 5000-Ω unbalance.

Frequency Response

The frequency response of an amplifier is determined by its *RC* time constants. The circuit for the ECG amplifier that we used is shown in Fig. 9. It employs three operational amplifiers in a standard instrumentation amplifier configuration. The relatively inexpensive components and minimum circuitry make it very attractive. An ECG amplifier of this type was suggested by Fairchild Semiconductor [16]. The diodes CR_{1-4} across the 100-μF capacitors are necessary because polarized capacitors will not operate correctly if the dc potential across them changes polarity. The diodes protect the capacitors by conducting when large dc potentials are present, as can occur when the input stages go in and out of saturation.

The low-frequency time constant is determined by $C_3/2$ times R_6 and is 3.3 s. The cutoff frequency is $1/2\pi\tau = 0.05$ Hz. The high-frequency response is attenuated twice. The first cutoff is determined by R_4C_1 and R_5C_2. The second cutoff is determined by R_8C_7 and $R_{11} + R_{10}, C_8$. R_{10} balances any circuit unsymmetry and adjusts for maximum CMRR. This amplifier meets the AHA specifications for frequency response, which are 0.05–100 Hz. Of course, a strip-chart recorder used as a readout device will usually limit the frequency response to about 100 Hz. Switch SW1 was included to quickly reset the baseline after the amplifier has saturated.

Other Circuits and Modifications

We will now examine some of the techniques that are or may be employed in a recording situation to eliminate interference.

60-Hz Filters: It is clear that good operating procedure and proper equipment design can eliminate interference from ECG recordings. Filters should not be necessary except in extreme conditions, but if they are employed the facts should be noted on the ECG record so that diagnostic errors will not result.

Preamplifiers Near the Electrodes: Several papers suggest preamplifiers (also called buffer amplifiers and cathode followers) near the electrode [4], [6], [7], [9], [15], [20]. The advantage of these is interference reduction by canceling the effects of electrode impedance unbalance. When we looked at the problem posed by the use of averaging networks, we found that we could employ individual buffer amplifiers for the three averaging leads rather than design an amplifier that has at least a 111-MΩ input impedance. These could be mounted in the ECG amplifier and would prevent the ac interference that results from the potential divider effect. Mounting the buffer amplifiers at the electrode seems to be an expensive solution to this problem. The reasons for employing buffer amplifiers at the electrode rather than in the equipment are 1) elimination of lead-flexing artifacts and lead-length limitation with shielded cable, and 2) elimination of interference caused by displacement currents entering the electrode leads. The differential amplifier input impedance can be much lower using buffer amplifiers because the buffer output impedances are on the order of 1 Ω, making any unbalances negligible. The high input impedances of buffer amplifiers solve the potential divider-effect problem, as illustrated earlier, by increasing the effective amplifier input impedance Z_{in}. These preamplifiers could introduce more interference than they eliminate if the potential gain of one were slightly different than the other. Assuming a CM potential of 10 mV, a pair of these preamplifiers mounted at the electrodes or on the equipment should be matched to 0.1 percent to meet our 1-percent interference criteria. Unity gain buffers of this tolerance are readily available. Insulated electrodes with preamplifiers have been proposed for long-term monitoring [9], [17]. The electrode–tissue interface impedance is much higher than with conventional electrodes so that a very high input impedance must be provided either at the electrode or the amplifier. We have not tested this kind of electrode.

Guarding Circuits–Input and Shields: This technique samples the CM potential within the amplifier and drives the inputs and/or the shields of the electrode leads through a cathode follower [14, p. 302]. Driving the inputs effectively raises the input impedance of the amplifier, but if the amplifier already had a sufficiently high input impedance, there is no noticeable improvement. Driving the shields cancels the effects of any capacitance between the shields and inner conductors at 60 Hz. This may improve the input impedance of an amplifier if the leads are very long, but otherwise it is better to simply ground the shields. Like the preamplifiers, these circuits improve the input specifications of an amplifier to reduce the potential divider effect.

