6

BIOPOTENTIAL AMPLIFIERS

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Amplifiers are an important part of modern instrumentation systems for measuring biopotentials. Such measurements involve voltages that often are at low levels, have high source impedances, or both. Amplifiers are required to increase signal strength while maintaining high fidelity. Amplifiers that have been designed specifically for this type of processing of biopotentials are known as *biopotential amplifiers*. In this chapter we examine some of the basic features of biopotential amplifiers and also look at specialized systems.

6.1 BASIC REQUIREMENTS

The essential function of a biopotential amplifier is to take a weak electric signal of biological origin and increase its amplitude so that it can be further processed, recorded, or displayed. Usually such amplifiers are in the form of voltage amplifiers, because they are capable of increasing the voltage level of a signal. Nonetheless, voltage amplifiers also serve to increase power levels, so they can be considered power amplifiers as well. In some cases, biopotential amplifiers are used to isolate the load from the source. In this situation, the amplifiers provide only current gain, leaving the voltage levels essentially unchanged.

To be useful biologically, all biopotential amplifiers must meet certain basic requirements. They must have high input impedance, so that they provide minimal loading of the signal being measured. The characteristics of biopotential electrodes can be affected by the electric load they see, which, combined with excessive loading, can result in distortion of the signal. Loading effects are minimized by making the amplifier input impedance as high as possible, thereby reducing this distortion. Modern biopotential amplifiers have input impedances of at least 10 MΩ.

The input circuit of a biopotential amplifier must also provide protection to the organism being studied. Any current or potential appearing across the amplifier input terminals that is produced by the amplifier is capable of affecting the biological potential being measured. In clinical systems, electric currents from the input terminals of a biopotential amplifier can result in

microshocks or macroshocks in the patient being studied—a situation that can have grave consequences. To avoid these problems, the amplifier should have isolation and protection circuitry, so that the current through the electrode circuit can be kept at safe levels and any artifact generated by such current can be minimized.

The output circuit of a biopotential amplifier does not present so many critical problems as the input circuit. Its principal function is to drive the amplifier load, usually an indicating or recording device, in such a way as to maintain maximal fidelity and range in this readout. Therefore, the output impedance of the amplifier must be low with respect to the load impedance, and the amplifier must be capable of supplying the power required by the load.

Biopotential amplifiers must operate in that portion of the frequency spectrum in which the biopotentials that they amplify exist. Because of the low level of such signals, it is important to limit the bandwidth of the amplifier so that it is just great enough to process the signal adequately. In this way, we can obtain optimal signal-to-noise ratios (SNRs). Biopotential signals usually have amplitudes of the order of a few millivolts or less. Such signals must be amplified to levels compatible with recording and display devices. This means that most biopotential amplifiers must have high gains—of the order of 1000 or greater.

Very frequently biopotential signals are obtained from bipolar electrodes. These electrodes are often symmetrically located, electrically, with respect to ground. Under such circumstances, the most appropriate biopotential amplifier is a differential one. Because such bipolar electrodes frequently have a common-mode voltage with respect to ground that is much larger than the signal amplitude, and because the symmetry with respect to ground can be distorted, such biopotential differential amplifiers must have high common-mode-rejection ratios to minimize interference due to the common-mode signal.

A final requirement for biopotential amplifiers that are used both in medical applications and in the laboratory is that they make quick calibration possible. In recording biopotentials, the scientist and clinician need to know not only the waveforms of these signals but also their amplitudes. To provide this information, the gain of the amplifier must be well calibrated. Frequently biopotential amplifiers have a standard signal source that can be momentarily connected to the input, automatically at the start of a measurement or manually at the push of a button, to check the calibration. Biopotential amplifiers that need to have adjustable gains usually have a switch by which different, carefully calibrated fixed gains can be selected, rather than having a continuous control (such as the volume control of an audio amplifier) for adjusting the gain. Thus the gain is always known, and there is no chance of its being accidentally varied by someone bumping the gain control.

Biopotential amplifiers have additional requirements that are application-specific and that can be ascertained from an examination of each application. To illustrate some of these, let us first consider the electrocardiogram (ECG), the most frequently used application of biopotential amplifiers.

6.2 THE ELECTROCARDIOGRAPH

To learn more about biopotential amplifiers, we shall examine a typical clinical electrocardiograph. First, let us review the ECG itself.

THE ECG

As we learned in Section 4.6, the beating heart generates an electric signal that can be used as a diagnostic tool for examining some of the functions of the heart. This electric activity of the heart can be approximately represented as a vector quantity. Thus we need to know the location at which signals are detected, as well as the time dependence of the amplitude of the signals. Electrocardiographers have developed a simple model to represent the electric activity of the heart. In this model, the heart consists of an electric dipole located in the partially conducting medium of the thorax. Figure 6.1 shows a typical example. Of course in reality the heart is a much more complicated electrophysiological entity, and far more complex models are needed to represent it.

This particular field and the dipole that produces it represent the electric activity of the heart at a specific instant. At the next instant the dipole can change its magnitude and its orientation, thereby causing a change in the electric field. Once we accept this simplified model, we need not draw a field plot every time we want to discuss the dipole field of the heart. Instead, we can represent it by its dipole moment, a vector directed from the negative charge to the positive charge and having a magnitude proportional to the amount of charge (either positive or negative) multiplied by the separation of the two charges. In electrocardiography this dipole moment, known as the *cardiac vector*, is represented by **M**, as shown in Figure 6.1. As we progress through a cardiac cycle, the magnitude and direction of **M** vary because the dipole field varies.

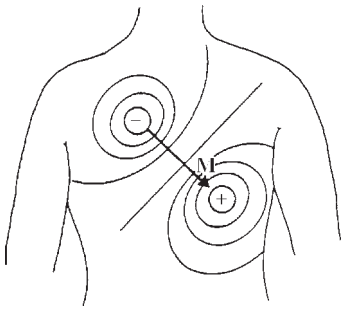


Figure 6.1 Rough sketch of the dipole field of the heart when the R wave is maximal The dipole consists of the points of equal positive and negative charge separated from one another and denoted by the dipole moment vector **M**.

The electric potentials generated by the heart appear throughout the body and on its surface. We determine potential differences by placing electrodes on the surface of the body and measuring the voltage between them, being careful to draw little current (ideally there should be no current at all, because current distorts the electric field that produces the potential differences). If the two electrodes are located on different equal-potential lines of the electric field of the heart, a nonzero potential difference or voltage is measured. Different pairs of electrodes at different locations generally yield different voltages because of the spatial dependence of the electric field of the heart. Thus it is important to have certain standard positions for clinical evaluation of the ECG. The limbs make fine guideposts for locating the ECG electrodes. We shall look at this in more detail later.

In the simplified dipole model of the heart, it would be convenient if we could predict the voltage, or at least its waveform, in a particular set of electrodes at a particular instant of time when the cardiac vector is known. We can do this if we define a *lead vector* for the pair of electrodes. This vector is a unit vector that defines the direction a constant-magnitude cardiac vector must have to generate maximal voltage in the particular pair of electrodes. A pair of electrodes, or combination of several electrodes through a resistive network that gives an equivalent pair, is referred to as a *lead*.

For a cardiac vector **M**, as shown in Figure 6.2, the voltage induced in a lead represented by the lead vector $\mathbf{a}_1$ is given by the component of **M** in the direction of $\mathbf{a}_1$. In vector algebra, this can be denoted by the dot product

$$v_{a1} = \mathbf{M} \cdot \mathbf{a}_1 \qquad \text{or} \qquad v_{a1} = |\mathbf{M}| \cos\theta \tag{6.1}$$

Where v_{a1} is the scalar voltage seen in the lead that has the vector $\mathbf{a}_1$. Let us consider another lead, represented by the lead vector $\mathbf{a}_2$, as seen in Figure 6.2. In this case, the vector is oriented in space so as to be perpendicular to the

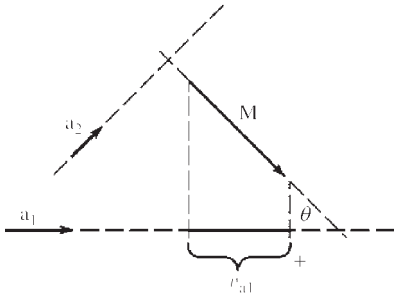


Figure 6.2 Relationships between the two lead vectors $\mathbf{a}_1$ and $\mathbf{a}_2$ and the cardiac vector **M**. The component of **M** in the direction of $\mathbf{a}_1$ is given by the dot product of these two vectors and denoted on the figure by v_{a1}. Lead vector $\mathbf{a}_2$ is perpendicular to the cardiac vector, so no voltage component is seen in this lead.

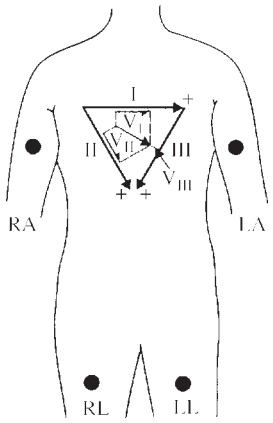


Figure 6.3 Cardiologists use a standard notation such that the direction of the lead vector for lead I is 0°, that of lead II is 60°, and that of lead III is 120°. An example of a cardiac vector at 30° with its scalar components seen for each lead is shown.

cardiac vector **M**. The component of **M** along the direction of $\mathbf{a}_2$ is zero, so no voltage is seen in this lead as a result of the cardiac vector. If we measured the ECG generated by **M** using one of the two leads shown in Figure 6.2 alone, we could not describe the cardiac vector uniquely. However, by using two leads with different lead vectors, both of which lie in the same plane as the cardiac vector such as $\mathbf{a}_1$ and $\mathbf{a}_2$, we can describe **M**.

In clinical electrocardiography, more than one lead must be recorded to describe the heart's electric activity fully. In practice, several leads are taken in the *frontal plane* (the plane of your body that is parallel to the ground when you are lying on your back) and the *transverse plane* (the plane of your body that is parallel to the ground when you are standing erect).

Three basic leads make up *the frontal-plane* ECG. These are derived from the various permutations of pairs of electrodes when one electrode is located on the right arm (RA in Figure 6.3), the left arm (LA), and the left leg (LL). Very often an electrode is also placed on the right leg (RL) and grounded or connected to special circuits, as shown in Figure 6.15. The resulting three leads are lead I, LA to RA; lead II, LL to RA; and lead III, LL to LA. The lead vectors that are formed can be approximated as an equilateral triangle, known as *Einthoven's triangle*, in the frontal plane of the body, as shown in Figure 6.3. Because the scalar signal on each lead of Einthoven's triangle can be represented as a voltage source, we can write Kirchhoff's voltage law for the three leads.

$$\mathrm{I} - \mathrm{II} + \mathrm{III} = 0 \tag{6.2}$$

The components of a particular cardiac vector can be determined easily by placing the vector within the triangle and determining its projection along each

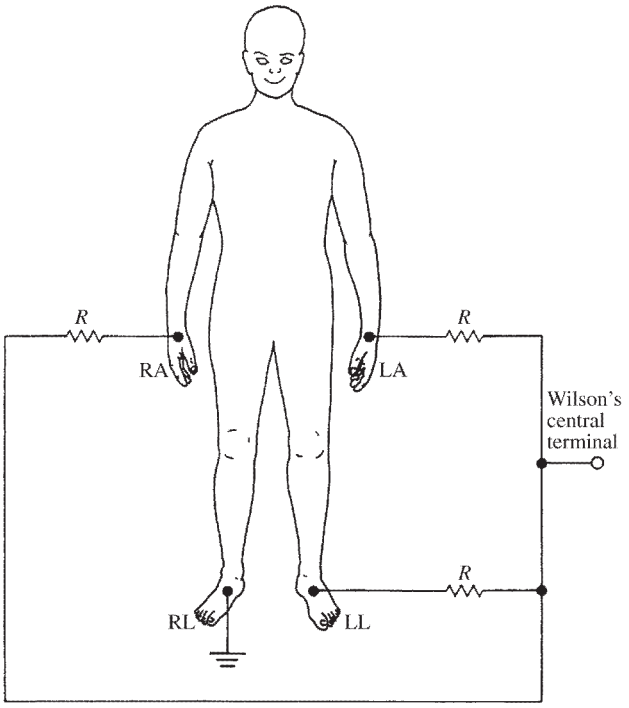


Figure 6.4 Connection of electrodes to the body to obtain Wilson's central terminal

side. The process can also be reversed, which enables us to determine the cardiac vector when we know the components along the three lead vectors, or at least two of them. It is this latter problem that usually concerns the electrocardiographer.

Three additional leads in the frontal plane—as well as a group of leads in the transverse plane—are routinely used in taking clinical ECGs. These leads are based on signals obtained from more than one pair of electrodes. They are often referred to as *unipolar leads*, because they consist of the potential appearing on one electrode taken with respect to an equivalent reference electrode, which is the average of the signals seen at two or more electrodes.

One such equivalent reference electrode is the *Wilson central terminal*, shown in Figure 6.4. Here the three limb electrodes just described are connected through equal-valued resistors to a common node. The voltage at this node, which is the Wilson central terminal, is the average of the voltages at each electrode. In practice, the values of the resistors should be at least 5 MΩ so that the loading of any particular lead will be minimal. Thus, a more practical approach is to use buffers (voltage followers, see Section 3.3) between each electrode and the equal-valued resistors. The signal between LA and the central point is known as VL, that at RA as VR, and that at the left foot as VF. Note that for each of these leads, one of the resistances R shunts the circuit

between the central terminal and the limb electrode. This tends to reduce the amplitude of the signal observed, and we can modify these leads to *augmented leads* by removing the connection between the limb being measured and the central terminal. This does not affect the direction of the lead vector but results in a 50% increase in amplitude of the signal.

