Biopotential Amplifiers

MECE 493-Biomedical Instrumentation

Dr. Kurtuluş Erinç Akdoğan
kurtuluserinc@cankaya.edu.tr

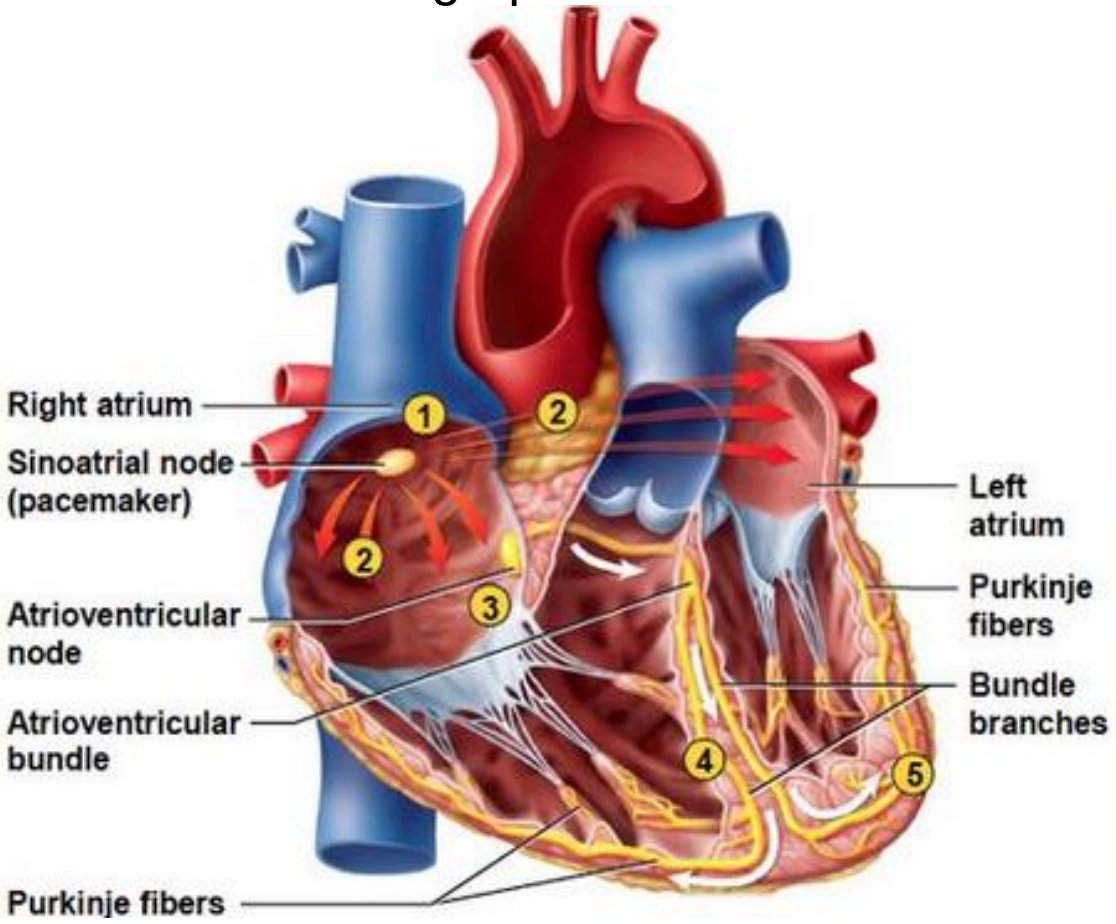
ÇANKAYA ÜNİVERSİTESİ
MEKATRONİK MÜHENDİSLİĞİ BÖLÜMÜ

Basic Requirements

- Use of biopotential amplifiers
  - Increase the amplitude (voltage, current, power) of biopotential signals
  - Isolation of load from source (*provides only current gain, leaving the voltage levels unchanged*)
- Basic requirements of biopotential amplifiers
  - High input impedance for minimal loading: > 10 MΩ (*not to distort measured signal*)
  - Input protection (*not to cause shocks in the patient*)
  - Isolation for safety (*current through the electrode can be kept at safe levels*)
  - Low output impedance (*to maintain maximal fidelity and range in the recording device*)
  - Optimal bandwidth for better SNR (*great enough to process interested biological signal bandwidth*)
  - Enough gain: ~ 1000 or more (*to make biological signals at millivolt levels compatible with recoding and display devices*)
  - High CMRR for differential input amplifiers (*biopotential signals are obtained from bipolar electrodes measuring difference of signals from two electrode and amplified by a differential amplifier*.)
  - Quick calibration (*gain must be calibrated prior to the measurement*)

Heart Conduction system

- To learn more about biopotential amplifiers, we shall examine a typical clinical elecrocardiograph.



1. SA node fires.
2. Excitation spreads through atrial myocardium.
3. AV node fires.
4. Excitation spreads down AV bundle.
5. Purkinje fibers distribute excitation through ventricular myocardium.

https://www.pinterest.com/mastersnowball/conduction-system-of-the-heart/

- Electric activity of heart can be modeled as an electric dipole located in the partially conducting medium of throax.
- Drawn diopole occurs at a specific instant. At the next instant, it can change its orientation and magnitude.
- By placing electrodes on the body surface, if two electrodes are located on different equal-potential lines of the electric field of the heart, voltage is measured.

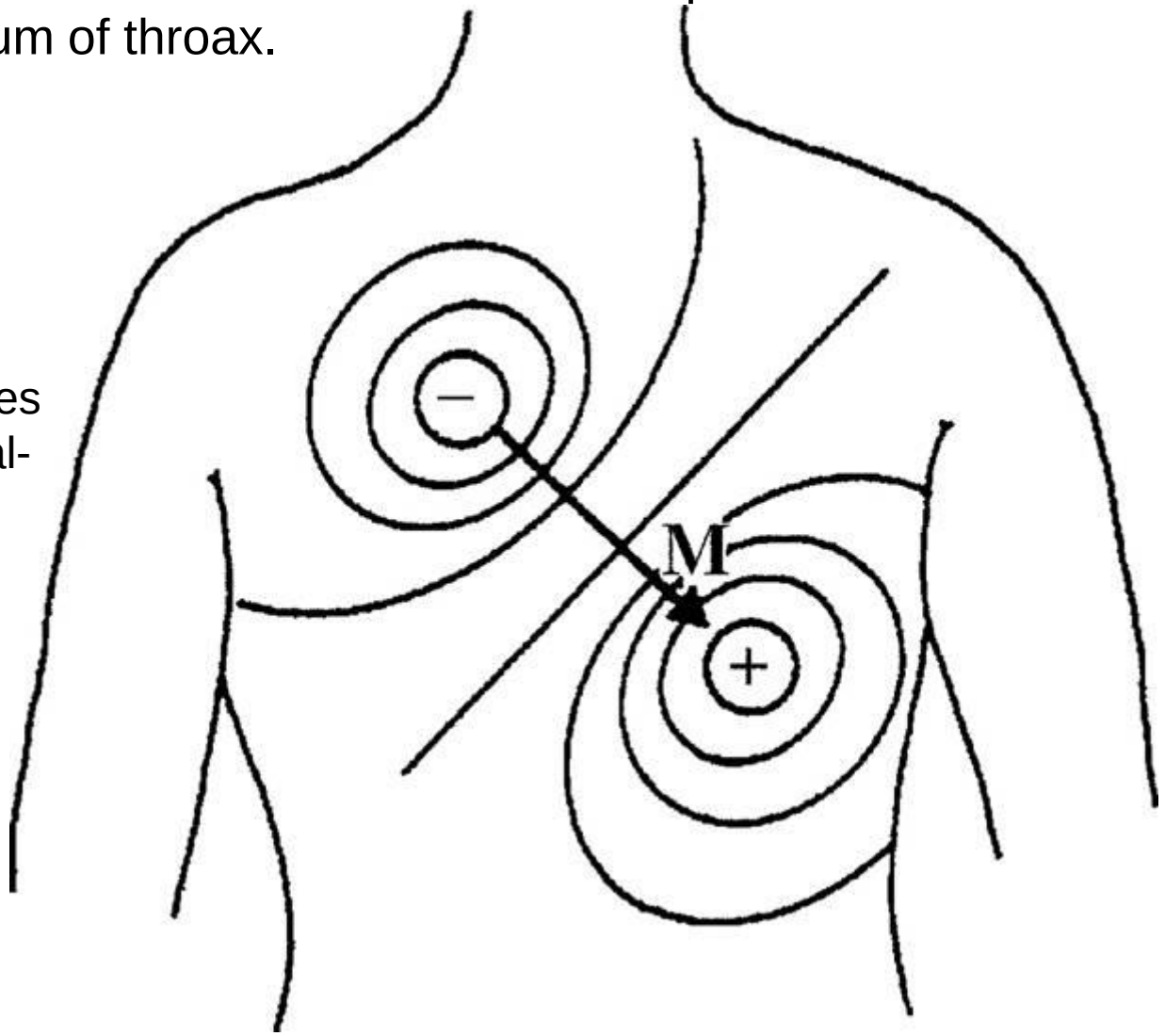
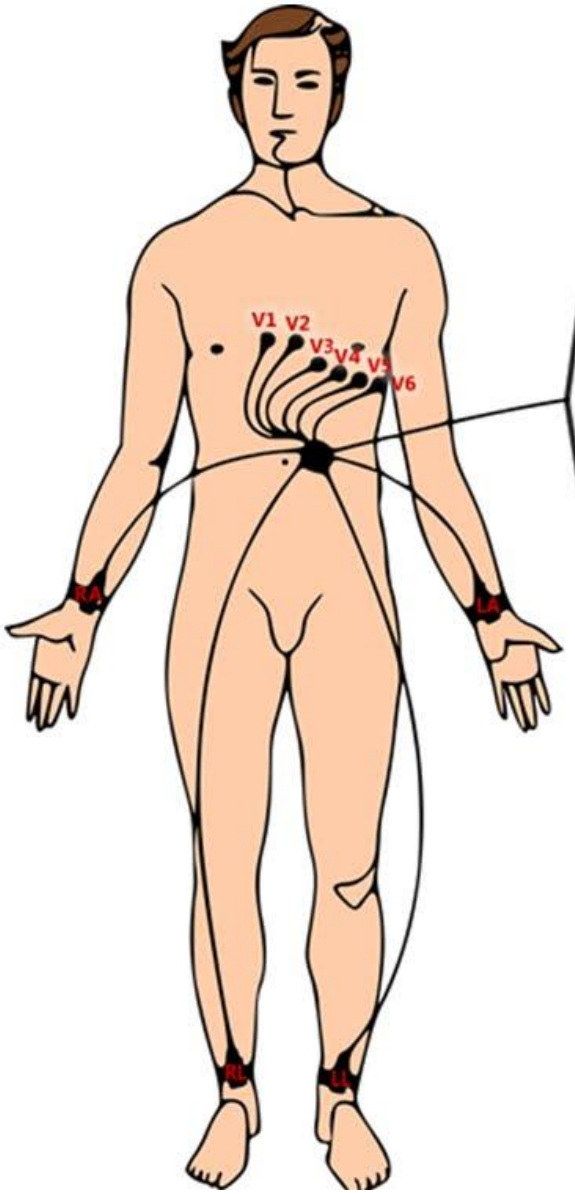
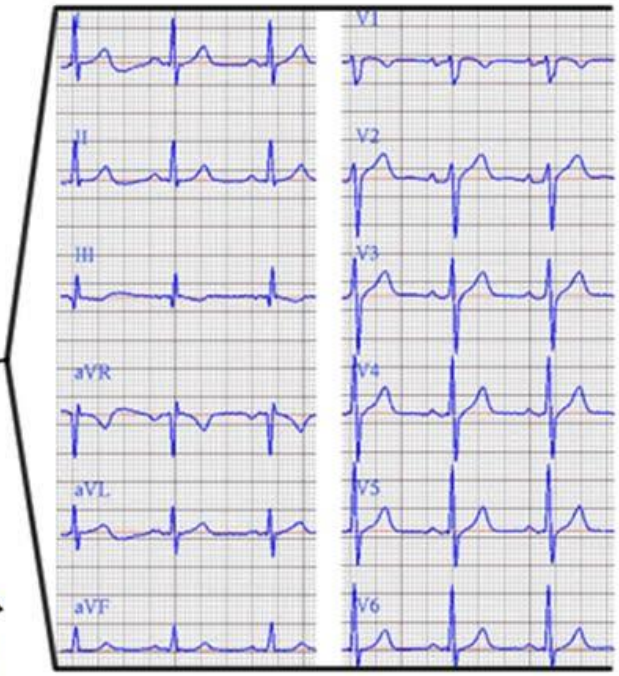
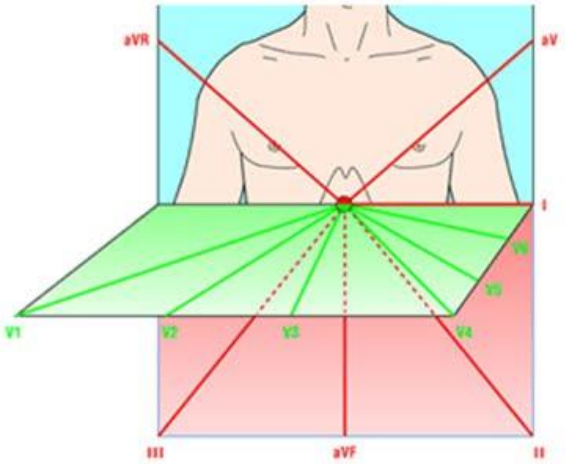


Figure 6.1 Rough sketch of the dipole field of the heart when the R wave is maximal. The dipole consists of the points of equal positive and negative charge separated from one another and denoted by the dipole moment vector **M.**

12-Lead ECG







V1 - 4th intercostal space
R sternal border

V2 - 4th intercostal space
L sternal border

V3 - Between leads V2 and V4.

V4 - 5th L intercostal space in midclavicular line

V5 - Horizontally even with V4, but in the anterior axillary line.

V6 - Horizontally even with V4 and V5 in the midaxillary line. (The midaxillary line is the imaginary line that extends down from the middle of the patient's armpit.)

http://www.ivline.org/2010/05/quick-guide-to-ecg.html

Why do we need standard ECG lead positions?

- 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, important to have certain standard positions for clinical evaluation of the ECG.
- The limbs make fine guideposts for locating the ECG electrodes.

What is a ECG lead vector?

- A pair of electrodes on the surface or its equivalence defines a lead
- Projection of the cardiac vector to a lead vector is scalar voltage of the lead.

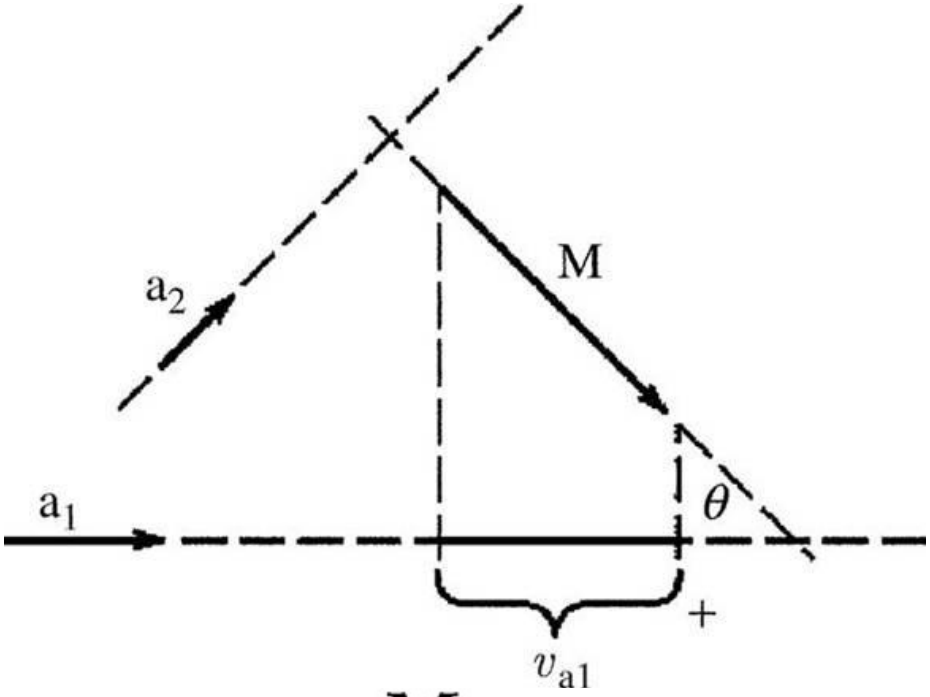


Figure 6.2 Relationships between the two lead vectors a1 and a2 and the cardiac vector M. The component of M in the direction of a1 is given by the dot product of these two vectors and denoted on the figure by va1. Lead vector a2 is perpendicular to the cardiac vector, so no voltage component is seen in this lead.

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

- To describe M uniquely, at least two leads a1 and a2 are required.

ECG Leads

- In clinical ectrocardiography, to describe the heart's electric activity fully.
- Frontal plane ECG
- Electrodes: RA, LA, LL, and RL (ground)
- Bipolar leads
  - Lead I: LA(+) ← RA(-),
  - Lead II: LL(+) ← RA(-),
  - Lead III: LL(+) ← LA(-)
- The lead vectors that are formed can be approximated as an equilateral triangle known as Eindhoven's triangle

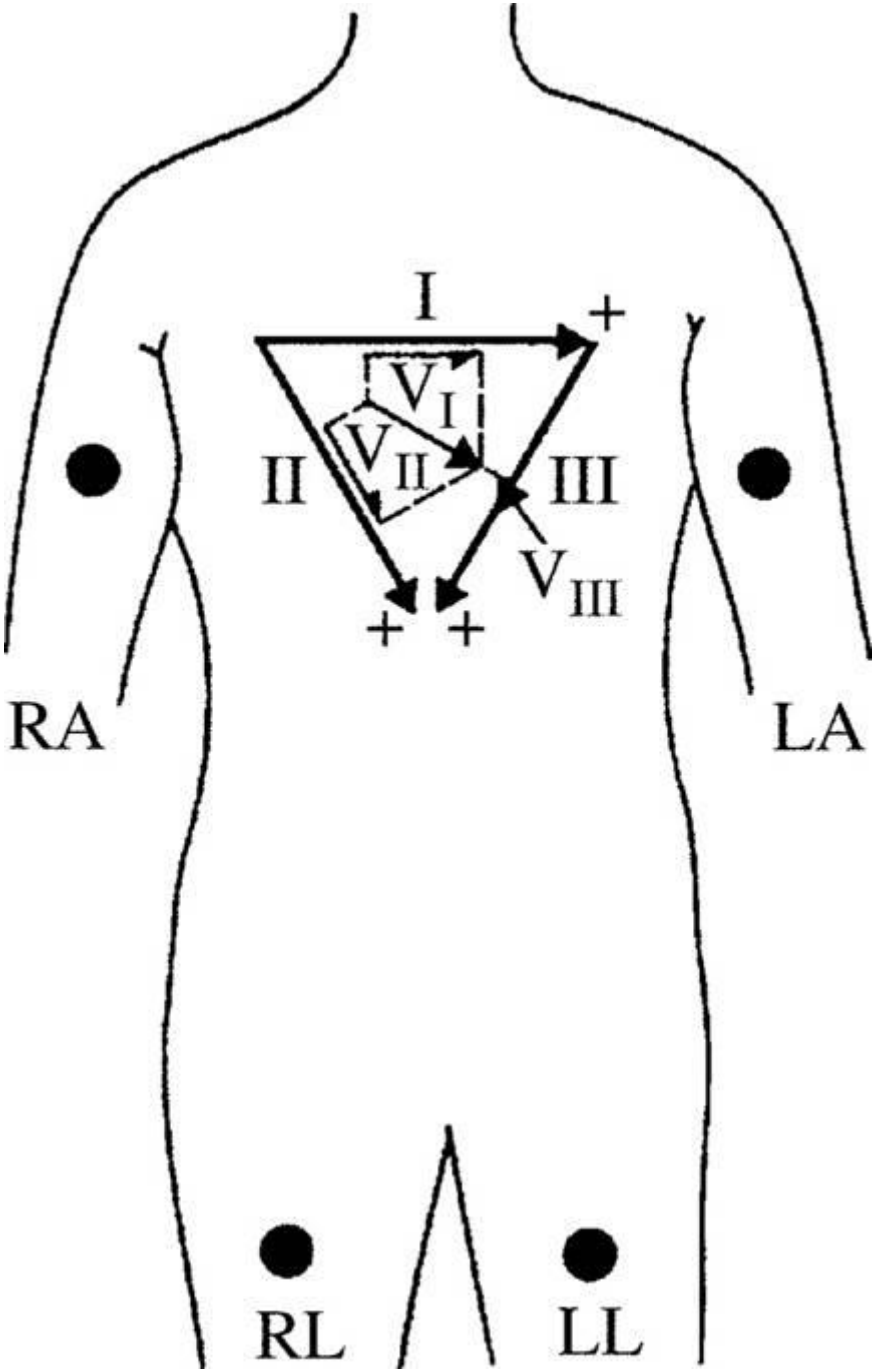
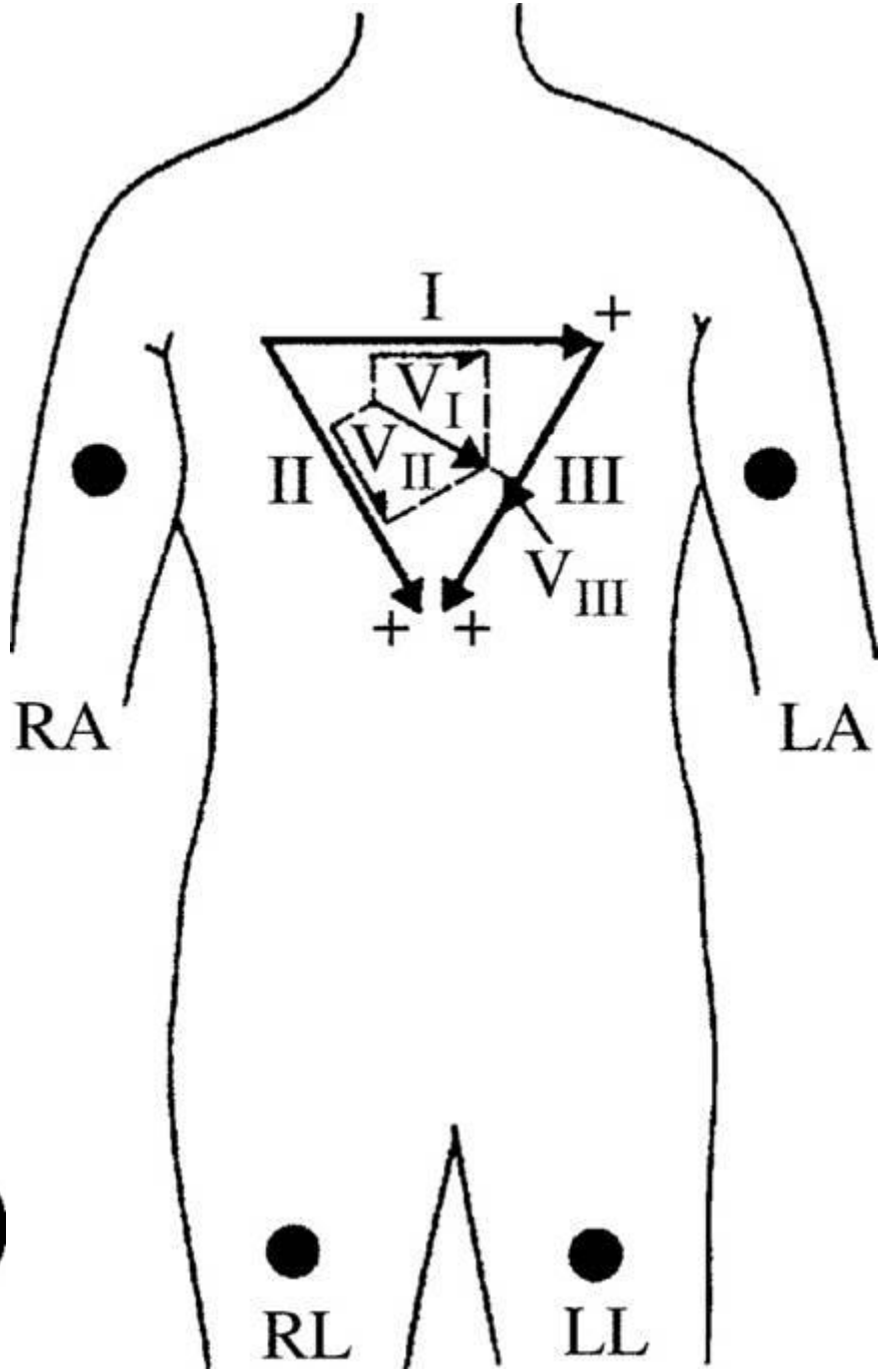


Figure 6.3 Cardiologists use a standard notation such that the direction of the lead vector for lead I is 0°, that of lead II is 60°, and that of lead III is 120°. An example of a cardiac vector at 30° with its scalar components seen for each lead is shown.

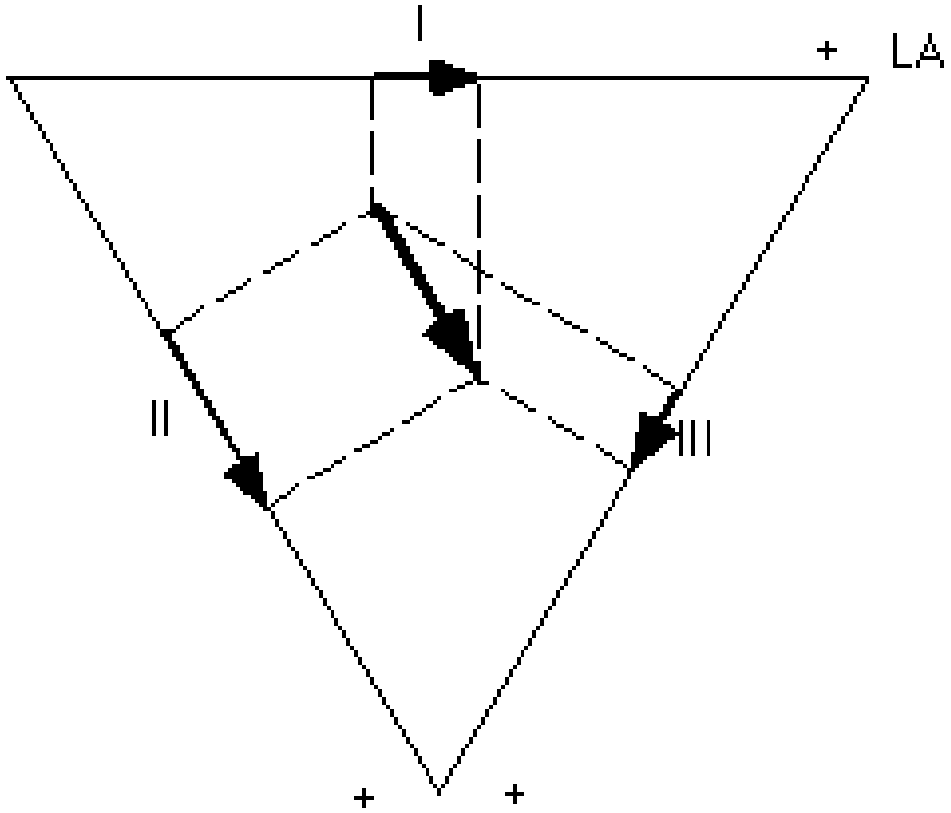


Kirchhoff's voltage law

$$I - II + III = 0$$

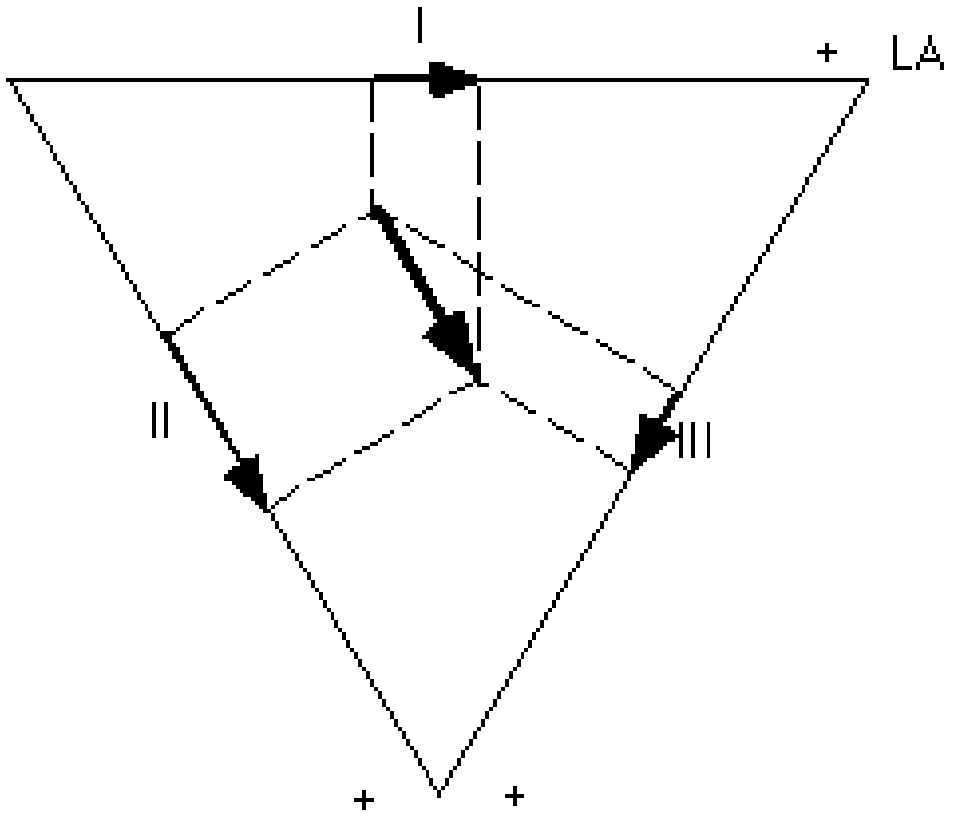
Problem

- You measured your ECG and found your 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.



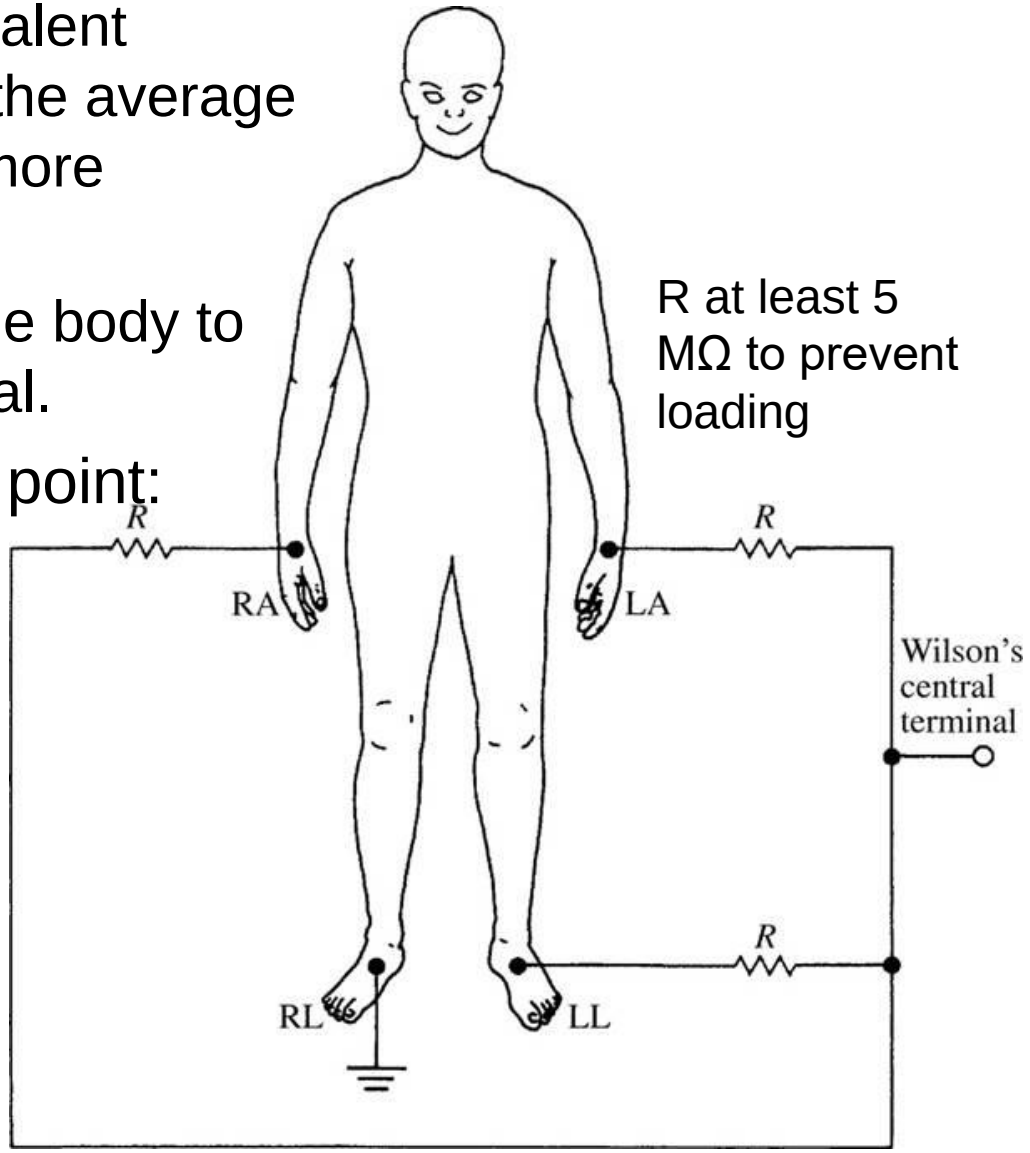
Problem

- What position of the cardiac vector during the R wave gives identical signals in leads II and III?



