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Radiometric sensing of biological layered media

Fernando Bardati and Domenico Solimini

Dipartimento di Ingegneria Elettronica, Facoltà di Ingegneria, II Università di Roma (Tor Vergata)

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The retrieval of temperature profiles in layered living tissues from microwave radiometric measurements is considered. The brightness temperature of a layered biological structure is expressed as a function of the thermal distribution in the tissues for different microwave frequencies, angles of observation, and polarization of the emitted waves. A "coherent" approach to the radiative transfer problem is followed to obtain the weighting functions in closed form for use in the inverse problem. Finally, the extraction of the thermal profiles from brightness data sets by use of Kalman filtering is discussed, with particular reference to the detectability of deep thermal anomalies.

INTRODUCTION

In recent years, microwave thermal emission from layered media has been extensively considered. The applications that have been envisaged have been largely concerned with the remote sensing of the earth's surface, where layering is caused by ice or snow or by the morphology of the soil itself [*England*, 1974, 1975; *Tsang et al.*, 1975; *Tsang and Kong*, 1975, 1976, 1980; *Njoku and Kong*, 1977; *Djermakoye and Kong*, 1979; *Chuang et al.*, 1980; *Fung and Chen*, 1981]. The use of microwave radiometry for determining the temperature of biological tissues has also attracted much attention because of the importance of such a noninvasive technique in diagnosis and therapy. Experiments have been carried out at different frequencies to give evidence of thermal anomalies for cancer detection [*Edrich*, 1979; *Myers et al.*, 1979; *Nguyen et al.*, 1980; *Mamouni et al.*, 1981]. A numerical analysis has also been conducted to ascertain the feasibility of the temperature determination in a layered model of the biological structure [*Edenhofer*, 1981]. The approach that has been followed in this work is based on the radiative transfer theory and on suitable assumptions both in the models and in the temperature distribution within the layers.

Indeed, several aspects of the problem deserve some discussion. The thermal state of the living tissues is to be determined from measurements of the brightness temperature of the biological structure. Microwave emission from various regions of the human body is essentially affected by the local layered character of the tissues. The surface layers of man in general consist of skin-fat-muscle or skin-fat-muscle-bone-muscle arrangements [*Barber et al.*, 1979]. Typical thicknesses of the layers are in the millimeter and centimeter range, so that interference effects leading to partially coherent emission processes significantly affect the apparent brightness of the structure at microwave frequencies [*Blinn et al.*, 1972; *England and Johnson*, 1977; *Carver*, 1977]. In both diagnostic and therapeutic applications, rather detailed temperature distributions must be determined. Indeed, pathological processes may concern small regions of the tissues, and, on the other hand, in hyperthermia, for instance, hot spots of small extent must be monitored. This requirement puts some constraints on the spatial resolution which is to be attained by the temperature retrieval procedure, still maintaining the temperature accuracy required by the actual application.

In this paper the emission of electromagnetic power from a plane parallel three-layer (skin-fat-muscle) model of human living tissues is considered. The brightness temperature has been determined for a wide range of frequencies in the microwave band, where a coherent approach [*Stogryn*, 1970] to the thermal emission problem is needed. The relative contribution to the brightness from each tissue is also given for the various frequencies. An expression which relates the emitted radiation intensity to the temperature profile within the tissues has been worked out with the purpose of determining the coherent radiative transfer kernels at the different fre-

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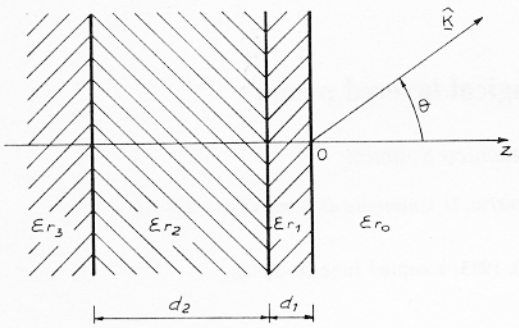

