INTERNATIONAL JOURNAL OF RESEARCH AND DEVELOPMENT IN ENGINEERING SCIENCES
e-ISSN: 2582 - 4201 Volume: 2 | Issue: 1 | Year: 2020

INVESTIGATIONS ON THE CHARACTERISTICS OF BIO-TISSUES FOR ELECTROMAGNETIC ENERGY

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Abstract: Human body is one of the most beautiful and complex systems in nature. It is characterized by non-homogeneous medium characteristics. In particular the conductivity, permittivity and permeability vary from location to location within the body. As body consists of different tissues like fat, muscles, liver, heart, skin and bone etc., they have different fundamental parameters. When there is a tumor within the body different techniques are adopted for the treatment. It is also possible to treat tumors like cancer by the application of electromagnetic energy. However, the electromagnetic waves penetrate different tissues only to a limited distance.

In case of brain tumors it has been practice to apply ultrasonic energy or Electromagnetic energy after removing skull, but Electromagnetic energy can be applied for treating such tumors without removing the skull. As Electromagnetic energy is at a certain frequency penetrates through the skull. In order to develop a suitable source of such energies the effects of Electromagnetic energy on different tissues are investigated thoroughly in the present work.

The work is presented in five chapters. In first chapter, an exhaustive survey of literature is made an Electromagnetic energy, bio tissues and their behavior. Second chapter deals with the computed results on the absorption properties of biological tissues. In chapter three penetration characteristics of different tissues are investigated. Equivalent network of human body and its interpretations are presented in chapter four. In chapter five impedance characteristics of biological tissues are presented.

The investigations carried out in the present work are extremely useful for the design of Electromagnetic energy sources for the treatment of malignant tumors. The data presented is exhaustive and in the future course of work collaboration with some of the hospitals will be undertaken for respective measurements.

Keywords: Bio-Tissues, Electromagnetic Energy, Tumors.

1. INTRODUCTION

The electromagnetic waves are generated from alternatively excited conductors. They can be in the form of wire, rod or planar conductors. High frequency/excitation produces electromagnetic waves with lower wavelength. On the other hand, low frequency excitations produce Electromagnetic waves of higher wavelength. Such Electromagnetic waves are characterized by several properties.

It is well-known from the open literature that radio frequency electrical signals above 10 KHz are not found to excite nerves or muscles [1]. In view of this, the electro surgery which is a major therapeutic model, Hald [3], and Schwan raised some contradictions on the above findings. This is due to the existence of minor skeletal muscle excitations. The muscle spasm, cardiac ventricular fibrillation and pacemaker malfunction are described during electro surgical applications. These are attributed to rectification effects of the electrode tissue interface [2].

It is confirmed that the electrical neuromuscular excitation can be induced at the frequencies above 10 KHz if the current density is sufficient. Some experiments are carried out by Lacourse et al., [7] by applying minimum currents to produce direct muscle stimulations as a function of frequency. Frequency is changed from 100Hz to 50 KHz. From the results obtained, it is found that minimum current range between 0.5mA to 500mA. As frequency is increased the minimum required current is also increased. The frequency of electrosurgical output current should be higher than the range at which shock on neuromuscular stimulation can occur.As reported in the open literature [4] for frequency above 10 KHz the predominant tissue effect is the generation of heat without muscle excitation. In view of this, the frequency above 100 KHz is an accepted surgical value. However the most commercial electro surgical units operate in the range of 100 KHz to 3MHz.

2. PRELIMINARY INVESTIGATION

Electromagnetic waves are in general non-ionizing; they are extremely useful to induce hyperthermia for cancer treatment. Some types of devices are used to induce local hyperthermia which is based on the wave guides [5]. They are multi model and ridged applicators [7]. Johnson et al., described low profile applicators for local heating of tissues. The energy storage takes place in the near field region of an antenna of lossless medium [8]. Where in the E and H fields are in true quadrature, thus creating a reactive loading effect; only the far field contributed to loss by radiation. When the medium is

lossy, power is also extracted from the near field region this occurs when hyperthermia applicators are in direct contact with the tissue. The near field zone is a function of frequency for antennas in the free space [3]. The near fields to small probes are analyzed by Anderson et al., the operating frequencies of an applicator is reduced the near field contribution to tissue heating becomes increasingly dormant for given geometry.

