BIO AMPLIFIER

- 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 micro shocks or macro shocks 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 the such current can be minimized.
- 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.

PREAMPLIFIERS

Differential amplifier is one which will reject any common mode signal that appears simultaneously at both amplifier input terminals and amplifies only the voltage difference that appears across its input terminals. Most of the amplifiers used for measuring bioelectric signals are of the differential type and are also sometimes referred to as bio-potential amplifiers.

AC coupled amplifiers have a limited frequency response and are, therefore, used only for special medical applications such as electrocardiograph machine. For recording electrocardiograms, an ac amplifier with a sensitivity, giving 0.5 mV/cm, and frequency response up to 1 kHz and an input impedance of 2 to 5 MW is used. For such applications as EEG and EMG, more sensitive ac amplifiers are required, giving a chart sensitivity of say 50 μV/cm with a high input impedance of over 10 MW.

Carrier amplifiers are used with transducers which require an external source of excitation. They are characterized by high gain, negligible drift, extremely low noise and the ability to operate with resistive, inductive or capacitive type transducers. They essentially contain a carrier oscillator, a bridge balance and calibration circuit, a high gain ac amplifier, a phase sensitive detector and a dc output amplifier.

DC amplifiers are generally of the negative feedback type and are used for medium gain applications down to about 1 mV signal levels for full scale. They are not practical for very low-level applications because of dc drift and poor common-mode rejection capabilities. They are usually employed as pen drive amplifiers in direct writing recorders.

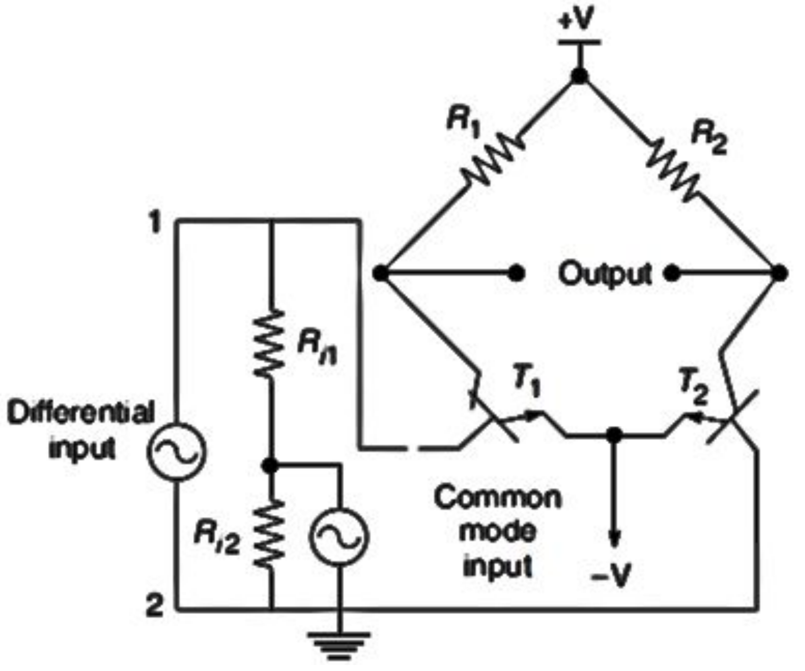
Chopper input dc amplifiers are preferred for low level inputs to instrumentation systems because of their high sensitivity, negligible drift and excellent common mode rejection capability. Their high frequency response is limited to about one half of the input chopper frequency.

Chopper-stabilized dc amplifiers are used for low level but preferably wideband applications such as oscilloscopes, tape recorders and light beam oscilloscope recorders. These are complex amplifiers having three amplifiers incorporated in the module. This includes an ac amplifier for signals above about 20 Hz, a dc chopper input amplifier for signals from about 20 Hz down to dc plus a wideband feedback stabilized dc amplifier.

DC bridge amplifiers are employed with resistive transducers which require an external source of excitation. Essentially, the amplifier comprises of a stable dc excitation source, a bridge balance and calibration unit, a high gain differential dc amplifier and a dc output amplifier. They can be used as conventional dc high gain amplifiers and offer operating simplicity and high frequency response. These amplifiers are necessary for transducers used to measure temperature and blood pressure. The sensitivity in these cases may be 50 mV/cm with an input impedance of 50 kW.

BIOPOTENTIAL AMPLIFIERS

- These amplifiers are designed to use in input stage (Preamplifiers)
- They have three input terminals out of which one is arranged at the reference potential and the other two are live terminals.
- The differential amplifier is employed when it is necessary to measure the voltage difference between two points, both varying in amplitude at different rates and in different patterns.
- Heart-generated voltages picked up by means of electrodes on the arms and legs, and brain-generated voltages picked up by the electrodes on the scalp are typical examples of signals whose measurement requires the use of differential amplifiers.



Typical differential amplifier configuration

- The differential amplifier is an excellent device for use in the bio-medical recording systems. Its excellence lies in its ability to reject common-mode interference signals which are invariably picked up by electrodes from the body along with the useful bioelectric signals.
- Also, as a direct coupled amplifier, it has good stability and versatility. High stability is achieved because it can be insensitive to temperature changes which are often the source of excessive drift in other configurations.
- It is versatile in that it may be adapted for a good many applications, e. g. applications requiring floating inputs and outputs or for applications where grounded inputs and/or outputs are desirable.

Working of a differential amplifier

Two transistors with their respective collector resistances (R_1 and R_2) form a bridge circuit. If the two resistors and the characteristics of the two transistors are identical, the bridge is perfectly balanced and the potential difference across the output terminals is zero.

Let us now apply a signal at the input terminals 1 and 2 of this circuit. The signal is to be such that at each input terminal, it is equal in amplitude but opposite in phase with reference to the ground.

This signal is known as the differential mode signal. Because of this signal, if the collector current of T_1 increases, the collector current of T_2 will decrease by the same amount, and the collector voltage of T_1 will decrease while that of T_2 will increase. This results in a difference in voltage between the two output terminals that is proportional to the gain of the transistors.

On the other hand, if the signal applied to each input terminal is equal in amplitude and is in the same phase (called the common-mode input signal), the change in current flow through both transistors will be identical, the bridge will remain balanced, and the voltage between the output terminals will remain zero.

Thus, the circuit provides high gain for differential mode signals and no output at all for common-mode signals. Resistances R_{i1} and R_{i2} are current limiting resistances for common-mode signals.

CMRR (common-mode rejection ratio):

- The ability of the amplifier to reject these common voltages on its two input leads is known as common-mode rejection and is specified as the ratio of common-mode input to differential input to elicit the same response.

- CMRR is an important specification referred to the differential amplifier and is normally expressed as decibels. CMRR of the preamplifiers should be as high as possible so that only the wanted signals find a way through the amplifier and all unwanted signals get rejected in the preamplifier stage.

