Measurement of Radio Frequency Permittivity of Biological Tissues with an Open-Ended Coaxial Line: Part II—Experimental Results

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Abstract—The permittivity of several reference liquids and selected biological tissues *in vivo* was measured in the frequency range from 0.01 to 1 GHz. Open-ended coaxial line sensors and computer-controlled network analyzer systems, described in a companion paper, were used. The results were analyzed and compared with the estimated uncertainties. The described method proved to be convenient, fast, and relatively accurate for *in vivo* measurements.

I. INTRODUCTION

AN OPEN-ENDED coaxial line, which has been previously analyzed [1] is a viable sensor for *in vivo* permittivity measurements at radio frequencies (RF). When used with a computer-controlled network analyzer this method offers convenience and provides good accuracy of measurements [1]. The line offers several advantages when compared with the short antenna sensor [2]. These include better compatibility with the measured material, no necessity for custom designed components (short-circuit and matched load), and simple relationships between the permittivity and the measured reflection coefficient.

There exists a relatively large data bank of the *in vitro* dielectric properties of biological substances, but only recently have a few biological substances been measured *in vivo* [3]. The permittivity of tissues *in vivo* is often different from that *in vitro* [3]. Knowledge of these properties is important in evaluating potential hazard of RF radiation, and in biomedical applications such as hyperthermia for cancer treatment and radiometry for cancer detection.

In this paper calibration methods of the open-ended coaxial line sensor are described. The accuracy of the method as limited by systematic and nonsystematic errors is evaluated experimentally and compared with theoretical estimates given in [1] for several reference substances.

Finally, the permittivity data obtained *in vivo* for muscle (skeletal and smooth), liver, kidney, spleen, and pancreas of a cat are reported.

II. EXPERIMENTAL PROCEDURES

A. Network Analyzer Error Correction and Check

The accuracy of the reflection coefficient measurement by the network analyzer was improved by a standard error correction procedure [4]. The reference plane for the open-line sensors is at the end of the line [1], and this poses some practical difficulties. As discussed elsewhere [1], for the probe to be compatible with *in vivo* measurements, there should be no "ground plane" around the line opening, and therefore a flat metal surface may not provide a good electrical short-circuit. The procedure employed in this work was the placement, under pressure, of a flexible metal foil backed by a thick rubber slab. The matched load is connected at the probe-connector plane rather than the reference plane (end of the line). This is an acceptable solution as long as the connector of the probe can be assumed perfect, and the changes of the line impedance are small; APC-7 connector used with the test probes when properly mounted introduces negligibly small reflections at frequencies of interest.

The source-match correction [4] includes not only discontinuities inside the network analyzer, but also all the discontinuities associated with connectors, cables, etc., between the network analyzer input port and the probe. Therefore, the number of transitions should be minimized, flexible cables eliminated, and only the best quality air line sections used when necessary.

B. Capacitance Evaluation

The total capacitance of the open line C_T, the fringe capacitance associated with the air part of the sensor C_0, and the fringe capacitance due to the fringe field inside the teflon-filled line C_f must be known. Two experimental methods have been used to determine the capacitances C_0 and C_f. In the first method a transmission-type cavity is formed from a coaxial line which is used as a probe in such

a way that the open-ended line (probe) constitutes a part of the cavity. Measurements of the resonant frequencies when the cavity end is open- and short-circuited allow the determination of the total capacitance $C_T = C_0 + C_f$ [5]. If the measurements are performed for the line open into air and also into a well known dielectric, then C_0 and C_f can be separated as follows:

$$C_0 = \frac{C_{T2} - C_{T1}}{\epsilon_2' - 1} \qquad (1)$$

$$C_f = C_{T1}\epsilon_2' - C_{T2} \qquad (2)$$

where C_{T1} is the capacitance (determined from the measured resonance frequency) of the line open into air, and C_{T2} is the capacitance of the line in contact (immersed) with a low-loss dielectric having known dielectric constance ϵ_2'. Dielectrics such as Teflon or carbon tetrachloride can be conveniently used.