The pattern of energy deposition when radio-frequency applicators are used for treatment of cancer by hyperthermia is computed by Hagmann [5]. A 180 cell block model of man is used for this purpose. The pattern of aberrant heating is found to depend upon the positions of arms and legs. It appears to be much less pronounced for treatment of the thigh or upper arm than it is for treatment of the abdomen. Clinical attempts at intracranial hyperthermia have been reported by Heimburger et al., are using external Ultra sound system. Samaras et al., used an implanted microwave antenna system. To administer intra cranial hyperthermia for brain tumor therapy, the following problems happen, those all are solved by Samaras they are:

- Heat delivery
- Temperature measurement
- Temperature control
- Pre and post treatment estimation of spatial thermal field distributions.

Silberman et al., investigated the feasibility of using the magnetic loop induction for non-invasive intracranial hyperthermia. Theoretical and empherical methods are used for the above purpose. Bini et al., [7] proposed polyacrylamide for simulating electrical behaviour of biological tissues. It is found to have good optical transparency, solid elastic, easily shapeable. It is also low cost and readily available; its thermal properties are investigated. The power transferred to a biological material is represented by an RC load. The equivalent-circuit parameters R and C are related to the material properties. Bio electric current in one dimensional tissues is measured by using a room temperature amplifier and a wire-wound, ferrite-core toroid to detect the associated biomagnetic field. The characteristics of toroid amplifier are presented by Gielen et al., [8]. The studies are made on the above system for measuring current in cardiac and skeletal muscle. Factors affecting the sensitivity of openable toroids are also considered. Although the circuit is valid for individual tissues, it is felt that it can also be applied to the whole body. The parameters Rm and Cm can be obtained from the measured values or parallel body resistance R_P as a function of frequency as follows.

In the frequency range 10-500 kHz, at 10 kHz, C_m represents an open circuit

$$R_P = R_e$$

This equals the measured parallel 'resistance at 10 kHz. At 500 kHz, C_m represents a short circuit

$$R_P = \frac{R_e R_m}{R_e + R_m}$$

This equals measured parallel resistance at 500 kHz. Therefore, Rm can be calculated from (2). The equivalent parallel body resistance R_P and capacitance C_P are represented by (3) and (4) in the frequency range 10-500 kHz .

$$R_P = \frac{R_e\left(1 + w^2 C_m^2 R_m^2\right)}{1 + w^2 C_m^2 R_m \left(R_e + R_m\right)}$$

$$C_p = \frac{C_m}{1 + w^2 C_m^2 R_m^2}$$

Here, $w = 2\pi f$ is the applied frequency. At the β-dispersion frequency w_o

$$w_0 R_m C_m = 1$$

$$R_p \mid_{w=w_o} = R_0 \frac{2R_e R_m}{R_e + 2R_m}$$

Here R_0 is output resistance of body

The frequency corresponding to R_0 can be obtained by interpolation from the measured values of parallel resistance R_P versus frequency. Knowing w_0, C_m can be obtained from [7,5]. C_b and R_b can be similarly calculated. There is a clinical need for a new type of compact applicator that has a low profile, low leakage, and efficiently transfers electromagnetic power to tissue without requiring bulky electrical matching components.

3. EXISTING SYSTEM

The calculations show that the greatest differential absorption occurs at frequencies between the dominant relaxations frequencies in the two tissues. From rat mammary gland tumour data, the calculations show an optimum frequency range of about 100-500 MHz for microwave hyperthermia treatment of at least these types of tumour. Theoretical Analyses of Electromagnetic Fields in Biological Tissues: One of the most vexing problems in studies involving the interaction of Electromagnetic fields and living biological systems and tissues was the quantification of the fields induced in the tissues by nearby sources. They described a method for rapid evaluation of these fields in tissues of arbitrary shape and characteristics when they were exposed to various sources including plane wave, aperture, slot, and dipole sources. The method, valid for both far- and near-zone fields, involves the use of a thermograph camera for recording temperature distributions produced by energy absorption in phantom models of the tissue structures. The magnitude of the electric field may then be obtained anywhere on the model as a function of the square root of the magnitude of the calculated heating pattern [4].

