EVALUATION OF A METHOD FOR DETERMINING THE EFFECTS OF LOCAL MICROWAVE EXPOSURE ON LOCAL TISSUE BLOOD FLOW

BY

BRIAN DAVID HORWITZ

B.S., University of Illinois, 1983

THESIS

Submitted in partial fulfillment of the requirements
for the degree of Master of Science in Electrical Engineering
in the Graduate College of the
University of Illinois at Urbana-Champaign, 1985

Urbana, Illinois

University of Illinois at Urbana-Champaign

DEPARTMENTAL FORMAT APPROVAL

THIS IS TO CERTIFY THAT THE FORMAT AND QUALITY OF PRESENTATION OF THE THESIS SUBMITTED BY BRIAN DAVID HORWITZ **AS ONE OF THE REQUIREMENTS FOR THE DEGREE OF** MASTER OF SCIENCE **IS ACCEPTABLE TO THE** DEPARTMENT OF ELECTRICAL ENGINEERING.

(Full Name of Department, Division or Unit)

22 May 1985

Date of Approval

Lilian W Beck

Departmental Representative

ACKNOWLEDGEMENT

The author would like to acknowledge the Horwitz family; without their constant love and support this work could not have been undertaken.

The author is indebted to his advisors, Professors Richard L. Magin and Kenneth R. Holmes, for suggesting the research topic and providing continuous guidance and support during the research. Their many helpful suggestions and criticisms are gratefully acknowledged.

Appreciation is extended to Mrs. Wanda Elliott and Mr. Bob Cicone for assistance in manuscript preparation, to Mr. Billy McNeill for constructing the necessary mechanical apparatus, and to Mr. Walter Bottji for assistance with surgical preparations.

This work was supported by a contract from the U.S. Environmental Protection Agency.

TABLE OF CONTENTS

CHAPTER 1

INTRODUCTION

The effects of electromagnetic waves on living systems has been an area of intense research for many years. Currently, debate is centered on the establishment of appropriate exposure standards for humans under all exposure conditions. For this reason, ongoing research has been undertaken which is attempting to isolate any physiological changes induced by electromagnetic waves. One such change which workers have attempted to isolate is in local tissue blood perfusion. This thesis describes work directed toward the development of a small implantable probe for the determination of local blood perfusion in tissues exposed to microwave radiation. A small pulsed-heated thermistor probe was used for blood flow measurements in this investigation. This technique is known as the Thermal Pulse Decay Method. These studies were designed to evaluate the feasibility of operating this probe in microwave fields. Initially, the necessary microwave and electronic equipment was assembled for testing the operation of the thermistor probe in the presence of intense microwave fields at 2,450 MHz. Additional system modifications and design anaylsis were performed as required in the course of these studies. The next stage of this study consisted of experimental measurements of the electrical response of the thermistor probe in the presence of microwaves. These studies were performed with the probes positioned with a fixed orientation with respect to the microwave fields and implanted in muscle-tissue equivalent dielectric material. The final stage of

the investigation consisted of studies in which different probe designs and orientation parameters were studied *in vivo*, in anesthetized animals.

The Thermal Pulse Decay (TPD) method is a technique for determination of local tissue thermal conductivity and blood perfusion. A description of the method has been described elsewhere [1,2]. Briefly, electrical power is applied for a short time to a small thermistor bead which has been previously inserted into the tissue. The resistance of the thermistor converts the electrical energy into heat at a point in the tissue. Subsequent to the power pulse, the embedded thermistor serves to measure the decay of tissue temperature. A computer assisted data acquisition system is used to sample (10 samples per sec) the temperature during the measuring time. Since the heated tissue volume continues to increase with time due to thermal diffusion, the sampling volume of a probe can be varied by selecting the appropriate sampling time. For a typical measurement time of 10 sec, the effective measurement volume has a radius in excess of 2 mm, which is eight times larger than the probe radius (0.25 mm). The measurement volume was approximately two orders of magnitude greater than the probe volume. Thus, tissue adjacent to the probe that may be traumatized constitutes only a small fraction of the sampling volume.

Following data collection a linear regression is performed on the temperature data. The perfusion is determined from a comparison between the measured temperature decay and a model generated temperature decay. The variance is estimated from the

measured and generated values; this will be referred to as the goodness of fit. When the generated temperature values lie on the measured cooling curve, the goodness of fit is small. A small goodness of fit indicates a small variance between the measured and modeled temperatures. Large values of the goodness of fit indicate poor data. For data obtained from in vivo experiments using the TPD method, typical values for the goodness of fit lie between 0.00001 and 0.001. When the goodness of fit exceeds 0.01 the measured temperature data are too erratic for calculating a faithful regression line from them. This indicates that the perfusion calculated from this temperature data set is not valid, and the sample point is discarded. Thus, the goodness of fit is an indication of the confidence placed on the calculated perfusion.

Interactions between thermometry systems and electromagnetic fields have been examined by several investigators [3-10]. The work has involved cataloging the response of temperature sensors to incident electromagnetic fields, and investigating methods to minimize undesired responses. Electromagnetic fields may cause three basic problems to any thermometry system: electromagnetic interference (noise) induced in the thermometry electronics and cabling associated with the sensor, excessive and artifactual heating of sensors constructed of resistive material (thermistors and thermocouples), and perturbation of electromagnetic fields by metallic components in the sensor. All of these problems can be reduced, or eliminated, if the sensor and its cables can be kept perpendicular to the electric field vector. In the living

animal, however, this remedy is complicated by the physical necessity of inserting the temperature probe into tissue in vivo, and making connections to remote thermometry electronics, coupled with the normal movement of the living tissues. The movement of a sensor, in which artifactual heating from electromagnetic waves occurs, will be detected as a change in temperature. This would be an erroneous measurement. Another complication encountered in tissue irradiation with electromagnetic waves is that frequently the near field of an antenna is used, as in the case of a contact antenna. The near field is characterized as having electric field components in all three principal directions, thereby rendering it impossible to orient the probe perpendicular to the electric field. The only solution to this particular problem is to ensure that the sensor is in the far-field of any radiating antenna.

The work performed under this contract has demonstrated several things. First, the TPD method is a new blood perfusion measurement technique; its nearly continuous perfusion measurements have revealed physiological changes in local tissue perfusion never before observed. There has not been enough time to investigate these physiological changes and determine what constitutes normal variations. This contributed significantly to the investigators' general inability to distinguish between physiological and microwave induced perfusion variations. Second, the probe-field interaction is a complex, nonlinear process which may not be avoidable and causes the system to yield unreliable data during microwave exposure periods. This factor

may render the present probe design unsuitable for use with the TPD method for measuring perfusion in a microwave field.

CHAPTER 2

IN VITRO MATERIALS AND METHODS

A simplified block diagram of the experimental system used in the present investigation is shown in Figure 1 (figures appear at the end of the thesis). The microwave source used in all the studies (Raytheon, Model PGM10X1) was provided by the EPA. It generates cw microwaves at 2,450 MHz at powers up to 100 W. The in vitro experiments were conducted in a constant temperature water bath using a 3.2 cm diameter direct contact microwave diathermy applicator, called the Elmed-35 (AT-502, Elmed, Inc., Addison, IL). Both the thermistor probe and a Vitek temperature sensing probe are implanted in tissue phantom material for these tests. The Vitek probe provides stable, accurate, and reliable temperature information for comparision with thermistor probe measurements made during each experiment [12].

An analog low pass filter was designed to remove high frequency components from the electrical signals generated in the thermistor probe by the electromagnetic waves. A circuit diagram for this filter and its transfer characteristic are shown in Figure 2. This circuit was used to provide the analog-to-digital converter on the DEC LSI-11 computer with stable signals used for the blood flow analysis.

The physical relationship between the thermistor probe and the antenna is shown in Figure 3. The separation distance (d) between the antenna and the probes can be easily changed by a mechanical adjustment. The orientation of the thermistor probe

with respect to the microwave fields is an important experimental parameter. The position of the antenna shown in Figure 3 results in a field orientation with the probes parallel to the direction of wave propagation. The electrical response of the thermistor probe was evaluated with the probe oriented parallel to either the electric field (E) or the magnetic field (H) or in the direction of propagation (K). Since the thermistor probe, unlike the Vitek probe, was not initially designed for use in the presence of microwave fields, these measurments provided information on the operation of the probe under microwave irradiation conditions.

CHAPTER 3

IN VITRO RESULTS

Preliminary studies of the effects of electromagnetic irradiation on the thermistor probe were performed using an air-coupled microwave antenna. Large variations were observed in the output of the irradiated thermistor probe, of the kind illustrated in Figure 4. These fluctuations are due to electromagnetic interference between the microwave field and the probe leads, as well as unshielded electronic components. Since the principal aim of this study is the evaluation of the TPD method and the present thermistor probe design in the tissues of living animals, all future *in vitro* studies were conducted using a direct contact antenna submerged in a water bath which eliminated some of this interference problem.

The response of the Vitek probe and the thermistor probe to 10 W of microwave power is shown in Figure 5. In Figure 5, the probe response sensitivity is approximately 0.6 ^{o}C/volt. These data were collected when both probes were parallel to the magnetic field (H) and located 2 cm from the antenna. The surrounding medium was a tissue phantom. The Vitek probe measures a linear temperature rise as expected for this 130 sec exposure. The thermistor probe also shows a linear response, but with a transient artifact due to an interaction between the probe and the microwave fields.

The probe artifact is seen only at the beginning and the end of the exposure. Because of the short duration of the artifact for this probe orientation, the artifact should not interfere with the pulse-heating of the thermistor bead that is needed for a determination of blood flow. However, this is not the case when the probe is positioned parallel to the electric field (E). Under the same experimental conditions a much larger artifact is seen when the microwave power is turned on and off (Figure 6). This orientation would be expected to produce the maximum interaction because the thermistor probe was parallel with the incident electric field (E) thus causing maximum artifactual heating of the probe, and maximum noise injected into the system through the probe leads. Additional experiments of this type indicated that the magnitude of the artifact depends on the relative orientation of the probe with the fields, the distance of the probe from the antenna and the amount of radiated microwave power. Increasing microwave power or moving the probe closer to the antenna results in larger artifacts. For example, Figure 7 illustrates the results when the probe is positioned parallel to the direction of propagation (K) at 1 cm from the antenna for 10 W of forward microwave power. The response is still linear but rapidly increases to a voltage level outside of the data sample window. In Figure 7, the probe response sensitivity is approximately 0.1 ^{o}C/volt.

