

ELECTRICAL PROPERTIES OF BIOLOGICAL TISSUE

Work code: Biotissue

ANNOTATION

The paper briefly examines the structure of biological tissue, its primary properties, and describes the welding process, with an emphasis on its impact on the tissue. The purpose of this study is to analyze the electrical properties of biological tissue during welding, specifically examining its impedance as a key characteristic. The relevance is due to the insufficient study of the electrical parameters of tissues during welding, which complicates the modeling and design of appropriate equipment. A selection of impedance measurement methods is presented, and their feasibility is substantiated. Experimental studies have shown that at the operating frequency, the impedance of biological tissue is predominantly active due to a slight phase shift. Thus, the results allow simplifying the mathematical modeling of tissue as an electrical load, which increases the accuracy and efficiency of the design of electric welding systems.

Keywords: *biological tissue, impedance, electric welding, measuring transducers, electrical properties, phase shift, electrical load.*

CONTENTS

INTRODUCTION	3
1 OVERVIEW OF THE PROPERTIES OF BIOLOGICAL TISSUE	5
2 WELDING PROCESS OVERVIEW	9
3 BIOLOGICAL TISSUE AS AN ELECTRICAL LOAD.....	11
4 IMPEDANCE MEASUREMENT METHODS	14
4.1 Measuring transducers based on the potentiometric measurement method	14
4.2 Measuring transducers based on the resonance measurement method ..	16
4.3 Measuring transducers based on transfer function analysis	17
4.4 Measuring transducers based on bridge measurement methods.....	18
5 EXPERIMENTAL IMPEDANCE MEASUREMENT.....	20
5.1 Structure of the installation	20
5.2 Installation calibration	21
5.3 Experimental research	23
CONCLUSIONS	25
LIST OF REFERENCES	26
APPENDIX A. The results of the experiment.....	28

INTRODUCTION

Welding of biological tissues is a promising area of surgery that has a number of advantages over surgical tissue connection. Important advantages of welding include the speed of tissue connection, reduced blood loss, the ability to connect tissues without the use of suture materials, and complete postoperative rehabilitation [1]. These factors have become key to the development of welding in surgery. However, despite its proven effectiveness, this method of tissue connection continues to attract the attention of scientists, particularly in the field of electronics. Research focuses on both the welding process itself and the devices used to perform it. Since systems for welding biological tissues require a high level of reliability and stability, improving these systems remains a relevant area of research. This could expand the possibilities for using welding systems in surgery, increase their portability, and improve the results of surgical interventions, contributing to the wider use of welding systems.

An important component for analyzing and designing electronic systems for welding is the study of biological tissue as a load. It is known that biological tissues are characterized by active and capacitive resistance [1]. However, changes in the physiological state of the tissue lead to independent changes in these parameters [2]. This significantly complicates the task of designing and testing electronic equipment, since the reaction and properties of living tissues differ significantly from non-living tissues. The study of living tissues is an important aspect of improving the efficiency of the welding process and eliminating possible negative reactions of the tissue to the action of the electric current. A clear understanding of the nature of tissue impedance and its dependence on current frequency is necessary, since the welding process is based on the use of tissue electrical resistance. This will allow for the improvement of the mathematical model of biological tissue, which would simplify experimental work with electrical equipment.

The study aims to analyze the impedance of biological tissues as an electrical load for welding systems and to determine its dependence on the power level and duration of the electrosurgical process.

The object of the study is biological tissue. The subject is the impedance characteristics of the tissue during the welding process.

The scientific novelty of the obtained results lies in the study of changes in the impedance of tissues that undergo changes in structure and physical properties during the welding process. The results obtained will provide a deeper understanding of the properties of biological tissues, specifically during the welding process, which will enable the construction of an appropriate mathematical model for the development and design of electrosurgical systems. This research is important for improving the results of surgical operations, since welding equipment can be designed taking into account changes in tissue properties under the influence of electric current.

1 OVERVIEW OF THE PROPERTIES OF BIOLOGICAL TISSUE

Since the formation of a connection during the welding of living tissues occurs due to changes in their structure and physical properties, it is worth starting with a brief overview of the structure of tissues and cells of living organisms to improve understanding of the welding process.

Tissues are groups of cells that have a similar structure and perform a common function. A cell is the smallest functional component of an organism and the basic unit of life [3, 4]. Figure 1.1 shows the structure of a cell.

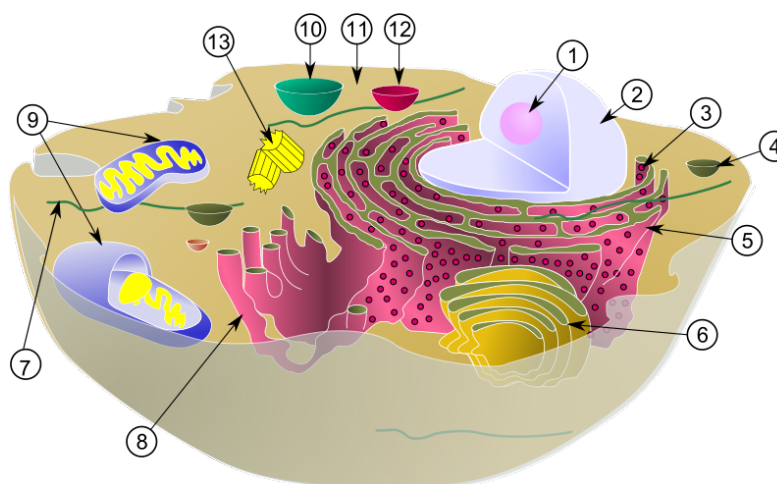


Fig. 1.1. Cell structure and its main components: 1 – Nucleolus; 2 – Nucleus; 3 – Ribosome; 4 – Vesicle; 5 – Rough endoplasmic reticulum; 6 – Golgi apparatus; 7 – Cytoskeleton; 8 – Smooth endoplasmic reticulum; 9 – Mitochondria; 10 – Vacuole; 11 – Cytosol; 12 – Lysosome; 13 – Centriole [5]

In the context of tissue welding, the most important components of a cell are the plasma membrane and cytoplasm. The membrane is the outer boundary of the cell and its direct contact with neighboring cells and the external environment. It consists mainly of proteins and lipids. The cytoplasm is the internal environment of the cell, which includes the cytosol and organelles. The cytosol is the aqueous part of the cytoplasm, containing various dissolved substances, ions, and small molecules [3, 4].