Electromagnetic Characteristics of Human Tissues: The dielectric properties of body tissues at radio frequencies and microwave frequencies had been examined by several authors. It is not possible to make measurements on live tissue, because of the need to hold the measuring device in contact with the tissue. For this reason, all studies use dead tissue. In excised animal tissue, mostly ovine, from freshly killed sheep, along with human autopsy materials is used. In addition, human skin and tongue in vivo is measured since the dielectric parameters vary significantly with frequency, a model based on the summation of the expression shown below:

$$\varepsilon(\omega) = \varepsilon_{\infty} + \sum_{m=1}^{4} \frac{\Delta\varepsilon_m}{1 + (j\omega\tau_m)^{(1-\alpha_m)}} + \frac{\sigma_j}{j\omega\varepsilon_0}$$

j is the ionic conductivity and *m* are material parameters for each dispersion region. There were significant differences in relative permittivity and conductivity across a wide frequency range. However, it can be seen that the variations in the range from 100MHz to approximately 1GHz were relatively small. There were also variations in tissue properties with age because of the variation in water content. However, the penetration depth does change significantly across this frequency range. At 100MHz, penetration depths were significant, and thus frequencies in this range were used for penetration into the body for communications with medical implants. As the frequency increases, the penetration depth reduces [6].

Electromagnetic Radiation Hazards

The depth of penetration into and subsequent absorption of a wave by the body were determined by the relative dielectric constant (ϵ_r) and conductivity (u) of the body tissues. These were shown as a function of frequency. The absorption of energy was determined by the attenuation constant. Because the body was a good conductor, the skin depth was the relatively large and of the body results in a significant reflection of the incident wave at the body/air boundary. Understanding Penetration Depths of Different RF Modes. Radiofrequency energy is a form of electromagnetic energy. When applied to tissue, it produces electromagnetic fields, causing the oscillation of molecules within the tissue, which results in the generation of heat. Radiofrequency can be applied to tissue by using either two points on the tip of a probe (bipolar) or between a single electrode tip and a grounding plate (monopolar). Since bipolar energy is more localized less energy is required to achieve the same heating effect. Radiofrequency aesthetic effects are based on mild heating of the skin's underlying network of collagen and elastin fibers which can lead to collagen shrinkage and thickening, improving skin's firmness and elasticity [5].

The penetration depth of RF energy depends on the electromagnetic wave frequency, the permeability of the tissue and the tissue's conductivity. This dependence is determined by the following equation:

$$\delta \approx \frac{1}{\sqrt{\pi f \mu \sigma}}$$

here : δ = Standard Depth of Penetration (mm)
f = Test Frequency (Hz)
μ = Magnetic Permeability (H/mm)
σ = Electrical Conductivity (% IACS)

From this equation it can be seen that the depth of penetration in millimeters is inversely proportional to the square root of the frequency. Hence, lower frequencies have a higher penetration rate. Electric currents flowing through a human body exhibit varying effects ranging from little or no perceptible effect to shocking sensation, to possibility of current flow. The effect of the current flow is a function of the applied voltage level, the duration of exposure, and the resistance and impedance of the body and the other available current paths. In General, the longer the contact duration, the higher the voltage and or lower the impedance of the body, the greater the chances that the electric current will exceed the level necessary for human or animal perception. The voltage level is significant in determining whether or not an energized object will have the ability to deliver current through a body [8].

The complete impedance of the model is the services summation of the individual impedances for each of the sections. The electromagnetic energy deposited in different tissues of each section can be obtained from the corresponding impedances and current. Dielectric Properties of VX-2 Carcinoma versus Normal Liver Tissue.

The dielectric properties of the tumor and normal tissue are distinctly different, by factors of 6-7.5 in the conductivity and 2-5 in the permittivity. These differences are striking in the figures, even considering the compressed logarithmic scale that is employed. The permittivity of the normal tissue falls off considerably faster than that of the tumor with frequency above 100 kHz.

4. PROPOSED SYSTEM

The development of models to represent the response of the human body is involved and complex. In the simplest model, two types of impedances are used:

- Skin impedance
- Internal impedance.