The augmented leads—known as aVL, aVR, and aVF—are illustrated in Figure 6.5, which also illustrates their lead vectors, along with those of leads I, II, and III. Note that when the negative direction for aVR is considered with

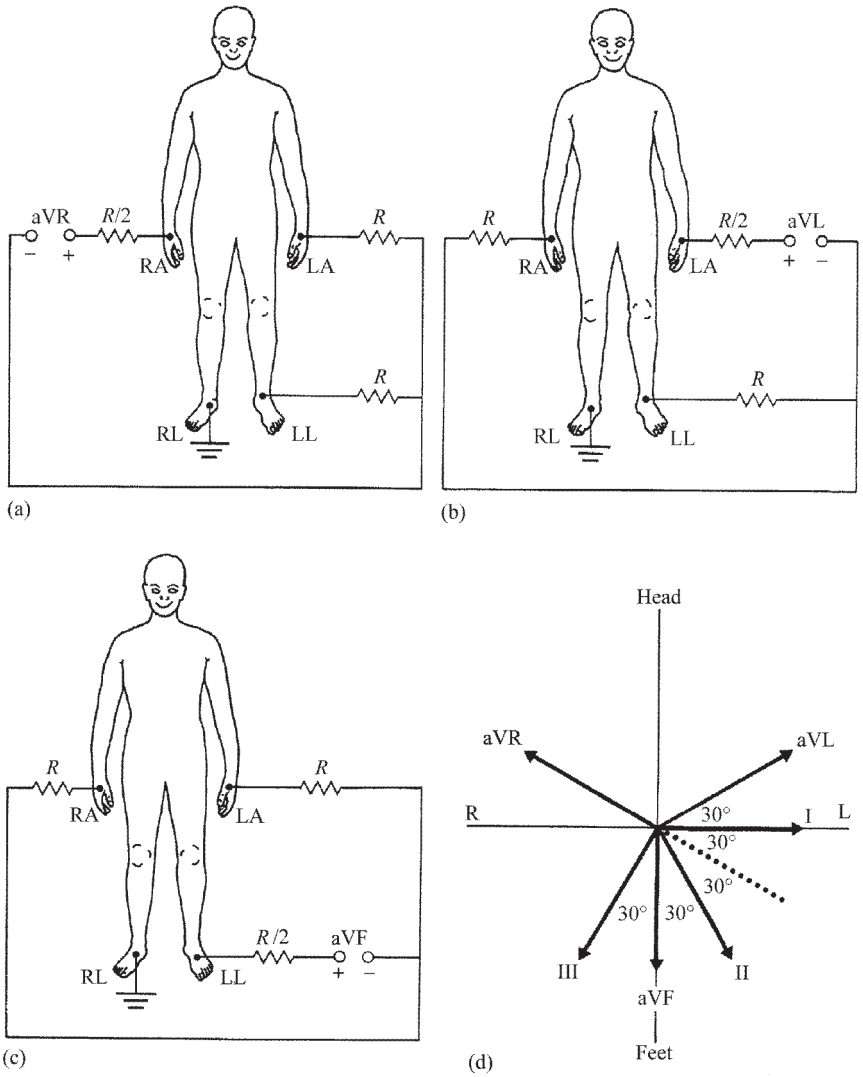


Figure 6.5 (a), (b), (c) Connections of electrodes for the three augmented limb leads. (d) Vector diagram showing standard and augmented lead-vector directions in the frontal plane.

the other five, all six vectors are equally spaced, by 30°. It is thus possible for the cardiologist looking at an ECG consisting of these six leads to estimate the position of the cardiac vector by seeing which of the six leads has the greatest signal amplitude at that point in the cardiac cycle.

EXAMPLE 6.1 Show that the voltage in lead aVR is 50% greater than that in lead VR at the same instant.

ANSWER Considering the connections for aVR and VR, we can draw the equivalent circuits of Figure E6.1(a) and (b). The voltages between each limb and ground are v_a, v_b, and v_c. When no current is drawn by the voltage measurement circuit, the negative terminal for aVR (the modified Wilson's central terminal) is at a voltage of v'_w with respect to ground, which can be determined as follows:

$$\begin{aligned} i_1 &= \frac{v_b - v_c}{2R} \\ v'_w &= i_1 R + v_c = \frac{v_b - v_c}{2R} R + v_c = \frac{v_b - v_c}{2} \end{aligned} \tag{E6.1}$$

Because no current is drawn, the positive aVR terminal (the right arm) is at a voltage v_a with respect to ground. Then aVR is

$$\text{aVR} = v_a - \frac{v_b + v_c}{2} = \frac{2v_a - v_b - v_c}{2} \tag{E6.2}$$

We can determine VR from Figure E6.1(b). To find the Wilson's central terminal voltage v_w, we simplify the circuit by taking the Thévenin equivalent circuit of the two right-hand branches. This gives the circuit shown in Figure E6.1(c) where v'_w comes from (E6.1). Now v_w is

$$v_w = \frac{v_a - v'_w}{3R/2} \frac{R}{2} + v'_w = \frac{v_a + 2v'_w}{3} \tag{E6.3}$$

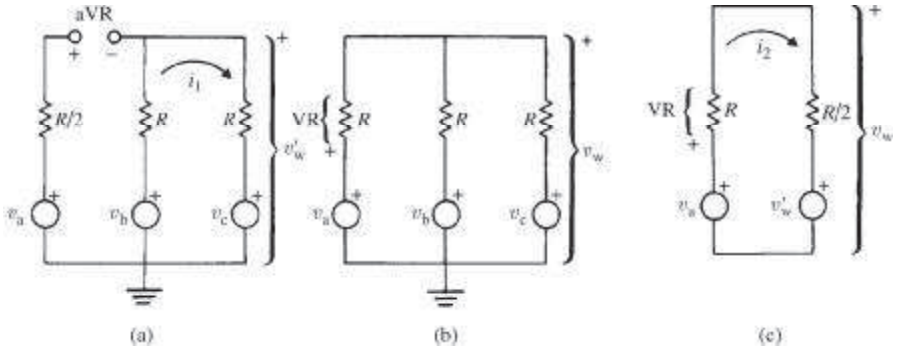


Figure E6.1 (a) aVR, (b) VR, and (c) simplified circuit of (a).

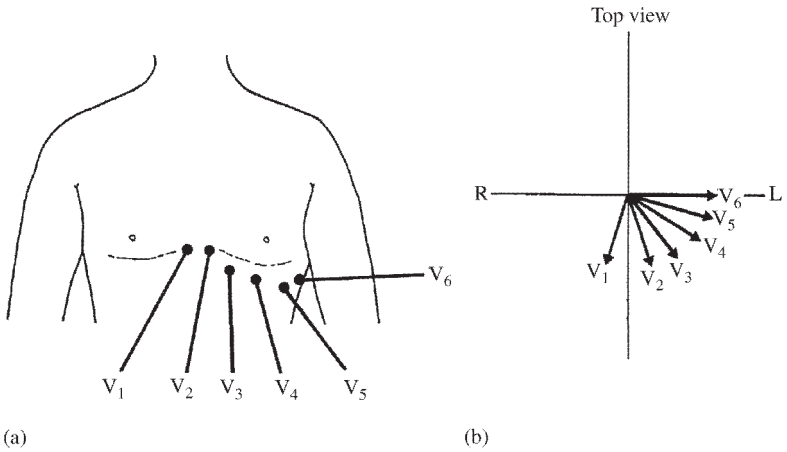


Figure 6.6 (a) Positions of precordial leads on the chest wall. (b) Directions of precordial lead vectors in the transverse plane.

$$v_w = \frac{v_a + 2(v_b + v_c)/2}{3} = \frac{v_a + v_b + v_c}{3} \tag{E6.4}$$

Thus

$$\mathrm{VR} = v_a - v_w = \frac{2v_a - v_b - v_c}{3} \tag{E6.5}$$

which shows that

$$\mathrm{aVR} = \frac{3}{2}\mathrm{VR}$$

When physicians look at the ECG in the transverse plane, they use *precordial* (chest) leads. They place an electrode at various anatomically defined positions on the chest wall, as shown in Figure 6.6. The potential between this electrode and Wilson's central terminal is the electrocardiogram for that particular lead. Figure 6.6 also shows the lead-vector positions. Physicians can obtain ECGs from the posterior side of the heart by means of an electrode placed in the esophagus. This structure passes directly behind the heart, and the potential between the esophageal electrode and Wilson's central terminal gives a posterior lead.

SPECIFIC REQUIREMENTS OF THE ELECTROCARDIOGRAPH

Because the electrocardiograph is widely used as a diagnostic tool and there are several manufacturers of this instrument, standardization is necessary. Standard requirements for electrocardiographs have been developed over the years (Bailey *et al.* 1990; Anonymous, 1991). Table 6.1 gives a summary of performance requirements from the most recent of these (Anonymous, 1991). These recommendations are a part of a voluntary standard. The Food and

Drug Administration is planning to develop mandatory standards for frequently employed instruments such as the electrocardiograph.

FUNCTIONAL BLOCKS OF THE ELECTROCARDIOGRAPH

Figure 6.7 shows a block diagram of a typical clinical electrocardiograph. To understand the overall operation of the system, let us consider each block separately.

1. *Protection circuit:* This circuit includes protection devices so that the high voltages that may appear across the input to the electrocardiograph under certain conditions do not damage it.
2. *Lead selector:* Each electrode connected to the patient is attached to the lead selector of the electrocardiograph. The function of this block is to determine which electrodes are necessary for a particular lead and to connect them to the remainder of the circuit. It is this part of the electrocardiograph in which the connections for the central terminal are made. This block can be controlled by the operator or by the

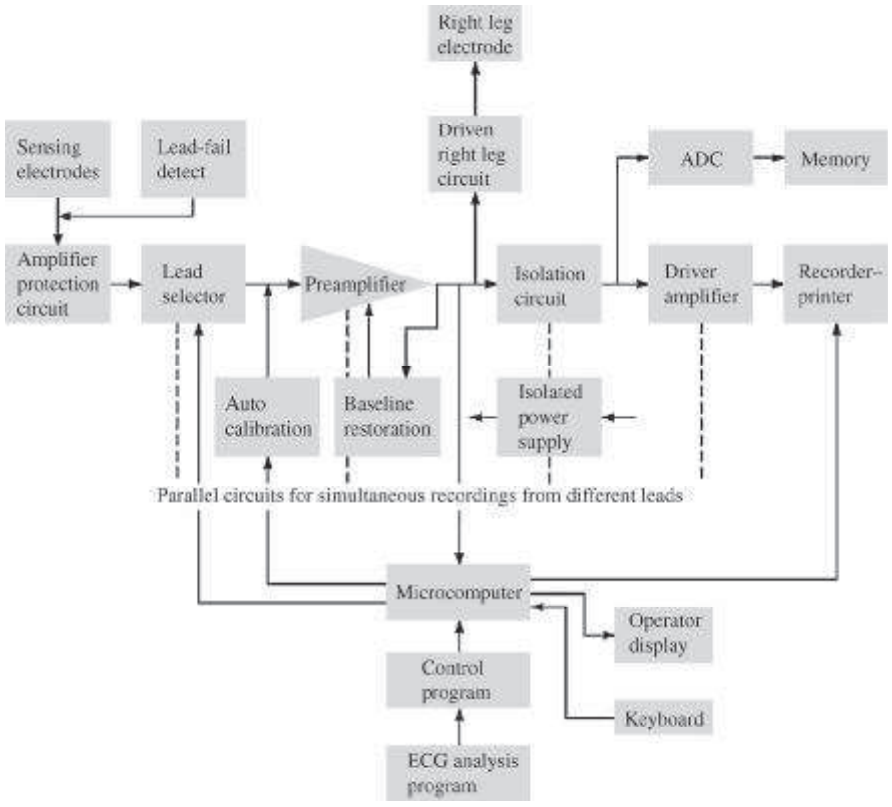


Figure 6.7 Block diagram of an electrocardiograph

microcomputer of the electrocardiograph when it is operated in automatic mode. It selects one or more leads to be recorded. In automatic mode, each of the 12 standard leads is recorded for a short duration such as 10 s.

3. *Calibration signal:* A 1 mV calibration signal is momentarily introduced into the electrocardiograph for each channel that is recorded.
4. *Preamplifier:* The input preamplifier stage carries out the initial amplification of the ECG. This stage should have very high input impedance and a high common-mode-rejection ratio (CMRR). A typical preamplifier stage is the differential amplifier that consists of three operational amplifiers (op amps), shown in Figure 3.5. A gain-control switch is often included as a part of this stage.
5. *Isolation circuit:* The circuitry of this block contains a barrier to the passage of current from the power line (50 or 60 Hz). For example, if the patient came in contact with a 120 V line, this barrier would prevent dangerous currents from flowing from the patient through the amplifier to the ground of the recorder or microcomputer.
6. *Driven-right-leg circuit:* This circuit provides a reference point on the patient that normally is at ground potential. This connection is made to an electrode on the patient's right leg. Details on this circuit are given in Section 6.5.
7. *Driver amplifier:* Circuitry in this block amplifies the ECG to a level at which it can appropriately record the signal on the recorder. Its input should be ac coupled so that offset voltages amplified by the preamplifier are not seen at its input. These dc voltages, when amplified by this stage, might cause it to saturate. This stage also carries out the bandpass filtering of the electrocardiograph to give the frequency characteristics described in Table 6.1. Also it often has a zero-offset control that is used to position the signal on the recorder. This control adjusts the dc level of the output signal.
8. *Memory system:* Many modern electrocardiographs store electrocardiograms in memory as well as printing them out on a recorder. The signal is first digitized by an analog-to-digital converter (ADC), and then samples from each lead are stored in memory. Patient information entered via the keyboard is also stored. The microcomputer controls this storage activity.
9. *Microcomputer:* The microcomputer controls the overall operation of the electrocardiograph. The operator can select several modes of operation by invoking a particular program. For example, she or he can ask the microcomputer to generate the standard 12-lead electrocardiogram by selecting three simultaneous 10 s segments of the six frontal plane leads followed by three 10 s segments of the six transverse plane leads. The microcomputer in some machines can also perform a preliminary analysis of the electrocardiogram to determine the heart rate, recognize some types of arrhythmia, calculate the axes of various features of the electrocardiogram, and determine intervals between these features. A keyboard and an alphanumeric display enable the operator to communicate with the microcomputer.

