Respiration

Respiration and breathing are words that are used interchangeably in biomedical engineering texts. Strictly speaking, respiration takes place at the cellular level and is the chemical process of releasing energy from (complex) organic compounds. It can be both aerobic (using oxygen) and anaerobic (not using oxygen) although only the former can be sustained for significant periods in most animals. Breathing is defined as the ventilation movements that are needed in some larger animals so that efficient gas exchange (the movement of oxygen into an organism and carbon dioxide out of an organism) can take place. It involves airways, ribs, intercostal muscles, the diaphragm and the lungs. These are shown in schematic representation in Fig 15 below.

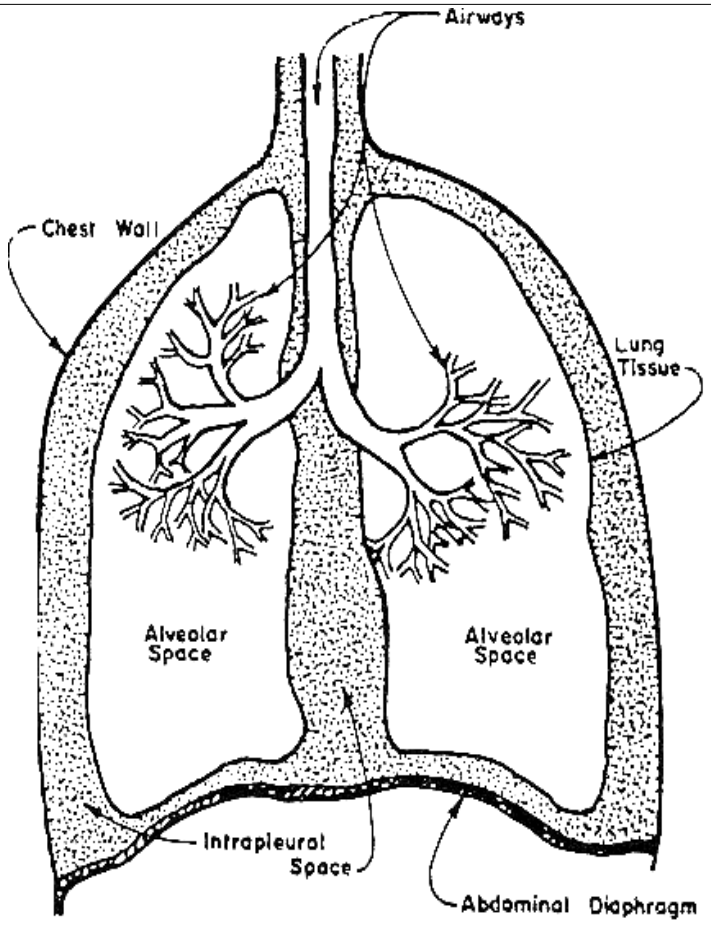


Figure 15: Schematic representation of the respiratory system

The purpose of breathing is therefore to enable gas exchange to take place. Gases are exchanged between red blood cells and air in very small gas-filled cavities called *alveoli.* There are approximately 300 million alveoli in the adult lung with a total alveolar surface area of 50 to 100 m^2. The alveoli have very thin walls ensuring that the numerous small blood vessels (capillaries) that surround them are very close to the inhaled air, maximising the potential for gas exchange. When someone breathes in, air enters the lungs and eventually the alveoli (via the oral and nasal cavities, the trachea, the bronchial tree and the bronchioles); when someone exhales, air is forced out of the lungs and so the total volume of air in the lungs varies with inhalation and exhalation.

Measuring Respiration

There are two main parameters of interest: the frequency (or rate) of breaths and the depth of these breaths. Unlike the heart, the lungs do not generate biopotentials which can easily be detected and so alternative methods must be designed. Two examples of these are the measurement of air flow with an appropriate sensor and *plethysmographic* techniques. (The term "plethysmography" refers, in general, to measurement of the volume or change in volume of a portion of the body.)

Measurement of airflow

Air flow can be detected in a variety of ways but these require a mask over the nose or tube in the mouth. For example, a heated element can be inserted in a mouth tube. As air flows over the element, it responds by cooling. This in turn means that a greater current is needed to maintain the element at a constant temperature. This current is

proportional to the airflow through the tube. The flow signal can also be integrated in order to determine the volume that has passed through the tube. The obstructive nature of the tube makes it impossible to use this technique for routine monitoring of respiration in patients, however.