In the second method the capacitances are determined from the input reflection coefficient measured by the computer-controlled network analyzer when dielectrics of well known properties are in contact with the probe. An approximate value of the total capacitance C_T has to be known to perform error corrections for the system. The dielectrics used for calibration not only have to be of known permittivity, but their permittivity should be such that at a selected measurement frequency, the optimum capacitance condition is satisfied [1]. Otherwise, large errors may result. As a first approximation C_f can be assumed to be equal to zero, and either ϵ' or ϵ'' of the calibration material can be used to determine C_0 [5]. The following relationships are employed, the first for ϵ' and the second for ϵ'':

$$C_0 = \frac{-2\Gamma\sin\phi}{\omega Z_0 \epsilon' (1 + 2\Gamma\cos\phi + \Gamma^2)} \qquad (3)$$

or

$$C_0 = \frac{1 - 2\Gamma}{\omega Z_0 \epsilon'' (1 + 2\Gamma\cos\phi + \Gamma^2)} \qquad (4)$$

where Γ is the magnitude and ϕ is the phase of the input reflection coefficient, and ϵ' and ϵ'' are the dielectric constant and loss factor of the calibration dielectric, respectively.

Saline solutions and water are good calibration dielectrics at frequencies 0.1–1 GHz due to the small uncertainty in measurements [1]. Calibration at higher frequencies to determine C_0 should be avoided since above a certain frequency (determined by the coaxial line dimensions), C_T is a function of frequency.

Determination of C_f is more difficult since low dielectric constant materials have to be used and for these the accuracy of measurements is poor

$$C_f = \frac{-2\Gamma\sin\phi}{\omega Z_0 (1 + 2\Gamma\cos\phi + \Gamma^2)} - \epsilon'C_0. \qquad (5)$$

The process of determining final values of C_0, C_f, and C_T requires a few iterations. After C_0 and C_f are determined, a new value of C_T is used in the error correction procedure and the calibration is repeated.

C. Reference Materials

Several materials of well-characterized dielectric properties used as references to evaluate the method and compare the errors in the results with the estimated uncertainties of measurements [1]. Distilled water and methanol were used along with saline solutions of 0.02, 0.08, and 0.25 molarity. The dielectric constant and loss factor (or conductivity) of saline solutions can be calculated as a function of molarity [6]. A 0.02-molarity NaCl solution has a conductivity approximately equal to that of biological tissues of low water content (e.g., fat, bone) while a 0.08-molarity NaCl solution has conductivity and dielectric constant approximately equal to that of biological tissue of high water content (e.g., muscle). For the 0.25-molarity NaCl solution the capacitance of the experimental probe was close to the optimum value for the range of frequencies of interest. The temperature of the reference materials was measured with an estimated uncertainty of ±0.1°C.

D. In Vivo Tissue Measurements

All *in vivo* measurements were done on a female feline (cat) under anesthesia induced with pentobarbital (35 mg/kg) and maintained with ketamine HCl (30 mg/kg). The animal was surgically prepared and monitored on a life support system. Rectal temperature was monitored and controlled with a heating pad and heat lamp. During the first two hours after anesthesia the temperature was 35°C, later rising to 36.5°C (normal 37.5°C).

Various tissues were surgically exposed for measurements. Skeletal muscle was exposed following a skin incision in the thigh. Smooth muscle and internal organs were exposed following a medial abdominal incision. In all cases, tissue exposed to air was maintained moist with an application of normal saline. Just before the tissue was placed in contact with the test probe it was blotted to remove the surplus saline. Tissue temperature was measured with accuracy of ±0.2°C using a digital temperature probe (Vitek) immediately prior to or following the dielectric measurements. At least three different locations of the same tissue type were measured. Consecutive measurements at the same location yielded identical data, and therefore were not done routinely, but only occasionally, to check the system operation. In cases where the tissue was too thin (less than 5 mm) it was folded double to secure at least 5 mm of the tissue under the test probe.

III. EXPERIMENTAL RESULTS

A. Probe Capacitance

The capacitances of the 8.3-mm (0.325-in) probe, measured by the resonance method, were $C_0 = 0.046 \pm 0.005$ pF and $C_f = 0.009 \pm 0.0005$ pF. Teflon was used as a test dielectric and the measurements were performed at three frequencies, 1.2, 1.8, and 2.4 GHz.