Table 6.1 Summary of Performance Requirements for Electrocardiographs (Anonymous, 1991)

Section	Requirement Description	Min/max	Units	Min/Max Value
3.2.1	Operating conditions:			
	Line voltage	Range	V rms	104 to 1127
	Frequency	Range	Hz	60 ± 1
	Temperature	Range	°C	25 ± 10
	Relative humidity	Range	%	50 ± 20
	Atmospheric pressure	Range	Pa	7×10^4 to 10.6×10^4
3.2.2	Lead definition (number of leads):	NA	NA	Table 3
	Single-channel	Min	NA	7
	Three-channel	Min	NA	12
3.2.3	Input Dynamic Range:			
	Range of linear operations of input signal	Min	mV	±5
	Slew rate change	Max	mV/s	320
	DC offset voltage range	Min	mV	±300
	Allowed variation of amplitude with dc offset	Max	%	±5
3.2.4	Gain control, accuracy, and stability:			
	Gain selections	Min	mm/mV	20, 10, 5
	Gain error	Max	%	5
	Manual override of automatic gain control	NA	NA	NA
	Gain change rate/min	Max	%/min	±0.33
	Total gain change/h	Max	%	±3
3.2.5	Time base selection and accuracy:			
	Time base selections	Min	mm/s	25, 50
	Time base error	Max	%	±5
3.2.6	Output display:			
	General	NA	NA	per 3.2.3
	Width of display	Min	mm	40
	Trace visibility (writing rates)	Max	mm/s	1600
	Trace width (permanent record only)	Max	mm	1
	Departure from time axis alignment	Max	mm	0.5
		Max	ms	10
	Preruled paper division	Min	div/cm	10
	Error of rulings	Max	%	±2
	Time marker error	Max	%	±2
3.2.7	Accuracy of input signal reproduction:			
	Overall error for signals	Max	%	±5
	Up to ±5 mV and 125 mV/s	Max	μV	±40

Table 6.1 *(Continued)*

Section	Requirement Description	Min/max	Units	Min/Max Value
	Upper cut-off frequency (3 dB)	Min	Hz	150
	Response to 20 ms, 1.5 mV triangular input	Min	mm	13.5
	Response after 3 mV, 100 ms impulse	Max	mV	0.1
		Max	mV/s	0.30
	Error in lead weighting factors	Max	%	5
	Hysteresis after 15 mm deflection from baseline	Max	mm	0.5
3.2.8	Standardizing voltage:			
	Nominal value	NA	mV	1.0
	Rise time	Max	ms	1
	Decay time	Min	s	100
	Amplitude error‘	Max	%	±5
3.2.9	Input impedance at 10 Hz (each lead)	Min	megohms	2.5
3.2.10	DC current (any input lead)	Max	μA	0.1
	DC current (any patient electrode)	Max	μA	1.0
3.2.11	Common-Mode Rejection:			
	Allowable noise with 20 V, 60 Hz and ±300 mV dc and 51 kΩ	Max	mm	10
	Imbalance	Max	mV	1
3.2.12	System noise:			30
	RTI, *p-p*	Max	μV	30
	Multichannel crosstalk	Max	%	2
3.2.13	Baseline control and stability:			
	Return time after reset	Max	s	3
	Return time after lead switch	Max	s	1
	Baseline stability:			
	Baseline drift rate RTI	Max	μV/s	10
	Total baseline drift RTI (2 min period)	Max	μV	500
3.2.14	Overload protection:			
	No damage from differential voltage, 60 Hz, 1 Vp-p, 10 s application	Min	V	1
	No damage from simulated defibrillator discharges:			
	Overvoltage	N/A	V	5000
	Energy	N/A	J	360
	Recovery time	Max	s	8
	Energy reduction by defibrillator shunting	Max	%	10
	Transfer of charge through defibrillator chassis	Max	μC	100

(*Continued*)

Table 6.1 *(Continued)*

Section	Requirement Description	Min/max	Units	Min/Max Value
	ECG display in presence of pacemaker pulses:			
	Amplitude	Range	mV	2 to 250
	Pulse duration	Range	ms	0.1 to 2.0
	Rise time	Max	μs	100
	Frequency	Max	pulses/min	100
3.2.15	Risk current (isolated patient connection)	Max	μA	10
		As per applicable document 2.11		
3.2.16	Auxiliary output (if provided):			
	No damage from short circuit risk	Max	μA	10
	Current (isolated patient connection)	As per applicable document 2.1.1		

10. *Recorder–printer:* This block provides a hard copy of the recorded ECG signal. It also prints out patient identification, clinical information entered by the operator, and the results of the automatic analysis of the electrocardiogram. Although analog oscillograph-type recorders were employed for this function in the past, modern electrocardiographs make use of thermal or electrostatic recording techniques in which the only moving part is the paper being transported under the print head (Vermariën, 2006). Digitized electrocardiograms can also be stored in permanent memory such as flash memory or optically based disk media such as CDs or DVDs.

6.3 PROBLEMS FREQUENTLY ENCOUNTERED

There are many factors that must be taken into consideration in the design and application of the electrocardiograph as well as other biopotential amplifiers. These factors are important not only to the biomedical engineer, but also to the individual who operates the instrument and the physician who interprets the recorded information. In the following paragraphs, we shall describe a few of the more common problems encountered and shall indicate some of their causes.

FREQUENCY DISTORTION

The electrocardiograph does not always meet the frequency-response standards we have described. When this happens, frequency distortion is seen in the ECG.

High-frequency distortion rounds off the sharp corners of the waveforms and diminishes the amplitude of the QRS complex.

An instrument that has a frequency response of 1 to 150 Hz shows *low-frequency distortion*. The baseline is no longer horizontal, especially immediately following any event in the tracing. Monophasic waves in the ECG appear to be more biphasic.

SATURATION OR CUTOFF DISTORTION

High offset voltages at the electrodes or improperly adjusted amplifiers in the electrocardiograph can produce saturation or cutoff distortion that can greatly modify the appearance of the ECG. The combination of input-signal amplitude and offset voltage drives the amplifier into saturation during a portion of the QRS complex (Section 3.2). The peaks of the QRS complex are cut off because the output of the amplifier cannot exceed the saturation voltage.

In a similar occurrence, the lower portions of the ECG are cut off. This can result from negative saturation of the amplifier. In this case only a portion of the S wave may be cut off. In extreme cases of this type of distortion even the P and T waves may be below the cutoff level such that only the R wave appears.

GROUND LOOPS

Patients who are having their ECGs taken on either a clinical electrocardiograph or continuously on a cardiac monitor are often connected to other pieces of electric apparatus. Each electric device has its own ground connection either through the power line or, in some cases, through a heavy ground wire attached to some ground point in the room.

A *ground loop* can exist when two machines are connected to the patient. Both the electrocardiograph and a second machine have a ground electrode attached to the patient. The electrocardiograph is grounded through the power line at a particular socket. The second machine is also grounded through the power line, but it is plugged into an entirely different outlet across the room, which has a different ground. If one ground is at a slightly higher potential than the other ground, a current from one ground flows through the patient to the ground electrode of the electrocardiograph and along its lead wire to the other ground. In addition to this current's presenting a safety problem, it can elevate the patient's body potential to some voltage above the lowest ground to which the instrumentation is attached. This produces common-mode voltages on the electrocardiograph that, if it has a poor CMRR, can increase the amount of interference seen.

OPEN LEAD WIRES

Frequently one of the wires connecting a biopotential electrode to the electrocardiograph becomes disconnected from its electrode or breaks as a result of excessively rough handling, in which case the electrode is no longer connected

to the electrocardiograph. Relatively high potentials can often be induced in the open wire as a result of electric fields emanating from the power lines or other sources in the vicinity of the machine. This causes a wide, peak-to-peak deflection of the trace on the recorder at the power-line frequency, as well as, of course, signal loss. Such a situation also arises when an electrode is not making good contact with the patient. A circuit for detecting poor electrode contact is described in Section 6.9.

ARTIFACT FROM LARGE ELECTRIC TRANSIENTS

In some situations in which a patient is having an ECG taken, cardiac defibrillation may be required (Section 13.2). In such a case, a high-voltage high-current electric pulse is applied to the chest of the patient so that transient potentials can be observed across the electrodes. These potentials can be several orders of magnitude higher than the normal potentials encountered in the ECG. Other electric sources can cause similar transients. When this situation occurs, it can cause an abrupt deflection in the ECG, as shown in Figure 6.8. This is due to the saturation of the amplifiers in the electrocardiograph caused by the relatively high-amplitude pulse or step at its input. This pulse is sufficiently large to cause the buildup of charge on coupling capacitances in the amplifier, resulting in its remaining saturated for a finite period of time following the pulse and then slowly drifting back to the original baseline with a time constant determined by the low corner frequency of the amplifier. An example of the slowly recovering waveform is shown in Figure 6.8 at a reduced amplitude and time scale to demonstrate the transient.

Transients of the type just described can be generated by means other than defibrillation. Serious artifact caused by motion of the electrodes can produce variations in potential greater than ECG potentials. Another source of artifact is the patient's encountering a built-up static electric charge that can be partially discharged through the body. Older electrocardiographs exhibit a similar transient when they are switched manually from one lead

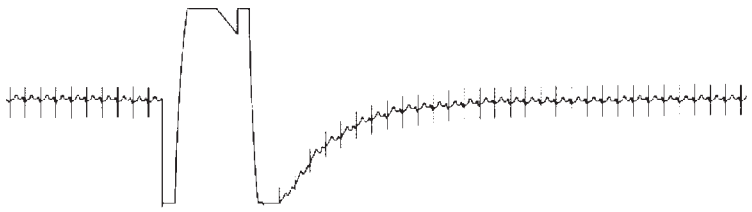


Figure 6.8 Effect of a voltage transient on an ECG recorded on an electrocardiograph in which the transient causes the amplifier to saturate and a finite period of time is required for the charge to bleed off enough to bring the ECG back into the amplifier's active region of operation. This is followed by a first-order recovery of the system.

to another, because there are different offset potentials at each electrode. This is usually not seen on newer machines that switch leads automatically, because voltages due to excess charge are discharged during the switching process.

This problem is greatly alleviated by reducing the source of the artifact. Because we do not have time to disconnect an electrocardiograph when a patient is being defibrillated, we can include electronic protection circuitry, such as that described in Section 6.4, in the machine itself. In this way, we can limit the maximal input voltage across the ECG amplifier so as to minimize the saturation and charge buildup effects due to the high-voltage input signals. This results in a more rapid return to normal operation following the transient. Such circuitry is also important in protecting the electrocardiograph from any damage that might be caused by these pulses.

Artifact caused by static electric charge on personnel can be lessened noticeably by reducing the buildup of static charge through the use of conductive clothing, shoes, and flooring, as well as by having personnel touch the bed before touching the patient. Motion artifact from the electrodes can be decreased by using the techniques described in Chapter 5.

EXAMPLE 6.2 An electrocardiograph has a broad frequency response so that its amplifier has a first-order time constant of 16 s. The electrocardiograph amplifier has a broad dynamic range of input voltages, but any input voltage greater than $\pm 2\,\text{mV}$ will be out the range of its display and cut off. While recording the ECG of a patient, a transient occurs that has an amplitude of 10 mV, and this causes the ECG to fall out of the range of the instrument's display. If the ECG R wave has an amplitude of 1 mV, how long will it take for the entire signal to be visible on the display?

ANSWER For the entire amplitude range of the ECG to be visible on the display, its baseline must be at a voltage of $2\,\text{mV} - 1\,\text{mV} = 1\,\text{mV}$. The recovery voltage at the amplifier will follow first-order exponential decay as given by

$$v = 10\,\text{mV}\, e^{-t/16\,\text{s}} \tag{E6.6}$$

This voltage must drop to 1 mV for the entire ECG waveform to be visible, so

$$\begin{aligned} 1\,\text{mV} &= 10\,\text{mV}\, e^{-t/16\,\text{s}} \\ 0.1 &= e^{-t/16\,\text{s}} \end{aligned} \tag{E6.7}$$

Solving for t, we find

$$\ln(0.1) = -\frac{t}{16\,\text{s}} = -2.303 \tag{E6.8}$$

and

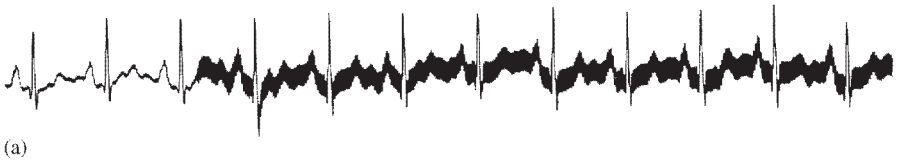
$$t = 36.8\,\text{s}.$$

INTERFERENCE FROM ELECTRIC DEVICES

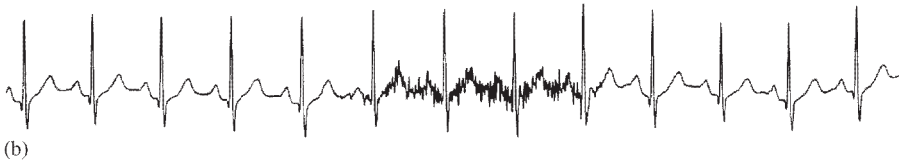
A major source of interference when one is recording or monitoring the ECG is the electric-power system. Besides providing power to the electrocardiograph itself, power lines are connected to other pieces of equipment and appliances in the typical hospital room or physician's office. There are also power lines in the walls, floor, and ceiling running past the room to other points in the building. These power lines can affect the recording of the ECG and introduce interference at the line frequency in the recorded trace, as illustrated in Figure 6.9(a). Such interference appears on the recordings as a result of two mechanisms, each operating singly or, in some cases, both operating together.

Electric-field coupling between the power lines and the electrocardiograph and/or the patient is a result of the electric fields surrounding main power lines and the power cords connecting different pieces of apparatus to electric outlets. These fields can be present even when the apparatus is not turned on, because current is not necessary to establish the electric field. These fields couple into the patient, the lead wires, and the electrocardiograph itself. It is almost as though small capacitors joined these entities to the power lines, as shown by the crude model in Figure 6.10.

The current through the capacitance C_3 coupling the ungrounded side of the power line and the electrocardiograph itself flows to ground and does not cause interference. C_1 represents the capacitance between the power line and one of the leads. Current i_{d1} does not flow into the electrocardiograph because of its high input impedance, but rather through the skin–electrode impedances Z_1 and Z_G and the subject being measured to ground. Similarly, i_{d2} flows through Z_2 and Z_G and the subject to ground. Body impedance, which is about 500 Ω, can be neglected when compared with the other



(a)



(b)

Figure 6.9 (a) A 60 Hz power-line interference. (b) Electromyographic interference on the ECG. Severe 60 Hz interference is also shown on the bottom tracing in Figure 4.13.

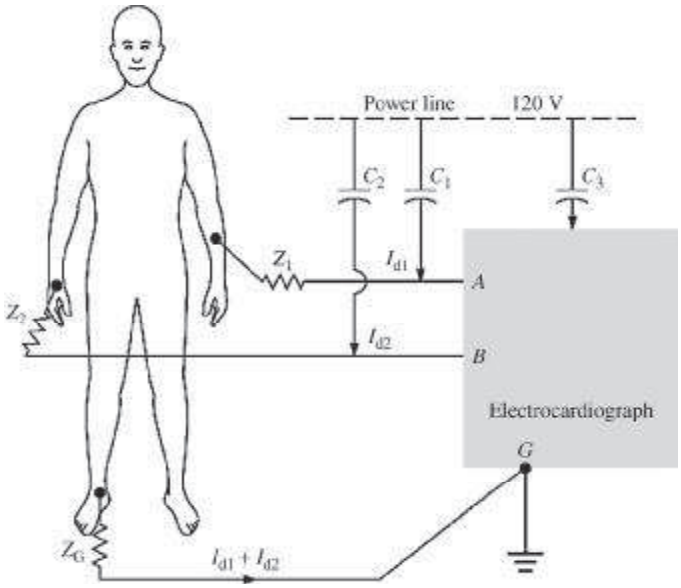


Figure 6.10 A mechanism of electric-field pickup of an electrocardiograph resulting from the power line. Coupling capacitance between the hot side of the power line and lead wires causes current to flow through skin–electrode impedances on its way to ground.

impedances shown. The voltage amplified is that appearing between inputs A and B, $v_A - v_B$.

$$v_A - v_B = i_{d1}Z_1 - i_{d2}Z_2 \tag{6.3}$$

Huhta and Webster (1973) suggest that if the two leads run near each other, $i_{d1} \cong i_{d2}$. In this case,

$$v_A - v_B = i_{d1}(Z_1 - Z_2) \tag{6.4}$$

Values measured for 9 m cables show that $i_d \cong 6\,\text{nA}$, although this value will be dependent on the room and the location of other equipment and power lines. Skin–electrode impedances may differ by as much as 20 kΩ. Hence

$$v_A - v_B = (6\,\text{nA})(20\,\text{k}\Omega) = 120\,\mu\text{V} \tag{6.5}$$

which would be an objectionable level of interference. This can be minimized by shielding the leads and grounding each shield at the electrocardiograph. This is done, in fact, in most modern electrocardiographs. Lowering skin–electrode impedances is also helpful.