In order to use the thermistor probe in the pulse-heating method for measurements of blood flow, the tissue temperature or the temperature gradient must be constant with time during the

applied heating pulse. This criterion can be satisfied using the existing thermistor probe by adjusting the microwave power level, the field orientation and antenna-tissue spacing. The data shown in Figure 8 illustrate an experimental measurement conducted in the dielectric phantom material. The cooling response of the thermistor immediately following the heating pulse provides the information for determination of the thermal diffusivity and perfusion. It was recognized that if the probe moves or if an electrical artifact is present while the cooling curve is being sampled, the artifactual heating of the probe will change, thus placing a discontinuity in the cooling curve. Such a perfusion measurment will have a large variance, and will be rejected by the curve fitting algorithms as an invalid data set.

The preliminary studies of the probe-field interaction demonstrated the need to reduce the interference between the probe leads and the incident fields, as well as the need to reduce thermal disturbances to the system. The studies also helped develop the irradiation system and networks used to match the antenna to the tissue.

CHAPTER 4

IN VIVO MATERIALS AND METHODS

After the preliminary study of the probe-field interaction, the measurements in the study proceeded along two parallel experimental paths. These paths were the development of improved probe shielding and the selection of an appropriate microwave antenna. The probe shielding was necessary to reduce the interference introduced into the system by the microwave field interacting with the wire leads and the thermistor probe. The antenna selection process included an examination of antennas having different heating patterns. The antennas examined were the Elmed-35, a planar spiral design and an E-plane sectoral horn design. The Elmed-35 is a circular, direct contact (near field) antenna having extremely narrow beamwidth. It has a 2 cm diameter spot heating pattern. The planar spiral has a circular aperture with a radius of 7.5 mm and a narrow beamwidth. The far field of this antenna starts approximately 1 mm from the aperture. All tissues irradiated by the planar spiral were in the far-field. The E-plane sectoral horn has a rectangular aperture with a 14 mm diagonal. It has a broad beamwidth, and can heat a large area. The far-field begins at 3.2 cm from the aperture; all tissues irradiated by this antenna were in this far-field region. The results of this phase of the study were used to select the antenna which delivered the microwave power efficiently and was minimally perturbing to the TPD System.

4.1. Experimental Procedure

The left or right kidneys of dogs (8-10 kg) and rabbits (2.7-4.9 kg) were used for the *in vivo* studies. All animals were fasted for 18-24 hr prior to surgery. Dogs were anesthetized with an i.v. injection of sodium pentobarbital (30 mg/kg). Rabbits were anesthetized with an i.m. injection of a mixture of xylazine (6 mg/kg) and Ketaset (60 mg/kg). All animals were placed on a heating pad to help maintain thermal balance. A surgical level of anesthesia was maintained with supplemental infusions of the respective anesthetic. The femoral artery or the carotid artery was cannulated to provide a measure of the systemic blood pressure (Gould Statham, P23ID) in both dogs and rabbits.

A lateral laparotomy was performed to expose the kidney, and one to six thermistor microprobes were inserted into the kidney cortex such that most of the probes would be positioned within the irradiated area. The probes were inserted into the kidney parallel to the direction of wave propagation (K) which is perpendicular to the electric field vector (E). The wire leads were also placed perpendicular to the (E) vector. This assumes that the kidney and probes were in the far-field of the antenna used. This condition was met for both the planar spiral and the E-plane sectoral horn antennas. However, it could not be met when using the Elmed-35 direct contact antenna. The Vitek temperature probe was placed on, or just beneath, the surface of the kidney nearest to the antenna and held in place with a suture. It provided accurate temperature information during the

microwave irradiation. The exposed kidney was then covered with a sheet of plastic wrap to minimize the heat loss via evaporative cooling and to aid in the maintenance of thermal homeostasis. The animal was then covered by a specially constructed Faraday cage which shielded the TPD instrumentation from interference by the microwaves. The frame of the cage also supported the antenna above the kidney, as shown in Figure 9. Both the planar spiral and Elmed-35 antennas were used in the experiments with dogs. The E-plane sectoral horn was used in the experiments with rabbits.

Once the tissue temperature stabilized, data sampling began. Blood flow measurements (samples) were typically made every 3 min for the duration of all the experiments. Each animal served as its own control. Measurements were made during the preexposure (control), the microwave exposure, and the postexposure periods. In the first experiment presented, there were two preexposure periods and two exposure periods since this experiment investigated the effect on the TPD system of the two antennas. Both blood pressure and tissue temperature were continuously monitored throughout the experiment. Following the postexposure period, the animal was sacrificed, and the kidney was removed for gross inspection and dissection. The dissection confirmed probe positions. The attainment of a constant tissue temperature or a tissue temperature which changes linearly with time before data sampling starts is crucial to the TPD method. The basic assumptions of the TPD method are that the system under study is near thermal steady state and that large temperature gradients do

not exist in the tissue. It is for this reason that before the exposure or postexposure sampling begins, the tissue is given time to reach a thermal steady state.

4.2. Data Analysis

Data obtained from individual perfusion probes were well behaved at some sampling times, but subject to extreme variations at other times. Because of persistent electrical noise and artifacts it was also difficult to collect valid data during a microwave exposure. For example, the blood pressure and the local tissue blood perfusion obtained from the perfusion probes during one experiment are shown in Figure 10. The data were self consistent, and there were six good measurements obtained from 16 samples during the microwave exposure. However, such results were not always the case. Figure 11 shows data obtained from other probes during the same experiment. Note the extreme variations in the data obtained from channel 1 during the microwave exposure. During this exposure the leads were completely shielded with saline soaked gauze, so at least in part, the reason for this variation in the perfusion data is most likely due to a change in the artifactual heating of the probe between perfusion samples. This is probably a result of a movement of the thermistor probe caused by normal, physiological movement of the animal's kidney. It was suggested that such data did not reflect an accurate assessment of the local tissue perfusion, but rather demonstrated that the electrical response of the TPD probe was being disturbed by the microwaves. In the same experiment, no data points were available from channel 6

during the exposure period. Again, since the leads were well shielded, this was probably due to a change in the artifactual heating of the thermistor probe. It can be seen that this probe was not broken, for it provided good data during the postexposure period. The data in this figure are typical of data obtained during an exposure period. In reviewing the local tissue perfusion data it becomes apparent that careful consideration must be given to the data presentation.

An ideal experiment would be one which presents nearly continuous, faithful measurements of the perfusion at several sites in the kidney. before, during, and after irradiation. However, since some of the data were highly variable and may not have reflected true perfusion measurements, while other data were discontinuous because many of the samples had a poor goodness of fit, it was decided to smooth the data by averaging. Two averaging schemes were chosen. One of these, called average total perfusion (PT), is a running average of discrete samples of local kidney perfusion integrated over a large volume of the kidney cortex. This average is calculated using the following equation:

$$\text{Average Total Perfusion} = PT(t) = \frac{1}{n}\sum_{i=1}^{n} P_i(t) \quad t = 0,3,6,\ldots T \qquad (1)$$

where n is the number of probes, Pi is the perfusion data obtained from the ith probe at one sample time, t is time in minutes, and T is the total duration of the experiment in minutes. The second average. called average local perfusion

(P_{li}), calculates the mean perfusion measured by each probe over the period of interest. It is given by

$$\text{Average Local Perfusion} = P_{li} = \frac{3}{T+3} \sum_{m=0}^{M} P_i(3m) \quad i = 1,2,\ldots n \qquad (2)$$

where $t = 3m$, $M = \frac{T+3}{3}$

where T is the duration of the period of interest (i.e., preexposure, exposure, postexposure). The three arises from the duration of the sampling interval and $P_i(3m)$ is the perfusion sample obtained from the i^{th} probe at a given time. The P_{li} will show any change in the average local perfusion measured by a particular probe due to the presence of the microwaves. It also smooths out the extreme variations in the average data. When average total perfusion is plotted against time, the absence of a data point every 3 min on the perfusion line indicates that the data on all channels at that instant in time had a goodness of fit which exceeded 0.01. This may indicate that all the TPD probes were severely perturbed at that time.

A test for statistical significance was performed on P_{li} to determine whether the perfusion changed significantly from one period to the next. The Smith-Satterthwaite test, a hypothesis testing procedure, are a t-test performed on populations having differing variances [12]. The data which did change significantly from the previous period are denoted by an asterisk above the error bar in the respective figures.

CHAPTER 5

RESULTS AND DISCUSSION

The data from Experiment 1 are presented in Figures 12 and 13. The experimental subject was a dog. Shown in Figure 12 is an overview of the entire experiment including the P_T, microwave power, tissue temperature and blood pressure. In the first microwave exposure period the planar spiral antenna was used. During the second exposure period the Elmed-35 contact antenna was used. When using the planar spiral, high power levels were needed to raise the tissue temperature, whereas the Elmed-35 needed much less power to achieve similar temperature increases. This demonstrates an impedance mismatch between the planar spiral and the air-tissue system. It was found that during the first exposure, increasing microwave power levels resulted in a decrease in valid data points. The increased variability of the data suggests that there was a direct relationship between the radiated power level and the introduction of noise into the system. During the second exposure there were fewer data points; this shows that in the near-field of the Elmed-35 antenna there is strong coupling between the field and probe which causes a severe perturbation of the TPD system. The data presented in Figure 13 show the P_{1i} during the four parts of this experiment. Only the perfusion in channel 4 changed significantly ($P < 0.05$) between the control period and the first microwave exposure. Five probes were positioned within the microwave field, and one probe was placed in a portion of the kidney which was not exposed to microwaves. This unexposed probe is shown in Figure 13 to be

in channel 4. Since the probe was not in the microwave field, the change in perfusion cannot be directly attributed to the direct effect of microwave irradiation. However, it could be due to an effect which the increased temperature had on kidney vasculature, circulating hormones, or vasoactive substances.

During the experiment illustrated in Figure 13 there is a lack of data during exposure 2. Probes 1 and 2 had excessive noise levels, and hence yielded no data during this exposure, but they did yield data once the microwaves were turned off. Channel 3 gave only one valid data point during this period, while channel 4, which was outside the field, was the only probe to yield data during this period. The poor data are due to the aforementioned probe-field interaction because the probe is in the near-field, along with the spot heating pattern which creates large thermal gradients within the tissue. The results of this experiment demonstrate that the Elmed-35 contact antenna is not the proper choice for the system. The planar spiral, while it does increase the standard deviations, seemed to be an acceptable choice from the viewpoint of thermal perturbations. However, the large amount of power needed to achieve a desired tissue temperature increase demonstrated that the radiation efficiency of this antenna is poor. Two new antenna criteria were defined based on the results of this experiment. First, the antenna must avoid spot heating. The best way to meet this condition was to find an antenna with a beamwidth broad enough to irradiate the entire kidney, heating it evenly. Second, the antenna must be an efficient tissue heater. These criteria led to the selection of

an E-plane sectoral horn antenna. The radiation pattern of this antenna is shown in Figure 14. If the H-plane is oriented parallel to the long axis of the kidney, then an entire small animal kidney can be heated. The relatively small kidneys of rabbits satisfy this condition, so the remainder of the experiments were performed on rabbits.

