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# A Nonperturbing Temperature Sensor for Measurements in Electromagnetic Fields*

T. C. Rozzell†, C. C. Johnson‡, C. H. Durney‡,
J. L. Lords‡ and R. G. Olsen‡

ABSTRACT

*An electro-optical temperature measuring device which neither perturbs electromagnetic fields nor causes hot spots has been developed for use in monitoring temperature in biological systems during microwave irradiation.*

## Introduction

Usual methods of measuring internal temperature in biological media employ metallic sensors (such as thermistors, thermometers, etc.) or utilize calorimetric techniques [1]. The presence of metallic structures concentrates electromagnetic (EM) fields and produces unwanted localized hot spots [2]. In order to properly assess biological responses and effectively determine electromagnetic power deposition, it is desirable to be able to measure temperature in the exposed medium during the course of exposure to the EM field. An alternative technique in which the temperature sensor is inserted into a fixed glass tube, after exposure, suffers from the fact that the measurements are discontinuous, the results are subject to error in extrapolating the measured temperature back to the temperature just post-irradiation, and, in dynamic systems, small changes may go undetected.

One attempt to overcome the aforementioned problems has recently been reported [3] in which the microwave integrated circuit strip line with lead wires 5$\mu$ wide and 2000Å thick spaced 5$\mu$ apart was used to continuously monitor brain temperature in experimental animals.

This paper discusses a temperature sensing probe which is constructed completely of dielectric materials, including liquid crystals, optical fibers, polyvinyl chloride sheathing, and a nested glass container. The liquid crystal/optic fiber (LCOF) probe causes minimal perturbation of electromagnetic fields in the microwave frequency range and does not contribute heat to surrounding tissue.

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* Manuscript received June 18, 1974.
† Office of Naval Research, Arlington, Viginia 22217.
‡ University of Utah, Salt Lake City, Utah.
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## Description of the Probe

Some design details and preliminary results have already been reported [4]. Basically, the probe consists of two bundles of polymethyl methacrylate optic fibers enclosed in a thin wall polyvinyl chloride sheath. These optic fiber bundles transmit light to and receive light from the sensing tip. At the sensor tip of the optic fiber bundles, a thin liquid crystal film or bulk liquid crystal is held in place sandwiched between two glass or teflon cups. The liquid crystal film is placed 0.5 - 1.0 mm from the tip of the optic fibers. The optic fiber bundles consist of 5 - 7 fibers of 0.125 mm diameter each. About one-half of the fibers are used to transmit light from a red (0.685 micron) gallium-arsenide-phosphide light emitting diode (LED) to illuminate the liquid crystal film. The remaining fibers carry the reflected light out to a phototransistor. The phototransistor generates a voltage proportional to the reflected light intensity, which value in turn is affected by the temperature change taking place in the liquid crystal film. The temperature or the output voltage is read directly on a meter.

A block diagram of the electro-optical illumination and read-out unit is shown in Figure 1. A clock generator sends pulses to the sample pulse generator for

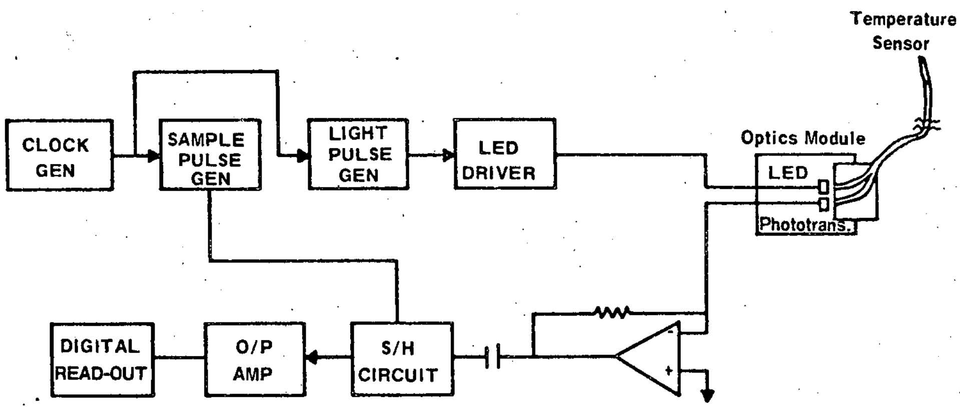


Figure 1 Block diagram of the electro-optic illumination and read-out electronics unit. The temperature probe is connected to the optics module via a snap-in/snap-out connector with position set screw.

