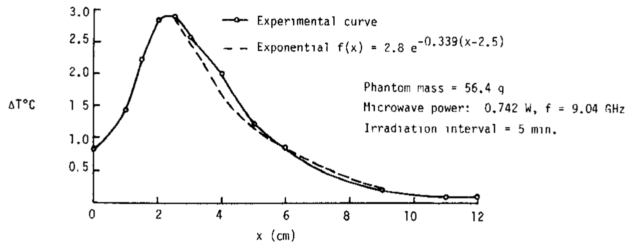


Fig. 7. Heating distribution along the central axis of sample holder.

$U$-band), and $c$ is the specific heat of the sample (1 cal/g °C for water). Hence, neglecting heat loss by the fluid, the expected rate of heating is 0.0325 °C/s for $E$-band and 0.0204 °C/s for $U$-band. Initial values of the heating rate in Fig. 6 are seen to decrease with increasing flow rate as heat transfer from the sample holder is increased.

To determine the distribution of microwave energy deposition within the sample holders, we have measured local heating within a scaled sample holder at $X$-band. Convective heat transfer was eliminated by using phantom compositions with appropriately scaled dielectric constant and conductivity [4] in place of water. Fig. 7 shows the temperature profile along the central axis $x$ of the scaled sample holder. The decrease in energy deposition with axial distance is approximately exponential as theoretically expected.

## IV. Conclusions

A broad-band temperature-controlled system for the study of cellular bioeffects of microwaves has been developed. The system coupling efficiency is greater than 99 percent over the frequencies used, and it is simple to operate.

Opposing thermocouples allow the measurement of temperature differences as small as $\pm 0.01$°C. Compensatory heating of the control sample holders suggests that temperature differences over the course of 30 min of exposure can be kept within $\pm 0.02$°C by use of appropriate feedback control circuits. This degree of accuracy is desirable in the study of microwave bioeffects.

To date, this system has been successfully employed in investigation of possible effects of microwaves on induction of Lambda phage in lysogenic strains of *E. coli* (both wild type [5] and a temperature-sensitive mutant) and back mutation of *Salmonella Typhimurium* cells (strain types TA1535 and TA1538) possessing $\text{Hi}\bar{\text{s}}$ mutations.

Our system has been designed for use at $U$- and $E$-bands, but it can be readily scaled for use at other waveguide bands.

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# Microwave Diffraction Tomography for Biomedical Applications

J. CH. BOLOMEY, A. IZADNEGAHDAR, L. JOFRE, CH. PICHOT, G. PERONNET, and M. SOLAIMANI

*Abstract*—**This short paper points out the realizability of a method suitable for quasi real-time coherent microwave tomography for biomedical applications.**

## I. Introduction

Recent experiments have shown the potential interest of active microwave imaging in biomedical applications [1]–[3]. The success of this new technique is largely dependent on the practical possibility of realizing rapid tomographic reconstruction and at low cost. This paper describes a promising method providing a good compromise between rapidity and cost. Its salient points consist in the principles of both the equipments and algorithms used in view of tomographic reconstruction. Firstly, the electromagnetic field probing is very fast and inexpensive compared to other conventional solutions using electronic or mechanical scanning. Secondly, instead of assuming straight line propagation, as is usually done in computerized tomography [3], diffraction effects are taken into account.

## II. Tomographic Algorithm

The tomographic process is achieved by simulating a focusing system characterized by small field depth and a variable focal length as depicted in Fig. 1. The focusing system transforms the divergent wavefront scattered by the organ under test into a convergent one from which an image can be derived. For a given focal length, the image corresponds to a thin organ slice. The cross section of the organ can be deduced from the different slices obtained by changing the focal length. The slice images are calculated from the field distribution measured in the aperture plane of the simulated focusing system. The reconstruction depends on the illumination and on the observation: the field at the organ location must be as uniform as possible and the length of the observation domain must be $2D+d$ where $d$ is the length of the organ slice in the direction parallel to the observation line and $D$ is the distance between the observation line and the slice.

The organ is seen by its equivalent currents $\boldsymbol{J}$ given by

$$\boldsymbol{J}(x,y)=\left[K^2(x,y)-K_m^2\right]\boldsymbol{E}_t(x,y) \tag{1}$$

where $\boldsymbol{E}_t(x,t)$ and $K(x,y)$ are, respectively, the total field and the wavenumber inside the organ, and $K_m$ is the wavenumber of the homogeneous surrounding medium.

The equivalent currents $J$ are responsible for the scattered field $\boldsymbol{E}_S$. For instance, in the simple case in cylindrical objects illuminated by a plane wave, the electric field of which is parallel to the cylinder $Z$-axis, the scattered field is related to the equivalent current by

$$\boldsymbol{E}_S(x,y)=\int_S \boldsymbol{J}(x,y)\,H_0^{(2)}\left(K_m\sqrt{(x-x')^2+(y-y')^2}\right)dx'\,dy' \tag{2}$$

where $H_0^{(2)}$ is the Hankel function of order zero and of the second

Manuscript received February 17, 1982; revised May 13, 1982. This work was supported in part by DGRST, under Contract TLB-81-M0909.