A new configuration of the microwave radiation system including the horn antenna was then tested, and the results are presented in Figures 15 and 16. The average kidney P_T, the blood pressure, tissue temperature, and microwave power are shown in Figure 15. The perfusion tended to follow the blood pressure both before and after the exposure, but not during the exposure. These results suggest that during the microwave exposure the perfusion is altered. However, the data were extremely variable during the exposure period, giving large standard deviations. This suggests that the microwave fields were interacting with the TPD probes directly, or indirectly, through some thermal perturbation. The P_{li} for this experiment is presented in Figure 16. The large standard deviations in the preexposure period are due to highly variable data obtained during the first 25 min of the experiments. The variability, which was due to unknown causes, was no longer evident after 25 min and very good control data were obtained for the rest of the preexposure period. The local tissue perfusion was observed to decrease in direct proportion to the overall decrease in systemic blood pressure. This lowering of perfusion may be due, in part, to the large thermal stress placed on the kidney by the $3^{o}C$ temperature rise.

These temperature effects cannot be separated from the effect of systemic blood pressure. The data show that all channels of perfusion data changed ($P < 0.05$) from the exposure period to the postexposure period.

During this experiment, a new probe design was tested. It consisted of 3/16 inch diameter braided copper shield which covered the probe leads to within 5 mm of the temperature sensing thermistor. With the probe inserted into the tissue, the shielding rested on the surface of the kidney. This probe produced no data other than preexposure data, not presented in the figures. Upon inspection of the probe and kidney after the experiment, a large hemorrhage was observed around the probe. The tissue appeared to have been severely overheated, while the probe was irrepairably damaged. It appeared that the braided shield perturbed the microwave field, concentrating the field energy on the kidney at the probe site. This significantly heated the surrounding tissue structures and the imbedded probe. The experiment served as a guideline for new shielding techniques, since it demonstrated the undesirable effects caused by the relatively bulky configuration of the braided shielding.

The system performance was adequate with respect to the antenna selection. However, it was deemed necessary to modify the design of the lead-wire arrangement in order to reduce the probe-field interaction. The next design was a four wire probe. It consisted of a wire, of the same type used to energize the probe, wound around the other three wires. The spacing between adjacent windings was kept much less than a wavelength and the

fourth wire was grounded. The purpose of this was to eliminate the lead-field interaction, without increasing the lead diameter to a point where it would perturb the incident field. This probe was then tested against the standard three wire probes, and the results are presented in Figures 17 and 18. The average total perfusion data are shown in Figure 17. During the exposure period there were only seven good samples for each probe, when 13 samples were taken. The postexposure data are also very poor, although upon inspection all of the probes were functioning properly. This poor postexposure data may be attributed to tissue damage due to the high temperature reached during the exposure period. High temperatures must be avoided, especially for long exposure periods. The average local tissue perfusion for this experiment is presented in Figure 18. The four wire probe data are in channel 3. This probe measured a change ($P < 0.05$) in perfusion between the preexposure and exposure periods. This, coupled with the lower standard deviation, seemed to indicate that proper shielding of the probe leads would improve the system performance. However, even the four wire probe gave many poor samples during the exposure period.

For the next experiment the leads were tucked under saline soaked gauze for shielding, and both three and four wire probes were used. The data for this experiment are shown in Figures 19 and 20. As shown in Figure 19, the P_T follows the blood pressure throughout most of the experiment. There were very few points discarded during the experiment, and the data points shown in Figure 19 which do not have error bars are points where both

channels of available data agreed with each other exactly. These very good data can be partially attributed to the complete shielding of the probe leads by the saline soaked gauze. Also, there was more time allowed for the system to reach thermal steady state before and after the exposure. Another factor could be the smaller temperature change during the exposure, and the relatively low maximum temperature. Data presented in Figure 20 show that the Pli data do not change significantly during, or following microwave exposure at the level used in this study. These data indicates that perfusion does not change as a result of microwave exposure. The perfusion measured by a four wire probe is shown in this figure to be in channel 4. Its standard deviations are large during the exposure which demonstrates that lead shielding is not a complete solution to the interference problem. although the data suggest that there are significant improvements when shielding is provided, as outlined above.

Figures 21 and 22 present the data from the last experiment performed under this contract. From the data presented in Figure 21 it can be seen that the initial temperature of the rabbit was low, and that the maximum temperature never exceeded 40 degrees. All probe leads were placed under saline soaked gauze for shielding. The data were highly variable during the exposure period as demonstrated by the very large standard deviations. Also, at several sample points, data were available from only one probe (the points without error bars). The data presented in Figure 21 show that sampling can take place in a microwave field, as evidenced by the great number of valid data points obtained

during the microwave exposure. However, the data so obtained are variable. The average local perfusion data, shown in Figure 22, indicate the same increased variability in the data during the exposure period. It is for this reason that the large changes in measured perfusion shown are not statistically significant. The data having the smallest variations are shown in this figure to be in channel 4. These data were obtained from a four wire probe and again show that the four wire probe provides greater shielding than the three wire probe. However, it is still subject to loss of data points and variations due to changes in the amount of probe self heating.

CHAPTER 6

CONCLUSION

In the course of this investigation the response of a tissue implantable thermistor probe for determining blood perfusion was tested, in the presence of microwave fields, under both *in vitro* and *in vivo* conditions. The results reported here suggest that under very carefully controlled conditions of electrical shielding the TPD method may offer the opportunity to evaluate local tissue perfusion in the microwave field. However, the use of a sensor which is sensitive to microwaves is a significant deterrent to its general use in situations where special shielding conditions must be implemented. Nevertheless, the efforts made to shield the probe leads from the incident fields have resulted in an improvement in probe response. The elimination of thermal gradients within the irradiated tissues has also led to improved system performance. Although proper probe orientation can eliminate artifactual heating of the thermistor, the movement of living subjects generally cannot be avoided. This move results in constantly changing probe orientation with respect to the incident microwave fields and alters the amount of artifactual heating. The data obtained in the present investigtion support the conclusion that the present probe design is not suitable for general use in a microwave field.

There are several probe designs which could be investigated in the future. A non-interacting TPD probe could be constructed from an optical temperature probe, such as the Luxtron Florescent Temperature Probe. The optical probe design could be modified to allow for pulse heating of the tissue by a laser light source. This probe would not perturb the incident fields, nor be perturbed by these fields. Utilizing this probe, the TPD method may provide measurements of local tissue blood perfusion during a microwave exposure. Another probe design which may be suitable for this application is a probe having separate circuits for temperature sensing and pulse heating. If the temperature sensing circuit consists of a high resistance thermistor bead having very lossy leads, such as the Vitek probe, it could accurately measure temperature under all exposure conditions. Meanwhile, the same thermistor could be pulse heated by another set of lossless leads which could be grounded when they are not supplying the power pulse. This probe could be readily adapted for use with the TPD method.

Other methods of measuring tissue perfusion are being developed here at the University of Illinois. Professor M. M. Chen has recently proposed a technique based on the basic principals of heat transfer. This method predicts that the rate at which local tissue temperature changes following an increase in the global tissue heating (due to irradiation) can be shown to reflect a measure of local tissue convective (perfusion) heat transfer. This technique would utilize several Vitek temperature probes to monitor the tissue heating. These of course are

unperturbed by the microwave fields and would serve as the entire electronic system, other than data aquisition. In conclusion, there has been considerble progress on this contract towards a solution to the probe interaction problems. However, it does not appear feasible to use the currently developed TPD probe for measurements of tissue perfusion in the presence of intense microwave fields.

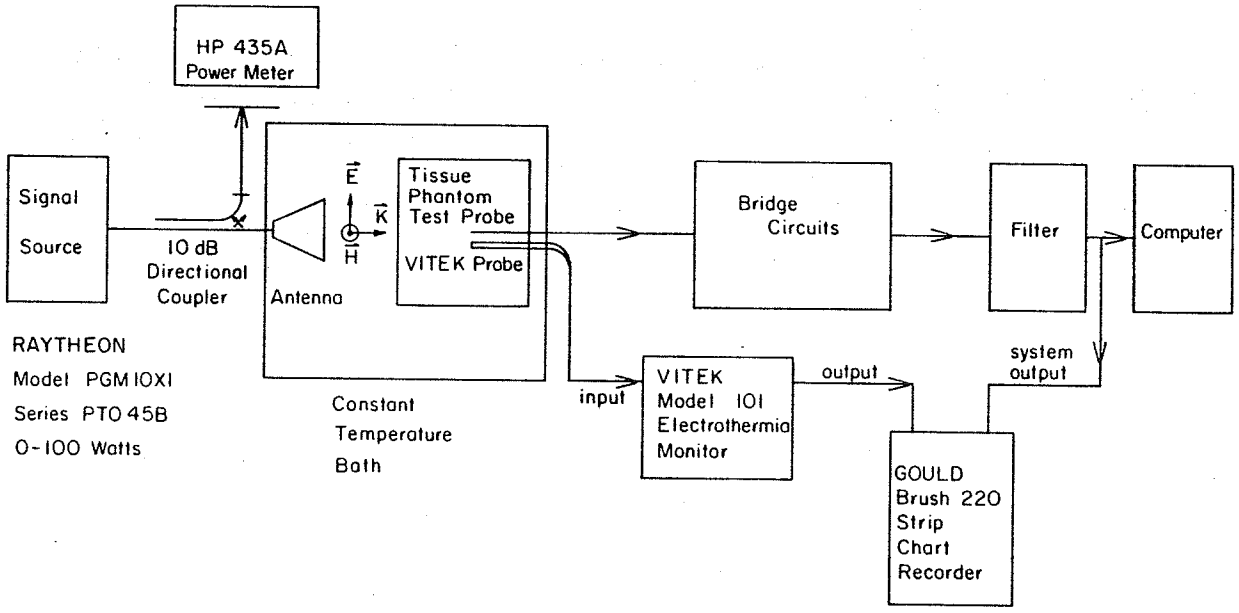

Figure 1. System block diagram.

