WEARABLES

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Hilary Term

In this course, we will study the principles and the design of wearables for the non-invasive measurement of physiological parameters known as the "vital signs". We will cover four of the most important vital signs that can be measured non-invasively:

1. The electrocardiogram (ECG)
   - Electrodes and the conversion of ionic potentials to electronic potentials
   - ECG instrumentation amplifiers
   - Noise reduction using the "driven right leg" circuit
   - [A quick, unexamined, look at electroencephalography (EEG)]

2. Respiration measurement
   - Electrical impedance changes of the chest cavity due to breathing
   - 2 or 4 electrode measurement?
   - 4-electrode method of measuring electrical impedance: principles, amplifier and signal processing circuits

3. Oxygen saturation using pulse oximetry
   - The optical characteristics of oxygenated and deoxygenated blood
   - Principles of pulse oximetry
   - Electronic implementation of sub-systems for pulse oximetry, including separation of the d.c. and a.c. components.
   - Detailed circuits for pulse oximetry.

4. Non-invasive blood pressure measurement
   - Theory and circuitry for two methods:
     1. The traditional method: Korotkoff sounds
     2. The current method: oscillometry

The electrocardiogram (ECG)

The heart has four chambers: left atrium, right atrium, left ventricle and right ventricle. Located at the top of the right atrium are a group of cells which act as the primary pacemaker for the heart. Through a complex change of ionic concentration across the cell membranes, an extra-cellular potential field is established. This potential field excites the neighbouring cells, resulting in cell-to-cell electrical propagation within the heart. Since the thorax acts as a conductive medium, the potential field generated by the heart propagates to the body surface.

If a pair of surface electrodes, attached to the left and right arms of a human subject, are connected to a high-input impedance differential amplifier, an electrical signal which varies in time with the heart beat will be observed at the output of the amplifier (see Figure 1). This signal, which has a peak amplitude, before amplification, of the order of 1mV, is known as the electrocardiogram (ECG).

Figure 1: An ECG trace. 1, 2, 3 ... are consecutive heart beats. The heart rate, in beats per minute, is $60/\Delta T$

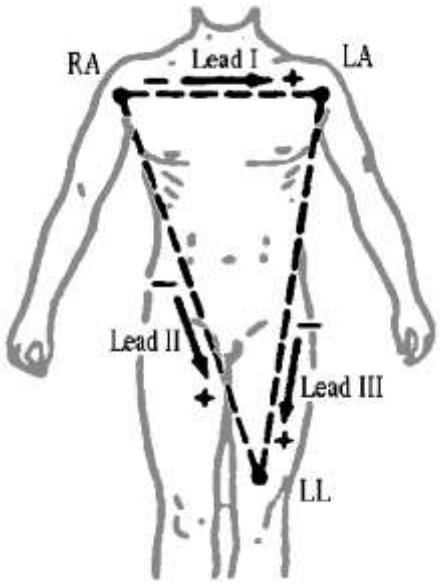
In order to record the ECG, we need a transducer capable of converting the ionic potentials generated within the body into electronic potentials that can be measured by conventional electronic instrumentation. Such a transducer consists of a pair of *electrodes*, which measure the ionic potential difference between their respective points of application on the body surface. Electrodes may be classified either as polarisable, in

which case they behave as capacitors, or non-polarisable, in which case they behave as resistors. Common electrodes have characteristics that lie between these extremes; the silver-silver chloride electrode discussed below approximates more closely to a non-polarisable electrode.

Equivalent circuit of a system for recording the ECG

Electrode placement

The most obvious way to record the ECG is between the right arm (RA) and the left arm (LA) although another two combinations using the left leg (LL) are also used clinically (RA-LL and LA-LL). Figure 2 summarises this.



V_I = (potential at LA) – (potential at RA)

V_{II} = (potential at LL) – (potential at RA)

V_{III} = (potential at LL) – (potential at LA)

Figure 2: Typical electrode placemen

A third electrode is also used to connect the patient to the common ground of the instrumentation. Usually, this ground electrode is attached to the right leg.

Silver-silver chloride electrode

Electrodes for recording biopotentials are composed of a metal (usually silver for ECG measurement), and a salt of the metal (usually silver chloride). In addition, some form of electrode paste or jelly is applied between the electrode (normally a flat silver disc) and the skin. The combination of the ionic electrode paste and the silver metal of the electrode forms a local solution of the metal in the paste at the electrode-skin interface (also referred to as the electrode-tissue interface or electrode-electrolyte interface). Hence, some of the silver dissolves into solution producing Ag^+ ions:

$$Ag \rightarrow Ag^+ + e^-$$

Ionic equilibrium takes place when the electric field set up by the dissolving ions is balanced by the forces of the concentration gradient. At this point, there is a monomolecular layer of Ag^+ ions at the surface of the electrode and a corresponding layer of Cl^- ions adjacent to this. This combination is called the *electrode double layer* and there is a potential drop E across this layer, called the *half-cell potential* (0.8V in the case of the *Ag-AgCl* electrode).

Equivalent circuit of electrode interface

The double layer of charges of opposite sign separated by a dielectric constitutes a form of capacitance, say C. However, since the *Ag-AgCl* electrode behaves mostly as a non-polarisable electrode, the main component of the impedance is resistive, say R_1.

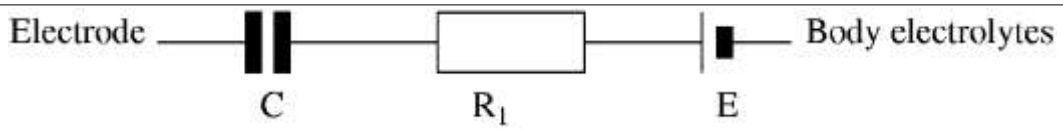


Figure 3: Simple R-C model of electrode-electrolyte interface

The series model in Figure 3 needs to be modified to account for the fact that the impedance does not increase to infinity as the frequency tends to zero. This is done by adding a parallel resistance R_2 (as shown in Figure 4) which accounts for the electrochemical processes taking place at the electrode-electrolyte interface. The values of R_1, R_2 and C depend on the electrode area, surface condition, current density and the type and concentration of electrode paste used. (Typical values are $R_1 = 2k\Omega$, $R_2 = 10k\Omega$ and $C = 10\mu F$.)

