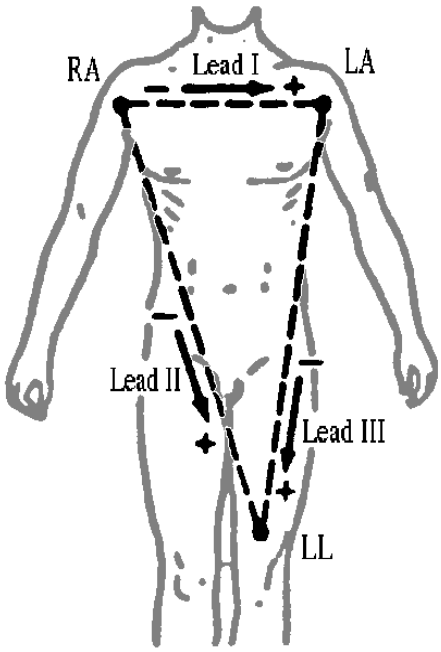
ECG Instrumentation

In order to record the ECG, we need a transducer capable of converting the ionic potentials generated within the body into electronic potentials which 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 9 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 9: Typical electrode placement

Another 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 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 dialectric constitutes a form a 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.

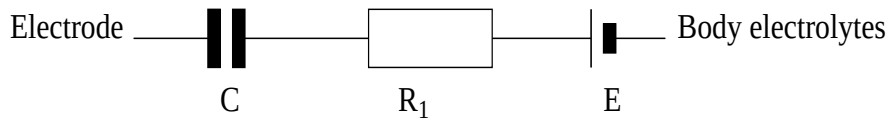


Figure 10: Simple R−C model of electrode−electrolyte interface

The series model in Figure 10 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 11) 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 = 2kΩ, R_2=10kΩ and *C*=10μF.)

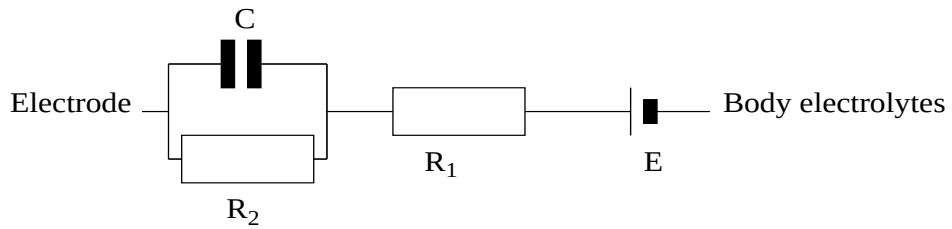


Figure 11: Equivalent circuit of the Ag−AgCl interface

Movement artefact

If the electrode is moved with respect to the elctrolyte, 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 are in contact with an electrolyte and one moves while the other remains stationary, a potential difference appears between the two during this motion. This potential is referred to as *moverment artefact* and can be a serious cause of interference in the measurement of ECG (or any other biopotential).

Overall equivalent circuit

Using the simple model of the electrode−electrolyte interface of Figure 11 as well as the even simpler model previously developed for the electrical activity of the heart, we can put together an equivalent circuit which models the impedance seen by the input stage of an ECG system. This overall equivalent circuit is shown in Figure 12.

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

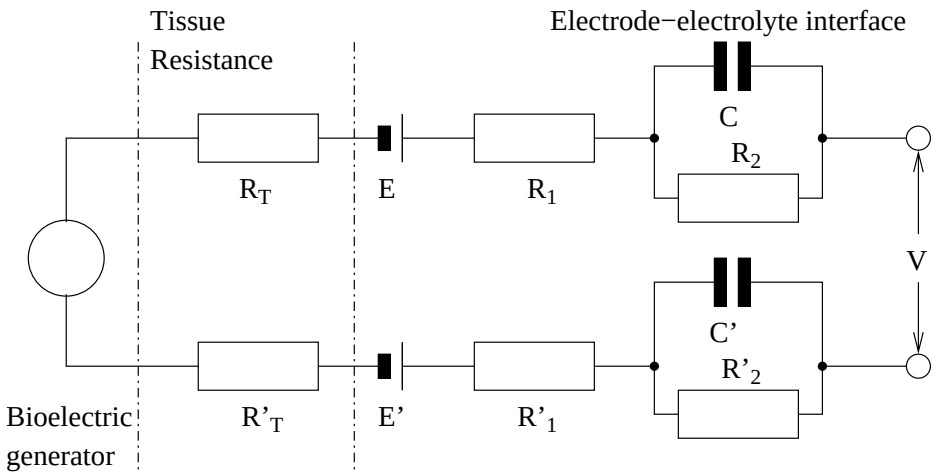


Figure 12: Equivalent circuit for tissue and electrode system

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 12 (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, *ie* 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

You have already met these amplifiers in the Core Course. The standard circuit, if more than 60dB of common−mode rejection is required, is the 3 op−amp instrumentation amplifier shown in Figure 13 and this circuit is used in most ECG machines.