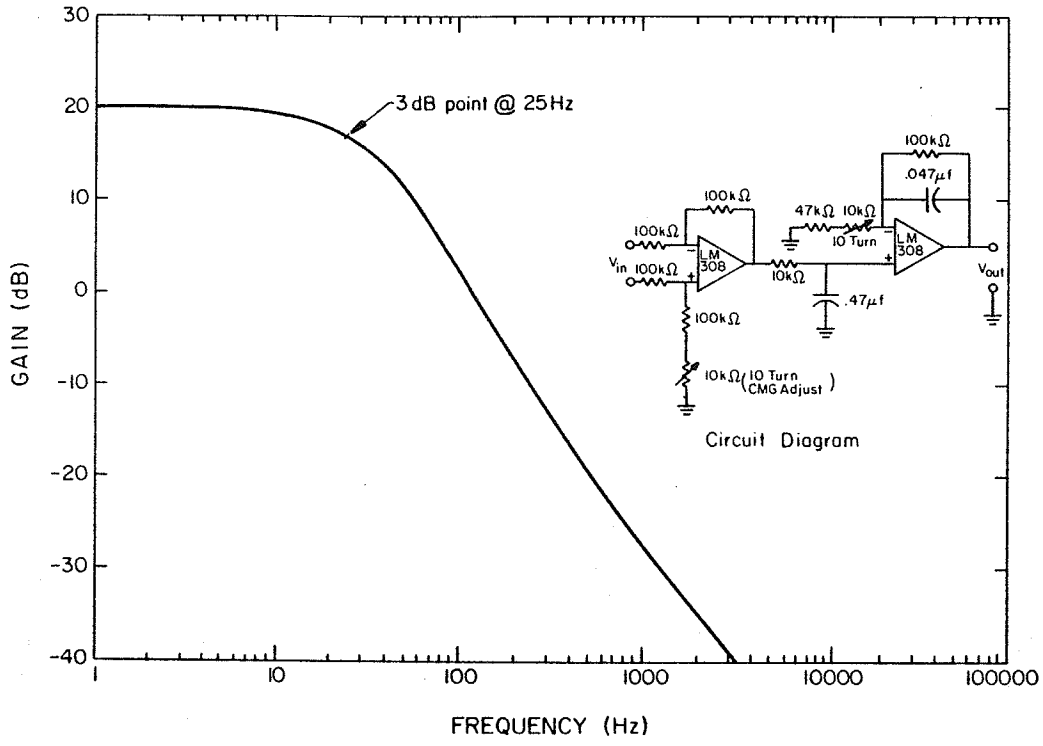

Figure 2. Active filter transfer characteristics, V_{out}/V_{in}.

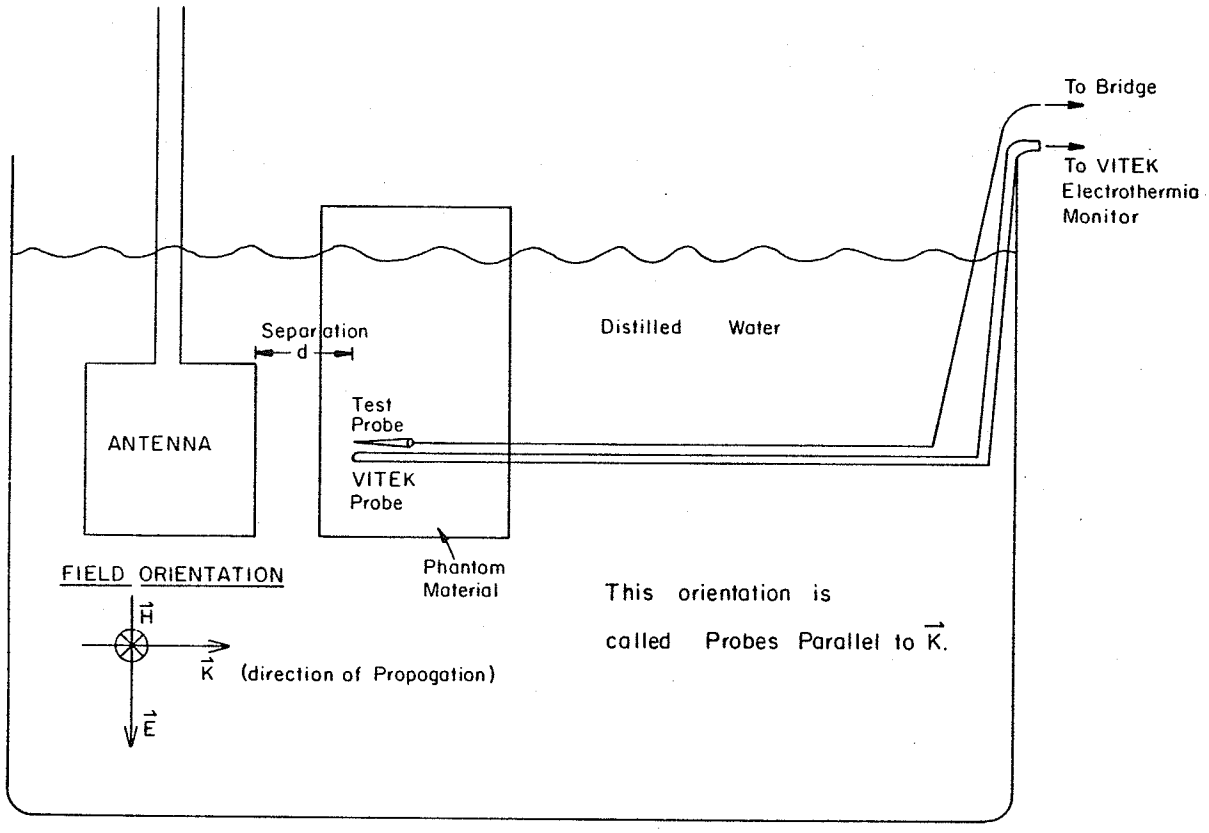

Figure 3. Physical relationship of probe to antenna.

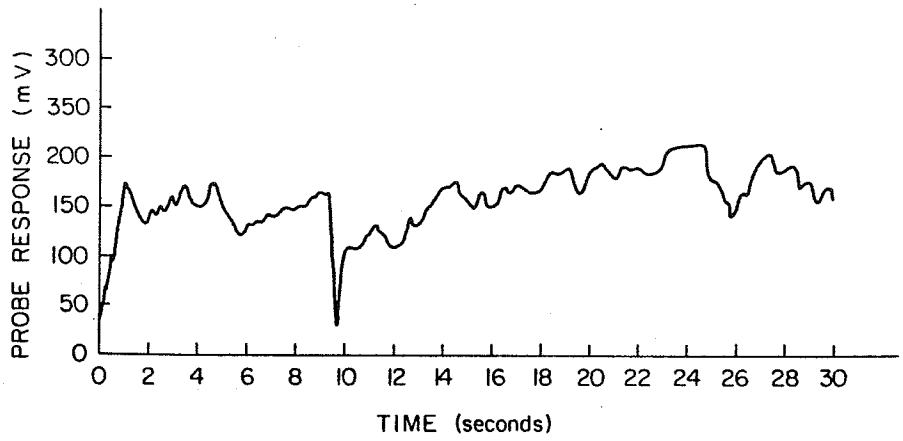

Figure 4. Thermistor probe parallel to E with 2 cm air gap.

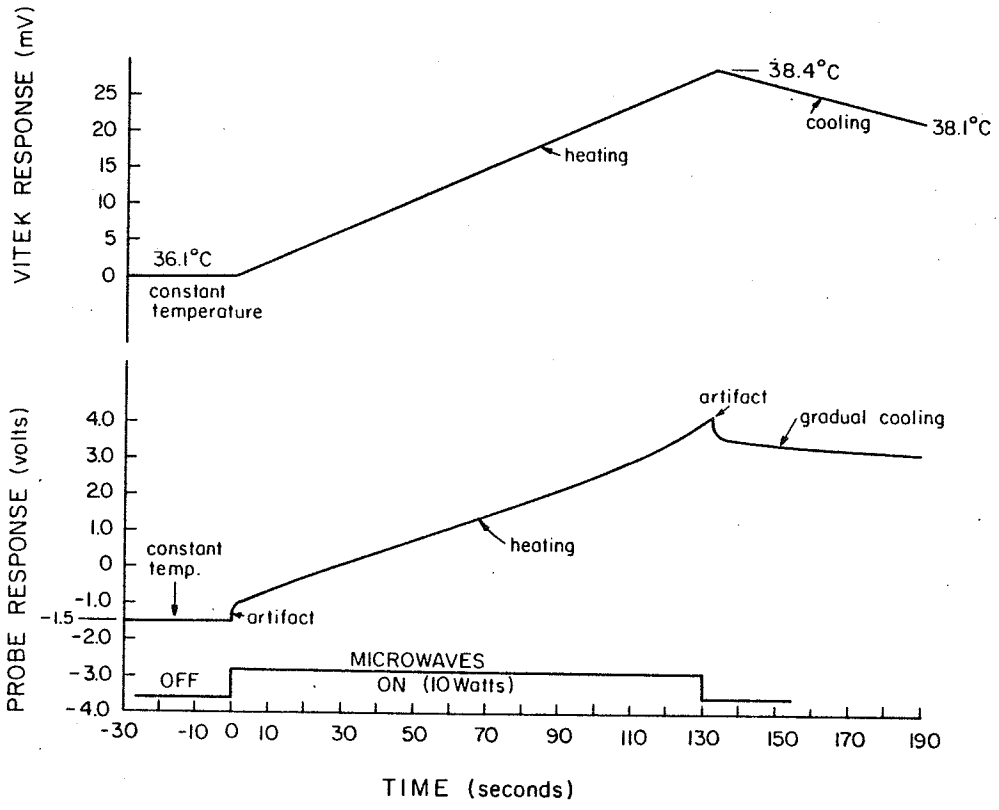

Figure 5. Thermistor probe parallel to H at 2 cm from antenna.

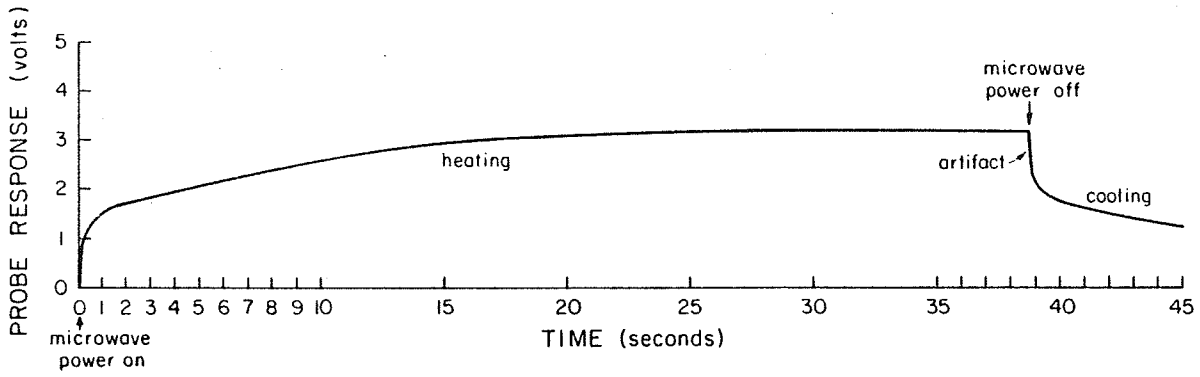

Figure 6. Thermistor probe parallel to E at 2 cm from antenna.