the sample-hold circuit and to the LED pulse generator which activates the LED driver, The LED is pulsed at a rate of 100 pulses per second with pulse widths of about 10 microseconds. Current pulses of about 1 - 3 amperes are sent to the LED. The reflected light pulses are amplified and sent to the sample-hold circuit to read pulse height. This pulse height is delivered to the output meter.

## Thermographic Camera Tests

The extent of temperature perturbation by the probe in a 7 cm diameter brain sphere was investigated by placing a probe mock-up consisting of the fibers and sheath along the center line of the sphere as shown in Figure 2. The sphere is split in two halves to allow quick access to the sphere interior for presentation to the thermographic camera, and the probe was held between these two halves. The thermographic camera presents a C-scan showing a two-

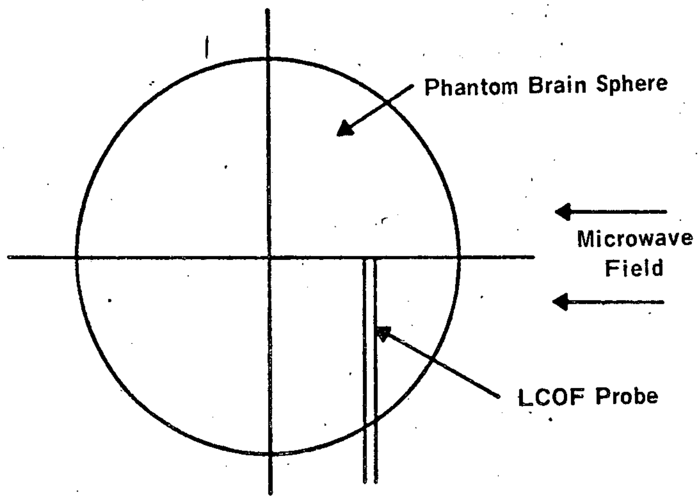


Figure 2 The LCOF probe mock-up is placed in a phantom brain sphere as shown to test for temperature perturbation effects as sensed by the thermographic camera.

dimensional temperature map, and a B-scan temperature versus distance along the sphere axis.

Tests were conducted at 918 MHz and 2450 MHz. A reference sphere with no probe was exposed and measured as a standard. Then the LCOF probe was inserted as shown in Figure 2 and tested. For comparison, a very small thermistor bead was fabricated and tested with idealized normal saline fluid columns for electrical leads. The test results are shown in Figures 3, 4 and 5. Figure 3 shows a C-scan on the left and B-scan on the right for the reference

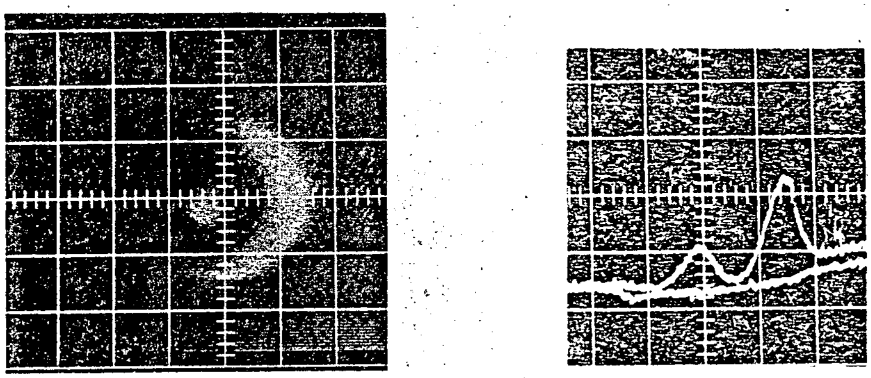

Figure 3 Reference sphere of 2450 MHz.