Fig. 1. Geometrical configuration of the biological model; $-d_1 < z < 0$, skin; $-(d_1 + d_2) < z < -d_1$, fat; and $z < -(d_1 + d_2)$, muscle.

quencies. The inverse problem of retrieving the temperature profile from a set of radiometric data by solving the resulting Fredholm integral equation of the first kind has been considered. Use has been made of Kalman filtering, which besides its good retrieving characteristics, exhibits the additional advantage of recursiveness, which is useful whenever the temporal evolution of the thermal profile is of interest, as is the case, for instance, in hyperthermic treatments.

EMISSION FROM LAYERED LIVING TISSUES

To investigate some significant characteristics of microwave emission from living structures, a three-layer plane parallel model has been adopted. This model in several instances is adequate to represent the basic electromagnetic behavior of actual biological structures and, in addition, leads to fairly simple analytical and numerical formulations [*Guy*, 1971]. In recent papers, two different approaches have been followed to investigate microwave emission from inhomogeneous media, with particular reference to the emission from the earth surface [*Schmugge and Choudhury*, 1981]. In the "incoherent" approach the intensity of the emitted radiation is obtained from a solution of the radiative transfer equation, while the "coherent" approach makes use of solutions of Maxwell's equations with the pertinent boundary conditions. Thermal emission is intrinsically an incoherent process. However, reflections which may be caused by nonrandom inhomogeneities in a stratified emitting medium produce interference effects, so that the emerging radiation exhibits partial coherence. Choice of either model relies on the characteristic thickness of the inhomogeneities [*Carver*, 1977]. The model adopted for the biological structure is a skin-fat-muscle arrangement (Figure 1), separated by fairly smooth boundaries, with negligible embedded inhomogeneities. Since typical thicknesses of the layers are in the millimeter or, at most, in the centimeter range, the effect of multiple reflections from the boundaries of the different kinds of tissues can be important. For this reason, a coherent approach must be followed to evaluate microwave emission from the assumed model.

According to *Stogryn*'s [1970] fundamental approach, a fluctuating current **J** can be defined within the lossy tissues to represent the local source of radiation. The electromagnetic field radiated into the lossless half space $z > 0$ (Figure 1) can be expressed as a function of **J** through the dyadic Green's function of the pertinent boundary value problem. Being an incoherent emission process, the source is characterized by the second-order moment of the current, which, in turn, depends on the local thermodynamic temperature and on the imaginary part of the dielectric constant of the tissues [*Landau and Lifshitz*, 1960]. The radiation intensity and, in turn, the brightness temperature T_B, is obtained from the second-order moment of the field, according to the following equation:

$$T_{Bp}(\hat{\mathbf{k}}, \omega) = 4 \cos\theta \left(\frac{\omega}{c}\right)^3 \varepsilon_{r0}^{1/2} \sum_{j,l,m} p_j p_l \int_{-\infty}^{0} A_{jm}(z, k_x, k_y) \cdot A_{lm}^*(z, k_x, k_y) \operatorname{Im}[\varepsilon_r(z, \omega)] T(z)\, dz \quad (1)$$

In (1) the Rayleigh-Jeans approximation has been used, and symbols have the following meaning: T_{Bp} is the brightness temperature corresponding to polarization $\mathbf{p} = \sum_i p_i \hat{\mathbf{x}}_i$, where $\hat{\mathbf{x}}_i$ are unit vectors of coordinate axes; $\hat{\mathbf{k}}$ is a unit vector in the direction of observation, forming an angle θ with the normal to the layers, while $\mathbf{k}$ is the corresponding wave vector; ω is the angular frequency; ε_{r0} and ε_r are the relative dielectric constants of the medium for $z > 0$ and of the tissues, respectively (magnetic permeability is μ_0 everywhere); c is the speed of light in vacuum; A_{jl} is the pertinent spectral component of the dyadic Green's function which can be determined by solving an inhomogeneous Helmholtz equation with the pertinent boundary conditions at the interfaces between the different layers of the structure and by taking the translational symmetry into account. The three-layer plane model of the biological structure permits the brightness temperature to be easily obtained in closed form as a function of the thermodynamic temperature of the tissues, thus allowing the introduction