The skin is a layered structure, with both resistance and capacitance. The resistance is non-linear in voltage and time. The impedance of the material interior may be considered as resistive, comparable to a similar volume of saline at human body concentration. The skin capacitance causes impedance to decrease with frequency, as AC current shunts the high skin resistance. The impedance of the human body can be broken down into the impedances of the various body parts, resulting in an equivalent circuit for the electrical path through the body[319].

The skin resistance is dependent upon the portion of the body (thicker or thinner), the wetness of the skin, and the area and pressure of contact. Skin resistance is divided into two parts, the surface layer parallel resistance, R_p, series resistance R_s. The spreading resistance covers the zone where the current spreads out into the body from the contact area. Skin resistivity (R_p+R_s) is in the range of 60-1200KW-cm^2 skin resistance decreases with the time of contact. After approximately 20 min, the resistivity of the dry and wet skin becomes equal, at about 2/3 of wet skin resistivity.

The human body may be considered as a resistive, piecewise homogeneous and linear volume conductor. Most of the tissue is isotropic. The muscle is, however, strongly anisotropic, and the brain tissue is anisotropic as well. Table 4.1 summarizes the tissue resistivity values of a number of components of the human body. Resistivity values for various tissues. The objective of a brain stimulator is to interact with neural tissue in order to realize few required activity in the brain. The interaction between the electrical stimulator and the neural tissue is selected to take place in the electrical domain. When employing a stimulator it is of special importance to have a thorough understanding of how neural tissue works and more

even important: how it answers to electrical impulses. A fundamental understanding of communication within the nervous system is required. Communications via the human nervous system eventuate by a propagation of nerve signals or action potentials from nerve fiber to the neural pathway [10].

Tissue	Resistivity ρ[Ωm]	Remarks
Brain	2.2	Gray matter
	6.8	white matter
Cerebrospinal	5.8	average
fluid	0.7	*Hct* = 45
Blood	1.6	longitudinal
Plasma	0.7	transverse
Heart muscle	2.5	longitudinal
Skeletal muscle	5.6	transverse
Liver	1.9	
Lung	13.2	
Fat	7	longitudinal
Bone	11.2	circumferential
	21.7	radial (at 100 kHz)
	25	
	177	
	15	
	158	
	215	

Table.1: List of Resistivity of different tissues in human body

Bioelectric Source and its Electric Field: The human body may be considered as a resistive, piecewise homogeneous and linear volume conductor. Most of the tissue is isotropic. The muscle is however, strongly anisotropic and the brain tissue is anisotropic as well. The resistivity of blood depends strongly on the hematocrit, Hmct (which denotes the percent volume of the red blood cells in whole blood) [320]. This dependence has an exponential nature and was given in Equation 4.1

$$\rho = 0.537\, e^{0.025\text{Hmct}}$$

Hugo Fricke studied theoretically the electric conductivity of a suspension of spheroids [321]. When applying this method to the conductivity of blood, we obtain what was called the Maxwell-Fricke equation:

$$\rho = 0.586\frac{1+0.0125\text{Hmct}}{1-0.01\text{Hmct}}$$

where ρ = resistivity of blood [Ωm]

Hmct = hematocrit [%]

Both of these equations give very accurate values. The correlation coefficient of Equation 4.1 to empirical measurements was r = 0.989. Because the best fitting curve to the measured resistivity values was slightly nonlinear in a semi logarithmic plot, Equation 4.2 gives better values with very low or very high hematocrit values. The resistivity of blood was also a function of the movement of the blood [332]. This effect is often neglected in practice.

Passage of Stimulus Current Through the Tissue: This type of damaging process is found to be the most critical process. The damage is related with the axonal activity and or the depolarization and hyper polarization of the cell membrane. However, it is not fully understood what process within the cells is responsible for the induced damage. It is only known that this type of damage is related to neuronal activity.

Regulated Voltage and Regulated Current Stimulation: The electronic circuit used to deliver the applied stimulus may be either a constant voltage device or a constant current device, and this will have a direct impact on the properties of excitation. In general, regulated current stimulators should be used, as this enables direct control of the extracellular electric field. However, consideration of the risk for tissue damage suggests that regulated voltage stimulators may be more appropriate for surface stimulation [4].

