3. Measurement of Respiration Using Bio-Impedance:

Impedance Pneumography

Overview

This chapter provides the basic principle of bio-impedance measurements and different types of instruments for the same. The conventional direct and indirect methods are also explained in brief. The relation between change in impedance and the change in volume is explained with the help of bipolar and tetrapolar electrode system. The impedance and frequency characteristics are explained with the help of equivalence circuit and plot of impedance and frequency. The basic equations are designed for the impedance how to measure in the electrical conduction through the biological matter.
3.1 Impedance Measuring Circuit

There are two basic impedance measuring circuits: the bridge and bipolar & tetrapolar circuits. With the bridge circuit, the bridge is adjusted (balanced) so that the output voltage is zero when the impedance (and phase angle) of R and C are equivalent to those of $Z_0$ is shown in Figure 3.1. Then the output is proportional to $\Delta Z$. the bridge circuit can be converted to a constant current circuit by making $Z_1$ high with respect to $Z_0 + \Delta Z$. thus, the current through the subject is independent of the subject impedance ($Z_0$).

With the bipolar symmetrical constant current circuit, the two resistances (R, R) are made high with respective $Z_0 + \Delta Z$, is shown in Figure 3.2. Therefore, the output voltage ($E_{out}$) is linearly proportional to $Z_0 + \Delta Z$. this condition identifies constant current operation, in which the current through the subject is independent of the subject impedance ($Z_0$).

![Impedance Bridge Diagram](image)

*Figure 3.1: Impedance Bridge*

With the tetra-polar constant current, current is admitted to the subject by two electrodes (1, 4) connected to a voltage source via two resistors (R, R) that are high value with respect to impedance appearing between electrode 1 and 4. Alternatively, an active constant current source can be used. The output voltage ($E_{out}$) is obtained from two potential measuring electrodes (2, 3) and is proportional to $Z_0 + \Delta Z$. 

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The properties of the bridge and constant current bipolar and tetra polar constant currents will now be described.

![Diagram of Symmetrical Bipolar constant Current Circuit.](image)

**Figure 3.2:** Symmetrical Bipolar constant Current Circuit.

### 3.1.1 Bridge Circuit

When two electrodes are employed, the impedance bridge circuit diagramed is shown in Figure 3.1 can be used. In such an arrangement, the oscillator voltage $E$ is applied to two opposite corners of the bridge and the detector is connected to the other two. When the ratio arms $Z_1$, $Z_2$ are resistors of equal value, the impedance bridge becomes Comparison Bridge. The balancing arm $RC$ may consist of parallel resistance and capacitance decide unit that are adjusted to balance the bridge for the basal $(Z_0)$ impedance between the electrode terminals.

![Diagram of Tetrapolar constant Current Circuit.](image)

**Figure 3.3:** Tetrapolar constant Current Circuit.

At balance, the value of the balance arm gives the equivalence parallel resistive and reactive components of the tissue electrode circuit. If $R$ and $C$ are placed in series and
the bridge balanced, the new values of R and C give the equivalent series circuit between the electrode terminals. These equivalents are valid only for the frequency employed. With most bridges the impedance is measured in terms of parallel or a series equivalent, but not both. Often it is desirable to transform one equivalent into the other; for example, if in a given case \(R_s\) and \(C_s\) are the series resistance and capacitance at a particular frequency \(f\), the parallel equivalents \(R_p\) and \(C_p\) at the same frequency are given by

\[ R_p = \frac{1 + (2\pi f C_s R_s)^2}{4\pi^2 f^2 R_s C_s^2} \quad R_p = R_s + \frac{X_s^2}{R_s} \]  

\[ C_p = \frac{C_s}{1 + (2\pi f C_s R_s)^2} \quad X_p = X_s + \frac{R_s^2}{X_s} \]  

\[ R_p = \frac{R_p}{1 + (2\pi f C_s R_p)^2} \quad C_p = C_p \left[ 1 + \left( \frac{1}{(2\pi f C_p R_p)^2} \right) \right] \]

Frequently it is desirable to carry out the reverse process that is to express a series circuit in terms of a parallel one. Rearranging of the expressions just given provides the relationships.