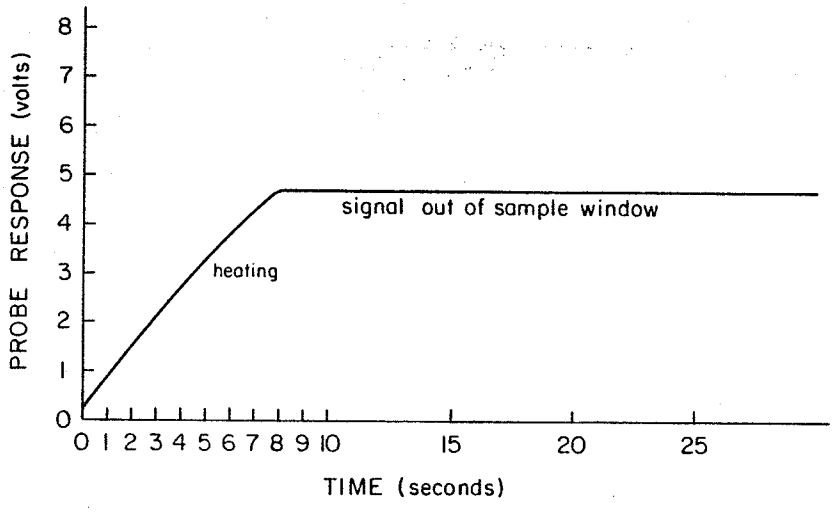

Figure 7. Thermistor probe parallel to K at 1 cm from antenna.

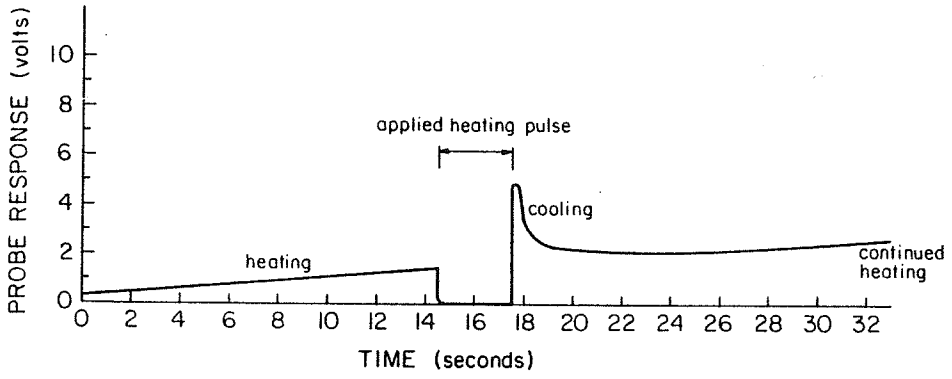

Figure 8. Thermistor probe parallel to E at 2 cm from antenna.

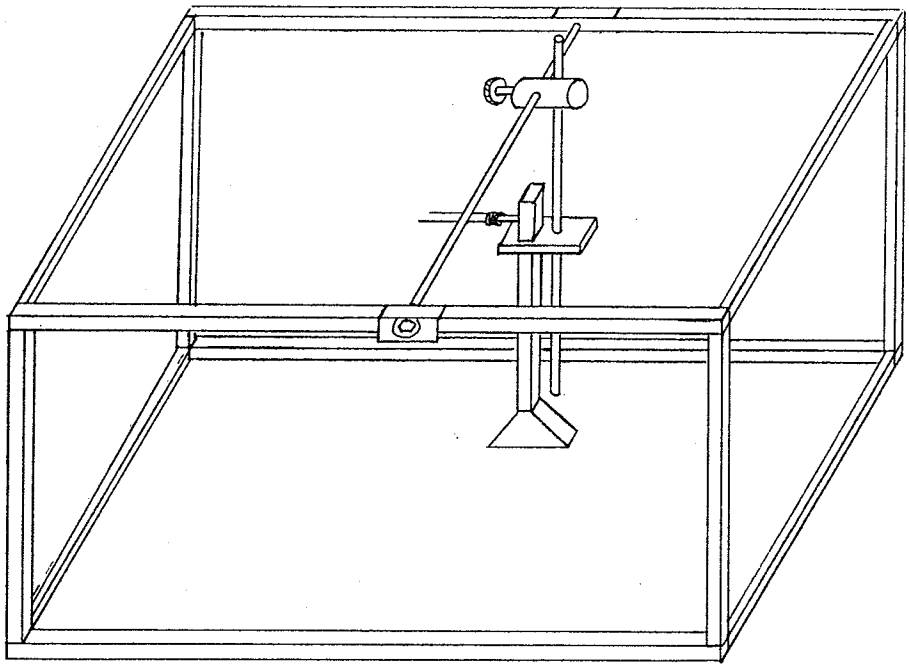

Figure 9. Screen covered Faraday cage supporting antenna positioner and E-plane sectoral horn.

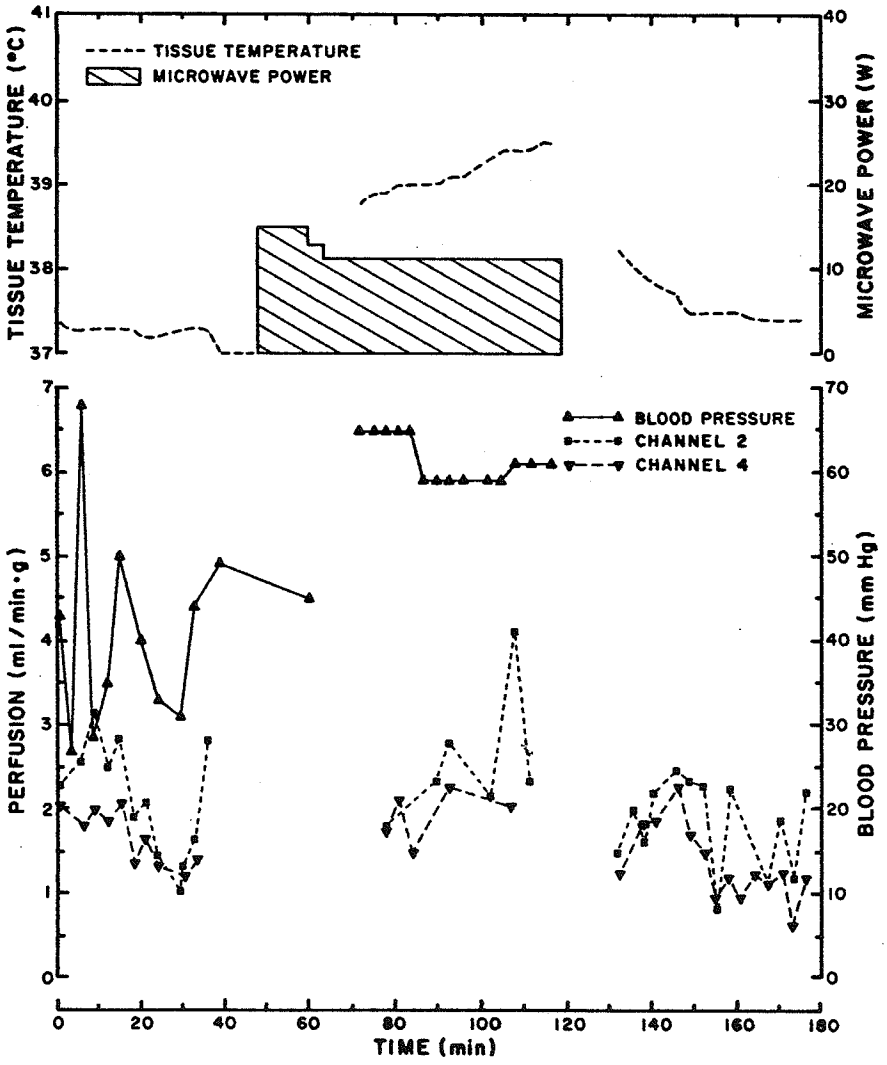

Figure 10. Local kidney blood perfusion (ml/min g), microwave power (W), tissue temperature (°C), and blood pressure (mmHg) plotted vs. time (min).

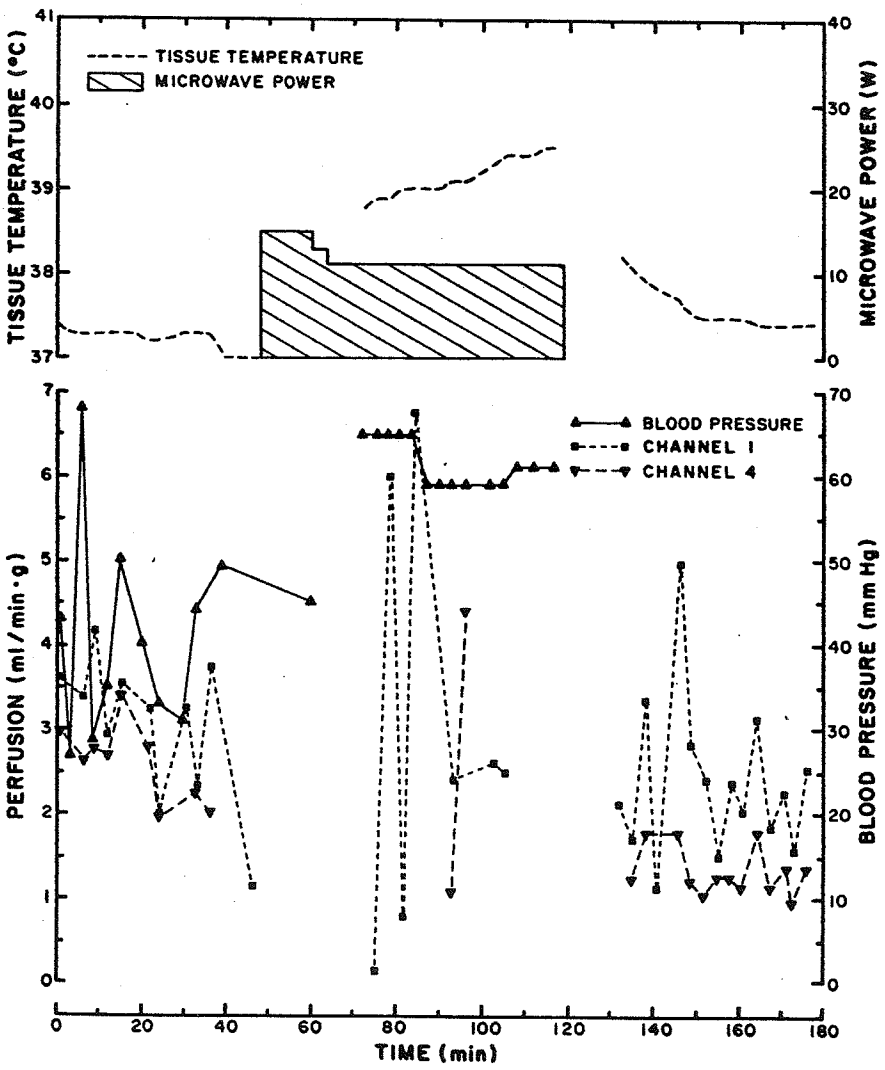

Figure 11. Local kidney blood perusion (ml/min g), microwave power (W), tissue temperature (°C), and blood pressure (mmHg) plotted vs. time (min).