Research into the electrical characteristics of living tissues has been conducted for over 100 years. It has been established that cell membranes have capacitive

properties due to their structure. Their main component is a double phospholipid layer. Since lipids do not conduct electricity, they act as a dielectric, separating the conductive media - the intracellular and intercellular spaces. This structure determines capacitive behavior, since membranes can accumulate electrical charge like a capacitor, where the lipid layer is an insulator and the conductive media are electrodes [2].

The structure of tissue influences its electrical properties. Intracellular fluid contains ions and dissolved molecules that determine its electrical conductivity, while cell membranes act as obstacles to charge movement. The extracellular space is also conductive due to its ion content. This confirms that tissue is a heterogeneous conductor [5]. From an electrical perspective, it can be modeled as an electrical circuit with a parallel connection of active resistance (the electrolytic resistance of the cytoplasm and intercellular space) and capacitance caused by the membrane.

Besides general electrical characteristics, an important part of research is understanding how the physiological state of tissue affects its electrical properties. Specifically, experiments show that the impedance of dead muscle tissue membranes is much lower than that of living tissue. This is likely due to the breakdown of tissue cell structures after death. Cytoplasmic resistance stays nearly the same, membrane capacitance increases, and polarization resistance decreases. Measurements of liver cell traits showed stable parameters, unlike muscle tissue. These findings suggest that changes in tissue electrical properties depend on how permeable the cell membranes are and the ion concentration in the cytoplasm. During different physiological processes or conditions, these parameters can vary independently of each other [2]. The results confirm the need for further research on the electrical properties of biological tissues to better understand the links between changes in cell membrane structure and their electrical behavior.

These properties of biological tissues, together with the developers' understanding of the basic phenomena occurring during high-frequency welding, have made it possible to create an approximate model of the high-frequency electric welding process. The main phenomena affecting welding include the electrical conductivity of

tissue, the dependence of conductivity on temperature, the formation of a conductive channel in living tissues, and the temperature denaturation of proteins contained in living tissue. Let us consider each aspect in more detail [1]:

1) The electrical conductivity of tissue is determined by the electrolyte content in tissue fluids and the structure of the tissue itself (epithelium has the lowest conductivity, while muscle tissue is characterized by significant resistance due to the dense arrangement of cells);

2) The conductivity of living tissue has a positive temperature coefficient, i.e., its conductivity increases with increasing temperature;

3) The formation of a conductive channel in living tissues is caused by damage to cell membranes under the influence of electric current. Loss of membrane integrity leads to a sharp decrease in tissue impedance, which creates conditions for the passage of electric current;

4) Temperature denaturation of proteins occurs when they are heated above 55 °C. The formation of a connection during surgery must be as quick as possible, so the tissue must be heated to a temperature at which protein coagulation will occur in a few seconds.

Electric current breaks through cell membranes, but it has been found that they are capable of recovering if the electric field strength drops. A high modulation frequency is required to keep the cell in a broken state. It has also been experimentally investigated that increasing the voltage or duration of the current does not allow the desired connection to be obtained, as this leads to thermal damage. This is explained by the fact that protein coagulation in cells occurs before the destruction of lipid membranes, i.e., before the formation of a single protein space in the tissues being connected [1].

Although the described properties of biological tissues are not exhaustive, they have made it possible to understand the basic physical processes that occur during tissue electrofusion. This became the basis for further experimental research and technical improvements that led to the modern understanding of the welding mechanism. In particular, the modulation of high-frequency voltage by rectangular

pulses with a frequency of several thousand hertz to destroy cell membranes before coagulation begins, which ensures the effectiveness of welding [1].

2 WELDING PROCESS OVERVIEW

The algorithm for joining tissues during electric welding is similar to the algorithm for contact electric welding of metals, since the joint is formed as a result of a change in the structure of the material under the influence of electric current. However, in surgery, it is not metals that are in contact, but living tissues. The general algorithm for joining tissues consists of compressing them with electrodes and passing a high-frequency current through them, which leads to heating and denaturation of proteins, thereby forming a strong connection [1].

This process not only depends on the direct influence of current but also causes changes in the electrical properties of tissues. As a result of tissue heating, significant changes occur in their impedance. When it reaches its minimum value, protein coagulation begins, leading to an increase in impedance. After the impedance stabilizes, the tissue temperature does not increase, regardless of the thickness and physical parameters of the tissue. As soon as the connection becomes strong enough, the energy supply is stopped [1].

Existing welding complexes allow surgical interventions to be performed in four modes:

- automatic electric welding – to obtain a sealed seam without suture materials or in cases where tissue suturing with suture materials is impossible;
- sealing of blood vessels – performed during tissue separation and organ mobilization, amputations, and tumor removal;
- high-frequency electric welding of large areas of living tissue – used when removing large areas of tissue with simultaneous sealing or stopping of bleeding (for example, during lung resection with simultaneous sealing or during the removal of a large tumor with stopping of bleeding);
- cutting tissue with simultaneous occlusion of small vessels – for dissection of muscles or removal of part of an organ [1].

These modes of operation are caused by phenomena associated with the production of thermal energy at the cellular level. In particular, cellular water is heated

when the current density is sufficient in the target tissue, causing it to boil and resulting in the rupture of the cell membrane. When this energy is directed along a blade or wire, the result is electrosurgical cutting. At lower current densities, a less intense reaction occurs, resulting in coagulation and drying of the tissue without cutting. Thus, electrosurgery uses the heat generated by the passage of current through tissue to achieve controlled thermal damage to tissues, resulting in cutting and coagulation [1, 6].