With the bridge circuit, the changes in impedance reflecting the physiological event produce a varying output voltage \((E_{\text{out}})\), which, after amplification and demodulation, is displayed to produce a record related to changes in the physiological event. It must be emphasized that an output is obtained if there is a change in either the resistive or the resistive components or in both. If it is desired to examine the magnitude of each component individually, a phase sensitive detector is required.
If the bridge is operated at the balance point and without the use of a phase sensitive detector, an output voltage is obtained if the impedance being measured increases or decreases. Under this operating condition, direction indication is lost. However, if after the bridge has been initially balanced, it is then unbalanced slightly by addition on a small resistance in series with the impedance being measured, the output voltage from the bridge will increase and decrease with the impedance being measured. The amount of resistance added is dictated by the maximum change in impedance required to drive the bridge toward the balance point.

Precautions must be taken in using the bridge circuit to avoid having the size of the output voltage become dependent on the magnitude of \( Z_0 \), the basal impedance. For the same \( \Delta Z \), if \( Z_0 \) is small, more current will flow through \( Z_1 \) and \( Z_0 \), and the output \( (E_{\text{out}}) \) will be large. This sensitivity dependence on \( Z_0 \), can be eliminated by making \( Z_1 \) much greater than \( Z_0 + \Delta Z \), which allows the bridge to have the properties of constant current.

### 3.1.2 Bipolar Constant Current Circuit

The constant current bipolar electrode system is depicted in Figure 3.2. Current from the oscillator is fed symmetrically to the electrodes through two resistances \( R \), \( R \), which is high value with respect to the total impedance between the electrode terminals. With this circuit configuration the current through the subject is determined by these resistances and the oscillator voltage \( E \) and it is independent of the electrode subject impedance \( (Z_0 + \Delta Z) \). The detector is connected across the electrodes, and the voltage present is a function of the basal impedance between the electrode \( Z_0 \) and any change \( \Delta Z \) due to the physiological event. It can easily be shown that the voltage across the electrode is given by

\[
E_{\text{out}} = \frac{E(Z_0 + \Delta Z)}{2R + Z_0 + \Delta Z}
\]  

(3.4)

If \( R \) is made much greater than \( Z_0 \) and \( \Delta Z \) is much less \( Z_0 \) then

\[
E_{\text{out}} \approx \frac{E(Z_0 + \Delta Z)}{2R} \approx \frac{EZ_0}{2R} + \frac{E\Delta Z}{2R}
\]

(3.5)
Demodulation of these signals after amplification yields a large constant signal \((E_0/2R)\) plus a smaller one \((E \Delta Z/2R)\) proportional to the impedance change due to the physiological event. The larger signal, reflecting the basal impedance of the electrode subject circuit, is often eliminated from the output by blocking this dc component with a capacitive or by canceling it with opposing voltage.

### 3.1.3 Tetrapolar Constant Current Circuit

The tetrapolar circuit is shown in Figure 3.3 for measuring impedance permits attainment of the highest accuracy because when properly applied with a constant current source, it eliminates all electrodes – subject impedance errors. The tetrapolar circuit was introduced by Bouty in 1884 to measure the resistivity of electrolytes with high accuracy.

With the tetrapolar circuit, current \(I\) is injected into the tissue by two current injecting electrodes (1 and 4). The tissue impedance dependent voltage \((E_{\text{out}})\) is obtained from two potential measuring electrodes (2 and 3). When a constant current source is used to inject the measuring current, its value is independent of the impedance of the two electrode subject impedances \((Z_{1s} \text{ and } Z_{4s})\) and that of the tissue between electrode 1 and 4 \((Z_{1s-4})\). This requirement is met when the output impedance \((2R)\) of the current source is large with respect to \(Z_{1s} + Z_{s1-4} + Z_{4s}\).

When the current injectors (1, 4) are widely spaced with respect to the potential measuring electrodes (2, 3), there arises the best opportunity to achieve a uniform current distribution between the potential measuring electrodes.