Tissue Damage

Clinical implementations of neural prostheses have not encountered decreased prosthesis performance as a result of tissue damage. However, continued safe use of these devices requires an understanding of the mechanisms responsible for tissue damage. Most generally, tissue damage can be classified as either passive or active, although it is not always possible to differentiate between these two damage modes. Passive damage can result from surgical trauma and chemical and mechanical bio-incompatibility of the implanted device. Active damage results from electrochemical reaction products formed at the tissue-electrode interface and from physiological changes associated with neural excitation. Currently, noble metal conductors and medical grade synthetic insulators, combined with exacting cleaning procedures ensure a minimal tissue response [8].

These materials are also able to withstand the harsh environment within the body without degradation. Mechanical damage may result from surgical trauma such as nerve stretching or from the presence of the electrode(s). Furthermore, damage to peripheral nerves may result from compression during postsurgical edema, or from stretching if the electrode becomes anchored by scar tissue. Similarly, electrode arrays implanted on the surface of the brain have resulted in local mechanical damage including tissue compression, mening eal fibrosis, and molecular layer and fiber deformation. Insertion of penetrating microelectrodes may result in vascular as well as neural damage. By minimizing electrode diameter, and beveling rather than sharpening the electrode tip, mechanical trauma has been minimized.

Tissue damage due to the passage of current may arise from electrochemical reaction products formed at the electrode-tissue interface and/or from physiological changes in the neural and surrounding tissue that are associated with neural excitation. The finding that charge per phase and charge density are co-factors in determining the threshold for neural injury by stimulation in both the peripheral nervous system and the central nervous system suggests that both electrochemical and physiological mechanisms contribute to neural damage. The number and spatial extent of neurons activated are related to the charge injected in each stimulus pulse, whereas the charge density contributes to the type and rate of electrochemical reactions that occur at the electrode-tissue interface [2].

Tissue damage may result from the products of electrochemical reactions at the electrode-tissue interface where the charge carriers change from electrons in the metal to ions in the tissue. Reactions that take place are dependent on the potential of the electrode as well as the time course of the stimulus pulse, and may be accelerated by increased current density. The electrode interface can be modeled by the parallel combination of a capacitor (C), representing the double-layer capacitance, and a series of diode-like elements, each representing an electrochemical reaction. The voltage

developed across the electrode-tissue interface (Ve) is determined by the amount of charge in the stimulus pulse (Q), because

V = QIC

V= voltage across the interface

Q= amount of charge in the stimulus pulse

I= current density

C= parallel combination of a capacitor

The voltage across the interface determines which chemical reactions will take place or, in the simple model, which diodes will turn on. The electrode capacitance is determined by the properties of the material and is proportional to the electrode area. Therefore, the potential developed across the interface is proportional to electrode area. This relationship is the basis for the correlation between charge density and tissue damage and the assertion that the charge density is an indirect measure of the electrochemical contribution to tissue damage. If the interface voltage is kept within certain limits then chemical reactions can be avoided, and all charge transfer will occur by the charging and discharging of the double-layer capacitance. However, in many instances, the electrode capacitance is not sufficient to store the charge necessary for the desired excitation without the electrode voltage reaching levels where reactions will occur. The principal approach to control the interface voltage has been the use of charge-balanced biphasic stimuli that have two phases that contain equal and opposite charge. However, even with charge-balanced pulses it is possible that the interface voltage may reach levels where electrochemical reactions can occur. If charge is lost through a reaction diode turned on in the first phase of the pulse, then the charge delivered in the second phase of the pulse will exceed the charge still on the capacitor and cause the electrode voltage to overshoot zero. This is the basis for imbalanced biphasic stimulation, which has been shown to reduce electrode corrosion and enables greater charge densities without damaging muscle tissue. The correlation of charge per phase and tissue damage is thought to result from physiological changes associated with neural excitation. The number of excited neurons is dependent on the charge delivered in the stimulus pulse. Thus, physiological changes occurring because of synchronous activation of a population of neurons increase as the number of excited neurons increases. This relationship may create the correlation between charge per phase and neural damage. There is substantial support for this hypothesis in both the peripheral nervous system and the central nervous system.