Figure 6.11 shows that current also flows from the power line directly into the body. This displacement current i_{db} flows through the ground impedance

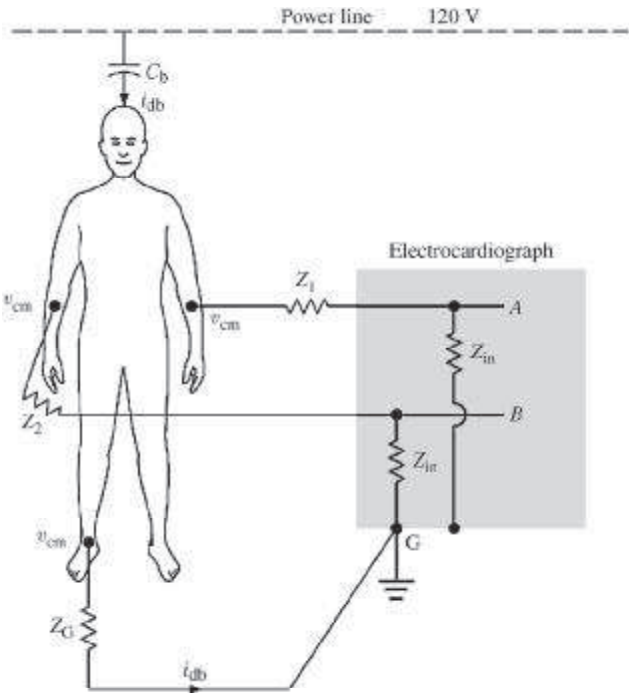


Figure 6.11 Current flows from the power line through the body and ground impedance, thus creating a common-mode voltage everywhere on the body. Z_{in} is not only resistive but, as a result of RF bypass capacitors at the amplifier input, has a reactive component as well.

Z_G to ground. The resulting voltage drop causes a common-mode voltage v_{cm} to appear throughout the body.

$$v_{cm} = i_{db} Z_G \tag{6.6}$$

Substituting typical values yields

$$v_{cm} = (0.2\,\mu A)(50\,k\Omega) = 10\,mV \tag{6.7}$$

In poor electrical environments in which $i_{db} > 1\,\mu A$, v_{cm} can be greater than 50 mV. For a perfect amplifier, this would cause no problem, because a differential amplifier rejects common-mode voltages (Section 3.4). However, real amplifiers have finite input impedances Z_{in}. Thus v_{cm} is decreased because of the attenuator action of the skin–electrode impedances and Z_{in}. That is,

$$v_A - v_B = v_{cm}\left(\frac{Z_{in}}{Z_{in} + Z_1} - \frac{Z_{in}}{Z_{in} + Z_2}\right) \tag{6.8}$$

Because Z_1 and Z_2 are much less than Z_{in},

$$v_A - v_B = v_{cm}\left(\frac{Z_2 - Z_1}{Z_{in}}\right) \tag{6.9}$$

Substituting typical values yields

$$v_A - v_B = (10\,\text{mV})(20\,\text{k}\Omega/5\,\text{M}\Omega) = 40\,\mu\text{V} \tag{E6.10}$$

which would be noticeable on an ECG and would be very objectionable on an EEG. This interference can be minimized by lowering skin–electrode impedance and raising amplifier input impedance.

Thus we see that the difference between the skin–electrode impedances is an important consideration in the design of biopotential amplifiers. Some common-mode voltage is always present, so the input imbalance and Z_{in} are critical factors determining the common-mode rejection, no matter how good the differential amplifier itself is.

EXAMPLE 6.3 A clinical staff member has attached a patient to an electroencephalograph (EEG machine) for a sleep study that continuously displays that patient's EEG on a computer screen and stores it in memory. This staff member accidently used two different types of electrodes for the EEG lead, and each electrode had a different source impedance. One had a relatively low impedance of 1500 Ω at EEG frequencies, while the other had a higher impedance of 4700 Ω. A ground electrode having an impedance of 2500 Ω was also used. The input impedance of each differential input of the EEG machine to ground was 10 MΩ, and the instrument had a CMRR of 80 dB. The power-line displacement current to the patient was measured at 400 nA. The amplitude of the patient's EEG was 12 μV.

a. How much common-mode voltage will be seen on this patient and will it significantly interfere with the EEG signal?
b. How much power-line interference will be seen on the patient's EEG?

ANSWER The common-mode voltage will be determined by the displacement current through the ground electrode impedance Z_G [see (6.6)].

a.
$$v_{cm} = 400 \times 10^{-9}\,\text{A}(2500\,\Omega) = 10^{-3}\,\text{V} \tag{E6.9}$$

The EEG machine's CMRR is 80 dB, which means that its differential gain is 10^4 times greater than its common-mode gain. Thus even though the signal-to-common-mode-noise ratio is 12/1000 at the EEG machine's input, it will be 120/1 at its readout. This should be sufficiently high to allow clinical interpretation of the EEG signal.

b. Since the common-mode interference is low, any power-line interference seen will be the result of the unbalanced impedances of the EEG electrodes. This will result in a differential signal as determined by 6.17.

$$v_a - v_b = 10^{-3}\left(\frac{4{,}700\,\Omega - 1{,}500\,\Omega}{10^6\,\Omega}\right) = 3.2 \times 10^{-6}\,\text{V} = 3.2\,\mu\text{V} \qquad \text{(E6.10)}$$

This is small compared to the 100 μV amplitude of the EEG signal and would be noticeable but tolerable interference.

The other source of interference from power lines is magnetic induction. Current in power lines establishes a *magnetic field* in the vicinity of the line. Magnetic fields can also sometimes originate from transformers and ballasts in fluorescent lights or electric appliances and other apparatus. If such magnetic fields pass through the effective single-turn coil produced by the electrocardiograph, lead wires, and the patient, as shown in Figure 6.12, a voltage is induced in this loop. This voltage is proportional to the magnetic-field strength and the area of the effective single-turn coil. It can be reduced (1) by reducing the magnetic field through the use of magnetic shielding, (2) by keeping the electrocardiograph and leads away from potential magnetic-field regions (both of which are rather difficult to achieve in practice), or (3) by reducing the effective area of the single-turn coil.

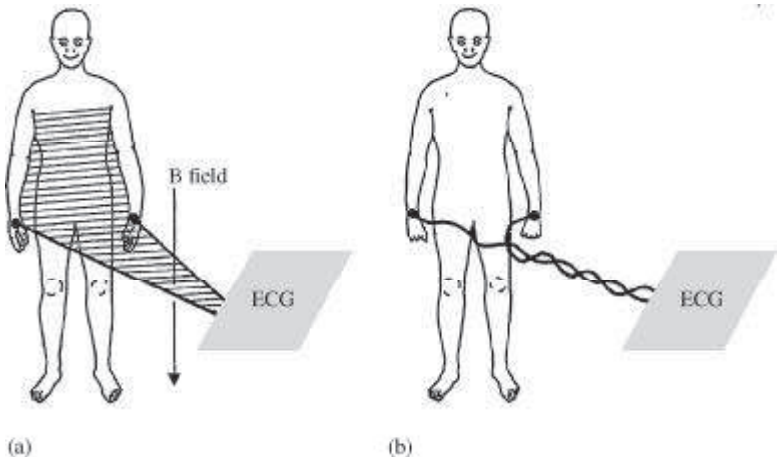


Figure 6.12 Magnetic-field pickup by the electrocardiograph (a) Lead wires for lead I make a closed loop (shaded area) when patient and electrocardiograph are considered in the circuit. The change in magnetic field passing through this area induces a current in the loop. (b) This effect can be minimized by twisting the lead wires together and keeping them close to the body in order to subtend a much smaller area.

This last approach can be achieved easily by twisting the lead wires together over as much as possible of the distance between the electrocardiograph and the patient.

OTHER SOURCES OF ELECTRIC INTERFERENCE

Electric interference from sources other than the power lines can also affect the electrocardiograph. *Electromagnetic interference* from nearby high-power radio, television, or radar facilities can be picked up and rectified by the *p–n* junctions of the transistors in the electrocardiograph and sometimes even by the electrode–electrolyte interface on the patient. Lower power electromagnetic interference can arise from local sources such as wireless devices including mobile telephones and wireless computing networks. The lead wires and the patient serve as an antenna in either case. Once the signal is detected, the demodulated signal appears as interference on the electrocardiogram. It was thought that such interference from mobile telephones could interfere with patient monitoring equipment in hospitals, but a study at the Mayo Clinic has shown this not to be a problem at their institution (Tri *et al.,* 2007).

Electromagnetic interference can also be generated by high-frequency generators in the hospital itself. Electrosurgical and diathermy (Section 13.9) equipment is a frequent offender. Grobstein and Gatzke (1977) show both the proper use of electrosurgical equipment and the design of an ECG amplifier required to minimize interference. Electromagnetic radiation can be generated from x-ray machines or switches and relays on heavy-duty electric equipment in the hospital as well. Even arcing in a fluorescent light that is flickering and in need of replacement can produce serious interference.

Electromagnetic interference can usually be minimized by shunting the input terminals to the electrocardiograph amplifier with a small capacitor of approximately 200 pF. The reactance of this capacitor is quite high over the frequency range of the ECG, so it does not appreciably lower the input impedance of the electrocardiograph. However, with today's modern high-input-impedance machines, it is important to make sure that this is really the case. At radiofrequencies, its reactance is low enough to cause effective shorting of the electromagnetic interference picked up by the lead wires and to keep it from reaching the transistors in the amplifier.

There is also a source of electric interference located within the body itself that can have an effect on ECGs. There is always some skeletal muscle located between the electrodes making up a lead of the electrocardiograph. Any time this muscle is contracting, it generates its own electromyographic signal that can be picked up by the lead along with the ECG and can result in interference on the ECG, as shown in Figure 6.9(b). When we look only at the ECG and not at the patient, it is sometimes difficult to determine whether interference of this type is muscle interference or the result of electromagnetic radiation. However, while the ECG is being taken, we can easily separate the two sources, because the EMG interference is associated with the patient's muscle contractions that can be observed when we look at the patient.

6.4 TRANSIENT PROTECTION

The isolation circuits described in Section 14.9 are primarily for the protection of the patient in that they eliminate the hazard of electric shock resulting from interaction among the patient, the electrocardiograph, and other electric devices in the patient's environment. There are also times when other equipment attached to the patient can present a risk to the machine. For example, in the operating suite, patients undergoing surgery usually have their ECGs continuously monitored during the procedure. If the surgical procedure involves the use of an electrosurgical unit (Section 13.9), it can introduce onto the patient relatively high voltages that can enter the electrocardiograph or cardiac monitor through the patient's electrodes. If the ground connection to the electrosurgical unit is faulty or if higher-than-normal resistance is present, the patient's voltage with respect to ground can become quite high during coagulation or cutting. These high potentials enter the electrocardiograph or cardiac monitor and can be large enough to damage the electronic circuitry. They can also cause severe transients, of the type shown in Figure 6.8.

Ideally, cardiac monitors and electrocardiographs should be designed so that they are unaffected by such transients. Unfortunately, this cannot be achieved completely. However, it is possible to reduce the effects of these electric transients and to protect the equipment from serious damage. Figure 6.13 shows the basic arrangement of such protective circuits. Two-terminal voltage-limiting devices are connected between each patient electrode and electric ground.

Figure 6.14(a) shows the typical current–voltage characteristic of such a device. At voltages less than V_b, the breakdown voltage, the device allows very little current to flow and ideally appears as an open circuit. Once the voltage across the device attempts to exceed V_b, the characteristics of the device sharply change, and current passes through the device to such an extent that the voltage cannot exceed V_b as a result of the voltage drop across the

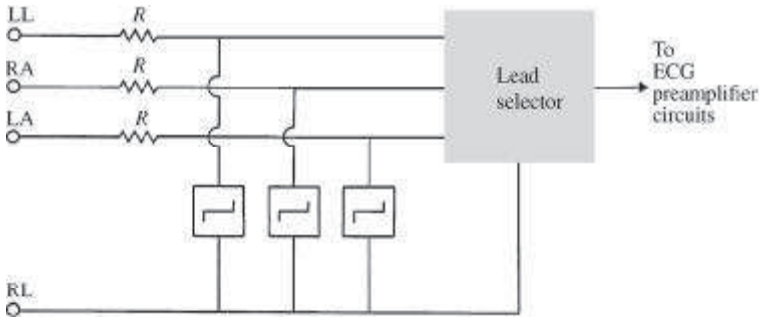


Figure 6.13 A voltage-protection scheme at the input of an electrocardiograph to protect the machine from high-voltage transients. Circuit elements connected across limb leads on left-hand side are voltage-limiting devices.

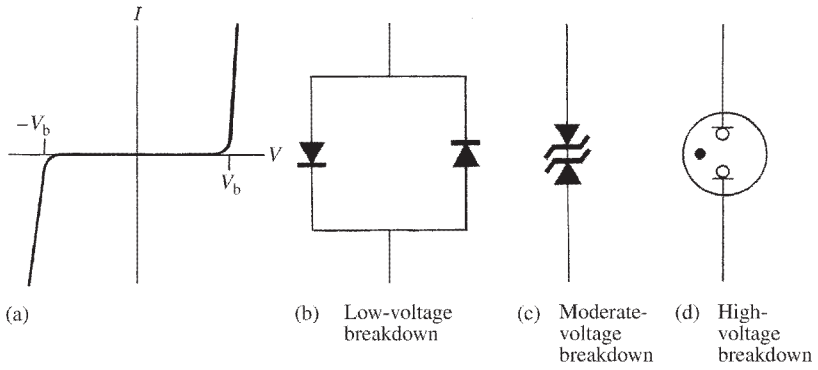


Figure 6.14 Voltage-limiting devices (a) Current–voltage characteristics of a voltage-limiting device. (b) Parallel silicon-diode voltage-limiting circuit. (c) Back-to-back silicon zener-diode voltage-limiting circuit. (d) Gas-discharge tube (neon light) voltage-limiting circuit element.

series resistors R (in Figure 6.13). Under these conditions, the device appears to behave as a short circuit in series with a constant-voltage source of magnitude V_b.

In practice, there are several ways to achieve a characteristic approaching this idealized characteristic. Figure 6.14 indicates three of these. *Parallel silicon diodes*, as shown in Figure 6.14(b), give a characteristic with a breakdown voltage of approximately 600 mV. The diodes are connected such that the terminal voltage on one has a polarity opposite that on the other. Thus, when the voltage reaches approximately 600 mV, one of the diodes is forward-biased. And even though the other is reverse-biased, its bias voltage is limited to the forward voltage drop. When the voltage across the network is reversed, the roles of the two diodes are reversed, again limiting the voltage across the network to approximately 600 mV. The transition from nonconducting state to conducting state, however, is not so sharp as shown in the characteristic curve, and signal distortion can begin to appear from these diodes at voltages of approximately 300 mV. Although the ECG itself does not approach such a voltage, it is possible under extreme conditions for dc-offset potentials of that order of magnitude to result from faulty electrodes. The main advantage of this circuit is its low breakdown voltage; the maximal transients at the amplifier input are only approximately 600 mV peak amplitude.

