Plethysmography

The literal meaning of Plethysmography is “Recording of instantaneous volume of an object/ body part”. It has, however become a synonym for “indirect assessment of blood volume changes in any part of the body from changes in the electrical parameter like impedance of the body segment”.

Photoelectric Plethysmography

Photo-plethysmography (PPG) provides a qualitative assessment of changes in cutaneous blood volume. It is based on the determination of the optical properties of a selected skin area. The PPG electrode is composed of an infrared light emitting diode and a photosensor. Light transmitted into the skin is scattered and absorbed by tissue in the illuminated field. Blood, being more opaque than surrounding tissue, attenuates the reflected light in greater proportion. The intensity of reflected light changes with tissue blood density. The measurement is localized to the microvasculature of the cutaneous layer underlying the electrode. The instrument may be used to measure arterial pulsations or transient venous volumes.
To measure changes in venous volume, the voltage signal generated in the photo sensor is amplified through a DC coupled circuit, which dampens higher frequency arterial pulsations, leaving a low frequency response with a longer time constant. This produces a relatively stable tracing, which corresponds to blood density in the underlying tissue. The PPG is able to monitor alterations in tissue venous filling as the patient is asked to perform certain manoeuvres. There are two different types of probes that can be used for PPG measurements:

1. *Reflection probe*: This type of probe can also be used for the venous test. The light emitting and sensitive parts are located side by side in one probe. The photosensors detect the light, which is backscattered from the tissue of the skin. Due to the body’s anatomy, the PPG sensors can only detect the pulse waves in areas that contain many arterio-venous anastomoses such as the fingers, toes, earlobes, or some regions of the face.

2. *Transmission probe*: In these probes, the photosensors are located on the opposite side of the light emitting parts. The tissue is located between them. This limits the field of application to locations where the light can penetrate all the way through the tissue (e.g., fingers, toes, or earlobes). In contrast to the reflection probe, the main sources of pulsation also contain the large vessels making these sensors especially useful for peripheral blood pressure measurements.

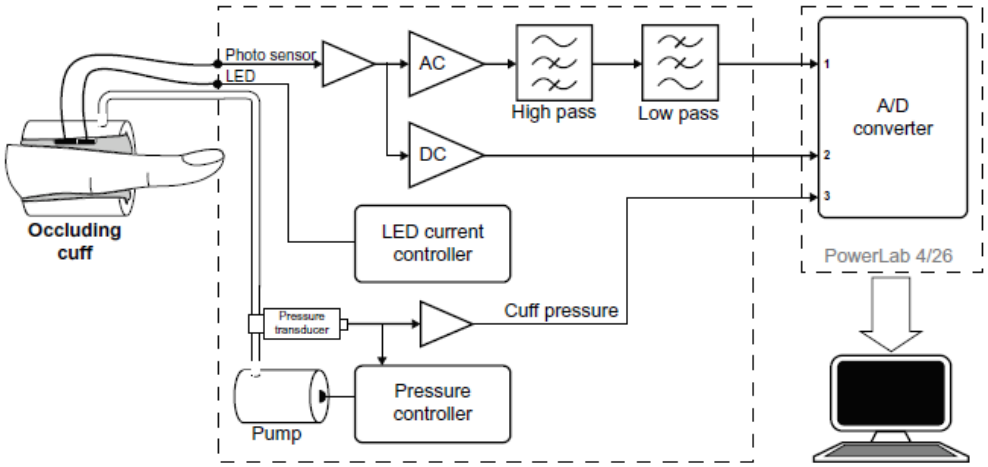


Fig. 1 Schematic block diagram of a photoplethysmograph system for the human finger.

Notes: The light-emitting diode (LED) system generates an infrared light directed toward the finger. The pressure generator induces a vascular volume change through the occluding cuff. The photo sensor detects the reflected light from the finger. The electric current from the photo sensor is decomposed into an alternating current (AC) and a direct current (DC). The AC current is then filtered through a high-pass and a low-pass filter. The A/D converter converts the AC, DC, and cuff-pressure input signals to the digital outputs before storing them on the personal computer.

Strain Gauge Plethysmography

A strain gauge is a mechanical transducer that expresses deformation as a change in electrical resistance. For clinical plethysmographic studies, the transducer is an elastic tube filled with mercury or an indium-gallium metal alloy conductor. Stretching the strain gauge decreases the diameter of the conductor, which increases its electrical resistance. The transducer is calibrated by incrementally stretching the conductor, then measuring the change in voltage that is produced.
When wrapped around a limb segment, say the calf or thigh, the strain gauge provides a circumferential measurement that is used to compute an area. The “slice volume” of the limb segment changes as the calf volume expands and contracts. The strain gauge can be used for
both arterial and venous applications.
Strain gauge measurements are typically determined from voltage tracings. Measurement error may be caused by temperature changes in the conductor, which affects electrical resistance, shifting of the transducer with limb repositioning, and selecting a limb area less sensitive to volume expansion.