Equivalent Model for Electrode Tissue Interface:

The physical behavior and the generation of an activation pulse of a cell membrane is described quite accurate using the advanced membrane model, incorporating the conductance, a membrane capacitance and a Nernst potential voltage source. This model was needed to understand the basic working principle of neural cells. To design an electric circuit which is interacting with neural cells this model however is not suitable, since it does not describe the electric properties of the cells. In this section a model is obtained for the impedance of the tissue seen from the terminals of the electrodes. As will be seen shortly, the interface between the electrodes and the tissue is the most challenging to the model. More than a century ago the first research has been conducted towards establishing a model for the electrode-tissue system. It was discovered that the response of the interface was exponential and frequency dependent in nature and that it was therefore reasonable to model the system with a series RC circuit as depicted [334]. The electrode interface model is represented in below figure

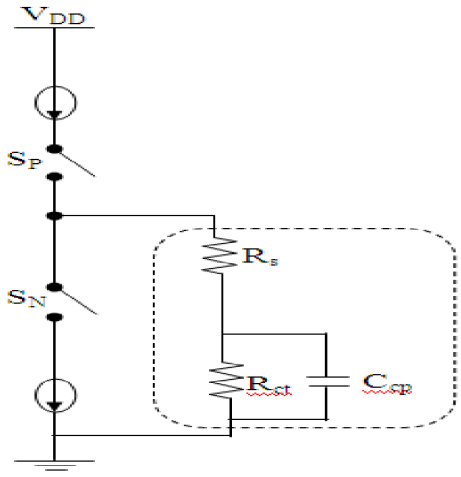

Figure.1: Elctrode interface model

In the equivalent circuit, R_s represents resistance of the connections, R_{ct} is the charge transfer resistor, C_{cp} is the capacitors which represents charge polarization. The conducting of the charged particles occurs typically through oxidation – reduction reactions at the electrode[351*]. Both components can be coupled to a physical meaning.

a. Double Layer Capacitor C_{dl} : The current (charge) in the electrode is a result of the movement of electrons. In the electrolyte, the current originates due to the movement of ions. This means that at the electrode-electrolyte interface there is an interaction between two types of charged particles: electrons and ions. The interface which separates them is therefore capacitive in nature (C).

b. The Series Resistance R_s: The resistive component R corresponds to the resistance of the electrode leads and the tissue. In practice the resistance of the electrolyte will be much higher and is therefore the dominant component. However, it was also discovered in that the electrode-electrolyte interface is much more complicated than just a simple linear first order RC circuit.

Safe Electrical Stimulation Techniques: Electrical Stimulation is to pass a charge through the cell membrane such that an action potential is generated. Electrical stimulation can be achieved by various methods such as constant voltage stimulation, constant current stimulation, switched capacitor stimulation and high frequency current switching stimulation. By injecting a large potential over a long period of time duration, charge is transferred under the electrode and strong currents flow which cause electrode dissolution, gazing as well as tissue damage, all of which should be avoided over all circumstances. If there is a generation of voltage across the electrode such that irreversible chemical reactions occur, then the electrical stimulation becomes ineffective. Additionally, the amount of charge delivered per phase and charge density delivered to the electrode are factors that need to be considered to ensure safe electrical stimulation.

Biphasic Current Stimulation and Residual Voltage: A very straightforward way to avoid this charge accumulation is the adoption of Biphasic Stimulation Pulses with Current Sources.

One of the best methods of stimulation is to use a negative current pulse to initiate the action potential in the cell followed by an interface pulse period, T_i to allow the action potential to propagate, and finally a positive current pulse to neutralize the charge sent in the first phase. The stimulation signal is sent at a frequency no greater than the refractory period of the nerve cell. The refractory period in an excitable cell is the minimum time which the cell shall not generate at action potential[335]. Typical balanced biphasic current stimulation topology shown below figure 4.2.