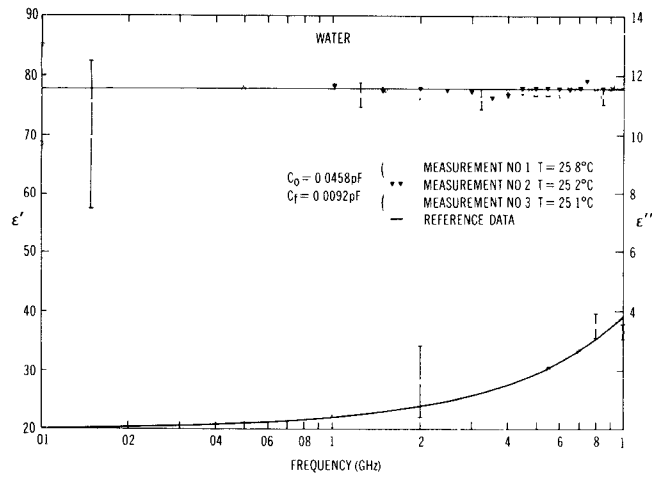Measurements were then made of the permittivity of

Fig. 1. Permittivity of water as compared with the reference data drawn as a continuous line. Vertical bars associated with the experimental points show the uncertainties of measurements determined based on systematic errors ($\Delta\Gamma = 0.003$, $\Delta\phi = 0.3°$).

TABLE I
CAPACITANCE C_0 OF THE 8.33-MM (0.325-IN) PROBE

Calibration Material and Parameter (in brackets)	C_0 Average (pF)	Standard Deviation (%)	Maximum Deviation (%)
Water (dielectric constant)	0.0458	±1.2	+1.9 −3.0
0.25M NaCl (dielectric constant)	0.0455	±0.6	+1.3 −0.7
0.25M NaCl (conductivity)	0.0461	±0.2	±0.3
Methanol (dielectric constant)	0.0468	±2.5	+3.8 −6.1

TABLE II
THE AVERAGE DIFFERENCE IN PERCENT BETWEEN THE MEASURED
PERMITTIVITY OF A VERY THICK AND LIMITED THICKNESS SAMPLE
AS A FUNCTION OF THE SAMPLE THICKNESS

Sample	Thickness (mm)			
	10	7.5	5	3
Water (distilled)	0	1.4	13	-
0.08M NaCl	0	0	1.0	3

various standard liquids using the network analyzer (HP 8410B). The results for C_0 obtained with these standard dielectrics are summarized in Table I. The measurements were performed for each dielectric at ten frequencies between 100 and 110 MHz.

The fringe capacitance, $C_f = 0.0092$ pF, was determined using carbon tetrachloride, Teflon, and air as reference materials. Typical standard deviations in these measurements were ±15 percent. The total capacitance was, therefore, $C_T = 0.055$ pF. A good agreement between the capacitance values obtained by the two methods was observed.

B. Sample Size

A simple experiment was conducted to find the minimum thickness of the sample in contact with the 8.3-mm probe which is equivalent to an infinite sample. Measurements were performed for water and a 0.08-molarity saline solution, simulating high water content tissues. The results are summarized in Table II. The measurements were performed at frequencies from 0.01 to 1 GHz, and the effect of the sample thickness was found to be relatively insensitive frequency.

C. Reference Materials

The dielectric constant and loss factor of distilled water measured with the two probes are shown in Fig. 1. The solid lines show the data calculated from the Debye equation for 25°C

$$\epsilon = \epsilon_\infty + \frac{\epsilon_0 - \epsilon_\infty}{1 + (j\omega\tau)^{1-\alpha}} \qquad (6)$$

with $\epsilon_0 = 78.3$, $\epsilon_\infty = 4.6$, $\tau = 8.07$ ps, and $\alpha = 0.014$ [7]. The points indicate the experimental data. Each set of measurements was taken after a separate error correction procedure. It should be noted that two measurement systems were used, the system based on the HP8407 network analyzer at frequencies 10–100 MHz, and the system based on the HP8410 network analyzer at frequencies 100 MHz–1 GHz [1]. The vertical bars indicate the uncertainty in measurements due to systematic errors in the measured reflection coefficient. The system errors were $\Delta\Gamma = 0.033$ and $\Delta\phi = 0.3°$ [1]. The experimentally determined loss factor data at frequencies below 100 MHz is not shown as it was subject to large uncertainties as expected [1].

The permittivity of methanol obtained experimentally is compared with the reference data [8] in Fig. 2. Generally, for both water and methanol the measured values lie within the estimated uncertainty limits [1] as compared with the reference data. Similar results were obtained for the selected saline solutions; however, as predicted, smaller errors accompanied the measurements [1].