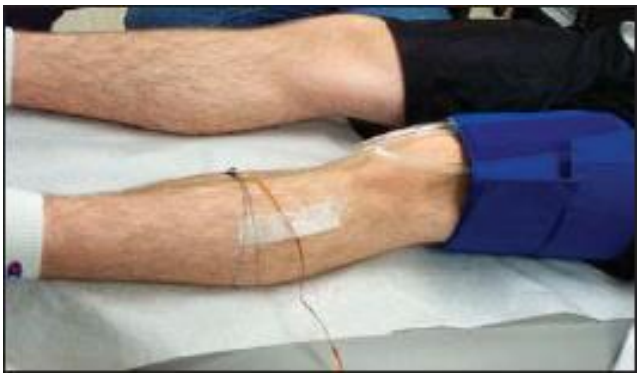
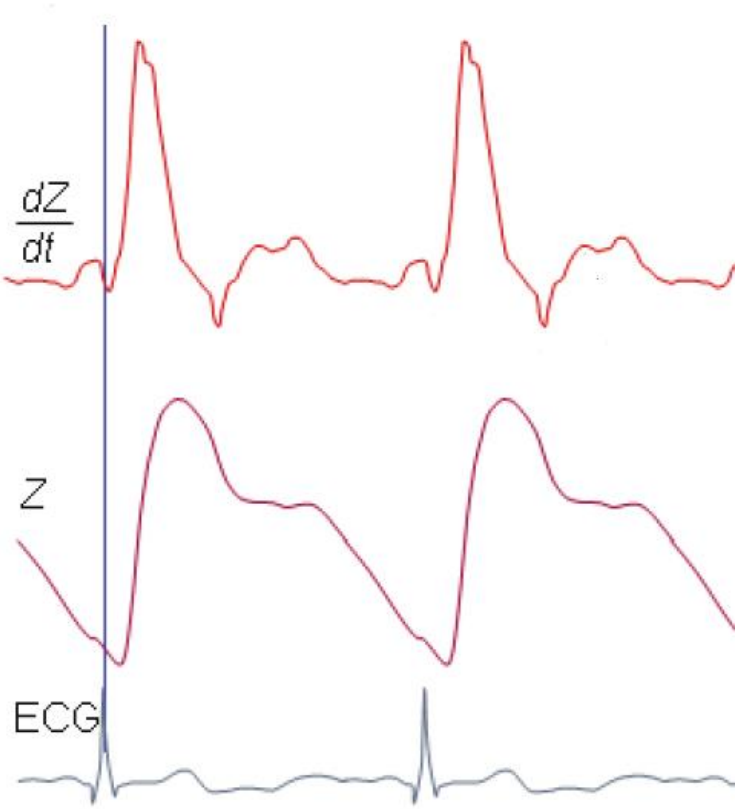


Fig. 2 Circumferential strain gauge measurement

Impedance Plethysmography

Impedance plethysmography (IPG) is used to measure electrolyte fluid volume through changes in electrical conductivity. The instrument passes a weak (1 mA) alternating current through a limb and measures the electrical resistance to current flow. To perform the measurement, four conductive bands are taped around the limb as outer and inner pairs of electrodes. The outer pair applies an alternating current. The inner pair is used to measure electrical resistance. The technique is termed
IPG to coincide with the use of an alternating current, which has impedance composed of resistive, capacitive, and inductive elements.
The major portion of the impedance remains resistive. The current is imperceptible to the patient because of its low amperage and the 22-100 kHz alternating current reduces its interference with systemic neuromuscular processes.
The results are quantitative in resistive units but not easily translated into volume. Electrolyte concentration, red blood cell velocity, haematocrit, and other factors make the conversion inexact. Still, short duration volume changes in the order of seconds to minutes are principally associated with shifts in venous blood volume. Even though extensively used for acute deep vein thrombosis detection, it has not fared well in the transition to chronic venous insufficiency examination.

A typical impedance measuring system is comprised of a sine-wave oscillator followed by voltage to current converter. This converter outputs sinusoidal current of constant amplitude (1-10 mA) which can be passed though the body segment with the help of two band electrodes called the current electrodes I_1 and I_2. Voltage signal developed along the current path is sensed with the help of another pair of electrodes called the sensing electrodes or voltage electrodes V1 and V2 as shown in Fig. 3. The amplitude of the signal thus obtained is directly proportional to the electrical impedance of the body segment between the electrodes V1 and V2. The amplification and detection of this signal yields an output signal, which is proportional to instantaneous impedance (Z) of the body segment. Initial value of the impedance, also known as **basal impedance (Z_o)** is obtained from a sample and hold circuit and is numerically displayed on the panel. Small changes in the impedance of the body segment caused by physiological processes like blood circulation, respiration etc, are obtained by subtracting the initial value of the impedance from the instantaneous impedance and is called the $\Delta Z(t)$ waveform. The Z is also differentiated with respect to time to get the rate of change of impedance or the dZ/dt waveform. By convention $-\Delta Z(t)$ and –dZ/dt are recorded on a strip chartrecorder to relate these waveforms with blood volume changes directly and are colloquially called $\Delta Z(t)$ and dZ/dt waveforms.