Because the breakdown voltage of this circuit is too small, it is usually increased simply by connecting two or three diodes in series instead of using single diodes in each branch. This has the advantage of not only increasing the breakdown voltage by multiplying the initial 600 mV by the number of diodes in series but also increasing the resistance of the circuit, both in the conducting and the nonconducting state.

When we want higher breakdown voltages, we can use the circuit of Figure 6.14(c). This circuit consists of two silicon diodes, usually *zener diodes*, connected back to back. When a voltage is connected across this circuit, one of

the diodes is biased in the forward direction and the other in the reverse direction. The breakdown voltage in the forward direction is approximately 600 mV, but that in the reverse direction is much higher. It generally covers the range of 2 to 20 V. Thus this circuit does not conduct until its terminal voltage exceeds the reverse breakdown of the diode by approximately 600 mV. Again, when the polarity of the circuit terminal voltage is reversed, the roles of the two diodes are interchanged.

A device that gives an even higher breakdown voltage is the *gas-discharge tube* illustrated in Figure 6.14(d). This device appears as an open circuit until it reaches its breakdown voltage. It then switches to the conducting state and maintains a voltage that is usually several volts less than the breakdown voltage. Breakdown voltages ranging from 50 to 90 V are typical for this device. This breakdown voltage is considered high for the input to most electrocardiographic amplifiers. Thus it is important to include a circuit element such as a resistor between the gas-discharge tube and the amplifier input to limit the amplifier's input current.

Designers of biopotential amplifiers often use miniature neon lamps as voltage limiters. They are essentially gas discharge tubes and are very inexpensive and have a symmetric characteristic, requiring only a single device per electrode pair. Their resistance in the nonconducting state is nearly infinite, so there is no loading effect on the electrodes—a feature that is most desirable when the biopotential amplifier has very high input impedance.

6.5 COMMON-MODE AND OTHER INTERFERENCE-REDUCTION CIRCUITS

As we noted earlier, common-mode voltages can be responsible for much of the interference in biopotential amplifiers. Although having an amplifier with a high CMRR minimizes the effects of common-mode voltages, a better approach to this problem is to discover the source of the voltage and try to eliminate it. In this section, we shall look at some of the sources of this and other types of interference to discover ways in which they can be minimized.

ELECTRIC- AND MAGNETIC-FIELD INTERFERENCE

As we saw in Section 6.3, electric interference can be introduced in systems of biopotential measurement through capacitive coupling and magnetic induction. We can minimize these interfering signals by trying to eliminate the sources of the signals via shielding techniques. Electrostatic shielding is accomplished by placing a grounded conducting plane between the source of the electric field and the measurement system. The measurement of very-low-level biopotentials, such as the EEG, has traditionally been carried out in a shielded enclosure containing either continuous solid-metal panels or at least grounded copper screening to minimize interference. Today, high-quality

differential instrumentation amplifiers with high CMRRs make such shielding unnecessary.

This type of shielding is ineffective for magnetic fields unless the metal panels have a high permeability (such as sheet steel or mumetal, a high permeability alloy). In other words, the panels must be good magnetic conductors as well as good electric conductors. Such rooms are available to provide magnetic shielding, but a much less expensive way of achieving a reduction of magnetically induced signals is to reduce the effective surface area between the differential inputs to the biopotential amplifier, in the case of differential signals, and between the inputs and ground, in the case of common-mode signals. Something as simple as a twisted pair of lead wires, as illustrated in Figure 6.12(b), may greatly improve the situation.

DRIVEN-RIGHT-LEG SYSTEM

In most modern electrocardiographic systems, the patient is not grounded at all. Instead, the right-leg electrode is connected (as shown in Figure 6.15) to the output of an auxiliary op amp. The common-mode voltage on the body is sensed by the two averaging resistors R_a, inverted, amplified, and fed back to the right leg. This negative feedback drives the common-mode voltage to a

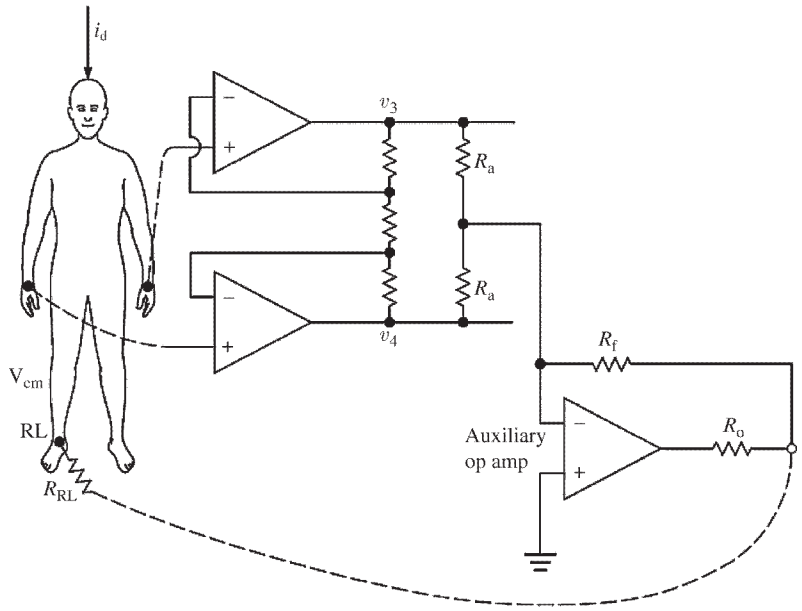


Figure 6.15 Driven-right-leg circuit for minimizing common-mode interference The circuit derives common-mode voltage from a pair of averaging resistors connected to v_3 and v_4 in Figure 3.5. The right leg is not grounded but is connected to output of the auxiliary op amp.

low value. The body's displacement current flows not to ground but rather to the op-amp output circuit. This reduces the interference as far as the ECG amplifier is concerned and effectively grounds the patient (Winter and Webster, 1983).

The circuit can also provide some electric safety. If an abnormally high voltage should appear between the patient and ground as a result of electric leakage or other cause, the auxiliary op amp in Figure 6.15 saturates. This effectively ungrounds the patient, because the amplifier can no longer drive the right leg. Now the parallel resistances R_f and R_o are between the patient and ground. They can be several megohms in value—large enough to limit the current. These resistances do not protect the patient, however, because 120 V on the patient would break down the op-amp transistors of the ECG amplifier, and large currents would flow to ground.

EXAMPLE 6.4 Determine the common-mode voltage v_{cm} on the patient in the driven-right-leg circuit of Figure 6.15 when a displacement current i_d flows to the patient from the power lines. Choose appropriate values for the resistances in the circuit so that the common-mode voltage is minimal and there is only a high-resistance path to ground when the auxiliary op amp saturates. What is v_{cm} for this circuit when $i_d = 0.2\,\mu A$?

ANSWER The equivalent circuit for the circuit of Figure 6.15 is shown in Figure E6.2. Note that because the common-mode gain of the input stage is 1 (Section 3.4) and because the input stage as shown has a very high input impedance, v_{cm} at the input is isolated from the output circuit. R_{RL} represents the resistance of the right-leg electrode. Summing the currents at the negative input of the op amp, we get

$$\frac{2v_{cm}}{R_a} + \frac{v_o}{R_f} = 0 \tag{E6.11}$$

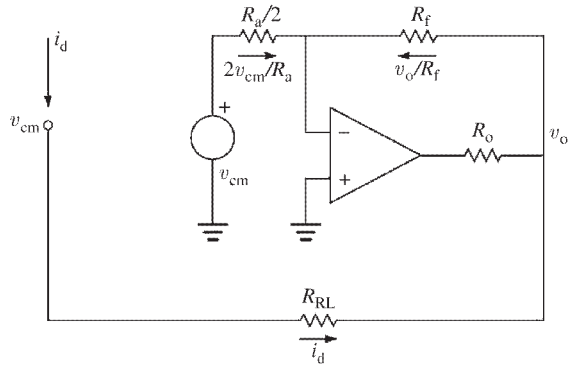


Figure E6.2 Equivalent circuit of driven-right-leg system of Figure 6.15.

This gives

$$v_o = -\frac{2R_f}{R_a} v_{cm} \tag{E6.12}$$

but

$$v_{cm} = R_{RL} i_d + v_o \tag{E6.13}$$

Thus, substituting (E6.2) into (E6.3) yields

$$v_{cm} = \frac{R_{RL} i_d}{1 + 2R_f/R_a} \tag{E6.14}$$

The effective resistance between the right leg and ground is the resistance of the right-leg electrode divided by 1 plus the gain of the auxiliary op-amp circuit. When the amplifier saturates, as would occur during a large transient v_{cm}, its output appears as the saturation voltage v_s. The right leg is now connected to ground through this source and the parallel resistances R_f and R_o. To limit the current, R_f and R_o should be large. Values as high as 5 MΩ are used.

When the amplifier is not saturated, we would like v_{cm} to be as small as possible or, in other words, to be an effective low-resistance path to ground. This can be achieved by making R_f large and R_a relatively small. R_f can be equal to R_o, but R_a can be much smaller.

A typical value of R_a would be 25 kΩ. A worst-case electrode resistance R_{RL} would be 100 kΩ. The effective resistance between the right leg and ground would then be

$$\frac{100\,\text{k}\Omega}{1 + \dfrac{2 \times 5\,\text{M}\Omega}{25\,\text{k}\Omega}} = 249\,\Omega$$

For the 0.2 μA displacement current, the common-mode voltage is

$$v_{cm} = 249\,\Omega \times 0.2\,\mu\text{A} = 50\,\mu\text{V}$$

6.6 AMPLIFIERS FOR OTHER BIOPOTENTIAL SIGNALS

Up to this point we have stressed biopotential amplifiers for the ECG. Amplifiers for use with other biopotentials are essentially the same. However, other signals do put different constraints on some aspects of the amplifier. The frequency content of different biopotentials covers different portions of the spectrum. Some biopotentials have higher amplitudes than others. Both these facts place gain and

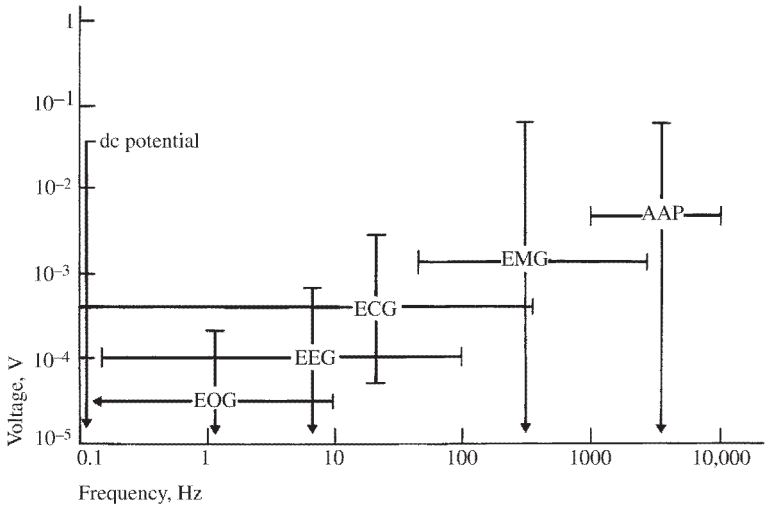


Figure 6.16 Voltage and frequency ranges of some common biopotential signals; dc potentials include intracellular voltages as well as voltages measured from several points on the body. EOG is the electro-oculogram, EEG is the electroencephalogram, ECG is the electrocardiogram, EMG is the electromyogram, and AAP is the axon action potential. [From J. M. R. Delgado, "Electrodes for Extracellular Recording and Stimulation." In W. L. Nastuk (ed.), *Physical Techniques in Biological Research*. New York: Academic Press, 1964.]

frequency-response constraints on the amplifiers used. Figure 6.16 shows the ranges of amplitudes and frequencies covered by several of the common biopotential signals. Depending on the signal, frequencies range from dc to about 10 kHz. Amplitudes can range from tens of microvolts to approximately 100 mV. The amplifier for a particular biopotential must be designed to handle that potential and to provide an appropriate signal at its output.

The electrodes used to obtain the biopotential place certain constraints on the amplifier input stage. To achieve the most effective signal transfer, the amplifier must be matched to the electrodes. Also, the amplifier input circuit must not promote the generation of artifact by the electrode, as could occur with excessive bias current. Let us look at a few requirements placed on different types of biopotential amplifiers by the measurement being made.

ELECTROMYOGRAPHY AMPLIFIER

Figure 6.16 shows that electromyographic signals range in frequency from 25 Hz to several kilohertz. Signal amplitudes range from 100 μV to 90 mV, depending on the type of signal and electrodes used. Thus electromyography (EMG) amplifiers must have a wider frequency response than ECG amplifiers, but they do not have to cover so low a frequency range as the ECGs. This is desirable because motion artifact contains mostly low frequencies that can be

filtered more effectively in EMG amplifiers than in ECG amplifiers without affecting the signal.

If skin-surface electrodes are used to detect the EMG, the levels of signals are generally low, having peak amplitudes of the order of 0.1 to 1 mV. Electrode impedance is relatively low, ranging from about 200 to 5000 Ω, depending on the type of electrode, the electrode–electrolyte interface, and the frequency at which the impedance is determined. Thus the amplifier must have somewhat higher gain than the ECG amplifier for the same output-signal range, and its input characteristics should be almost the same as those of the ECG amplifier. When intramuscular needle electrodes are used, the EMG signals can be an order of magnitude stronger, thus requiring an order of magnitude less gain. Furthermore, the surface area of the EMG needle electrode is much less than that of the surface electrode, so its source impedance is higher. Therefore, a higher amplifier input impedance is desirable for quality signal reproduction.

AMPLIFIERS FOR USE WITH GLASS MICROPIPETTE INTRACELLULAR ELECTRODES

Intracellular electrodes or microelectrodes that can measure the potential across the cell membrane generally detect potentials on the order of 50 to 100 mV. Their small size and small effective surface-contact area give them a very high source impedance, and their geometry results in a relatively large shunting capacitance. These features place on the amplifier the constraint of requiring an extremely high input impedance. Furthermore, the high shunting capacitance of the electrode itself affects the frequency-response characteristics of the system. Often positive-feedback schemes are used in the biopotential amplifier to provide an effective negative capacitance that can compensate for the high shunt capacitance of the source.