Relative errors in the conductivity measurements of the saline solutions are shown in Fig. 3. The relative uncertainty in the conductivity is equal to the relative uncertainty in the loss factor. The actual errors are within the estimated limits.

The measurements on the reference materials have shown that when proper care is exercised during the measure-
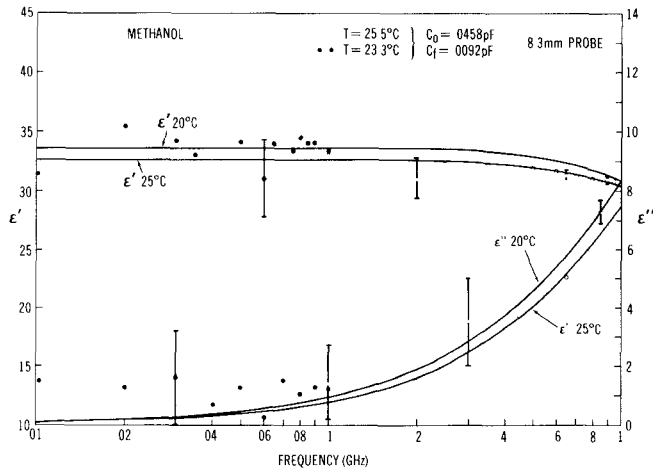
Fig. 2. Permittivity of methanol as compared with the reference data drawn as a continuous line. Vertical bars show the uncertainties of measurements ($\Delta\Gamma = 0.003$, $\Delta\phi = 0.3°$).
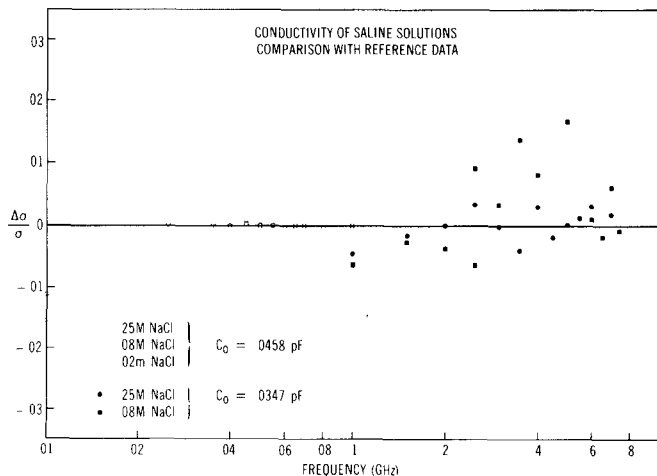

Fig. 4. Dielectric constant and conductivity of skeletal muscle tissue of a cat, measured with the 8.3-mm probe.

Fig. 3. Typical errors in the conductivity measurements for three saline solutions.

Fig. 5. Dielectric constant and conductivity of smooth muscle of a cat, measured with the 8.3-mm probe.

ments to minimize nonsystematic errors, the results deviate from the reference data by an amount not more than was estimated on the basis of systematic errors only [1]. However, during the course of measurements nonsystematic errors were encountered on several occasions. The most common errors were due to the following factors.

1) There could be imperfect contact with the test sample, e.g., formation of a small air bubble was occasionally observed while measuring water samples. This effect was clearly evident as the measured permittivity values at all frequencies were smaller than the actual values.

2) Temperature drift, particularly due to changes in the length of the coaxial line connectors was consistently observed. Therefore, the measurement system was turned on for at least 3 h before the tests were performed.

3) Imperfect connections during calibration usually resulted in large correction parameters and were eliminated before proceeding to further tests.

