SAM-TDR-64-5

THORACIC CAGE IMPEDANCE MEASUREMENTS

Tissue Resistivity in Vivo and Transthoracic Impedance at 100 kc.

TECHNICAL DOCUMENTARY REPORT NO. SAM-TDR-64-5

March 1964

USAF School of Aerospace Medicine
Aerospace Medical Division (AFSC)
Brooks Air Force Base, Texas

Task No. 775801

(Prepared under contract No. AF 41(657)-403 by the University of Minnesota, Minneapolis, Minnesota)

FOREWORD

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ABSTRACT

Average volume resistivities for thoracic tissues at an exciting frequency of 100 kc. are required for studies of the current flux paths in transthoracic impedance plethysmography. By using a uniquely designed probe and anesthetized dogs, an average tissue resistivity was found for skeletal muscle to be approximately 400 ohm cm.; heart muscle, 450 ohm cm.; lung tissue, 1,200 to 1,500 ohm cm.; kidney and liver, 600 ohm cm.; fat, 1,000 to 2,000 ohm cm.; and for whole canine blood, 150 ohm cm. at a hematocrit of 45. With the exception of whole blood, all values varied significantly between dogs and between individual readings. The values for lung and liver tissues and for fat were noted to have particularly large variations between readings and between animals.

Canine skeletal muscle tissue had a linear electrical resistivity characteristic for excitation current densities up to 80 ma./cm.^{2} Representative normal, human skin on the biceps was shown to contribute approximately $4.5 \times 10^{-3} \times \epsilon^{150^\circ}$ mhos/cm.^{2} of electrode surface area in series with the transthoracic impedance, with possible modifications required for electrode geometry and electrode paste.

This technical documentary report has been reviewed and is approved.

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1. INTRODUCTION

It is believed that significant applications of electrical impedance plethysmography will be based on some knowledge of the current flux paths through the irregular geometry of the human thorax. As an initial step toward the determination of these flux paths, the electrical properties of the relevant tissues must be established.

This report summarizes an investigation of the electrical resistance properties of living thoracic tissues at an excitation frequency of 100 kc. It also describes experiments which established the linear characteristics of tissue resistivity, the frequency characteristics of the transthoracic impedance, and the relevant impedance contributed by the skin. The investigations were designed with the specific purpose of providing information for the study of current flux paths in the human thorax for various surface electrode configurations and locations. The 100 kc. excitation frequency was selected on the basis of initial considerations and is justified later in the report. The data were taken on human subjects and, for the tissue measurements, on anesthetized dogs. The definition of volume resistivity is not as intuitive when applied to biologic tissue as when applied to a metal or a semiconductor material. Consequently, it is important to clarify the concept as it is considered in this report. Determining the resistance of a composite volume of various tissue subvolumes implies primarily the use of average tissue resistivities. It is the intent of this report, therefore, to consider the various thoracic tissues as homogeneous, isotropic, electrically linear volumes, so that the concept of an average resistivity for any one tissue volume has validity. An attempt to determine current flux paths in a complex geometry such as the human thorax by considering subtissue structures (e.g., organ nonhomogeneities) does not appear rational at this time. Some investigators have discussed the anisotropic characteristics of muscle tissue; however, the relatively nonuniform orientation of muscle structures in the thorax tends to minimize, if not cancel, this effect. A justification of the linearity assumption is made later in this report.

Received for publication on 23 October 1963.

An experimental procedure was designed around a test probe which provided relatively rapid measurements of tissue resistivity for current flux paths which were approximately spherical in geometry. A number of measurements were made for a variety of test-probe insertions in each tissue, the resulting average value then interpreted as an average tissue-volume resistivity. This procedure was most subject to question when applied to lung tissue.

The investigation of tissue impedance concentrated on the measurement of tissue resistivity, with minimum attention to the reactive effects (tissue dielectric constant) for an examination of current flux paths in the thorax; this appeared justified as a first approximation. In general, transthoracic impedance measurements at 100 kc. showed a phase angle of 10° to 15°. Maximum experimental values found in tissue have not exceeded 35°; this extreme value, however, was suspected of indicating polarization effects. Other investigators reported phase angles of less than 5° for frequencies below 1 kc. (1).

Schwan (2) summarized work that had been completed in the field of passive electrical properties of tissue up to 1957. Concerning the resistivity of thoracic and abdominal tissues, he concluded that, because of large standard deviations for measurements in a given tissue, there was no significant difference between the resistivity of cardiac and skeletal muscle, liver, and lung. Although no specific results were published at 100 kc., Schwan stated that the resistivity values for these tissues at this frequency were approximately one-half the resistivity values that were obtained at 1 kc., or 375 to 500 ohm cm. It is important to note that these values were based on the use of a test probe which did not necessarily measure average tissue resistivity. Specifically, the probe he used consisted of two electrode surfaces mounted on a long, small-caliber catheter. When inserted into the tissue or into arteries or veins of an organ, the probe was particularly susceptible to conductive shunts of the tissue adjacent to the probe, fluid accumulations along the probe, and anisotropic effects (3).

In a recent study, a tetrapolar probe arrangement was used to determine the resistivity of thoracic tissues (3). While the reported results tend to confirm data of the present studies, the work was based on the use of a voltage pulse (similar to the QRS potential of the ECG voltage waveform) rather than on the use of a single excitation frequency. As the frequency dependence of tissue resistivity is well known (2), these results are of questionable interpretation with respect to the stated objective of this investigation.

The probe designed for the tissue resistivity investigations is described in section 3, along with the probe calibration procedure. The results of resistivity measurements on skeletal muscle are described in section 4 as are other thoracic organ tissues and the effects of organ surface and connective tissue on larger volume-resistance measurements. The linearity of muscle tissue at 100 kc., investigated as a separate project (4), is summarized in section 5. The effects of skin resistance on transthoracic impedance measurements are considered in section 6. The frequency characteristics of the transthoracic impedance, specifically with respect to cardiac activity studies, are discussed in section 7.