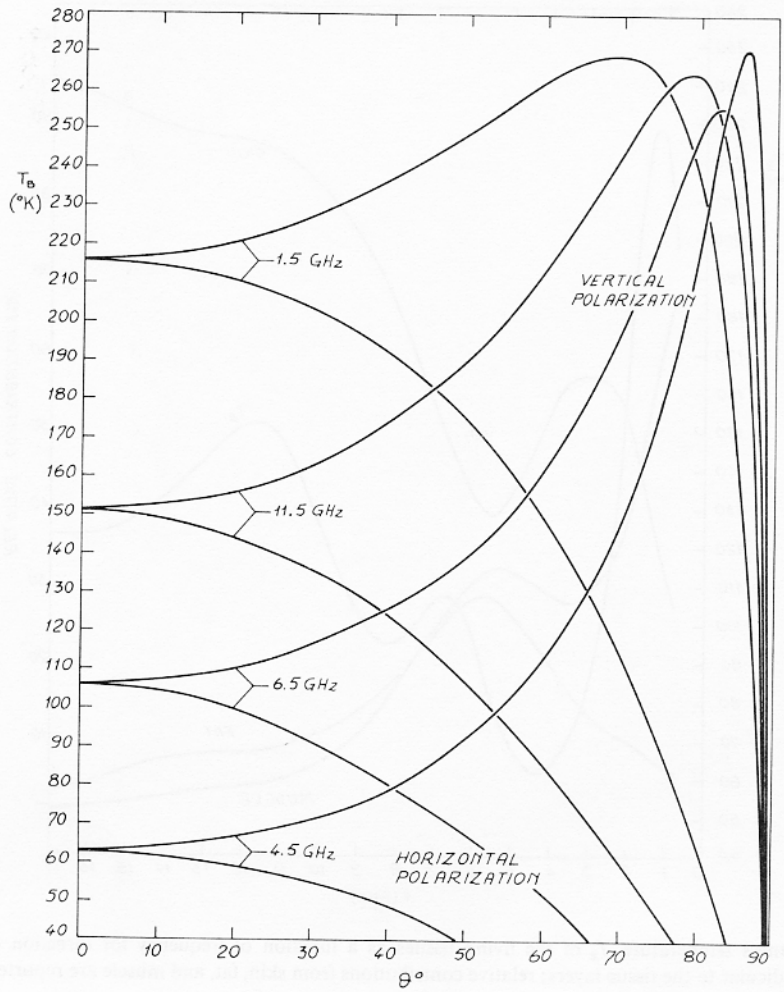

Fig. 2. Brightness temperature T_B of the living tissues as a function of observation angle θ for horizontal and vertical polarization at four frequencies.

of the weighting functions (see next section). In the case of more complicated structures, such as layers with continuously varying permittivities, explicit closed-form solutions are difficult to obtain, and other approaches, based essentially on the discretization of the problem, can be followed [*Tsang et al.*, 1975; *Wilheit*, 1978].