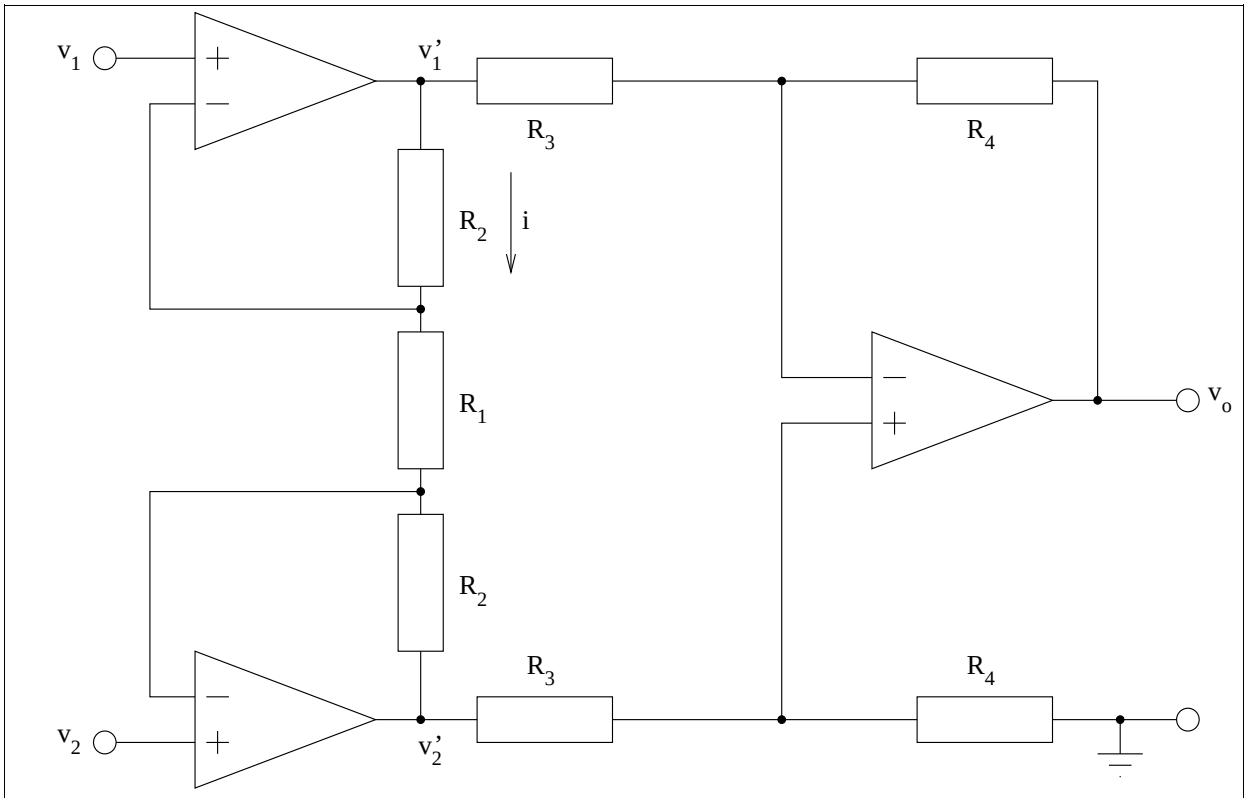


Figure 13: Standard ECG instrumentation amplifier

Revision

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

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

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

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

ie the first two op−amps and associated resistors give a differential gain A_d of $1 + \frac{2R_2}{R_1}$.

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 followers provide a differential gain but pass common−mode signals at unity gain (*ie* $A_{cm} = 1$).

The output stage generates a single−ended output and eliminates any remaining common−mode signal. The overall common−mode rejection ratio (CMRR) is given by:

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

ie 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 impedance), 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).

Hence the output voltage from the differential amplifier 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.