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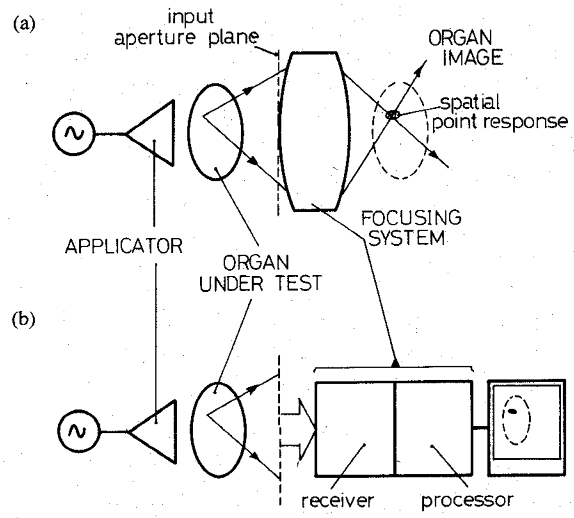


Fig. 1. (a) Principle of the tomographic system and (b) its practical realization.

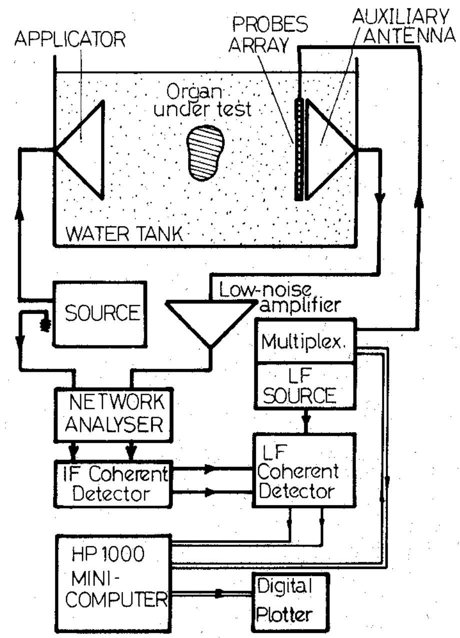


Fig. 2. Principle of field probing by means of the modulated scattered technique.

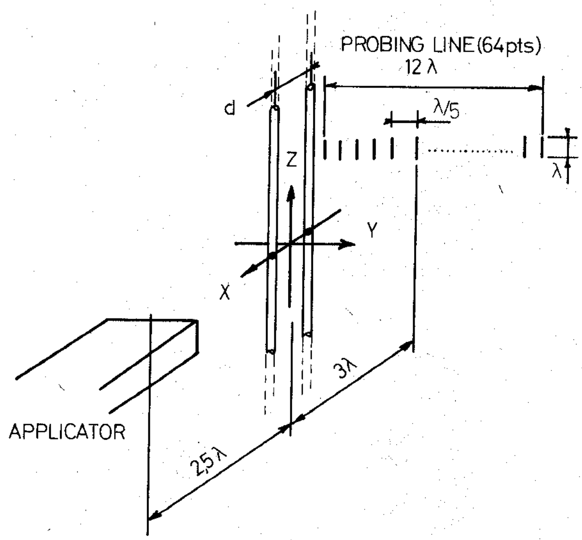


Fig. 3. Experimental setup for resolution estimation at 3 GHz ($\lambda = 11.6$ mm).

kind. For three-dimensional problems, the relation is more complicated [10]. But, in any case, such relations can be used to reconstruct $\boldsymbol{J}$ from $\boldsymbol{E}_S$. More precisely, the algorithms which have been used are based on the angular spectrum method [4]. They are described in a recently published paper [11].

The image (the reconstructed current) appears as the convolution product between the point-spread function of the focusing system and the equivalent current distribution induced in the organ [11]. The extent of the point-spread function is primarily related to the dimensions of the aperture plane compared to the wavelength. Its spread in the direction normal to aperture plane is larger than in the direction parallel to it. This effect can be compensated by imaging the organ by means of different illuminations (two or more) and by combining them [5], [9], [11]. The net resulting point response has then approximately the same extent and, hence, the same resolution in all directions.

## III. Field Probing Method

The previously described process requires the knowledge of the electric field. Moving the probe is time consuming; probe arrays are complex and expensive because of the high number of elements to be considered and because of the necessary connection between them and a receiver through a microwave low-level multiplexer. The modulated scattering technique [6], [7] provides the rapidity of the probe arrays without the need for multiplexing. The array is constituted of probes loaded by nonlinear elements. Modulating the load of a given probe results in a diffracted signal. This signal, proportional to the field to be measured, can be collected by an auxiliary antenna. Modulating one after one the probes allow us to record the field, in phase and amplitude, on the array surface (Fig. 2). This modulation provides a means for distinguishing the useful signal from the clutter and makes easier its detection. Such a measurement technique requires a high-sensitivity receiver.