The voltage \((E_{\text{out}})\) appearing between the potential measuring electrodes (2, 3) is proportional to the current injected \((I)\), the impedance of the tissue between the potential measuring electrodes \((Z_{s2-3})\), and their separation. Therefore, \(E_{\text{out}} = IZ_{s2-3}\). Now, if a voltage measuring instrument has an input impedance that is high \((Z_{2s}, Z_{3s})\) and that of the tissue between the potential measuring electrodes \((Z_{s2-3})\), the potential measuring instrument indicates the true voltage \(E_{2-3}\); therefore

\[
E_{2-3} = IZ_{s2-3}; \quad Z_{s2-3} = Z_0 + \Delta Z \quad \text{and} \quad E_{2-3} = E_{\text{out}} \quad (3.6)
\]
The method of calibrating the tetrapolar system is simple and merely requires disconnecting all four electrodes from the subject and connecting the potential measuring electrodes (2, 3) to a calibrated noninductive resistor. Electrode 1 is then connected to electrode 2, and electrode 4 is connected to electrode 3. Thus the tetrapolar system is converted to bipolar system for calibration.

Figure 3.4: Techniques for balancing to permit direct-coupled recording with bipolar and Tetrapolar constant Current Circuits.

When the bipolar or tetrapolar systems are used to measure physiological events by impedance, the desired signal usually rides on a large undesired (basal) signal \(Z_0\), which must be removed. If (after demodulation) only the change in impedance is desired, capacitive coupling with an adequately long time constant is satisfactory.
However, if both the basal impedance $Z_0$ and its change $\Delta Z$ are of interest, it is necessary to subtract the large output signal that represents $Z_0$. Although easy to do with a bucking voltage applied after the impedance signal is demodulated, the baseline recording will reflect the value of $Z_0$ plus any instability in the amplitude of the constant current source. This problem can be eliminated by deriving the bucking voltage from the constant current source; use of this technique was described by Worley and Geddes (1992) and is applicable to the bipolar and tetrapolar methods.

Figure 3.4 illustrates block diagrams for bipolar and tetrapolar impedance measuring circuits in which a constant source is created by using an oscillator connected to a step up transformer delivering current to the subject via two resistors that are high in value with respect to the impedance of the subject. The oscillator is also connected to a buffer amplifier ($A_b$), which delivers its output to a rectifier / filter (demodulator) circuit with along time constant. From this circuit is derived a direct voltage proportional to the basal impedance $Z_0$. This signal is amplified and fed into a summing amplifier ($A_s$) to subtract the basal impedance signal from the $Z_0 + \Delta Z$ signal. The $Z_0$ balance (BAL) control adjusts the amount of canceling that is applied.

### 3.1 Electrical Conduction in Biological Matter

Biological tissues such as muscles, bones, etc. and biological fluids such as blood, urine, cerebrospinal fluid etc. are neither good conductors of electricity as a metal nor a bad conductor as wood. This intermediate property of the biological matter makes its measurement feasible by simple principle/instruments. The conductivity of biological fluids is more than that of tissues due to abundant charge carriers in the former. Essentially conductivity is the measure of the ability of electrical charge to move in the material under the influence of an electrical field (electric voltage applied between two points in the material). The conductivity ($\sigma$) is either expressed directly or by its reciprocal, known as resistivity ($\rho$).

For the material, obeying Ohms law, resistivity is defined as the resistance offered by 1 cm$^3$ of the material to the flow of electric current. Conductivity $\rho$ is independent of the geometrical configuration of the material and can be calculated from the inherent properties of the material as follows. If $N$ is the number of charge carrier available in
the material per unit volume, $Q$ is the charge (coulombs) per charge carriers and $\mu$ is the mobility of the charge carrier or the net velocity imparted to the charge carrier by an electric field of unit strength, the conductivity of the material is given as:

$$\sigma = \frac{NQ}{\mu}$$  \hspace{1cm} (3.7)

$$\rho = \frac{1}{NQ \mu}$$  \hspace{1cm} (3.8)

$N$ and $Q$ are constants for a material; however $\mu$ is temperature dependent and $\sigma$ or $\rho$ therefore needed to be specified at a particular temperature. From the value of $\rho$ it is possible to calculate the electrical resistance offered by the matter of known geometry. If $L$ (cm) is the length and $A$ (cm$^2$) is the area of cross section of a homogeneous cylindrical conductor, resistance ($R$) is given as:

$$R = \frac{L}{A}$$  \hspace{1cm} (3.9)

$R$ can be accurately measured for any object using Ohms Law. In this case constant current is passed through the object and the voltage developed across is measured. The ratio of voltage and current gives the value of $R$. Since $L$ and $A$ can be physically measured, above equation can therefore be used for the determination of $\rho$.