2. SUMMARY

An indication of the average volume resistivities of thoracic tissues at 100 kc. is needed for the development of the current flux paths in studies of transthoracic impedance plethysmography. As no directly applicable results have been found in the literature, an extensive investigation was undertaken and is described in this report.

A unique probe was designed to investigate the volume resistivity of tissue in vivo. The probe minimized the influence of low-resistance body fluids on the tissue measurements, facilitated relatively numerous and rapid measurements, and indicated an average (isotropic) value of resistivity. Its design also permitted resistance measurements between organs. The use of a measurement box for volume resistance provided a satisfactory check on the probe measurements of muscle tissue.

The resistivity of skeletal muscle was established by probe measurements in the canine adductor muscle. An average resistivity at 100 kc. was indicated to be approximately 400 ohm cm., based on 88 measurements from 5 dogs. The maximum value of the five standard deviations computed from the data taken for each of the animals, however, was 66 ohm cm. Individual, extreme measurements were 250 and 500 ohm cm. No significant change in skeletal muscle resistivity was detected because of an interrupted blood flow or the injection of a vasodilator drug. The 100 kc. resistivity of lung tissue was found to be approximately 1,500 ohm cm. and 1,200 ohm cm. for an approximately normally inflated and totally deflated (open chest) condition. Average values for individual animals were subject to variations of approximately 70% of these values, while individual measurements were distributed over a wider range. An average change of resistivity of 25% was noted between the inflated and totally deflated conditions. The average values of other tissue

resistivities were established to be approximately 450 ohm cm. for heart muscle, with individual values occurring between 350 and 550 ohm cm.; from 1,000 to 3,000 ohm cm. for fat with considerable variation noted for individual measurements; and 600 ohm cm. for both kidney and liver, with variations from 300 to 900 ohm cm. existing for liver tissue. Average resistivity values of 150 and 120 ohm cm. were found for whole canine blood with a hematocrit of 45 and 35, respectively.

An independent study, reviewed in section 5, indicated that experimentation carried out in skeletal muscle tissue in vivo with a maximum average electrode current density less than 80 ma./cm.^{2} at 100 kc. is in the linear range of the tissue resistivity.

By using specially designed surface electrodes applied to the biceps of a subject, a skin admittance of $4.5 \times 10^{-3} \times \epsilon^{160°}$ mhos/cm.^{2} of electrode-skin area, $\pm$ 10%, was found to be in series with the impedance attributed to the internal tissue of the body. An added resistance may have to be included, depending on the geometry of the surface electrode and electrode paste used.

Concerning the use of 100 kc. for the excitation frequency of transthoracic impedance studies, data indicated that this frequency provided approximately the maximum change of impedance over a cardiac cycle for a ventricular-volume-shaped impedance waveform and, at least, for representative electrode types and positions used in other phases of this study.

3. TISSUE PROBE

As the volume resistivities of blood and other body fluids are lower than the resistivities of most tissues, it is necessary that the procedures used for measuring tissue resistivity in vivo minimize the undesirable effects of local bleeding. The probe designed for this investigation is shown in figure 1. The electrodes consisted of two cylindric surfaces with parallel axes spaced 3/8 in. apart. The probe was constructed by removing the syringe

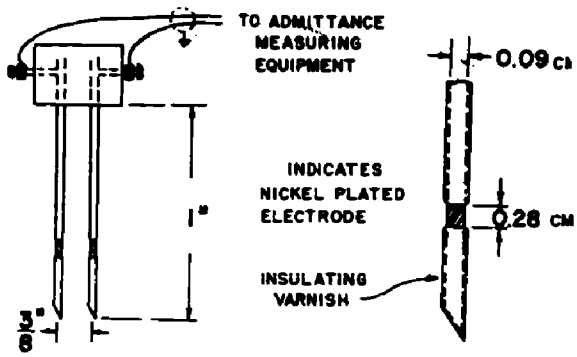


FIGURE 1

Tissue probe for resistivity determinations.

fitting from two 20-gage stainless steel hypodermic needles. Electrical contact was made through brass screws, which also anchored the needles to the acrylic block. The exposed portions of the needles were insulated with 4 or more coats of Epoxylite electrode insulator, approximately 80 μ thick, and the tips were sealed. After coating, the insulation was removed from a portion of each needle to make the electrode surfaces, which were then nickel-plated. The electrode-insulation junction was beveled to prevent tearing of the tissues during insertion.

The tissue probe was used in conjunction with a General Radio impedance comparator, model 1605A, a precision bridge with internal signal source and null indicator. When inserted into a tissue and connected to the comparator, the probe became the unknown impedance in one arm of a standard electrical bridge circuit.

Specifically, the probe was designed to allow many tissue measurements to be made in rapid sequence, to minimize the disturbance of the tissue and the electrical shunting effects of blood and other body fluids between the electrode surfaces, and to provide an average (isotropic) value of resistivity. Two electrode (needle) spacings were used, a 3/4-in. spacing for measurements in large tissue volumes and for interorgan and intermuscular measurements, and a 3/8-in. spacing to minimize the error in small volume tissues due to current flux boundary conditions.

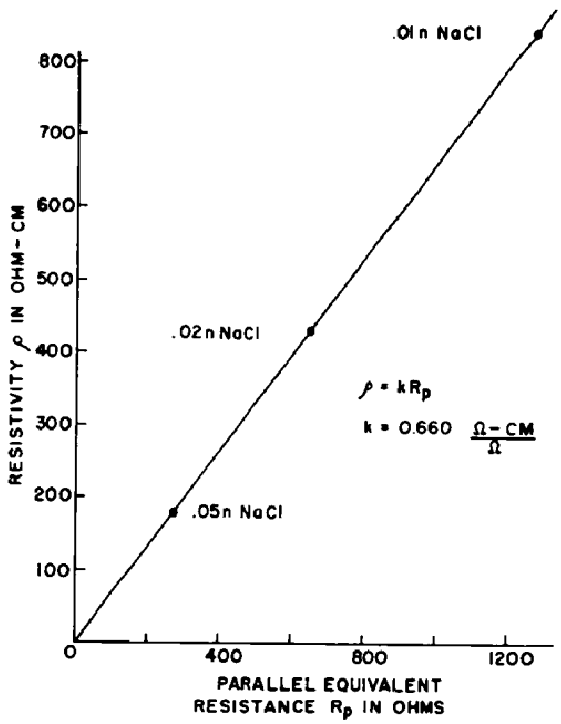


FIGURE 2

Tissue probe calibration in NaCl solutions.