3 Additional lead – Unipolar leads

- 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.
- Connection of electrodes to the body to obtain Wilson's central terminal.
- The signal between central point:
  - and LA is VL
  - and RA is VR,
  - and LL is VF.



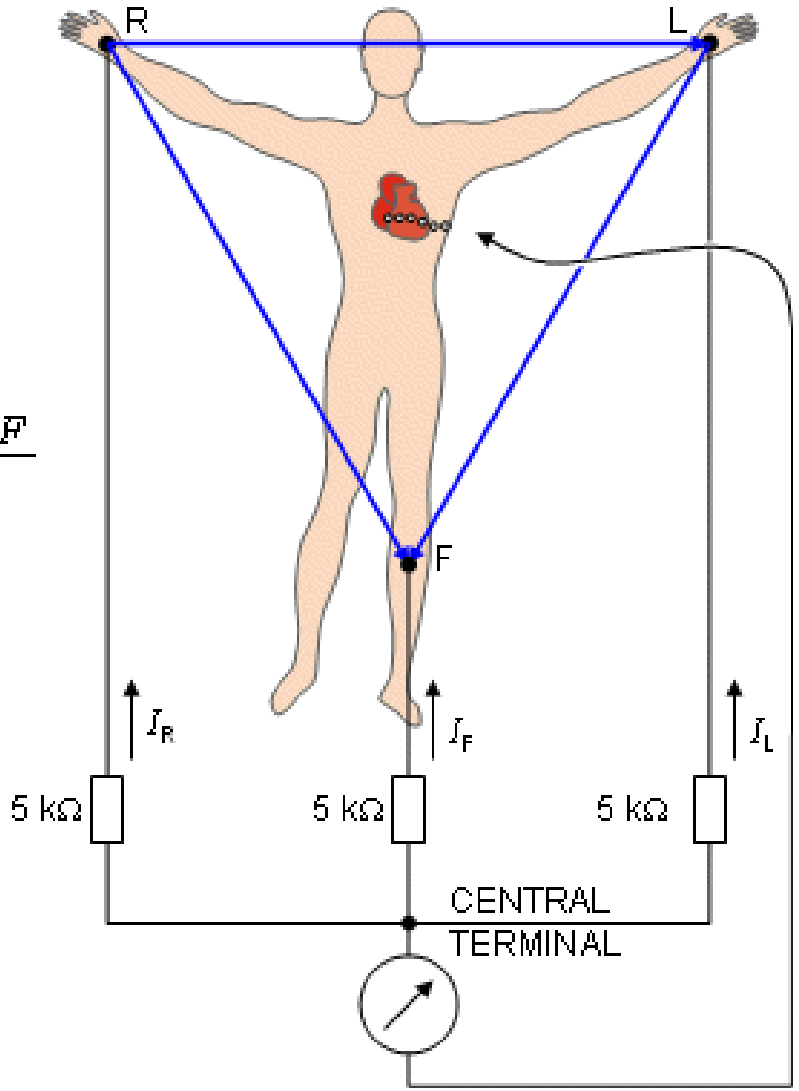
WILSON CENTRAL TERMINAL

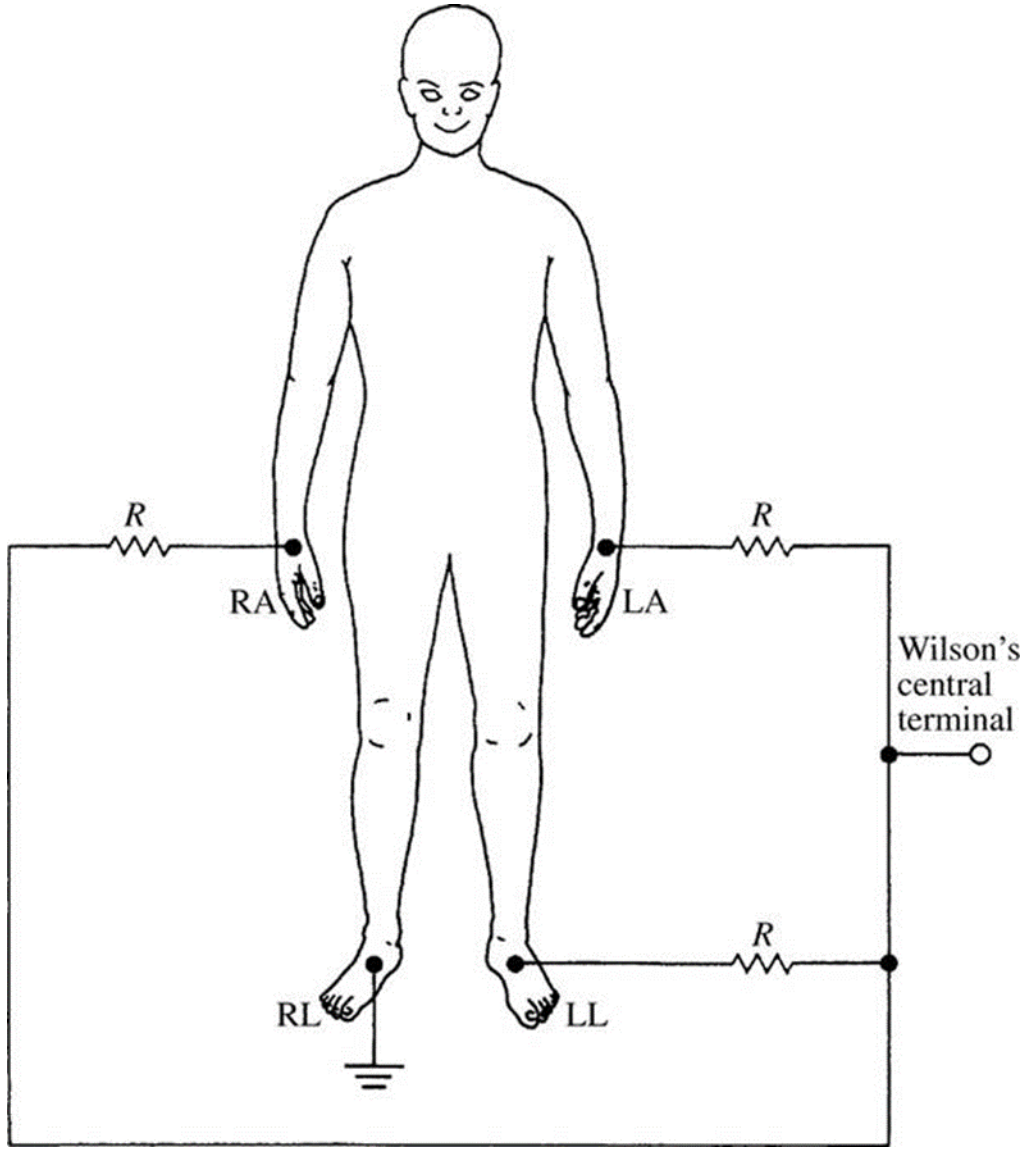
- Wilson suggested that unipolar potentials should be measured with respect to central terminal as reference.
- Assume an ideal voltmeter between CT and ground. There is no current between CT and the electrode.Then,

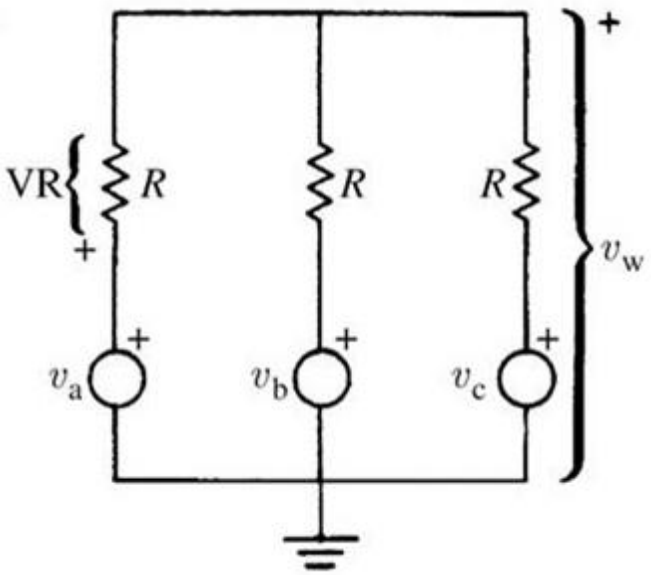
$$I_R + I_L + I_F = \frac{\Phi_{CT} - \Phi_R}{5000} + \frac{\Phi_{CT} - \Phi_L}{5000} + \frac{\Phi_{CT} - \Phi_F}{5000}$$

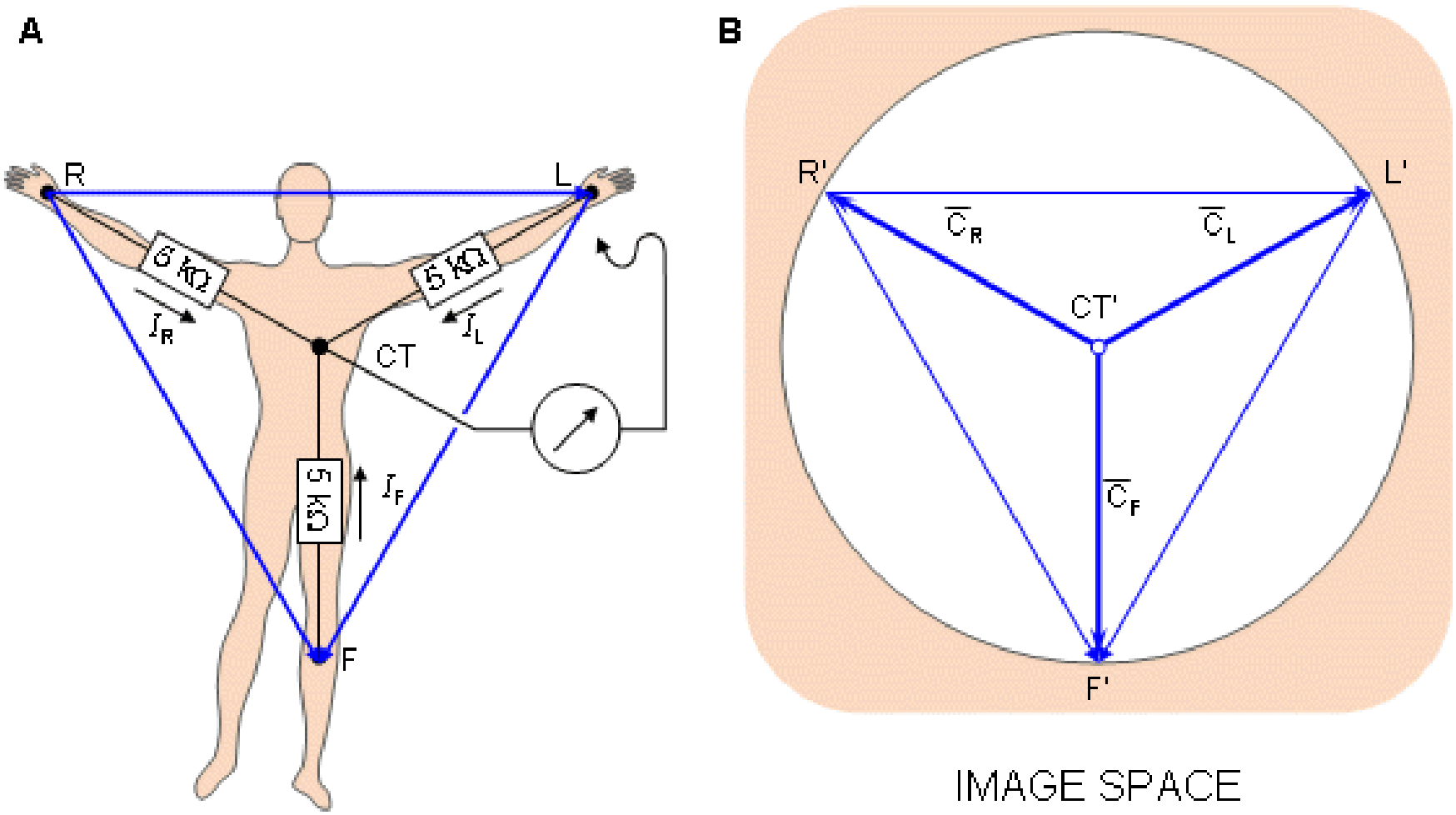
$$\Phi_{CT} = \frac{\Phi_R + \Phi_L + \Phi_F}{3}$$

- Since the central terminal potential is the average of the extremity potentials it can be argued that it is then somewhat independent of any one in particular and therefore a satisfactory reference.




R
R
RA
LA
Wilson's
central
terminal
R
RL
LL


VR
R
R
R
+
+
+
+
+
v_a
v_b
v_c
v_w



- A) The circuit of the Wilson central terminal (CT).
- (B) The location of the Wilson central terminal in the image space (CT'). It is located in the center of the Einthoven triangle.

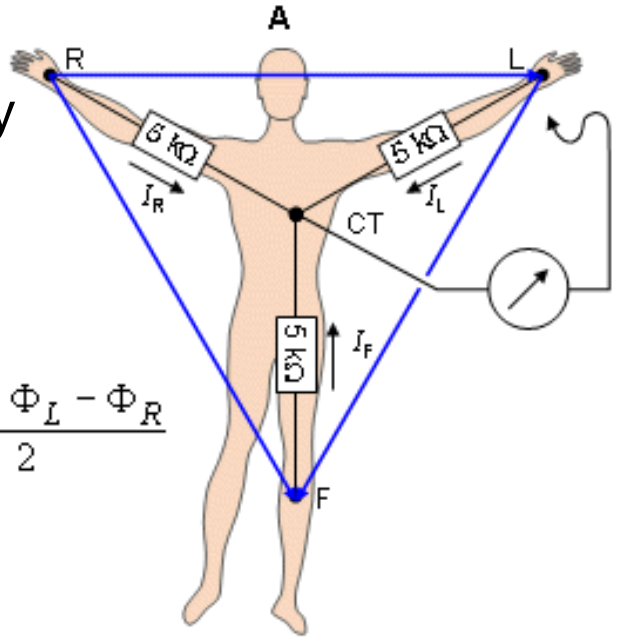
Augmented Leads

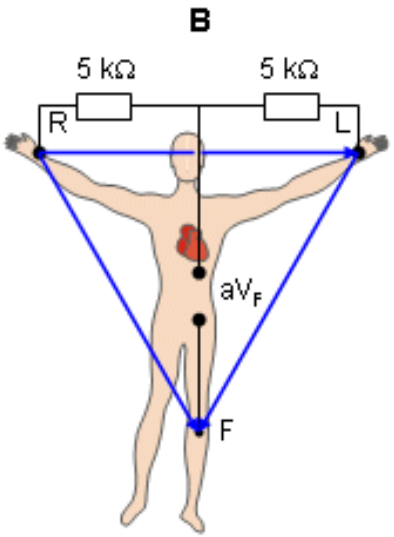
- Three additional limb leads, VR, VL, and VF are obtained by measuring the potential between each limb electrode and the Wilson central terminal.

$$V_F = \Phi_F - \Phi_{CT} = \frac{2\Phi_F - \Phi_R - \Phi_L}{3}$$

- These signals can be augmented (amplified) by omitting that resistance from the Wilson central terminal, which is connected to the measurement electrode.

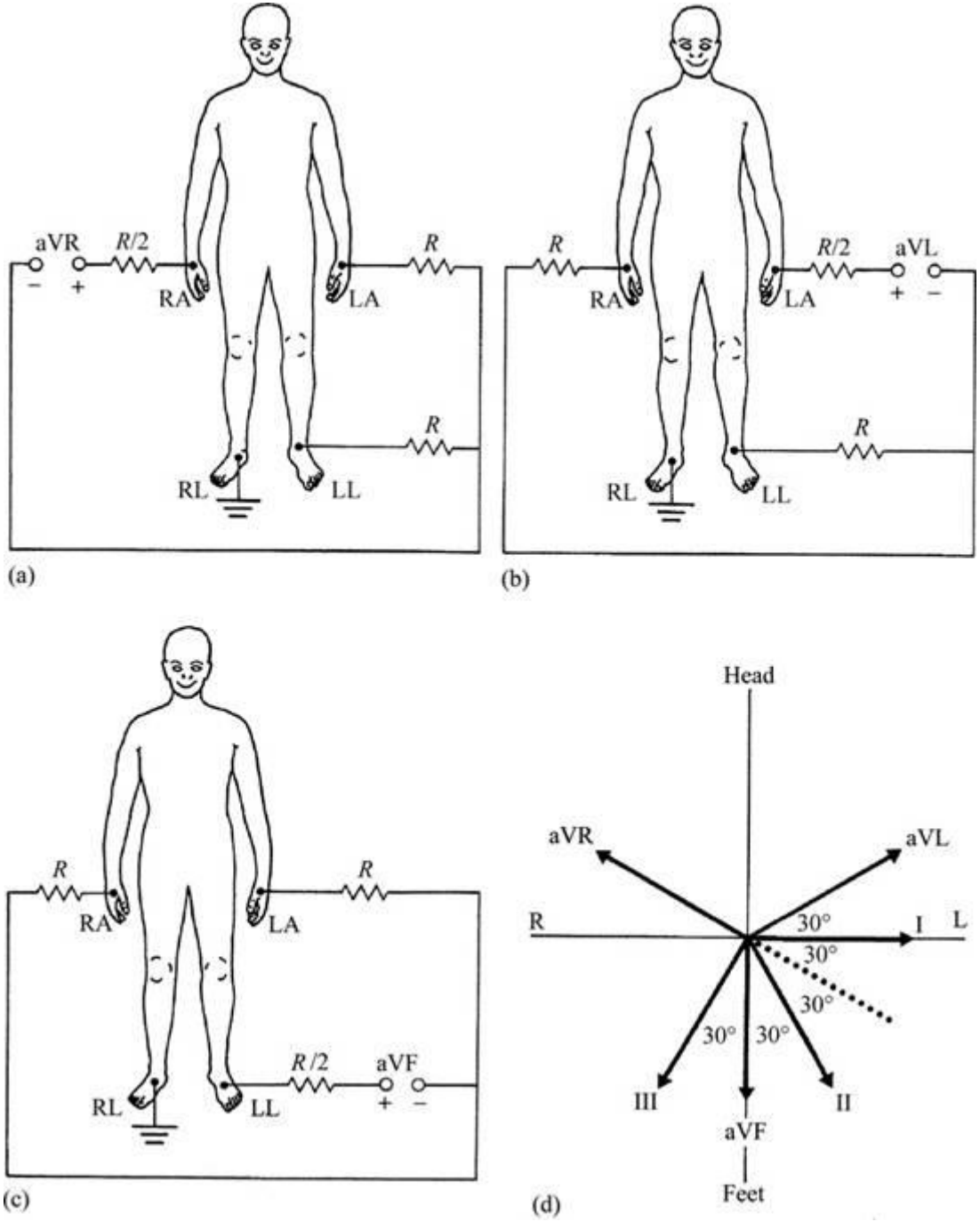
- In this way, the aforementioned three leads, VR, VL, and VF may be replaced with a new set of leads that are called augmented leads aVR, aVL, and aVF



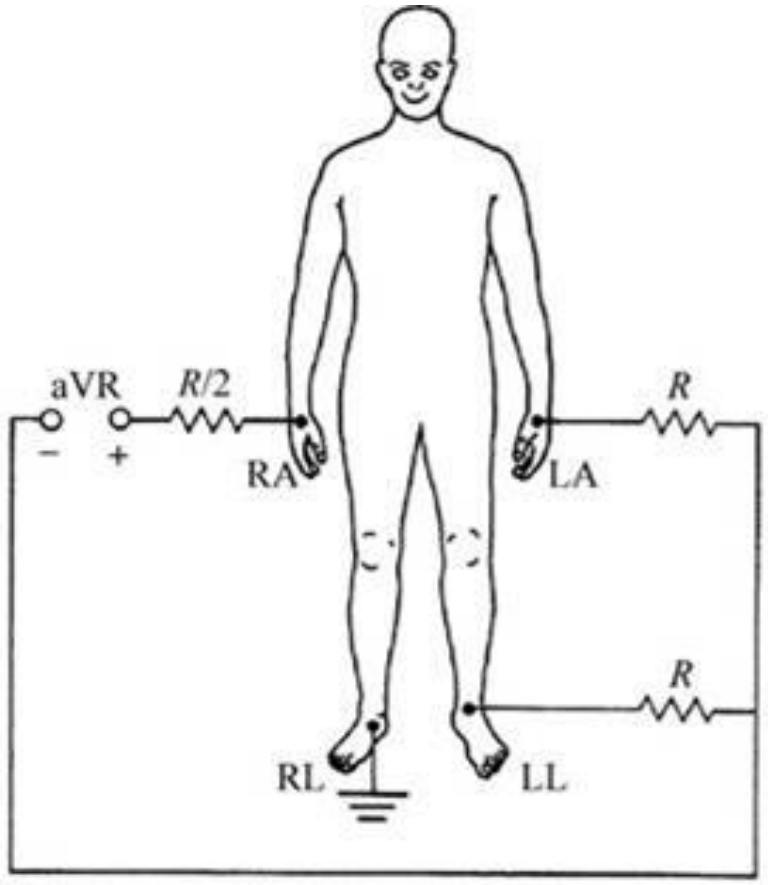


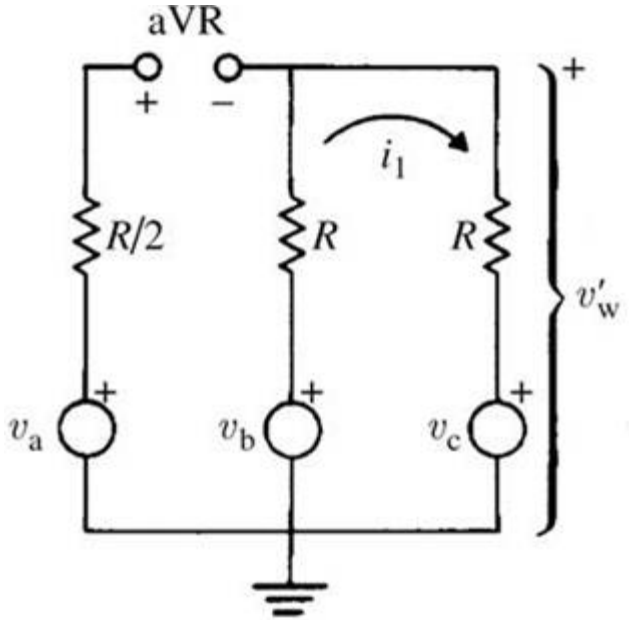
$$V_{aV_F} = \Phi_F - \Phi_{CT/aV_F} = \Phi_F - \frac{\Phi_L + \Phi_R}{2} = \frac{2\Phi_F - \Phi_L - \Phi_R}{2}$$

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.



Mathematical Relationship Between *VL* and *AVL*





$$i_1 = \frac{v_b - v_c}{2R}$$

$$v'_w = i_1 R + v_c = \frac{v_b - v_c}{2R} R + v_c = \frac{v_b - v_c}{2}$$

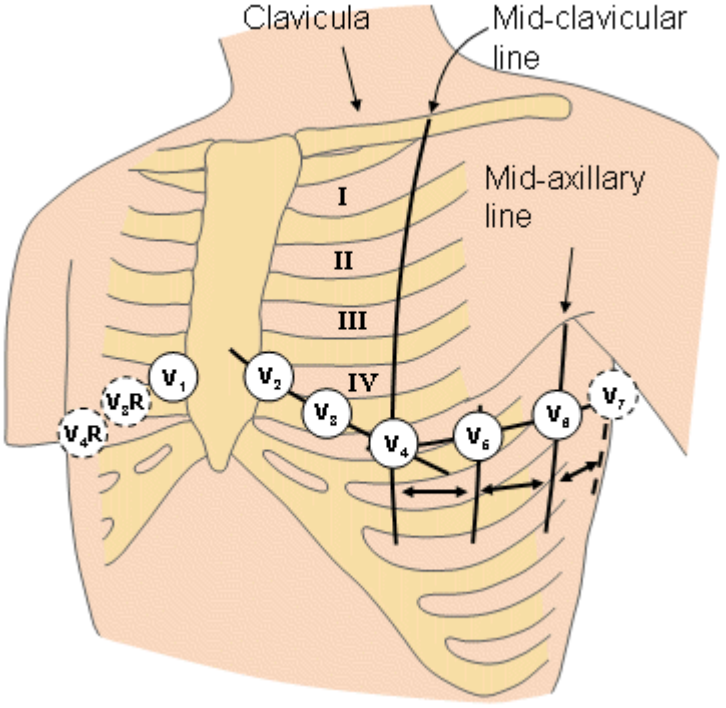
$$\text{aVR} = v_a - \frac{v_b + v_c}{2} = \frac{2v_a - v_b - v_c}{2}$$

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

$$\text{VR} = v_a - v_w = \frac{2v_a - v_b - v_c}{3}$$

PRECORDIAL LEADS

- For measuring the potentials close to the heart, Wilson introduced the precordial leads (chest leads) in 1944.
- These leads, V1-V6 are located over the left chest.

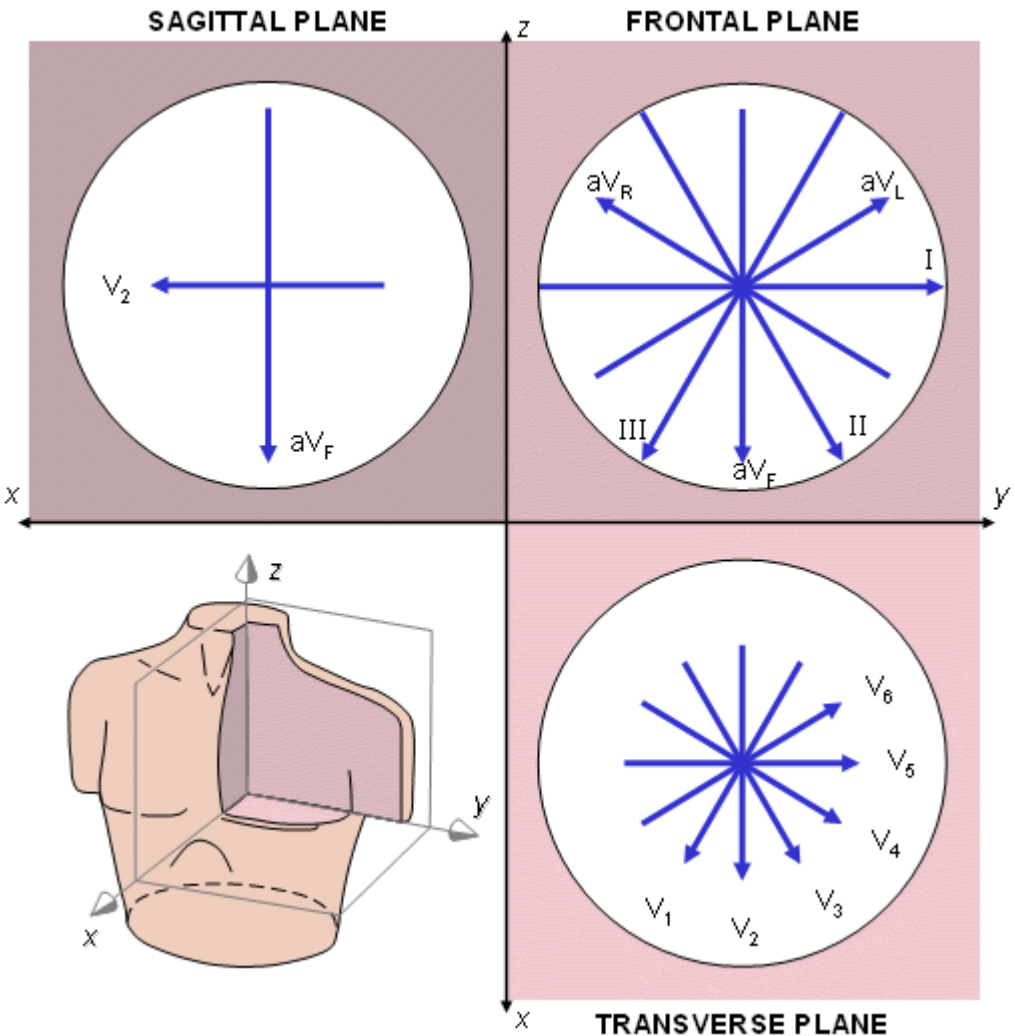


THE INFORMATION CONTENT OF THE 12-LEAD SYSTEM

- The most commonly used clinical ECG-system, the 12-lead ECG system, consists of the following 12 leads, which are
  - I, II, III
  - aV_R, aV_L, aV_F
  - V_1, V_2, V_3, V_4, V_5, V_6
- Of these 12 leads, the first six are derived from the same three measurement points. Therefore, any two of these six leads include exactly the same information as the other four.
- Over 90% of the heart's electric activity can be explained with a dipole source model (Geselowitz, 1964). To evaluate this dipole, it is sufficient to measure its three independent components. In principle, two of the limb leads (I, II, III) could reflect the frontal plane components, whereas one precordial lead could be chosen for the anterior-posterior component.

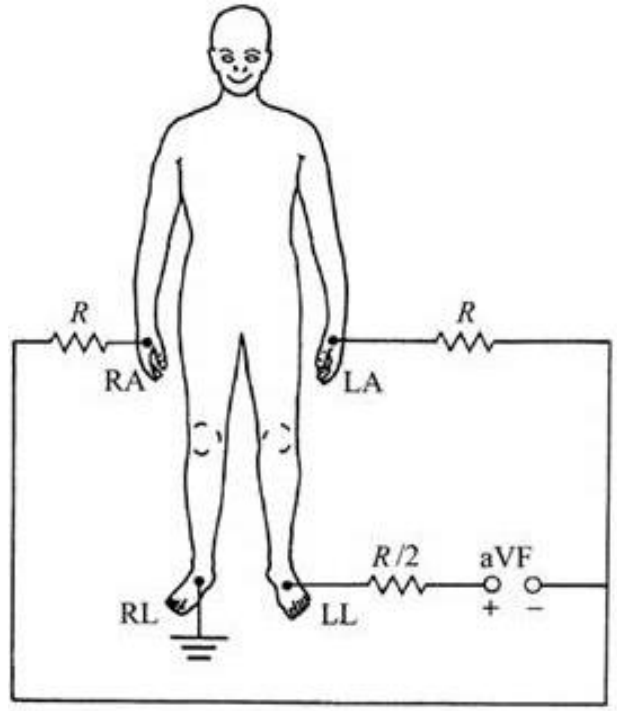
- The lead V2 would be a very good precordial lead choice since it is directed closest to the x axis while aVF is parallel to z axis
- The 12-lead ECG system could be thought to have three independent leads and nine redundant leads.

- However, in fact, the precordial leads detect also nondipolar components, which have diagnostic significance because they are located close to the frontal part of the heart.
- Therefore, the 12-lead ECG system has eight truly independent and four redundant leads.

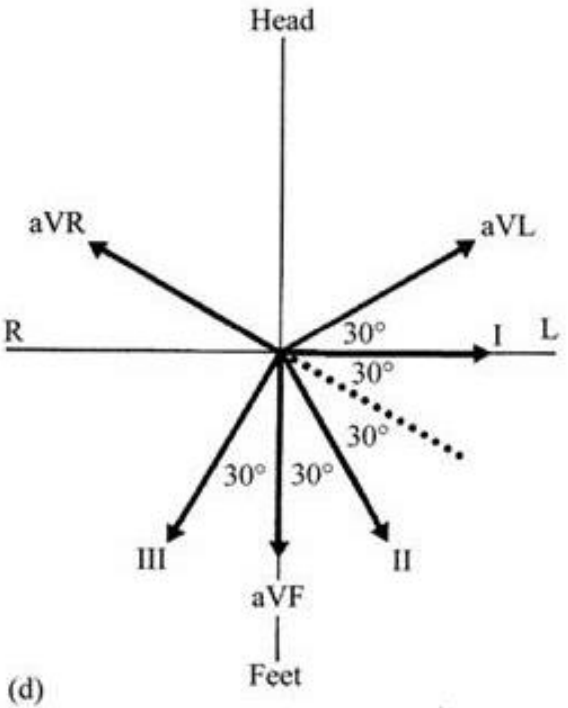


Problem:

The direct measurement of augmented foot lead aVF is shown below. Sketch the body, show the placement of electrodes, any resistors, amplifiers that may be used prior to the differential amplifier input and connections to the differential amplifier input.