Similarly Kubicek [49] derived formula of $Z0$ for volume stroke, in blood flow, measurements. Biological materials have slightly different behavior than the Ohmic conductors described above. Measurement of $\rho$ in biological materials is complicated.

(a) Electrolytic nature of the fluids,
(b) Distribution of the material in the suspension,
(c) Orientation of membranes which are relatively poor conductors.

For instance, in skeletal muscle, the transverse resistivity is much higher than the longitudinal resistivity. Furthermore, the resistivity of blood is highly dependent on the haematocrit and on whether the blood is in motion. Table 3.1 gives the resistivity of various biological materials as described by Baker et al [50].
Table 3.1: Table shows the resistivity of various biological materials.

<table>
<thead>
<tr>
<th>Sr. No.</th>
<th>Biological Material</th>
<th>Resistivity in Ohm-cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.</td>
<td>Urine</td>
<td>30</td>
</tr>
<tr>
<td>2.</td>
<td>Plasma</td>
<td>63</td>
</tr>
<tr>
<td>3.</td>
<td>Cerebrospinal Fluid</td>
<td>65</td>
</tr>
<tr>
<td>4.</td>
<td>Blood</td>
<td>150</td>
</tr>
<tr>
<td>5.</td>
<td>Skeletal Muscle</td>
<td>300</td>
</tr>
<tr>
<td>6.</td>
<td>Cardiac Muscle</td>
<td>750</td>
</tr>
<tr>
<td>7.</td>
<td>Lung</td>
<td>1275</td>
</tr>
<tr>
<td>8.</td>
<td>Fat</td>
<td>2500</td>
</tr>
<tr>
<td>9.</td>
<td>Bone</td>
<td>16600</td>
</tr>
</tbody>
</table>

33 Electrode System

When the impedance technique is used to measure a physiological event, one of two methods is employed. With both, the physiological event is placed between the measuring electrodes in such a way that the event alerts the current density distribution between the electrodes, thus manifesting itself as change impedance. In one method, the electrodes are in ohmic contact with the preparation and a relatively low impedance circuit is formed; in the other, the electrodes are insulated from the subject and relatively high impedance capacitive circuit results.

Although the circuits described earlier are those most frequently employed, in some cases in which movements has been converted into corresponding changes in capacitance the variation in capacitance has been employed to frequency modulate a carrier. On a few occasions a direct voltage was applied to the electrodes and the physiological event altered the conductivity and/or capacitance of the circuit, thereby modulating the direct current.
33.1 Guard Electrode Technique

When conducting property of a specimen is measured with electrodes that are small with respect to its size, the current spreads beyond the electrodes as in Figure 3.5(a). This situation which often exists physiological events which are measured by impedance plagued the early physicists seeking an accurate value for the resistivity of a material. Current spread also occurs when a small electrode is used in conjunction with a large electrode as shown in Figure 3.5 (b). To achieve a uniform current density distribution under an electrode, a guard ring electrode is applied. Figure 3.5(a) & 3.5(b) shows that the current distribution with small and large electrodes applied to a conducting specimen extends well beyond both electrodes. To obtain a uniform current density distribution under the smaller electrode, a concentric guard electrode is placed around the smaller electrode and the potential of guard; the guard electrode is maintained at the potential of the smaller electrode used for current measurement as shown in Figure 3.5 (c). Note that this technique prevents current spread from the smaller electrode [50] connected to the current meter and achieves a more uniform current density distribution in the specimen. The current flowing in the guard electrode is not measured; it merely aids in providing a uniform current density distribution around the main current path in the specimen. The modern method of applying a guard electrode employs an operational amplifier of unity gain to drive the guard electrode; this technique is illustrated in Figure 3.5 (d).

The guard electrode technique is sometimes applied when the constant voltage system is used to measure impedance or impedance change. Graham reported its use in the measurement of respiration in the human with a chest-to-back electrode arrangement operated at 50 kHz.