The probe constant was determined by measuring the probe resistance in 3 NaCl solutions of known resistivities (fig. 2). The slope of the curve established the probe constant. Some of these calibration curves did not intersect the origin, suggesting probe polarization. The value of equivalent series resistance varied between 0 and 50 ohms. Schwan and Kay (1) found that the polarization effect in tissue was approximately twice as large as in saline. Since a typical value for the probe resistance in tissue was 500 ohms and the maximum value for polarization resistance in saline was 50 ohms, the polarization error in tissue resistivity was estimated to be less than 20%. The error in the probe constant was less than 5%, making the total experimental error less than 25%.

Edge effects due to the location of tissue boundaries near the electrode were estimated by varying the spacing between the probe to the saline-glass boundary in the calibrating cell. It was found that, if the smallest distance from an electrode to a saline boundary was greater than the distance between the electrodes, the increase in the measured value of the probe resistance, R_p, was less than ½% over the value of R_p measured in an essentially infinite medium of the same resistivity. In all resistivity measurements, an attempt was made to insure that the minimum electrode-tissue boundary spacing was greater than the electrode separation. The calibration and use of the probe assumed that the calibrating solution and the tissue samples were large enough that the nature of the current field boundary did not significantly affect the measured values of R_p.

It was also recognized that if the tissue probe was inserted into an infinite volume, the measured probe resistance would be determined significantly by the material adjacent to the probe surface. From field-mapping considerations in a homogeneous medium, however, approximately 50% of the total voltage drop was found to occur over about 30% of the ⅜-in. spacing between the electrode surfaces. It was felt that this represented a near optimum design relative to minimum tissue disturbance caused by the probe insertion and minimum boundary and parallel field effects.

All measurements of tissue resistivity made with the probe were performed on mongrel adult dogs, male or female, 20 to 50 lb., placed on an electrically insulated table. The dogs were kept in a state of surgical anesthesia throughout the experiment by repeated injections.

The choice of a series or parallel equivalent circuit to characterize the externally measurable properties of a volume of tissue, as measured with a probe for example, is arbitrary and independent of the geometry of the volume, the electrode, or the properties of the tissue. By convention, however, a parallel equivalent circuit was used to define the intrinsic electrical properties of the tissue (2). Therefore, for sinusoidal steady-state excitation of a volume of tissue, the following definitions were made:

R_p, the measured parallel resistance in ohms,

$G = (R_p)^{-1}$, the conductance in $(\text{ohms})^{-1}$,

C, the measured parallel equivalent capacitance in farads, and

$Y = G - i\omega C$, the parallel admittance in $(\text{ohms})^{-1}$,

where

$i = \sqrt{-1}$ and $\omega = 2\pi f$, the excitation frequency in $(\text{seconds})^{-1}$.

By assuming homogeneous, isotropic, linear, passive muscle tissue, and for ar arbitrary but fixed electrode geometry, the relation between the measured value of R_p and the intrinsic resistivity, ρ (ohm cm.), was expressed as:

$$[R_p = a_1\rho]$$

with a_1 constant with respect to the voltage, but dependent on the electrode and volume geometry, and determined experimentally.

4. RESISTIVITY OF TISSUE IN VIVO

Skeletal muscle tissue

An incision was made through the skin and subcutaneous tissue of an anesthetized dog to expose the medial aspect of the adductor muscles and gracilis muscles. The adductor group of muscles was selected for the measurements of the skeletal muscle resistivity because of the relatively large volume to surface area—the ratio necessary to reduce probe edge effects. After measuring the resistance between the electrodes in a 0.02 *N* NaCl solution, the electrodes were inserted through the gracilis muscle into the adductor muscles. Up to 14-probe resistance values could be recorded for separate probe insertions into each adductor group for each dog without the insertions occurring too close to one another, or too close to the muscle surface. The probe was calibrated between each measurement; if the calibrating value of R_p differed by more than 5% before and after a tissue resistivity measurement, the intervening data were discarded.

A significant problem was encountered when recording R_p for skeletal muscle; the value of R_p decreased with time for any given probe location. A typical curve of this

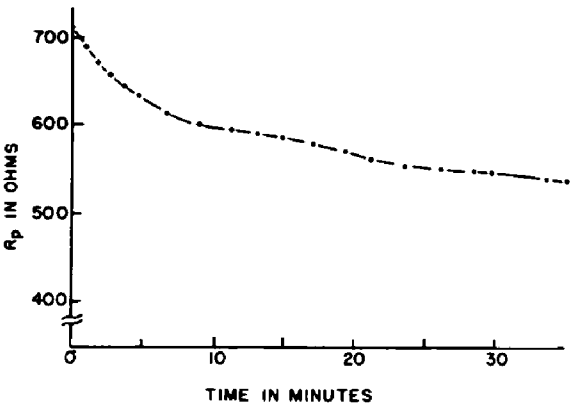


FIGURE 3

Change in R_p following probe insertion.

phenomenon is shown in figure 3. It was hypothesized that the change in R_p was caused by an accumulation of low-resistance blood and interstitial fluid around the electrode surface. This possibility was investigated by using a modified probe design similar to that shown in figure 1, but made with 15-gage needles which were insulated on the inside surface and not plugged. After this hollow-needle probe was inserted into the tissue, physiologic (0. 16 *N*) saline was injected in doses of 0.01 cc. through the hollow needle into the region immediately surrounding the electrode surface. The resulting R_p decreased appreciably after each injection, as indicated in figure 4. While this procedure did not provide conclusive evidence that fluid accumulation was the cause of the change of probe resistance with time, it strongly supported the hypothesis. In general, all reported experimental values of tissue resistance were made within 1 minute after the probe was inserted into the tissue to minimize the possible effects of fluid accumulation on the data.