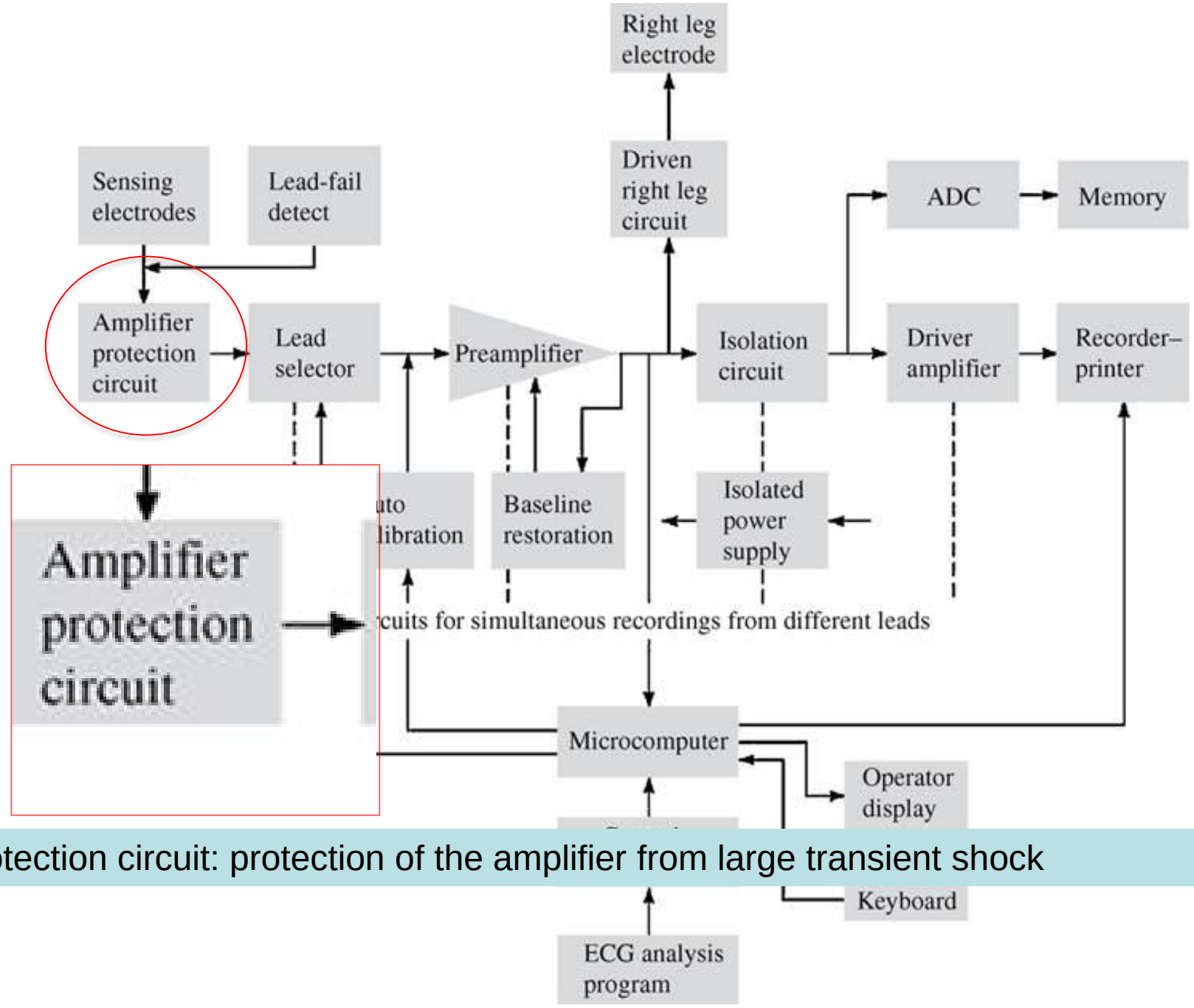


(c)



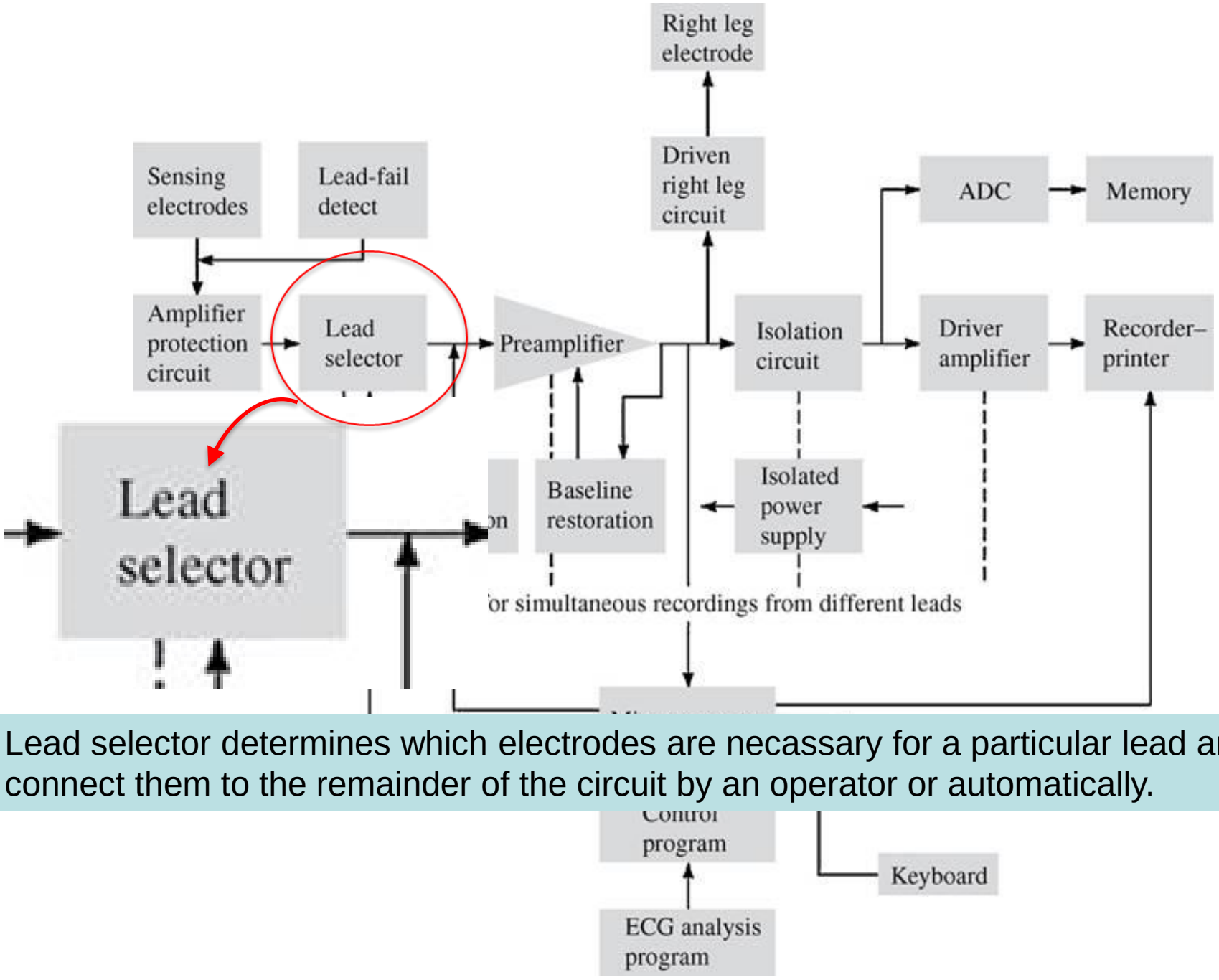
(d)

Block diagram of an electrocardiograph



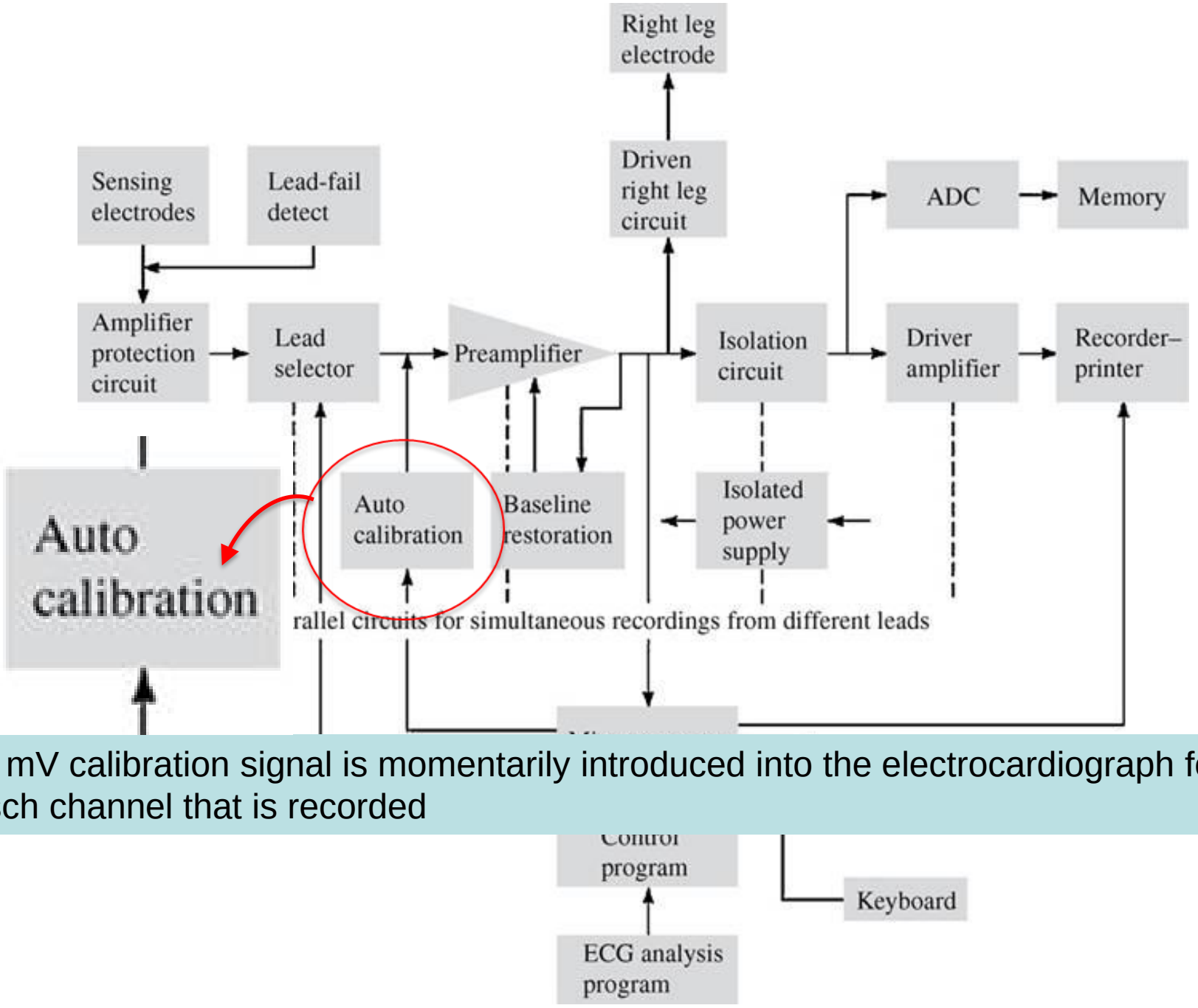
Protection circuit: protection of the amplifier from large transient shock

Block diagram of an electrocardiograph



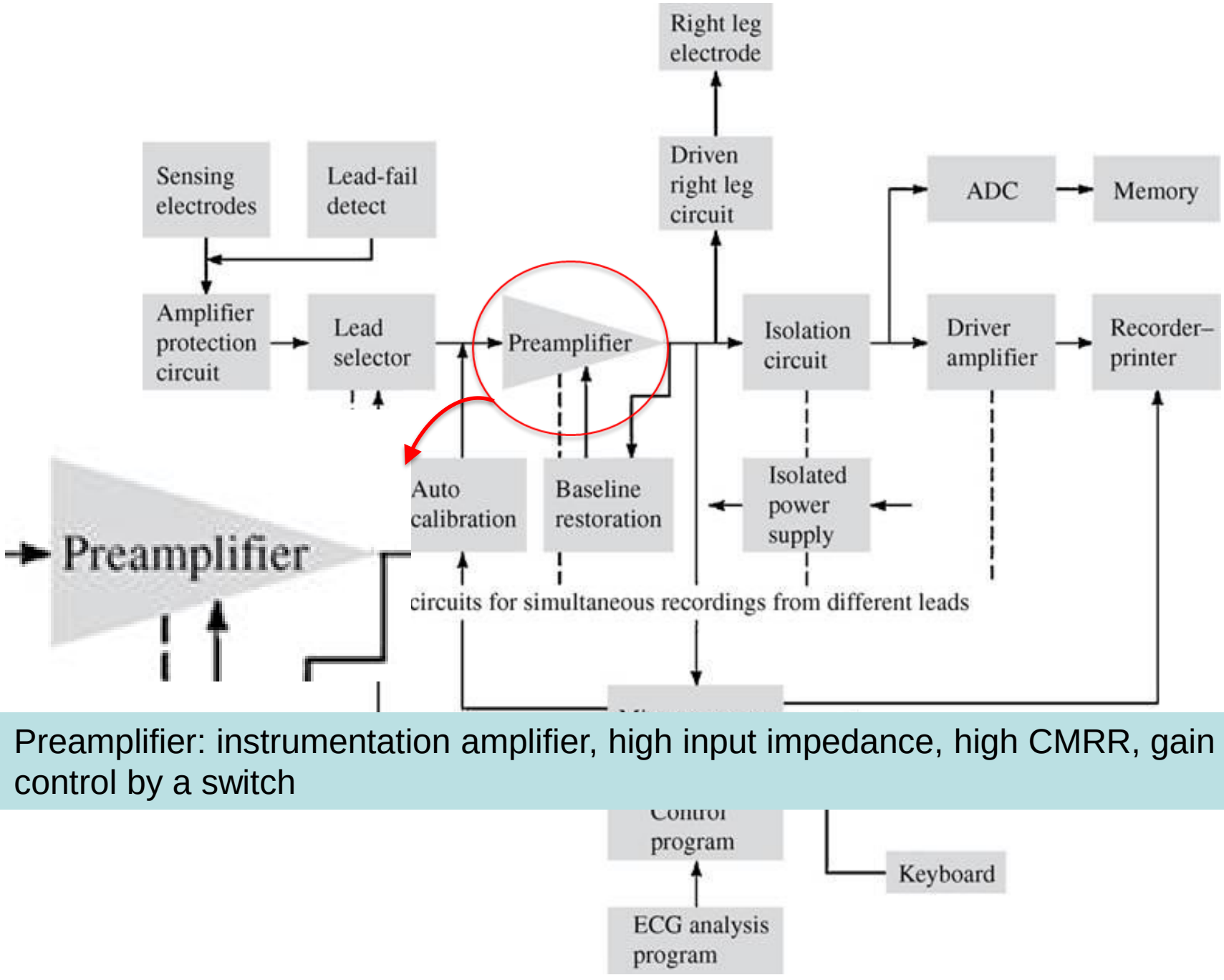
Lead selector determines which electrodes are necassary for a particular lead and to connect them to the remainder of the circuit by an operator or automatically.

Block diagram of an electrocardiograph



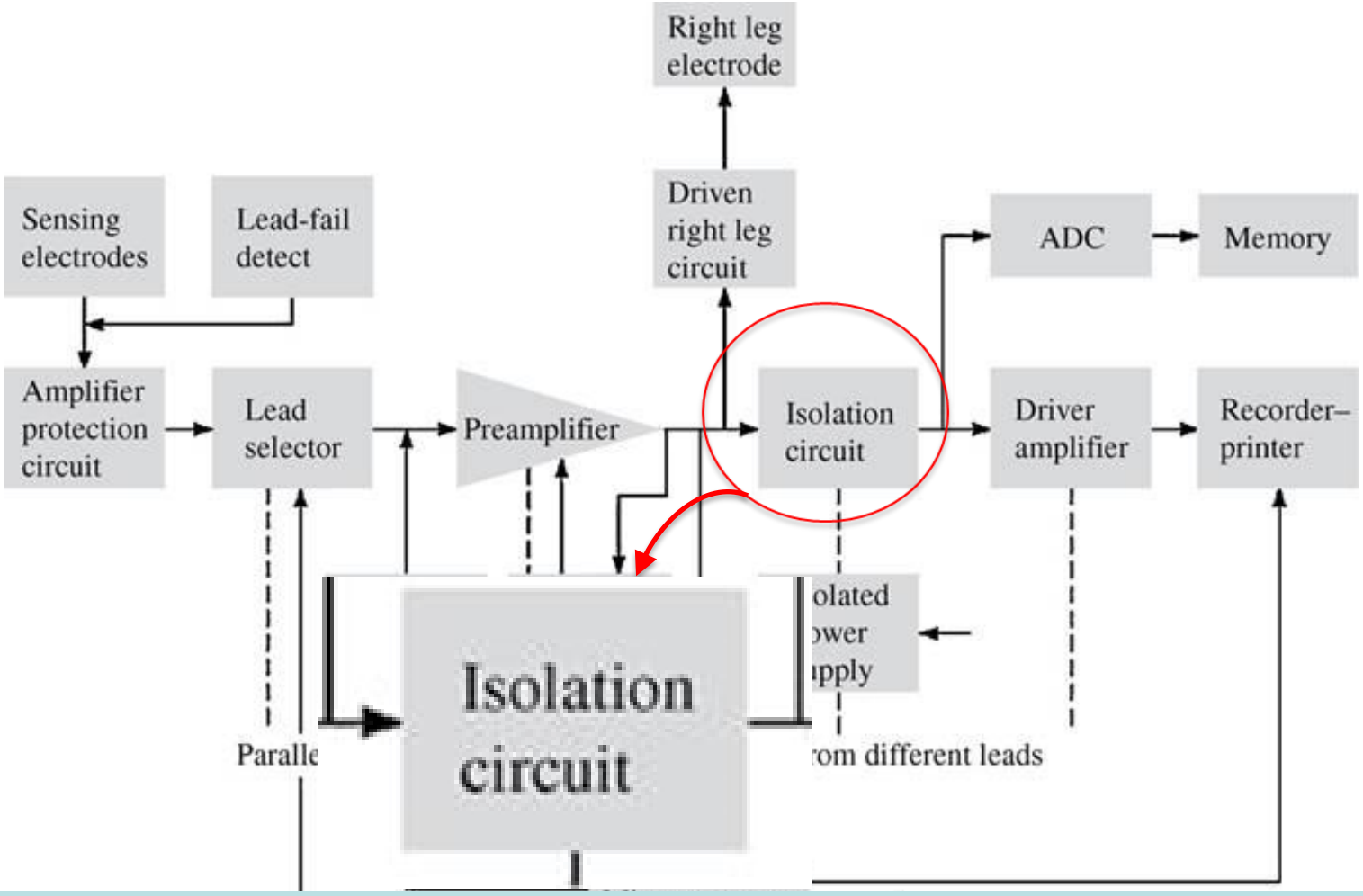
A 1 mV calibration signal is momentarily introduced into the electrocardiograph for easch channel that is recorded

Block diagram of an electrocardiograph



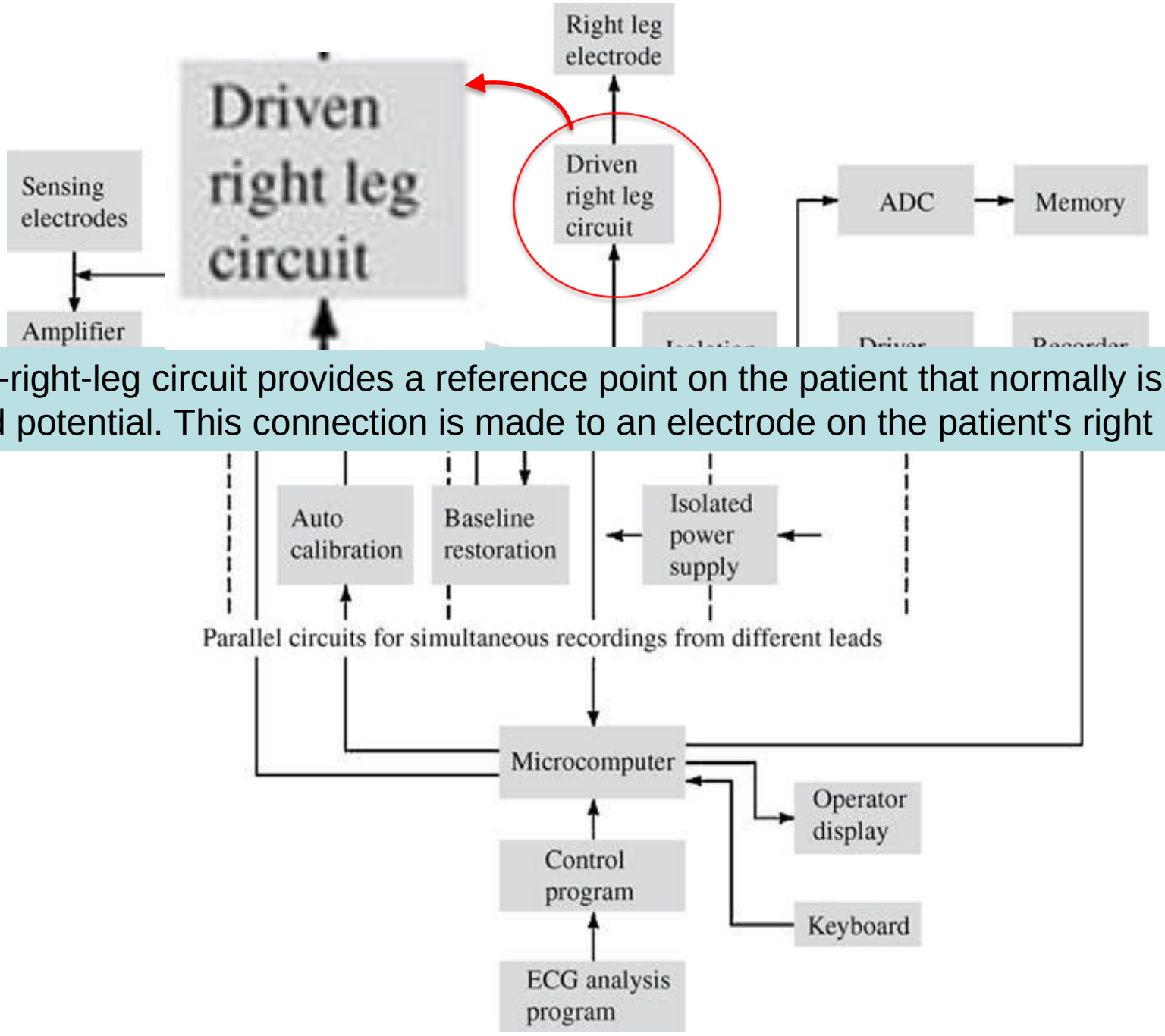
Preamplifier: instrumentation amplifier, high input impedance, high CMRR, gain control by a switch

Block diagram of an electrocardiograph



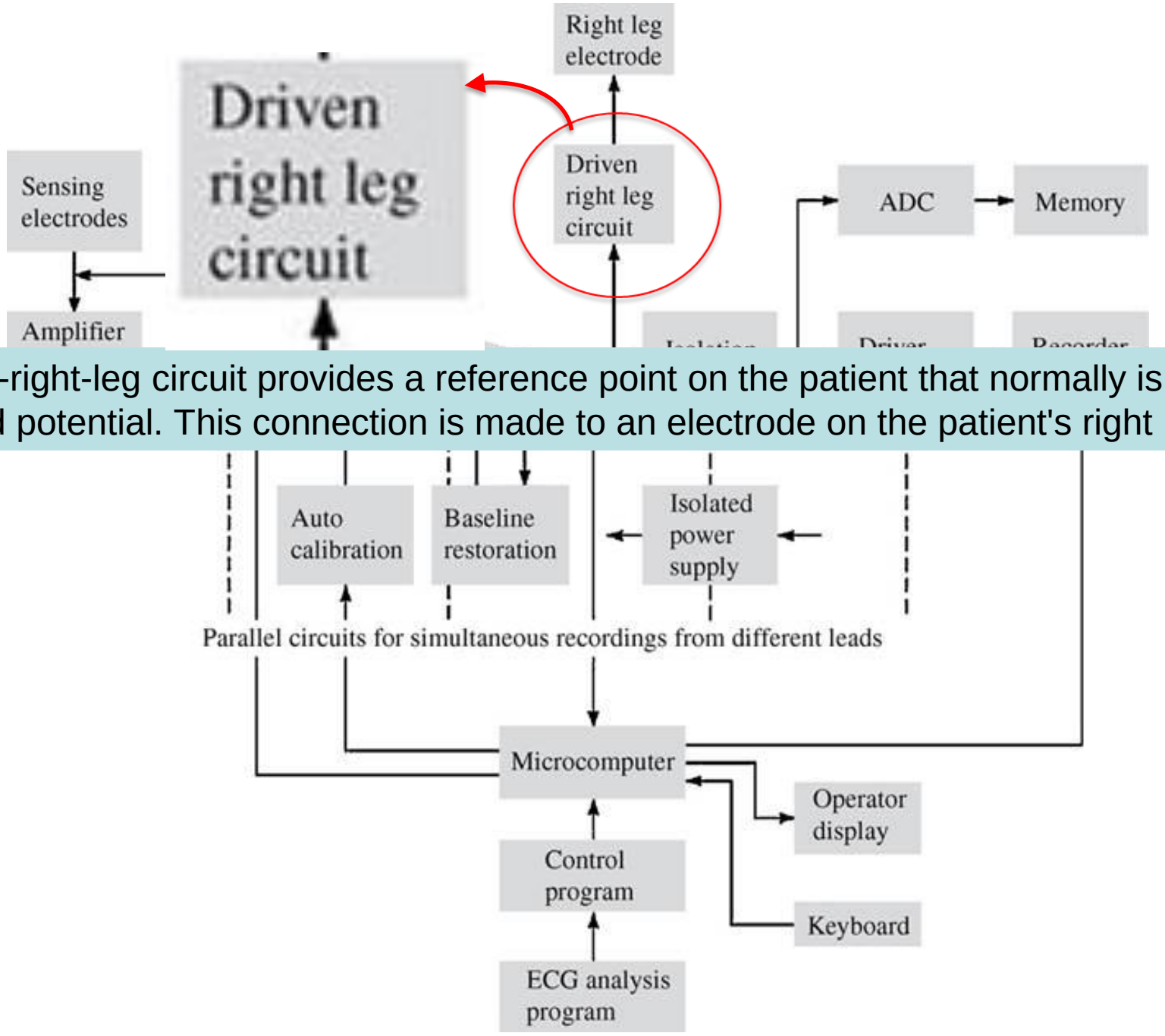
Isolation circuit: safety of the patient, separate patient ground from earth ground tto prevent dangerous currents from flowing from the patient.

Block diagram of an electrocardiograph



Driven-right-leg 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.

Block diagram of an electrocardiograph



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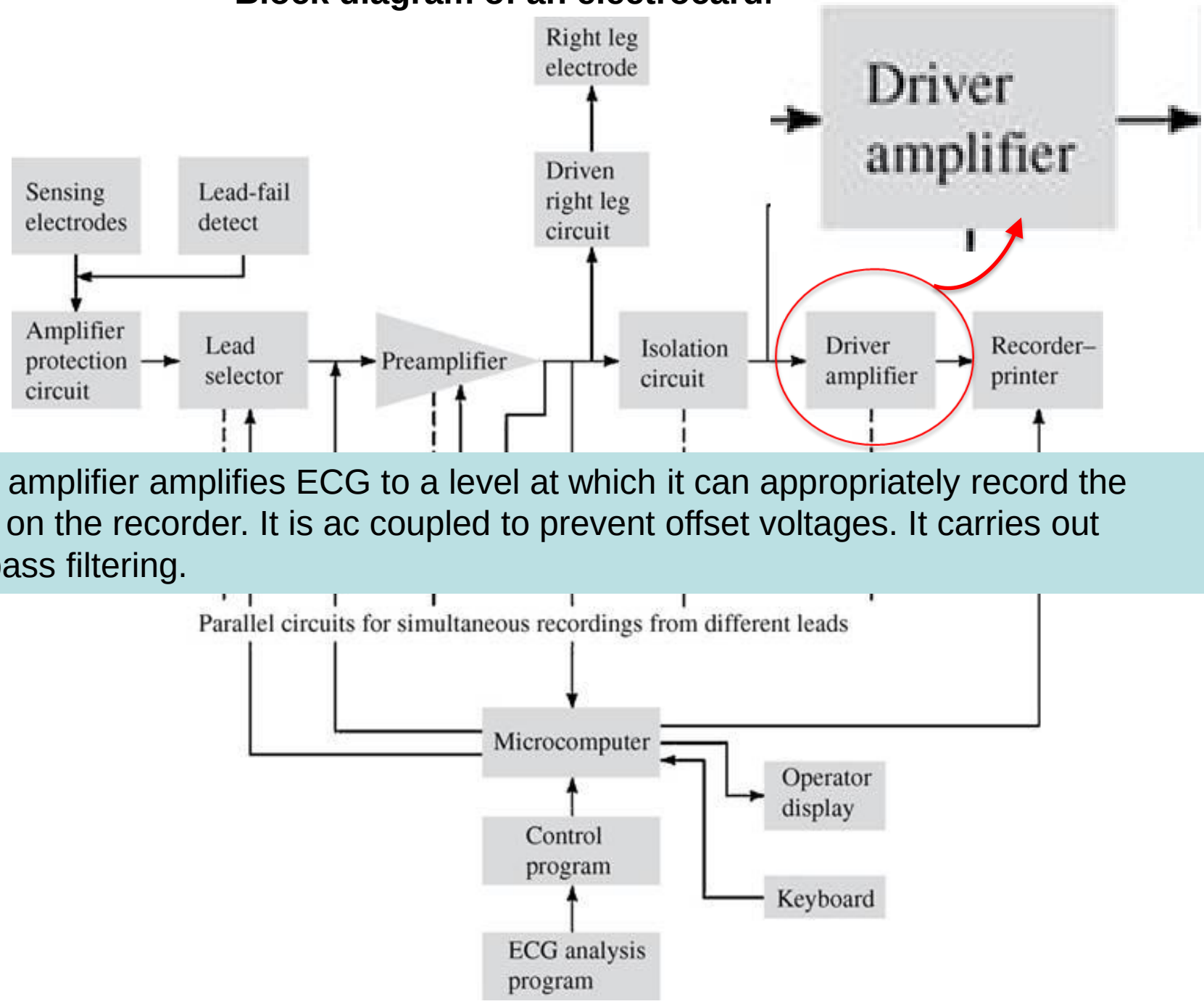
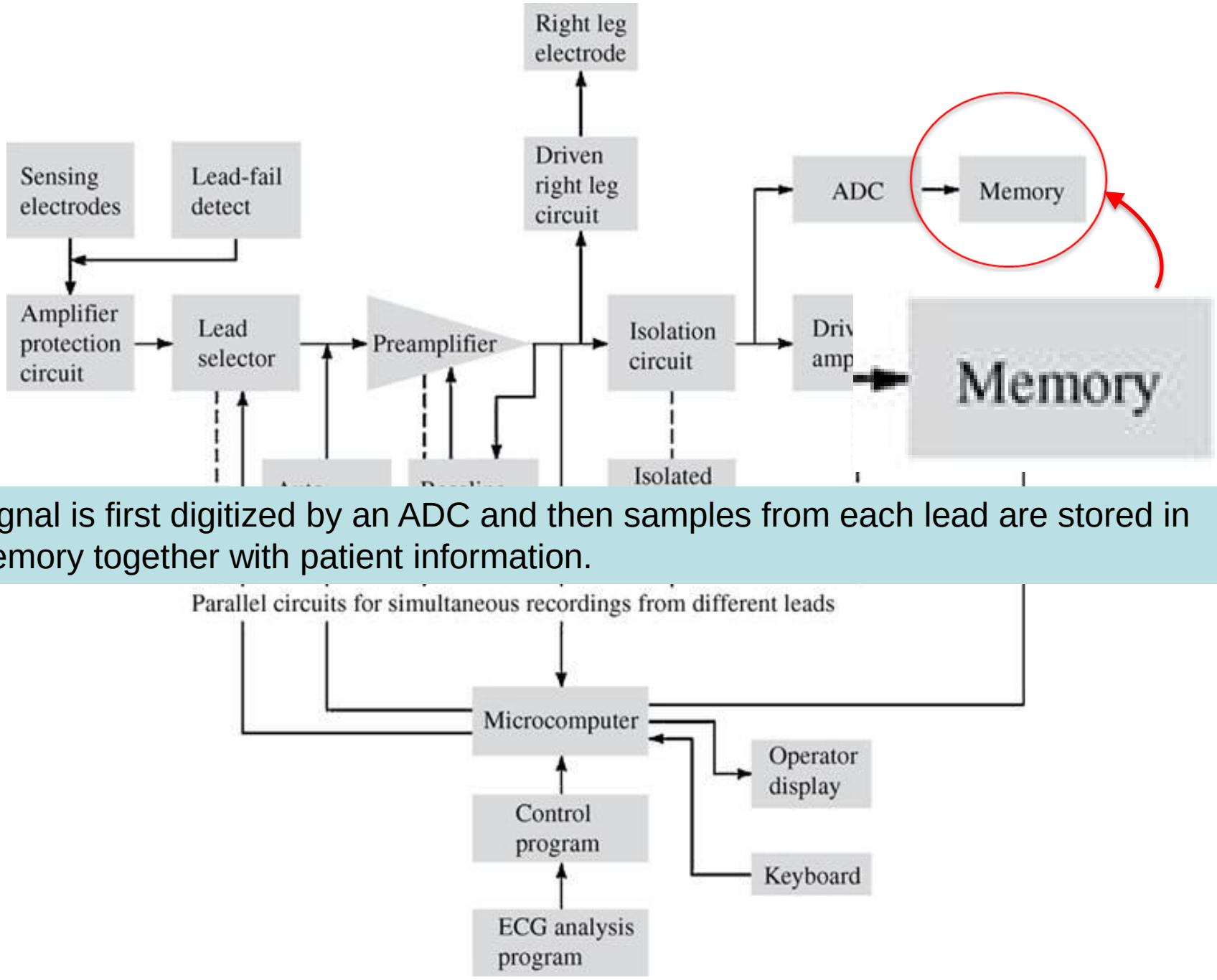
Block diagram of an electrocardi