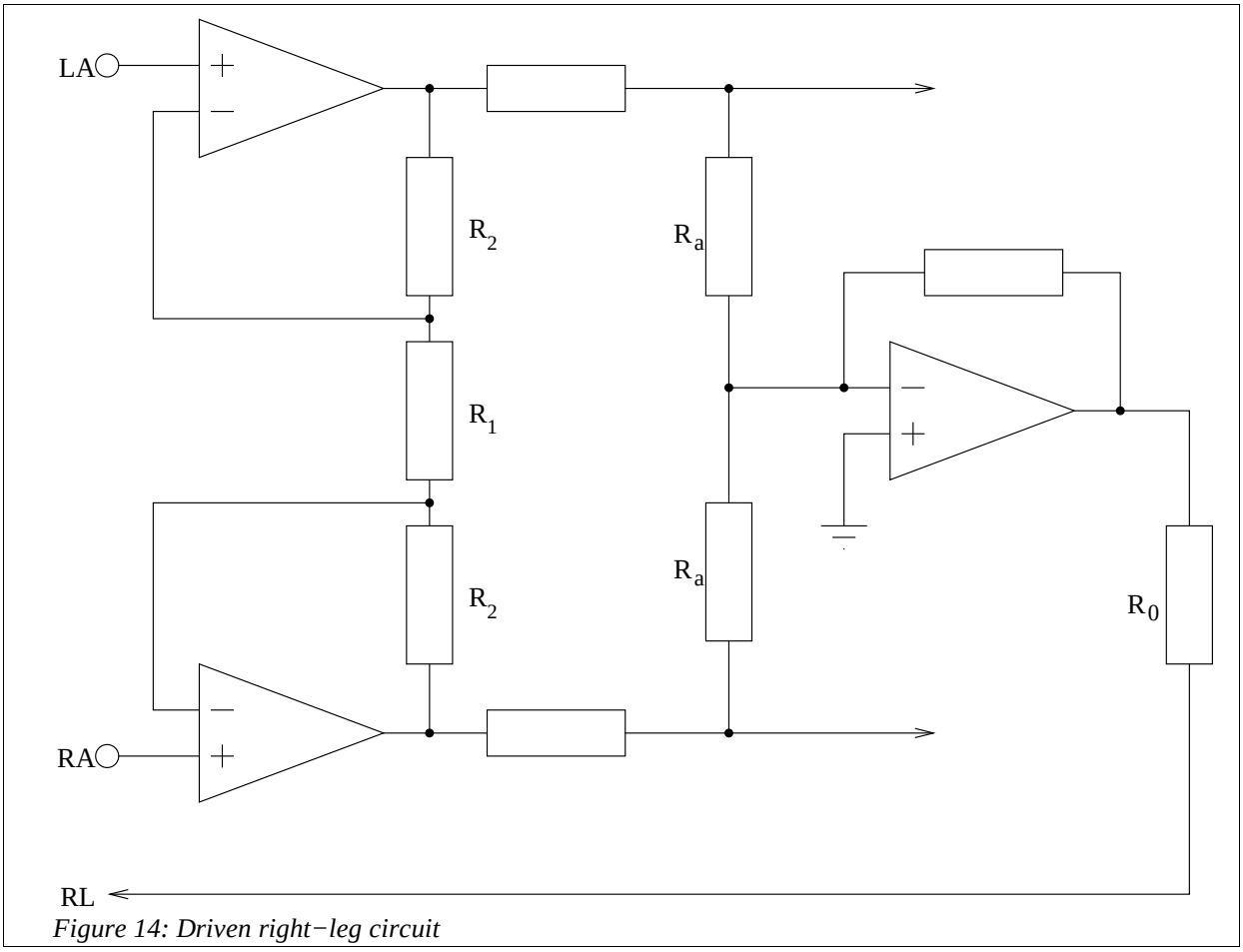


Figure 14: Driven right−leg circuit

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 14 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.

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 means, 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 examples, 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. In general, cardiac muscle damage, or infarcts, are correlated with loss of amplitude. 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 SA node. 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 (*eg* beat−to−beat interval histograms) are often used although advances in memory technology is reducing the need for these.
- *Exercise stress ECGs* in which the patient is taken close to maximum heart rate by exercising, for example on a tread mill. 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 ecletrodes).

Although foetal monitoring may add complexity to the (already stressful) process of giving birth, the timing of dips in the foetal heart rate in relation to contractions in the mother is known to be an indicator of foetal distress. Similarly, changes in the shape 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 also 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. It should be obvious by now that the most easily distinguished feature of the ECG is the QRS complex, which is a sharp spike.