Experimental results for skeletal muscle tissue are given in table I. It is noted that a significant variation of the resistivity occurs both between individual measurements and between animals. The total experimental average was 395 ohm cm., with maximum and minimum data values at approximately 250 to 500 ohm cm. The extreme values for the standard

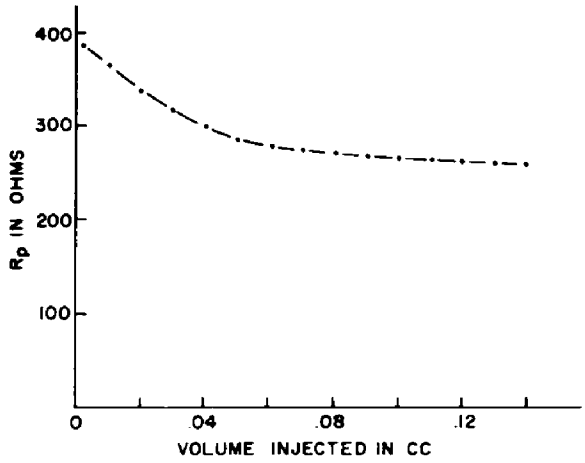


FIGURE 4

Decrease in R_p with injected saline.

TABLE I

Skeletal muscle resistivity

Experiment	Average ρ	S.D.	Number of data
1	425	35.2	9
2	401	64.6	26
3	348	66.6	20
4	401	18.0	15
5	413	26.7	18
Total	395 ohm cm.		88

deviation of data taken on any one animal were 18 and 66.7 ohm cm.

To provide a check on the resistivity established with the tissue probe as being representative of an average volume resistivity, the resistivity of muscle was also found with the aid of a Lucite muscle box (fig. 5). This box was calibrated by a procedure similar to that used for the probes, the accuracy of the calibration estimated at 10%, including the effects of open-ended construction. An excised canine, semitendinous muscle was placed in the box to fill a determined volume. The resistivity of the muscle for the animal used for experiment 5 (table I) was found to be 435 ohm cm. and 447 ohm cm. on two separate trials. It was concluded that these data adequately confirmed the validity of the probe measurements for

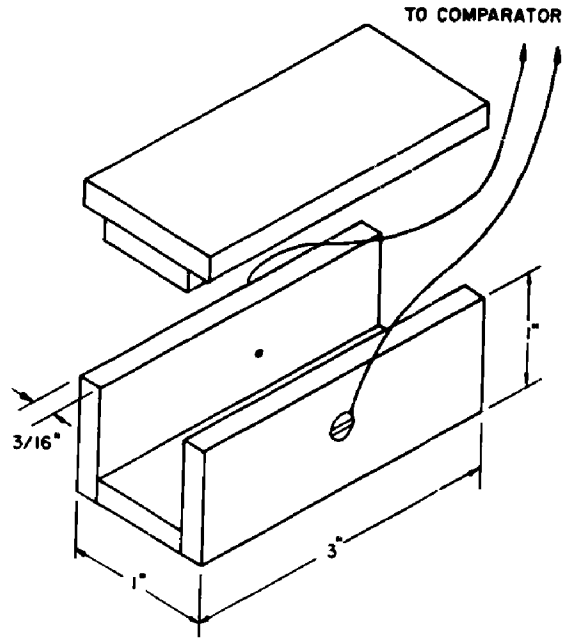


FIGURE 5

Calibrated muscle box for resistivity measurements.

obtaining a gross average resistivity of skeletal muscle tissue.

A series of experiments was performed on an exposed canine gracilis muscle to determine the influence of blood flow on the measured (average) value of resistivity. First, the femoral artery and the femoral vein were separately occluded, while R_p was continually recorded. Any resulting change in R_p due to either occlusion was not observable (less than ½%). Under conditions of normal circulation, but with the muscle exposed and the probe inserted into the muscle, up to 40 μg. of norepinephrine was injected into the jugular vein. The value of probe resistance recorded continuously before and after the injection was not observed to change as a result of this injection. As these changes of less than ½% of R_p were not significant, relative to the general experimental technic, it was concluded that the average resistivity of the skeletal muscle is independent of normal, or even abnormal, variations of blood flow.

TABLE II

Heart muscle resistivity

Experiment	Average ρ	Number of data
1	455	11
2	432	7
3	432	6
4	474	2
5	500	7
Total	456 ohm cm.	33

Nonskeletal muscle tissue

For the determination of tissue resistivities of thoracic organs, dogs were artificially respired and the internal organs were exposed.

Heart muscle tissue measurements were taken in the left ventricle wall within 2 minutes after euthanasia was performed on the animal with sodium pentobarbital, and the heart stopped beating. This procedure was assumed to provide satisfactory data, as independent measurements on the adductor muscle indicated negligible change in resistivity immediately after death (also see Schwan and Kay (1)). The measurements of heart tissue were considered less accurate than the measurements of skeletal muscle due to possible boundary effects of the smaller relevant volumes.

Experimental results for heart muscle are summarized in table II. The 33 separate data points for the 5 experiments refer to separate probe insertions. While the value of these data averaged 456 ohm cm., the minimum and maximum values were 276 and 674 ohm cm. For any one animal, these data indicated an average resistivity of 450 ohm cm., with most individual readings between 350 and 550 ohm cm.

The measurement of lung resistivity placed greatest demands on the averaging technic of the tissue probe because of the lung volume nonhomogeneity. During the experiments, the lungs were artificially ventilated so that the lung air volume varied between approximately full normal inflation and total deflation (complete collapse). Some values of lung resistivity were also measured when the lungs were held at approximately one-half the inflated volume, in an attempt to duplicate the condition of normal expiration in an intact animal.