Without the guard applied, the basal impedance was 400 Ω, which increased to 404 Ω with inspiration. When the electrode on the back was guarded, the basal impedance was 8000 Ω and the same inspiratory volume increased the impedance by 80 Ω. In this application, which measured the ventilation of one lung, a 20 fold respiratory impedance change was obtained by using the guard electrode technique. Application of the guard increased the measured basal transthoracic impedance. The ratio of the change in respiratory transthoracic resistance using the guard electrode technique to
that with simple bipolar electrodes varied with the size of the guarding electrodes; ratios ranging from 39 to 67.5 were obtained.

Figure 3.5: Current spread in a conductor and the use of a concentric guard electrode $G$ to achieve uniform current density distribution. a. Current spread in a conductor with similar electrodes; b. Current spread with different sized electrodes; c. Current spread with guard electrode maintained at the same potential as measuring electrode, $M$; d. Guard electrode driven by a unity gain amplifier ($A$).

33.2 Electrodeless impedance measurement

By the use of electromagnetic induction, it is possible to measure the resistivity of a conducting substance without applying electrodes; this technique is used quite extensively in geophysical exploration. The first to apply this method to obtain signals reflecting respiration and pulsatile blood flow were Tarjan & McFee, who
employed three identical rigidly mounted coaxial coils separated by a distance approximately equal to their radii. The central coil was energized with 100 kHz current, and the two outer coils were connected in series opposition, as in Figure 3.7. The arrangement therefore constitutes a differential transformer. With the three-coil assembly well removed from conducting materials, amplitude and phase adjustments were made to obtain the smallest possible unbalanced signal from the coils connected in series opposition.

When such a balanced three coil assembly is brought near conducting material, a current is induced in it. The magnitude of the current is proportional to the conductivity. The induced current produces a magnetic field of its own, which alters the voltage induced in one or both of the two pickup coils. With fixed geometry, the unbalanced voltage can be detected and processed to provide a measure of the conductance of the material and any variations it experiences.

![Diagram](image)

**Figure 3.6**: Symmetrically guarded concentric bipolar for measuring transthoracic impedance

Tarjan & McFee detected respiration in the human by mounting the coil assembly a short distance from the thorax. By carefully placing the axis of the coil assembly over the heart and requesting that the subject hold his or her breath at full inspiration, recording strikingly similar to ventricular volume changes were
obtained. With the assembly placed near the head, the investigators recorded pulsatile changes reflecting cerebral blood flow. In this application, they were even able to detect metallic tooth fillings.

Figure 3.7: The electrodeless method of measuring impedance and impedance change using electromagnetic induction.

The electrodeless method for detecting impedance changes exhibits some interesting characteristics. For many reasons, the elimination of electrodes provides great practicability and allows nonstressful measurements to be made. Magnetic fields penetrate readily into biological tissue, and subintegumental events can be detected more easily than by the use of surface electrodes. However, with electromagnetic induction, the magnitudes of the signals provided by physiological events are small, and they decrease with increasing distance from the coil assembly; thus care must be exercised to obtain the optimum placement of the coil assembly that constitutes the transducer. If artifacts are to be avoided, the subject must not move with respect to the coli assembly. It is also important to prevent the induced current from attaining a magnitude sufficient to stimulate irritable tissue or to cause heating. Finally, as with all other impedance-measuring techniques; it is difficult to calibrate the signal obtained in terms of the
magnitude of the physiological event.

3.4 Selection of Parameters for Respiration Measurement Using Impedance Pneumography

3.4.1 Characteristics of impedance pneumography

Practically it is probably the most attractive feature of the impedance method for the measurement of respiration. Nothing is simpler than affixing electrodes to a subject and connecting them to the recording equipment. Since the impedance change is related to the volume of air moved, the system must be calibrated according to standards. The calibration requires the use of a Spirometer or volume measuring instrument. After removing the calibrating device respiratory volume can be measured without obstructing the air-stream. Another attractive feature of the method is availability of the electrocardiogram and cardiac impedance signal from the same electrode.

3.4.2 Impedance Volume Relation

The relationship between impedance change ($\Delta Z$) and volume of air moved ($\Delta V$) is approximately linear.