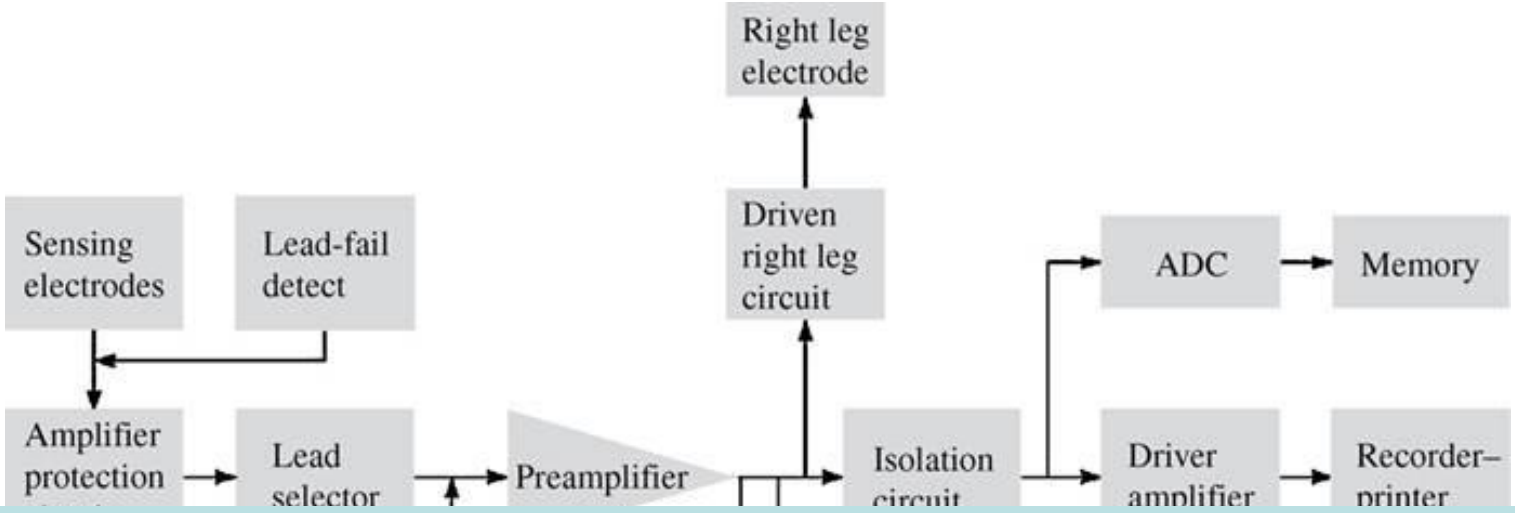
Driver amplifier amplifies ECG to a level at which it can appropriately record the signal on the recorder. It is ac coupled to prevent offset voltages. It carries out bandpass filtering.

Block diagram of an electrocardiograph

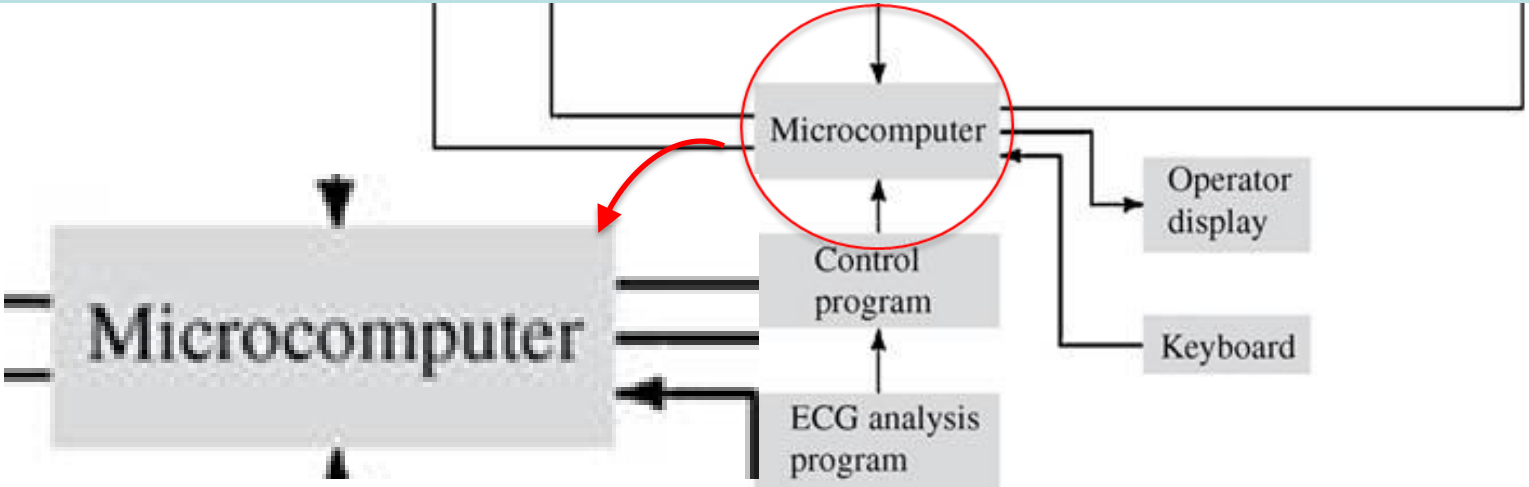


The signal is first digitized by an ADC and then samples from each lead are stored in the memory together with patient information.

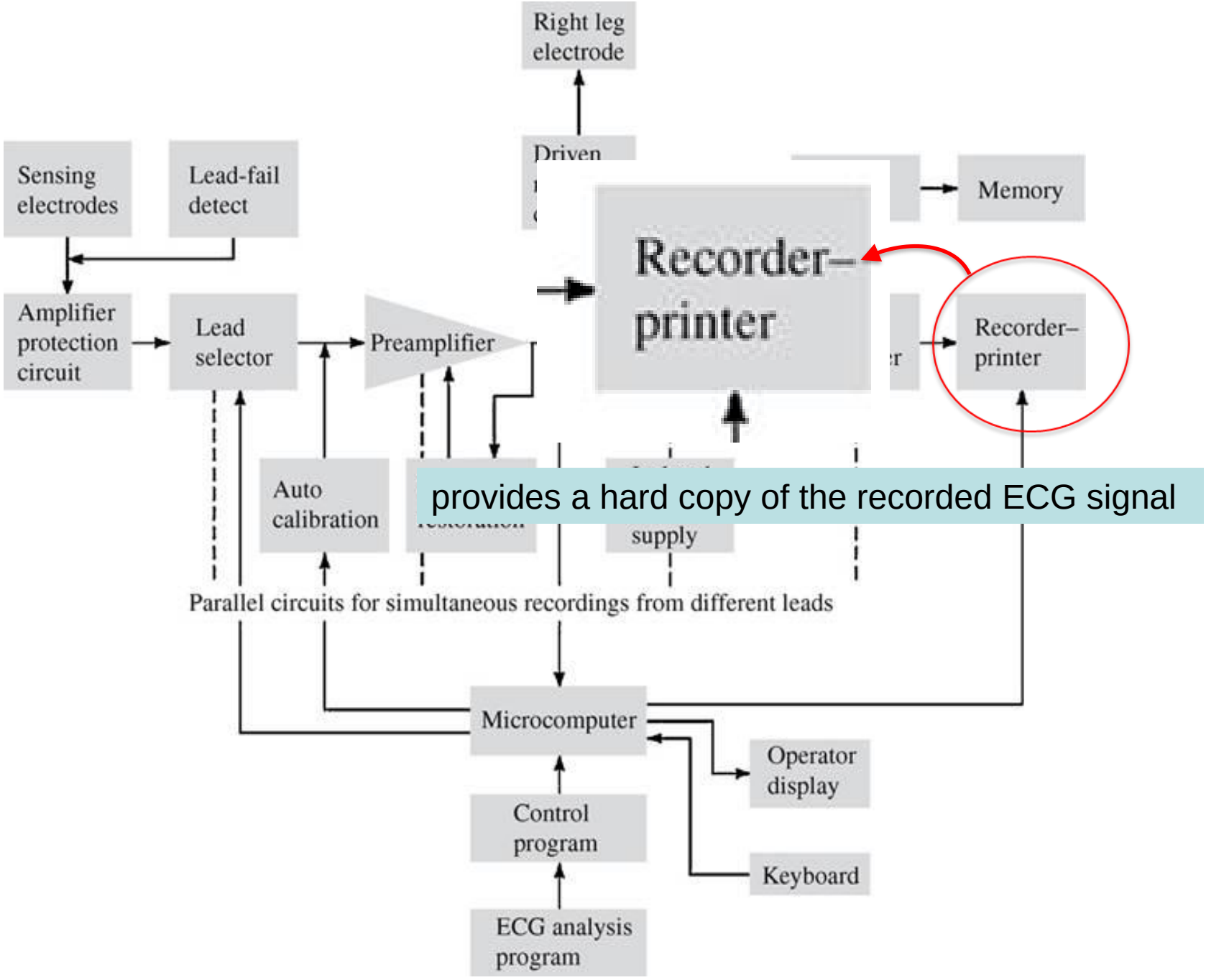
Block diagram of an electrocardiograph



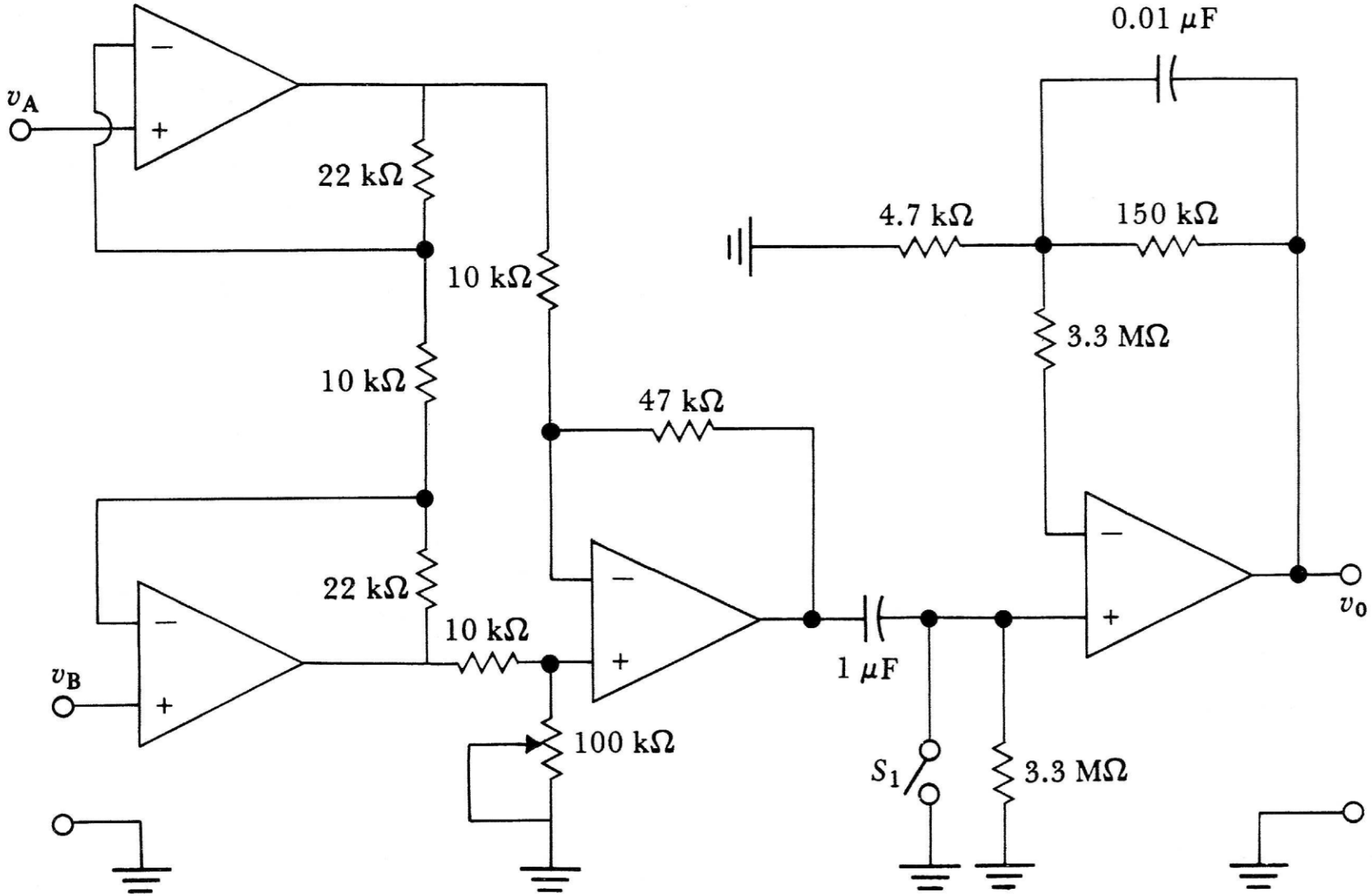
- Microcomputer: control, digital signal processing, storage, user interface, communication, etc.
- It can be programmed to generate the standard 12-lead electrocardiogram by selecting three simultaneous 10s segments of the six frontal plane leads.
- It can make preliminary analysis to determine the heart rate, recognize arrhythmia. It can communicate with user via a keyboard and screen.



Block diagram of an electrocardiograph



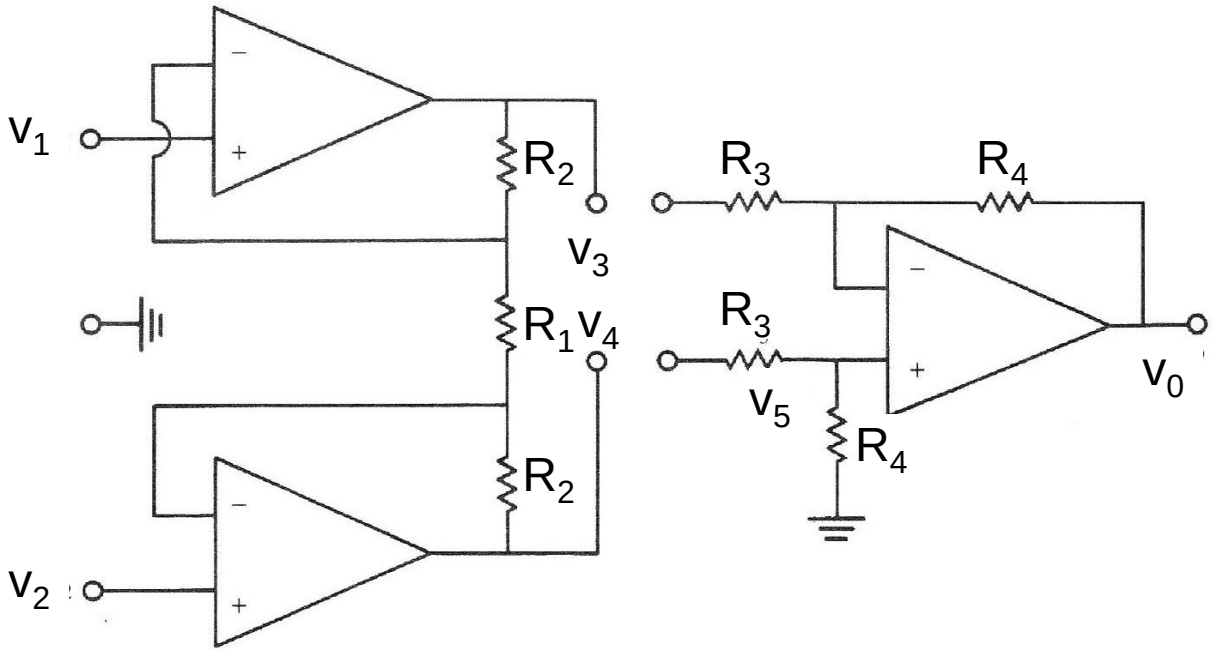
The circuit of an ECG amplifier



DIFFERENTIAL AMPLIFIER

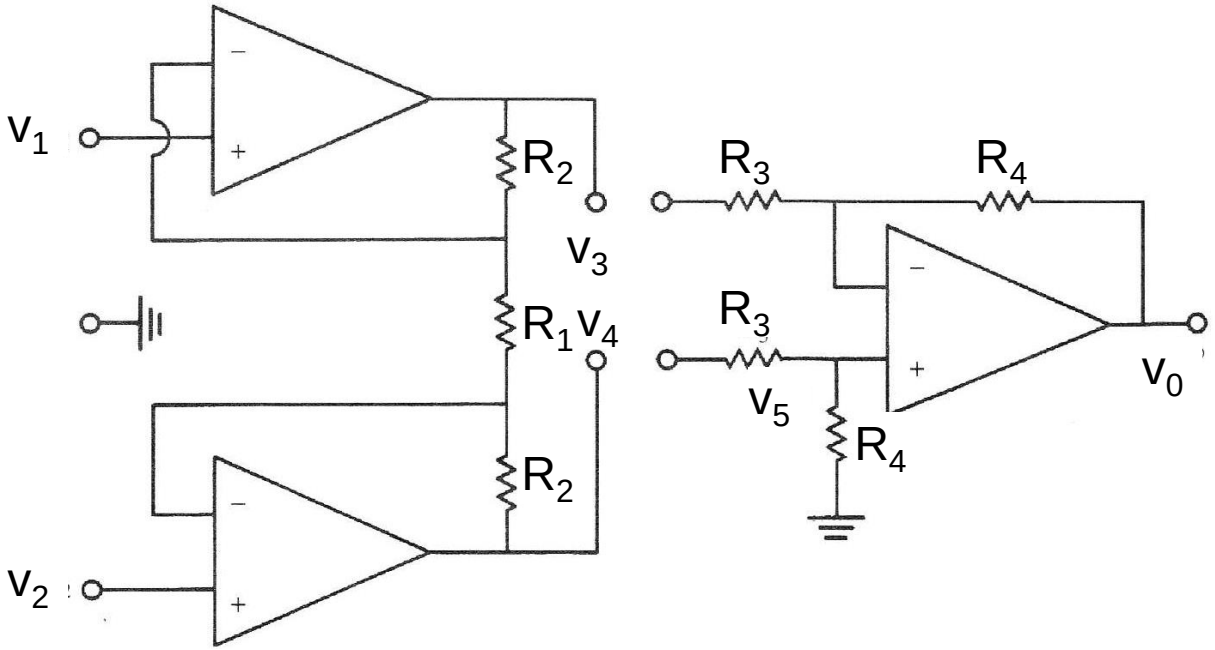
- The right side of figure shows a simple one-op-amp differential amplifier.
- By Rule 2, $v_5 = \dfrac{v_4 R_4}{R_3 + R_4}$ By Rule 1, $i = \dfrac{v_3 - v_5}{R_3} = \dfrac{v_5 - v_o}{R_4}$

$$v_o = \frac{(v_4 - v_3)R_4}{R_3}$$



DIFFERENTIAL AMPLIFIER

- When $v_4=v_3$, the differential amplifier-circuit (not op-amp) common-mode gain G_c is 0.
- When $v_4 \neq v_3$ differential gain G_d is equal to R_4/R_3
- No differential amplifier perfectly rejects the common-mode voltage. To quantify this imperfection, we use the term common-mode rejection ratio (CMRR) is defined.
- CMMR=100 for some oscilloscope differential amplifiers
- CMMR>10,000 for a high-quality biopotential amplifier.

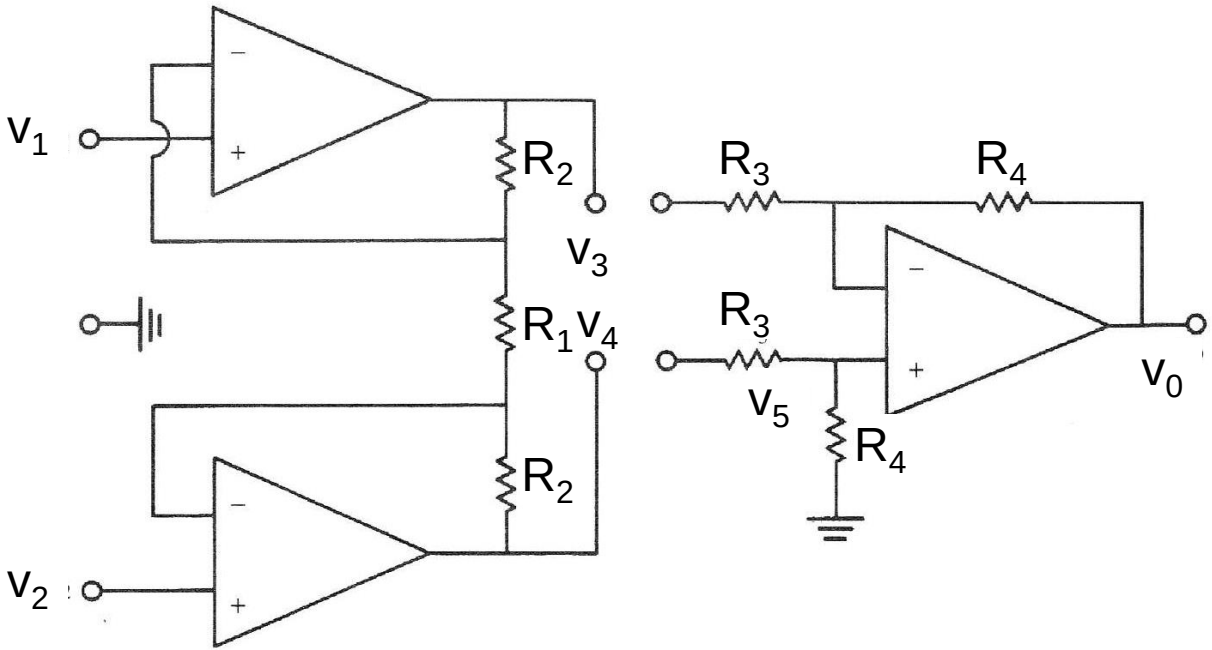


$$v_o = \frac{(v_4 - v_3)R_4}{R_3}$$

$$\text{CMRR} = \frac{G_d}{G_c}$$

THREE-OP-AMP DIFFERENTIAL AMPLIFIER

- Right half of the circuit has low input impedance.
- Adding two followers separately to inputs can increase the input impedance but does not affect CMMR.
- Using circuit on the left both CMMR and input impedance are increased.
- The resulting three-op-amp amplifler circuit is frequently called an instrumentation amplifier.
- This circuit finds wide use in measuring biopotentials because it rejects the large 60 Hz common-mode voltage that exists on the body.



$$G_d = \frac{v_3 - v_4}{v_1 - v_2} = \frac{2R_2 + R_1}{R_1}$$

High Input Resistance of Voltage Follower Configuration

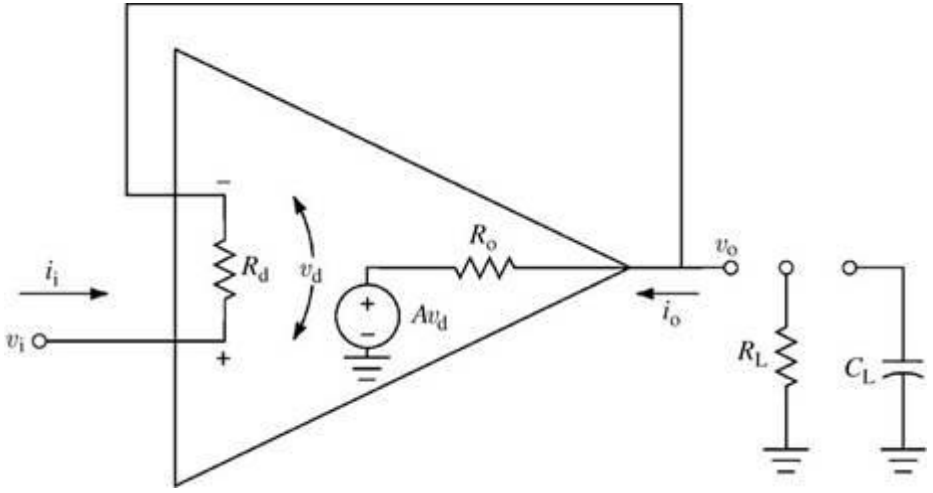
- In order to calculate the amplifier-circuit input resistance R_{ai}, assume a change in input voltage v_i.
- Amplifier-circuit input resistance R_{ai} is about $(10^5) \times (2\ \text{M}\Omega) = 200\ \text{G}\Omega$.
- The amplifier input impedance is much higher than the op-amp input impedance R_d.
- The amplifier output impedance is much smaller than the op-amp output impedance R_o.
- Noninverting amplifiers: R_{ai} is very high.
- Inverting amplifier: R_{ai} is usually small.

$$\Delta v_o = A\Delta v_d = A(\Delta v_i - \Delta v_o)$$

$$= \frac{A\Delta v_i}{A+1}$$

$$\Delta i_i = \frac{\Delta v_d}{R_d} = \frac{\Delta v_i - \Delta v_o}{R_d} = \frac{\Delta v_i}{(A+1)R_d}$$

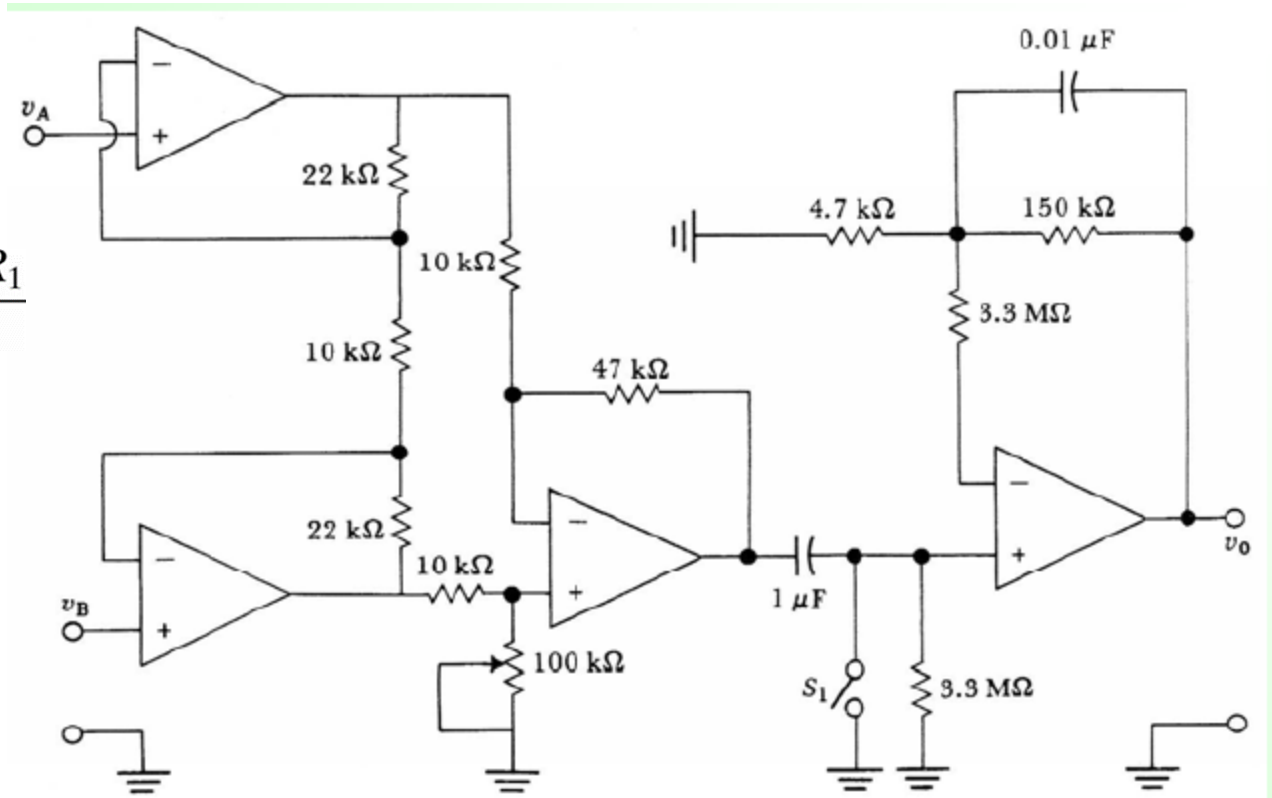
$$R_{ai} = \frac{\Delta v_i}{\Delta i_i} = (A+1)R_d \cong AR_d$$

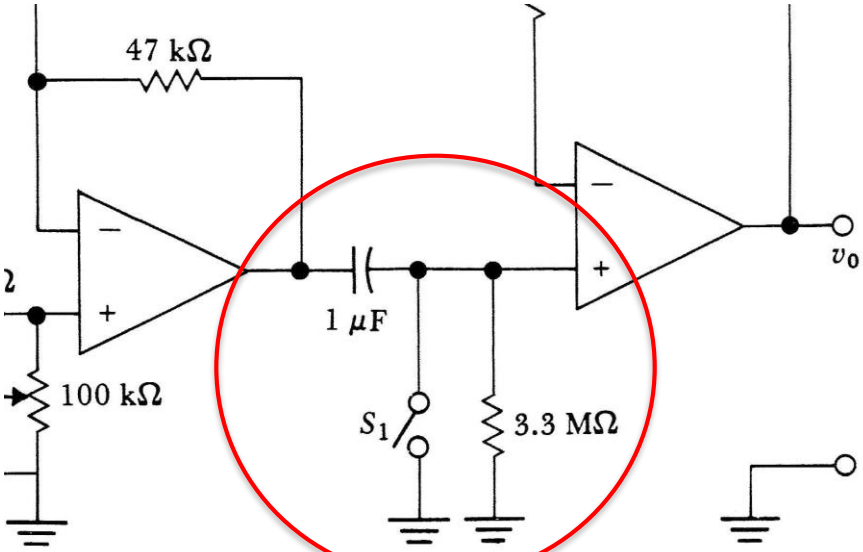


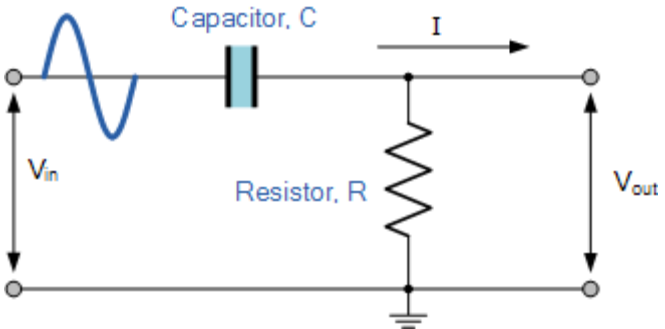
- High commonmode rejection is achieved by adjusting the potentiometer to about 47 kΩ.
- Electrodes may produce an offset potential of up to 0.3 V. Thus, to prevent saturation, the decoupled stages have a gain of only 25.
- Coupling capacitors are not placed at the input because this would block the op-amp bias current. Adding resistors to supply the bias current would lower the Zin.
- Coupling capacitors placed after the first op amps would have to be impractically large. Therefore, the single 1 μF coupling capacitor is aded after instrumentation amplifier.
- 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.