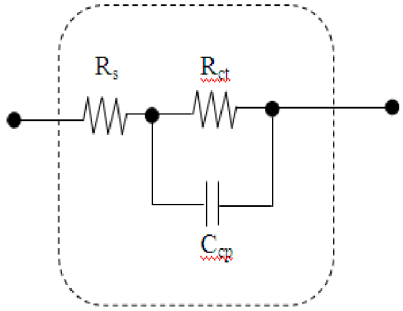

Figure .2: Typical balanced biphasic current stimulation topology

Ideally, biphasic stimulation pulse don't show net charge transfer, but due to mismatch this alleviates and the voltage over the simple electrode model (C_E, R_E) ends up with potential ΔV_E measurements show that DC currents must be kept below I_{dc} <10nA in order to prevent charge accumulation on the electrodes and resulting strong faradic currents. On the other hand, stimulation currents are known as large as 1mA thus, the biphasic current pulse matching should be in the range of

$$\frac{\Delta I}{I} \leq \frac{10nA}{1mA} = 0.001\%$$

This cannot be achieved in a circuit. Therefore, different charge balancing techniques were proposed. They can be broadly classified as passive and active charge balancing approaches.

5. RESULTS:

(1) Specifications Of Equivalent Electrode Model:
C=1µF , R_{ct}=17KΩ , R_s=2.2KΩ

S.No	Stimulated voltage applied (mV)	Output voltage from digital CRO (mV)
1	0	0
2	20	19.4
3	30	29.6
4	40	39.8
5	60	58.5
6	80	78.2
7	90	87.5
8	100	97.5
9	120	116.5
10	140	136.2

Table.2: Stimulated voltage and output output voltage response of skin tissue

(2) Specifications Of Equivalent Electrode Model:
C=100µF , R_{ct}=18KΩ, R_s=3.9KΩ

S.No	Stimulated Voltage Applied (mV)	Output Voltage from Digital CRO (mV)
1	0	0
2	20	19.8
3	30	29.8
4	40	39.7
5	60	59.5
6	80	79.4
7	90	89.2
8	100	99.3
9	120	119.2
10	140	138.9

Table.3: Stimulated voltage and output voltage response of skin tissue

(3) Specifications Of Equivalent Electrode Model:
C=0.1µF , R_{ct}=15KΩ, R_s=2.2KΩ

S.No	Stimulated Voltage applied (mV)	Output Voltage from Digital CRO (mV)
1	0	0
2	20	21.6
3	30	31.5
4	40	40.6
5	60	60.3
6	80	80.3
7	90	90.2
8	100	99.9
9	120	120.2
10	140	140.8

Table 4: Stimulated voltage and output output voltage response of skin tissue

(4) Specifications Of Equivalent Electrode Model:
C=1µF , R_{ct}=17.2KΩ,R_s=3.9KΩ

S.No	Stimulated Voltage Applied (mV)	Output Voltage from Digital CRO (mV)
1	0	0
2	20	20
3	30	29.9
4	40	39.9
5	60	59.9
6	80	79.7
7	90	89.8
8	100	99.8
9	120	119.9
10	140	139.6

Table 5 : Stimulated voltage and output output voltage response of skin tissue

(5) Specifications Of Equivalent Electrode Model:
C=1µF , R_{ct}=18.9KΩ, R_s=2.2KΩ

S.no	Stimulated Voltage Applied (mV)	Output Voltage from Digital CRO (mV)
1	0	0
2	20	20
3	30	29.9
4	40	39.8
5	60	59.7
6	80	79.5
7	90	89.4
8	100	99.3
9	120	119.1
10	140	139.0

Table 6 : Stimulated voltage and output output voltage response of skin tissue

Comparison of Electrode Interface Model Results with Experimental Data:

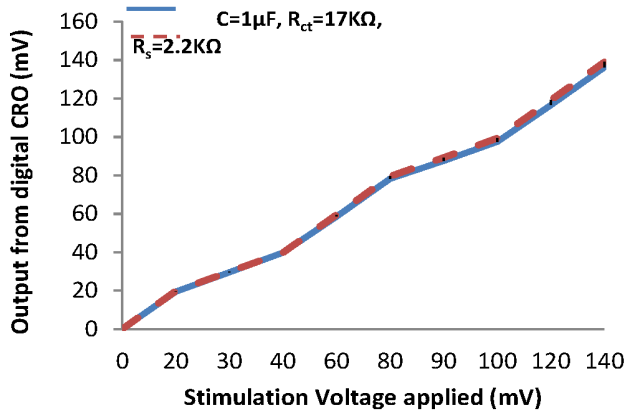

Figure .3: Stimulation effects on electrode model and its output voltage response of skin tissue