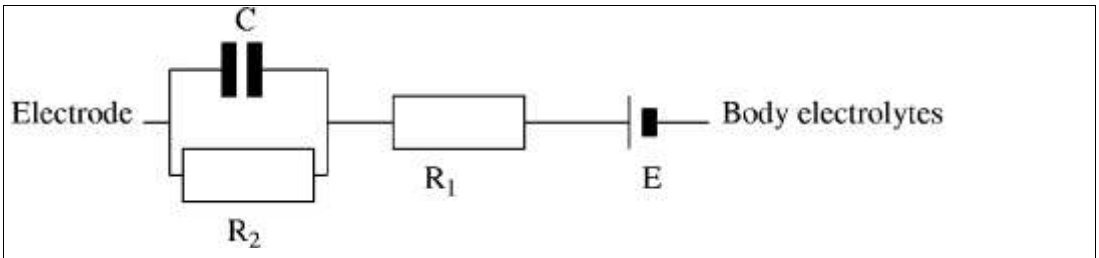


Figure 4: Equivalent circuit of the Ag-AgCl interface

Movement artefact

If the electrode is moved with respect to the electrolyte, this mechanically disturbs the distribution of charge at the interface and results in a momentary change of the half-cell potential until equilibrium can be re-established. If a pair of electrodes is in contact with an electrolyte and one of the electrodes moves while the other remains stationary, a potential difference appears between the two during this motion. This potential is referred to as *movement artefact* and can be a

serious cause of interference in the measurement of the ECG (or any other biopotential).

Overall equivalent circuit

If we represent the electrical activity of the heart by a voltage generator, model the tissues in the thorax as resistors and use the simple model of the electrode-electrolyte interface of Figure 4, we can put together an equivalent circuit which models the electrical impedance seen by the input stage of an ECG system. This overall equivalent circuit is shown in Figure 5 below.

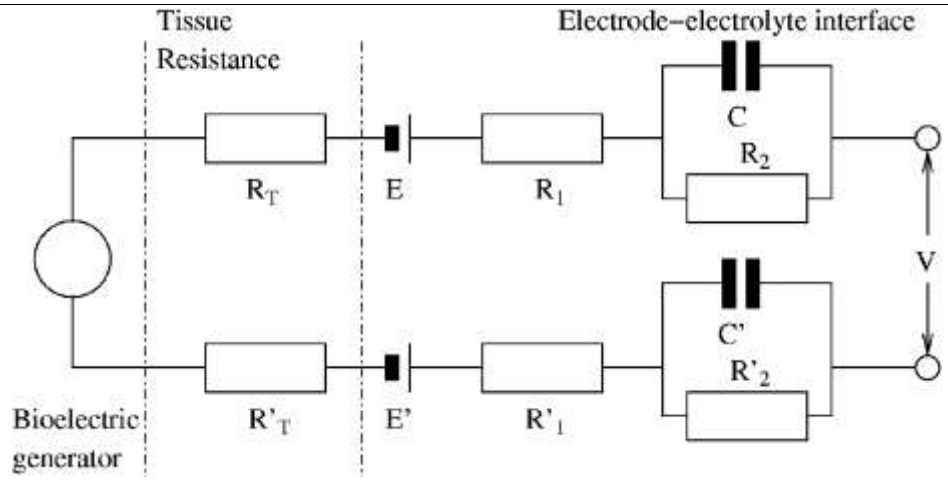


Figure 5: Equivalent circuit for tissue and electrode system

Although C and C', R_1 and R_1', R_2 and R_2' may not be exactly equal (different sites and modes of application on the skin), E should be equal to E' (same type of electrode). Hence V represents the actual difference of ionic potential between the two points on the body from which the ECG is being recorded.

ECG Amplifiers

The peak value of the voltage *V* in Figure 1 (corresponding to the R-wave of the ECG waveform) is typically 1mV. Thus amplification is required in order to increase the signal amplitude for further processing and for display (typically on either a chart recorder or a screen of some sort).

First problem - electric field interference

The ECG voltage *V* is not the only signal found at the input of the amplifier; one major source of interference is the electrical power system. Capacitance between power lines in the wall, floor and ceiling and nearby equipment couples current into the patient, wires and machine. This current flows through the skin-electrode impedances on the way to ground. The capacitance to these power line sources varies with proximity but is of the order of 50pF which corresponds to an impedance of 64MΩ at 50Hz. If the right leg is connected to the common ground of the amplifier through an electrode with contact impedance of, say, 5kΩ, the mains potential of 240V will appear as a ~20mV noise input. This value is well in excess of the ECG signal itself.

The key to extracting the desired ECG signal from the 50Hz noise is the fact that the ECG signal is the difference in potential between a pair of electrodes, i.e. a *differential* voltage. On the other hand, the 50Hz noise voltage is common to each electrode (it appears equally at both the Right Arm and Left Arm input terminals). Rejection of mains interference therefore depends on the use of a *differential amplifier* in the input stage of the ECG machine, the amount of rejection depending on the ability of the amplifier to reject *common-mode* voltages.

Differential Amplifiers

The standard circuit, if more than 60dB of common-mode rejection is required, is the 3 op-amp instrumentation amplifier shown in Figure 6 and this circuit is used in most ECG machines.