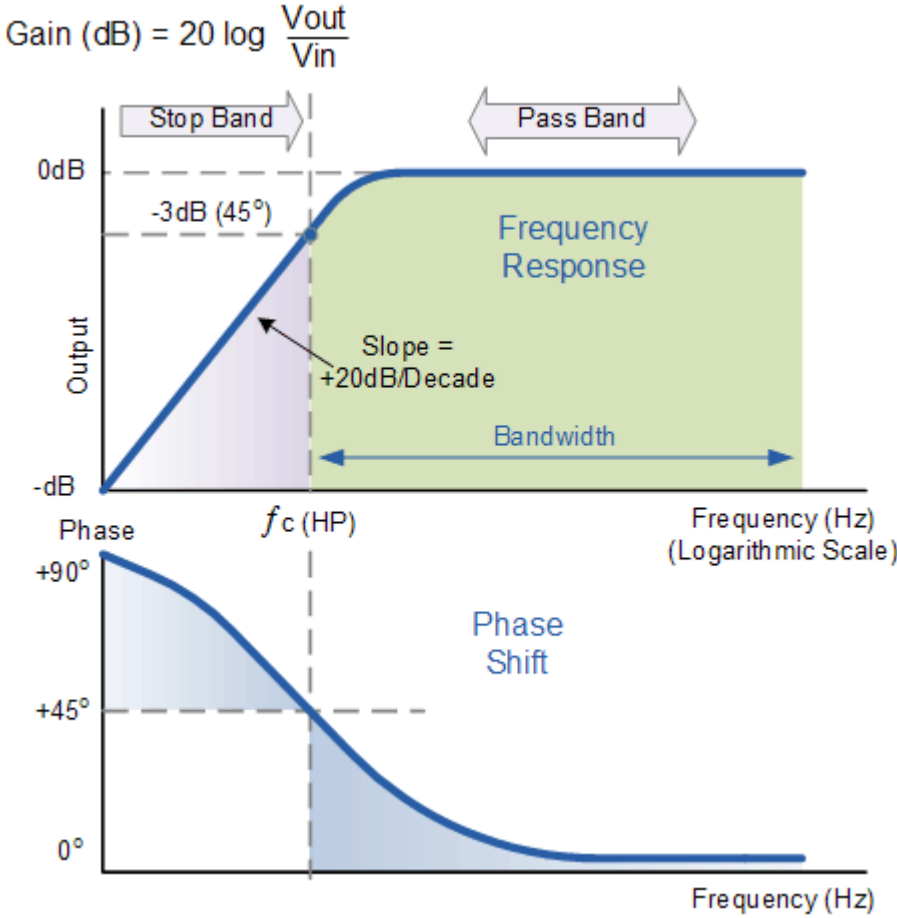
$$G_d = \frac{v_3 - v_4}{v_1 - v_2} = \frac{2R_2 + R_1}{R_1}$$

$$v_o = \frac{(v_4 - v_3)R_4}{R_3}$$







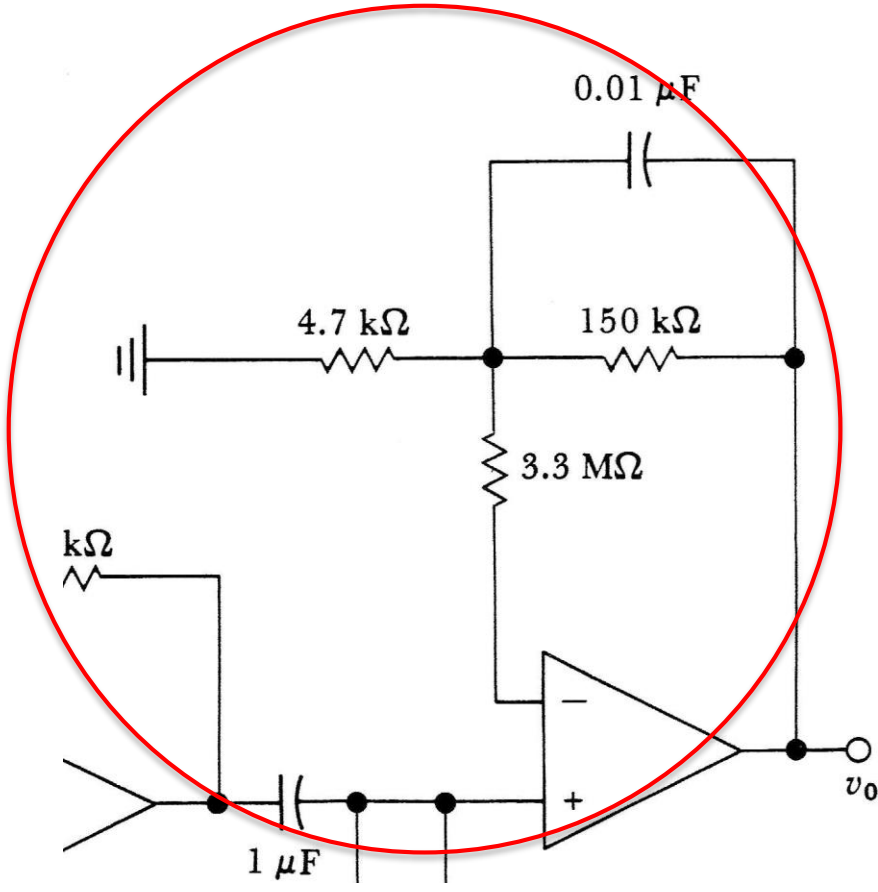


$$f_c = \frac{1}{2\pi RC} = 0.048\text{Hz}$$

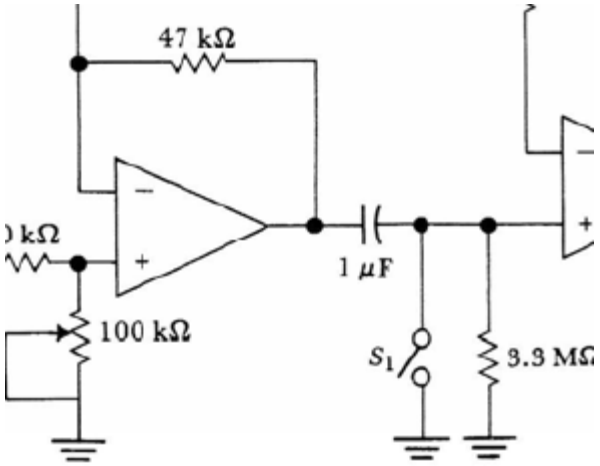
NonInverting Amplifier + Low Pass Filter

- The output stage is a noninverting amplifier that has a gain of 32.
- 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 100Hz.

$$f_c = \frac{1}{2\pi RC} = 106\text{Hz}$$

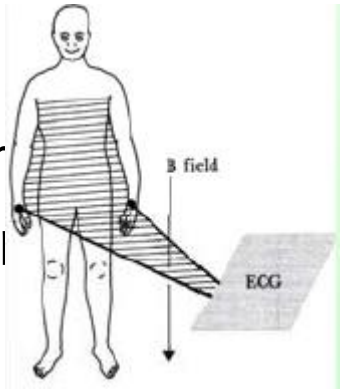


- Switch S1 may be momentarily closed to decrease the discharge time constant when the output saturates.
- This is required after defibrillation or lead switching to charge the 1μF capacitor rapidly to the new value and return the output to the linear region.
- We do not discharge the capacitor voltage 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.



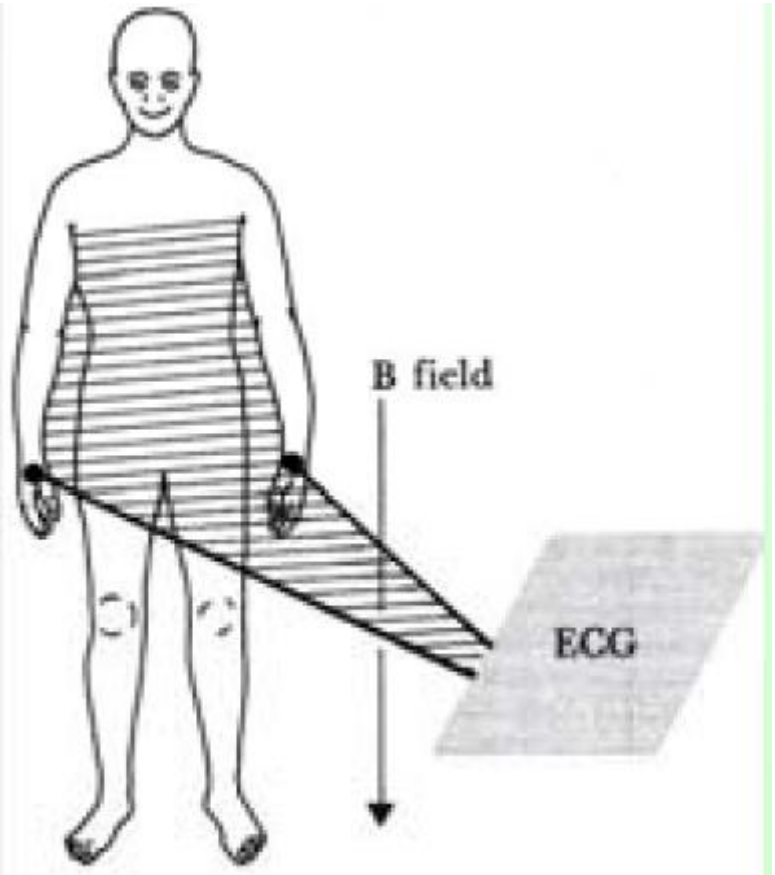
Frequent Problems

- Frequency Distortion
  - if filter specification does not match the frequency content of biopotential then the result is high and low frequency distortion
- Saturation or cut-off distortion
  - high electrode offset voltage or improperly calibrated amplifiers can drive the amplifier into saturation then the peaks of QRS waveforms are cut off
- Ground loops
  - Cause: multiple instruments with different ground potentials on one patient
  - May cause safety problem
  - Increased common mode voltage
- Electric/magnetic field coupling
  - open lead wires (floating connections) pick up EMI
  - long leads produce loop that picks up EMI (induces loop cur
- Interference from power lines (common mode interfe
  - can couple onto ECG signal



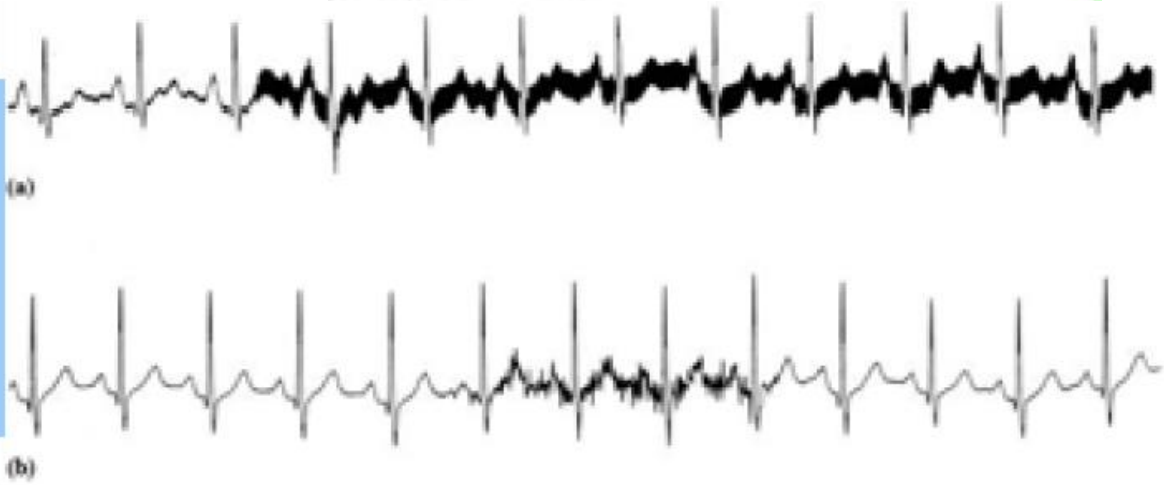
Frequent Problems

- Interference from power lines (common mode interference)
  - can couple onto ECG signal



60Hz supply noise

Coupled to ECG



Artifact from large electric transients

- Patient is having an ECG taken, cardiac defibrillation may be required.
- High-voltage high-current electric pulse is applied to the chest of the patient so that transient potentials can be observed across the electrodes.
- Potentials can be several orders of magnitude higher than the normal potentials encountered in the ECG causing saturation of amplifiers
- Pulse is sufficiently large to cause the buildup of charge on coupling capacitances in the amplifier,
- Results in 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.
- Protection circuit usually speeds up the recovery

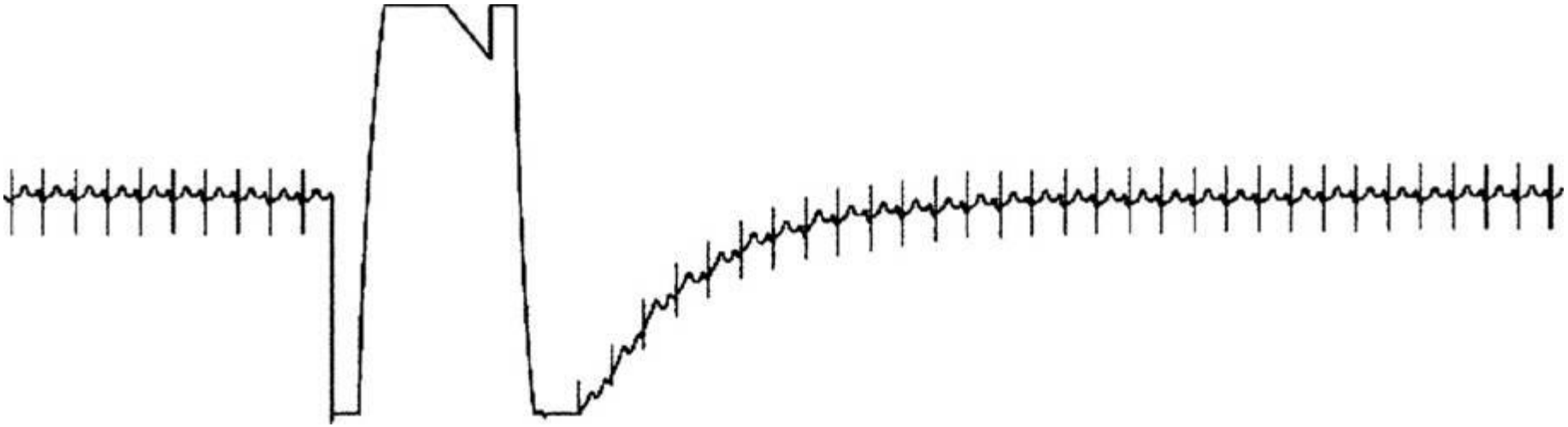


Figure 6.8 Effect of a voltage transient on an ECG recorded on an electrocardiograph in which the transient causes the amplifier to saturate, and a finite period of time is required for the charge to bleed off enough to bring the ECG back into the amplifier's active region of operation. This is followed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

Problem

You designed an ECG system for a clinician with a large gain such that any voltage greater than ±3 mV will be out of range for display on the computer. Your ECG system design follow a first order response with a time constant of 20 s (refer to chapter 1).

The clinician was recording ECG for a patient and switched from lead I to lead II. Because of different offset potential at each electrode a transient spike occurred of amplitude 20 mV.
The R wave of the patients ECG was 1 mV.

Calculate how long the clinician has to wait for the entire signal to be visible on the display? Is this acceptable (refer table 6.1)

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

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

Not acceptable!!!!!

Return time after lead switch	Max	s	1

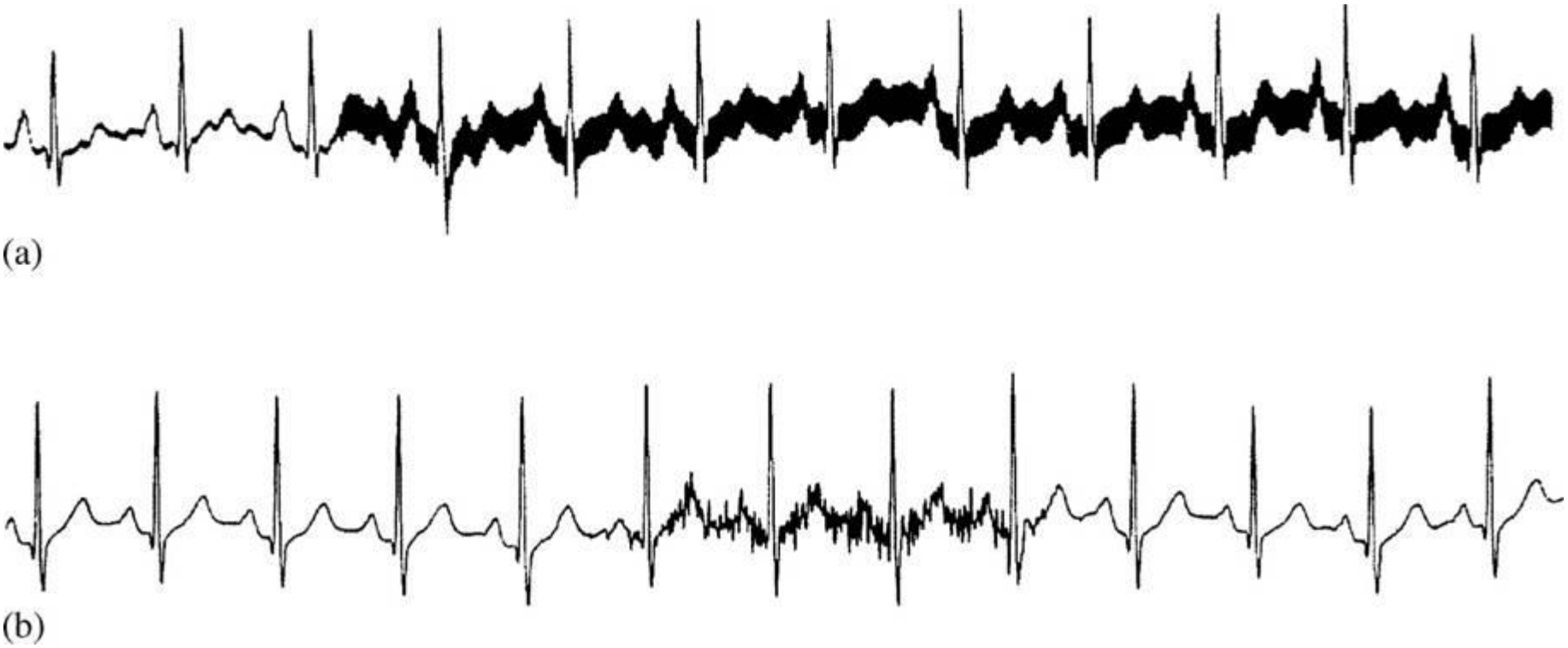


Figure 6.9 (a) 60 Hz power-line interference, (b) Electromyographic interference on the ECG.

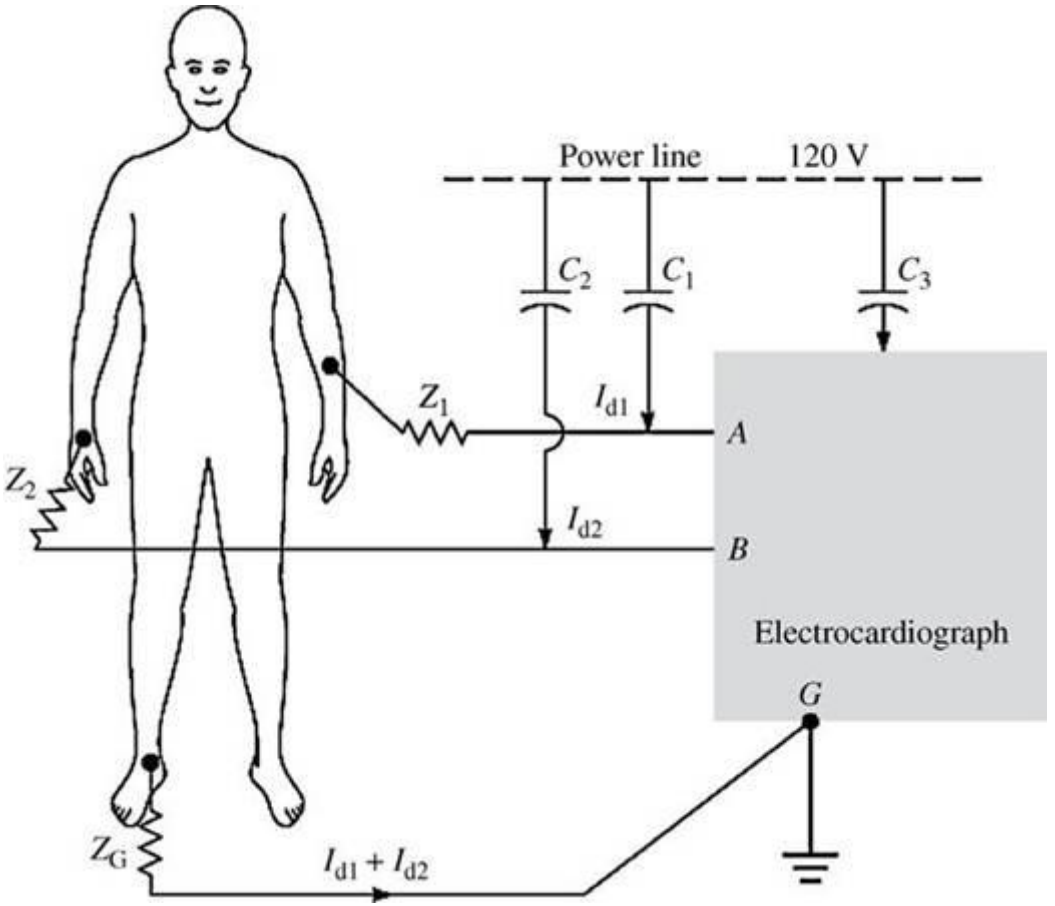
Problem

- You designed and shipped an Electrocardiograph to a clinician to acquire ECG. You selected a uA741 op-amps such that the input impedance of each differential input of your ECG machine to ground was 10 MΩ. You estimated that the CMRR for your device was 80 dB and the displacement current due skin-electrode impedance (Z1 and ZG) and (Z2 and ZG) and the subject to be about 6 nA.
- However, you forgot to impedance match the electrodes. Such that $ZG = Z1 = 1\ M\Omega$, $Z2 = 1.2\ M\Omega$

- Now you are worried that the interference caused due to mismatch of electrodes would create an objectionable amount of ECG.
- Show necessary calculations to confirm your worry.
- What actions you will take to minimize this noise?

Mechanism of electric-field pickup of an electrocardiograph resulting from the power line.

- 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 and C_2 represents the capacitance between the power line and the leads.



- i_{d1} and i_{d2} does not flow into the electrocardiograph because of its high input impedance,
- Flow through the skin–electrode impedances Z_1 and Z_G to ground and Z_2 and Z_G to ground respectively
- Body impedance, about 500 Ω, can be neglected when compared with the other impedances shown.
- Voltage amplified is that appearing between inputs *A* and *B*, v_A - v_B.

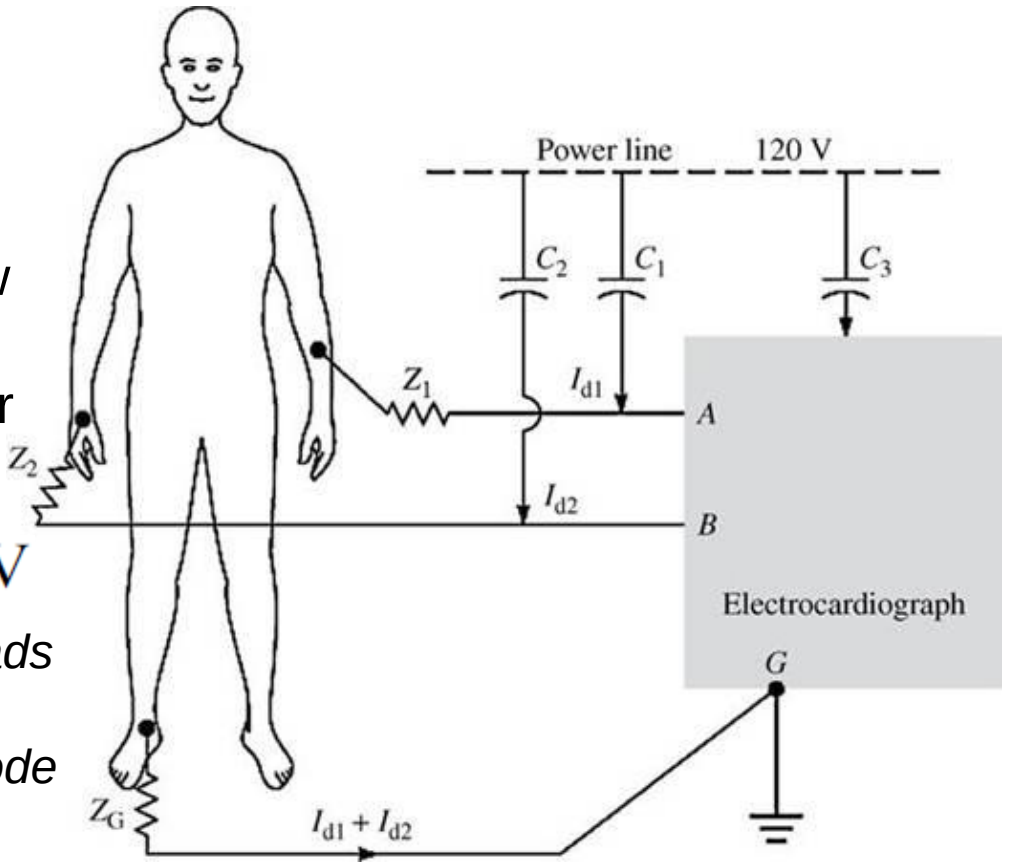
$$v_A - v_B = i_{d1}Z_1 - i_{d2}Z_2$$

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

Values measured for 9 m cables show that $i_d \cong 6$ nA,
Skin–electrode impedances may differ by as much as 20 kΩ

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

•To reduce this interference, shield the leads and ground each shield at the electrocardiograph and Lower skin-electrode impedance.

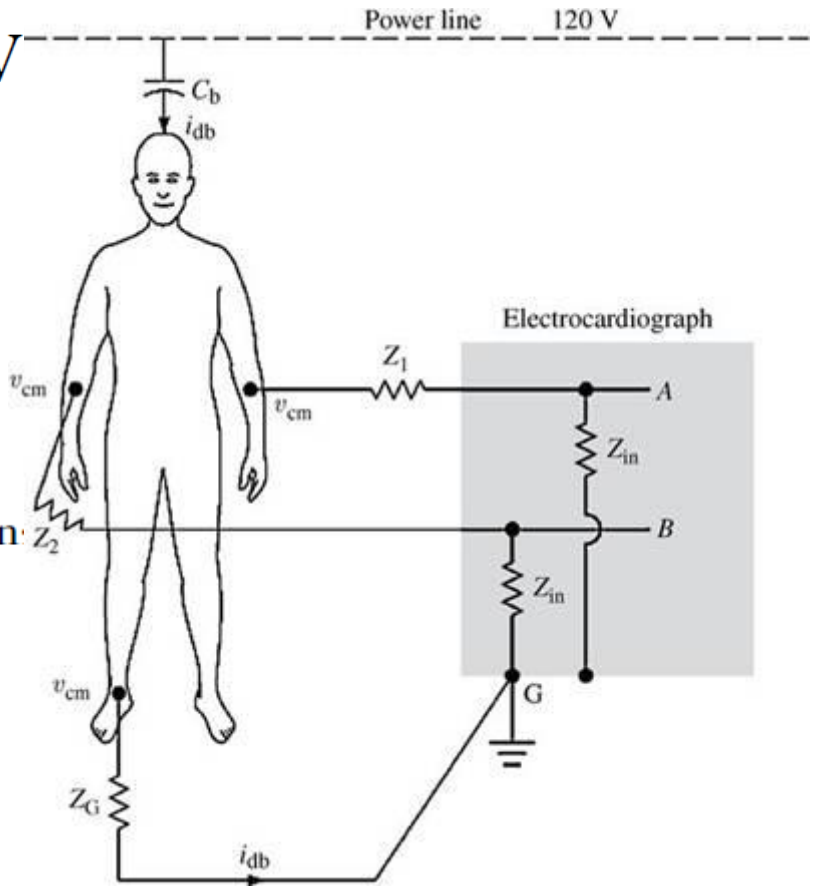


Problem

- But then you also realized that there is a displacement current that flows from the power line through the body and ground impedance.
- You estimated this displacement current (i_{db}) ≅ 200 nA.
- This create a common-mode voltage everywhere on the body.
- Would this common-mode voltage create objectionable interference on the ECG?
- Show necessary calculation to confirm this?
- What strategy would you implement to minimize this common-mode noise?

Current flows from the power line through the body and ground impedance, thus creating a common-mode voltage everywhere on the body. Substituting typical values yields

$$v_{cm} = i_{db} Z_G = (0.2\ \mu A)(50\ k\Omega) = 10\ mV$$



Amplifiers have finite input impedance, thus

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

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

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

Substituting typical values yields

$$v_A - v_B = (10\ mV)(20\ k\Omega / 5\ M\Omega) = 40\ \mu V$$

•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.

•So the input imbalance and *Zin are critical factors* determining the common-mode rejection, no matter how good the differential amplifier itself is.

EXAMPLE

- 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 1,500Ω at EEG frequencies while the other had a higher impedance of 4,700Ω. A ground electrode having an impedance of 2,500Ω was also used. The input impedance of each differential input of the EEG machine to ground was 10 MΩ and the instrument had a common mode rejection ratio 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 is

$$v_{cm} = i_{db} Z_G \qquad v_{cm} = 400 \times 10^{-9} \text{ A } (2{,}500\ \Omega) = 10^{-3} \text{ V}$$

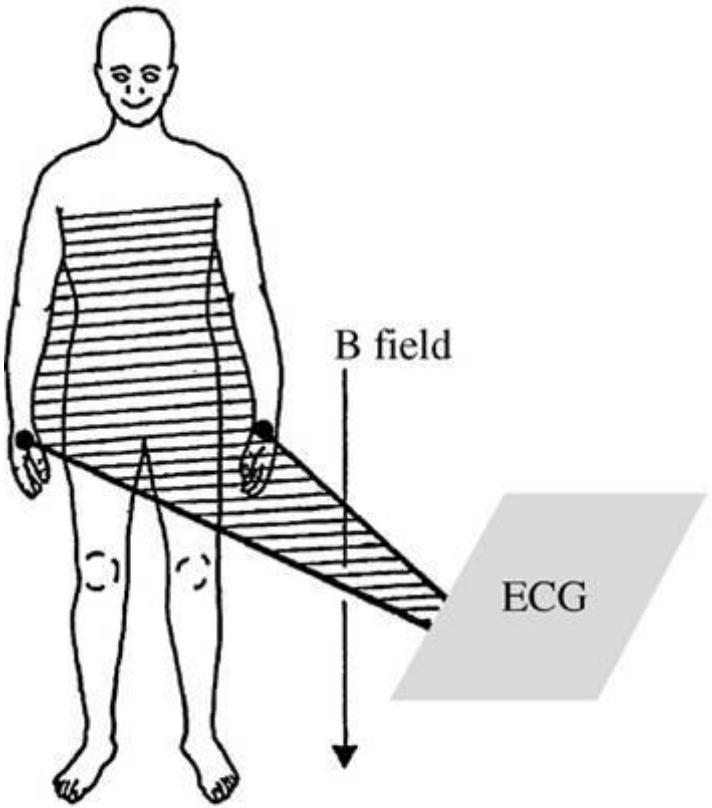
$$v_A - v_B = v_{cm} \left(\frac{Z_2 - Z_1}{Z_{in}} \right)$$

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

- This is noticeable compared to the 12 μV amplitude of the EEG signal.