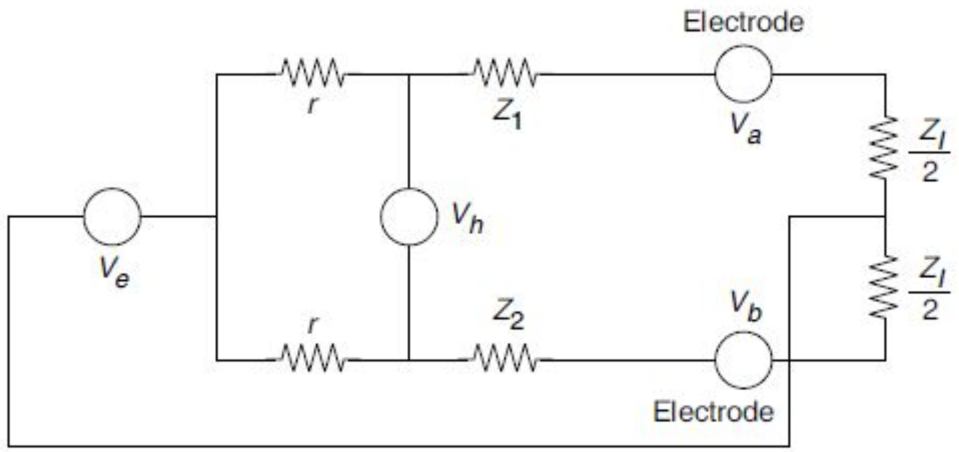
- A high rejection ratio is usually achieved by the use of a matched pair of transistors in the input stage of the preamplifier and a large 'tail' resistance in the long-tailed pair to provide maximum negative feedback for in-phase signals.

Even if the CMRR of an instrument is very high, still in some cases, in the hospitals, 50 Hz artefact can be seen in an ECG trace recording or monitor display.

A possible cause of the problem is source **impedance unbalance**; that is, the impedance of the electrodes as seen by the input of the ECG amplifier is not equal in all legs of the input.

The problem this creates is that some of the common mode voltage presented at each of the inputs is seen as a differential voltage and is amplified by the differential gain of the amplifier.

- V_h represents the voltage signal generated by the heart.
- V_e represents unwanted inphase signal picked up from the mains wiring and other sources.
- Z_i is the total input impedance of the preamplifier.
- Z_1 and Z_2 are the skin contact impedances of the electrodes.
- The resistance r represents tissue and blood resistance which is negligibly low as compared with other impedances.
- If the amplifier is perfectly balanced by equal inphase voltages, V_a and V_b, at the electrodes would give rise to a zero-output signal.
- However, the voltages V_a and V_b depend, in practice, on the values of Z_1 and Z_2. It can be shown that the electrical interference signal V_e will give rise to the same output signal as would a desired signal, from the patient, of amplitude



Equivalent circuit for the input of an ECG amplifier

Hence, the discrimination factor between desired and undesired signals (CMRR) is given by

$$\text{CMRR} = \frac{Z_1/2}{Z_2 - Z_1}$$

Assuming Z_i to be 10 MΩ and a difference of electrode skin contact impedance as 1KΩ, then

$$\text{CMRR} = \frac{Z_i}{2(Z_2 - Z_1)} = \frac{10 \times 10^6}{2 \times 1 \times 10^3} = 5000$$

In case the difference of electrode skin contact impedance is 5 KΩ, then

$$\text{CMRR} = \frac{Z_i}{2(Z_2 - Z_1)} = \frac{10 \times 10^6}{2 \times 5 \times 10^3} = 1000$$

Thus, we see that the impedance unbalance due to the electrodes on the ECG has reduced the common mode rejection from 5000 to 1000 only.

Further, let us suppose that Z_i is 100 MΩ and the difference in electrode skin contact impedance is 5 KΩ, then

$$\text{CMRR} = \frac{Z_i}{2(Z_2 - Z_1)} = \frac{10 \times 10^6}{2 \times 5 \times 10^3} = 10,000$$

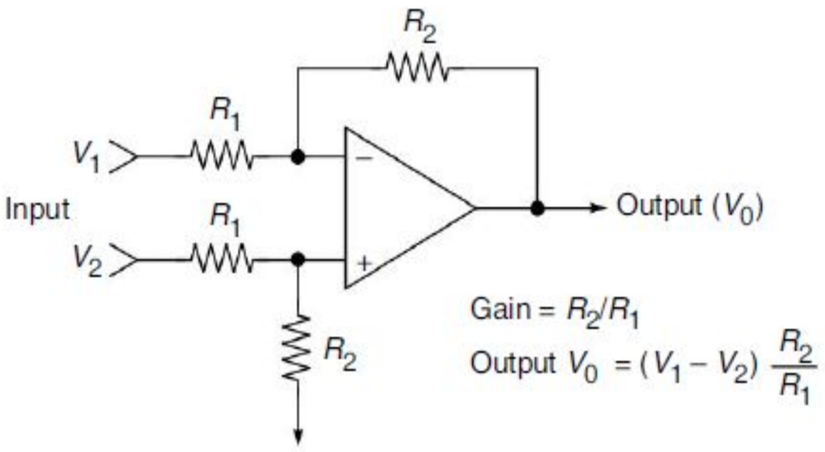
This shows that a high input impedance is very necessary in order to obtain a high CMRR.

Also, the electrode skin resistance should be low and as nearly equal as possible

The common mode rejection for most op-amps is typically between 60 dB and 100 dB. This may not be sufficient to reject common mode noise generally encountered in biomedical measurements.

Also, the input impedance is not very high to handle signals from high impedance sources. One method to increase the input impedance of the op-amp is to use field effect transistors (FET) in the input differential stage.

A more common approach is to use an instrumentation amplifier in the preamplifier stage



A single op-amp in a differential configuration

INSTRUMENTATION AMPLIFIER

The differential amplifier is well suited for most of the applications in biomedical measurements.

However, it has the following limitations:

- The amplifier has limited input impedance, and therefore, draws some current from the signal source and loads them to some extent.
- The CMRR of the amplifier may not exceed 60 dB in most cases, which is usually inadequate in modern biomedical instrumentation systems.