The frequency response of microelectrode amplifiers must be quite wide. Intracellular electrodes are often used to measure the dc potential difference across a cell membrane, so the amplifier must be capable of responding to dc signals. When excitable cell-membrane potentials are to be measured, such as in muscle cells and nerve cells, rise times can contain frequencies of the order of 10 kHz, and the amplifiers must be capable of passing these, too. The fact that the potentials are relatively high means that the voltage gain of the amplifier does not have to be as high as in previous examples.

A preamplifier circuit that is especially useful with microelectrodes is the negative-input-capacitance amplifier shown in Figure 6.17. The basic circuit consists of a low-gain, very-high-input-impedance, noninverting amplifier with a capacitor C_f providing positive feedback to the input. If we look at the equivalent circuit for this amplifier [Figure 6.17(b)], we can relate the input voltage and current:

$$v_i = \frac{1}{C_f}\int i_i\,dt + A_v v_i \tag{6.11}$$

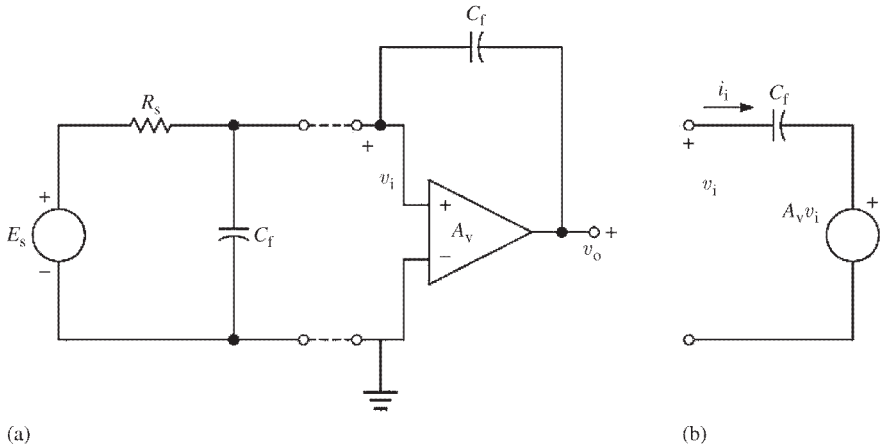


Figure 6.17 (a) Basic arrangement for negative-input-capacitance amplifier. Basic amplifier is on the right-hand side; equivalent source with lumped series resistance R_s and shunt capacitance C_s is on the left. (b) Equivalent circuit of basic negative-input-capacitance amplifier.

where A_v is the amplifier gain, provided the op amp itself draws no current. This equation can be rearranged as follows:

$$v_i = \frac{1}{(1 - A_v)C_f} \int i_i \, dt \tag{6.12}$$

Thus the equivalent capacitance at the amplifier input is $(1 - A_v)C_f$. If A_v is greater than unity, this equivalent capacitance is negative. The amplifier is connected to the microelectrode, with its high source resistance R_s. The shunt capacitance from the electrode and cable is C_s. The total circuit capacitance is

$$C = C_s + (1 - A_v)C_f \tag{6.13}$$

which is zero when

$$C_s = (A_v - 1)C_f \tag{6.14}$$

This condition can be met by adjusting either the amplifier gain A_v or the feedback capacitance C_f.

In practical negative-input-capacitance amplifiers, the idealized condition of (6.14) cannot be met, because the gain of any amplifier has some frequency dependence. There are thus frequencies where the input capacitance of the amplifier does not cancel the source capacitance, and the circuit does not have an ideal transient response. The amplifier employs positive feedback and does not have an ideal frequency response, so it is possible for the conditions of oscillation to be met at some frequency, and the amplifier will then become unstable. Thus it is important that the amplifier be carefully adjusted to meet

the condition of (6.14) as closely as possible without becoming unstable. Another consequence of the positive feedback is that the amplifier tends to be noisy. This is not a serious problem, however, because the voltages from microelectrodes are usually relatively high. A noninverting amplifier with adjustable gain greater than one can be used to drive a shield around the wire leading to the + input. Then the stray capacitance between the wire and the shield can serve as C_f and the shield minimizes interference.

ELECTROENCEPHALOGRAPH AMPLIFIERS

Figure 6.16 shows that the electroencephalograph (EEG) requires an amplifier with a frequency response of from 0.1 to 100 Hz. When surface electrodes are used, as in clinical electroencephalography, amplitudes of signals range from 25 to 100 μV. Thus amplifiers with relatively high gain are required. These electrodes are smaller than those used for the ECG, so they have somewhat higher source impedances, and a high input impedance is essential in the EEG amplifier. Because the signal levels are so small, common-mode voltages can have more serious effects. Therefore, more stringent efforts must be made to reduce common-mode interference, as well as to use amplifiers with higher CMRRs and low noise.

EXAMPLE 6.5 A small rural hospital would like to purchase an electroencephalograph but cannot afford to build a shielded room in which to measure patients' EEGs. A clinical engineer has determined that there can be common-mode noise on their patients with amplitudes as large as 100 mV. What must the minimum CMRR of their electroencephalograph be so that an EEG signal of 25 μV amplitude has no more than 1% common-mode noise?

ANSWER The SNR at the amplifier input can be as low as

$$\text{SNR} = \frac{25 \times 10^{-6}\,\text{V}}{10^{-1}\,\text{V}} = 2.5 \times 10^{-4} \tag{E6.15}$$

The SNR at the output or display of the electroencephalograph must be at least

$$\text{SNR} = (1\%)^{-1} = 100 \tag{E6.16}$$

The CMRR then must be the ratio of the output SNR to that at the input

$$\text{CMRR} = \frac{100}{2.5 \times 10^{-4}} = 4 \times 10^{5} \tag{E6.17}$$

or $20 \log_{10}\left(4 \times 10^{5}\right)\text{dB} = 112\,\text{dB}$.

This is within the range of CMRR available in high-quality differential amplifiers. (Pallás-Areny and Webster, 1990).

6.7 EXAMPLE OF A BIOPOTENTIAL PREAMPLIFIER

As we have seen, biopotential amplifiers can be used for a variety of signals. The gain and frequency response are two important variables that relate the amplifier to the particular signal. An important factor common to all amplifiers is the first stage, or preamplifier. This stage must have low noise, because its output must be amplified through the remaining stages of the amplifier, and any noise is amplified along with the signal. It must also be coupled directly to the electrodes (no series capacitors) to provide optimal low-frequency response as well as to minimize charging effects on coupling capacitors from input bias current. Of course, every attempt should be made to minimize this current. Even without coupling capacitors it can polarize the electrodes, resulting in polarization overpotentials that produce a large dc offset voltage at the amplifiers' input. This is why preamplifiers often have relatively low voltage gains. The offset potential is coupled directly to the input, so it could saturate high-gain preamplifiers, cutting out the signal altogether. To eliminate the saturating effects of this dc potential, the preamplifier can be capacitor-coupled to the remaining amplifier stages. A final consideration is that the preamplifier must have a very high input impedance, because it represents the load on the electrodes (Thakor, 1988).

Often, for safety reasons, the preamplifier either is electrically isolated from the remaining amplifier stages (and hence from the power lines) (Section 14.9) or is located near the signal source to minimize interference pickup on the high-impedance lead wires. In the latter case, we can use a battery-powered preamplifier with low power consumption or a power supply that is electrically isolated with this circuit.

Figure 6.18 shows the circuit of an ECG amplifier. The instrumentation amplifier of Figure 3.5 is used to provide very high input impedance. High common-mode rejection is achieved by adjusting the potentiometer to about 47 kΩ. Electrodes may produce an offset potential of up to 0.3 V. Thus, to prevent saturation, the dc-coupled stages have a gain of only 25. Coupling capacitors are not placed at the input because this would block the op-amp bias current. Adding resistors to supply the bias current would lower the Z_{in}. Coupling capacitors placed after the first op amps would have to be impractically large. Therefore, the single 1 μF coupling capacitor and the 3.3 MΩ resistor form a high-pass filter. The resulting 3.3 s time constant passes all frequencies above 0.05 Hz. The output stage is a noninverting amplifier that has a gain of 32 (Section 3.3).

A second 3.3 MΩ resistor is added to balance bias-current source impedances. The 150 kΩ and 0.01 μF low-pass filter attenuates frequencies above 106 Hz. Switch S_1 may be momentarily closed to decrease the discharge time constant when the output saturates. This is required after defibrillation or lead switching to charge the 1 μF capacitor rapidly to the new value and return the output to the linear region. We do *not* discharge the capacitor voltage to zero. Rather, we want the right end to be at 0 V when the left end is at the dc voltage determined by the electrode offset voltage. Switch closure may be automatic, via a circuit that detects when the output is in saturation, or it may be manual.

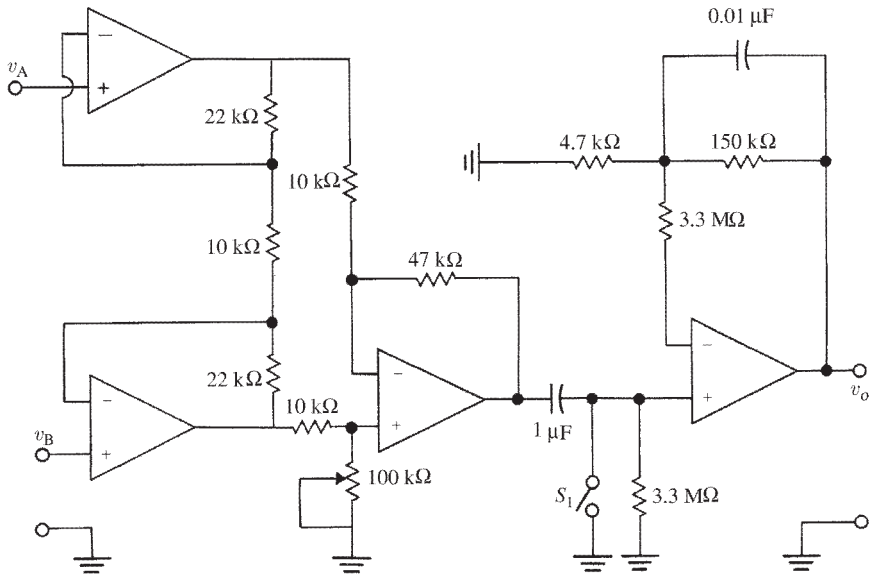


Figure 6.18 This ECG amplifier has a gain of 25 in the dc-coupled stages. The high-pass filter feeds a noninverting-amplifier stage that has a gain of 32. The total gain is $25 \times 32 = 800$. When μA 776 op amps were used, the circuit was found to have a CMRR of 86 dB at 100 Hz and a noise level of 40 mV peak to peak at the output. The frequency response was 0.05 to 106 Hz for ±3 dB and was flat over 4 to 40 Hz. A single op-amp chip, the LM 324, that contains four individual op amps could also be used in this circuit reducing the total parts count.

Although any general-purpose op amp such as the 741, 301, and 358 is satisfactory in this circuit, an op amp such as the 411, which has lower bias current, may be preferred.

Spinelli *et al.* (2004) developed an ECG amplifier based on standard low-power op amps and a single 5 V power supply. It accepts input offset voltages up to ±500 mV, yields a CMRR of 102 dB at 50 Hz, and provides a reset behavior for recovering from overloads or artifacts. Dobrev *et al.* (2008) describe a circuit that measures the ECG using two electrodes instead of the usual three.

6.8 OTHER BIOPOTENTIAL SIGNAL PROCESSORS

CARDIOTACHOMETERS

A cardiotachometer is a device for determining heart rate. The signal most frequently used is the ECG. However, software for deriving heart rate from signals such as the arterial pressure waveform, pulse oximeter pulse waves, or heart sounds has also been developed.

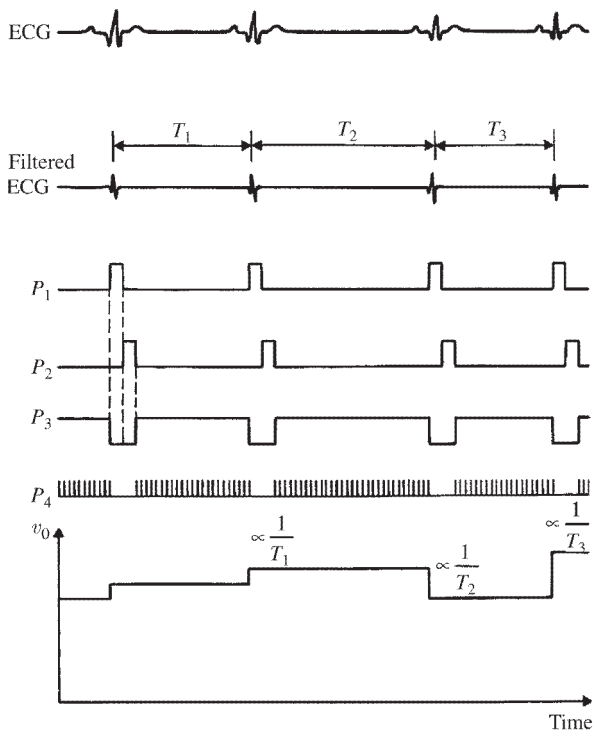


Figure 6.19 Timing diagram for beat-to-beat cardiotachometer

The beat-to-beat cardiotachometer determines the reciprocal of the time interval between heartbeats for each beat and presents it as the heart rate for that particular interval. Any slight variability in the interval between beats shows up as a variation in the instantaneous heart rate determined by this method.

Figure 6.19 shows the timing diagram beat-to-beat cardiotachometer. In software, the ECG initially passes through a bandpass filter, which passes QRS complexes while reducing artifact and most of the P and T waves. In one example, a threshold detector triggers the pulse P_1.

A 1 kHz clock signal enters a counting register whenever P_3 is high. Because P_3 is high during the interval between QRS complexes, the 1 ms pulses coming from the clock (P_4) accumulate in register 1 during this period. If the register is initially at zero, the number of pulses in the register by the time the next QRS complex arrives equals the number of milliseconds in the interval between this QRS complex and the previous one. Once the gate prohibits additional clock pulses from entering register 1, pulse P_1 enables the signal in this register to be stored in a second register, which serves as a memory. Software calculates v_o using

$$v_o = \frac{k}{T_R} \tag{6.15}$$

where k is a constant and T_R is the interval between QRS complexes. We see that v_o is proportional to the reciprocal of the beat-to-beat time interval of the original ECG; in other words, it is proportional to the heart rate. Note that this voltage shifts with each heart beat and that its amplitude is calculated from the duration of the previous beat-to-beat interval.

Alarm circuits can also be used with this type of cardiotachometer. These compare the signal in register 1 to determine whether an interval of longer than a preset value has occurred (this could happen if the heart rate were too low). Software can monitor the signal in register 2 to determine whether it is less than a preset value, a situation that would occur if the heart rate were too high. In either case, the software can then be used to activate appropriate alarms.

ELECTROMYOGRAM INTEGRATORS

It is frequently of interest to quantify the amount of EMG activity measured by a particular system of electrodes. Such quantification often assumes the form of taking the absolute value of the EMG and integrating it.