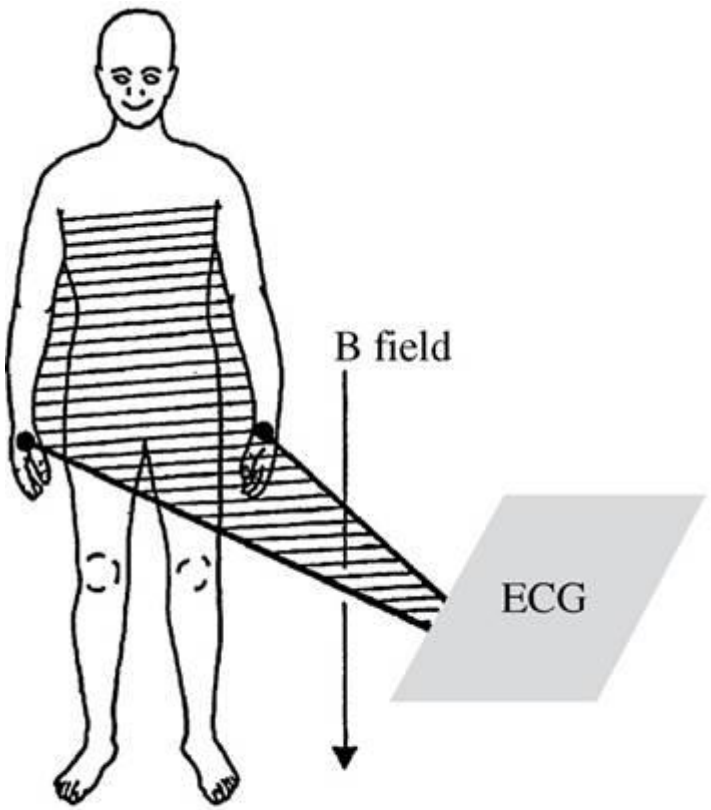
•Current in power lines establishes a *magnetic field* in the vicinity of the line.
•Magnetic fields can also sometimes originate from transformers and ballasts in fluorescent lights.
•If such magnetic fields pass through the effective single-turn coil produced by the electrocardiograph, lead wires, and the patient, voltage is induced in this loop.

•This voltage is proportional to the magnetic-field strength and the area of the effective single-turn coil.

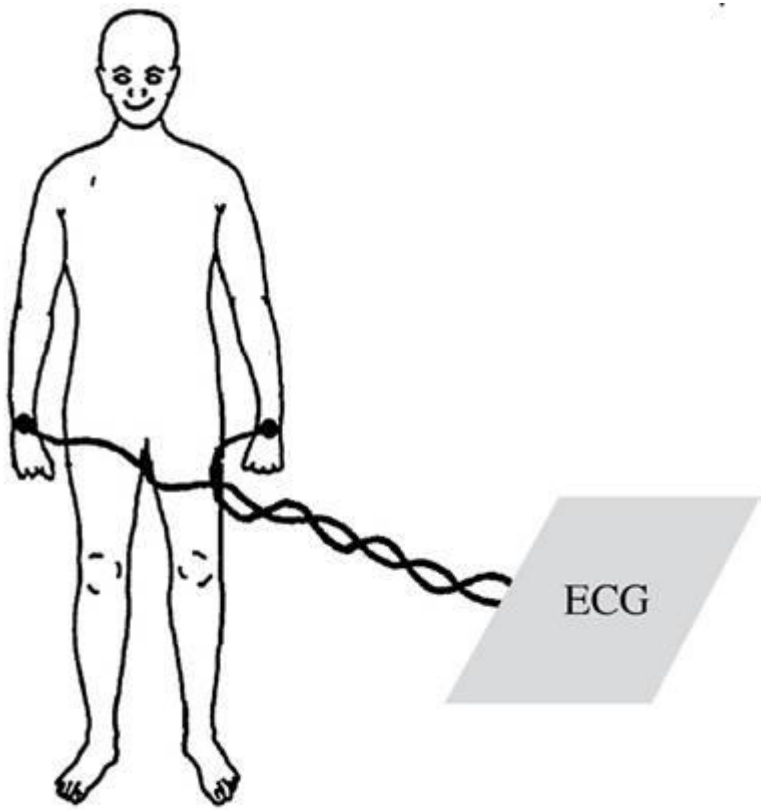


(a)

Induced voltage 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. by twisting the lead wires together and keeping them close to the body in order to subtend a much smaller area.



(a)



(b)

Transient Protection

- 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, 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.

•Figure shows the basic arrangement of such protective circuits. Two-terminal voltage-limiting devices are connected between each patient electrode and electric ground.
•After a certain voltage level these devices become short circuit and prevents high current flowing through ECG.

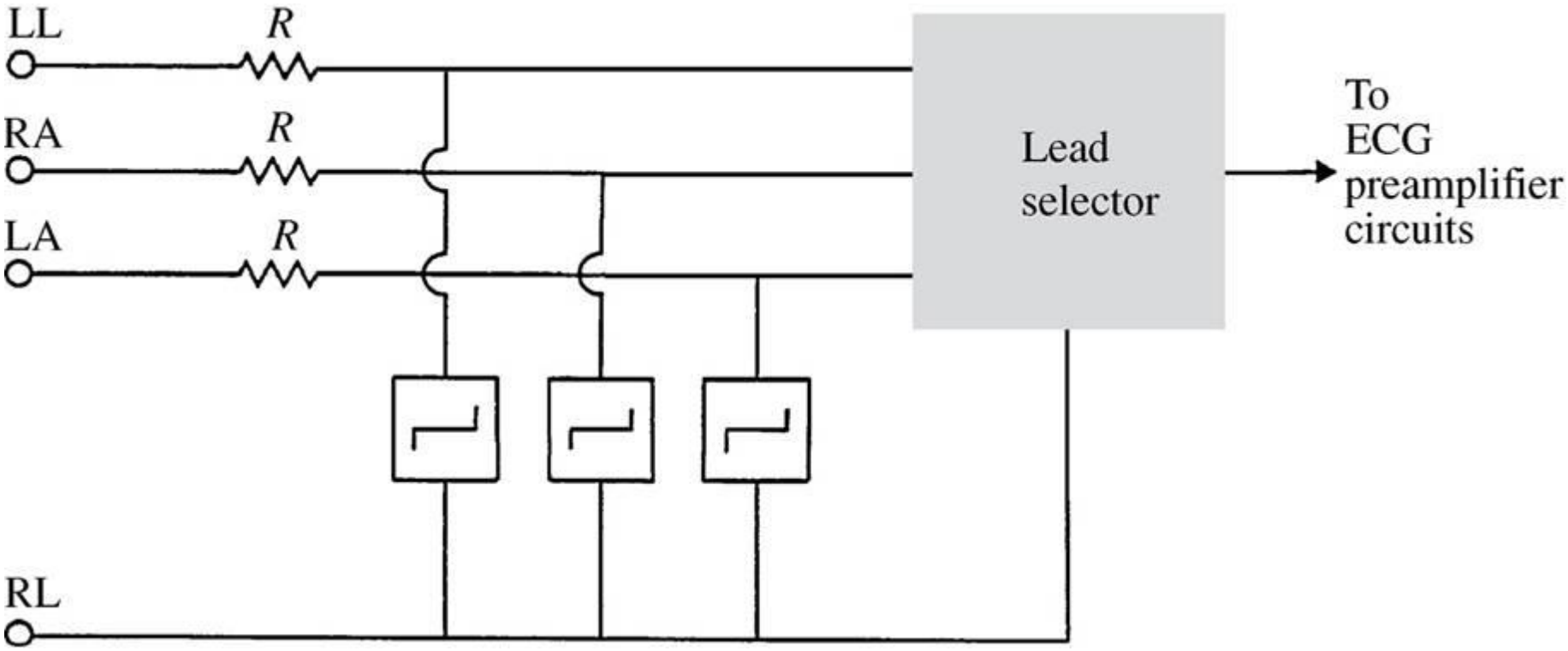


Figure 6.13 A voltage-protection scheme at the input of an electrocardiograph to protect the machine from high-voltage transients. Circuit elements connected across limb leads on left-hand side are voltage-limiting devices.

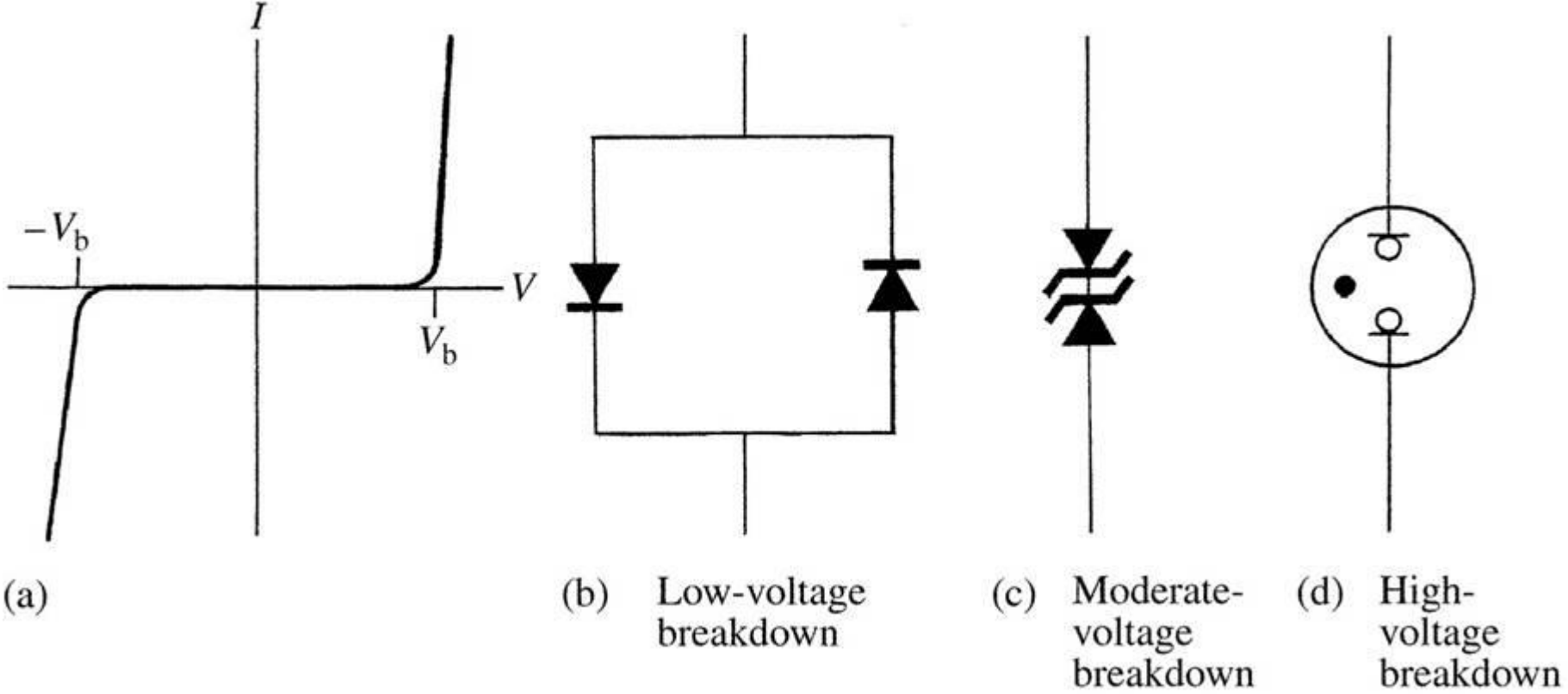
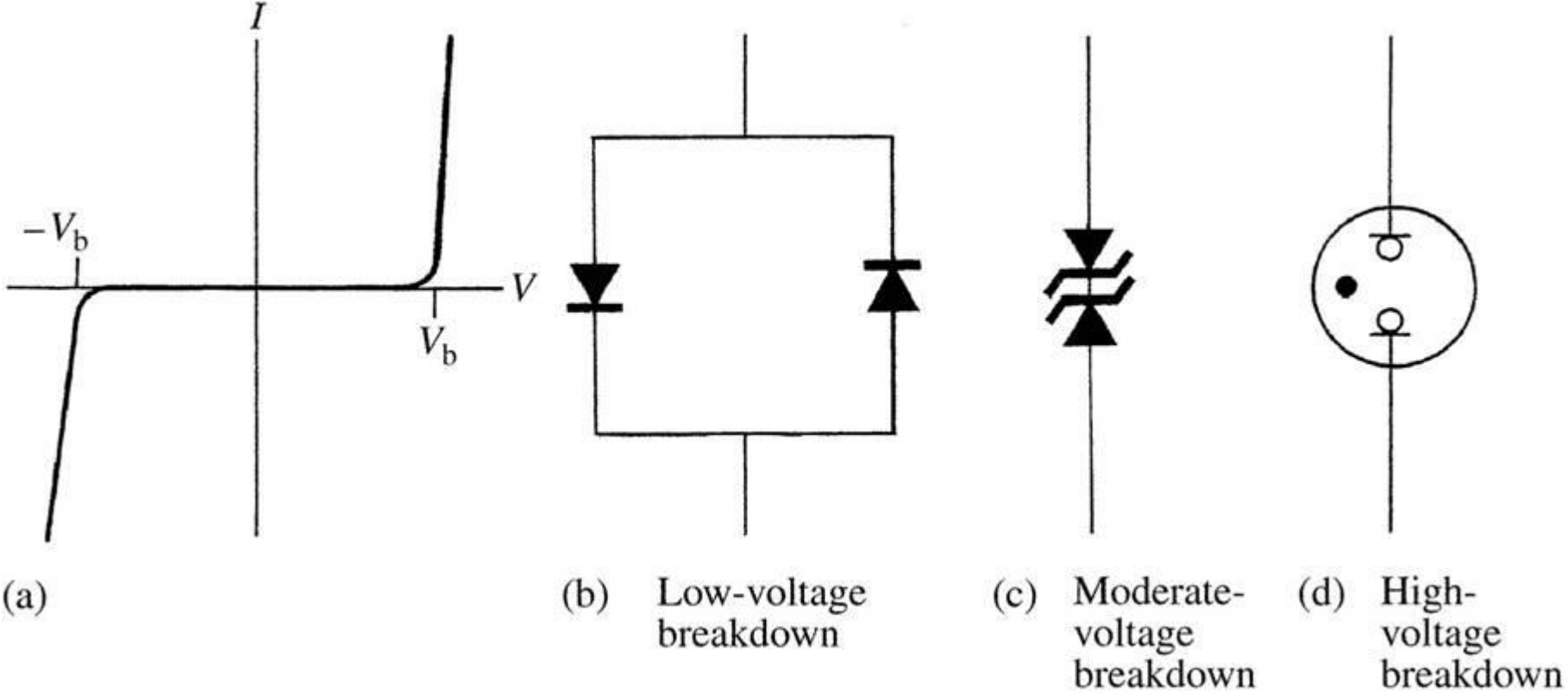


Figure 6.14 Voltage-limiting devices (a) Current–voltage characteristics of a voltage-limiting device, (b) Parallel silicon-diode voltage-limiting circuit, (c) Back-to-back silicon Zener-diode voltage-limiting circuit, (d) Gas-discharge tube (neon light) voltage-limiting circuit element.

•At voltages less than *Vb, 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 *Vb, the characteristics of the device sharply* change, and current passes through the device to such an extent that the voltage cannot exceed *Vb*

•Under these conditions, the device appears to behave as a short circuit in series with a constant-voltage source of magnitude Vb.



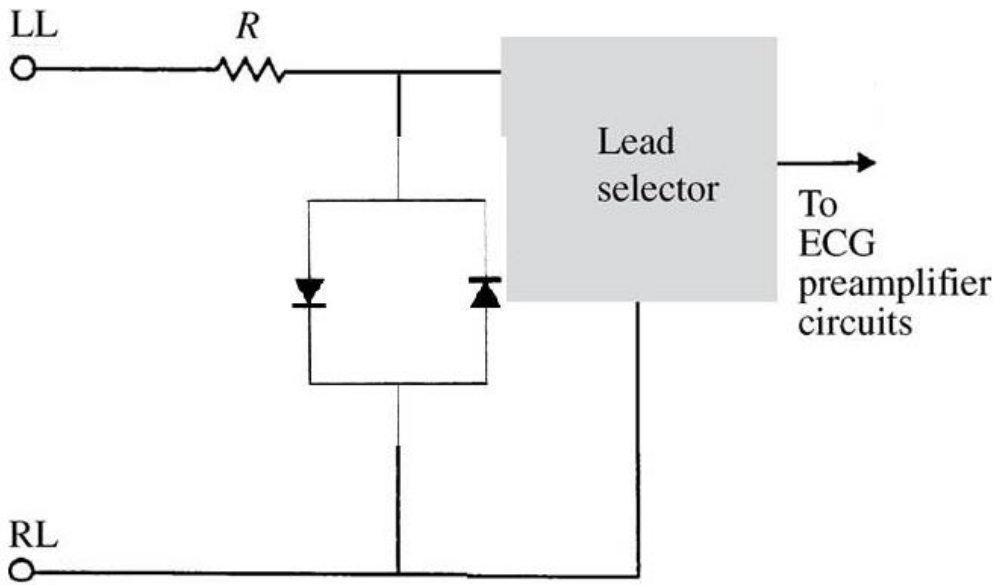
•**Parallel silicon diodes**, as shown in Figure 6.14(b), give a characteristic with a breakdown voltage of approximately 600 mV. the ECG itself does not approach such a voltage, it is possible under extreme conditions for dc-offset potentials.

•**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 in the range of 2 to 20 V.

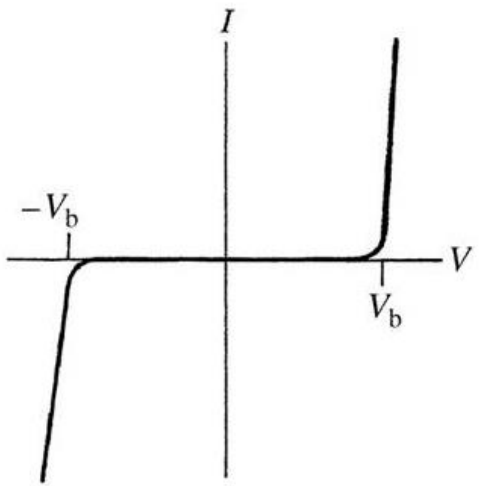
•**Gas-discharge tube** appears as an open circuit until it reaches its breakdown voltage. It then switches to the conducting state and maintains a voltage that is usually several volts less than the breakdown voltage. Breakdown voltages ranging from 50 to 90 V are typical for this device.

Problem

- Silicon diodes having a forward resistance of 2 Ω are to be used as voltage-limiting devices in the protection circuit of an electrocardiograph as shown in figure.
- If voltage transients as high as 500 V can appear at the electrocardiograph input during defibrillation, what is the minimal value of *R* that the designer can choose so that the voltage at the preamplifier input does not exceed 800 mV?
- Assume that the silicon diodes have a breakdown voltage Vb=600mV and I-V characteristics given below where kT/e=26mV and I_o=10^{-14}A



$$I = I_0 \exp(eV/kT)$$

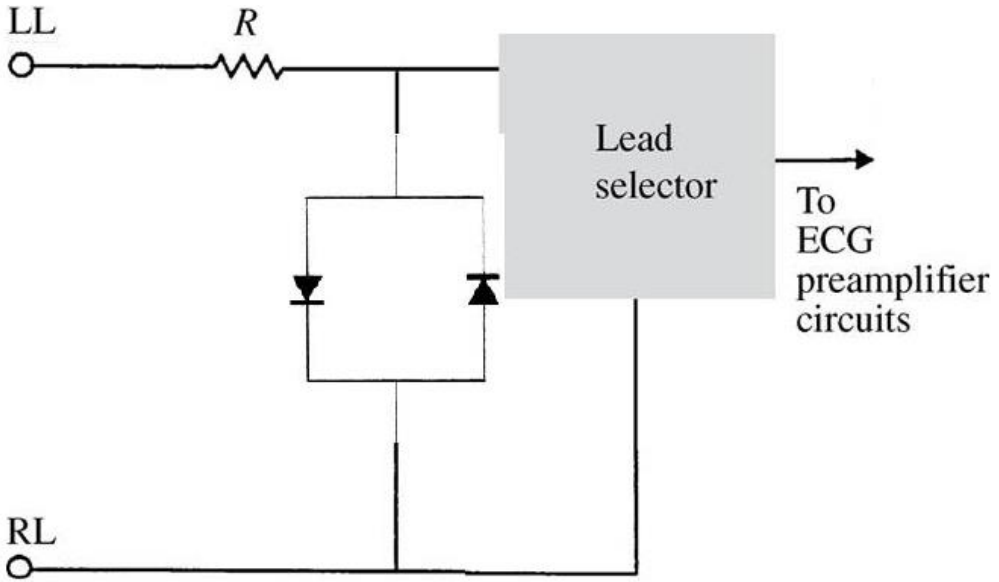


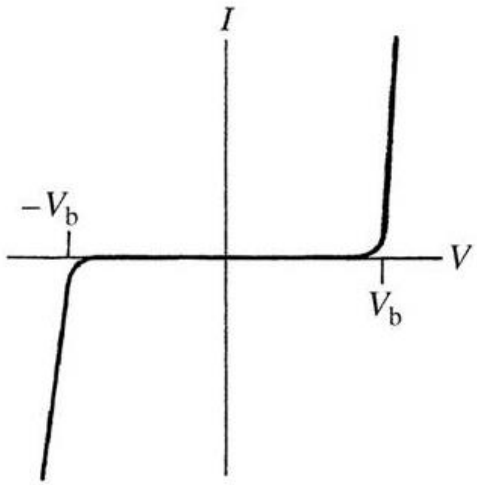
Solution

- If V=800mV then I=0.23A

$$I = I_0 \exp(eV/kT)$$

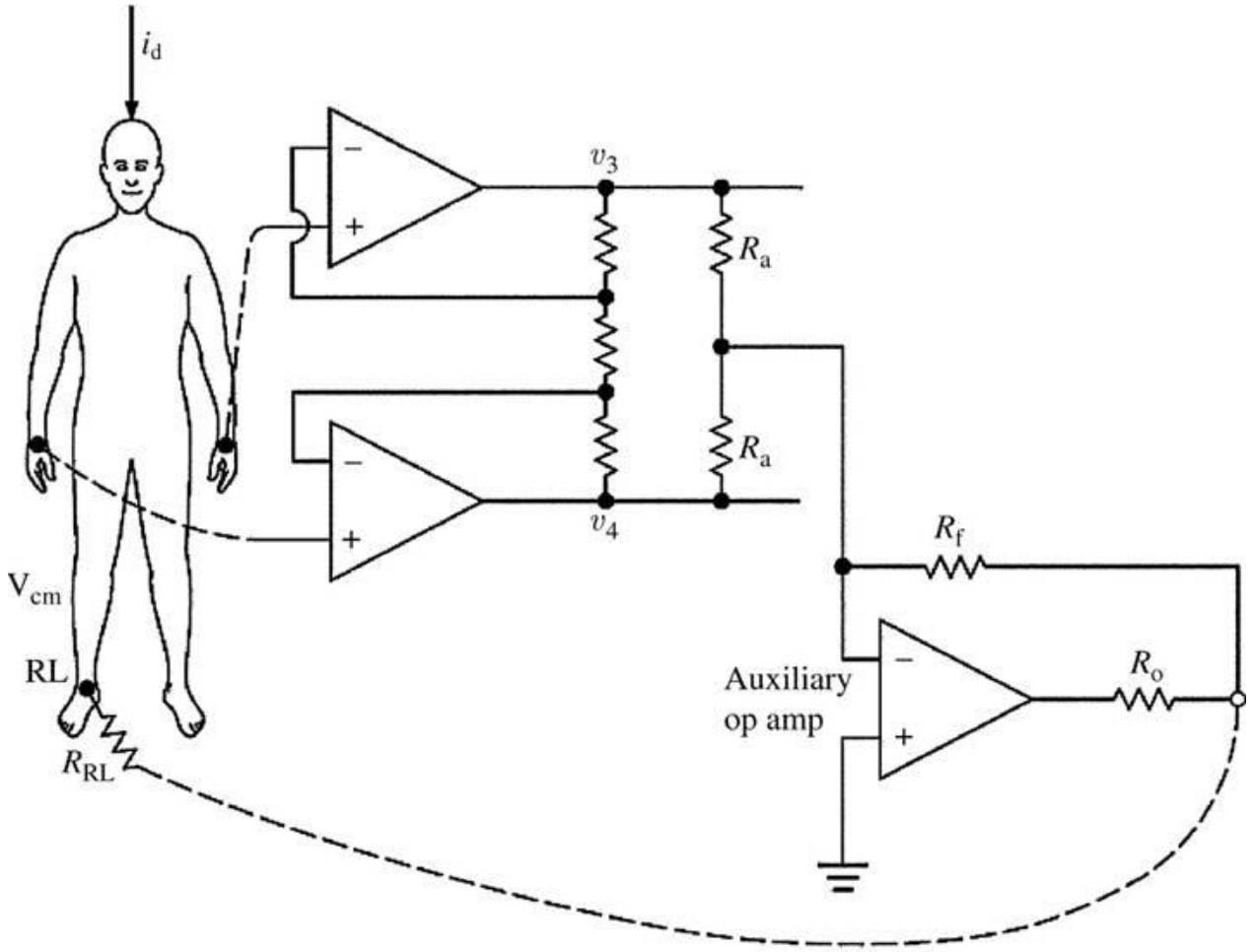
- where kT/e=26mV and I_o=10^{-14}A
- (500-0.8)/0.23=2170Ω





Driven-right-leg system

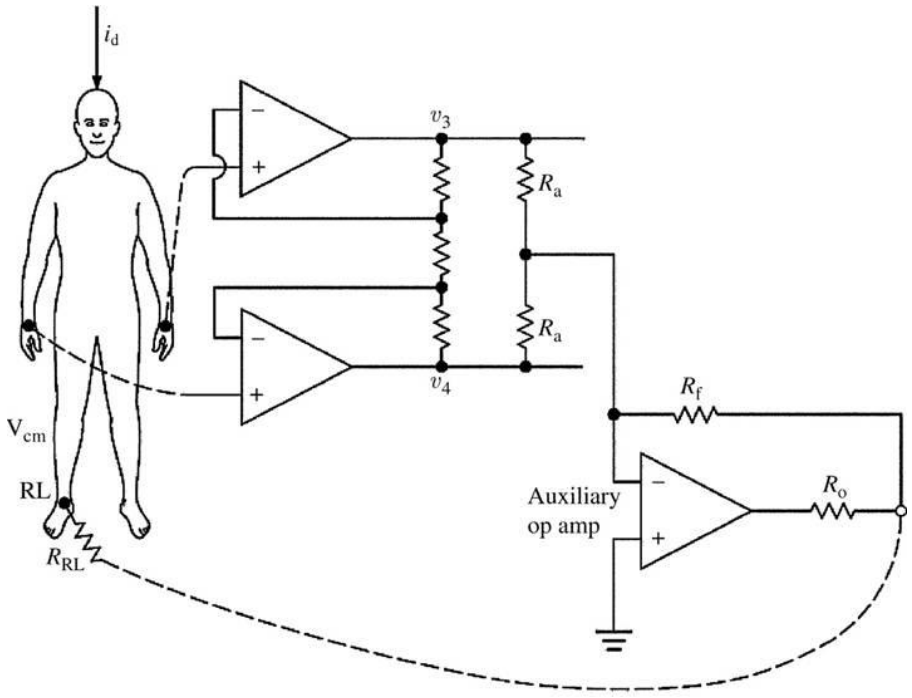
- In many modern electrocardiographic systems, the patient is not grounded at all. Instead, the right-leg electrode is to the output of an auxiliary op amp.



Driven-right-leg system

- 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 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



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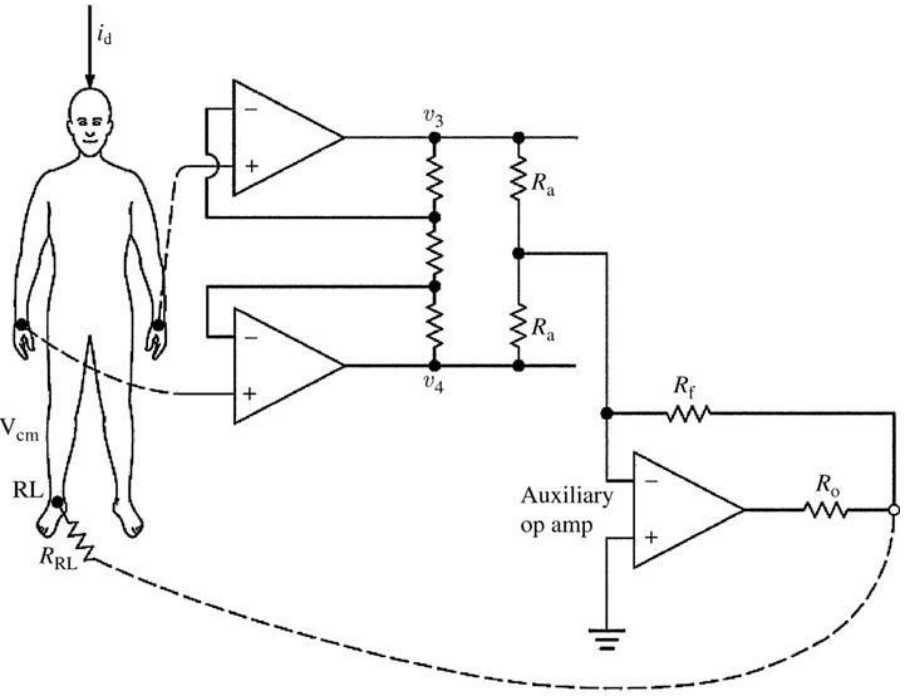
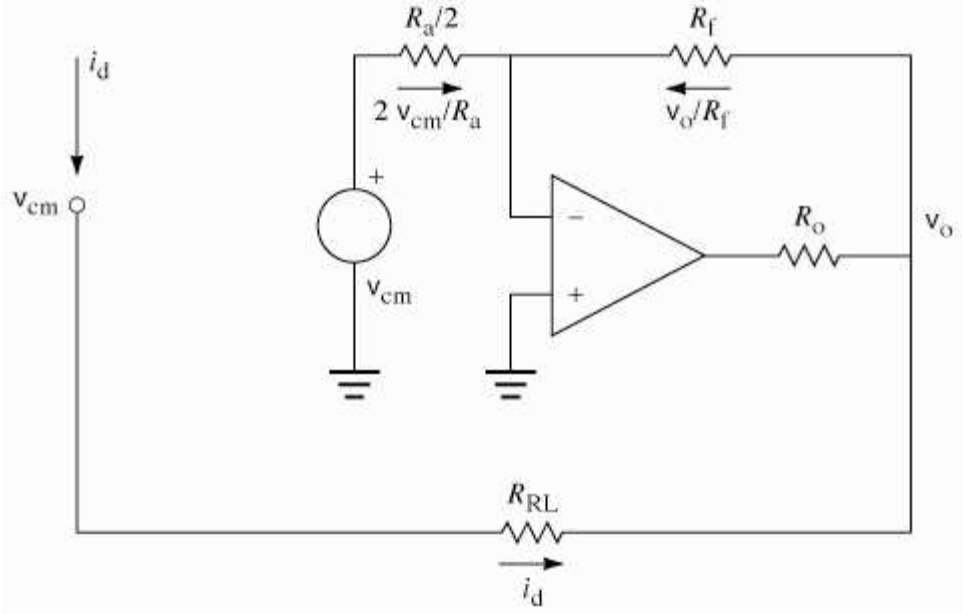


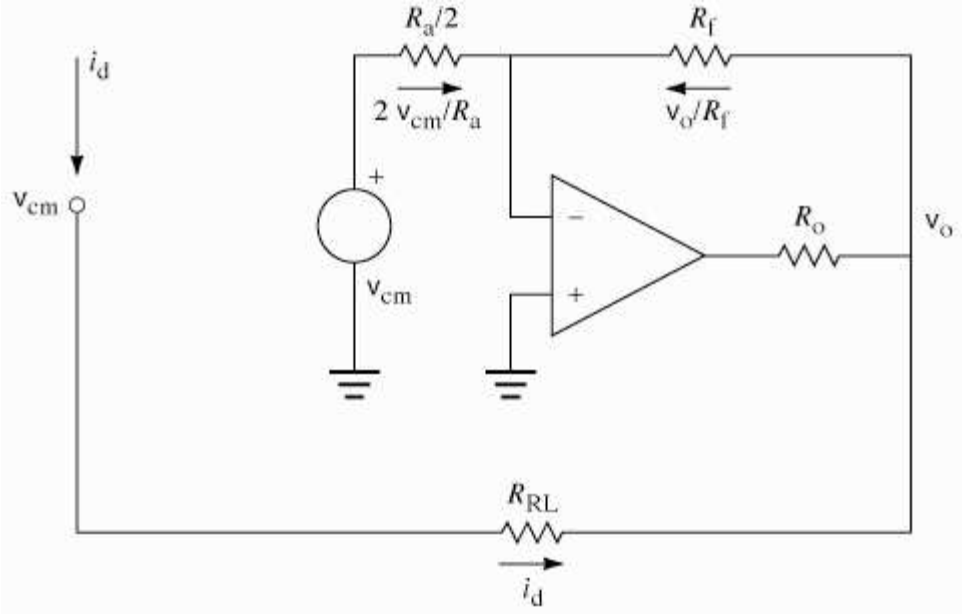


$$\frac{2v_{cm}}{R_a}+\frac{v_o}{R_f}=0 \qquad v_o=-\frac{2R_f}{R_a}v_{cm}$$

$$v_{cm}=R_{RL}i_d+v_o \qquad v_{cm}=\frac{R_{RL}i_d}{1+2R_f/R_a}$$

Driven-right-leg system

- When the amplifier saturates, as would occur during a large transient Vcm, its output appears as the saturation voltage Vs. The right leg is now connected to ground through this source and the parallel resistances Rf and Ro.
- To limit the current, Rf and Ro should be large. Values as high as 5 MΩ are used.