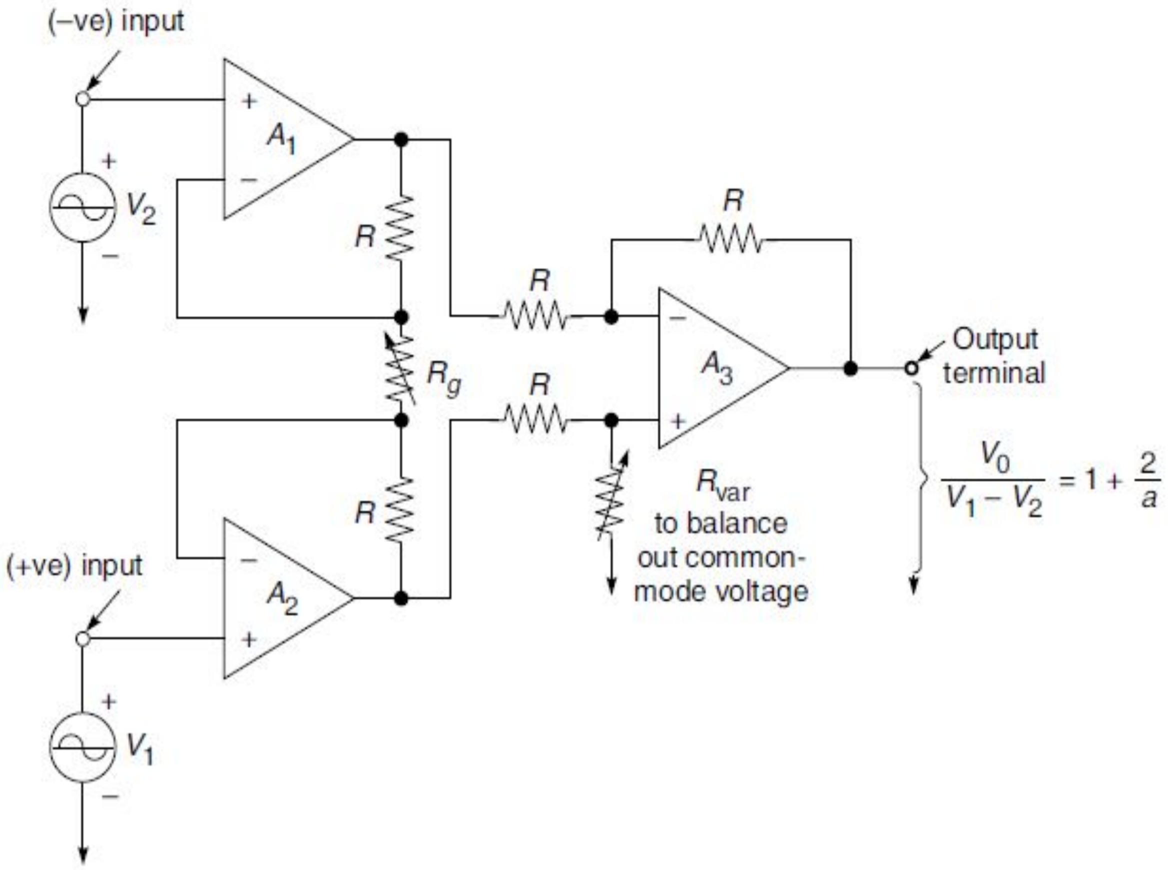
These limitations have been overcome with the availability of an improved version of the differential amplifier, known as an **instrumentation amplifier** which is a precision differential voltage gain device that is optimized for operation in an environment hostile to precision measurement.

It basically consists of three op-amps and seven resistors. Basically, connecting a buffered amplifier to a basic differential
The amplifier makes an instrumentation amplifier.

op-amp A_3 and its four equal resistors R form a differential amplifier with a gain of 1. Only A_3 resistors must be matched.

The variable resistance R_{var} is varied to balance out any common-mode voltage. Another resistor R_g, is used to set the gain using the formula

$$\frac{V_0}{V_1 - V_2} = 1 + \frac{2}{a}$$ where a = R_g/R.



V_1 is applied to the +ve input terminal and V_2 to the –ve input terminal. V_0 is proportional to the difference between the two input voltages.

The important characteristics of the instrumentation amplifier are:

- Voltage from different input ($V_1 – V_2$) to single ended output, is set by one resistor.
- The input resistance of both inputs is very high and does not change as gain is varied
- V_0 does not depend on common-mode voltage, but only on their difference

If the inputs are prone to high voltage spikes or fast swings, which the op-amps cannot cope with, they may be protected using back-to-back connected diodes at their inputs. However, this reduces the input impedance value substantially and limits the bandwidth.

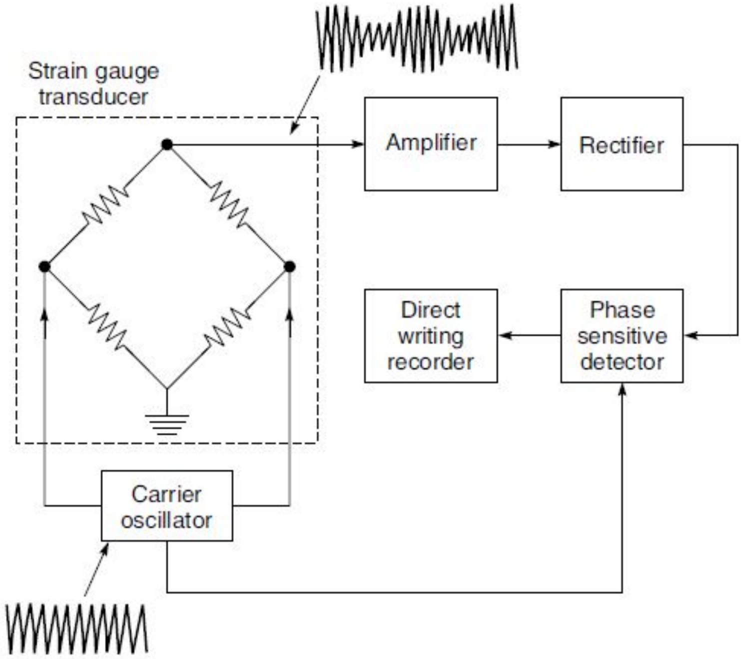
The instrumentation amplifier offers the following advantages for its applications in the biomedical field:

- **Extremely high input impedance**
- **Low bias and offset currents**
- **Less performance deterioration if source impedance changes**
- **Possibility of independent reference levels for source and amplifier**
- **Very high CMRR**
- **High slew rate**
- **Low power consumption**

Good quality instrumentation amplifiers have become available in single IC form such as μA725, ICL7605, LH0036, etc.

CARRIER AMPLIFIER

To obtain zero frequency response of the dc amplifier and the inherent stability of the capacitance coupled amplifier, a carrier type of amplifier is generally used.
The carrier amplifier consists of an oscillator and a capacitance coupled amplifier. The oscillator is used to energize the transducer with an alternating carrier voltage.
The transducers, which require ac excitation, are those whose impedance is not purely resistive.
Example can be of a capacitance-based pressure transducer whose impedance is mainly capacitative with a small resistive component.
The frequency of the excitation voltage is usually around 2.5 kHz.
The transducer shall change the amplitude of the carrier voltage in relation to the changes in the physiological variable being measured.
The output of the transducer, therefore, would be an amplitude modulated (AM) signal.
The modulated ac signal can then be fed to a multi-stage capacitance coupled amplifier.
The first stage produces amplification of the AM signal.
The second stage is so constructed that it can respond only to signal frequency of the carrier.
It can be further amplified in the following stage. After amplification, the signal is demodulated in a phase-sensitive demodulator circuit.
This helps to extract amplified signal voltage after the filter circuit. The voltage produced by the demodulator can then be applied to the driver stage of the writing system.



Carrier amplifiers can be used with a resistance strain gauge transducer such as a semiconductor strain gauge. When used with pressure gauges, a calibration control is provided on the carrier amplifier. This enables direct measurements of the blood pressure from the calibrated graphic recorder.