4) Occasionally, noise, lack of repeatability of connections, or other unidentifiable causes would result in the measurements at one frequency having a much greater
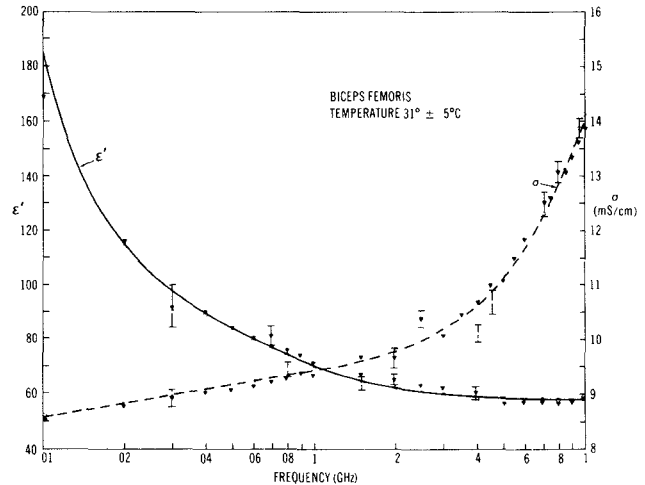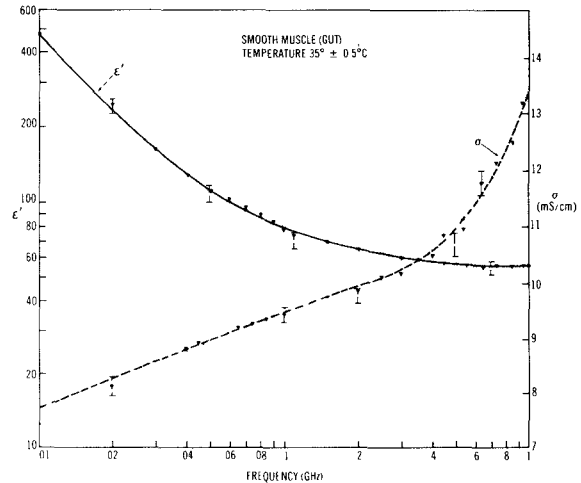
error than estimated. Normally, repetition of the error calibration procedure eliminated this error.

D. In Vivo Tissues

The permittivity of two types of muscle of a cat *in vivo* is shown in Figs. 4 and 5, while the results for internal organs kidney, spleen, and liver are provided in Figs. 6–8. The different symbols indicate typical data obtained for various parts of the same tissue. Consecutive measurements at the same tissue location produced virtually identical results. The solid curves were drawn to represent an approximate average of the experimental points, taking into account three or more sets of data in each frequency range and adjusted to match at 100 MHz. The vertical bars indicate the estimated uncertainties (shown only at representative frequencies).

The dielectric constant and loss factor of muscle tissues, biceps femoris, and smooth muscle (gut) do not vary appreciably from one sample to the other. Variations of the measured properties of such tissues as kidney, spleen, and liver indicate more nonuniformities.
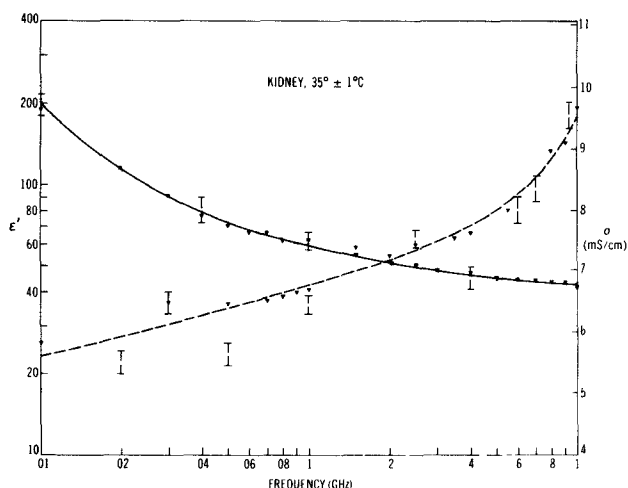
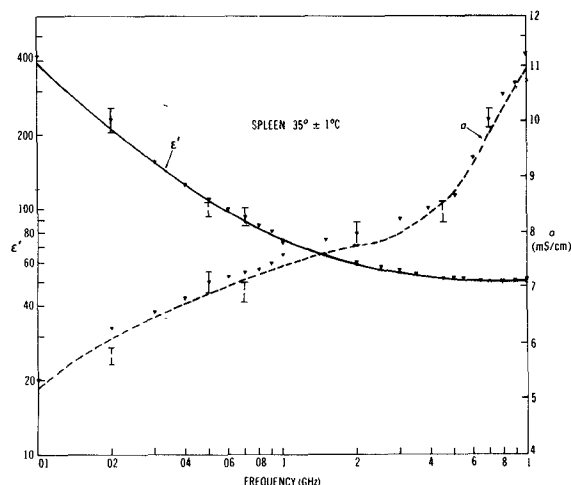Fig. 6. Dielectric constant and conductivity of kidney of a cat, measured with the 8.3-mm probe.