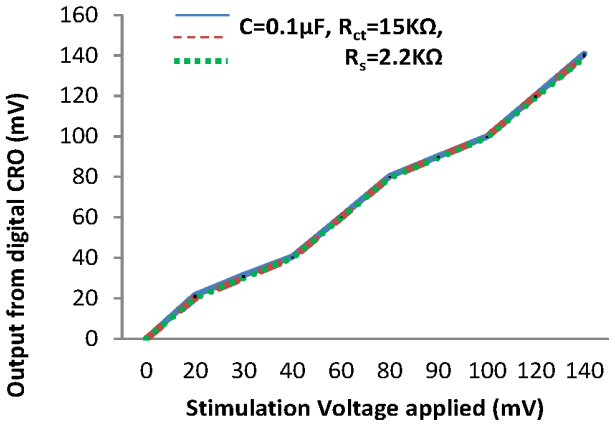

Figure .4: Stimulation effects on electrode model and its output voltage response of skin tissue

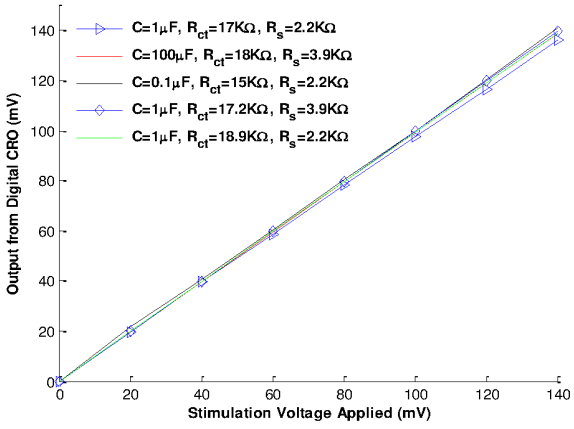

Figure .5 : Stimulation effects on electrode model and its output voltage response of skin tissue

6. CONCLUTION

It is evident from the First chapter the survey as literature, that there exists a lot of research on biological systems, tissue properties. However, literature survey gives an indication of several unaddressed problems. In view of this, several pairs of tissues are considered in chapter two and relative power absorption of tissues are considered for different polarizations. The results indicate that the power absorption ratio is found to vary from pair to pair. There are found to depend on frequency. The results on penetration characteristics indicate that the depth of penetration depends on frequency, permittivity, permeability and conductivity. The computed results obtained over a large frequency range indicate that frequency plays a major role. Moreover, the depth of penetration is found to vary from a few centimeters to tens of centimeters depending on the tissue.

Human body is represented by electrical equivalent in chapter four considering the realistic values of circuit elements. Response of skin tissue is presented. The results indicate linear variation between digital output and applied voltage.
From the results of chapter five it has been possible to conclude that there is a huge change in impedance characteristics with frequency for all tissues. The results on impedance characteristics are extremely useful for design of Electromagnetic energy sources.

An illustration overview on safe electrical stimulation of neural tissues and electrode charge, charge balancing approaches for functional neural stimulation and designed current stimulation electrode neural tissue interface model, compute the charge transfer resistance(R_{ct}). If R_{ct} is significant then the net residual voltage at the end of the anodic pulse will be non zero regardless of how well the both pulses of current matched. Therefore, the electrode residual voltage is briefly observed after every stimulation cycle and verified. If it remains within a predefined voltage range, if not an offset current is adapted in order to track the biphasic current mismatch in next stimulations.

The value of the modeled impedance changes highly due to the electrode/electrolyte interface and demonstrates generation of a net residual voltage can cause damage to the electrode. Hence,it is required to eliminate the residual voltage to zero. An active feedback mechanism can remove the residual voltage to zero.

Finally,it is concluded that the important aim of this proof of concept chapter is to explain that the residual voltage can be controlled by adjusting the width of the anodic pulse and various techniques provides system with valuable safety information about charge balanced and safe electrode condition.

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