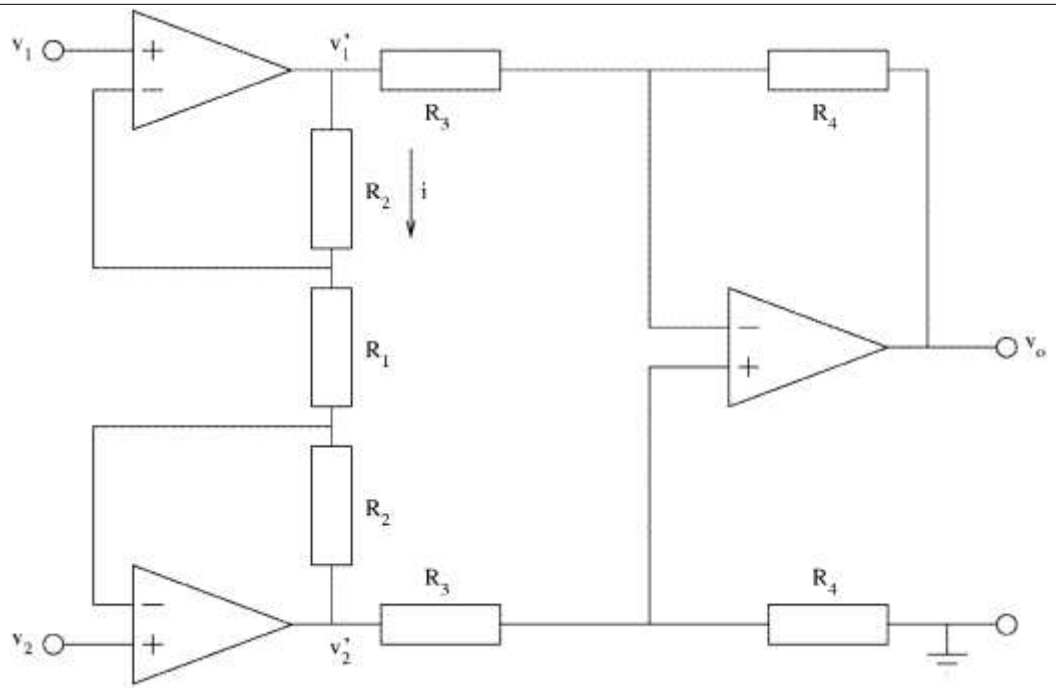


Figure 6: Standard ECG instrumentation amplifier

$$i = \frac{v_1' - v_1}{R_2} = \frac{v_1 - v_2}{R_1} = \frac{v_2 - v_2'}{R_2}$$

$$v_1' = \left(1 + \frac{R_2}{R_1}\right) v_1 - \frac{R_2}{R_1} v_2$$

$$v_2' = \left(1 + \frac{R_2}{R_1}\right) v_2 - \frac{R_2}{R_1} v_1$$

$$v_2' - v_1' = (v_2 - v_1)\left(1 + \frac{2R_2}{R_1}\right)$$

i.e. the first 2 op-amps and associated resistors give a differential gain A_d of $\mathbf{1} + \frac{\boldsymbol{2R_2}}{\boldsymbol{R_1}}$.

If $v_1 = v_2 = v_{cm}$, then $v_1' = v_2' = v_{cm}$ from the second and third equations above; thus the cross-coupled op-amp followers provide a differential gain but pass common-mode signals at unity gain ($A_{cm} = 1$).

The output stage (third op-amp with the R_3 and R_4 resistors) generates a single-ended output and eliminates any remaining common-mode signal. The overall common-mode rejection ratio (CMRR) is given by:

$$\text{CMRR} = \frac{\boldsymbol{A_{d1} \cdot A_{d2}}}{\boldsymbol{A_{cm1} \cdot A_{cm2}}}$$

i.e. the CMRR of the output stage multiplied by the differential gain of the input stage (since $A_{cm1} = 1$).

Second problem - magnetic induction

Current in magnetic fields induces voltage into the loop formed by the patient leads. The induced voltage is proportional to the field strength and the coil area. Reducing this interference requires that the field strength be reduced by moving the equipment and leads (difficult to do in practice) or that the coil area be reduced by twisting the lead wires together all along their length.

Third problem - source impedance unbalance

If there is a severe unbalance in the electrode-skin interface impedances (also known as the contact impedances), the body's common-mode potential will be higher at one input than at the other.

Hence a fraction of the common-mode voltage will be seen as a *differential* voltage and will be amplified by the differential gain of the amplifier (see question on the problem sheet).

The output voltage from the differential amplifier therefore consists of 3 components:

- the *desired* output due to amplification of the differential ECG signal;
- an *unwanted* component of the common-mode signal due to the fact that the common-mode rejection is not infinite;
- an *unwanted* component of the common-mode signal due to source impedance unbalance.

Modification to conventional ECG system

Driven right-leg circuitry

In modern ECG recording systems, the patient is often not grounded. Instead, the right leg electrode is connected as shown in Figure 7 to the output of an auxiliary op-amp. The common-mode voltage on the body is sensed by two averaging resistors R_a, inverted and fed back to the right leg through R_0. This circuit actually drives a very small amount of current (less than 1 μA) into the right leg to equal the displacement currents flowing in the body. The body therefore becomes a summing junction in a feedback loop and the negative feedback from this circuit drives the common-mode voltage to a low value.