For humans the coefficient ($\Delta Z/\Delta V$) depends on the size of the subject and the location of the electrodes. Baker and Geddes in their study obtained a fairly good linearity between $\Delta Z$ and $\Delta V$ for different electrode location. In this study they used bipolar electrodes. They obtained ($\Delta Z/\Delta V$) equal to 6.0 $\Omega/L$ for adults. The linearity between $\Delta Z$ and $\Delta V$ obtained in human subjects of different builds is presented graphically in Figure 3.8. This linearity was obtained with bipolar electrodes placed on midaxillary lines at different levels on the chest. Inspection of this illustration shows that the coefficient ($\Delta Z/\Delta V$) is the largest for adults of slight build.
Figure 3.8: **Transthoracic impedance changes ΔZ vs. respired volume measured ΔV with bipolar electrodes applied to subjects of light, medium and heavy builds.**

### 3.4.3 Impedance of Living Tissue

The smallest unit of living tissue is the single cell, which can be idealized as an electrolyte containing a variety of sub cellular structures necessary for metabolism repair and reproduction. These sub-cellular structures are completely enveloped by a membrane having a low electrical leakage in the resting state.

The cell membrane acts as a passive capacitance at resting stage. The value of such capacitance ranges from 0.1 to 12 μF/cm² with a typical value of about 1.0 μF/cm². The resistivity of cytoplasm ranges from 10 to 30000 Ω cm for minimum cells. Biological tissue consists of aggregation of cells of differing shapes bonded together and surrounded by tissue fluids containing electrolytes. Therefore the dc or low frequency current passing through a specimen of living tissue can pass around the cells as shown in Figure 3.9 (a) for the high frequency current the reactance of the
membrane capacitance of the cell is small, and current of through cytoplasm as well as environment fluid as shown in Figure 3.9 (b). Therefore low frequency tissue impedance is high and high frequency impedance is low as shown in Figure 3.9 (c).

Figure 3.9: Pathway for a) low frequency current and high frequency current for a cell in an electrolyte; c) the resulting idealized impedance – frequency characteristic.

The transition from the high to low value is also shown is the capacitance of characteristics for the type of tissue reflects which the capacitive nature of the cell membrane. These characteristic features of living tissue, recognized by Philippine
1920, lead to the concept of equivalent circuit. The circuit at the right side of the
Figure 3.10 is frequently used to describe the passive electrical behavior of biological
specimens.

![Circuit Diagram]

**Figure 3.10:**  
*a) Idealized equivalent circuit for living tissue; b) the impedance 
frequency characteristics for living tissue represented by the locus.*

### 3.4.4 Selection of Frequency and Current

Typical impedance frequency and phase frequency characteristic for a single frog egg
suspended in an electrolyte are shown in Figure 3.11. Figure 3.12 is the impedance
locus plot for the same specimen. Recalling that the resistivity of electrolytes is
constant over a considerable frequency range, it is interesting to note that the circuit
measure between the electrode terminals exhibit capacitive reactance. The centre of
the semicircle in the impedance – locus plot lies slightly below the resistance axis
because the cell membrane is not an ideal capacitor. A line between the point where
the arc of the semicircle crosses the resistance axis and the centre of the circle makes
an angle $\alpha$ with a perpendicular line through the centre of the circle. This angle is the
membrane phase angle. If the cell membrane were a perfect insulator, $\alpha$ would be $90^\circ$, 

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the tangent of $\alpha$ would be infinity, and the centre of the semicircle would be on the resistance axis.

Figure 3.11:  a) Impedance frequency characteristics of a single frog egg in an electrolyte; b) phase frequency characteristic.

Selection of frequency for measurement of the respiratory impedance change is very important. The high frequency current is used for starting consideration because due to low frequency current may stimulate the cell. It is noted that the change in transthoracic impedance from full inspiration to maximum expiration is particularly the same in the frequency range 50 KHz to 600 KHz.

When the impedance locus method is used to display the impedance-frequency characteristics of a biological specimen, the pattern obtained is slightly different from the previous Figure 3.12.
Figure 3.12: Impedance locus plot for a single frog egg suspended in an electrolyte

3.5 In-Vivo Impedance Measurement

In-vivo measurement in the body is practically not feasible by the method described above. This is because it is not possible to apply the end plate electrodes in uniform contact with area of cross-section of the body segment. For the technique to remain NON-INVASIVE it is necessary to use surface electrodes. Baker (1989) has shown the validity of using surface electrodes for in-vivo impedance measurement [50].