The raw EMG, amplified appropriately v_1, is fed to software, which in one example takes the absolute value. As indicated in the waveform of Figure 6.20, only positive-going signals v_2 result following this. The negative-going portions of the signal have been inverted, making them positive. Software then integrates the signal. Once the integrator output has exceeded a preset threshold level v_t, a comparator then reinitiates integration of the EMG until the cycle repeats itself.

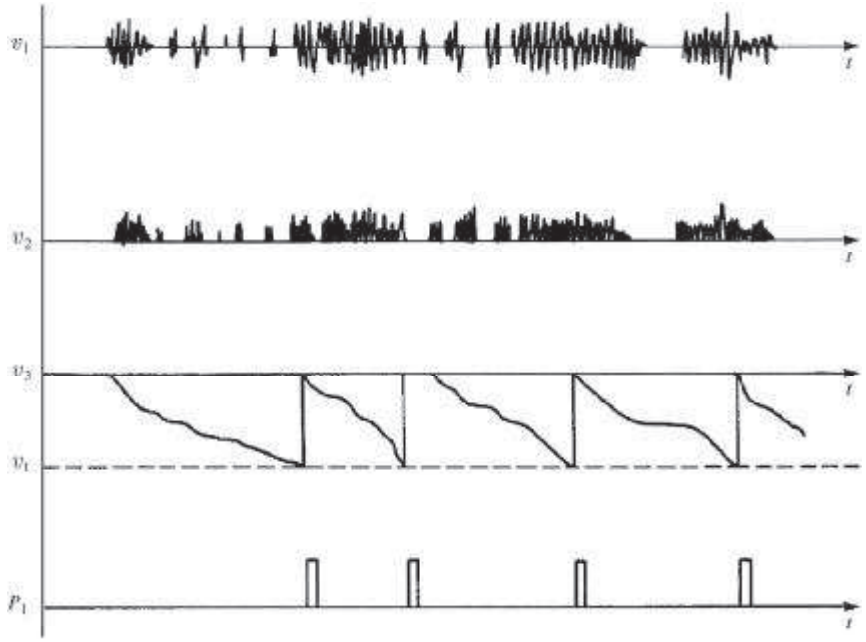


Figure 6.20 The various waveforms for the EMG integrator

We can view the output from the integrator in two ways. The actual voltage output from the integrator can be recorded on a conventional recorder or computer to give the actual integral at any instant. The total integral necessary to reset the integrator is known, so at any instant the integral equals the number of times the integrator has been reset, multiplied by this calibration constant, plus whatever is recorded as being in the integrator at that time. Another way to view the output of the integrator is to count the number of reset pulses P_t. We then determine the approximate integral by determining the number of resets over a specific time interval and calculating total activity.

EVOKED POTENTIALS AND SIGNAL AVERAGERS

Often in neurophysiology we are interested in looking at the neurological response to a particular stimulus. This response is electric in nature, and it frequently represents a very weak signal with a very poor signal-to-noise ratio (SNR). When the stimulus is repeated, the same or a very similar response is repeatedly elicited. This is the basis for biopotential signal processors that can obtain an enhanced response by means of repeated application of the stimulus (Childers, 1988).

Figure 6.21 shows how signal averaging works. The response to each stimulus is recorded. The time at which each stimulus occurs is considered the

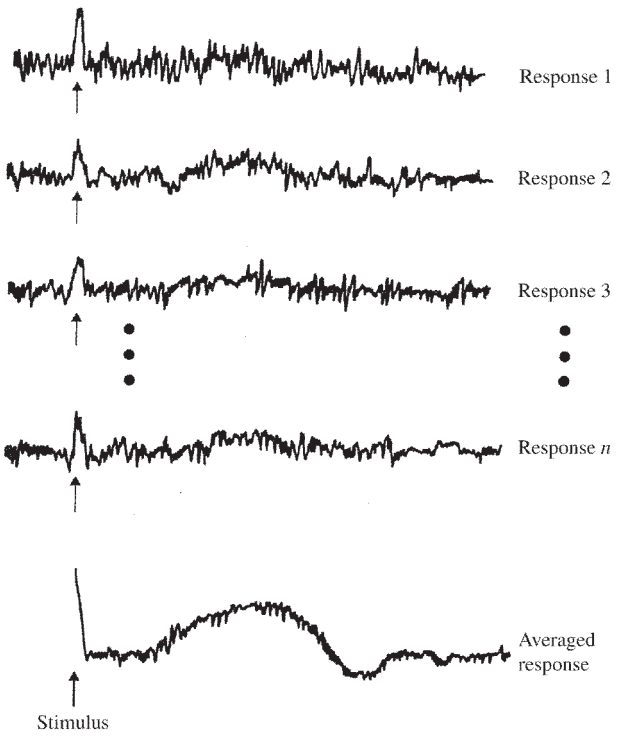


Figure 6.21 Signal-averaging technique for improving the SNR in signals that are repetitive or respond to a known stimulus.

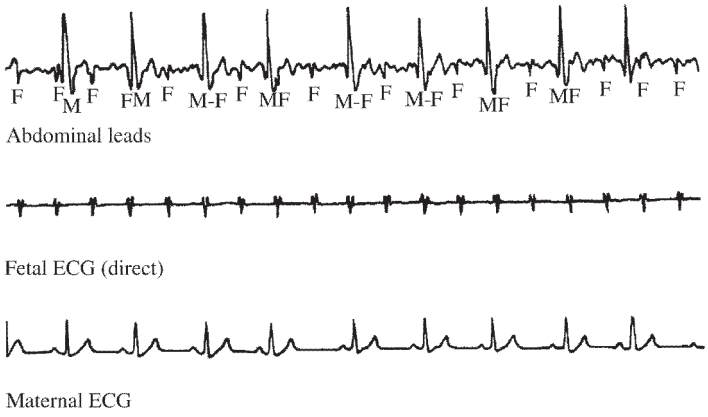


Figure 6.22 Typical fetal ECG obtained from the maternal abdomen F represents fetal QRS complexes; M represents maternal QRS complexes. Maternal ECG and fetal ECG (recorded directly from the fetus) are included for comparison. (From J. F. Roux, M. R. Neuman, and R. C. Goodlin, ''Monitoring of intrapartum phenomena.'' In *CRC Critical Reviews in Bioengineering*, 2, January 1975, pp. 119–158 Copyright CRC Press. Used by permission of CRC Press, Inc.)

reference time, and the values for each response at this reference time are summed to get the total response at the reference time. This process is repeated for the values of the responses sampled immediately after the reference time, and the sum is determined for this point in time after the stimulus. The process is then repeated for each sample point after the reference time so that a waveform that is the sum of the individual responses can be displayed, as shown at the bottom of Figure 6.22. The only limits on the number of samples that can be summed are the available memory for storing the responses and the time required to collect the data. Practical signal averaging algorithms can process more than 1000 repeated responses to extract the weak response waveform from the noise.

The noise on the individual responses is random with respect to the stimulus. This means that if a large enough sample is taken, some positive-going noise pulses at a particular instant after the stimulus partially cancel some negative-going noise spikes at the same instant. Thus the net sum of the noise at any instant following the stimulus increases as $\sqrt{n}$, where n is the number of responses. The evoked response, on the other hand, follows the same time course after each stimulus. Thus there is no cancellation in this signal as the individual responses are summed. Instead, the amplitude of the evoked response increases in direct proportion to n. By repetitive summing, one is thus able to enhance the SNR by the factor $n/\sqrt{n} = \sqrt{n}$.

This technique is frequently used with the EEG and ERG. As stated earlier in this chapter, EEGs obtained from surface electrodes are very weak and consequently can have a high noise component. When a repetitive stimulus

(such as electric shock, flashing light, or repeating sound) is applied to the test subject, it is difficult to ascertain the response in a directly recorded EEG. However, if we apply this signal summing or averaging technique, it is possible to obtain the evoked response.

EXAMPLE 6.6 The electroretinogram (ERG) from a patient had a response to a flash of light that was buried in the noise such that the SNR was 1:1. A computer can be used to average this response over many flashes to extract it from the noise. How many responses to flashes need to be averaged to improve the SNR to 10:1 (20 dB) and 100:1 (40 dB)?

ANSWER The SNR is improved by a factor of $\sqrt{n}$. so to get a 10-fold improvement we need

$$\begin{aligned} 10 &= \sqrt{n} \\ n &= (10)^2 = 100 \text{ samples averaged.} \end{aligned} \tag{E6.18}$$

For a 100-fold improvement we need

$$\begin{aligned} 100 &= \sqrt{n} \\ (100)^2 &= n = 10{,}000 \text{ samples averaged} \end{aligned} \tag{E6.19}$$

Signal averaging is usually performed on a computer. The basic scheme involves digitizing the signal and then locating the stimulus. The response is stored in memory. After the second application of the stimulus, the signal is digitized and stored, and the stimulus is located. The first sample of the response after the stimulus is added to the first sample of the response to the first stimulus, and the sum remains in memory. The second samples taken of each response are added, and so on. The summed signal can be displayed on an oscilloscope, a chart recorder or printer. The operator of the system can look at the sum after each application of the stimulus to determine how many stimuli are necessary to extract the signal from the noise adequately.

This technique can be used without applying the external stimulus. One example of its use is the recording of the ECG of a fetus. Although it is possible to record the fetal R waves from electrodes placed on the abdomen of the mother, artifacts generated by the ECG of the mother and other biopotentials, as well as by electrode noise, obscure the finer details of the fetal ECG. A signal-averaging technique similar to that we have described can be applied by using the fetal R wave in the same capacity as the stimulus. In this case the computer locates the R wave and averages several hundred milliseconds of the signal prior to it and several hundred milliseconds of the signal following it, in order to recover the complete P-QRS-T configuration of the fetal ECG. Such averaging techniques do not always work, however, because the various intervals of the fetal ECG, as well as the waveforms themselves, may change slightly from one beat to the next. The sum is an average of all the recorded

ECG configurations and might provide a waveform that does not indicate the single-beat ECG of the fetal heart.

FETAL ECG

As we have said, physicians can determine the ECG of a fetus from a pair of biopotential-sensing electrodes placed on the abdomen of the mother. Often it is necessary to try several different placements to get the best signal. Once the best placement is determined, we obtain a recording such as that shown in the top trace of Figure 6.22. For comparison, Figure 6.22 also shows a direct ECG of the same fetus and a direct ECG of the same mother. The fetal ECG signal is usually quite weak; it generally has an amplitude of around 50 μV or less. This makes it extremely difficult to record the heartbeat of the fetus by using electrodes attached to the abdomen of the mother during labor, when the mother is restless and motion artifact as well as EMG interfere. There is also considerable interference from the ECG of the mother (Neuman, 2006).

Note that the QRS complexes of the mother are much stronger than those of the fetus, which makes it difficult to determine the fetal heart rate electronically from recordings of this type. This information can be obtained manually, however, by measuring the fetal R–R interval on the chart and converting it to heart rate.

Several methods have been devised for improving the quality of fetal ECGs obtained by attaching electrodes to the mother's abdomen. In addition to the signal-averaging technique, physicians have applied various forms of anticoincidence detectors to eliminate the maternal QRS complexes (Offnet and Moisand, 1966). This method, as shown in the block diagram of Figure 6.23, uses at least three electrodes: one on the mother's chest, one at the upper part or fundus of the uterus, and one over the lower part of the uterus. The ECG of the mother is obtained from the top two electrodes, and the fetal-plus-maternal signal is obtained from the bottom two. The center

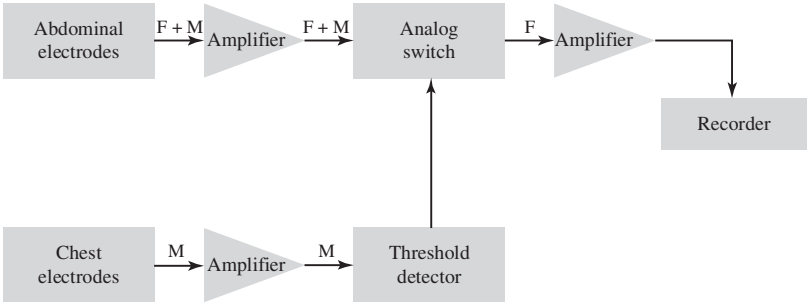


Figure 6.23 Block diagram of a scheme for isolating fetal ECG from an abdominal signal that contains both fetal and maternal ECGs. (From J. F. Roux, M. R. Neuman, and R. C. Goodlin, ''Monitoring of intrapartum phenomena.'' In *CRC Critical Reviews in Bioengineering*, 2, January 1975, pp. 119–158 Copyright CRC Press. Used by permission of CRC Press, Inc.)

electrode is common to both. A threshold detector determines the mother's QRS complexes and uses this information to turn off an analog switch between the electrodes recording the fetal ECG and the recording apparatus. Therefore, whenever a maternal QRS complex is detected, the signal from the abdominal leads is temporarily blocked until the end of the QRS complex, thereby eliminating it from the abdominal recording. Note that this technique also eliminates any fetal QRS complexes that occur simultaneously with the maternal ones. Modern systems incorporate computing circuits to recognize the absence of this fetal signal and to compensate for it when determining the fetal heart rate. One must always be cautious in using such a system since it can anticipate a fetal beat during a maternal QRS complex, but, in fact, the beat did not occur during the maternal QRS.

THE VECTORCARDIOGRAPH

In Section 6.2 we looked at the basis of the ECG and defined the cardiac vector. The ensuing description of the electrocardiograph showed how a particular component of the cardiac vector could be recorded. Such scalar ECGs are the type that are usually taken. However, we can obtain more information from a *vectorcardiogram* (VCG). A VCG shows a three-dimensional—or at least a two-dimensional—picture of the orientation and magnitude of the cardiac vector throughout the cardiac cycle. It is difficult for practical machines to display the VCG in three dimensions, but it is relatively simple to display it in two dimensions—or, in other words, its component in a particular plane of the body.

Special lead systems have been developed that can provide the x, y, and z components of the ECG. Any two of these can be fed into a vectorcardiograph to arrive at the VCG for the plane defined by the axes. The signal from the lead for one axis is connected to input 1, and that for the other enters input 2. These signals are then plotted one versus the other on the readout screen and/or they are printed. For each heartbeat, a vector loop representing the locus of the tip of the cardiac vector when its tail is at the origin is determined.

Because of the complexities of obtaining the vectorcardiogram and the difficulty that arises in interpreting the patterns, this technique is generally limited to special studies at tertiary-medical-care facilities or to use as a research tool. The scalar 12-lead electrocardiogram is employed for routine clinical studies.