Since ΔZ(t) and dZ/dt are produced by the physiological processes, it is possible to extract the changes produced by one particular process by either suppressing the other process or by signal processing techniques. For example to extract the signal produced by blood circulation, the subject under investigation can be instructed to hold his breath. On the other hand a low pass filter can suppress the changes caused by blood circulation and give the changes produced by respiration. Nyboer's equation, derived from parallel conductor theory, relates the blood volume changes (ΔV) with the changes in electrical impedance (ΔZ) as follows (Nyboer J.1960):

$$\frac{1}{Z_o} = \frac{1}{Z_B} + \frac{1}{Z_t}$$

$$taking\ differential, \frac{\Delta Z}{{Z_o}^2} = \frac{\Delta Z_B}{{Z_B}^2}$$

$$Z_B = \rho_B \frac{L}{A}$$

$$V = LA = \rho_B \frac{L^2}{Z_B}$$

$$\Delta V = -\rho_B \frac{L^2 \Delta Z_B}{{Z_B}^2} = -\rho_B \frac{L^2 \Delta Z}{{Z_o}^2}$$

where ρ_B is the resistivity of blood in ohm-cm and it is of the order of 160 Ωcm, L is the length between sensing electrodes and Z_o is the gross electrical impedance and the body segment called **basal impedance**.

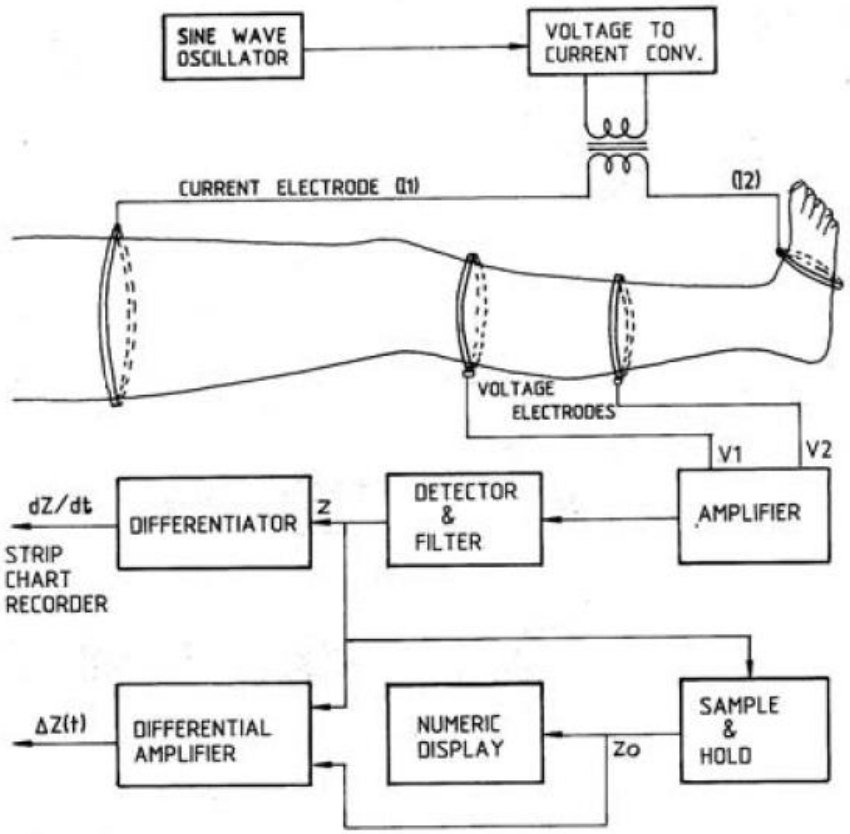


Fig. 3 *Block diagram of a typical impedance plethesmograph system. Constant amplitude sinusoidal current is passed through the body segment with the help of current electrodes I1 and I2. Voltage sensing electrodes V1 and V2 are applied at desired location on the body segment along the current path. The amplitude of sensed sinusoidal signal is directly proportional to the instantaneous impedance Z of the body segment between the sensing electrodes. Z is processed electronically yield basal impedance Zo, ΔZ(t) and dZ/dt waveform.*