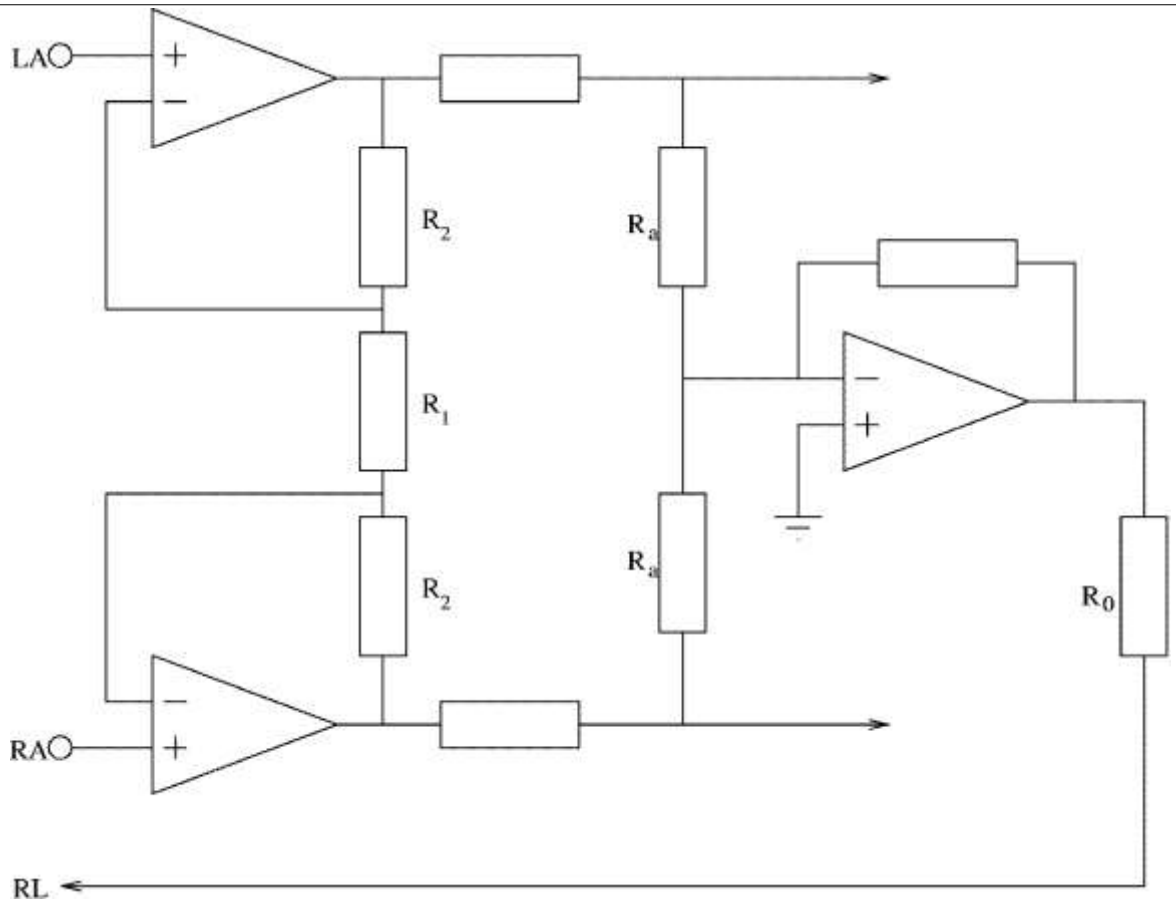


Figure 7: Driven right-leg circuit

The circuit also helps to increase the patient's safety. If an abnormally high voltage should appear between the patient and ground due to electrical leakage or other reasons, the auxiliary op-amp in the right leg circuit saturates. This effectively ungrounds the patient since the amplifier can no longer drive the right leg. The resistance R_0 between the patient and ground is usually several MΩ and is therefore large enough to protect the patient. With a 5 MΩ resistor, for example, and a supply voltage of 10 V, the amplifier will saturate at a current of approximately 2 μA.

Use of ECG for Diagnostic Purposes

As has already been mentioned, diagnostic information can be obtained from the ECG waveform, by analysis of the amplitude and relative timing of the various segments. The ECG is highly informative in the diagnosis of a heart attack (*myocardial infarct*). Insufficient blood supply to the cardiac cells due to a blockage in the coronary arteries (*ischaemic heart condition*) causes S-T segment elevation on the ECG.

Abnormal heart rates (arrhythmias) can be observed and treated; for examples slow rhythms (bradycardia) can be treated with stimulants or a pacemaker, whilst in the case of fast rhythms (tachycardia) depressants can be prescribed. *Ectopic beats* are beats which originate from a region of the heart other than the sino-atrial node (pacemaker cells). An ectopic beat in the ventricle causes an extra R-wave, indicative of a premature ventricular contraction (PVC).

These abnormal conditions are usually identified by one of two means:

- *Ambulatory monitoring* for up to 24 hours of patients who have been identified as being at risk of heart attacks. Data compression techniques (e.g. beat-to-beat interval histograms) are often used although advances in memory technology have considerably reduced the need for these.
- *Exercise stress ECGs* in which the patient is taken close to maximum heart rate by exercising, for example on a treadmill. Changes in the ECG waveform during this process give the cardiologist indications as to the efficiency and capacity of the heart's pumping action. PVCs may only occur when the body is under physical stress, as this makes demands for higher cardiac output. Exercise testing can also be used to assess the effectiveness of therapeutic and surgical treatments.

A more specialised application of ECG analysis is the detection of *foetal distress* prior to and during labour. An additional problem here is the separation of the maternal and foetal ECGs (adaptive filtering techniques are usually required, in addition to careful positioning of the electrodes).

Foetal monitoring is now routine in the developed world, and the timing of dips in the foetal heart rate in relation to the mother's contractions, as well as the amplitude of the dips, are thought to be important indicators of foetal state. Similarly, changes in the level of the S-T segment may give an indication of low oxygen supply to the foetus (*foetal hypoxia*). Other research has concentrated on monitoring foetal heart rate variability, low variabilities having been shown to correlate with foetal distress (a high degree of variability would seem to indicate that the foetus' thermoregulatory control processes are working properly).