The lung tissue data are summarized in table III. The average resistivity value for the inflated lung condition was 1,530 ohm cm., with individual animal averages ranging from approximately 860 to 2,500 ohm cm. Maximum and minimum values were 840 and 3,430 ohm cm. Corresponding resistivity values for the collapsed lungs were 1,220 ohm cm. for the total experimental average, approximately 800 to 2,100 ohm cm. for individual animal averages, and 675 to 2,800 ohm cm. for the limits of individual measurements. On the basis of these averages and very limited additional data taken near normal expiration volume, the value of resistivity for lung tissue in a normally expired intact animal would be expected to be between 1,300 and 1,400 ohm cm., but this value could vary significantly (0.5:1 or 1:2) for an individual animal. The

TABLE III

Lung tissue resistivity

Experiment	ρ, average			Number of data
	Full inspiration	Mid-inspiration	Complete expiration	
1	1,125		810	5
2	2,610	2,100		1
3	2,410		2,090	5
4	857		786	3
5	1,310		1,085	6
6	1,450	1,345	1,125	10
Total	1,530 ohm cm.		1,220 ohm cm.	30

TABLE IV

Blood resistivity, average of 2 and 3 measurements per experiment

Experiment	Average ρ	Hematocrit
1	158	40
	155	41
2	108	29
	118	33
3	129	40
4	120	36
5	153	47

average change of resistivity between approximately full normal inspiration volume and total deflation was found to be about 300 ohm cm. or 25% of the deflated value.

A very high value of R_p was occasionally found during the measurements of lung tissue resistivity. This may have been caused by electrode surfaces positioned in an air space. These data were discarded. It is possible that a varying amount of electrode surface contact with lung tissue was a partial cause of the large variation in these resistivity measurements.

The resistivity of whole canine blood was measured in a test cell immediately after removal and the addition of heparin. Averages of 2 or 3 readings for individual blood samples are given in table IV. The individual readings did not vary more than 4 ohms from the given average values. As it is recognized that the resistivity of blood is a function of the hematocrit (5), no additional average is given. A value of 150 to 155 ohm cm. for whole blood with a hematocrit between 45 and 50 and a value of 120 ohm cm. for whole blood with a hematocrit of about 35 are indicated from a plot of resistivity versus hematocrit. Again, the variation of average values between animals was found to be 10% or more. These resistivity values were consistent with other (unreported) data taken with the tissue probe inserted directly into the right ventricles.

While incidental to the investigation of the resistivities of thoracic tissues, additional tissue resistivity values were determined for fat, kidney, and liver. Both kidney and liver values averaged 600 ohm cm. The individual liver values varied widely between 300 and 900 ohm cm. as might be expected from the large number of bile- and blood-volume spaces. Data for fat were difficult to obtain because of volume considerations—i.e., probe-edge effects. Some values were indicated around 1,000 ohm cm., others considerably higher (2,000 to 3,000 ohm cm.), again varying widely for individual measurements.

By using a probe similar to that described in figure 1 but with a ¾-inch electrode spacing, measurements were made to determine (1) the effects of organ surface tissues and interfaces on interorgan resistance measurements, and (2) the effects of probe axis orientation on the values of R_p. One needle of the probe was inserted into each of two adjacent organs or tissues, bridging the connective tissue, surface tissues, and others. In no case was a significant difference noted on the measured probe resistance, R_p, due to surface tissues or organ interface structures. Furthermore, no significant difference was noted on the measured resistance because of the probe orientation. This last result supported the anisotropic assumption made relative to the average tissue resistivity measured by the probe.

5. TISSUE LINEARITY

An experimental study of the electrical characteristics of tissue in vivo requires a knowledge of the limits of linearity of the tissue. As pointed out by Schwan (2), little has been known about the linearity of tissue. This is particularly true for excitation frequencies around 100 kc. Consequently, the current density limit for muscle tissue linearity at 100 kc. was experimentally determined.

Procedure

As the impedance measured with the probe inserted into tissue decreased with time (fig. 3), probably due to the gradual development of a thin layer of blood and interstitial fluid around the electrodes, impedance measurements taken simultaneously at both low-current and high-current densities had to be

made. The difference of these two records was then used to detect the presence of a nonlinear characteristic. A constant voltage (0.4 v.) General Radio impedance comparator was used for the low-current density control. On the basis of preliminary measurements made with a General Radio 916-AL bridge and an oscillator and detector, it was established that muscle tissue was linear to within 0.5% when examined with the 1 ma. probe current generated by the comparator. This 1 ma. probe current corresponded to an average current density, J_s, of 13 ma./cm.^{2} over the surface of the electrodes. With a linear value of current density established as a control level, the current density was then raised above this value, using the GR bridge and oscillator, while measuring the impedance with oscilloscope and voltmeter combination (scope-meter). This was continued until a significant difference between the impedance measured with the comparator and the scope-meter was noted.

For these experiments, the probe was inserted into the central portion of the adductor group after the muscles had been exposed by a skin incision. Preliminary measurements of the tissue impedance indicated that, as J_s was increased to about 300 ma./cm.^{2}, the impedance decreased predominantly because of a decrease in R_p. Hence, R_p was used as the parameter for the investigation of tissue linearity.