## IV. Applications

The method has been applied to the problem of Fig. 3 consisting of deducing the cross section of an inhomogeneous cylinder from field measurement on a finite segment. For resolution estimation, the case of two parallel plexiglass rods ($\epsilon_r = 2.4$, $\varnothing = 6$ mm) has been considered. The rods are immersed in water ($\epsilon_r = 74$, $\sigma = 2.39$ s/m) in which the wavelength is $\lambda = 11.6$ mm at 3 GHz. The applicator is fed by a 10-mW generator with a standard $S$-band waveguide matched to water with a dielectric slab. The probe array is a linear array of 64 dipoles loaded by diodes. The space between adjacent dipoles is $\lambda/5$. The extent of the measurement line is 12.6 $\lambda$. The receiver chain [8] provides the real and imaginary parts of the field distribution. It consists of 1) a low-noise Société d'Etude du Radant amplifier, 2) a Hewlett–Packard network analyzer, 3) two coherent detection stages—the first one at the IF of the network analyzer, the second one at the modulation frequency (1 kHz). Data acquisition and processing are performed by a HP 1000 minicomputer.

## V. Results

Fig. 4 gives two examples of infinite rods reconstruction using two orthogonal measurements in the cross-section plane. Orthogonal illumination is simulated by rotating the rods around the $Z$-axis.

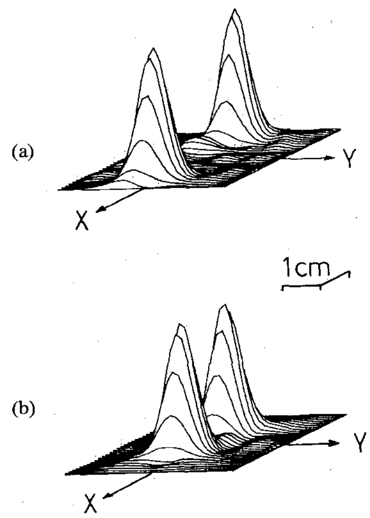


Fig. 4. Rods equivalent currents (cross section) as function of separation. (a) $d = 2.4\,\lambda$ (27.8 mm); (b) $d = 1.2\,\lambda$ (13.9 mm).

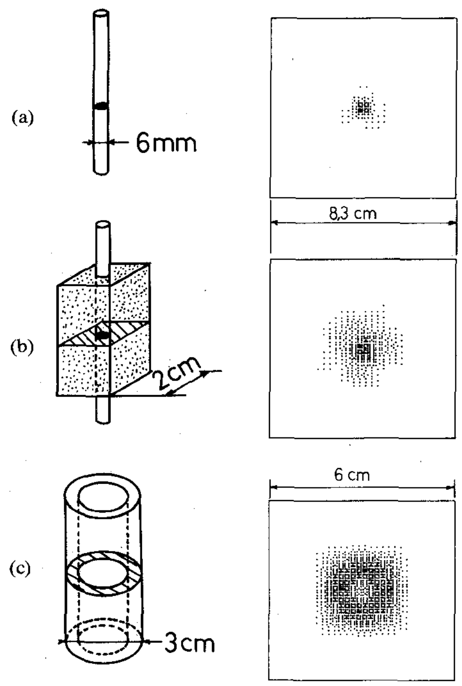


Fig. 5. Equivalent currents of dielectric bodies (cross section) in the case of four orthogonal illuminations. (a) Rod. (b) Rod surrounded by sponge. (c) Water filled plexiglas cylinder.

Fig. 5 shows three examples of various dielectric cylinders reconstruction using four orthogonal measurements.

As an indication of measurement speed, recording the field on a measurement line requires approximately one half of a second; the time required for deducing the graphs of Fig. 4 is 50 s on a HP 1000 with use of FTT algorithms but without any special processor.

## VI. Conclusion

This short paper shows the possibility of fast coherent tomography of biological media. In contradistinction to conventional computerized tomography, the algorithms take into account diffraction effects. The organs are viewed from their equivalent currents which depend both from the permittivity and from the illuminating field. Numerical simulations are in progress in order to estimate the usefulness of such a representation and the contrasts to be expected. The modulation scattering technique allows amplitude-phase recording at the rate of a few milliseconds per point. The sensitivity of the receiver is sufficient for atraumatic illumination levels, always less than 10 mW/cm$^2$. Multiple illuminations provide a spatial resolution of $\lambda/2$ where $\lambda$ is the wavelength in water ($\lambda = 1.16$ cm at 3 GHz).

## Acknowledgment

The authors wish to thank the Laboratoire de Thermologie Biomedicale (Strasbourg) and Société d'Etude du Radant (Orsay) for their collaboration in this work.

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# Numerical and Experimental Results for Near-Field Electromagnetic Absorption in Man

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***Abstract*— Experimental results are presented for the whole-body-average energy absorption and the internal *E*-fields in man exposed to leakage-type near fields. An empirical relationship, previously presented [1]**

Manuscript received January 6, 1982; revised May 18, 1982. This work was supported in part by the National Institute of Environmental Health Sciences, Research Triangle Park, NC, under Grant ES02304.

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