sphere at 2450 MHz. The two lines on the B-scan represent pre- and post-irradiation temperatures scanned through the center of the sphere. Figure 4 shows the results at 2450 MHz with the LCOF probe in place, indicating only about

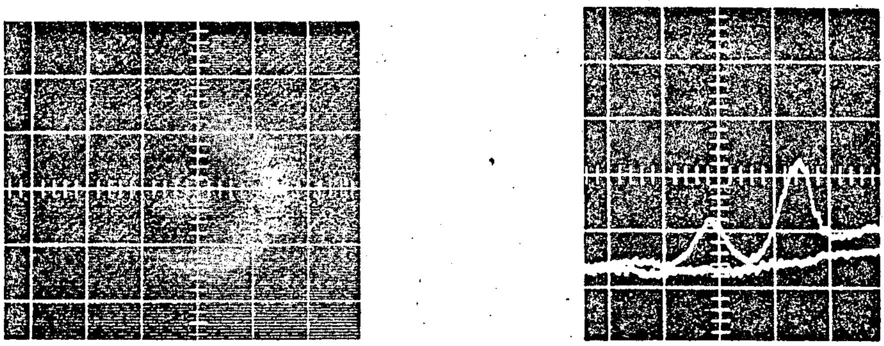

Figure 5 Sphere with thermistor bead and saline columns for leads at 2450 MHz. An 18% is measured.

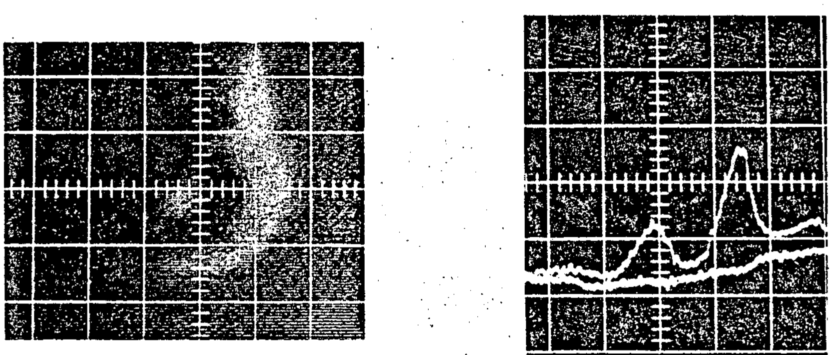

Figure 5 Sphere with thermistor bead and saline columns for leads at 2450 MHz. An 18% error in peak temperature is measured.

a 2% error in peak temperature and very similar shaped curves. Figure 5 shows the results with the thermistor bead probe at 2450 MHz. An 18% error in peak temperature is observed and the shape of the curve indicates localized artifactual heating has taken place. The heated mass outside the sphere is caused by the saline leads protruding from the sphere which are heated by the irradiation. These results show that even very short high conductance leads can cause significant temperature perturbation in tissue.

## Design Testing

The light reflection characteristics of the liquid crystal film as a function of temperature were found to be critically dependent upon distance of the film from the polished end of the optic fibers. The apparatus shown in Figure 6

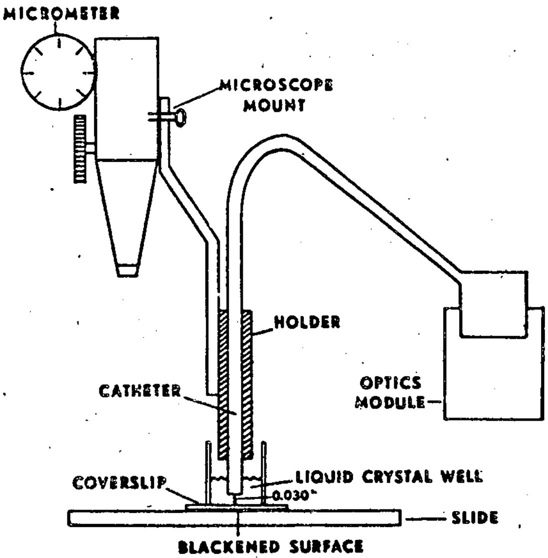


Figure 6 Apparatus used to determine the relationship between liquid crystal thickness and reflectance (R).