For the analysis and design of electronic systems for welding, it is important to study biological tissue as a load. Since complex changes in tissue properties occur during electrosurgery, they require further study.

3 BIOLOGICAL TISSUE AS AN ELECTRICAL LOAD

Impedance is an important characteristic that determines the electrical properties of the objects under study, which may be heterogeneous conductors, and is used to study their structural composition, structural features, and functional parameters [7]. It can change under the influence of alternating current depending on the anatomy of the tissue, composition, or signal frequency [8]. The path of current within the tissue depends on the frequency. At low frequencies, the resulting current path mainly surrounds the cells through extracellular fluids. In contrast, high-frequency alternating current passes through both extracellular and intracellular fluids, enabling the welding process [9].

Since cell membranes act as capacitors, the impedance of biological tissues Z becomes complex and depends on frequency f . Impedance is determined by the formula (3.1) [5]:

$$Z = R + jX, \quad (3.1)$$

where R is active resistance, X is reactive resistance, and j is the imaginary component.

Impedance can also be written in trigonometric form using the formula (3.2):

$$Z = |Z|(\cos\varphi + j\sin\varphi), \quad (3.2)$$

where $|Z|$ is the absolute value of impedance (modulus), φ is the phase.

The absolute value of impedance is determined by the formula (3.3):

$$|Z| = \sqrt{R^2 + X^2}. \quad (3.3)$$

Living tissue is a heterogeneous conductor with resistive capacitive impedance characteristics, which can be modeled by the equivalent circuit shown in Fig. 3.1. The elements of this circuit have physical meanings: R_E is the external resistance of cells, R_I is the internal resistance, C_M is the membrane capacitance [5].

The presence of membrane capacitance makes the impedance Z frequency-dependent and its active resistive R and active capacitive components X [5]:

$$Z = R_E \parallel \left(R_I + \frac{1}{j2\pi f C_M} \right) = \frac{R_E(1 + j2\pi f C_M R_I)}{1 + j2\pi f C_M (R_E + R_I)}. \quad (3.4)$$

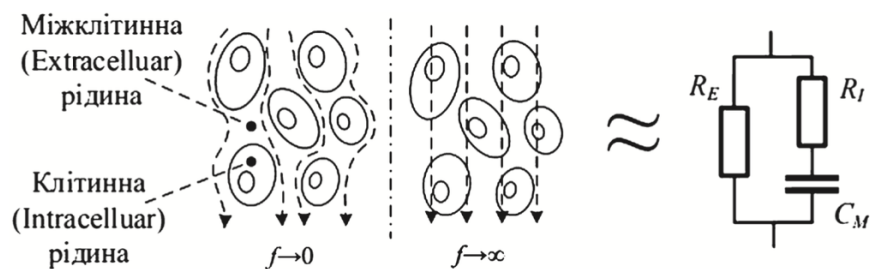


Fig. 3.1. Impedance model [5]

Equation (3.4) is difficult to use for constructing a model of biological tissue as a load. In addition, it is important to take into account changes in impedance during the welding process, as they directly affect the efficiency of welding.

The welding process is based on the use of electrical resistance of tissue, so impedance testing is important for improving the efficiency of this process and preventing possible negative reactions of tissue to the action of electric current. Tissue impedance measurement is used not only in electrosurgery, but also in medical diagnostics, in particular for differentiating between healthy and pathologically altered tissues, assessing organ condition, diagnosing tumors, monitoring physiological status, and developing sensors for detecting infectious and endocrine disorders [8-13].

Electrical impedance spectroscopy is used to measure impedance. This method is useful for the quantitative assessment of cellular changes. It can be used to identify and monitor cellular responses. Electrical impedance spectroscopy provides information about impedance over a wide range of frequencies, which is not available when using other non-invasive diagnostic methods. The principle of this method is as follows: an electrical stimulus is applied to the sample under study, and the response of the sample is measured as a function of the frequency of the applied signal [8, 10, 11]. There are two possible implementations of this method: a known voltage is applied to the sample and the resulting current is measured, or a current is applied to the sample and the resulting voltage drop across the sample is measured [11].

An example of the application of impedance measurement in electrosurgery is the adaptation of the power supplied to the tissue based on the impedance

characteristics of the tissue, which minimizes collateral thermal damage to tissues during cutting. Studies have shown that tissue impedance depends on the depth of electrode penetration and cutting speed. Its changes can be used to regulate the power of the inverter in real time [9].

The accumulated knowledge about impedance will help to better understand the properties of biological tissues during welding and can be used to develop mathematical models describing the behavior of tissues under the influence of electric current. This will allow the design of welding equipment that takes into account changes in the properties of biological tissue, which in turn will contribute to improving welding efficiency.

4 IMPEDANCE MEASUREMENT METHODS

In modern medical monitoring systems for diagnostic studies, measuring transducers are widely used to measure the electrical impedance of organs and tissues, which allows obtaining data on the state of the body [14].

To measure impedance, measuring transducers must have the following characteristics: a frequency ranges from 10^{-2} to 10^6 Hz with a minimum measurement time, since measuring currents can affect the object being measured [14].

The following types of measuring transducers can be distinguished: bridge, potentiometric measuring, vector measuring, distributed parameter line circuits, and impulse impedance measuring circuits [14].

Let us consider each of them in more detail.

4.1 Measuring transducers based on the potentiometric measurement method

There are three types of measuring transducers: two-electrode measuring transducer, transducer with vector meter, and transducer based on a line with distributed parameters [14].