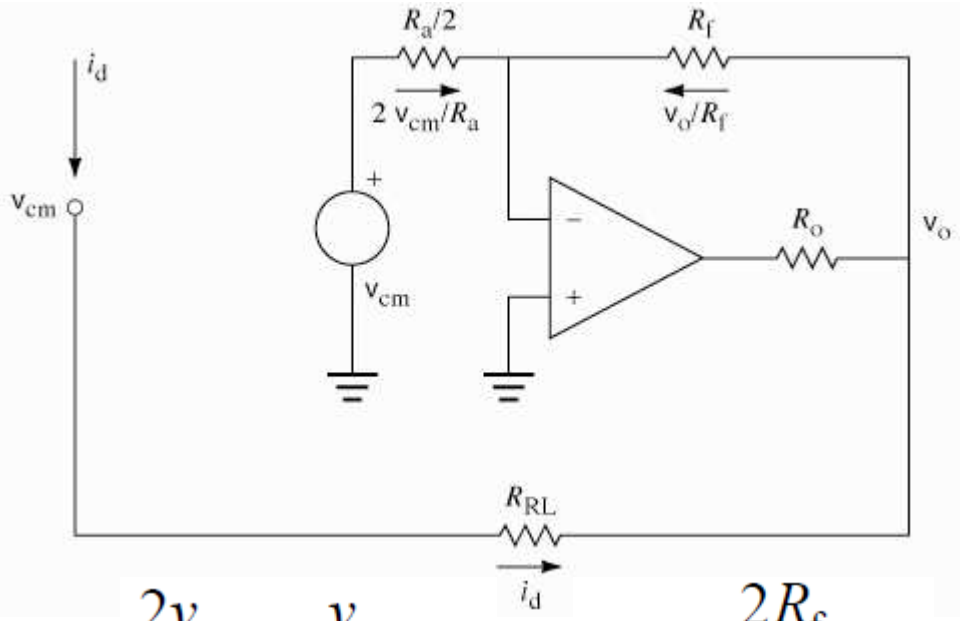


$$\frac{2v_{cm}}{R_a}+\frac{v_o}{R_f}=0 \qquad v_o=-\frac{2R_f}{R_a}v_{cm}$$

$$v_{cm}=R_{RL}i_d+v_o \qquad v_{cm}=\frac{R_{RL}i_d}{1+2R_f/R_a}$$

Driven-right-leg system

- When the amplifier is not saturated, we would like Vcm 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 Rf large and Ra relatively small. Rf can be equal to Ro, but Ra can be much smaller.
- A typical value of Ra 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
- For the 0.2 mA displacement current, the common-mode voltage is



$$\frac{2v_{cm}}{R_a}+\frac{v_o}{R_f}=0 \qquad v_o=-\frac{2R_f}{R_a}v_{cm}$$

$$v_{cm}=R_{RL}i_d+v_o \qquad v_{cm}=\frac{R_{RL}i_d}{1+2R_f/R_a}$$

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

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

Problem

- A driven right leg amplifier has R_a = 20 kΩ, R_f = 200 kΩ, R_{RL} = 1 MΩ, R_o = 200 kΩ, i_d = 0.5 μA.
- (a) Calculate V_{cm}.
- (b) Estimate risk current flowing through patient when the patient touches 120 V (assume 120 V short circuits all op amp inputs and outputs and all skin resistances).
- (c) Is this risk current acceptable (refer to Table 6.1)?

Answer

- A driven right leg amplifier has R_a = 20 kΩ, R_f = 200 kΩ, R_{RL} = 1 MΩ, R_o = 200 kΩ, i_d = 0.5 μA.
- (a) $$v_{cm} = \frac{R_{RL} i_d}{1 + 2R_f / R_a}$$
- (b) Rf and R0 become parallel resistors connecting to the ground.

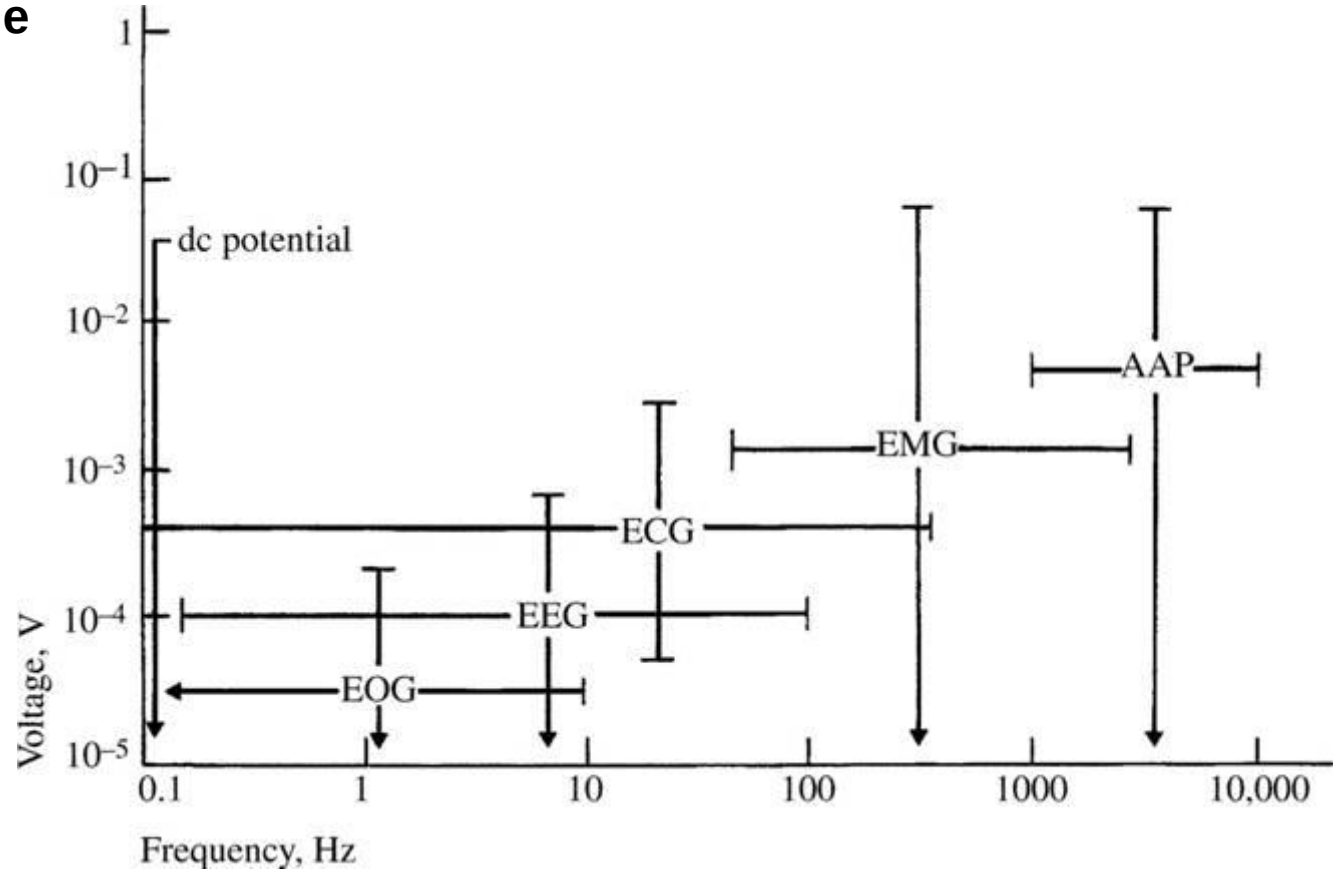
Table 6.1 Summary of Performance Requirements for Electrocardiographs (Anonymous, 1991)

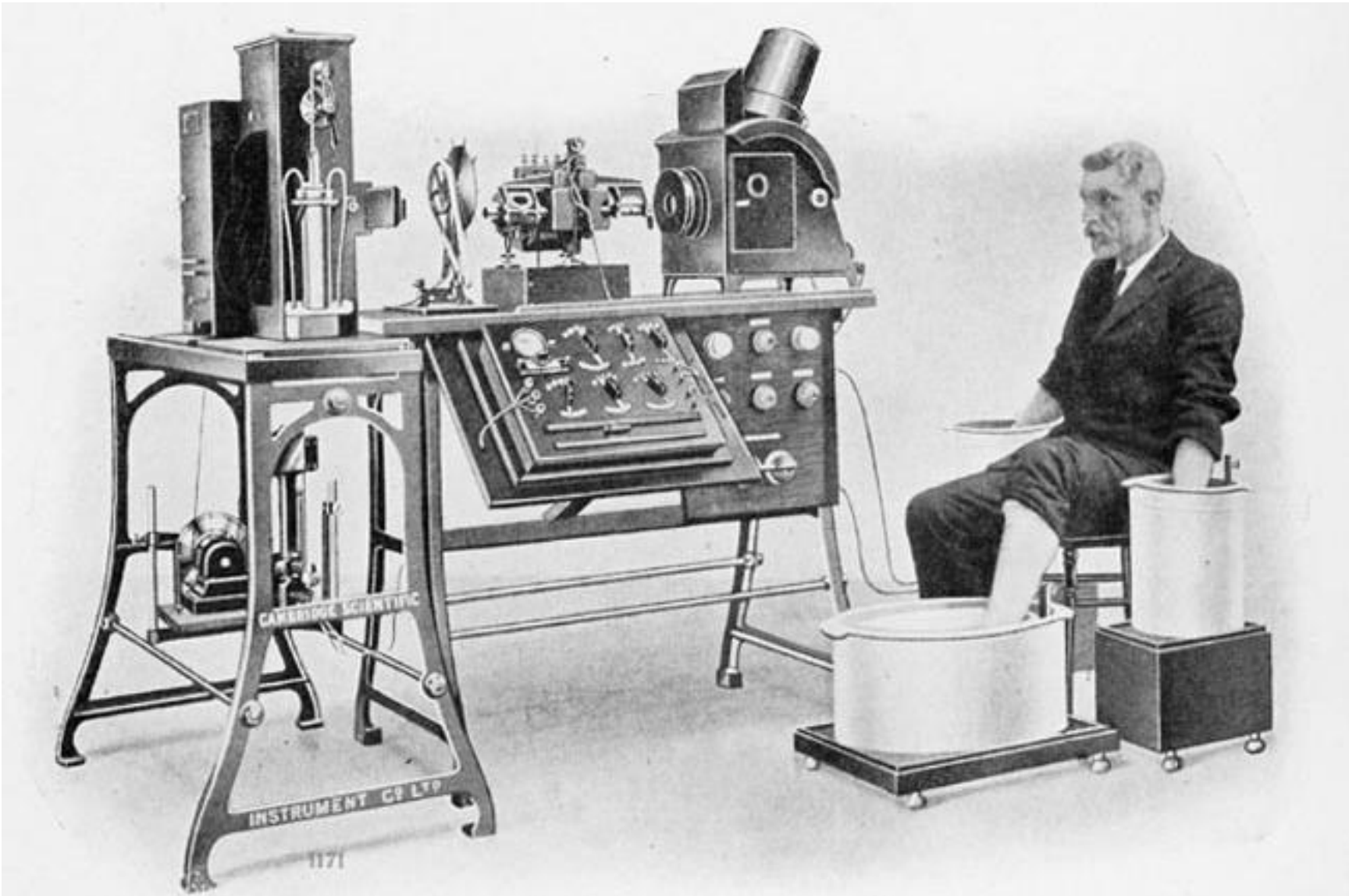
3.2.15	Risk current (isolated patient connection)	Max	μA	10
		As per applicable document 2.11		

AMPLIFIERS FOR OTHER BIOPOTENTIAL SIGNALS

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 frequency response constraints on the amplifiers used.

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.





PHOTOGRAPH OF A COMPLETE ELECTROCARDIOGRAPH, SHOWING THE MANNER IN WHICH THE ELECTRODES ARE ATTACHED TO THE PATIENT, IN THIS CASE THE HANDS AND ONE FOOT BEING IMMERSED IN JARS OF SALT SOLUTION

https://en.wikipedia.org/wiki/Willem_Einthoven#/media/

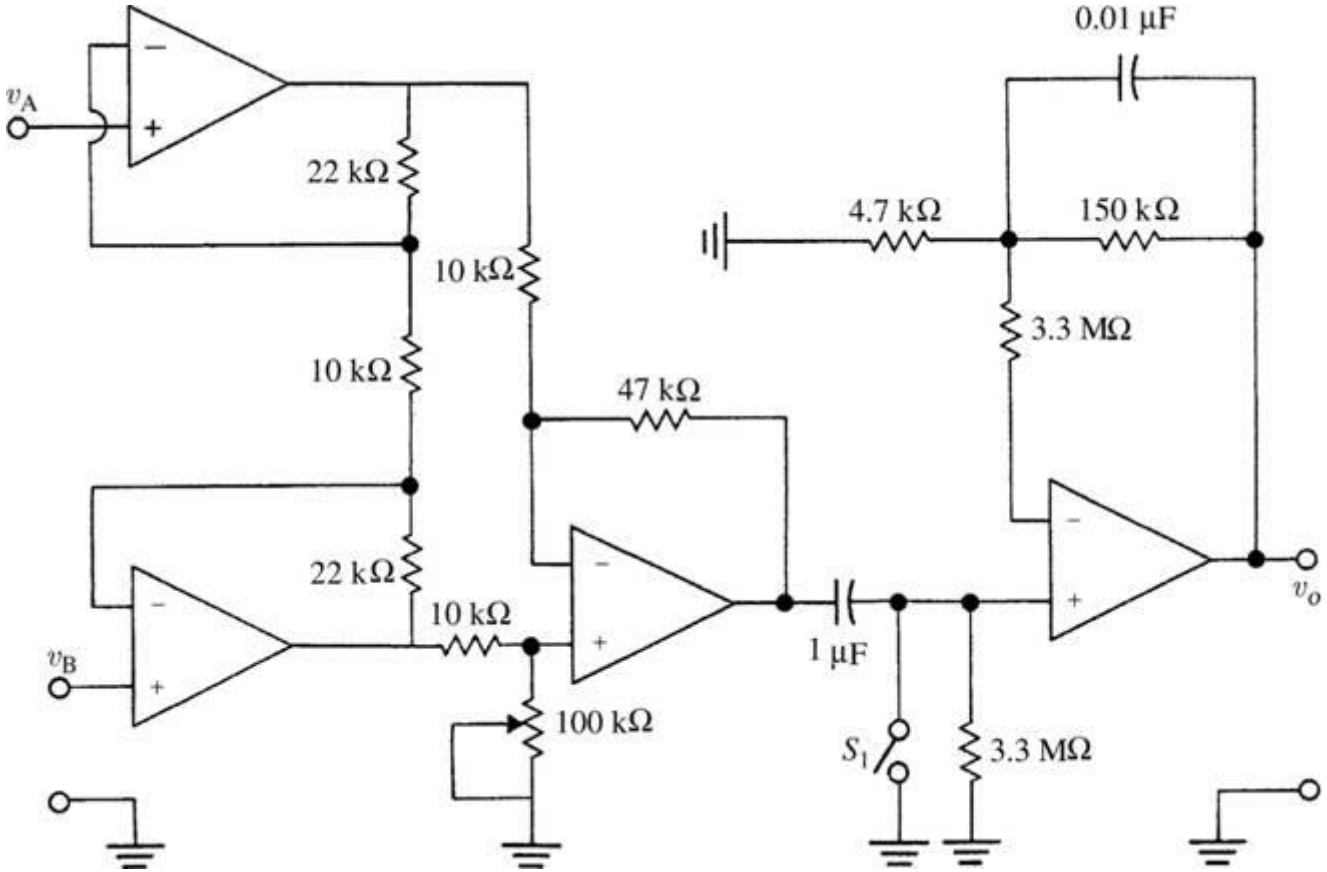


Figure 6.18 This ECG amplifier has a gain of 25 in the dc-coupled stages. The high-pass filter feeds a noninverting-amplifier stage that has a gain of 32. The total gain is 25 X 32 = 800. When mA 776 op amps were used, the circuit was found to have a CMRR of 86 dB at 100 Hz and a noise level of 40 mV peak to peak at the output. The frequency response was 0.04 to 150 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.

EMG AMPLIFIER

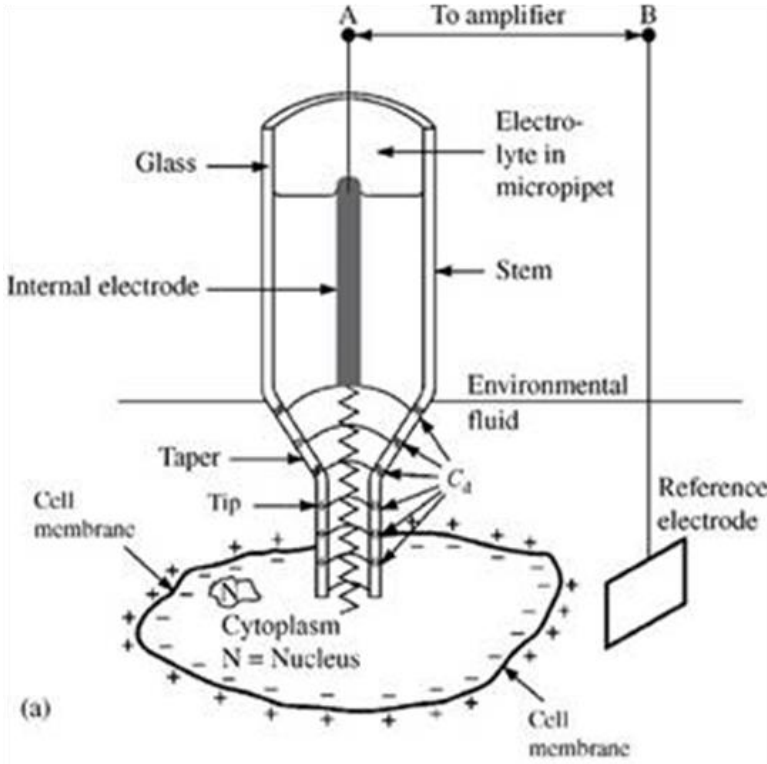
- Electromyographic signals range in frequency from 25 Hz to several kilohertz. Signal amplitudes range from 100 mV to 90 mV, depending on the type of signal and electrodes used.
- Thus EMG amplifiers must have a wider frequency response than ECG amplifiers, but they do not have to cover so low a frequency 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 5000Ω, 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.

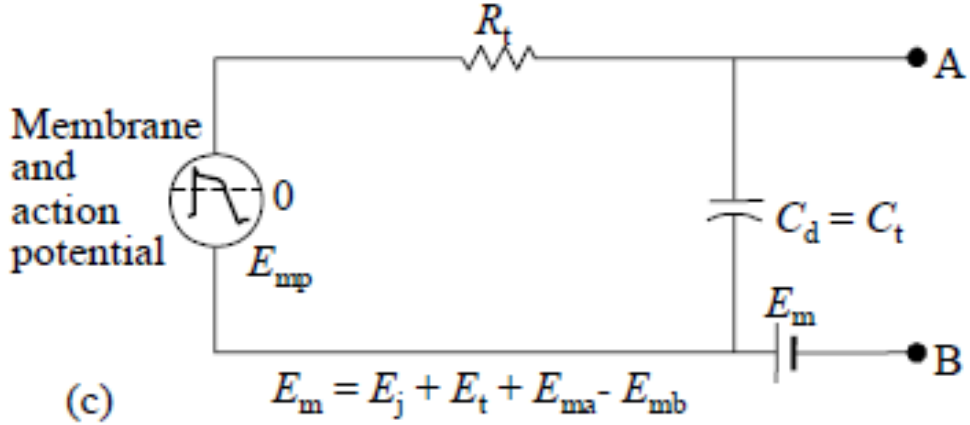
AMPLIFIERS FOR USE WITH GLASS MICROPIPET INTRACELLULAR ELECTRODES

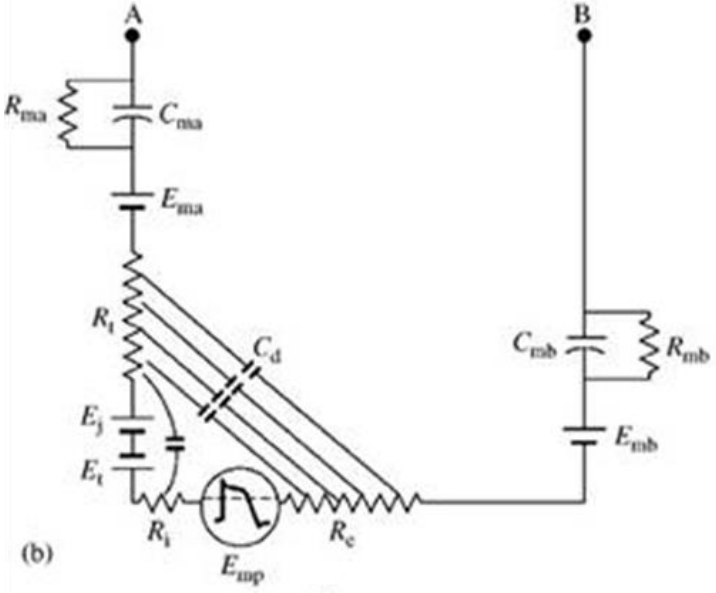
- Intracellular electrodes or microelectrodes that can measure the potential across the cell membrane generally detect potentials on the order of 50 to 100 mV.
- Their small size and small effective surface-contact area give them a **very high source impedance** and their geometry results in a relatively **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.

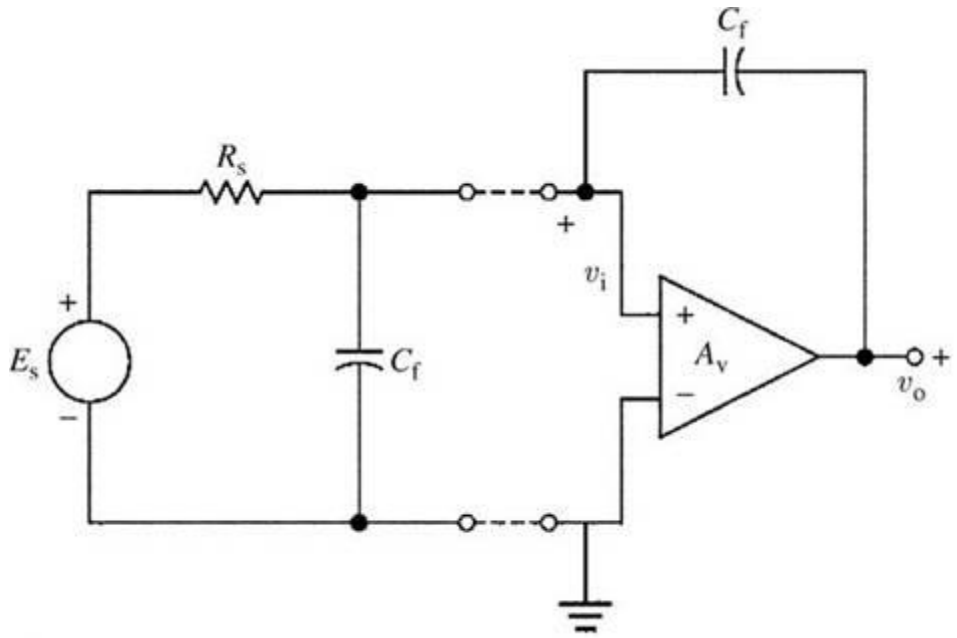
The equivalent Circuit for Micropipette Electrode

(a) Electrode with its tip placed within
a cell, showing the origin of distributed capacitance.
(b) Equivalent circuit for the situation in (a).
(c) Simplified equivalent circuit.









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

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

The total circuit capacitance is $C = C_s + (1 - A_v)C_f$

which is zero when $C_s = (A_v - 1)C_f$

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.

EEG AMPLIFIERS

- 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 common-mode-rejection ratios and low noise.

Example

- 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 common mode rejection ratio (CMRR) of their electroencephalograph be so that an EEG signal of 25 μV amplitude has no more than 1% common mode noise?

ANSWER

- The signal-to-noise ratio (SNR) at the amplifier input can be as low as

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

- The signal to noise ratio at the output or display of the electroencephalograph must be at least.

$$\text{SNR} = (1\%)^{-1} = 100$$

- The common mode rejection ratio then must be the ratio of the output signal to noise ratio to that at the input

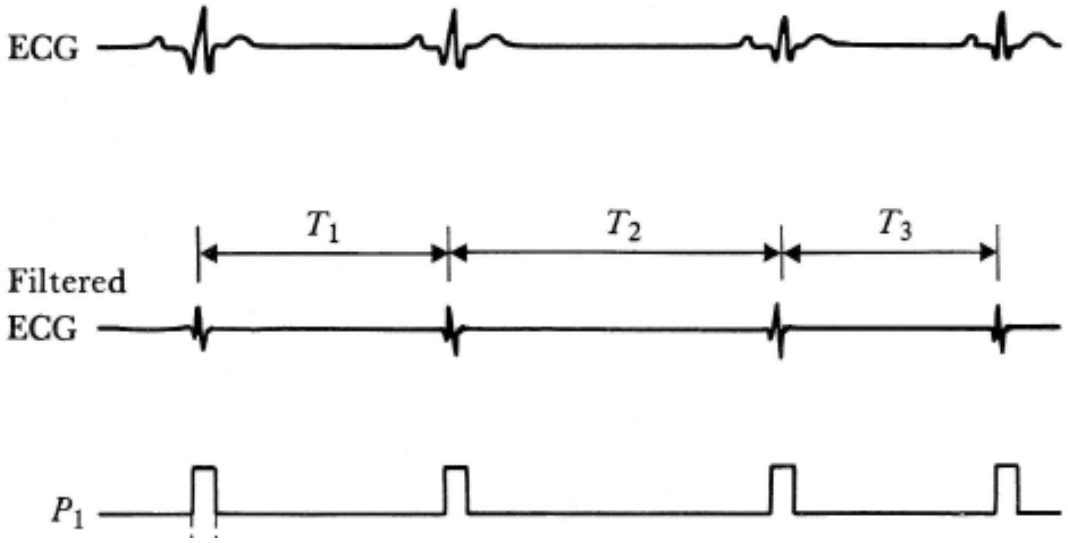
$$\text{CMRR} = \frac{100}{2.5 \times 10^{-4}} = 4 \times 10^{5} \quad \text{or} \quad 20 \log_{10}(4 \times 10^{5})\ \text{dB} = 112\ \text{dB}$$

- This is within the range of CMRR available in high-quality differential amplifiers.

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.
- 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.

- 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 P1.



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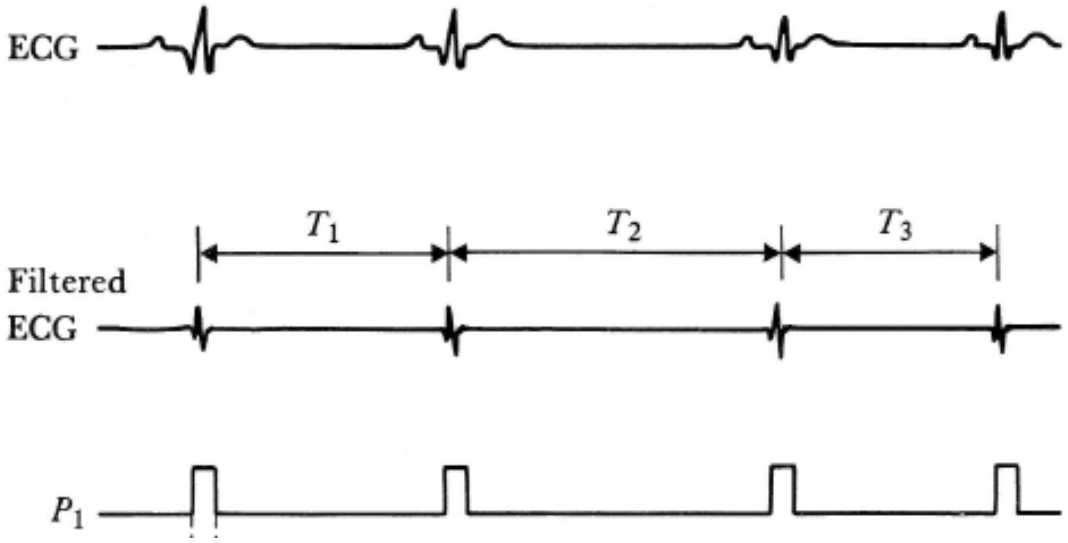


- A threshold detector triggers the pulse P1.
- A 1 kHz clock signal enters a counting register whenever P3 is high. Because P3 is high during the interval between QRS complexes, the 1 ms pulses coming from the clock (P4) accumulate in register 1 during this period otherwise nothing is recorded.
- Software calculates v_o using

$$v_{\mathrm{o}} = \frac{k}{T_{\mathrm{R}}}$$

- where k is a constant and TR is the interval between QRS complexes.

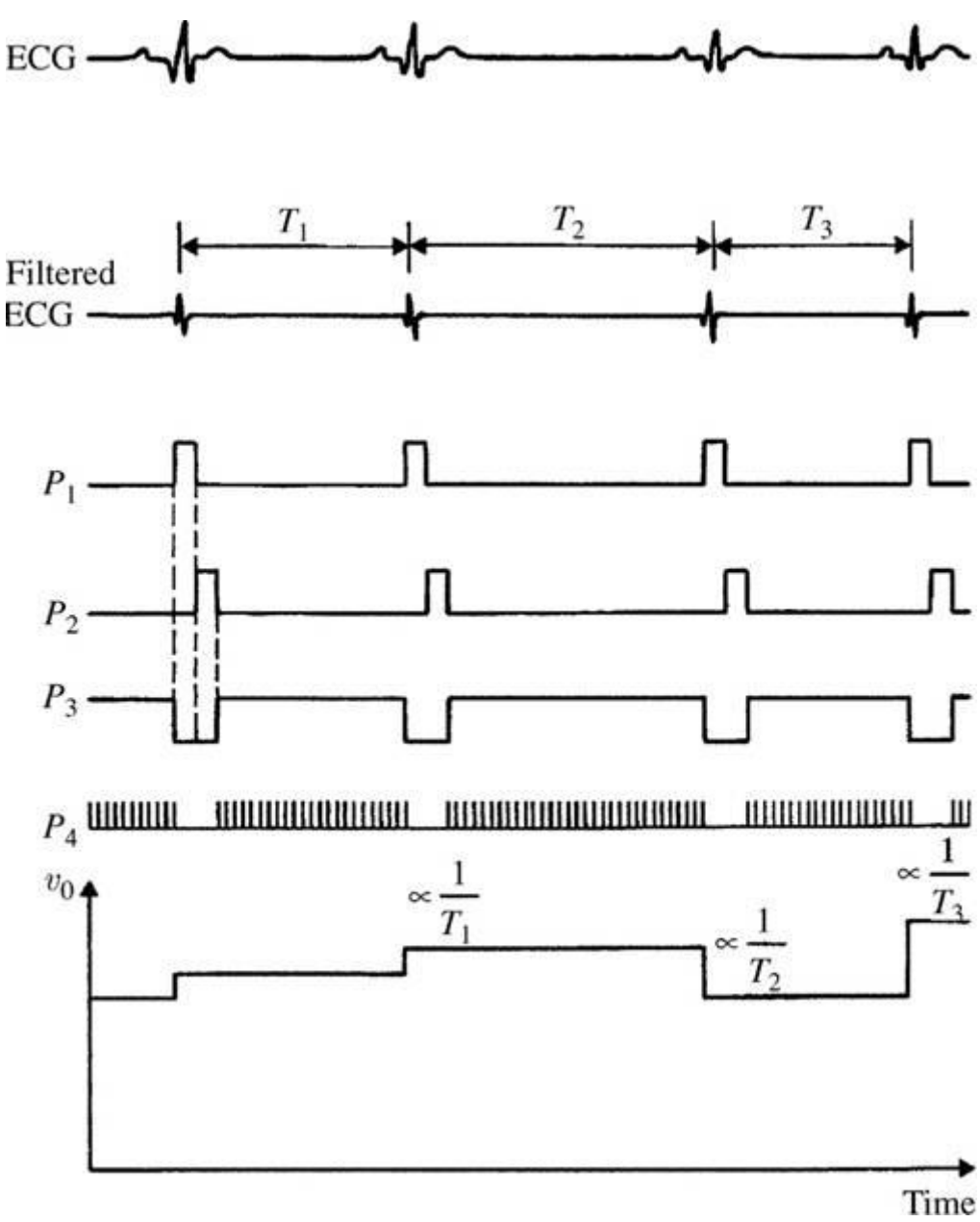


Figure 6.19 Timing diagram for beat-to-beat cardiotachometer to determine heart rate.