Lock-in amplifier is a useful version of the carrier technique designed for the measurement of low-level signals buried in noise. This type of amplifier, by having an extremely narrow-width output band in which the signal is carried, reduces wideband noise and increases the signal-to noise ratio. Thus, the difference between carrier amplifier and lock-in amplifier is that the former is a general-purpose instrument amplifier while the latter is designed to measure signals in a noisy background.

In principle, the lock-in amplifier works by synchronizing on a single frequency, called the reference frequency. This frequency is made to contain the signal of interest. The signal is modulated by the reference frequency in such a way that all the desired data is at the single reference frequency whereas the inevitable noise, being broadband, is at all frequencies. This permits the signal to be recovered from its noisy background.

CHOPPER AMPLIFIER

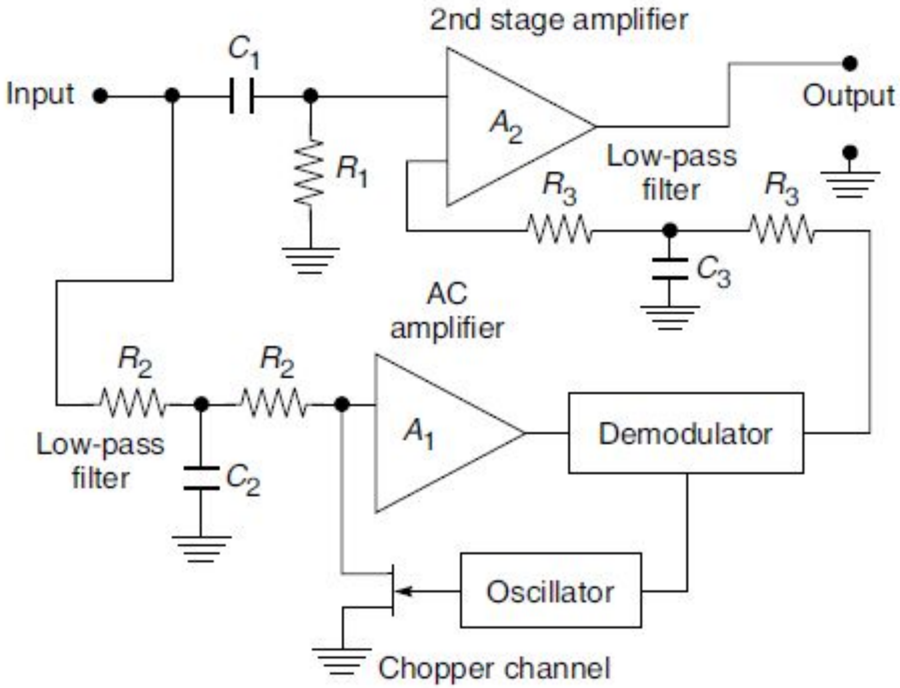
The chopper amplifier is a useful device in the field of medical electronics as it gives another solution to the problem of achieving adequate low frequency response while avoiding the drift problem inherent in direct coupled amplifiers.

This type of amplifier makes use of a chopping device, which converts a slowly varying direct current to an alternating form with amplitude proportional to the input direct current and with phase dependent on the polarity of the original
signal.

The alternating voltage is then amplified by a conventional ac amplifier whose output is rectified back to get an amplified direct current.

A chopper amplifier is an excellent device for signals of narrow bandwidth and reduces the drift problem.

Fig shows a simplified block diagram of a single-ended chopper-stabilized amplifier.



The amplifier achieves its ultra low dc offset voltage and bias current by chopping the low frequency components of the input signal, amplifying this chopped signal in an ac amplifier (A_1) and then demodulating the output of the ac amplifier.

The low frequency components are derived from the input signal by passing it through the low-pass filter, consisting of R_2, C_2 and R_2. The chopping signal is generated by the oscillator.

The filtered output is then further amplified in a second stage of dc amplification (A_2).

High frequency signals, which are filtered out at the input of the chopper channel, are coupled directly into the second stage amplifier.

The result of this technique is to reduce the dc offsets and drift of the second amplifier by a factor equal to the gain of the chopper channel.

The ac amplifier introduces no offsets. Minor offsets and bias currents exist due to imperfect chopping, but these are extremely small.

The amplifier modules contain the chopper channel, including switches and switch-driving oscillator built on the module; only the dc power is supplied externally.

The great strength of the chopper-stabilized amplifier is its insensitivity to component changes due to ageing, temperature change, power supply variation or other environmental factors.

Thus, it is usually the best choice where both offset voltage and bias current must be small over long periods of time or when there are significant environmental changes.

Both bias current and offset voltage can be externally nulled. Chopper amplifiers are available in both singleended as well as differential input configurations.

Chopper amplifiers find applications in the medical field in amplification of small dc signals of a few microvolts.

Such order of amplitudes are obtainable from transducers such as strain gauge pressure transducers, temperature sensors such as thermistors and strain gauge myographs, when they are used as arms of a dc Wheatstone bridge.

A chopper amplifier is also suitable for use with a thermocouple

ISOLATION AMPLIFIERS

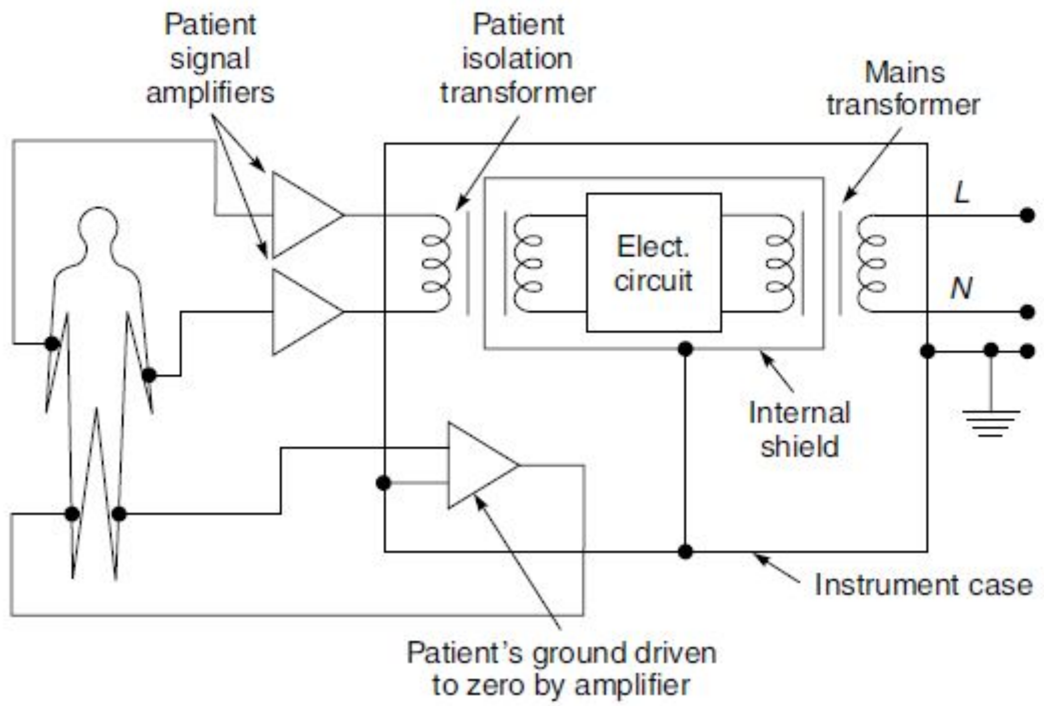
Isolation amplifiers are commonly used for providing protection against leakage currents in biomedical recorders such as ECG machine.