Inductance plethysmography

The variations in chest volume can be measured using an inductive plethysmograph, a device which employs a pair of wires, each attached in a zig-zag pattern to its own highly compliant belt. One belt is placed around the ribcage and the other around the abdomen, so that each wire forms a single loop, and the pair is excited by a low-level radio-frequency signal. Changes in a loop's cross-sectional area produce corresponding changes in self-inductance. After demodulation, an output is obtained proportional to the local cross-sectional area of the segment of the chest wall that is encircled by the loop.

Electrical impedance plethysmography

It is simple to attach electrodes to a body segment and measure the resulting electrical impedance Z of the tissue within the measuring volume. Since the resistivity ρ of the lungs increases as air enters the lungs during inspiration, the impedance of the lung tissue changes accordingly (since $Z = \rho l/A$ for a cylinder of length l and cross-sectional area A).

Choice of frequency for electrical impedance measurements

A current as high as 1 mA is required to achieve adequate signal-to-noise ratio. At low frequencies, a current of 1 mA causes an unpleasant shock. However, the amplitude of the current required for perception

increases with frequency. Therefore, frequencies above 20 kHz are used to avoid perception of the current.

The skin-electrode contact impedance decreases by a factor of about 100 as the frequency is increased from a few Hz up to 100 kHz. High frequencies are therefore used to decrease both the skin-electrode impedance and the undesirable changes in this impedance that result from movement of the patient. If a frequency much higher than 100 kHz were used however, the low impedances of the stray capacitances would make the design of the instrument very difficult. Hence a frequency between 20 and 100 kHz is usually adopted.

Number of electrodes

For reasons of economy and ease of use, some impedance plethysmographs use two electrodes. The current *i* flows through the same electrodes that are used to measure the voltage *v*. The movement of the chest cavity, however, causes artefactual changes in the skin-electrode contact impedance, which are superimposed on the desired impedance changes caused by the changes in resistivity of the lungs (and at the same frequency). Because the skin-electrode contact impedances are in series with the desired tissue impedance, it is impossible to separate the two and determine the actual changes in electrical impedance of the lungs.

To solve these problems, the four-electrode impedance plethysmograph shown in Figure 16 is used. The current flows through the two outer electrodes and voltage is sensed between two inner electrodes. With an infinite input-impedance voltage-sensing amplifier, errors caused by the variations in skin-electrode impedance are eliminated.

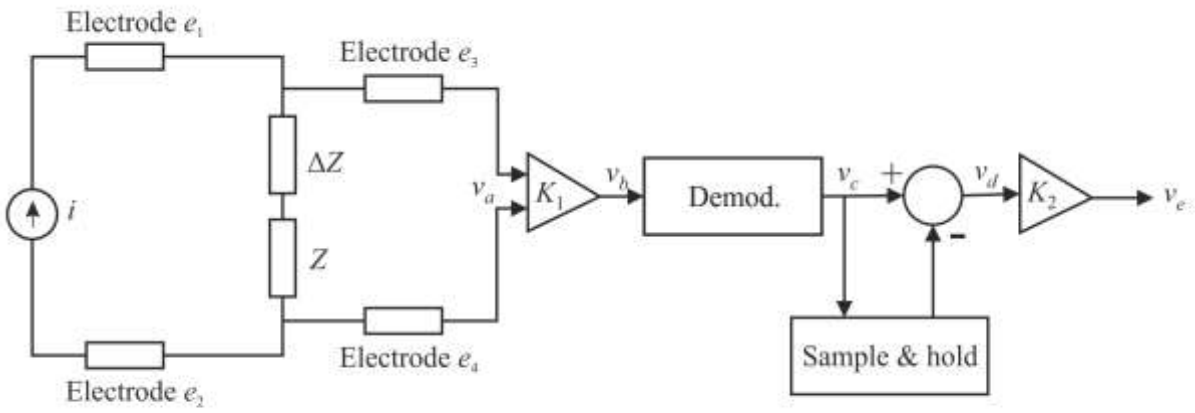


Figure 16: 4-electrode impedance plethysmograph. Current is injected through two outer electrodes, and voltage is sensed between two inner electrodes.

4-electrode electrical impedance plethysmograph

Define impedances Z_1-Z_4 as being associated with electrodes e_1-e_4. Ideally, the current source *i* causes a *constant* current to flow through *Z*, regardless of changes in ΔZ (the desired change in impedance due to the changes in resistivity of the lungs with breathing) or any other impedances.