Results

As the average current density at the surface of the electrodes in the adductor muscle, J_s, was increased from 3 to 375 ma./cm.^{2}, a significant (nonlinear) drop in R_p was noted (fig. 6). Since only the change in R_p, due to a change in J_s, was considered the nonlinear effect, the current density for the limit of linearity was taken as the value for which the corresponding R_p exceeded the R_p^* measured with the control current density by the (2%) experimental accuracy. This critical value of J_s can be considered a heating threshold for muscle, as no stimulation or other nonlinear phenomenon is expected at this frequency. For the typical experimental data shown in figure 6, the muscle was found to be linear for

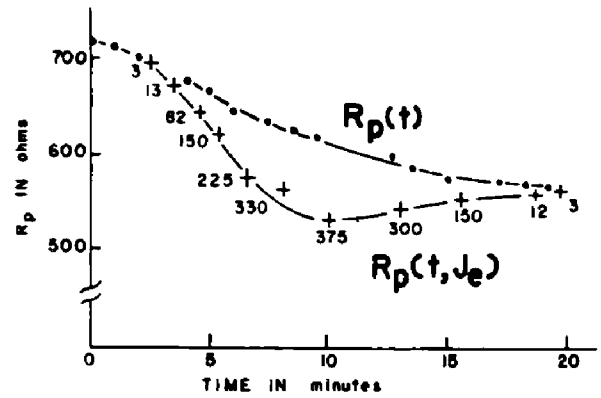


FIGURE 6

Change in R_p with increasing current density.

J_s less than 82 ma./cm.^{2} In four other experiments, with a total of 4 dogs, this threshold varied between 82 and 110 ma./cm.^{2}

It was concluded that tissue resistivity measurements and volume resistance studies carried out at 100 kc. with a maximum current density below 80 ma./cm.^{2} can be considered to be performed within the linear range. This conclusion justified the assumption of tissue linearity made in the introduction to this report.

6. NORMAL HUMAN SKIN IMPEDANCE AT 100 KC.

A series of measurements was made to obtain a representative admittance per unit area of normal skin at 100 kc. This was done to allow an estimate of the effect of skin on electrical impedance measurements made across the thoracic cage.

For purposes of thoracic impedance plethysmography, it is convenient to consider skin as composite layers of tissue which provide a relatively high impedance to a normal flow of alternating current from an electrode in contact with the skin surface. It is assumed that under the skin, tissue of relatively low resistivity exists in sufficient volume so that essentially all the electrode current passes through the skin in the volume directly below

the electrode and then spreads out in the volume under the skin according to the relevant conductivity and geometry.

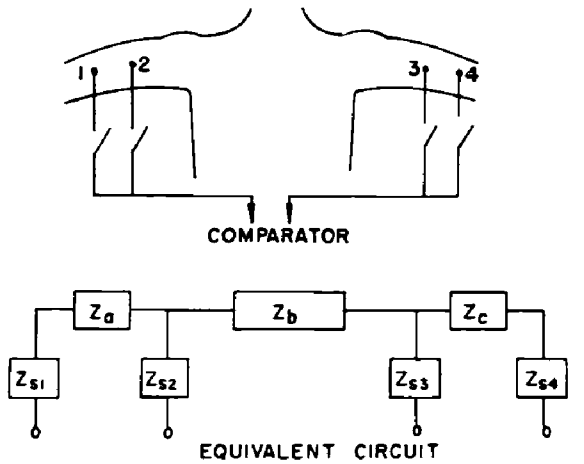
Because of difficulty in controlling internal-current flux paths in the human body, it was necessary to assume that the measured impedance values of the skin on the upper arm are representative values of skin impedance on the thorax. No attempt was made to correlate the skin impedance with skin thickness, sebaceous gland density, apparent thickness of subcutaneous fat, or proximity to bone structure. It was further assumed that the electrodes, the electrode paste, and the electrode-skin spacing all contributed negligible effects to the total measured impedance.

It was anticipated that the data reported in this section would be used to determine a skin impedance which can be subtracted from a total measured transthoracic impedance. For convenience of application, however, the data are presented in units of admittance per unit area of parallel R and C elements equivalent to the assumed lumped skin-series impedance.

Procedure

In general, electrodes were designed to provide a significant change in skin-surface area with minimum effect on the internal impedance. Specifically, the skin impedance was measured with two different electrode geometries. For each geometry, the change of impedance was noted as the electrode area was doubled. Electrodes were positioned on the upper arms with the corresponding current flux path passing through the shoulders and the upper portion of the thorax. This position was selected to avoid cardiac variations on the measured impedance, as well as to minimize the significance of a change in internal flux path as the electrode position and area were changed.

Two electrodes were placed close together on each upper arm, as shown in figure 7. The total electrode impedance was first measured across the thorax using one electrode on each arm. Then, adjacent electrodes on each arm were electrically connected and the impedance



ASSUME: $Z_a = Z_c = 0$, and

$Z_{s1} = Z_{s2} = Z_{s3} = Z_{s4} = Z_s$

THEN: $Z_{1-3} = Z_{1-4} = Z_{2-3} = Z_{2-4} = 2Z_s + Z_b$

$Z_{12-34} = Z_s + Z_b$

and $Z_{1-3} - Z_{12-34} = Z_s$

$$Z_s = \left(\frac{1}{R_s} + j\omega C_s\right)^{-1}$$

$$Y_{area} = \left(\frac{1}{R_s} + j\omega C_s\right)\frac{1}{A}$$

FIGURE 7

Measurement of skin resistance.

again was measured across the thorax. In effect, the area of each arm electrode was doubled, decreasing the skin impedance by a factor of two, while the internal body impedance between the electrodes remained essentially constant. The skin impedance was then calculated from the difference between the two measurements and converted to an admittance per unit electrode area (fig. 7).

The thorax impedance was measured between the four possible pairs of single electrodes to consider the effects of variations in the flux path between different electrode pairs. The resulting extreme values of the impedance were used to establish a tolerance on the computed admittance values.