1) Two-electrode measuring transducers measure the voltage drop across the object under study at a given current. Figure 4.1 shows a simplified diagram of a two-electrode measuring transducer based on the potentiometric measurement method. Its principle of operation is as follows: the voltage from the output of generator G is converted into electric current using resistor R_1 and the resistance of the object under study R_X . Provided that $R_1 \gg R_X$, the conversion function can be considered linear [14]:

$$U_{OUT} = \frac{R_X \cdot U_G}{K \cdot R_1} \quad (4.1)$$

The main disadvantage of a measuring transducer with an asymmetrical generator output is the distortion of readings if the object is grounded not at the electrode but at a point. In this case, measuring transducers with a symmetrical

generator output are used. Instead of a resistor R_1 , two resistors $\frac{1}{2}R_1$ are connected to the generator outputs. The signal amplifier has a balanced input. In addition, the symmetrical input eliminates the effect of common-mode interference [14].

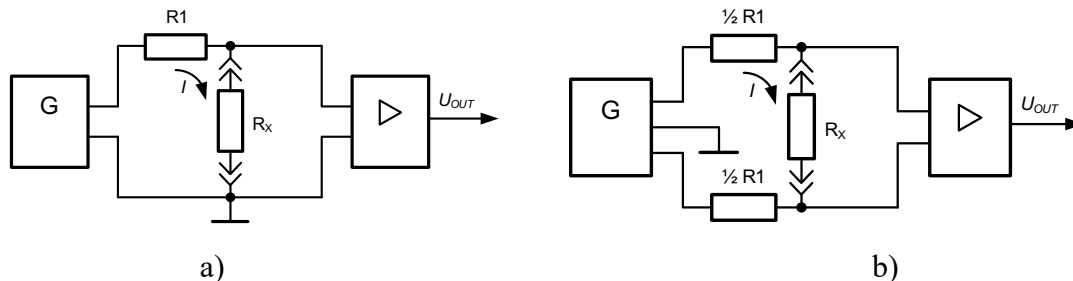


Fig. 4.1. Two-electrode measuring transducer based on the potentiometric measurement method: a) with an asymmetrical generator output, b) with a symmetrical generator output [14].

2) Transducers using a vector meter are used to study the AC impedance of the object under investigation. They allow the generator frequency to be varied over a wide range. Fig. 4.2 shows a diagram of such transducers. Principle of operation: a sinusoidal signal from the generator is fed to the object under study Z ; signals carrying information about the current and voltage at the object are pre-amplified; the amplitude detector determines the effective values of current and voltage, based on which the electrical impedance modulus is calculated; the phase detector determines the phase difference between voltage and current, which allows the impedance components to be determined [14].

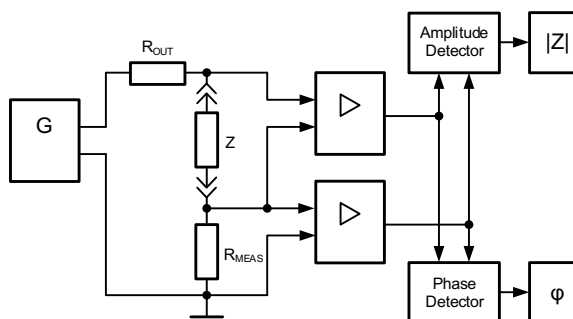


Fig. 4.2. Converter based on a vector meter [14]

3) Measuring transducers based on a line with distributed parameters are used to study the impedance properties of cellular and subcellular structures, which are most

pronounced in the frequency range from 10^6 to 10^9 Гц. Hz. The diagram of such a transducer is shown in Fig. 4.3. Electrical impedance is studied by measuring the ratio between the incident and reflected waves from the biological object under study connected to a distributed parameter line. Conversion function [14]:

$$Z = \frac{U_{INC} + U_{REF}}{U_{INC} - U_{REF}} Z_0, \quad (4.2)$$

where Z_0 is the line resistance; U_{INC} is the incident wave signal; U_{REF} is the reflected wave signal.

If we rewrite formula (4.2) using the reflection coefficient $r = \frac{U_{REF}}{U_{INC}}$, it will take the following form:

$$Z = \frac{1+r}{1-r} Z_0. \quad (4.3)$$

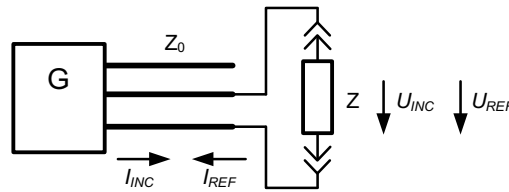


Fig. 4.3. Converter based on a line with distributed parameters [14]

The advantage of such measuring transducers for studying the electrical resistance of biological tissues lies in the simplicity and adequacy of the effect, while the disadvantages include possible changes in the electrical characteristics of the object under study due to the longer dependence of the electric current flow on the applied voltage [14].

4.2 Measuring transducers based on the resonance measurement method

Measuring transducers based on the resonance measurement method are mainly used at high frequencies, due to the fact that resonance phenomena are less pronounced at low frequencies, which leads to a decrease in measurement accuracy. The diagram of such a transducer is shown in Fig. 4.4. The measuring circuit consists of a high-frequency generator G and a measuring oscillating circuit LC , which includes reference

inductance and capacitance. The oscillating circuit LC is powered by a measuring generator, the frequency of which can be smoothly changed until resonance occurs. The moment of resonance is determined by the voltage value, and the resonance frequency is determined by the generator scale. To measure the impedance of the object under study, either a reference inductance or capacitance is connected. To determine the active component, a reference resistance R is connected to the oscillating circuit. Thus, after reaching the resonance frequency, both the active and reactive components of the impedance of the object under study can be determined [14].

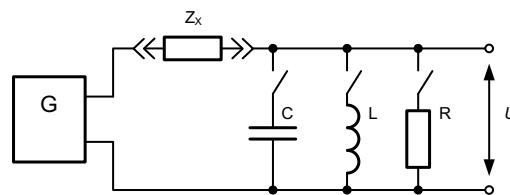


Fig. 4.4. Transducer based on the resonant measurement method [14]

The disadvantage is the use of this type of transducer at high frequencies, as it complicates the determination of the impedance of biological tissues over a wide frequency range [14].