All the above applications involve the analysis of the ECG waveform and the extraction of various features of the waveform. In each case, the heart rate provides information of value and needs to be calculated. There are two types of heart rate meters (also known as *cardiotachometers*):

- the *averaging* heart rate meter which calculates the average heart rate from a count or estimate of the number of beats over a period of time;
- the *beat-to-beat* heart rate meter which computes the reciprocal of the time interval between two consecutive heart beats and updates the information with each heart beat.

Heart rate meters

The easiest way to obtain the heart rate (usually given as beats per minute) is to count some identifying feature of the ECG. The most easily distinguished feature of the ECG is clearly the QRS complex, which is a sharp spike.

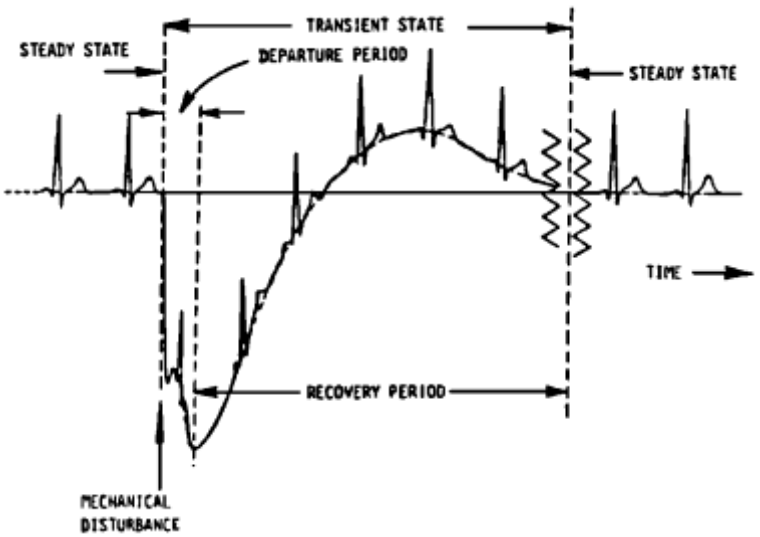


Figure 8: Effect of motion on ECG recording

There are 3 main problems in detecting the QRS complex:

- artefacts due to electrode motion (as shown in Figure 8)
- baseline wander (mainly due to breathing)
- T-waves with high frequency content

These problems can be considerably reduced by passing the ECG signal through a band-pass filter. Spectrum analysis of the ECG signal reveals that most of the frequencies present in the QRS complex lie near 20 Hz and hence a filter with a pass-band of, say, 10 to 40 Hz should maximise the QRS energy.

The circuit of Figure 9, which would be equally applicable to averaging or beat-to-beat meters, could be used to generate pulses coincident with the R-wave of the ECG waveform. The threshold circuit triggers the

pulse generator when the band-pass filter output exceeds a preset threshold level. The pulse width should be greater than the Q-S interval so that only one pulse can be generated per QRS complex. The pulse train can then be fed to either of the circuits of Figure 10 or Figure 11, depending on the type of heart rate meter required.

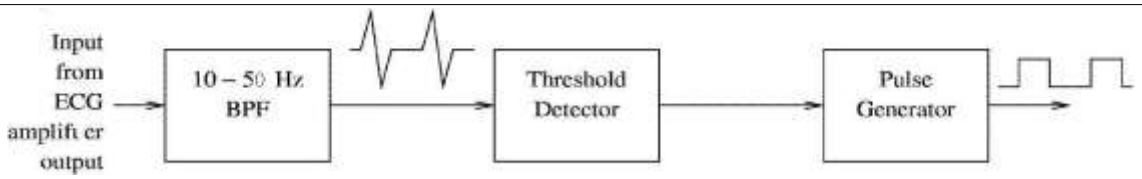


Figure 9: R-wave pulse generator

Averaging heart rate meter

An indication of the average heart rate can be obtained by feeding the R-wave pulses to the input of a low-pass filter, which determines the "average power" of the pulse train (and hence the pulse repetition frequency since the pulses are of constant width and amplitude). The higher the heart rate is, the larger the charge which builds up on the capacitor. The resistor also bleeds off some of the charge from the capacitor when there are no pulses at the input, giving the sawtooth waveform shown in Figure 10. The time constant of the R-C circuit should be several beats long (between 5 and 15 seconds, typically) to minimise the output ripple.

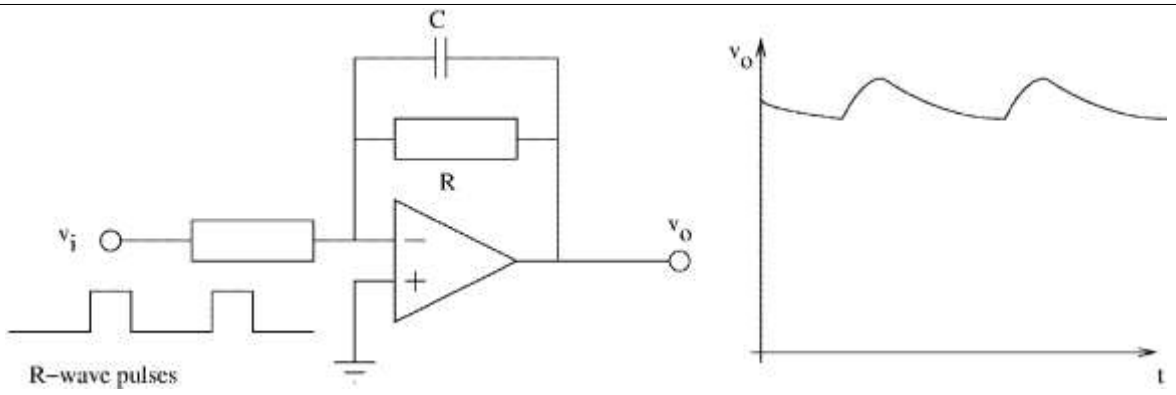


Figure 10: Low-pass filter for pulse counting in averaging heart rate meter

Beat-to-beat heart rate meter

This computes the time between each beat and inverts it, giving an instantaneous heart rate. The circuit and timing diagrams for such a meter which uses digital techniques to compute the heart rate to 12-bit accuracy are shown in Figure 11.