There are 3 main problems in detecting the QRS complex:

- artefacts due to electrode motion (as shown in Figure 15)

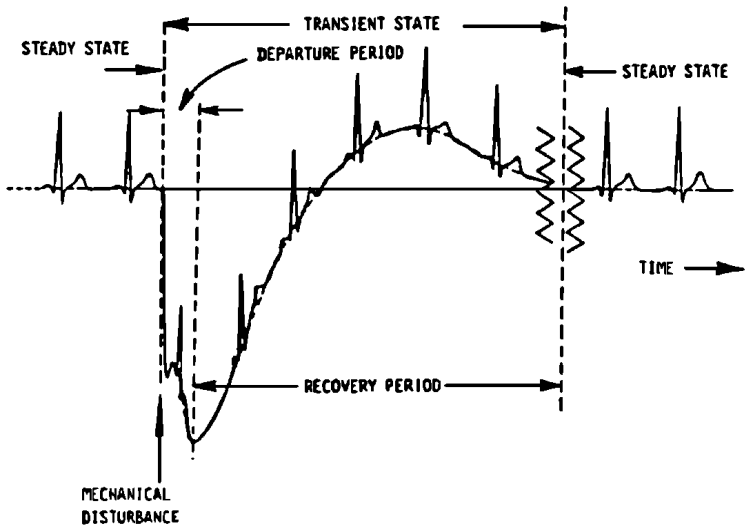


Figure 15: Effect of motion on ECG recording

- 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 band−pass of, say, 10 to 40 Hz would maximise the QRS energy.

The circuit of Figure 16, which would be equally applicable to averaging or beat−to−beat meters, could be used to generate pules coincident with the R−wave of the ECG waveform. The threshold circuit triggers the pulse generator when the band−pass filter outputs 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 17 or Figure 18, depending on the type of heart rate meter required.

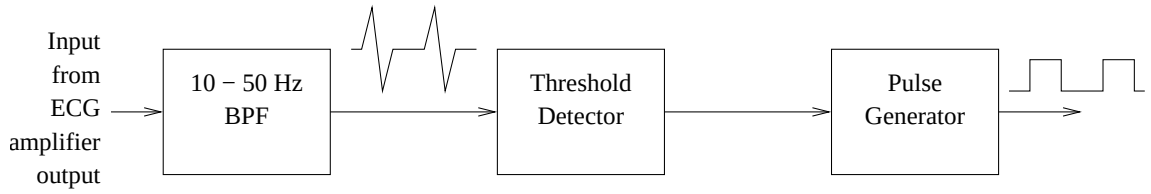


Figure 16: R−wave pulse generator

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, 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 17. The time constant of the R−C circuit should be several beats long (between 5 and 15 seconds, typically) to minimise the output ripple.

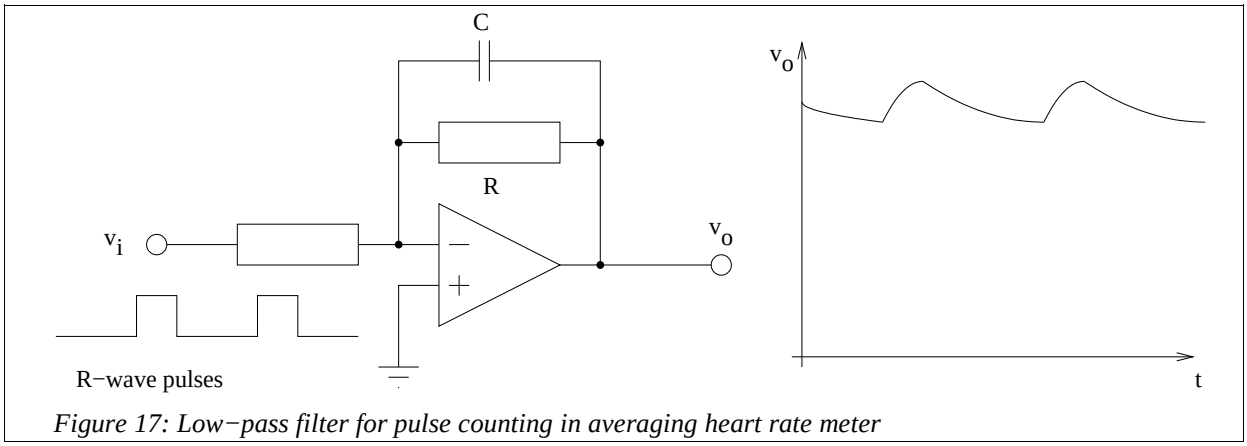


Figure 17: Low−pass filter for pulse counting in averaging heart rate meter

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 18.