was used to determine the relationship between the bulk liquid crystal thickness and reflectance (R). In this setup, a micrometer adjustment allowed for a very accurate measure of the distance as a rigidly held fiber bundle was moved in relation to a small container of liquid crystal with an opaque (black) bottom. As shown in Figure 7, the minimum distance to obtain maximum reflectance is about 25 - 30 mils. This was further verified by determining the relationship between reflectance (R) and temperature (in the range of 30 and 45°C) at various distances. Clearly, the family of curves in Figure 8 show that below 30 mils the slopes are too small to provide very much sensitivity. Since the read-out is proportional to reflectance, it is desired to achieve the maximum change in R per unit of temperature change.

This sensor has been designed to operate in the temperature range of 30 to 45°C since most ambient biological temperatures fall within this range. The liquid crystal mixture, which is a mixture of three purified, dehydrated cholesteryl compounds, is the determining factor in establishing the temperature range.

Initial tests have shown that when the temperature reaches approximately 50 - 55°C, the liquid crystals "clear" and deflectance drops sharply as shown in Figure 9. Between 35 and 40°C, however, the relationship between temperature and output of the phototransistor is such that temperature can be determined within 0.1 to 0.5°C. This degree of precision, if it can be maintained, should be sufficient for the majority of the envisioned applications. In the final design,

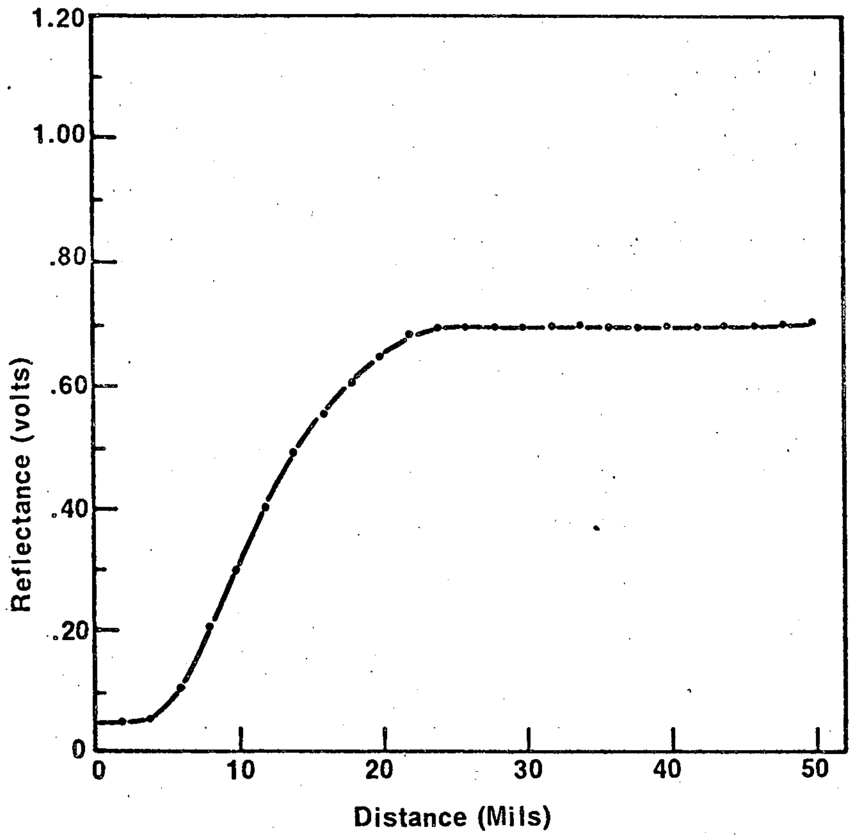


Figure 7 The relationship between the liquid crystal-optic fiber separation and reflectance (R).

it is hoped that the probe can be operated with a dependable accuracy of 0.1°C.