They break the ohmic continuity of electric signals between the input and output of the amplifier.

The isolation includes different supply voltage sources and different grounds on each side of the isolation barrier.

Three methods are used in the design of isolation amplifiers:

(i) transformer isolation
(ii) (ii) optical isolation
(iii) (iii) capacitive isolation.



Isolation amplifier (transformer type)

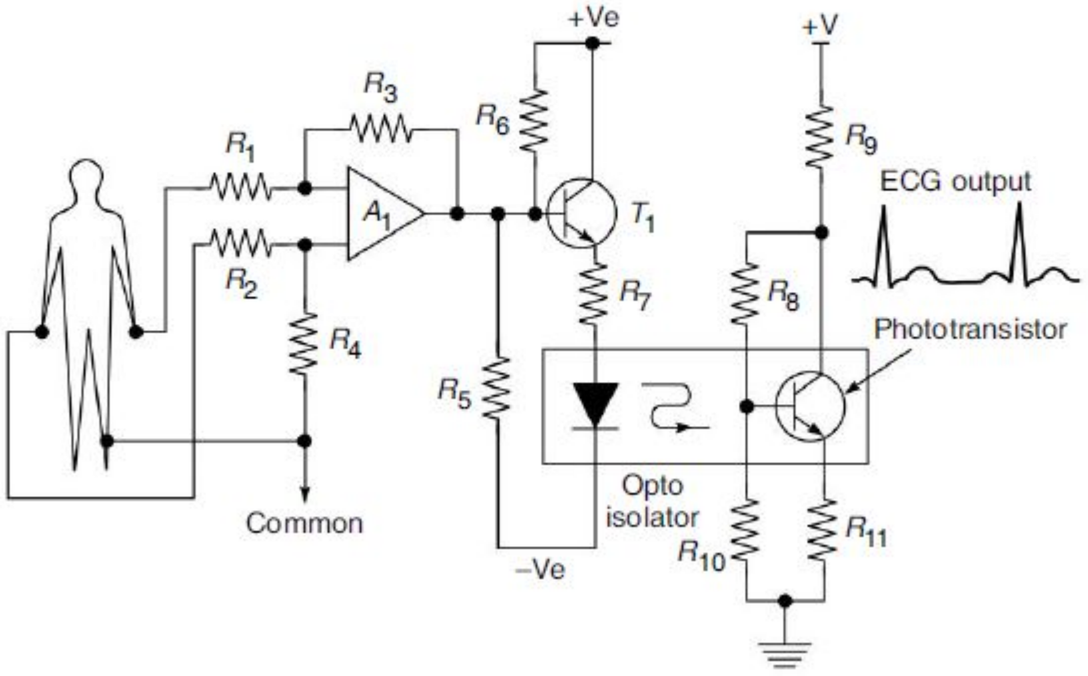
ISOLATION AMPLIFIERS

Isolation could also be achieved by optical means in which the patient is electrically connected with neither the hospital line nor the ground line.

A separate battery-operated circuit supplies power to the patient circuit and the signal of interest is converted into light by a light source (LED).

This light falls on a phototransistor on the output side, which converts the light signal again into an electrical signal; having its original frequency, amplitude and linearity.

No modulator/demodulator is needed because the signal is transmitted optically all the way.

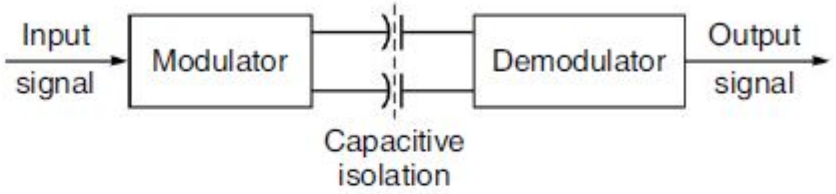


Optically isolated isolation amplifier

ISOLATION AMPLIFIERS

The capacitive method uses digital encoding of the input voltage and frequency modulation to send the signal across a differential capacitive barrier.

Separate power supply is needed on both sides of the barrier. Signals with bandwidths up to 70 kHz can be conveniently handled in this arrangement



The relative merits of the three types of isolation techniques are:

- All three types are in common use, though the transformer isolation amplifier is more popular.
- Opto-coupled amplifier uses a minimum number of components and is cost effective, followed by the transformer coupled amplifier. The capacitor coupled amplifier is the most expensive.
- Opto-isolated amplifiers offer the lowest isolation voltage (800 V continuous) between input and output; transformer coupled 1200 V and capacitance coupled 2200 V.
- Isolation resistance levels are of the order of 1010, 1011 and 1012 ohms for transformer coupled, opto-coupled and capacitance coupled amplifiers respectively.
- Gain stability and linearity are best for capacitance coupled versions-0.005%, and on par for the transformer and opto-coupled amplifier 0.02%.

- **THE ECG:** 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.

- 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 represents 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.

- The electric potentials generated by the heart appear throughout the body and on its surface.

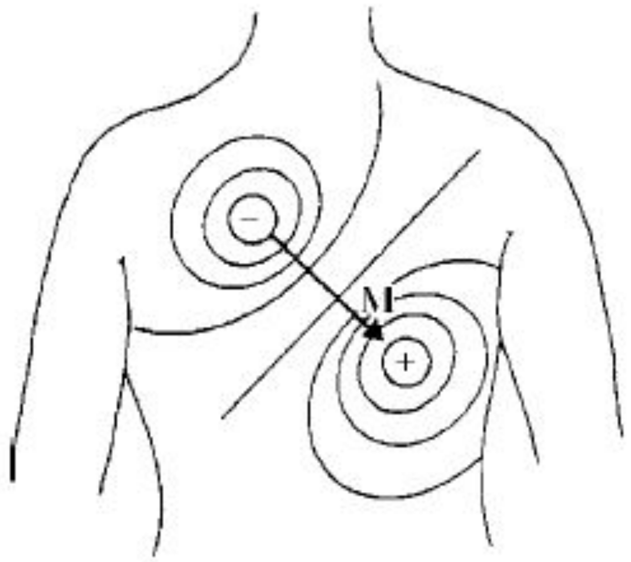
- We determine potential differences by placing electrodes on the surface of the body and measuring the voltage between them, being careful to draw little current (ideally there should be no current at all, because current distorts the electric field that produces the potential differences).

- 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.