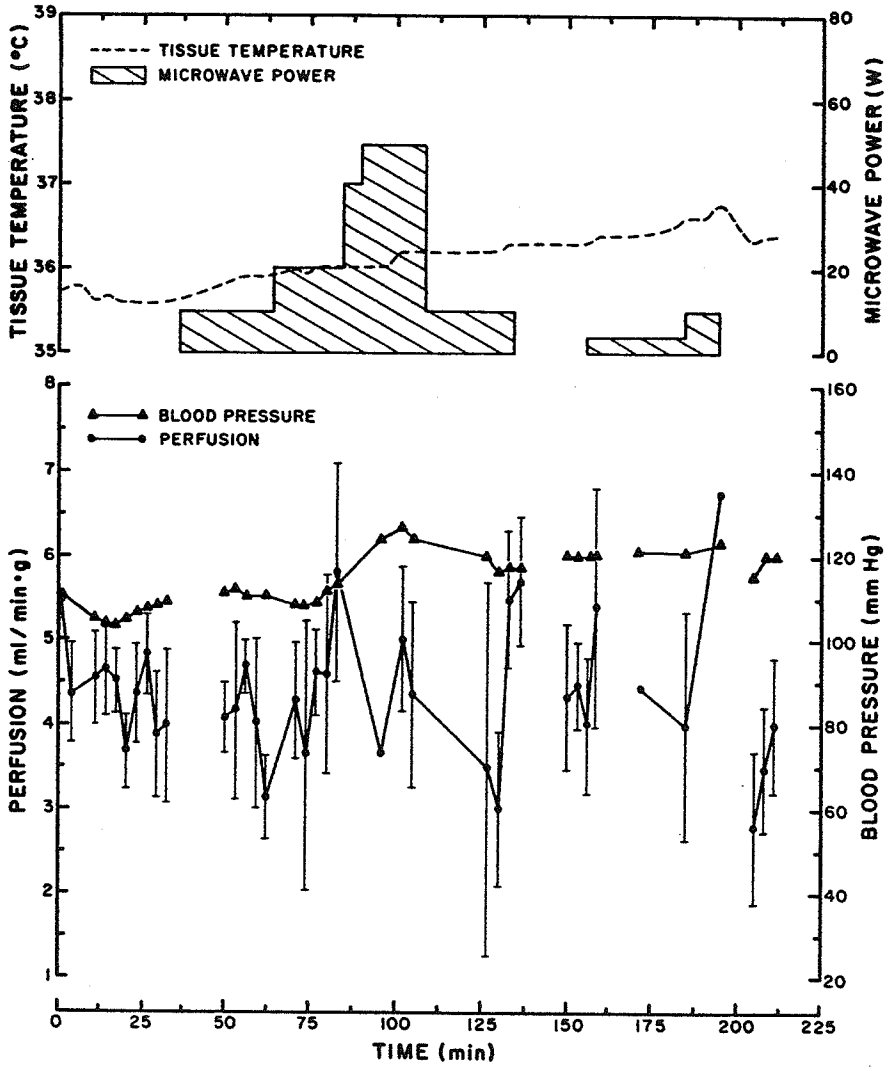

Figure 12. Experiment 1. Average total kidney perfusion (ml/min g), microwave power (W), tissue temperature (°C), and blood pressure (mmHg) plotted vs. time (min). Planar spiral antenna used during first exposure, Elmed-35 used during the second.

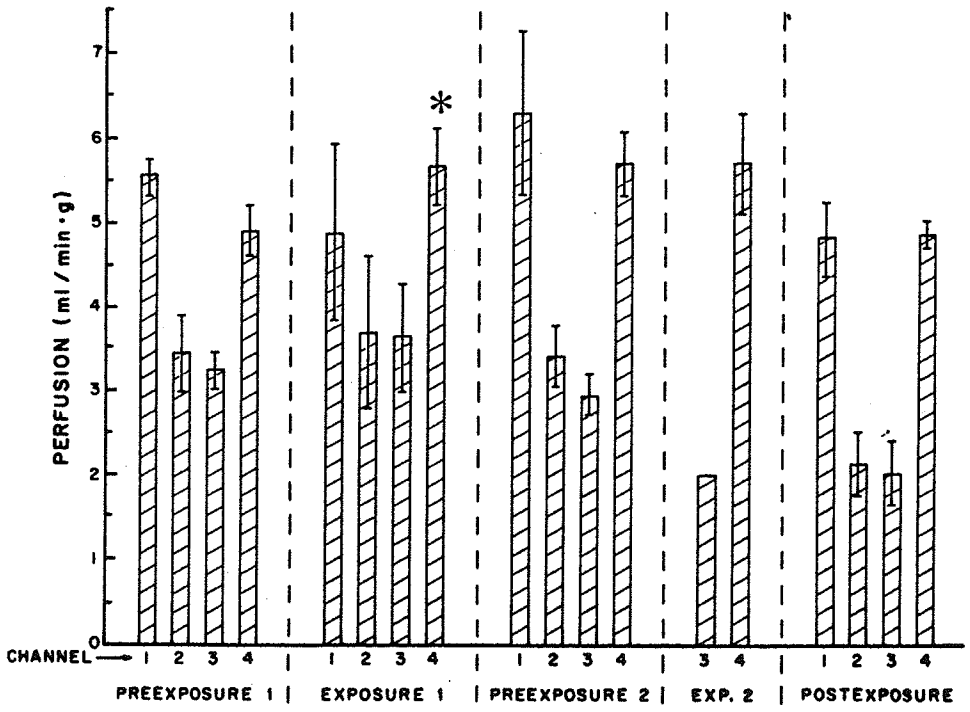

Figure 13. Experiment 1. Average local kidney perfusion (ml/min g).

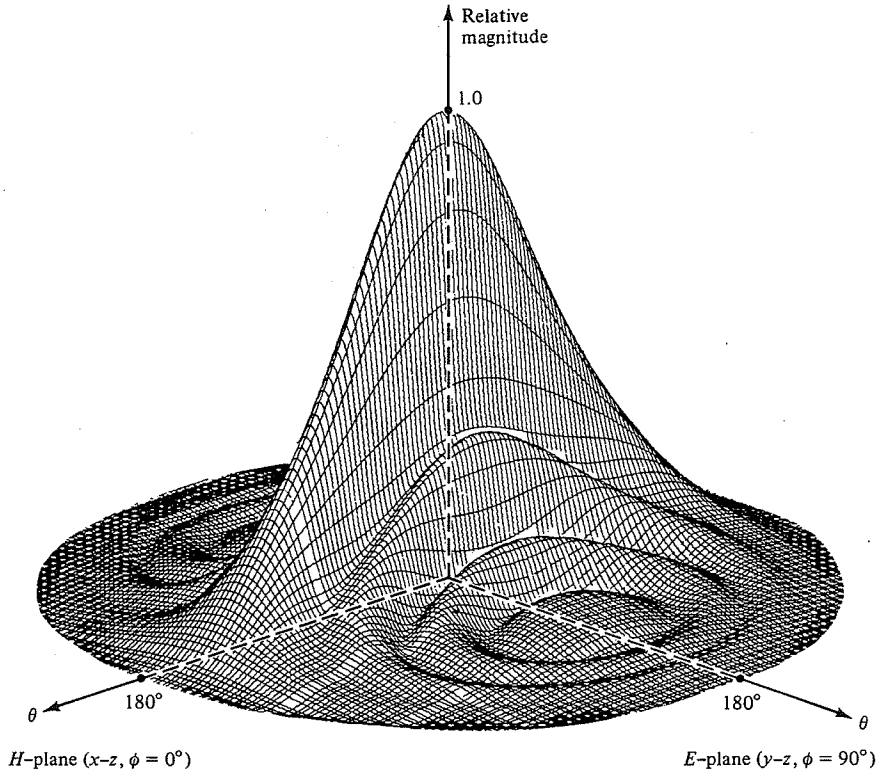

Figure 14. Three dimensional field pattern of E-plane sectoral horn antenna [13].

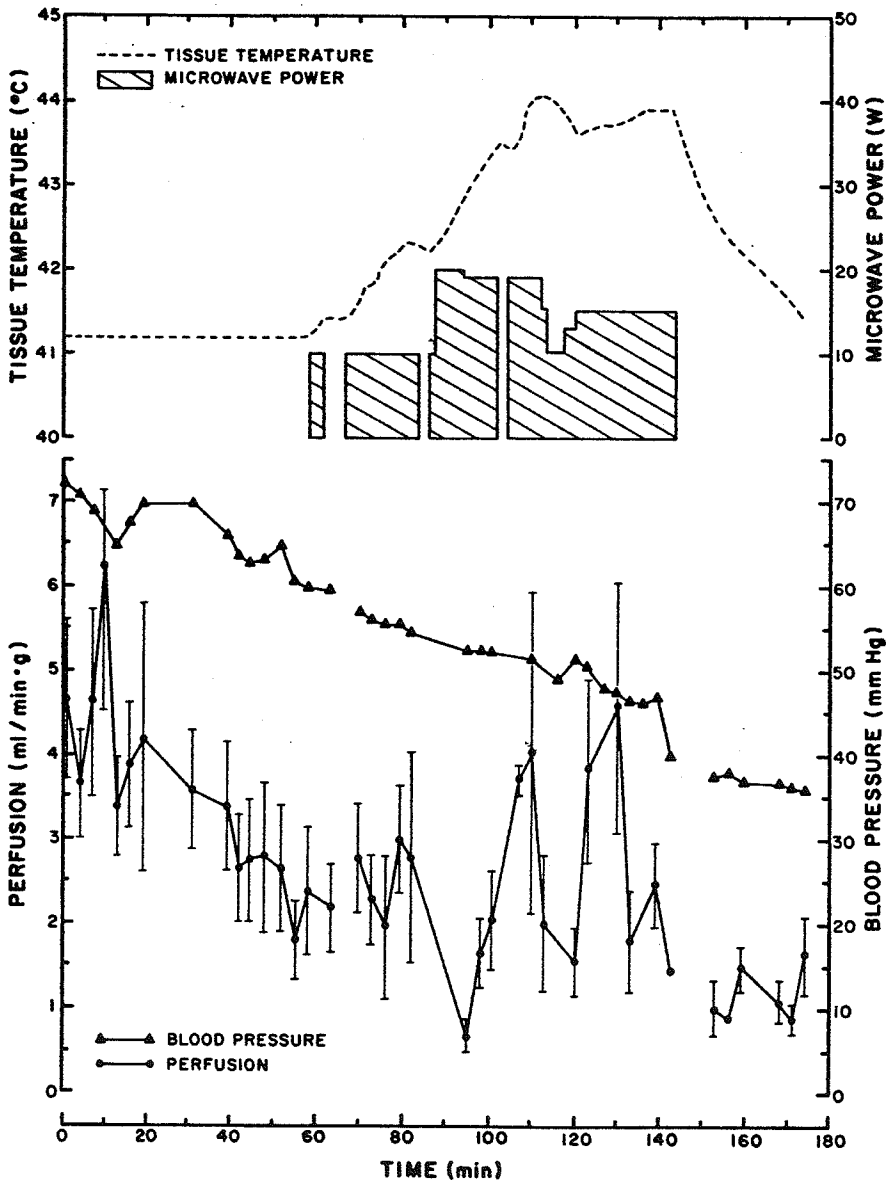

Figure 15. Experiment 2. Average total kidney perfusion (ml/min g), microwave power (W), tissue temperature (°C), and blood pressure (mmHg) plotted vs. time (min).