## Applications

The tip of the LCOF probe is approximately 1.75 mm in diameter. The length of the optic fibers can be as much as a meter or more. In its present form the fragility of the tip, with its thin glass covering, is such that the probe cannot be used to "drill" its own hole through solid tissue. Therefore, when used in muscle or other such firm tissue, it is necessary to clear a path for the probe in some manner. A probe of this size can be used with ease to measure rectal temperature of animals down to the size of large mice. Work continues in an effort to reduce the diameter of the tip, but 1 mm will probably represent the minimum size attainable.

This probe can easily be slipped into the chambers of excised animal hearts in cardiac studies. It can be inserted easily into brain tissue or placed surgically in a number of internal locations where it is desired to follow the course of microwave induced temperature changes. It has been successfully placed in the eye of a rabbit to monitor temperature during radiation in cataractogenesis studies.

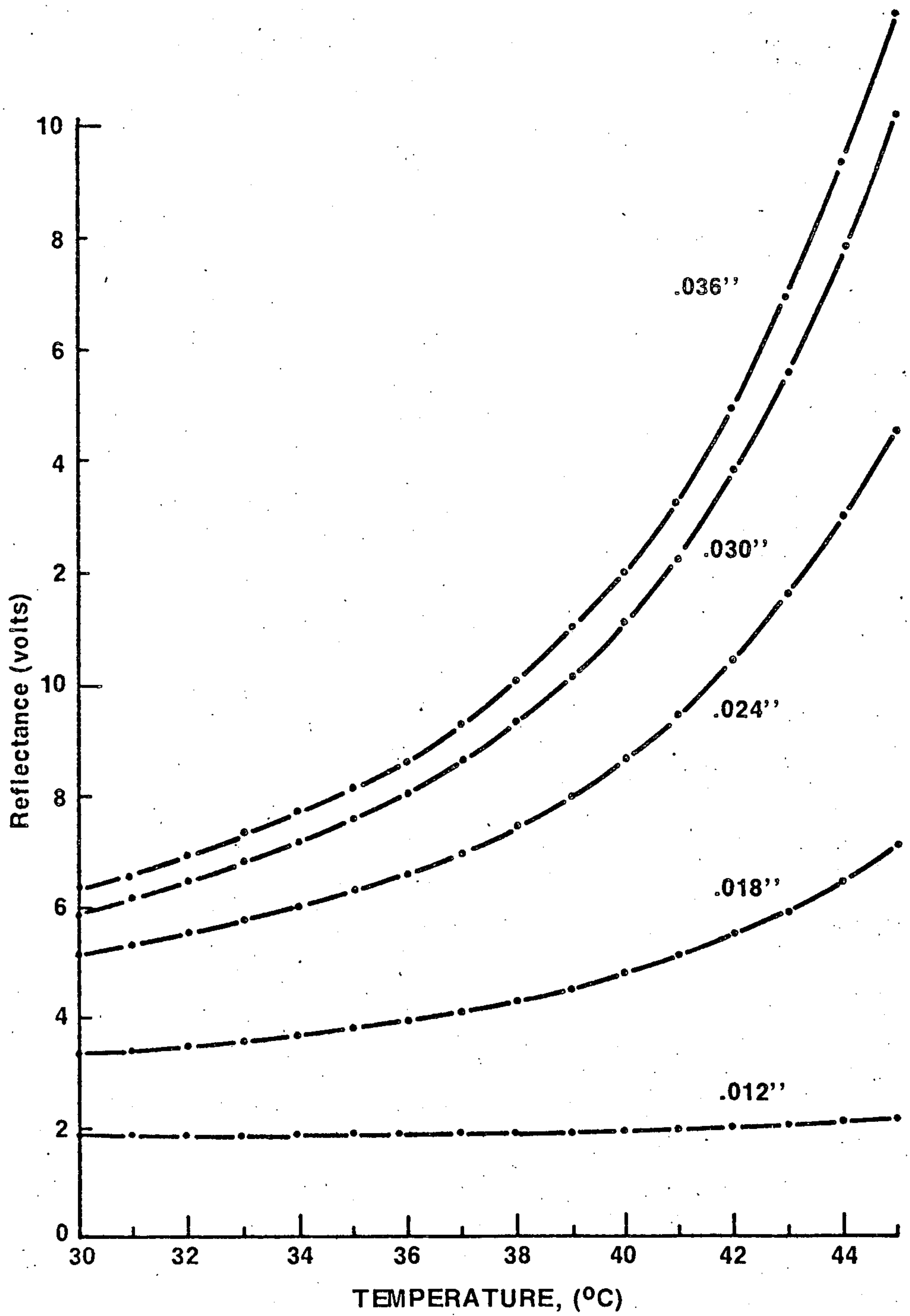


Figure 8 The relationship between reflectance (R) and temperature at various thicknesses of liquid crystal.