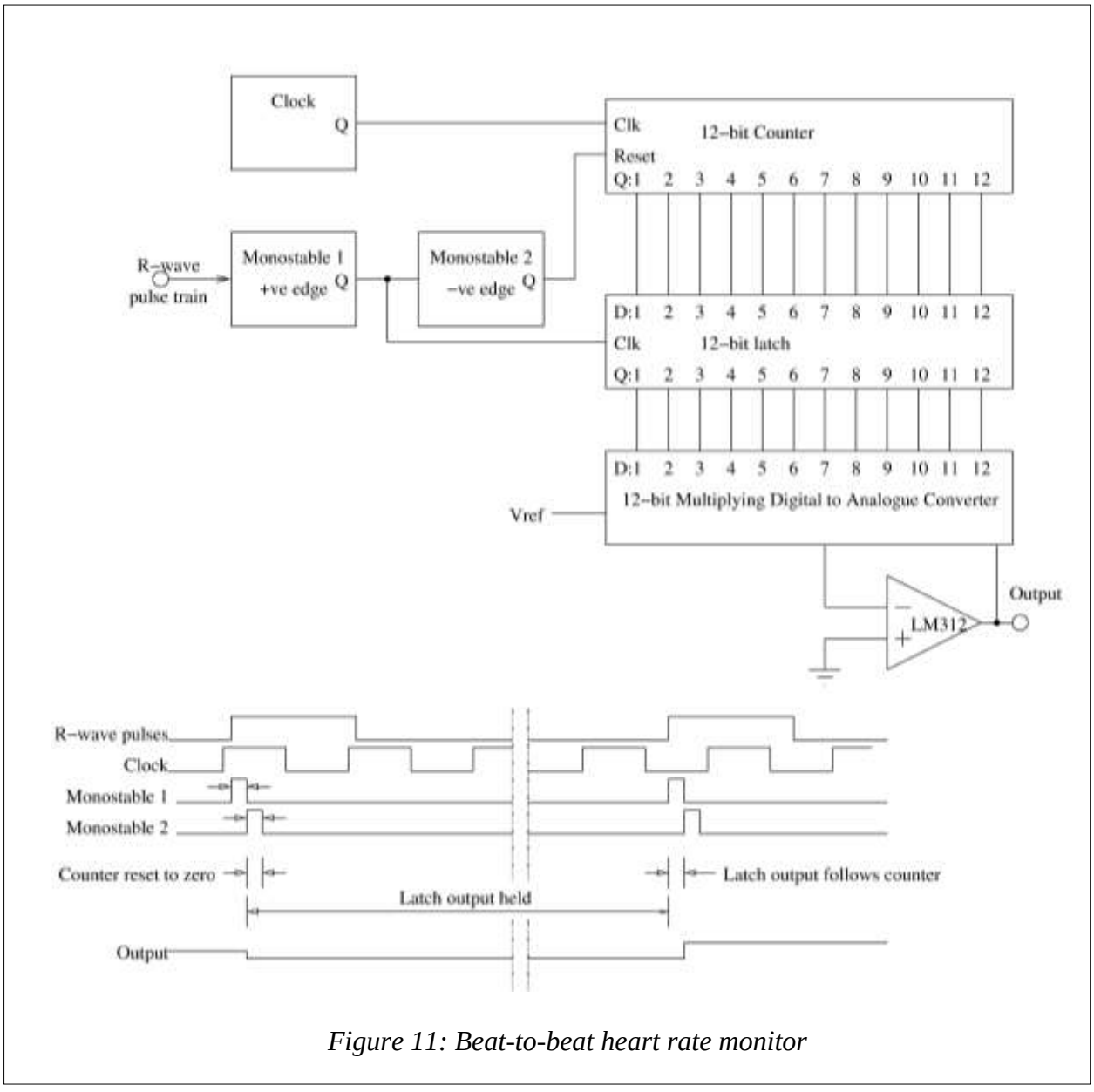


Figure 11: Beat-to-beat heart rate monitor

The leading edge of the R-wave pulse (from the circuit of Figure 9) triggers the first mono-stable, giving a pulse of 10 μs width, for example. The falling edge of this 10 μs pulse in turn triggers the second monostable.

The counter counts the clock pulses between consecutive R-wave pulses. Monostable 1 causes the contents of the counter to be loaded into the latch; monostable 2 then resets the counter to zero in readiness for the next cycle. The maximum clock frequency is set by the length of the monostable pulses as the load and reset pulses must take less than one clock cycle in order to maintain the specified accuracy.

The outputs of the latch are taken to a multiplying digital-to-analogue converter (DAC). The output voltage is proportional to the reference voltage multiplied by the reciprocal of the number of clock pulses in one period of the input signal.

Heart rate variability

Under resting conditions, the ECG of a healthy individual exhibits periodic variation in the R-R interval. This phenomenon, known as respiratory sinus arrhythmia (RSA), fluctuates with the phase of respiration – acceleration during inspiration, deceleration during expiration. During expiration, the vagus nerve is stimulated, which slows down the heart rate (the right vagus innervates the sinoatrial node and acts on it). During inspiration, the vagus nerve is not stimulated. Thus, the breathing rate can be estimated from the heart-rate of a patient, as investigated in a later part of the course.