Driven Right Leg [18]: This modification also samples the CM potential within the amplifier. See Fig. 10. Instead of

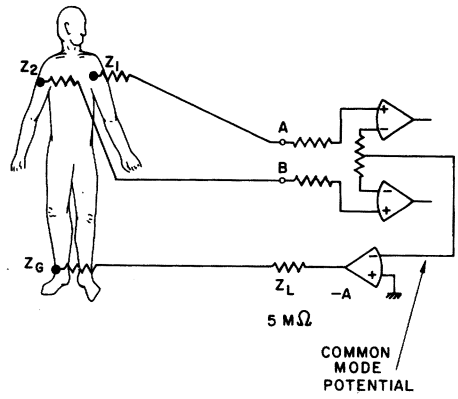

Fig. 10. Driven right-leg configuration.

directly grounding the patient, the right-leg electrode is connected through a current-limiting resistor to the right-leg amplifier, which amplifies the CM potential. This circuit actually drives a small amount of current into the right leg to equal the displacement currents flowing in the body. The body is the summing junction in a feedback loop so that the effect is to lower the CM potential on the body to a very low value. The amplifier is designed to saturate at currents above 2 μA so that shock hazard is minimized. The patient is isolated from ground by the 5-MΩ resistor. This is probably the best method for eliminating CM potential interference. Applications to EEG recording where the interface tolerance is much lower are immediately apparent. We feel that rather than employ very high input impedance amplifiers or buffer amplifiers to eliminate ac interference due to the potential divider effect, a better solution is to use the driven right-leg scheme, which requires only one additional amplifier. Remember that the other factors can still cause interference even though the effects of the CM potential are eliminated.

The buffer amplifiers may still be preferable, however, in those cases where the electrode impedance is high. In these cases, the problem is not interference, but rather signal distortion. An appreciable portion of the available signal appears across the electrode, and less than the desired amplitude appears at the amplifier.

Isolated Input: A newcomer to ECG recording might ask why it is necessary to ground the patient at all if we only wish to measure the difference in cardiac potentials at different locations. The answer is a ground is not necessary. Recent improvements have allowed not only the ECG amplifier but the power source as well to be completely isolated from the patient and input circuitry [18], [19]. This reduces any leakage current from the ECG equipment. If the patient is not grounded by some other path, there is no return for displacement currents in the body. This type of amplifier has no ground electrode attached to the patient and is becoming increasingly popular for reasons of patient safety. As when using radiotelemetry, only two electrodes are required, which is certainly more convenient. If the patient is grounded either inadvertently or on purpose, the amplifier will still not see a CM potential. This method will be sensitive to differential

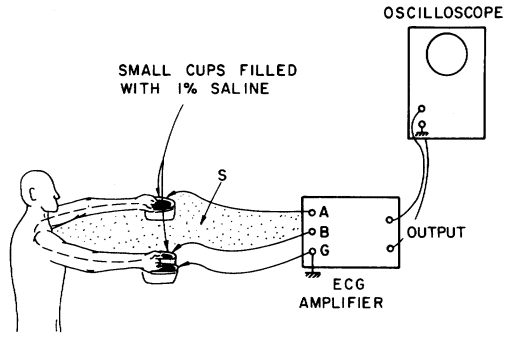

Fig. 11. Unusual ECG recording configuration used for testing a large number of people quickly.

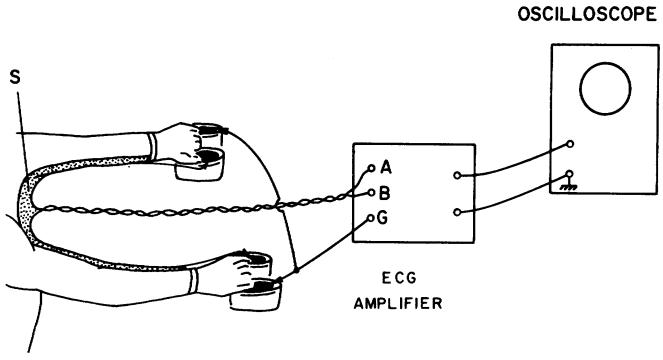

Fig. 12. Improved ECG recording configuration for reduced 60-Hz interference.

potentials on the body as a result of displacement currents flowing through the body's internal resistance and also magnetic induction.

Example

An example of how the variables described in the paper might be applied to an unusual recording situation is given below. We wished to find an inexpensive configuration to display ECG's quickly from a large number of people. This type of quick and portable test could conceivably be employed to record the ECG of all the people in a village or all the children in a grade school in a short amount of time. The configuration in Fig. 11 was first tested at an exposition to display the ECG's of the visiting public. The apparatus consists of an ECG amplifier, oscilloscope, and three silver-silverchloride electrodes mounted in small cups of 1-percent salt solution. Each person placed his fingertips in the cups and immediately could observe his own ECG in real time.

This system was not completely satisfactory because, in addition to the muscle potentials normal for any ECG taken from the hands, there was also high interference and some spurious radiations from other exhibits on the output waveform. The interference was about 200 μV at the input of the amplifier. This is 20 times greater than our 1-percent criterion. We proceeded to apply the quick tests given in this article to isolate and cure the problem.