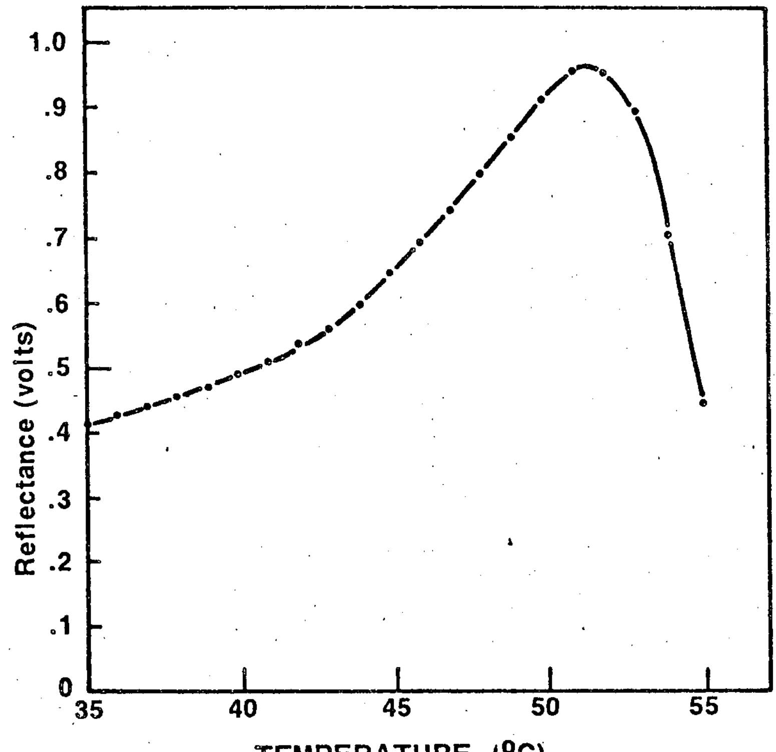


Figure 9 Reflectance as a function of temperature illustrating the drop in reflectance as the liquid crystals begin to clear at approximately 52°C.

The optic fibers terminate at a small connector which locks into an "optics module" which contains the LED, LED driver, phototransistor and preamplifier. This arrangement may allow for the probe to be chronically implanted in certain animals such as dogs, rabbits, monkeys, etc. The necessity for occasional calibration, however, may preclude this.

This type of temperature probe will also be useful in non-ionizing radiation dosimetry. It can be used to determine absorbed electromagnetic power density by measuring the time rate of change of temperature as a function of a step input of energy to the tissue of organ according to the relation

$$\overline{P} = 4.186 f c \Delta T / t$$

where $\overline{P}$ is the absorbed power density, f the tissue mass density, c the tissue specific heat, $\Delta T$ the temperature rise and t the time of irradiation.

**Acknowledgement**

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## References

1 Hunt, E., "Dosimetry for Whole Animal Microwave Irradiation," Health Physics (In Press).
2 Johnson, C. C. and Guy, A. W., "Non-ionizing Electromagnetic Wave Effects in Biological Materials and Systems," Proc. of the IEEE, 60(6), p. 692 (1972).
3 Larsen, L. E., Moore, R. A. and Acevedo, J., "An RF Decoupled Electrode for Measurement of Brain Temperature During Microwave Exposure," Proc. 1973 IEEE-G-MIT International Microwave Symposium, p. 262, June 1973.
4 Olsen, R., Johnson, C., Rozzell, T., Durney, C. and Lords, J., "A Microwave Transparent Temperature Sensor," Proc. 1973 International IEEE/GAP Symposium and USNC/URSI Meeting, Boulder, Colorado, August 1973.