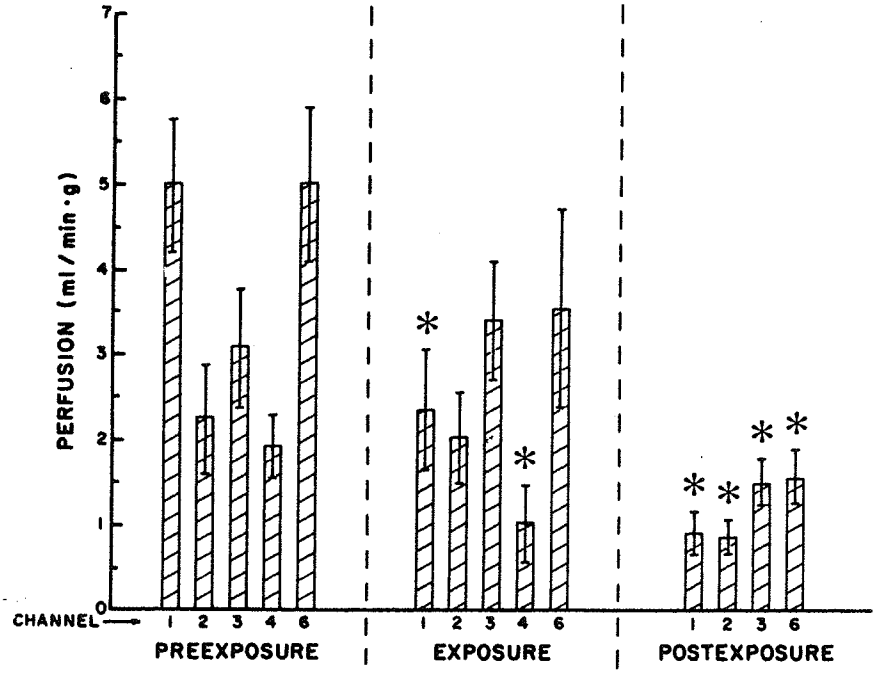

Figure 16. Experiment 2. Average local kidney perfusion (ml/min g).

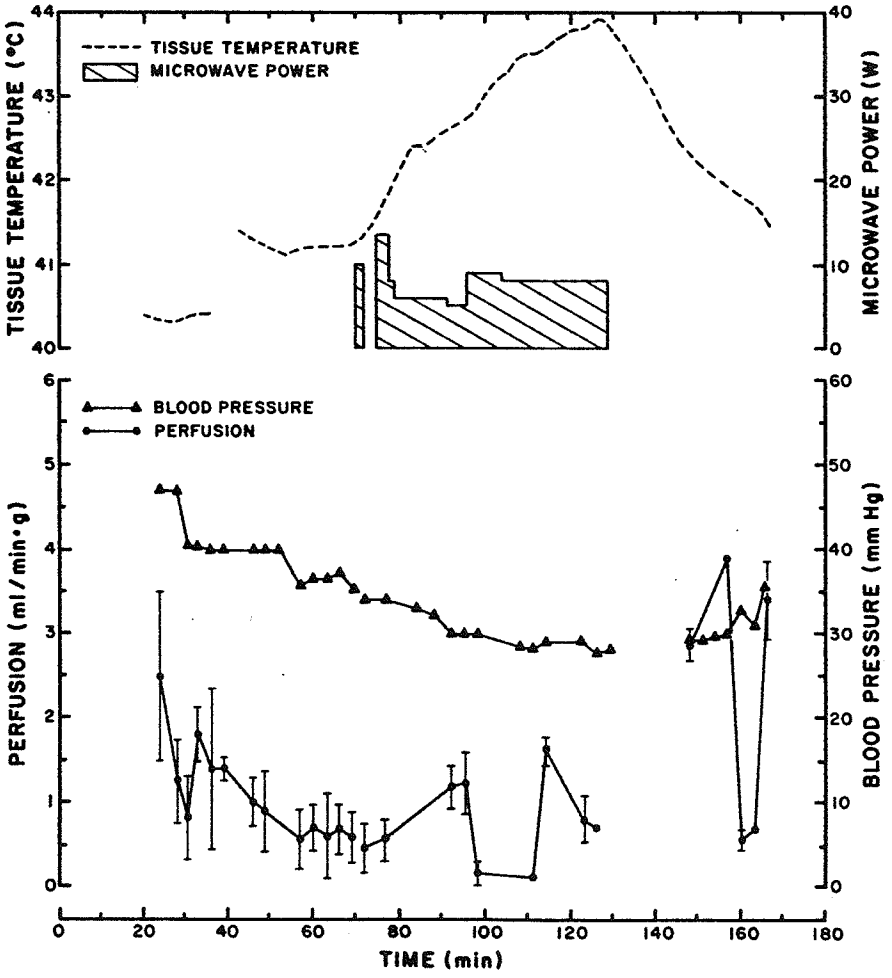

Figure 17. Experiment 3. Average total kidney perfusion (ml/min g), microwave power (W), tissue temperature (°C), and blood pressure (mmHg) plotted vs. time (min).

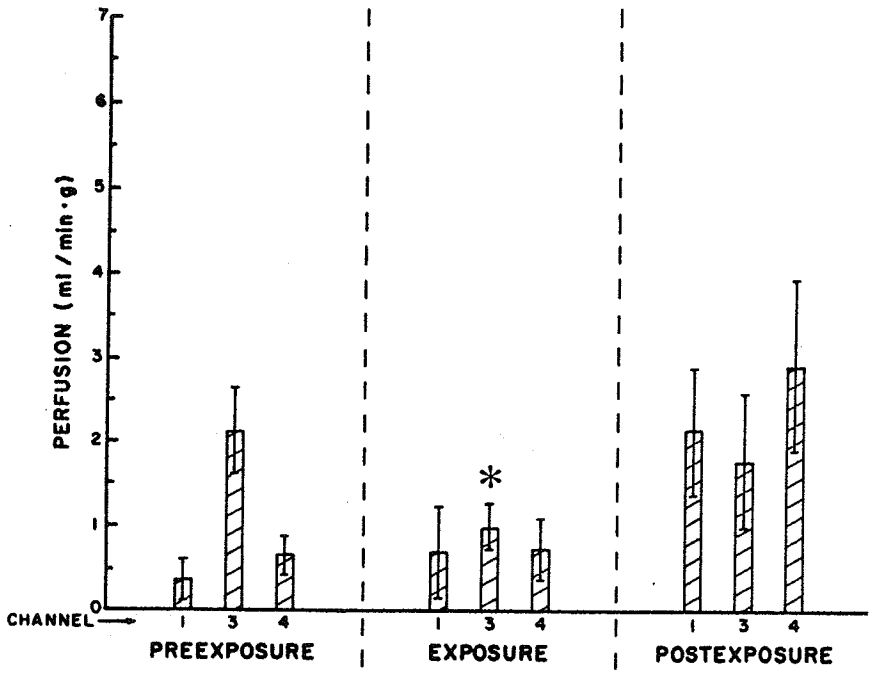

Figure 18. Experiment 3. Average local kidney perfusion (ml/min g). Four wire probe is in channel 3.

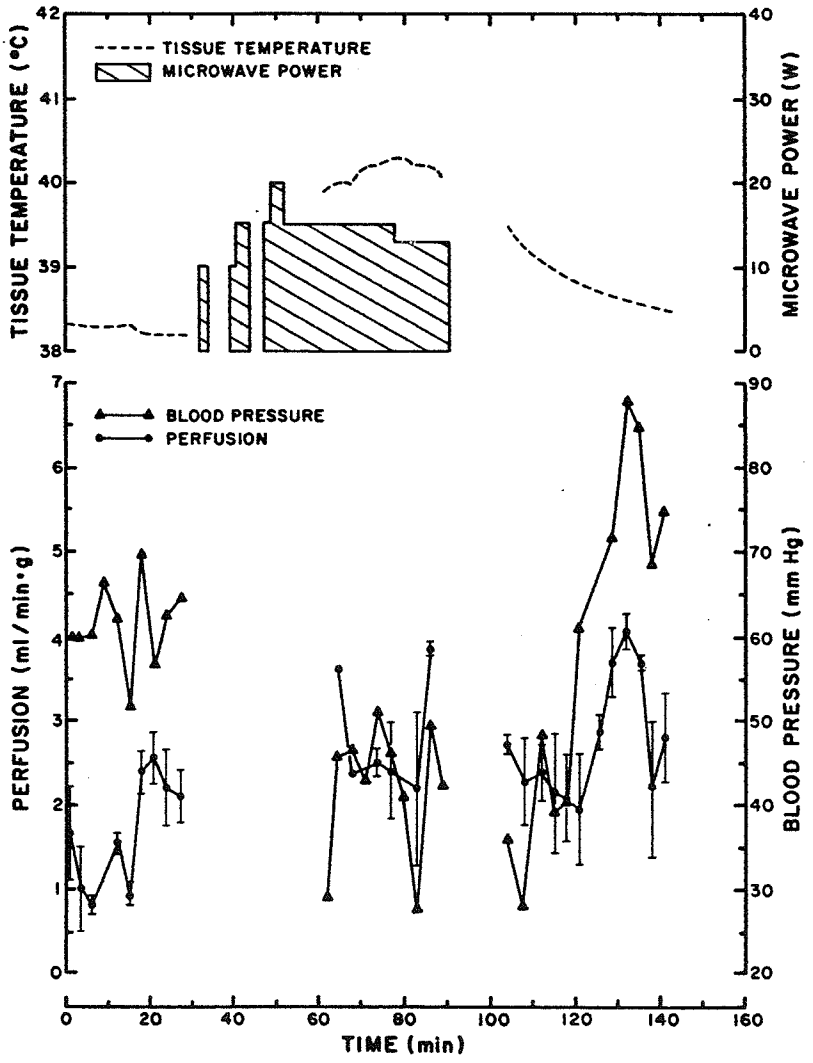

Figure 19. Experiment 4. Average total kidney perfusion (ml/min g), microwave power (W), tissue temperature (°C), and blood pressure (mmHg) plotted vs. time (min).