First we considered magnetic induction interference. The area of the loop in this case was S, indicated in Fig. 11. The electrode leads A and B were twisted and run to make the area as small as possible, as shown in Fig. 12. The interference decreased to 160 μV. Then we placed three fingers of one hand in the three cups to test for currents entering the leads and found the output to be free of interference. The electrode leads were shielded nevertheless to keep EM fields from causing problems (there was an amateur radio station operating nearby). We then checked to see if displacement currents into the body were causing the problem. Replacing the grounding cup with a standard chest electrode made the interference negligible. Because we did not wish to use electrodes that required removal of clothing or electrode paste, we added a second grounding electrode cup on the other hand.

The problem was the body displacement current flowing through the high resistance of the arm, hand, and finger. The second ground, on the other hand, provides another path to ground for this current and balances the potential difference due to the body-displacement current. This configuration has proved satisfactory. An improved configuration, using two electrodes instead of four, could be developed using radiotelemetry.

Summary

Interference can be described by the equation

$$V_{\text{total}} = \underbrace{KBS}_{\substack{\text{magnetic}\\ \text{induction}}} + \underbrace{I_{D1}Z_1 - I_{D2}Z_2}_{\substack{I_D \text{ in}\\ \text{leads}}} + \underbrace{K'I_DZ_I}_{\substack{I_D \text{ in}\\ \text{body}}} + \underbrace{I_DZ_G}_{\substack{\text{CM}\\ \text{ac}\\ \text{potential}}} \underbrace{\left[\frac{1}{X}\right]}_{\substack{\text{CMRR}\\ \text{factor}}} + \underbrace{\left(\frac{Z'_{\text{in}}}{Z'_{\text{in}}+Z_1} - \frac{Z''_{\text{in}}}{Z''_{\text{in}}+Z_2}\right)}_{\substack{\text{potential}\\ \text{divider}\\ \text{effect}}}. \quad (27)$$

It is apparent that not all the factors in (27) produce noticeable interference in all situations where interference is present. Operating practice and equipment design will determine which factors are the most important in any particular application. The tests to isolate these factors are designed to quickly locate the source of interference and eliminate it so that an interference-free ECG can be recorded.

We conclude that 1) magnetic induction is an often overlooked, but nonetheless important source of interference that can be prevented by merely twisting the input leads and running them close to the body; 2) interference from displacement current in the unshielded electrode leads is seldom a problem, unless there is a large unbalance in the electrode impedances or the leads are placed such that the displacement current into one lead is much greater than that into the other; and 3) interference caused by unsymmetrical electrode placement relative to differential ac potentials on the body can be reduced by moving the ground electrode. We find the driven right-leg amplifier design to be the cheapest and safest method of eliminating ac interference caused by electrode impedance unbalance or low amplifier CMRR and input impedance.

Do not tolerate interference in your ECG recordings. Spend some time applying these simple tests to isolate and remedy the factor that is causing the problem.

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An IC Piezoresistive Pressure Sensor for Biomedical Instrumentation

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Abstract—**A thin-diaphragm piezoresistive pressure sensor for biomedical instrumentation has been developed using monolithic integrated-circuit (IC) techniques. The piezoresistive effect has been chosen for this device because it provides an observable resistance change that is a linear function of pressure and is observable at low stress levels. A diaphragm is used as a stress magnifying device; its magnification is proportional to the square of the ratio of the diaphragm diameter to its thickness. The pressure-induced stresses in the diaphragm are sensed by properly oriented piezoresistors interconnected to form a bridge.**

An anisotropic etching technique is used for the formation of the diaphragms; this technique makes possible a novel thickness monitoring scheme that also acts as a chip separation etch. Sensors with diaphragm diameters of 0.5 mm and thickness of only 5 μm, surrounded by a 0.15-mm wide ring of thick silicon, have been batch fabricated using this technique. An intrinsic sensitivity of 14 μV/V supply/mmHg has been achieved.

Temperature drift in these sensors is dominated by the temperature dependence of the piezoresistive coefficient. A temperature-compensation circuit has been devised for these sensors by deriving a temperature-dependent signal that is pressure independent for the compensation of the temperature-dependent part of the bridge unbalance voltage.

These sensors, after being mounted on the tip of a small catheter, can be inserted into the biological system through the inner bore of a larger catheter that was formerly occupied by a guide wire. The sensors have been utilized for acute measurements of blood pressure in dogs with satisfactory results.

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Introduction

THE MOST COMMON technique for obtaining reliable pressure measurements in biological systems utilizes a flexible stainless steel guide wire about 1 mm in diameter that is inserted into the artery. This guide wire is pushed to the desired location under fluoroscopic monitoring, used as a