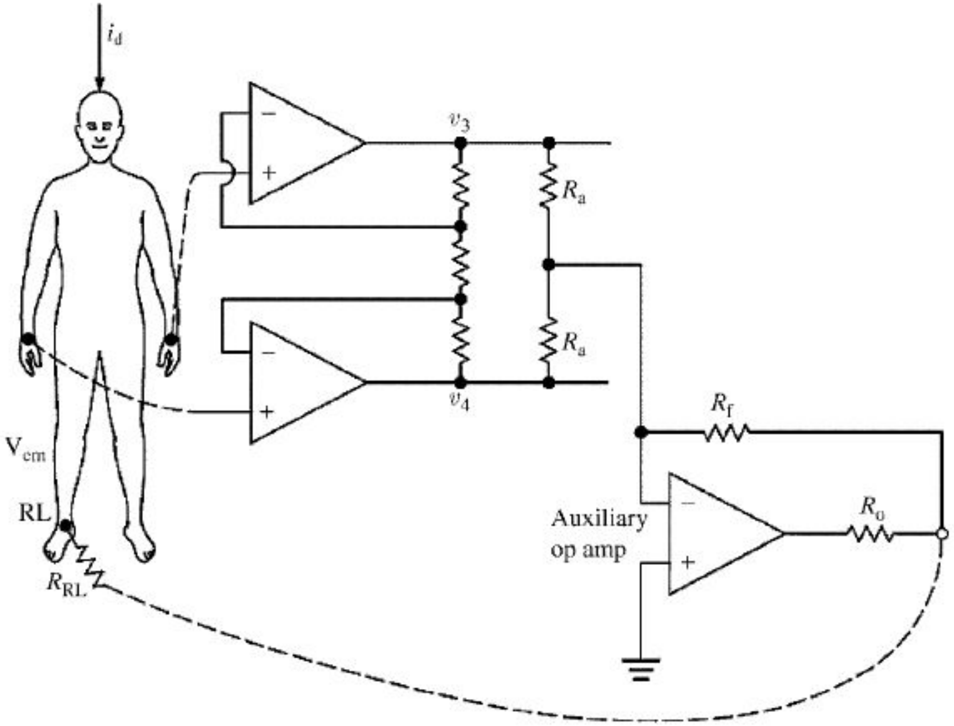
- The limbs make fine guideposts for locating the ECG electrodes.



RIGHT LEG DRIVEN ECG AMPLIFIER

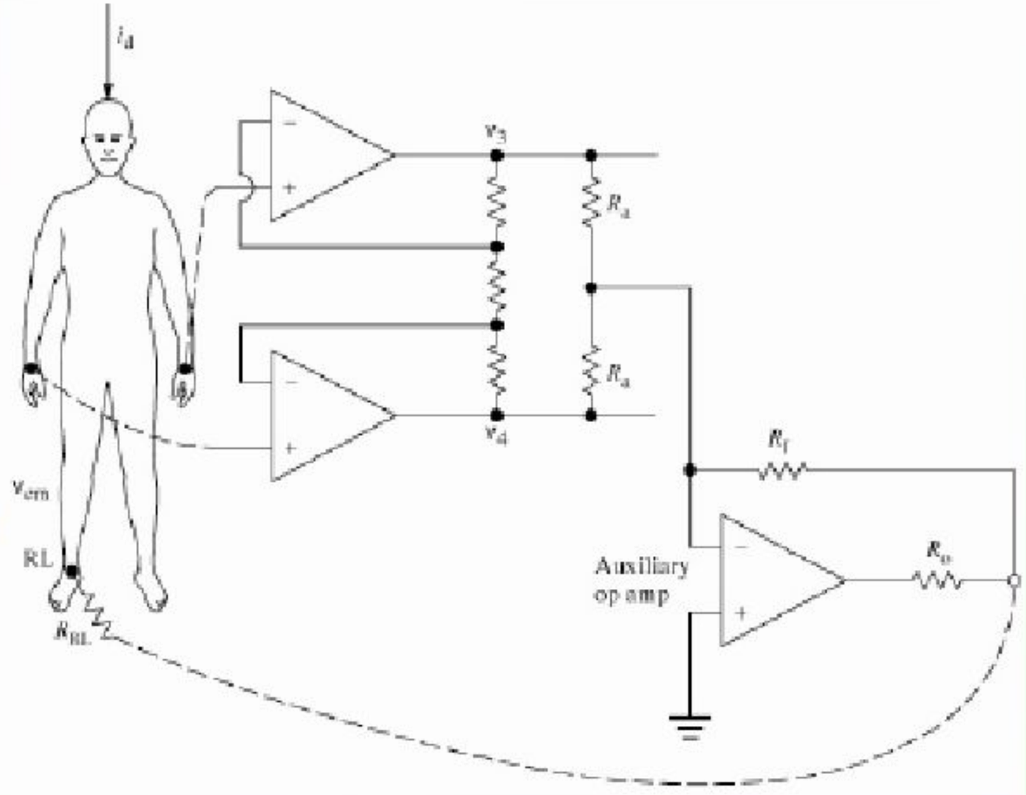
A **Driven Right Leg Circuit** or **DRL** circuit is an electric circuit that is often added to biological signal amplifiers to reduce Common-mode interference. Biological signal amplifiers such as ECG (Electrocardiogram) EEG (Electroencephalogram) or EMG circuits measure very small electrical signals emitted by the body, often as small as several micro-volts (millionths of a volt). Unfortunately, the patient's body can also act as an antenna which picks up electromagnetic interference, especially 50/60 Hz noise from electrical power lines. This interference can obscure the biological signals, making them very hard to measure. Right Leg Driver circuitry is used to eliminate interference noise by actively cancelling the interference. Other methods of noise control include:

- Faraday cage
- Twisting Wires
- High Gain Instrumentation Amplifier
- Filtering



Driven Right Leg System

- Motivation
  - reduce interference in amplifier
  - improve patient safety
- Approach
  - patient right leg tied to output of an auxiliary amp rather than ground
  - common mode voltage on body sensed by averaging resistors, Ra's & fed back to right leg
  - provides negative feedback to reduce common mode voltage
  - if high voltage appears between patient and ground, auxiliary amp effectively un-grounds the patient to stop current flow



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

- The common-mode voltage on the body is sensed by the two averaging resistors Ra, 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 the 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.

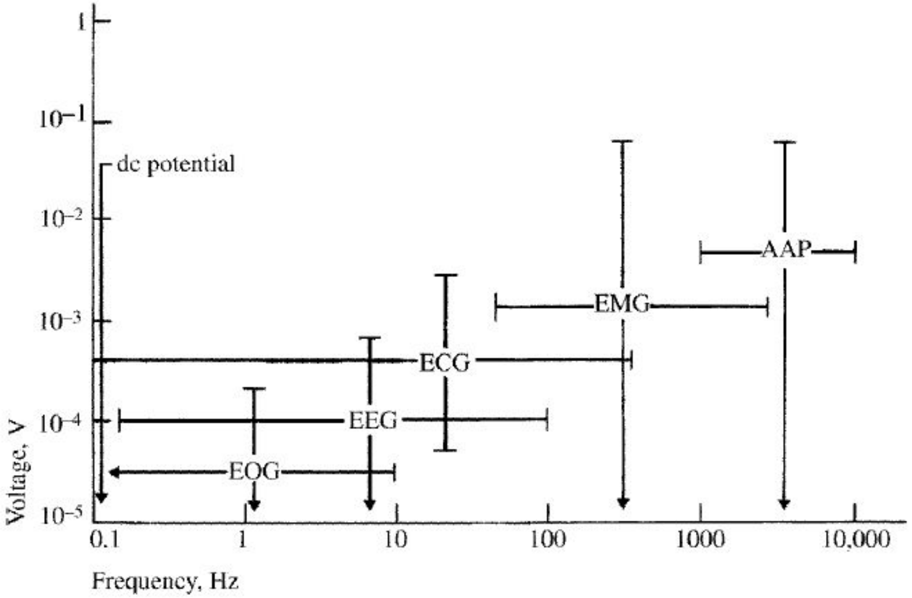
- The circuit can also provide some electric safety. If an abnormally high voltage should appear between the patient and the ground as a result of electric leakage or other cause, the auxiliary op amp saturates.