6.9 CARDIAC MONITORS

There are several clinical situations in which continuous observation of the ECG and heart rate is important to the care of the patient. Continuous observation of the ECG during the administration of anesthesia helps doctors monitor the patient's condition while he or she is undergoing medical

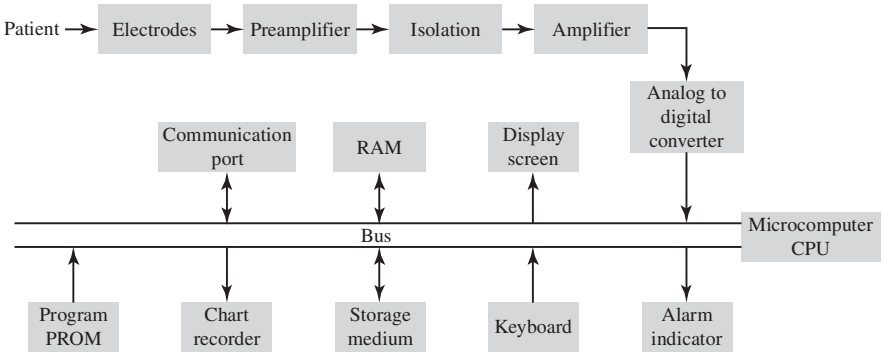


Figure 6.24 The cardiac monitor displays a continuous electrocardiogram and heart rate and also identifies alarm conditions.

procedures and during recovery from anesthesia. Constant monitoring of the ECG and heart rate of the myocardial-infarction patient during the danger period of several days following the initial incident has made possible the early detection of life-threatening cardiac arrhythmias. Continuous monitoring of the fetal heart rate during labor may help in the early detection of complications.

These and other clinical applications of continuous monitoring of the ECG and heart rate are made possible by *cardiac monitors*. Figure 6.24 shows the basic cardiac monitor in block-diagram form. Its front-end circuitry is similar to that of the electrocardiograph. A pair of electrodes, usually located on the anterior part of the chest, pick up the ECG and are connected by lead wires to the input circuit of the monitor. The input circuit contains circuitry, as described in Section 6.4, to protect the monitor from high-voltage transients that can occur during defibrillation procedures.

The next stage of the monitor is a standard biopotential amplifier designed to amplify the ECG. Although it is best to have the frequency-response characteristics described in Section 6.2, cardiac monitors often have a slightly narrower frequency response than would be acceptable for a diagnostic electrocardiograph. The reason for this is that much of the motion-artifact signal seen during movement of the patient is at very low frequencies. By filtering out some of these low frequencies, we can obtain a vast improvement in SNR and recording stability without seriously affecting the information that pertains to cardiac rhythm in the ECG. Frequency response should be from 0.67 to 40 Hz (Anonymous, 1992). Cardiac monitors should not trigger on pacemaker spikes, which continue even when the heart has stopped. To avoid double counting, cardiac monitors should not trigger on tall T waves (Anonymous, 1990).

Patient isolation circuitry (Section 6.2) is usually found in the circuit following an ECG preamplifier. This is followed by an additional amplifier to raise the signal to levels appropriate for further processing.

In most modern cardiac monitors, the amplified ECG signal is digitized by an ADC, and the remaining processing is carried out by a computer. The

digital signal is processed by a microcomputer in the monitor. This system block can perform many functions depending on the program that controls it. The digital signal can be filtered and displayed on a computer screen, the heart rate determined by cardiotachometer software, alarm conditions identified and alarms sounded, data stored in temporary or permanent memory, an ECG rhythm strip printed for review and charting, and communication of the data to other systems within or outside of the hospital.

Often a physician wants to have a permanent record of the ECG being monitored. For this reason, many cardiac monitors have a small chart recorder or graphic printer built into them that can be switched on by the operator or the computer to record a particularly interesting ECG as it appears on the screen.

It is often desirable to have a record of the events in the ECG that lead up to a serious arrhythmia. Such a record can be made if the digital signal is fed first to a memory loop, which delays the ECG signal by about 15 s. The output from the memory loop can then be fed to the printer or chart recorder where the hard copy is produced. Thus, when the operator of the monitor sees an interesting ECG waveform or the monitor itself detects a clinically significant arrhythmia, the information can be obtained from the memory loop to give a record of the events that led up to that particular pattern.

The heart rate is determined from the ECG using a computer algorithm that performs the function of a cardiotachometer. The output is displayed on a rate display so that the operator can immediately tell the patient's heart rate. Alarm circuitry to warn of high and low heart rate is also associated with this algorithm. The alarm system can also produce a hard copy of the events that led up to the alarm for analysis by clinicians. This can be a valuable aid to clinicians in selecting appropriate therapy for the alarm-producing event.

Most hospitals also utilize cardiac monitors in an organized system called an *intensive-care unit*. In such units, there are individual monitors at each patient's bedside that display the ECG in real time as well as the heart rate and any alarm conditions that have recently occurred. These individual monitors are connected to a central unit located at the nursing station that shows the ECGs for all patients being monitored, along with a heart-rate display and alarm indicator for each patient. A printer at the central station can be activated either locally or by remote control from the individual monitors at the patient's bedside.

Computer algorithms that can recognize cardiac arrhythmias and record the frequency of their occurrence are also included in cardiac monitors. The machines can also prepare hard-copy charts showing trends in the patient's monitored parameters and can keep records of various therapeutic measures taken by the clinical staff. The computer can also be a big help in the intensive-care unit by carrying out many observational and charting functions, thereby freeing the clinical staff to care for the patient (Nazeran, 2006).

The availability of microcomputers and high-capacity electronic memory has made it possible to monitor ambulatory patients with detection of cardiac arrhythmias. These monitors consist of an ECG amplifier that provides a signal to an ADC, where it is digitized and stored in memory for later download and

analysis. Such devices can collect data from ambulatory patients, and these data are analyzed later by a computer (Jurgen, 1976).

Microcomputers in cardiac monitors perform two basic functions, data management and data analysis. In the former case, the microcomputer controls the various components of the system and directs the transport of data from one block to another along the bus. Carrying out the second function involves the actual analysis of the electrocardiogram. It includes filtering and artifact reduction, identification of the various components of the electrocardiogram, determination of the heart rate, and identification of arrhythmias. More than one microcomputer can be used in a monitor system to carry out these functions. The microcomputer is under the software control. This makes it possible to update the monitor by replacing the software rather than modifying any hardware of the instrument.

The microcomputer can temporarily store the data, and an alternative medium such as a separate hard drive is used to archive selected incidents or the entire monitored data. There is also a staff interface to the system that consists of a keyboard and a display monitor.

Computerized cardiac monitors can be integrated into other hospital information systems. Frequently these monitors also have a network connection that enables them to interact with other information systems or to transmit data to physicians' offices located away from the intensive-care unit.

Ambulatory cardiac monitors are often used in the diagnosis and treatment of heart disease. The most frequently applied ambulatory monitor—the Holter monitor—includes a miniature digital recorder with electronic memory that the patient wears. These devices consist of a battery-powered ECG amplifier and recorder that are connected to electrodes placed on the patient's chest. The instrument is sufficiently small to allow the patient to wear it like a necklace, and the recorder memory can hold from 24 to 48 h of continuous ECG recording. Some recorders can collect data from three leads simultaneously so that vectorcardiograms can be stored. Special computerized playback units rapidly analyze the data files for cardiac arrhythmias and display these portions of the electrocardiogram on a computer screen or generate a hard-copy printout of them. The playback units also summarize the total recording in a report that indicates variables such as heart rate, variability in heart rate, type and number of arrhythmias, and amount of artifact.

Holter monitors are used by physicians to detect cardiac arrhythmias that occur infrequently in patients and are usually not detected during office or hospital examinations. Microelectronics has made it possible to make these monitor–recorders so small that they can be surgically implanted under the skin of patients or incorporated into other implanted devices such as pacemakers. The Medtronic Corp. (2008) Reveal Insertable Loop Recorder has a mass of only 17 g and can store up to 42 min of ECG. It can monitor a patient for up to 14 months with built-in electrodes. The recorder can either be activated by the patient when they experience the symptoms or be programmed to recognize and record significant events. By being implantable, the problems of patient compliance or electrode detachment are avoided.

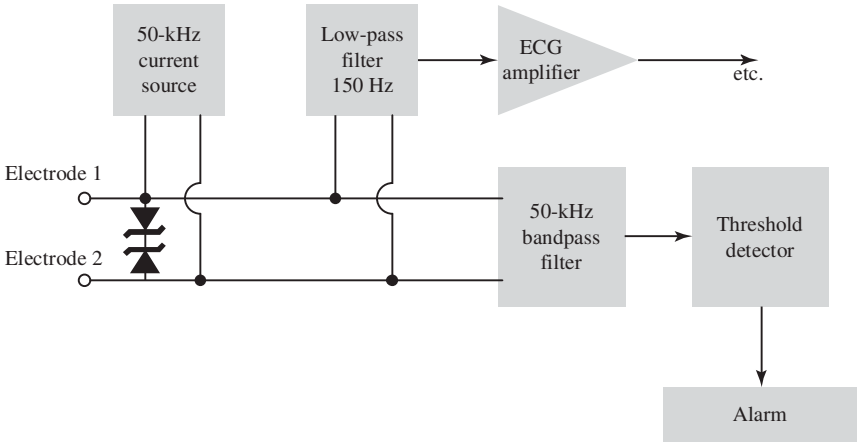


Figure 6.25 Block diagram of a system used with cardiac monitors to detect increased electrode impedance, lead wire failure, or electrode falloff.

Farwell *et al*. (2006) have shown how this technology can impact clinical evaluation of patients with fainting spells that could be the result of infrequent life-threatening cardiac arrhythmias.

In situations in which cardiac monitors are used to observe a patient's ECG over a long period of time, artifact and failure of the monitor can occur as a result of a poor electrode–patient interface. The longer the electrodes remain on the patient, the more often this occurs. In intensive-care units, electrodes are routinely changed—sometimes once a shift, sometimes once a day—to ensure against this type of breakdown. Most cardiac monitors also have alarm circuits that indicate when electrodes fall off the patient or the electrode–patient connection degenerates.

Figure 6.25 is a block diagram of a typical lead fall-off alarm. A 50 kHz high-impedance source is connected across the electrodes. Peak amplitudes of the current can be as great as 100 to 200 μA without any risk to the patient, because the microshock hazard to excitable tissue decreases as the frequency increases above 1 kHz (Figure 14.3). The current passes through the body between the electrodes, and, as long as there is good electrode contact, the voltage drop is relatively small. If the electrode connections become poor, as can happen when the electrolyte gel begins to dry, or if one of the electrodes falls off or the wire breaks, the impedance between the electrodes increases considerably. This causes the voltage produced by the 50 kHz source to rise. The high-frequency signal is separated from the ECG by the filtering scheme, as shown. The ECG passes through a low-pass filter with approximately a 150 Hz corner frequency, and is processed in the usual way. A bandpass filter with a 50 kHz center frequency passes the voltage resulting from the current source to a threshold detector. This detector sets off an alarm when the voltage exceeds a certain threshold, which would correspond to poor electrode contact. When an electrode falls off the patient, the interelectrode impedance should increase to

infinity, resulting in the possibility of 50 kHz voltages high enough to cause some damage to the electronic devices. For this reason, a high-voltage protection circuit, such as that described in Section 6.4, is frequently connected across the input terminals to the monitor. In the case shown in Figure 6.25, back-to-back zener diodes are used.

6.10 BIOTELEMETRY

Biopotential and other signals are often processed by radiotelemetry, a technique that provides a wireless link between the patient and the majority of the signal-processing components. By using a miniature radio transmitter attached to the patient to broadcast the information over a limited range, clinicians can monitor a patient or study a research animal while the subject has full mobility. This technique also provides the best method of isolating the patient from the recording equipment and power lines. For a single-channel system of biopotential radiotelemetry, a miniature battery-operated radio transmitter is connected to the electrodes on the patient. This transmitter broadcasts the biopotential over a limited range to a remotely located receiver, which detects the radio signals and recovers the signal for further processing. In this situation there is obviously negligible connection or stray capacitance between the electrode circuit connected to the radio transmitter and the rest of the instrumentation system. The receiving system can even be located in a room separate from the patient's. Hence the patient is completely isolated, and the only risk of electric shock that the patient runs is due to the battery-powered transmitter itself. Thus, if the transmitter power supply is kept at a low voltage, there is negligible risk to the patient.

Many types of radiotelemetry systems are used in biomedical instrumentation (Ziaie, 2006). The basic configuration of the system, however, is pretty much the same for all. A preamplifier amplifies the ECG signal to a level at which it can modulate the transmitter. Pulse-code modulation in the range of 100 to 500 MHz is the dominant method. The entire transmitter is powered by a small battery pack. It is carried by the patient and usually attached by means of a special harness. Ultraminiature radio transmitters can be attached by surgical tape directly to the patient's skin. In research with experimental animals, experimenters can surgically implant the tiny transmitters within the bodies of the animals so that no external connections or wires are required. Stuart Mackay of Boston University pioneered this technique many years ago (Mackay, 1970), and many applications from wildlife biology to clinical medicine followed. Today, the technique is routinely found in hospital intensive-care and step-down units for cardiac monitoring (Budinger, 2003).

In the receiving system, a pickup antenna receives the modulated signal. The signal is then demodulated to recover the original information from the carrier. The signal can be further amplified to provide a usable output. The

receiver system is generally powered directly from the power line, because it is in a permanent location and is not attached to the patient in any way.

The bandwidth of the system is determined by the rate at which it is sampled. Theoretically, the rate of sampling should be at least twice that of the highest-frequency component to be transmitted, but in practical circuits the rate of sampling is usually at least five times that of the highest-frequency component.

It is important to note that, although radiotelemetry systems provide ideal isolation with no patient ground required, they are not completely immune to problems of electric noise. Because coupling is achieved by a radiated electromagnetic signal, other electromagnetic signals at similar frequencies can interfere and cause artifacts. In extreme cases, these other signals can even bring about complete loss of signal.

In addition, the relative orientation between transmitting and receiving antennas is important. There can be orientations in which none of the signals radiated from the transmitting antenna are picked up by the receiving antenna. In such cases, there is no transmission of signals. In high-quality radiotelemetry systems, it is therefore important to have a means of indicating when signal interference or signal dropout is occurring. Such a signal makes it possible to take steps to rectify this problem and informs the clinical staff that the information being received is noise and should be disregarded.

The advent of wireless computer communication systems has affected biotelemetry as well. Telemetry systems capable of two-way communication utilize the standardized wireless computer connection protocols such as WiFi, Bluetooth and ZigBee. Complete transceiver (transmitter and receiver) systems for these protocols are available on a single integrated circuit chip, so very small wireless devices can now be realized. These can be incorporated into wireless sensing networks that can either be implanted in the body or incorporated into clothing. Although systems such as Bluetooth and ZigBe are limited to short range, external transponders can extend coverage.

PROBLEMS

6.1 What position of the cardiac vector at the peak of the R wave of an electrocardiogram gives the greatest sum of voltages for leads I, II, and III?
6.2 What position of the cardiac vector during the R wave gives identical signals in leads II and III? What does the ECG seen in lead I look like for this orientation of the vector?
6.3 An ECG has a scalar magnitude of 1 mV on lead II and a scalar magnitude of 0.5 mV on lead III. *Calculate* the scalar magnitude on lead I.
6.4 Design a system that has as inputs the *scalar* voltages of lead II and lead III and as output the *scalar* voltage of the cardiac vector **M**.