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 v1, is fed to software, which in one example takes the absolute value, resulting v2.
- Software then integrates the signal. Once the integrator output has exceeded a preset threshold level vt, a comparator then reinitiates integration of the EMG until the cycle repeats itself.

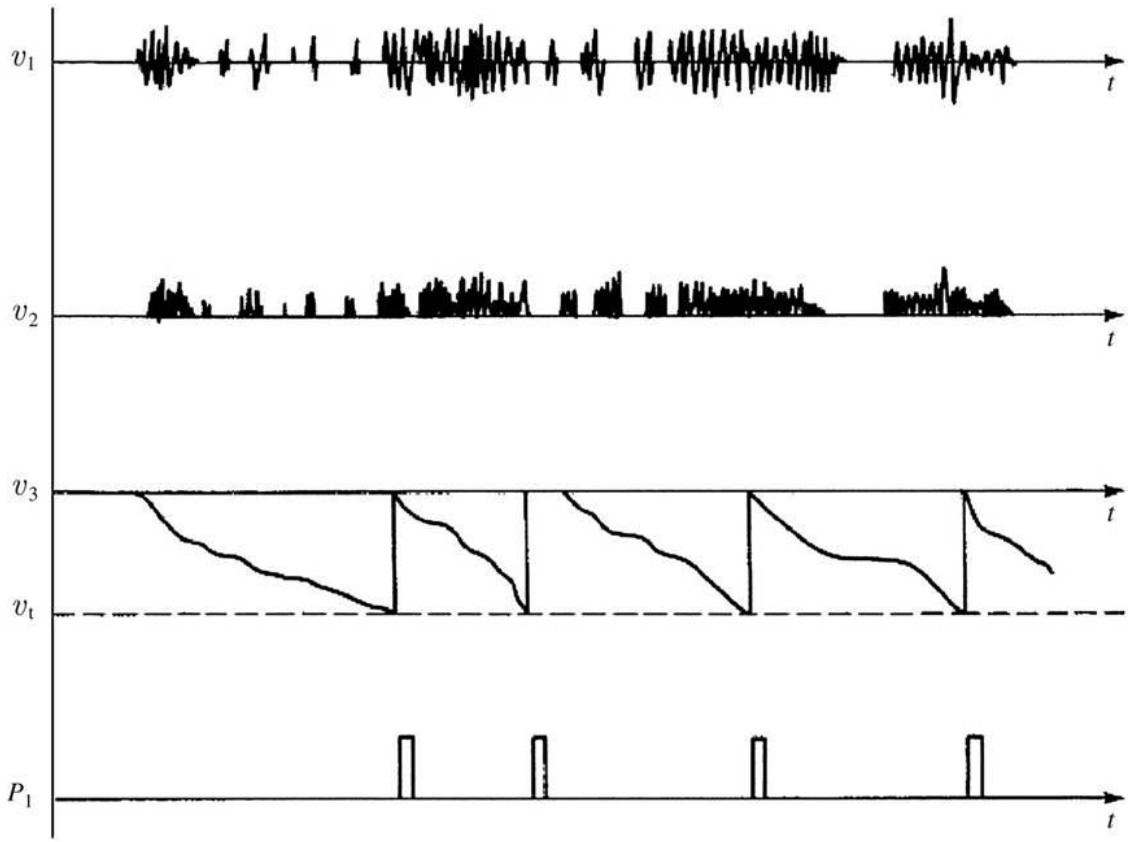


Figure 6.20 The various waveforms for the EMG integrator.

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).
- 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 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

$$n / \sqrt{n} = \sqrt{n}$$

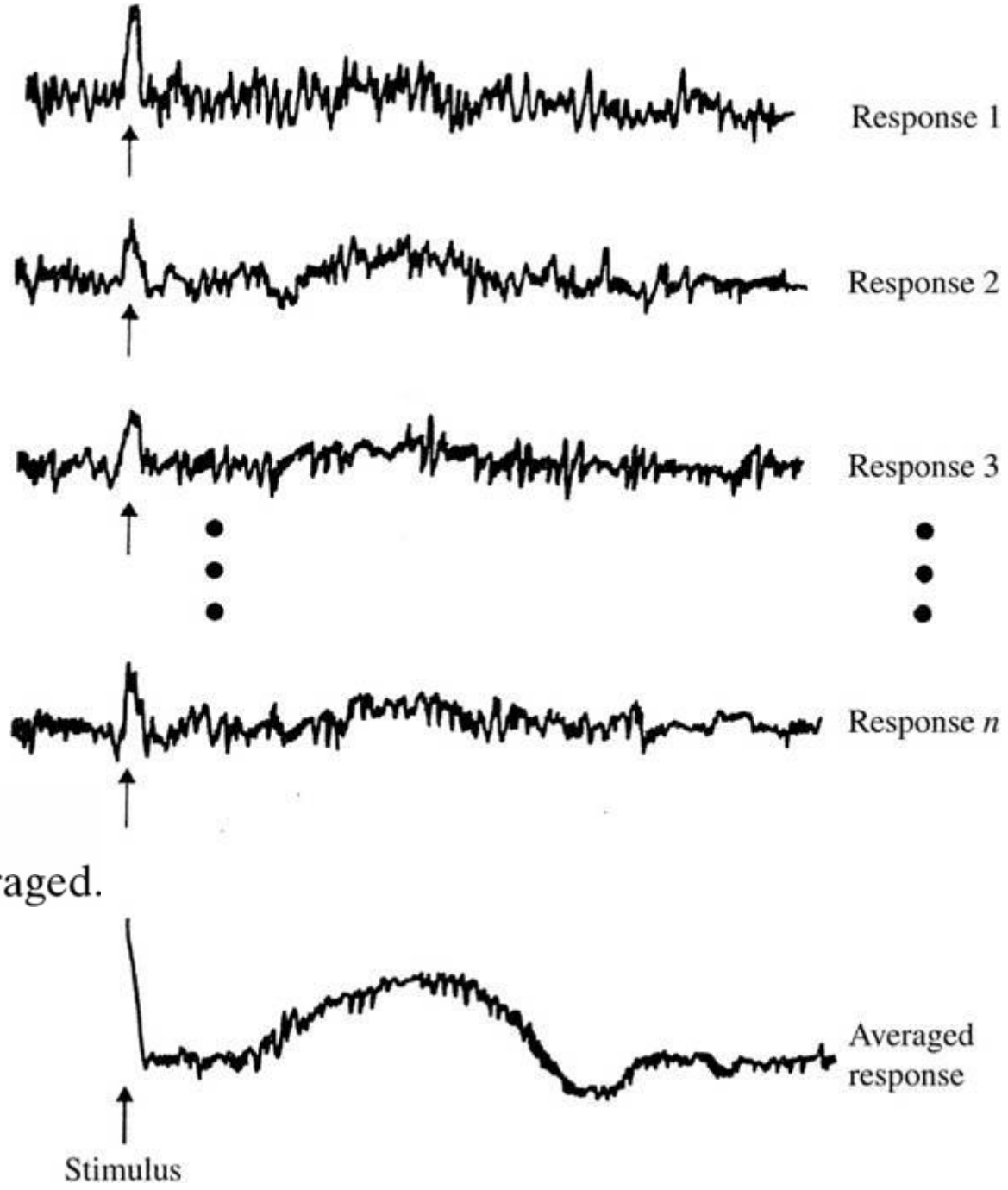


Figure 6.21 Signal-averaging technique for improving the SNR in signals that are repetitive or respond to a known stimulus.

$$10 = \sqrt{n}$$

$$n = (10)^2 = 100 \text{ samples averaged.}$$

EXAMPLE

- 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:

$$10 = \sqrt{n}$$

$$n = (10)^2 = 100 \text{ samples averaged.}$$

For a 100-fold improvement we need

$$100 = \sqrt{n}$$

$$(100)^2 = n = 10{,}000 \text{ samples averaged}$$

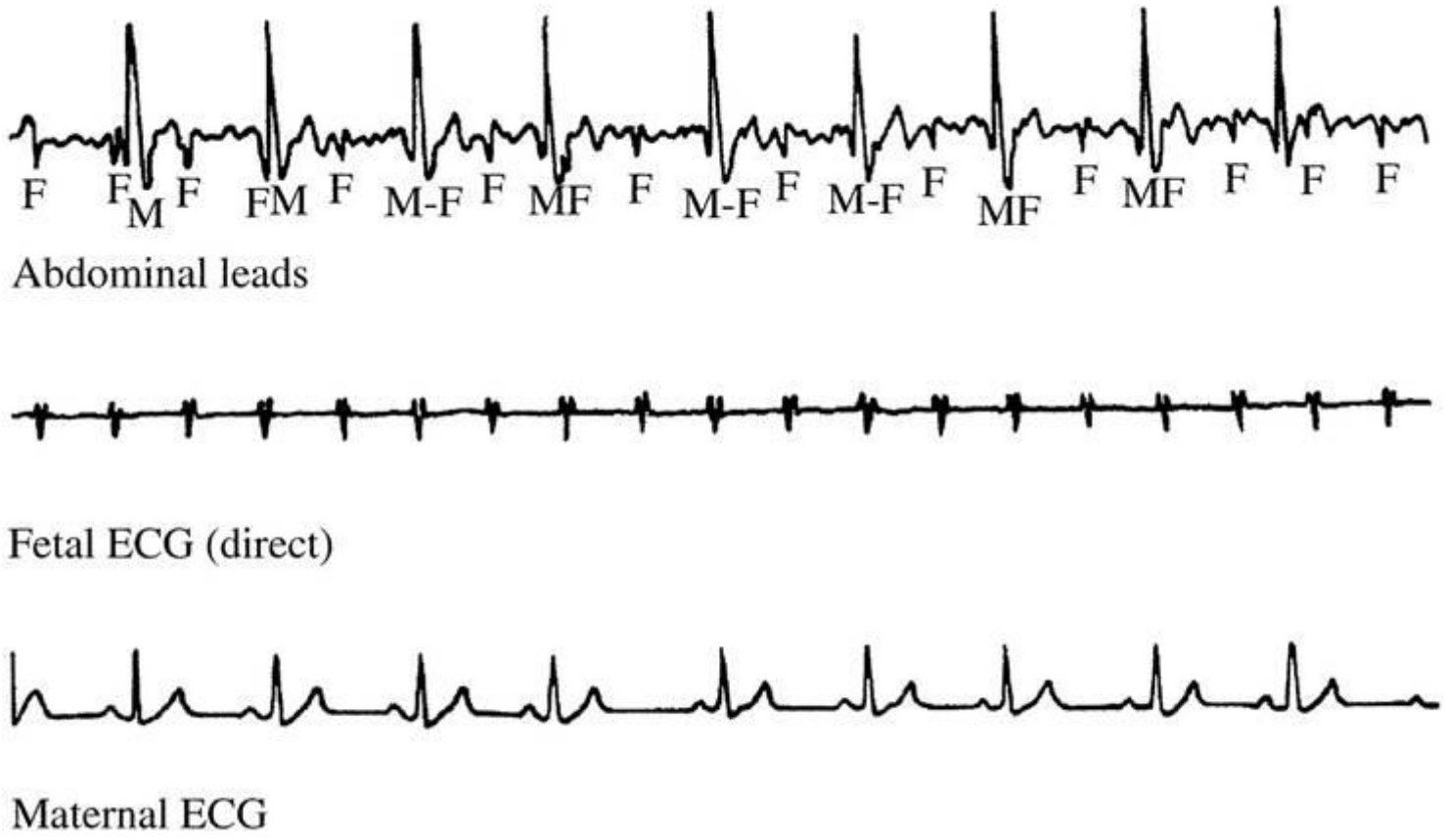


Figure 6.22 Typical fetal ECG obtained from the maternal abdomen. F represents fetal QRS complexes; M represents maternal QRS complexes. Maternal ECG and fetal ECG (recorded directly from the fetus) are included for comparison. (From "Monitoring of Intrapartum Phenomena," by J. F. Roux, M. R. Neuman,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

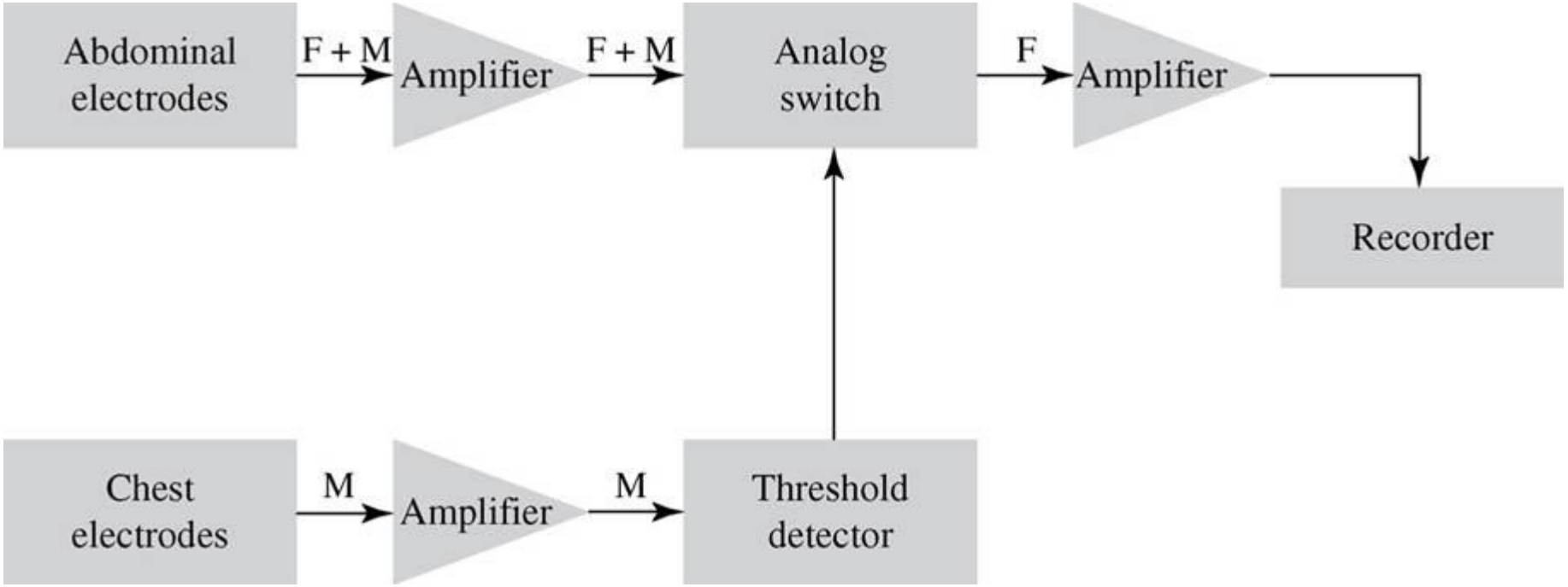


Figure 6.23 Block diagram of a scheme for isolating fetal ECG from an abdominal signal that contains both fetal and maternal ECGs. (From "Monitoring of Intrapartum Phenomena," by J. F. Roux, M. R. Neuman, and R. C. Goodlin, in *CRC Critical Reviews in Bioengineering,* 2, pp. 119–158, January 1975, ©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

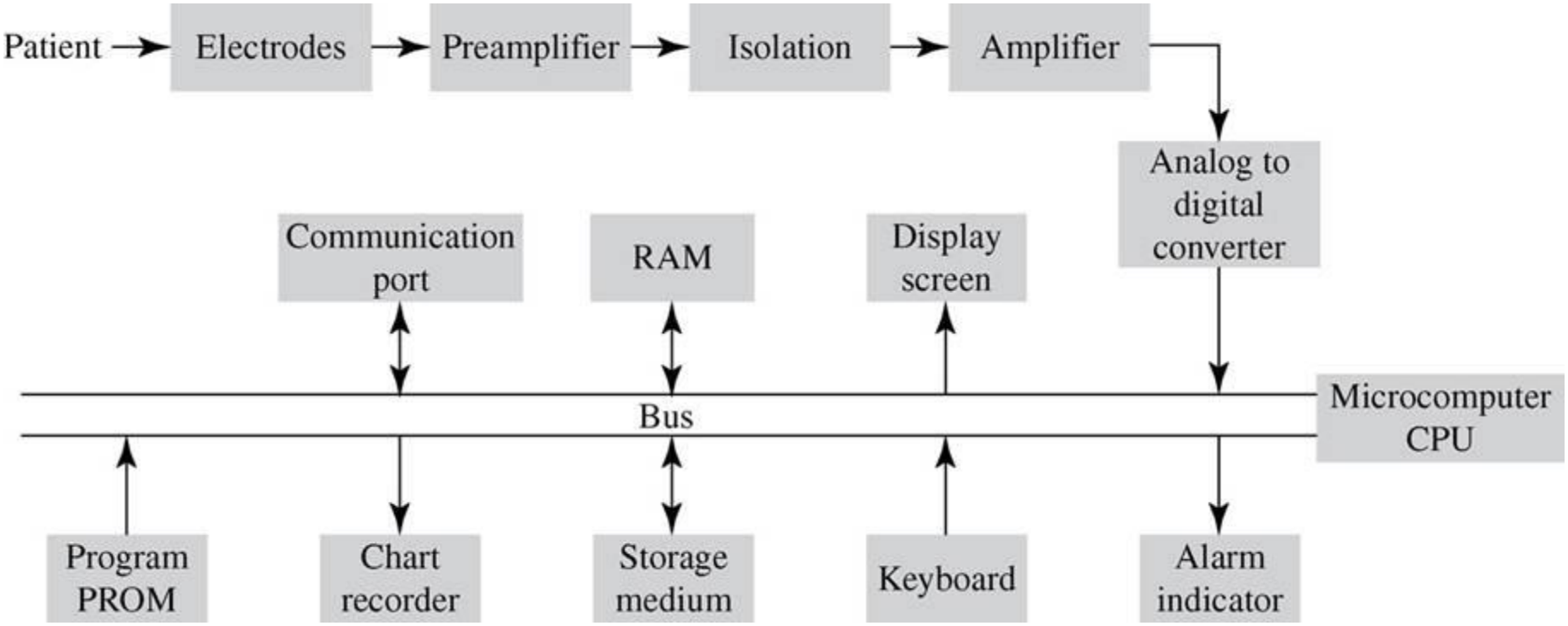


Figure 6.24 Block diagram of a cardiac monitor.

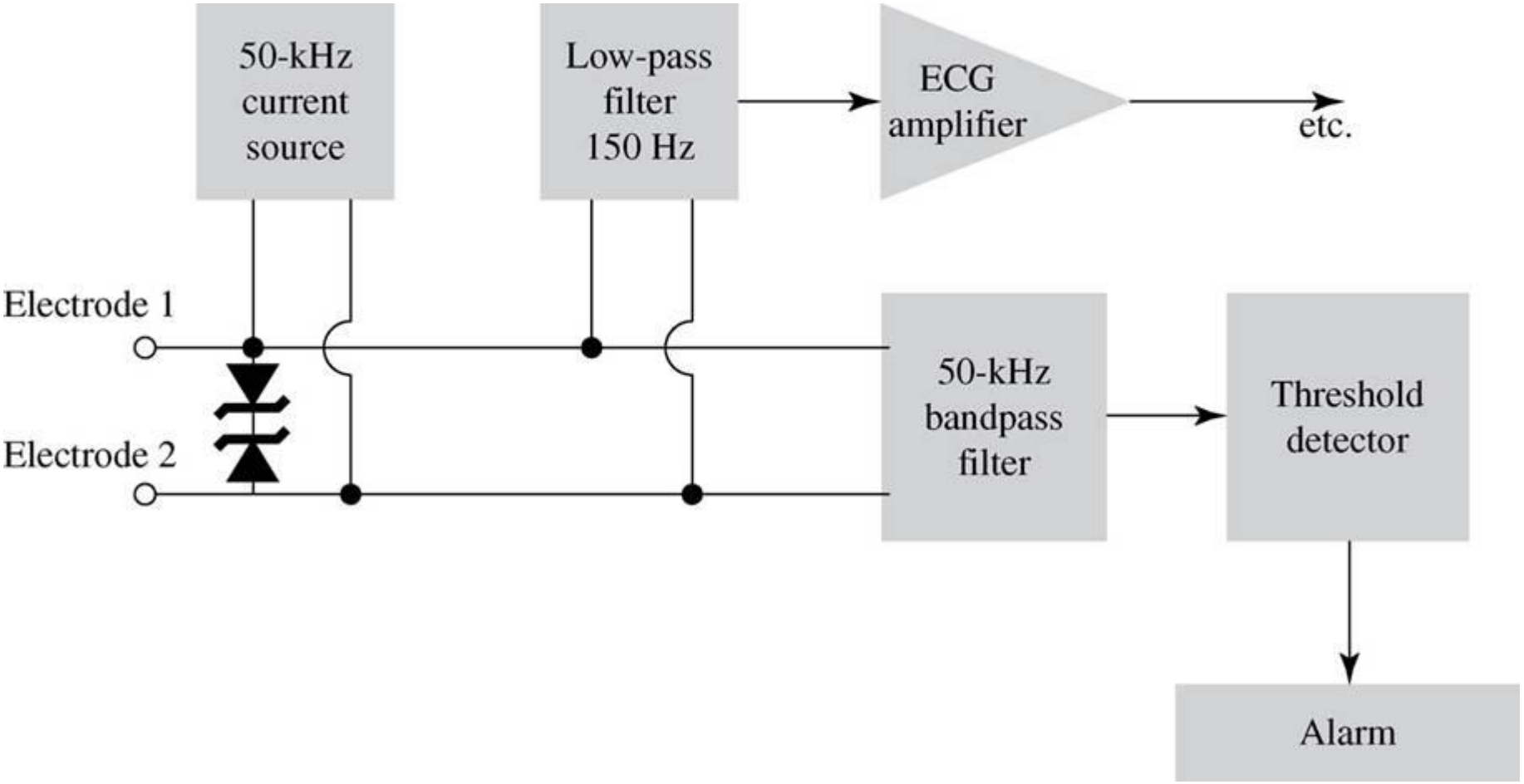


Figure 6.25 Block diagram of a system used with cardiac monitors to detect increased electrode impedance, lead wire failure, or electrode fall-off.

$$\frac{2v_{cm}}{R_a} + \frac{v_o}{R_f} = 0$$

$$v_o = -\frac{2R_f}{R_a} v_{cm}$$

$$v_{cm} = R_{RL} i_d + v_o$$

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

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

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

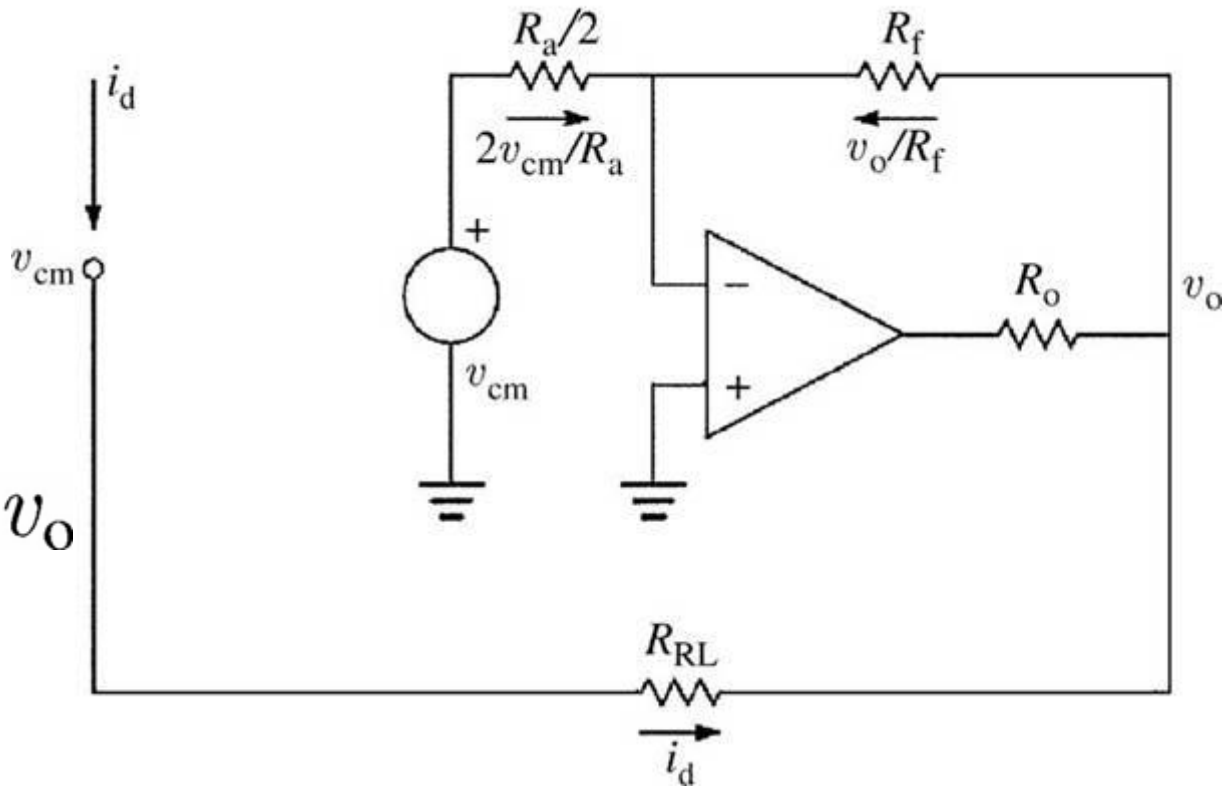
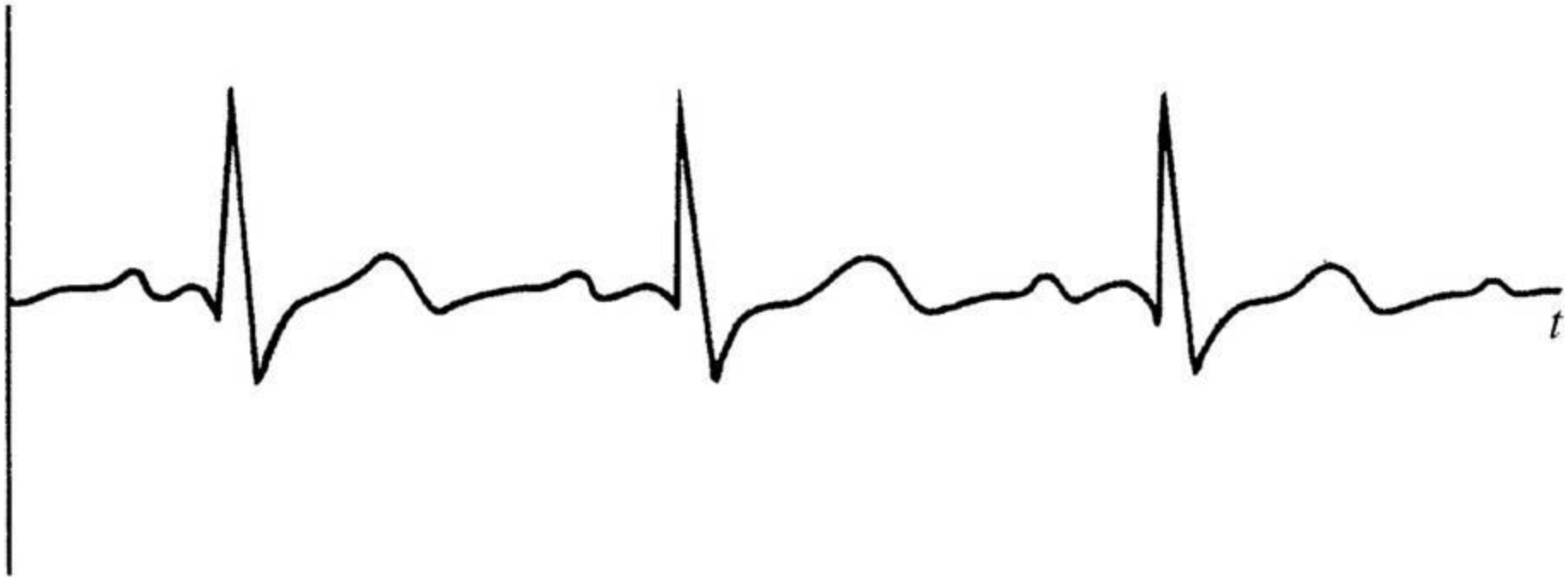
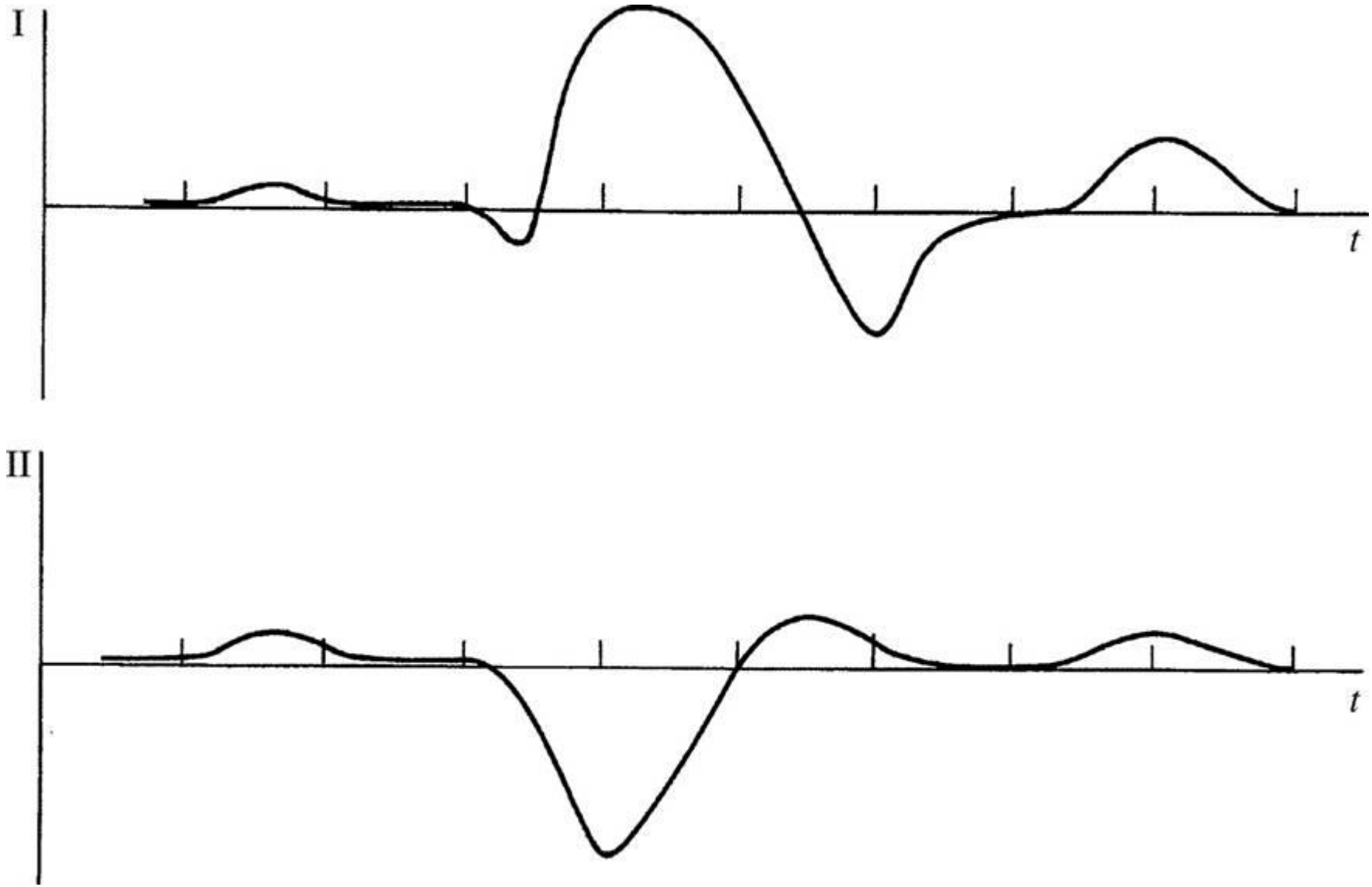


Figure E6.3 Equivalent circuit of driven-right-leg system of Figure 6.15.



The ECG shown is distorted as a result of an instrumentation problem. Discuss possible causes of this distortion, and suggest means of correcting the problem.



The figure shows ECGs from simultaneous leads I and II. Sketch the vector loop for this QRS complex in the frontal plane.