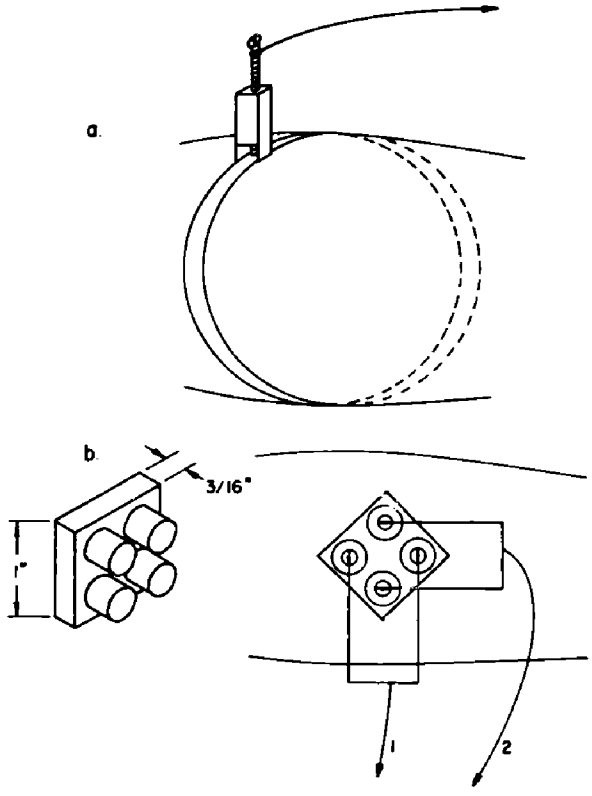


FIGURE 8

Electrodes for skin admittance measurements.

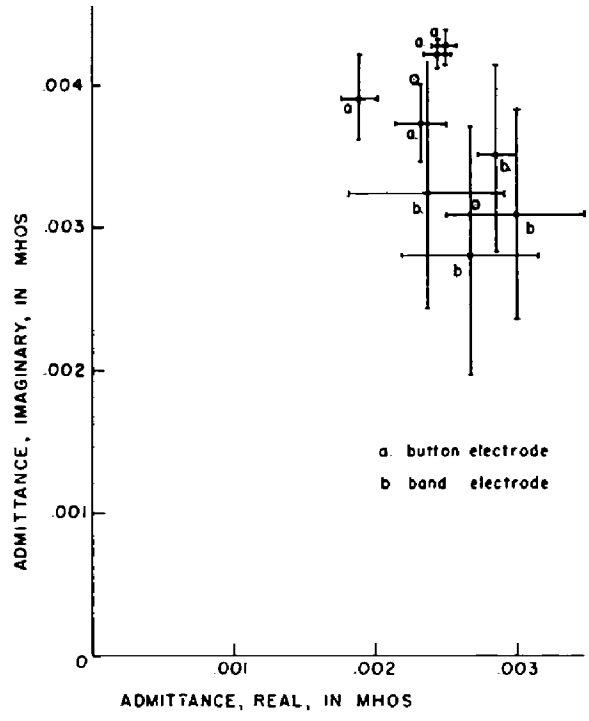


FIGURE 9

Skin admittance data.

One electrode geometry was a band 0.5 cm. wide and 0.0013 cm. thick, wrapped around the arm and fastened with a small screw clamp to insure complete contact with the skin (fig. 8a). The electrode was made from spring brass, platinum-plated on the surface which contacted the skin, and coated with a thin layer of electrode paste (Translyte, Electronic Medical Systems). Two bands were placed parallel to each other about ½ in. apart around each bicep, at locations 1 and 2 and locations 3 and 4 on figure 7.

An alternate electrode geometry was constructed with 4 cylindric brass buttons, ⅜ in. in diameter, mounted in a block of acrylic plastic. Diametrically opposed buttons were connected in parallel so that a single and double electrode area configuration consisted of two and four buttons, respectively (lead 1 or combined lead 1 and 2 in fig. 8b). The end of each button was platinum-plated, coated with electrode paste, and held in position on the arm with surgical tape.

With both electrode geometries, all visible electrode paste was removed after the electrode was positioned. The electrodes were held in place approximately 10 minutes before measurements were made to allow the skin to reach an equilibrium state with the paste.

Results

The results of two experiments on 1 subject are shown in figure 9. These data indicated approximate average values of $4.75 \times 10^{-3} \times \epsilon^{150°}$ mhos/cm.^{2} for the button electrodes and $4.00 \times 10^{-3} \times \epsilon^{150°}$ mhos/cm.^{2} for the band electrodes. One measurement made with an unplated band appears as the extreme right data point in figure 9.

There was a larger tolerance on the skin admittance computed from the band-electrode

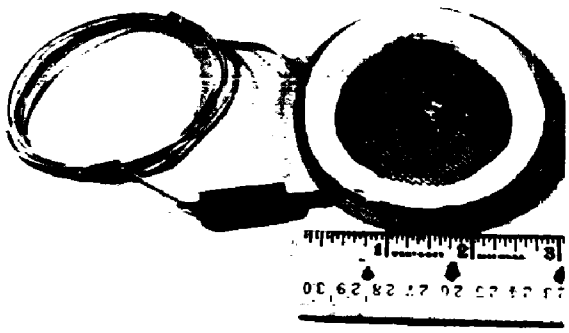


FIGURE 10

Transthoracic impedance electrode.

data than for the button-electrode data. This was partially explained by the fact that the band electrodes had such a comparatively large area that the total skin impedance associated with each electrode was small. Consequently, small errors in the impedance measurements caused large percentage errors in the computed skin admittance. Furthermore, the band electrodes were relatively sensitive to body movements, causing appreciable changes in the impedance for slight body motion.

For the type of surface-skin electrode often used for transthoracic impedance studies and shown in figure 10, it is important to recognize that the electrode paste used in the cup can add significantly to the resistance attributed to the combination of electrode and skin. For example, the values of resistivity for Translyte and Redux (Sanborn) electrode paste were found to be 430 ohm cm. and 12.5 ohm cm., respectively, at 100 kc. and room temperature. By using the Redux paste to completely fill a 1-in. diameter cup, negligible (less than 1 ohm) difference in skin resistance was found when the cup electrode either had the screen completely recessed or had the screen held flush to the skin surface. By using the Translyte paste, however, a 7.5-ohm increase per electrode for the flush screen and a 20-ohm increase per electrode for the recessed screen were noted, compared to the corresponding Redux paste values. The 7.5-ohm increase may be due to a higher skin-paste contact resistance, while the 20-ohm increase includes an added amount for the volume resistance of the higher resistivity paste.