Band electrodes (in the form of loop around the subject) for injecting current provide essentially uniform current distribution in the central portion of the object. Therefore another pair of band electrodes (sensing electrodes) is applied around the central portion of the object for measurement of voltage signal developed along the current path. This tetra-polar method not only insures the uniform current distribution between the potential measuring electrodes but also reduces considerably the polarization at sensing electrode. Use of spot electrode on the surface similar to those used for electro-cardiography produces extreme distortion of the current density
distribution and yields unreliable measurement. Thus the band electrodes, in the form of loop around the body segments, applied on the surface of the body appear to be the best choice for non-invasive measurement of the electric impedance in the human body. Kubicek (1974) and Mohpatra (1981) [48, 51] substituted the values of change in impedance for calculating the change in volume across the area of the given segment with electrical impedance technique.

The same principle was used for the measurement of respiration. As a person breathes continuously there is change in air volume in the chest/lungs area. Using the principle of impedance pneumography the constant amplitude sine wave current of high frequency is applied to the arms and the change in air volume – impedance is measured across the chest by using sensing electrode. Thus at the start, some basal impedance is noted and this will be taken as a reference, the next values of impedance will be measured and the previous (reference) impedance is subtracted from the next by using differential mode so that only change in the impedance will occur. This principle is used and further processed for the measurement, and diagnosis related to lungs.

3.5.1 Measurement of Respiration: Impedance Pneumography

Impedance pneumography is a method of determining changing tissue volumes in the body, based on the measurement of electric impedance at the body surface. Impedance pneumography employs low amplitude, high frequency (50 to 500 KHz) alternating current (AC) between two surface electrodes to record thoracic movements or volume changes at the rib cage during a respiratory cycle. Based on Ohm's law, the voltage drop across the electrodes is computed as impedance, which increases during inspiration and decreases during expiration. Impedance pneumography technique will be more preferable for the variability and which is non invasive method for respiration measurements. The inspiration and expiration will change the air volume which is proportional to the change in impedance if we apply constant current through the body.
3.5.2 Impedance and its Measurement

Resistance measurement becomes difficult in biological materials or any electrolytic substance as the application of steady electric field results in polarization at the electrodes. The factors responsible for such polarization are:

a) the capacitance formed by the electrodes with the biological specimen as the dielectric.

b) the capacitance effect of the double layer at the surface of the electrode and

c) the Faradic admittance in parallel with the double layer.

The difficulty is reduced to a large extent by employing time varying electric field, usually sinusoidal in place of DC current. The quality measured using sinusoidal field
is usually referred to as impedance \((Z)\), in place of resistance, due to the frequency dependant character of the former. The frequency of the time varying field is chosen between 20 KHz to 200 KHz for measurement on biological materials as the effects described above are reduced to less than 4% in this range (Price et al 1979).

Generally, constant current method or bridge method is employed for the measurement of resistance or impedance. Constant current method is based on Ohms law. In this method a constant DC current (for resistance measurement) or constant amplitude sinusoidal current (for impedance measurement) is passed through the object and the voltage signal developed across the conductor is measured as shown in Figure 3.14. Bhuta et al [52] explained these technical aspects of impedance pneumography.

In in-vitro measurement, the biological material is connected in a rectangular cell made of an insulator. The two open surfaces of the cell are closed by conducting end plate (electrodes), which are in contact uniformly with biological material. The resistance \((R)\) of the material is calculated from the voltage \((V)\) measured and the constant current \((I)\) applied. If \(L\) is the distance \((\text{cm})\) between the end plate electrodes and \(A\) is the area of cross-section \((\text{cm}^2)\) of the conductivity cell, the resistivity \((\rho)\) of the biological material is calculated using following equation:

\[
\rho = \frac{R.A}{L} \hspace{1cm} \text{(3.4)}
\]

![Figure 3.14: Principle of impedance measurement by constant current method.](image)
Basic impedance measurements are shown in Figure 3.14 with simple principle of impedance by applying constant current across the measurement of impedance.

![Graph showing impedance measurement parameters]

Figure 3.15: Normal Respiration Signal

The normal respiration waveform and its parameters are shown in Figure 3.15.

1. Inspiration: From A to B:
2. Expiration: From B to C
3. Respiration Time (Rt): From B to D:
4. Respiratory rate = \( \frac{60}{Rt} \) Breaths per Min.
5. Respiration Amplitude = \( R_{\text{amp}} \)