4.3 Measuring transducers based on transfer function analysis

The principle of operation of measuring transducers based on transfer function analysis consists in measuring the response of the object under study to a unit rectangular current pulse of a given amplitude, converting the obtained results from the time domain to the frequency domain using Laplace transformation, and determining the electrical resistance in the required frequency range [14, 15].

The structural diagram of a measuring transducer based on the pulse impedance measurement method is shown in Fig. 4.5. Pulse impedance measurement consists of the following steps [14]:

1. A single stepped pulse of electric current of a given amplitude is applied to the object under study.

2. The response of the system (voltage across the object) is measured and converted from the time domain to the frequency domain using a time-frequency conversion controller.

3. Based on the obtained voltage spectra and the known current pulse spectrum, the frequency characteristic of the electrical impedance of the object under study in the required frequency range is determined.

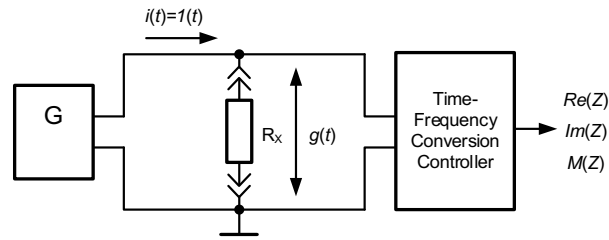


Fig. 4.5. Converter based on the pulse impedance measurement method [14]

The conversion function of such a converter is as follows [14]:

$$Z(j\omega) = j\omega \cdot F\{g(t)\} \quad (4.4)$$

Thus, the frequency response of the impedance of the object under study can be obtained based on the voltage across the object when it is subjected to a test current pulse [14].

The advantages of this method are: the ability to determine the impedance parameters of the object under study in a wide frequency range; short measurement time; rapid application of the test current to the object under study [14].

4.4 Measuring transducers based on bridge measurement methods

Measuring transducers based on bridge measurement methods consist of a test signal source, a bridge measuring circuit, and an amplifier [14].

Fig. 4.6 (a) shows a diagram of a measuring transducer designed to measure active resistances, and Fig. 4.6 (b) shows a diagram of a measuring transducer designed to measure the resistive and capacitive components of the impedance of the object under study [14].

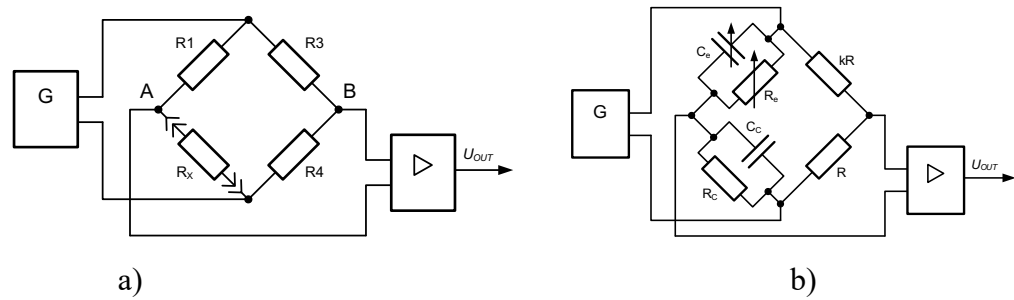


Fig. 4.6. Converter based on bridge methods: a) for measuring active resistances, b) for measuring resistive and capacitive components of impedance [14]

Bridge measuring circuits operate on the principle of bridge balancing. A test signal is applied to the input of the bridge circuit, which consists of the unknown impedance of the object under test and known reference impedances. By adjusting the parameters of the reference impedances, a bridge balance state is achieved in which the bridge output signal becomes zero or minimal. The value of the unknown impedance is determined based on the values of the reference impedances in the balance state [14].

The disadvantages of measuring transducers based on bridge methods include a small dynamic measurement range, the dependence of measurement accuracy on the sensitivity of the circuit to determining the bridge balance, and the dependence of measurement results on the power supply voltage. Despite this, their advantage is the high accuracy of impedance measurement at different frequencies [14].

Considering the advantages and limitations of various measuring transducers, a transducer based on the pulse impedance measurement method was deemed optimal. This method offers a short measurement time, which is particularly important for assessing the impedance of biological tissues. Since the bioelectrical impedance of such tissues varies across a wide frequency range, traditional methods like bridge and potentiometric measurements are less suitable due to their longer measurement durations. For further research, a two-electrode circuit was selected. Given the high currents and voltages involved in electrosurgical procedures, a simpler measurement approach is more appropriate. The simplicity of the circuit enables direct measurement of key parameters under experimental conditions [14].

5 EXPERIMENTAL IMPEDANCE MEASUREMENT

5.1 Structure of the installation

An experimental setup was created to determine the impedance properties of biological tissue during welding. It is shown in Fig. 5.1. It consists of an electrosurgical unit, a measuring shunt, a TDS2012C oscilloscope, a passive electrode (100x100 mm copper plate), and an active electrode (5x6 mm copper rod) with a fixed contact force. The test sample is muscle tissue from a freshly slaughtered pig. The test sample is placed between the passive and active electrodes. The setup allows measuring the phase shift between the voltage applied to the welded tissues and the current passing through them at the operating frequency of the electrocoagulator [14, 16].