7. FREQUENCY CHARACTERISTICS OF TRANSTHORACIC IMPEDANCE

With proper electrode locations on the thorax and with an external excitation of 100 kc., it is possible to obtain records of the change in transthoracic impedance, as a function of time during an expiratory pause, that are almost identical in waveform to a typical ventricular volume waveform. (As the electrode positions are varied, many other waveforms are obtained, presumably reflecting a combination of changing ventricular, atrial, aortic, and pulmonary blood volume patterns.) By changing the frequency of the applied electrical signal, a variation of the amplitude of the peak-to-peak impedance change was noted. The shape of this frequency curve suggested that a maximum existed in the range from 10 kc. to 1 mc. It was assumed that this maximum occurred at approximately the same frequency for other significant thorax electrode positions; consequently, frequency response data were taken for the electrode position that produced the best ventricular volume waveform at 100 kc.

Procedure

A General Radio 916-AL bridge was connected to a sitting subject by two 3-in. polyethylene cup electrodes, similar to that illustrated in figure 10. The cup electrodes were filled with Sanborn Redux electrode paste and placed on the thorax as follows: The left electrode was centered in the fourth intercostal space on the left mammary line and the right electrode centered in the fourth intercostal space midway between the anterior medial line and the right mammary line. The voltage out of the bridge (fig. 11) represented the degree of imbalance of the bridge and fluctuated synchronously with the subject's cardiac cycle. This voltage was amplified by a narrow-band amplifier (National HRO 60 receiver), filtered, calibrated, and recorded as the magnitude of the impedance change during systole. It was

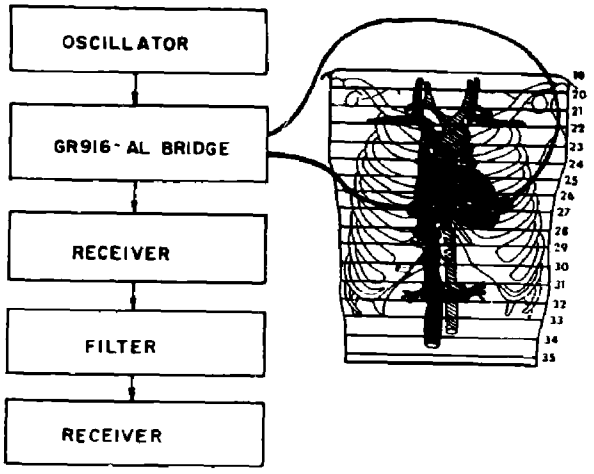


FIGURE 11

Instrumentation for the transthoracic impedance frequency characteristic.

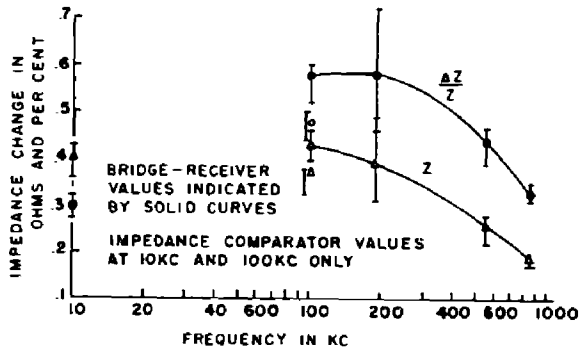


FIGURE 12

Frequency characteristic of the transthoracic impedance change over a cardiac cycle.

possible to examine the frequency range from 100 kc. to 1 mc. with this instrumentation.

For some information about the behavior of the impedance change with cardiac activity below 100 kc., a General Radio impedance comparator was connected directly to a recorder. The comparator excitation voltage was only available at the fixed frequencies of 10 kc. and 100 kc. While it was not possible to determine quantitatively that the curve between these frequencies was monotonically decreasing, this was apparent from related experiments.

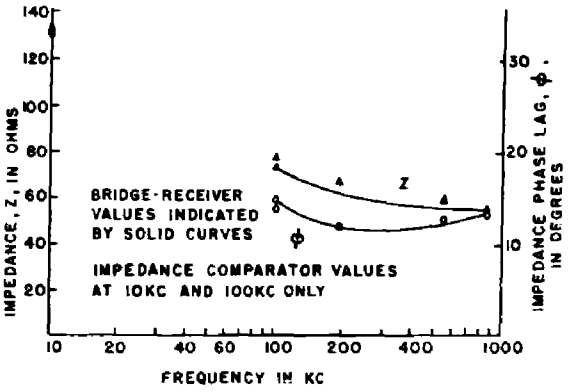


FIGURE 13

Transthoracic impedance and phase angle.

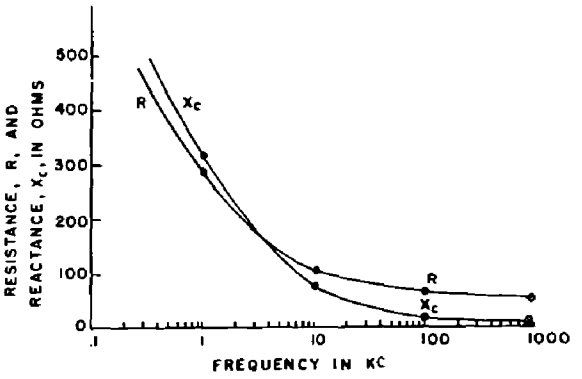


FIGURE 14

Transthoracic series resistance and reactance.

Results

The data are plotted in figure 12, the accuracy of each point indicated by a vertical line. The frequency at which the largest change in transthoracic impedance occurred during the cardiac cycle was at, or slightly below, 100 kc. The percentage change in transthoracic impedance during the cardiac cycle as a function of frequency is also plotted in figure 12. Corresponding average (balance) values of impedance, phase angle, and series resistance and reactance as functions of frequency are shown in figures 13 and 14.

On the basis of these data, the examination of cardiac activity related to stroke volume by transthoracic impedance measurements can be effectively concentrated at an excitation frequency of 100 kc.

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