- This effectively ungrounds the patient, because the amplifier can no longer drive the right leg. Now the parallel resistances R, and R, are between the patient and the 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.

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 frequency-response constraints on the amplifiers used.



- Figure shows the ranges of amplitudes and frequencies covered by several of the common bio potential signals. Depending on the signal, frequencies range from de 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.

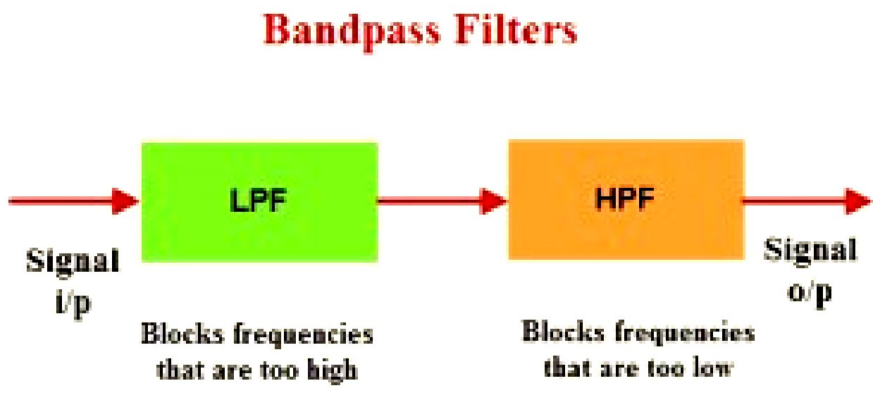
- Also, the amplifier input circuit must not promote the generation of artifact by the electrode, as could occur with excessive bias current.

FILTER

- A filter is a device in signal processing used to allow wanted frequency components from the signals and to remove unwanted ones.
- The background noise of the interfacing signal can be reduced by eliminating some frequencies, which is known as filtering.
- Filter circuits can be designed to combine the properties of the LPF (low pass filter) and HPF (high pass filter) into a single filter, which is known as a bandpass filter.
- This filter can be created by combining a low pass filter and a high pass filter.

TYPES OF FILTER

- Active or passive
- Time variant or time invariant
- Linear or non-linear
- Analog or digital

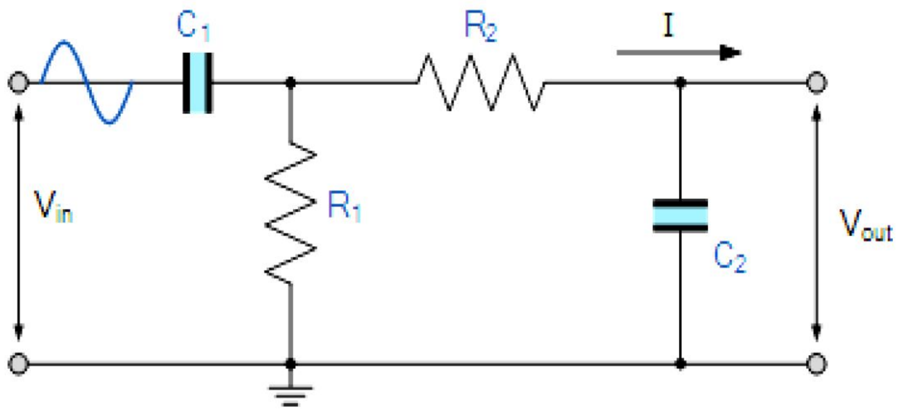


BANDPASS FILTER

• A band-pass filter (BPF) is an electronic circuit that allows signals between two precise frequencies to pass but separates signals at other frequencies.
• Some BPFs are involved in an external power source and utilize active components like transistors and ICs (integrated circuits), known as active bandpass filters.
• Other BPFs don't use any power source and only passive components like inductors and capacitors; these are known as passive bandpass filters.
• Bandpass filters are mainly involved in wireless transmitters and wireless receivers.
• The main objective of this filter in a transmitter is to limit the bandwidth of the o/p signal to the minimum required level and to convey data at the preferred speed and in the preferred form.
• In a receiver, a BPF (bandpass filter) allows signals within a preferred range of frequencies to be decoded while avoiding signals at redundant frequencies from getting through.
• A BPF also optimizes the S/N ratio (signal to noise) of a receiver.

Bandpass Filter Circuit Diagram

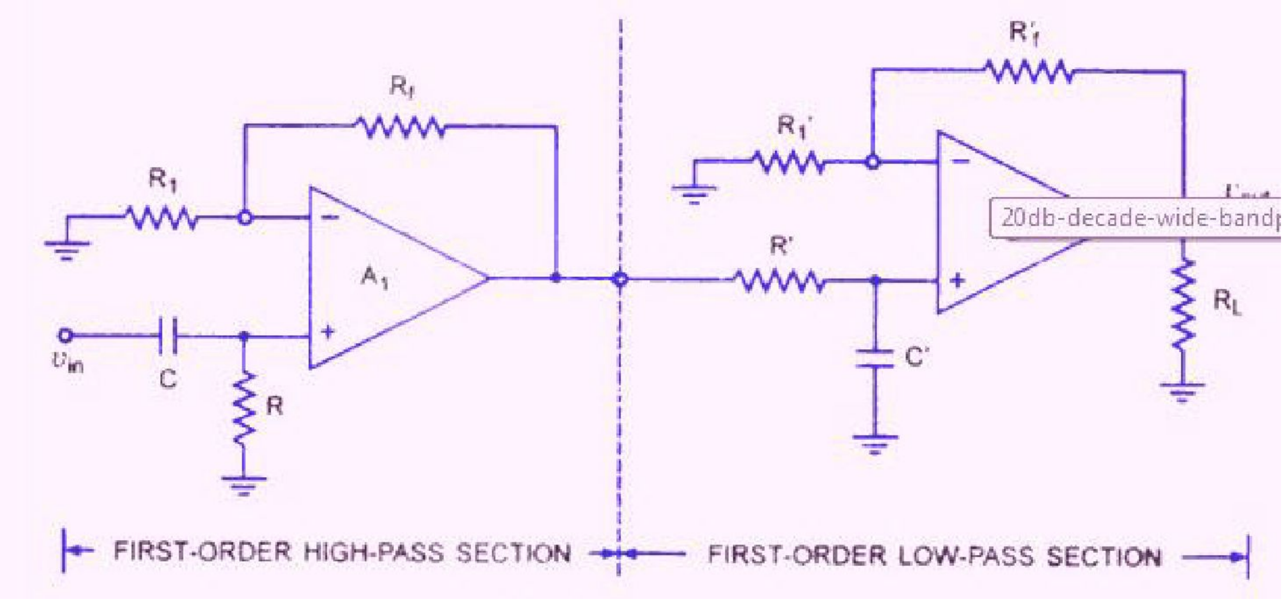
• Bandpass filters pass signals with a certain band of frequencies without deforming the i/p signal.
• The band of frequencies can be very wide, and it is normally known as the filter's bandwidth.
• The term bandwidth can be defined as the frequency range between two particular cutoff frequency points (fc).
• For widespread frequencies, bandwidth can be defined as, the difference between the lower cutoff frequency and the higher cutoff frequency points ($BW = f_H - f_L$).