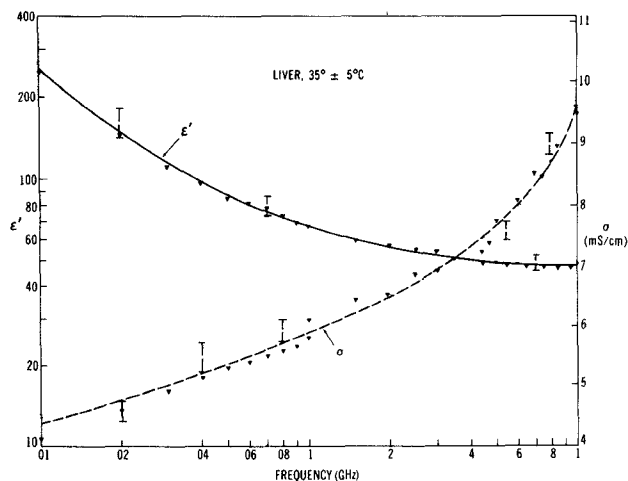Fig. 7. Dielectric constant and conductivity of spleen of a cat measured with the 8.3-mm probe.

Fig. 8. Dielectric constant and conductivity of liver of a cat measured with 8.3-mm probe.

The dielectric constant and loss factor of skeletal muscle (biceps femoris) are very close to *in vivo* data for canine muscle [2], [9]. No published data is available for smooth

muscle, whose dielectric properties at frequencies 10–100 MHz are somewhat different from those of skeletal muscle.

The dielectric constant of cat kidney was found to be lower at frequencies 0.1–1 GHz than that of a dog as reported previously [2], while the conductivity values are relatively close. It is not clear whether this difference in dielectric constant results from a species difference or a measurement difference.

The data obtained for liver and spleen can only be compared to *in vitro* data [3]. The differences in the properties obtained *in vivo* in this study and the properties obtained previously *in vitro* are of the same order of magnitude as the differences observed by Burdette *et al.* [2] for muscle tissue. They may be attributed to the different water content of spleen and liver for various species.

IV. CONCLUSIONS

The permittivity of distilled water, saline solutions, and methanol were measured using an open-ended coaxial line and computer-controlled network analyzer systems at frequencies 0.01–1 GHz. Results were found to be within the uncertainty limits estimated from measurement system errors when compared with the reference data. Measurements *in vivo* were performed on two types of muscle, and on the liver, kidney, and spleen of a cat. Highly reproducible and accurate data were obtained.

The open-ended coaxial line sensor built from a 8.3-mm Teflon line provides information about the permittivity of the sample in a zone with approximately the surface area of the end of the probe and with thickness of approximately 5 mm for high water content tissues.

The measurement method has proven to be very fast (time to obtain data at one frequency without the calibration is about 10 s; calibration takes about 10 min and should be done after a long warm up of 2 h or more), convenient, and accurate.

ACKNOWLEDGMENT

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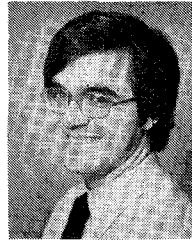
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M. A. Stuchly (M'71–SM'76), for a photograph and biography please see page 86 of this issue.

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Waveguide Technique for the Calibration of Miniature Implantable Electric-Field Probes for Use in Microwave-Bioeffects Studies

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Abstract —A new *S*-band waveguide technique has been developed for the calibration of miniature probes used in determining electric fields in biological tissues at 2.45 GHz. A section of waveguide is filled with tissue-equivalent liquid separated from the air-filled waveguide by a very thin (0.25-mm) planar dielectric spacer. The probe response is measured as a function of position on each side of the spacer and extrapolated to the interface. The ratio of probe response in air to that in test liquid is then determined assuming continuity of tangential *E*-field across the spacer. In the water–glycerol solution modelling wet tissue, the probes are 3.0 ± 0.6 times more sensitive to E^2 than in air. A wide variety of both wet and dry tissues may be simulated using liquids of different dielectric properties—a check on the properties is provided by comparing the measured depth of penetration of the wave in the liquid with the calculated value. Problems using the probes in biological tissues are also discussed.

I. INTRODUCTION

FOR MICROWAVE bioeffects research it is desirable to know the *local* rate of energy deposition at the site of action of the microwave radiation. The rate of energy deposition is usually expressed as a specific absorption rate