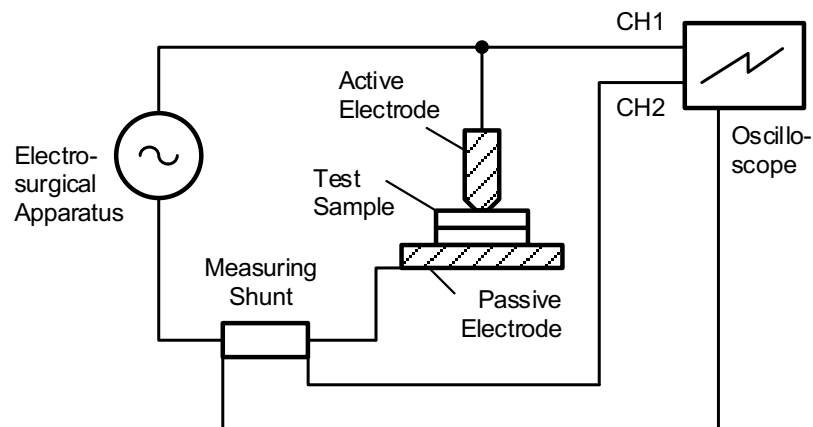


Fig. 5.1. Experimental setup [14, 16]

Fig. 5.2 shows a simplified structural diagram of an electric welding machine. The input module rectifies and filters the voltage from the power supply. The transistor regulator provides a stable voltage level for the high-frequency inverter, ensuring the voltage with the necessary parameters for the welding electrodes. Current meters and voltage meters normalize the measured values. CS is a current sensor [14].

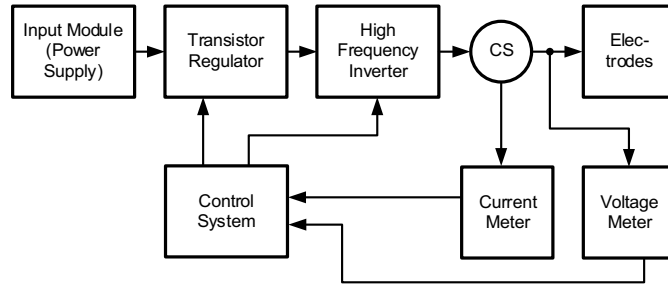


Fig. 5.2. Simplified structural diagram of an electric welding machine [14]

5.2 Installation calibration

Calibration was performed to reduce the influence of parasitic inductances of the elements and wires of the installation. The main calculations are given below [14].

1) Use of equivalent load.

To minimize parasitic inductance, a measuring shunt with a nominal value of 2.5 ohms was assembled from 17 MF02SFF metal film resistors of 43 ohms each, connected in parallel. This connection method reduces parasitic effects [14].

2) Calculation of parasitic inductances.

The total length of the connecting wires inside the electrocoagulator housing is 15-20 cm; outside the housing, no more than 15 cm. According to formula [17], the total inductance of the conductors is approximately 342-503 nH [14]:

$$L_{tot} = 2l \left(\ln \left(\left(\frac{2l}{d} \right) \left(1 + \sqrt{1 + \left(\frac{d}{2l} \right)^2} \right) \right) - \sqrt{1 + \left(\frac{d}{2l} \right)^2} + \frac{\mu_0}{4} + \frac{d}{2l} \right) \quad (5.1)$$

where L_{tot} is the total inductance; l is the length of the wires; d is the diameter of the wire; μ_0 is the magnetic constant.

The parasitic inductance of the metal film resistors L_{Rp} , used in the experiment, is approximately 2 nH [18]. The measured inductance of the component terminals L_{pin} ranges from 2 to 7 nH. The total inductance of the resistor L_{Rto} [14]:

$$L_{Rto} = L_{Rp} + L_{pin} = 4 \dots 9 \text{ nH.}$$

For the shunt section, inductance is defined as:

$$L_{shunt} = \frac{L_{Rto}}{17} = \frac{4 \dots 9}{17} = 0,24 \dots 0,53 \text{ nH.}$$

3) Calculation of phase shift.

Phase shifts between current and voltage caused by parasitic parameters were measured using an equivalent load instead of the test sample [14].

The equivalent resistance consisted of three sections connected in series, each containing 17 parallel-connected MF02SFF metal film resistors with a nominal value of 430 Ω . Taking into account previous calculations, the parasitic inductance of three sections connected in series is 0.72...1.59 nH, and the total inductance is 342.93...504.77 nH [14].

The reactive resistance of the circuit was calculated using the formula $X = 2\pi fL$; the phase shift was calculated using the formula $\varphi = \arctan\left(\frac{X}{R}\right)$. The calculation results are shown in Table 5.1 [14].

Table 5.1

Results of calibration parameter calculations

f , MHz	R , Ω	X , Ω	φ , $^\circ$
0.44	75	0.95...1.39	0.72...1.07
1.76	75	3.78...5.58	2.89...4.26
3.50	75	7.52...11.10	5.72...8.42

To verify the calculation results, an experimental study of phase shifts caused by parasitic inductances of the measuring device elements was conducted. Figure 5.3 shows the obtained oscillograms of the voltage across the shunt (channel 1, orange line) and the shunt current (channel 2, blue line) with time scales of 500 ns/cell (a) and 50 ns/cell (b) for a frequency of 0.44 MHz, a power level of 50 u.o. (1 u.o. \approx 1 W), and an exposure time of 2 s [14].

Table 5.2 shows the results of measuring the impedance properties of a circuit loaded with equivalent resistance. As can be seen from Table 5.2, the measured values of phase shift φ for frequencies of 0.44 MHz, 1.76 MHz, and 3.5 MHz are within the calculated values [14].