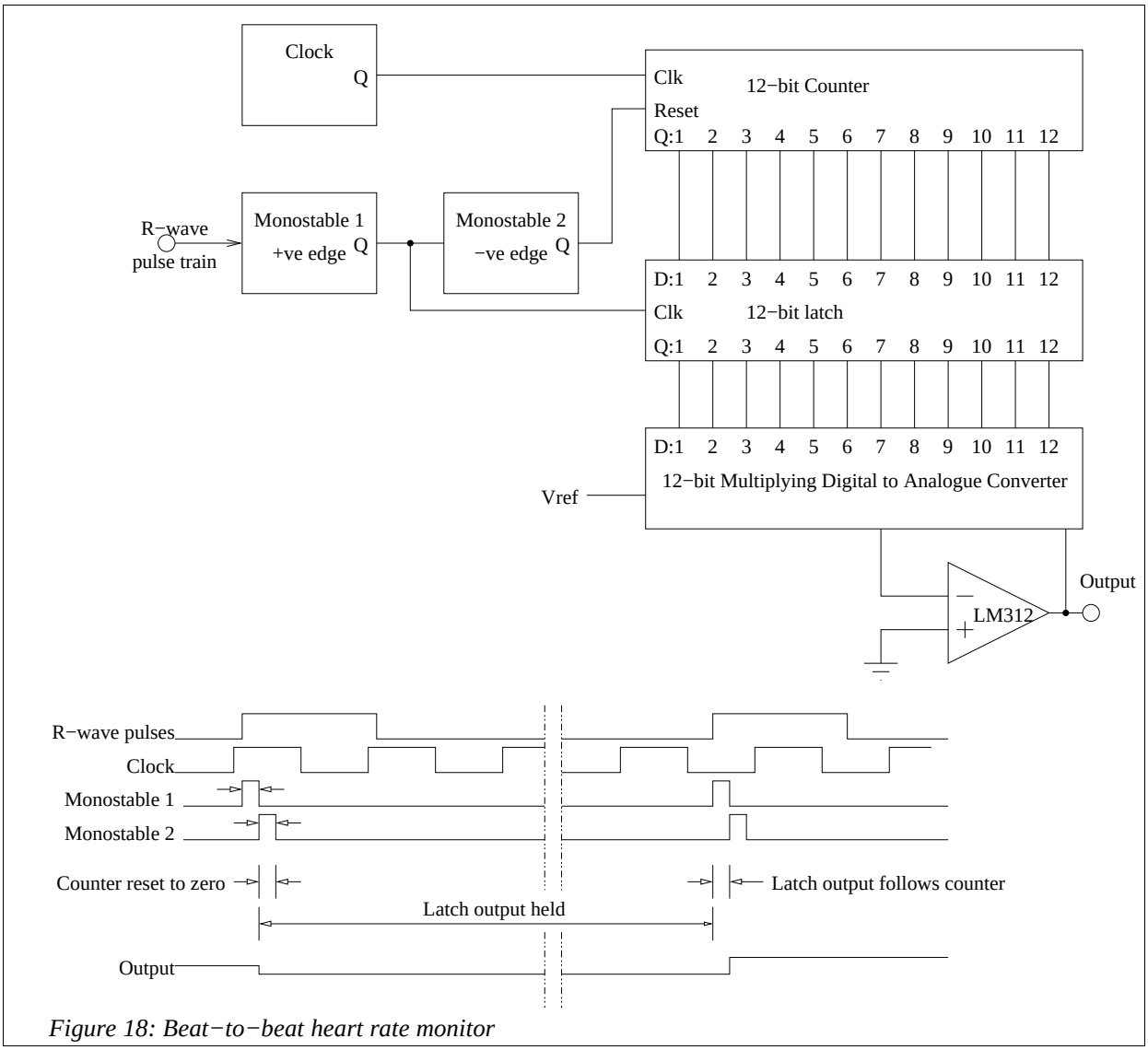


Figure 18: Beat−to−beat heart rate monitor

The leading edge of the R−wave pulse (from the circuit of Figure 16) triggers the first mono−stable, giving a pulse of 10 μs width in the examples given here. 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 ino 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), which is connected as a divider in the feedback loop of the LM312 op−amp. The output voltage is proportional to the reference voltage multiplied by the reciprocal of the numbers of clock pulses in one period of the input signal.