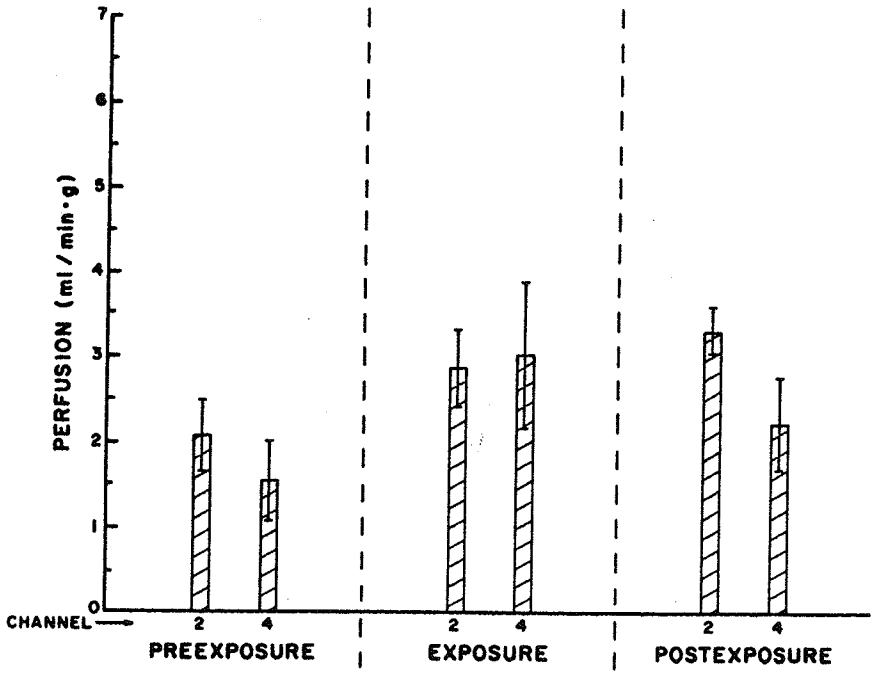

Figure 20. Experiment 4. Average local kidney perfusion (ml/min g). Four wire probe is in channel 4.

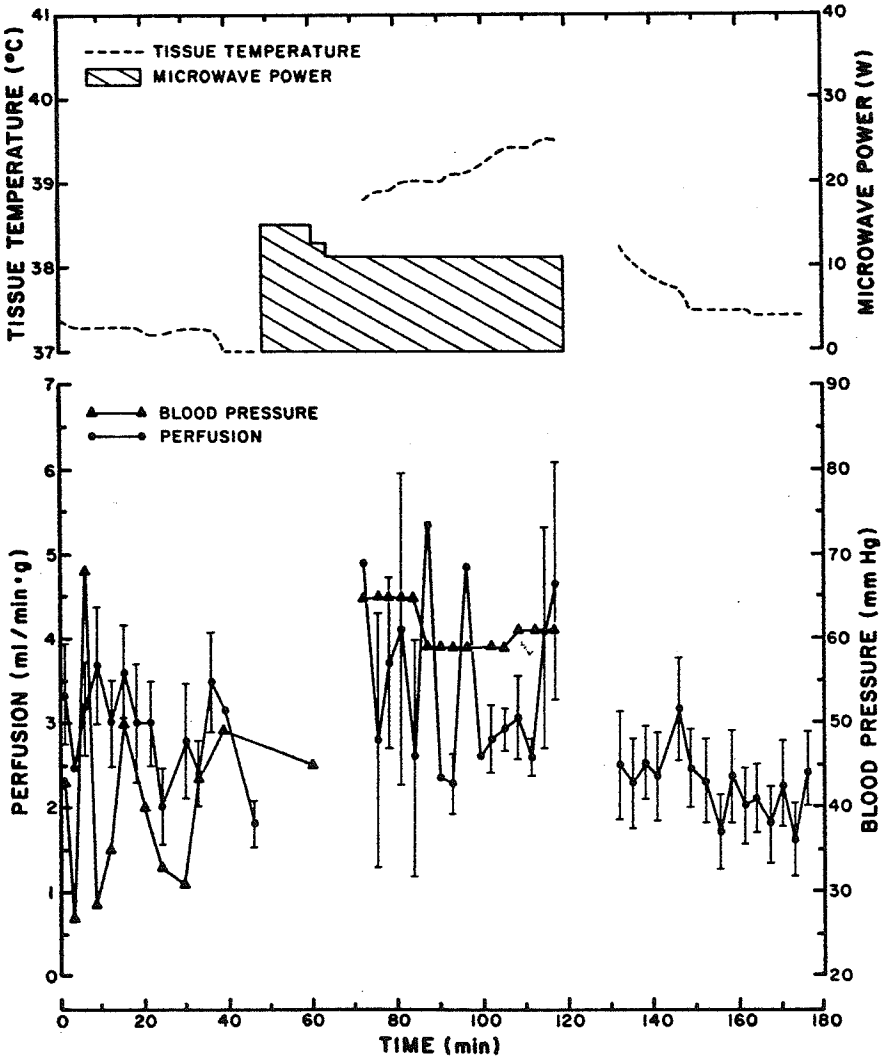

Figure 21. Experiment 5. Average total kidney perfusion (ml/min g), microwave power (W), tissue temperature (°C), and blood pressure (mmHg) plotted vs. time (min).

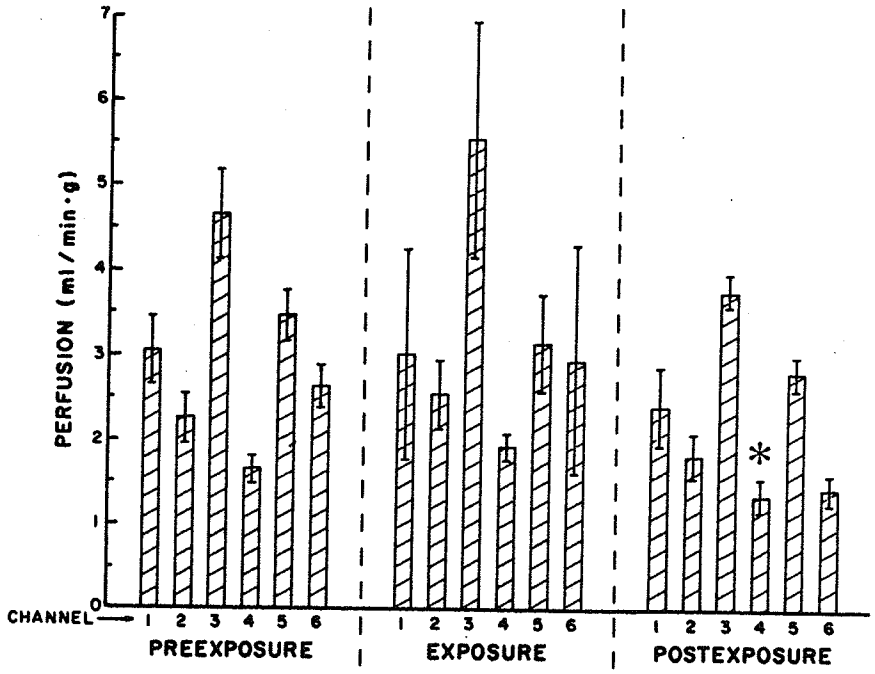

Figure 22. Experiment 5. Average local kidney perfusion (ml/min g). Four wire probe is in channel 4.

REFERENCES

1. Chen, M. M. and Holmes, K. R. "Thermal Pulse-Decay Method for Simultaneous Measurement of Thermal Conductivity and Local Blood Perfusion Rate of Living Tissues," Advances in Bioengineering, ASME, 1980, pp. 113-115.
2. Holmes, K. R. and Chen, M. M. "Local Tissue Heating:Microprobe Technique for Heat Transfer Parameter Evaluation," Measurement of Blood Flow and Local Tissue Energy Production by Thermal Methods, Evaluation of Methodology, Muller-Schauenburg, editor, Thieme-Stratton, Inc., New York, 1983, pp. 50-56.
3. Cetas, T. C. and Connor, W. G. "Hyperthermia Thermometry," J. Medical Physics, vol. 5, No. 2, p. 86, Mar./Apr. 1978.
4. Cetas, T. C. "A Birefringent Crystral Optical Thermometer for Measurements of Electromagnetically-Induced Heating," in Program and Abstracts 1975 Annual Meeting, International Union of Radio Science, 20-23 October 1975, Boulder, CO (USNC/URSI, National Academy of Sciences, Washington), p. 255.
5. Hahn, G. M. Hyperthermia and Cancer, Plenum Press, New York, 1982, p. 217.
6. Rozzell, T. C., Johnson, C. C., Durney, C. H., Lords, J. L., and Olsen, R. G. "A Nonperturbing Temperature Sensor for Measurements in Electromagnetic Fields," J. Microwave Power, vol. 9, pp. 241-249, 1974.

7. Johnson, C. C. and Guy, A. W. "Nonionizing Electromagnetic Wave Effects in Biological Materials and Systems," Proc. IEEE, vol 60, pp. 692-718, June 1972.

8. Larsen, L. E., Moore, R. A., and Acevedo, J. "A Microwave Decoupled Brain-Temperature Transducer," IEEE Trans. Microwave Theory Tech., vol. MTT-22, pp. 438-444, April 1974.

9. McAfee, R. D., Pazenavette, L. L., and Shubert, H. A. "Thermistor Probe Error in an X-band Microwave Field," J. Microwave Power, vol. 9, September 1974.

10. Magin, R. L., and Kantor, G. "Comparison of the Heating Patterns of Small Microwave (2450 MHz) Applicators," J. Bioengineering, vol. 1, pp. 493-509, 1977.

11. Bowman, R. R. "A Probe for Measuring Temperature is Radio-Frequency-Heated Material," IEEE Trans. Microwave Theory Tech., vol. MTT-24, pp. 43-45, January 1976.

12. Devore, J. L. Probability and Statistics for Engineering and the Sciences Brooks/Cole Publishing Co., Montery, CA, 1982.

13. Balanis, C. A. Antenna Theory Harper and Row, New York, 1982, p. 540.