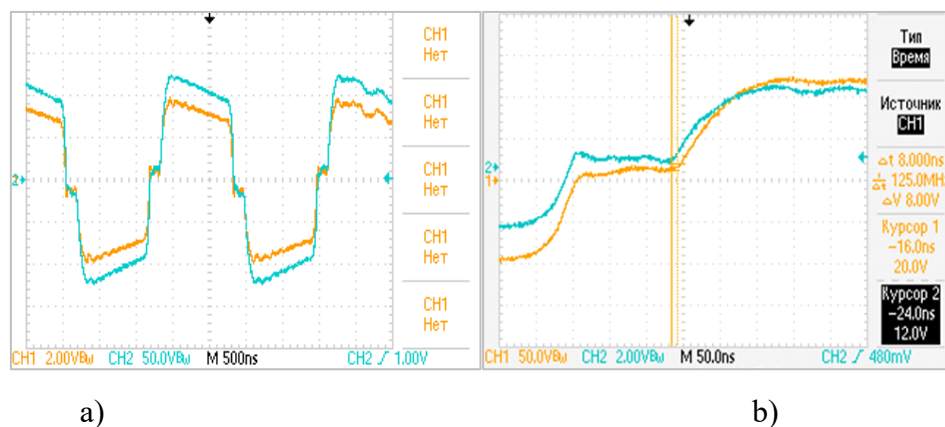


Fig. 5.3. Oscillograms of electromagnetic processes in an electrocoagulator with time scales of a) 500 ns/cell and b) 50 ns/cell for a frequency of 0.44 MHz and a power level of 50 u.o. [14]

Table 5.2

Results of calibration parameter measurements

f , MHz	R , Ω	U , V	I , A	φ , °
0.44	75	125	1.60	1.27
1.76	75	130	1.76	3.05
3.50	75	135	2.20	7.71

5.3 Experimental research

Experimental studies have made it possible to determine the phase shift at different power levels and durations of electrosurgical exposure. The object of the study was 5-7 mm thick sections of muscle tissue from a freshly slaughtered pig. Fig. 5.4 shows the oscillograms of voltages on a measuring shunt with a load for a frequency of 0.44 MHz, a power level of 50 conventional units, an exposure time of 2 s, with scales of a) 250 ns/cell, b) 50 ns/cell [14, 16].

The results of the study are presented in Appendix A Table 1 for a frequency of 0.44 MHz, in Table 2 for a frequency of 1.76 MHz, and in Table 3 for 3.5 MHz [14, 16].

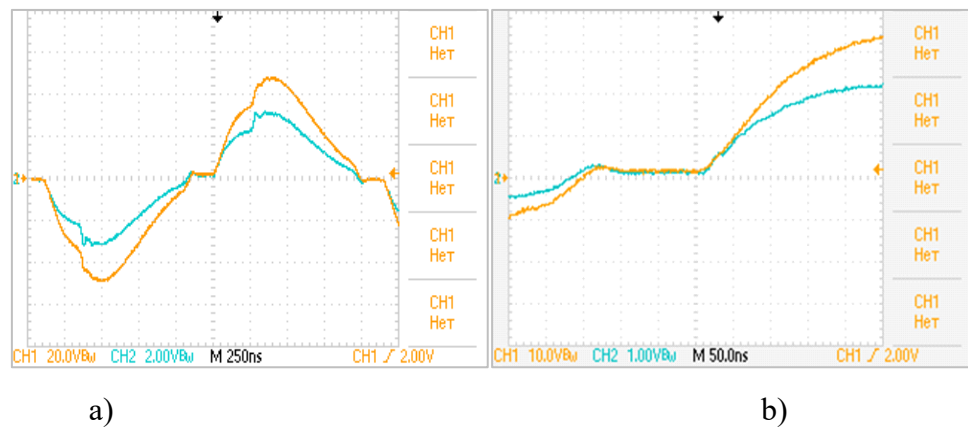


Fig. 5.4. Oscillograms of voltages on a measuring shunt with a load for a frequency of 0.44 MHz, a power level of 50 conventional units, and an exposure time of 2 s with scales of a) 250 ns/cell, b) 50 ns/cell [14, 16]

Since the phase shift between the current passing through the tissues under study and the voltage at the electrodes obtained during experimental welding of biological tissue samples at the operating frequencies of the electrocoagulation device (0.44 MHz) does not exceed one electrical degree for different values of power and duration of the electrosurgical procedure. This allows us to consider the nature of the impedance of biological tissue during welding to be active when solving problems of its modeling as a load on the electrocoagulator. This approach makes it possible to use lower-order equations for calculations, significantly simplifying mathematical models [14].

CONCLUSIONS

In the course of the work, literature on the subject of electric welding of tissues was analyzed, which made it possible to outline the current state of research in this field. In particular, it was found that due to the complexity of the processes occurring in the cell during welding, the differences between living and non-living tissue complicate research and require further study of the parameters of different types of tissue and their changes.

Particular attention was paid to tissue impedance, since the welding process is based on the use of the electrical resistance of tissue. Although there are articles in the literature related to the measurement of impedance and its use in electrosurgery, these studies focus on the use of impedance as a tool for monitoring and adapting electrosurgical processes, without directly analyzing changes in tissue properties during the welding process.

During experimental studies of the phase shift between voltage and current in the biological muscle tissue of a freshly slaughtered pig, it was found that it does not exceed one degree for a frequency of 0.44 MHz. This allows this parameter to be neglected when constructing mathematical models and to consider the tissue only as a model with active resistance. This approach greatly simplifies the mathematical modeling of the system through the use of lower-order equations and facilitates the analysis of the processes occurring during welding.

The results obtained are important for the design and testing of tissue welding equipment. Taking into account the effect of the welding process on tissue also contributes to the development of more accurate operating modes, which will improve the quality of surgical interventions.

Further research on this topic may include studying the impedance properties of different types of tissue, as well as developing equipment for accurate impedance measurement during electrosurgical welding and adapting equipment parameters for better tissue connection results.