Electroencephalography

(Unexamined)

In addition to ECG measurements, biopotentials can also be recorded from the brain (electroencephalography - EEG) or from muscles (electromyography - EMG). Electroencephalography has found many clinical uses, from the investigation of epileptic fits to sleep studies or the diagnosis of brain death. The EEG is the summation of neural depolarisations in the brain due to stimuli from the five senses as well as from normal brain activity. On the surface of the brain, these potentials are of the order of 10 mV; the amplitude of the EEG signal recorded with scalp electrodes, however, is 100 μV, at most, because of the attenuation caused by the skull.

EEG instrumentation is much the same as that used for recording the ECG, except that circuit design is even more critical because of the lower amplitude of the EEG signal. The frequency response of EEG differential amplifiers usually extends from 0.1 to 100 Hz.

EEG electrodes

The EEG is measured with *Ag-AgCl* electrodes which are placed in standard positions on the skull, at regular intervals along three lines: one from the nose to the back of the head, one from ear to ear and one around the circumference of the skull (see Figure 12).

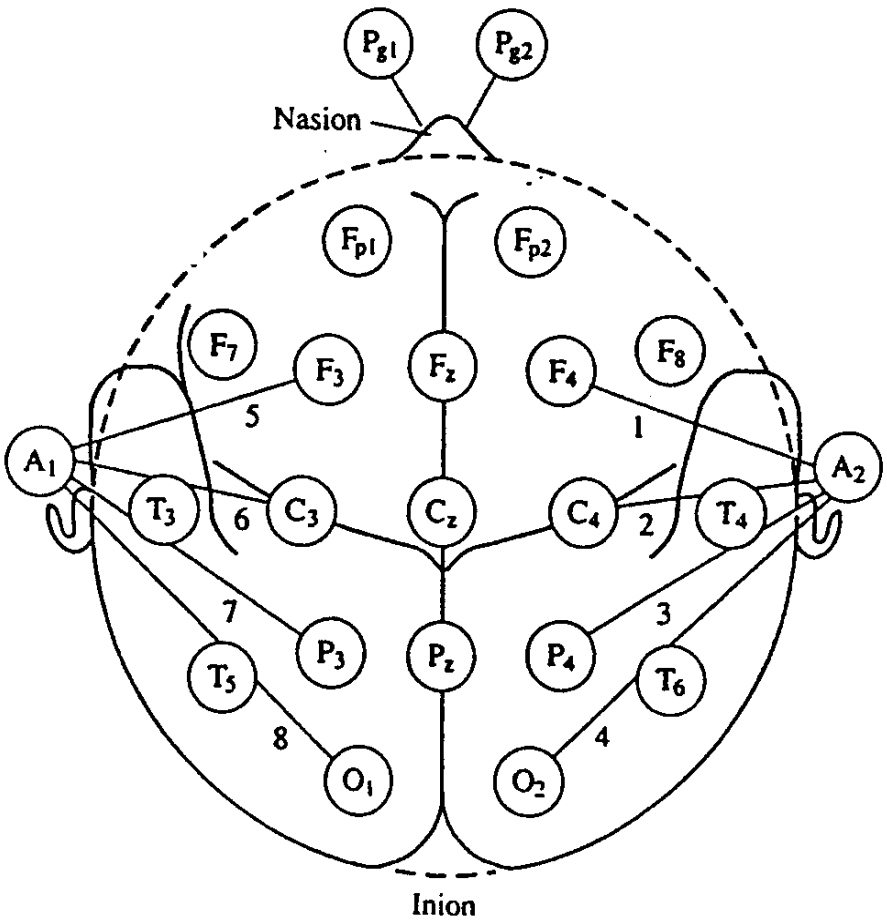


Figure 12: EEG electrode positions on the scalp

Characteristics of the EEG

The frequency content of the EEG varies with the state of alertness and mental activity. To assist in EEG analysis, the normal EEG range of 0.5 to 30 Hz has been subdivided into five bands (note, however, that there is some degree of variation in the exact cut-offs from one system to another):

Delta	δ	0.5 - 4 Hz
Theta	θ	4 - 8 Hz
Alpha	α	8 - 13 Hz
Beta	β	13 - 22 Hz
Gamma	γ	22 - 30 Hz