Electrodes Z_3 and Z_4 are used to sense the voltage. Ideally, the voltage amplifier has an input impedance sufficiently high that hardly any current flows through Z_3 and Z_4.

As with the ECG instrumentation seen earlier, there are also common-mode impedances to ground. These impedances can convert common-voltages to erroneous differential voltages unless the circuitry is carefully designed.

The output of the amplifier is a large 100 kHz signal, which is amplitude-modulated by a small amount $i\Delta Z$ at the breathing frequency. The $i\Delta Z$ component may be demodulated by any AM detector, for example a diode followed by a low-pass filter.

The demodulator produces an output $Z + \Delta Z$. Often ΔZ, which contains the useful information, is only 1/1000th of Z. A balancing voltage is therefore subtracted from the demodulated signal to give ΔZ. The balancing voltage is derived from the mean of the signal over a short window.

When the patient moves, however, Z changes by an amount much larger than ΔZ. An automatic reset system is needed to cope with this. Whenever ΔZ saturates its amplifier, a sample-and-hold circuit makes $Z + \Delta Z$, which momentarily resets ΔZ to zero. The sudden vertical-reset trace is easily distinguished from the slower-changing physiological data.

Electrical impedance plethysmography - conclusion

The advantages of electrical impedance plethysmography are that it is non-invasive and relatively simple to use. Its disadvantages are that it is not sufficiently accurate to obtain an accurate measurement of volume changes, except with individual patient calibration (which is not practical). For this reason, it is often used simply to extract breathing rate by peak detection on the filtered ΔZ signal (see Figure 17 below). In this application, it is often known as electrical impedance *pneumography* (movement of the chest) rather electrical impedance plethysmography since there is no attempt to measure volume change.

Of course, if only the rate (i.e. frequency) of respiration is required, then a two-electrode system is perfectly adequate. It is the output of such a system which is shown in Figure 17 from which it can be seen that the frequency of the patient's breaths can be determined from the trace. (Note also the smaller cardiac-synchronous component, at a higher frequency, which arises out of the movement of blood in the thorax during the cardiac cycle.)

Figure 17: An impedance plethysmography trace showing 10 breathing cycles

Diagnostic use of respiration information

Respiration information is extremely important in a number of clinical areas. In both the very young and the elderly, unpredictable cessations of breathing (*apnoeas*) occur which, if undetected, can result in death.

Respiration information is also useful for diagnosing two other complaints: obstuctive sleep apnoea (OSA) and Cheyne-Stokes respiration. During sleep, an OSA patient stops breathing for a period of about 30-60 seconds before biochemical "reflexes" take over, resulting in the patient waking up and starting to breathe. The patient then goes back to sleep but, about 30-60 seconds later, stops breathing again and so the cycle continues throughout the night.

Deriving breathing rate from cardiosynchronous waveforms

The ECG and the PPG from pulse oximeters (see later) are both cardiosynchronous waveforms. As described in the first lecture, these waveforms are modulated by breathing: (i) the breathing rate can therefore be estimated by examining the period of the sinusoidal variation of the heart rate (from RSA).

Additionally, (ii) there exists amplitude modulation of the ECG and PPG; thus, breathing rate can also be estimated by examining the period of the sinusoidal variation of the envelope of these waveforms.

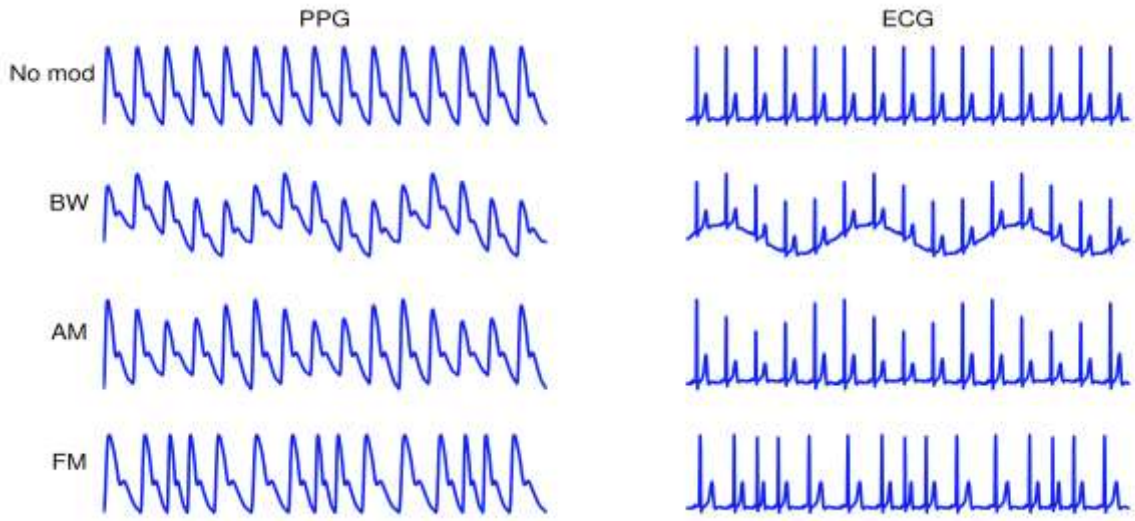
Finally, (iii) there exists a variation in the baseline of the ECG and PPG with breathing; the breathing rate can thus be estimated period by identifying the period of this sinusoidal variation.