LIST OF REFERENCES

1. Тканиннозбережна високочастотна електрозварювальна хірургія: Атлас / за ред. Б. Є. Патона, О. М. Іванової. Київ : Наукова думка, 2009. 200 с.
2. Philippson M., “Les lois de la résistance électrique des tissus vivants,” Bull. Cl.Sci. Acad. R. Belg., ser. 5, vol. 7, no. 7, pp. 387–403, Jul. 1921.
3. Elbourne S. Cell and body tissue physiology. British Journal of Nursing. 2022. Vol. 31, no. 13. P. 696-702. URL: <https://doi.org/10.12968/bjon.2022.31.13.696>.
4. Гістологія людини / О. Д. Луцик та ін. Київ: Книга плюс, 2003. 592 с.
5. Dubko A., Yamnenko I., Stepenko S., Bondarenko O. Impedance analysis of biological tissues for their welding. 2023 IEEE 4th KhPI Week on Advanced Technology (KhPIWeek). 2-6 October 2023, Kharkiv, Ukraine / IEEE. URL: <https://doi.org/10.1109/KhPIWeek61412.2023.10312990>.
6. Benias P.C, Carr-Locke D.L. ERCP Third edition. 11 – Principles of Electrosurgery // edited by Baron T. H., Kozarek R.A., Carr-Locke D.L. 2019. P. 86-92.
7. Grimnes, S. Bioimpedance and Bioelectricity Basics. JG Martinsen – San Diego, CA: Academic Press, 2000. – 749 p.
8. Bera T. K. Bioelectrical Impedance and The Frequency Dependent Current Conduction Through Biological Tissues: A Short Review. 3rd International Conference on Communication Systems (ICCS-2017). 14-16 October 2018, Rajasthan, India / IOP Conference Series: Materials Science and Engineering. URL: <https://doi.org/10.1088/1757-899X/331/1/012005>.
9. Bao C., Mazumder, S.K. Output Power Computation and Adaptation Strategy of an Electrosurgery Inverter for Reduced Collateral Tissue Damage. IEEE Transactions on Biomedical Engineering. 2023. Vol. 70, no. 6. P. 1729–1740. URL: <https://doi.org/10.1109/TBME.2022.3225271>.
10. Dean D., Thillaiyan R., Machado D., Sundararajan R. Electrical Impedance Spectroscopy Study of Biological Tissues. Journal of Electrostatics. 2008. Vol. 66, no. 3-4. P. 165-177. URL: <https://doi.org/10.1016/j.elstat.2007.11.005>.

11. Bounik R., Cardes F., Ulasan H., Modena M.M., Hierlemann A. Impedance imaging of cells and tissues: Design and applications. BME Frontiers. 2022. Vol. 2022, no. 1. P. 1-21. URL: <https://doi.org/10.34133/2022/9857485>.

12. Abasi S., Aggas J.R., Garayar-Leyva G.G., Walther B.K., Guiseppi-Elie A. Bioelectrical Impedance Spectroscopy for Monitoring Mammalian Cells and Tissues under Different Frequency Domains: A Review. ACS Measurement Science Au. 2022. Vol. 3, no. 1. URL: <https://doi.org/10.1021/acsmesuresciau.2c00044>.

13. Дуваров Ю., Калашнікова Л., Дубко А. Застосування імпедансометрії для визначення електрофізичних параметрів м'яких біологічних тканин. 2024. Біомедична інженерія і технологія. №16. URL: <https://doi.org/10.20535/.2024.16.317937>.

14. Dziuba I., Bondarenko Y., Koroliuk T., Dubko A., Stepenko S., Bondarenko O. Experimental Determination of Biological Tissue Impedance for Electrosurgical Process. 2024 IEEE 6th International Conference on Modern Electrical and Energy Systems.

15. Grimnes S. and Martinsen Ø.J. Bioimpedance and Bioelectricity Basics. 3rd edition. Academic Press. 2014. URL: <https://doi.org/10.1016/C2012-0-06951-7>.

16. [REDACTED]

[REDACTED]

17. Coil Inductance Calculator & An Essential Guide for Electrical Engineers <https://www.keysight.com/used/us/en/knowledge/calculators/inductance-calculator>.

18. Grover F.W. Inductance Calculations: Working Formulas and Tables. Dover Publications, 2004, 286 p.

APPENDIX A. The results of the experiment

Table 1

Results of the experiment for 0.44 MHz

Exposure time t, s	Output power P	Voltage U, V	Average current I, A	Average resistance R, Ohm	Phase shift between voltage and current φ , °
0,12	10	22	0,80	27,50	<1
0,22		20	0,76	26,30	
0,60		21	0,80	26,25	
1,10		20	0,80	25,00	
2,40		18	0,80	22,50	
6,00		21	0,78	26,92	
20,0		19	0,72	26,38	
35,0		18	0,76	23,68	
0,12	50	44	1,60	27,50	
0,22		38	1,60	23,75	
0,60		35	1,60	21,88	
1,10		30	1,60	18,75	
2,40		24	1,60	15,00	
6,00		23	1,45	15,87	
20,0		32	0,40	80,00	
0,12	280	70	3,60	19,44	
0,22		60	3,60	16,67	
0,60		50	3,60	13,89	
1,10		50	3,65	13,69	
2,40		50	3,70	13,51	

Table 2

Results of the experiment for 1.76 MHz

Exposure time t, s	Output power P	Voltage U, V	Average current I, A	Average resistance R, Ohm	Phase shift between voltage and current φ , °
2	25	21	0.60	35.00	2.54
4		21	0.64	32.81	2.54
10		19	0.64	29.69	2.54
30		17	0.68	25.00	2.54
2	50	33	1.00	33.00	3.16
10		28	1.08	25.93	2.54
2	75	38	1.20	31.67	3.81
4		30	1.20	25.00	3.16
10		40	1.20	33.33	3.81

Table 3

Results of the experiment for 3.5 MHz

Exposure time t, s	Output power P	Voltage U, V	Average current I, A	Average resistance R, Ohm	Phase shift between voltage and current φ , °
2	25	30	1.00	30.00	1.26
4		25	1.00	25.00	<1
10		16	1.00	16.00	1.26
20		20	0.88	22.73	2.53
2	50	42	1.68	28.57	5.05
4		56	1.60	35.00	6.32
10		40	1.60	25.00	3.79
2	75	40	1.60	20.00	3.16
4		50	1.60	25.00	1.89