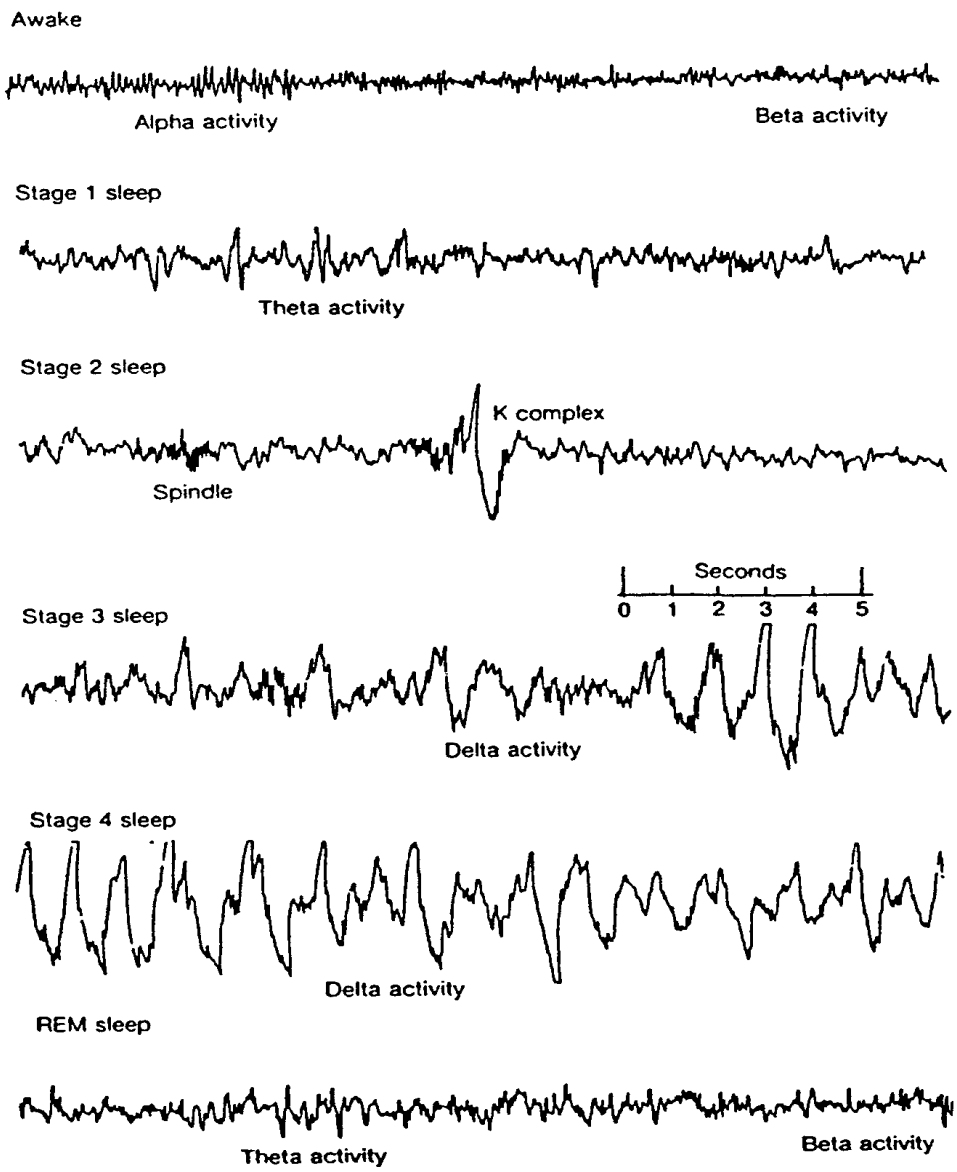


Figure 13: EEG activity for various sleep stages

What does the EEG look like in different brain states? Figure 13 shows the EEG through various stages (or depths) of sleep. The lower trace shows that, if the subject is dreaming, the EEG exhibits rapid, low-voltage waves that resemble those obtained in alert subjects (rapid-eye-movement or REM sleep). During the transition from REM to non-dreaming (deep) sleep, the EEG sometimes exhibits bursts of alpha-like activity, called *spindles*. These are periodic waves which have a frequency between 12 and 14 Hz and an amplitude of around 10 μV.

They last between 0.5 and 1 second but no-one has yet come up with a satisfactory explanation for their origin.

Diagnostic uses of the EEG

The main clinical purpose of the EEG is to help physicians diagnose disease. The pathological states or diseases usually diagnosed using the EEG are brain death, epilepsy and sleep disorders. The sustained absence of EEG signals is a clinical measure of brain death and the EEG can also be used as one of the indicators for deciding to carry out an organ transplant.

The pathological EEG during tonic-clonic seizures (formerly known as *grand mal* epilepsy) is characterised by high-magnitude synchronous waves of about 10 Hz. Absence seizures (formerly known as *petit mal* epilepsy) exhibit a spike and slow-wave pattern with a repetition frequency of about 3 Hz. Figure 14 shows the EEG waveforms recorded from different sites during one such seizure.

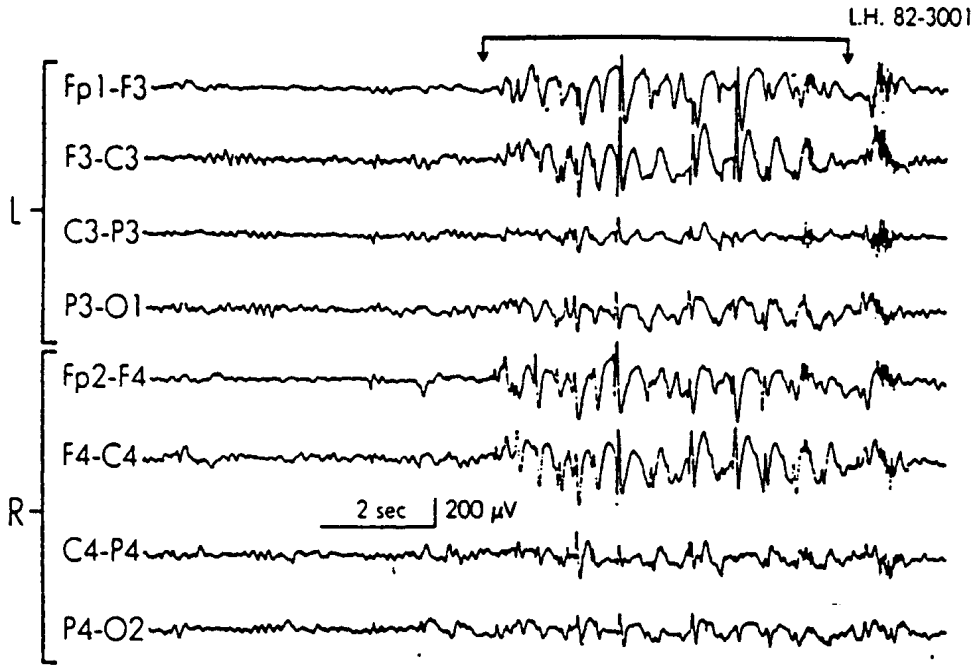


Figure 14: EEG during an epileptic seizure

Evoked Potentials

This is a technique whereby a stimulus, such as a light flash or loud click, is applied to the body's sensory system and the change in the EEG signal recorded from a particular area of the brain. Normal EEG activity, however, masks the brain's response to a single stimulus; *repetitive* stimuli have to be used and the evoked response is distinguished from the background activity by using the technique of *signal averaging*.