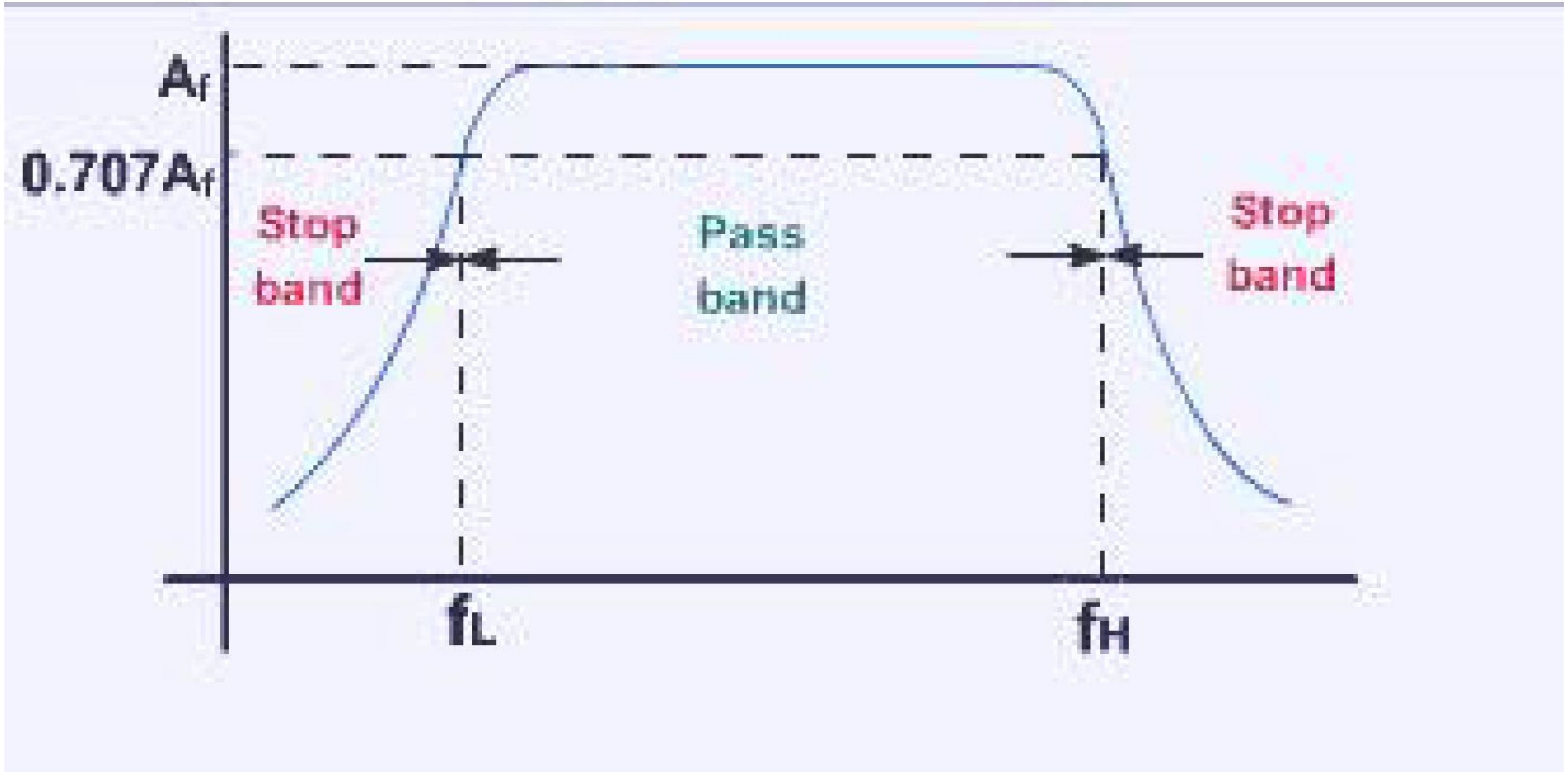
• The cutoff frequency of the LPF must be higher than the cutoff frequency of the HPF for the proper working of the pass band filter.
• An ideal bandpass filter can also be used to filter out or isolate certain frequencies that recline in a specific band of frequencies.
• For instance, BPFs are known as second-order filters because they have two reactive components.
• The circuit diagram of the bandpass filter is built with the capacitors by placing one capacitor in the LPF (low pass filter) circuit and the other capacitor in the HPF (high pass filter) circuit.
Types of Bandpass Filters Bandpass filters are categorized into two types:
• Wide bandpass filter
• Narrow bandpass filter.

Wide Bandpass Filter

- A wide bandpass filter (WBF) formed by cascading low-pass and high-pass sections which is generally an alternative circuit for ease of design and performance.
- It is realized by a number of feasible circuits. A ± 20 db/ decade bandpass filter formed by first-order low-pass and high-pass sections can be cascaded.
- In the same way, a ± 40 db/decade bandpass filter can be formed by connecting a second-order low-pass and high-pass filter in series.
- The order of the BPF (bandpass filter) is governed by the order of the low-pass and high-pass filters it has.
- A ± 20 db/decade wide bandpass filter is composed of a first-order high-pass filter (HPF).



BPF Frequency Response



Narrow Bandpass Filter

• Narrow bandpass filter is named as a multiple-feedback filter as it consists of two feedback paths.
• The operational amplifier is used in the inverting mode.
• Generally, the narrow bandpass filter (BPF) is designed for precise values of fc (center frequency) and Q or center frequency and BW.
• The required components of the circuit are determined from the following relationships.
• Each of the C1 & C2 can be taken equal to C for design calculation simplifications.

R1 = Q / 2pfc CAf
R1 = Q / 2pfc C (2Q2-Af)
R3 = Q / pfc C
Where _Af‘ is the gain at the _fc‘ center frequency and is given as **Af = R3/2R1**

However, the gain Af must satisfy the following condition.
Af < 2 Q2
• The _fc_ (center frequency) of the various feedback filters can be altered to a new frequency _fc_ without changing the gain or bandwidth.
• This is attained simply by altering R2 to R‘2 so that

R’2 = R2 [fc/f’c]2