These effects (i)-(iii) are labelled as follows in the figure below:

(i) Frequency modulation (FM), caused by RSA;

(ii) Amplitude modulation (AM); and

(iii) Baseline wander (BW).



Electrical Safety

Electrical safety is a very important issue in biomedical instrumentation. The physiological effect of an electrical current, depending on its amplitude, frequency, duration and which part of the body it acts on, can be hazardous and sometimes fatal.

Physiological effects of electricity

Electrolysis

Electrolysis takes place when direct current is passed through any medium which contains free ions. If two electrodes are placed on the skin and a direct current is passed, the positively charged ions in the tissue underneath will migrate to the negative electrode and the negatively changed ions to the positive electrode. This will disturb normal cell functions and small ulcers will be formed beneath the electrodes, for example if a d.c. current of 100 μA is applied for a few minutes.

The recommendations for safety standards for patient-connected equipment drawn up by the International Electrotechnical Commission (IEC601) defines a ‘direct current’ as a current with frequency less than 0.1Hz and limits the direct current that can flow between electrodes to 10 μA.

Neural stimulation

There is a normal potential difference across a nerve membrane. An action potential will occur if this potential difference is reversed for a certain period of time. Electrical current is able to reverse this potential difference and hence cause neural stimulation. As a result a pain will be

felt if a sensory nerve has been stimulated, or a muscle will contract if a motor nerve has been stimulated.

The major hazards are the stimulation of heart and skeletal muscle, either directly or by the stimulation of motor nerves. The co-ordinated pumping activity of the heart can be disrupted by currents that pass through the heart (ventricular fibrillation), which, if not rectified, will almost certainly cause death. Neural stimulation is the major hazard from which patients connected to biomedical instrumentation need to be protected.

The effect of neural stimulation by a hazard source (e.g. a high voltage) is dependent largely on the local conditions where the voltage is applied. If this high voltage is applied to dry skin, the high impedance of the dry skin (10 – 100 kΩ) results in a low current flow and consequently a low hazard. But if the skin is wet, the skin impedance could drop to as low as 1 kΩ, which can result in a much more hazardous current. Some studies have shown that the threshold of electrical current for ventricular fibrillation at 50 Hz for an average-sized human varies from about 75 mA to 400 mA. IEC601 limits the current flow through contact to the skin to 0.5 mA with a single fault in the equipment.

If a current is passed through two electrodes which are attached to, say, the arms, the current will be distributed throughout the body and only a fraction will actually flow through the heart. But, if directly applied to the heart wall, a current as small as 100 μA can cause ventricular fibrillation. Therefor*e* for equipment that can be connected directly to the heart *(NB: invasive* monitoring), IEC601 limits the current to 10 μA under normal operating conditions and 50 μA with a single fault. An

example of such invasive monitoring is a cardiac catheter for measuring beat-by-beat Left Ventricular Blood Pressure.

Note that neural stimulation becomes increasingly difficult at frequencies above 1 kHz, hence the choice of frequency in electrical impedance plethysmography.

Tissue heating

The major effect of high-frequency electrical currents is that of heating. The local effect of heating depends not only on the current amplitude, frequency and the time of exposure, but also on factors such as the depth of the tissue and the blood flow.

Therapeutic uses of electrical shock

Electrical shock is also deliberately applied to patients in clinical practice for therapeutic purposes. Most of these applications make use of the neural stimulation effect, for example pacemakers, defibrillators (to stop ventricular fibrillation), or Implantable Stimulators for Neuromuscular Control (to help paralysed patients regain some neuromuscular control). Defibrillators can either be external to the body or implanted (Implantable Cardiac Defibrillators – ICDs).