The brightness temperature of a three-layer (skin-fat-muscle) biological structure has been computed for various frequencies in the microwave band. Values for the dielectric constants of the three types of tissues have been obtained from published data [*Stuchly and Stuchly*, 1980; *Schwan and Foster*, 1980]. In the range of frequencies that has been considered (0.975–17 GHz), both the real and the imaginary parts of the dielectric constants of the three types of tissues undergo fairly large variations. For the skin the real part varies between 44 at the lowest frequency and 32 at the highest, for the fat between 6 and 4, and for the muscle between 51 and 33. The corresponding values of the imaginary parts are 18 and 23, 1.5 and 0.6, and 24 and 23, respectively. As far as the thicknesses of the layers are concerned, a 2-mm-thick skin and a 1-cm-thick layer of fat have been assumed, while the muscular tissue has been considered to have infinite extent. Indeed, the thicknesses of the biological layers are fairly dependent both on the particular region of the human body considered and on the morphology of the single individuals. On the one hand, the assumed widths of the layers are reference numbers used in some studies of electromagnetic interaction with models of man [*Massoudi et al.*, 1979], and on the other hand, they are consistent with anatomic volumetric data, at least

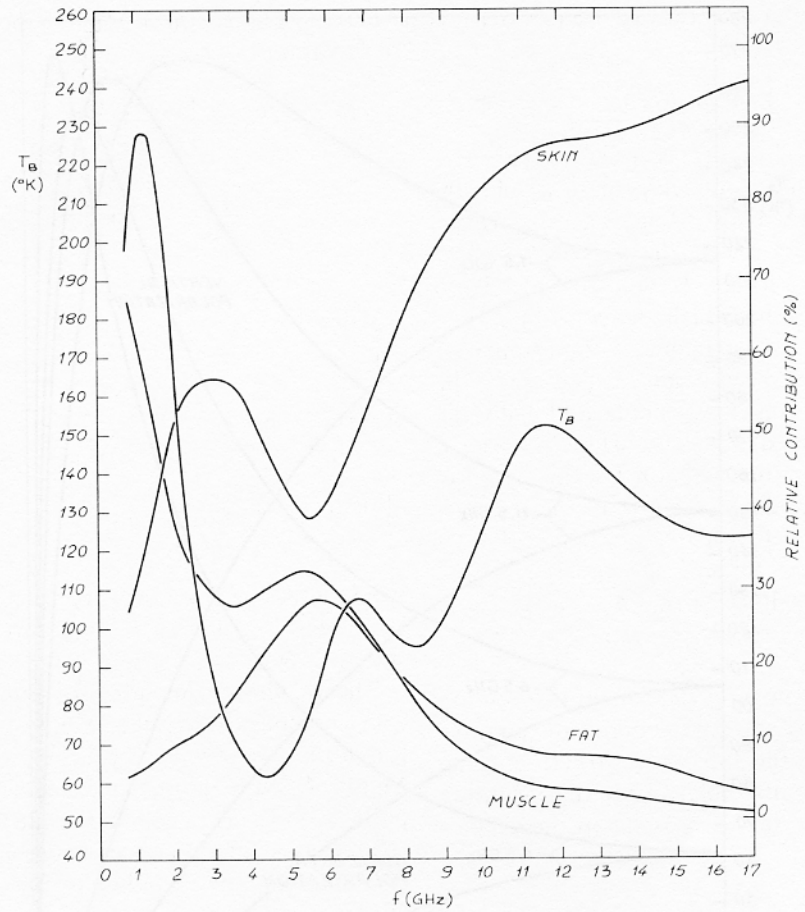

Fig. 3. Brightness temperature T_B of the living tissues as a function of frequency for direction of observation perpendicular to the tissue layers; relative contributions from skin, fat, and muscle are reported also.

for the trunk section of the human body [*Huckaba and Tam*, 1980]. Because of the highly absorbing nature of muscular tissues, the radiation originated from the deepest layers does not contribute appreciably to the brightness, so that the assumption of an infinitely thick muscle does not alter the emitting behavior of the structure. In Figure 2 the brightness temperatures for horizontal and vertical polarizations are reported as functions of the observation angle for four frequencies. The results refer to the physiological distribution of temperature in the living tissues obtained by an appropriate thermal model [*Jain*, 1980] when the ambient air temperature is 30°C. The dependence on frequency both of the pseudo-Brewster angle and of the brightness temperature at given angles is an interesting feature of the diagrams. In Figure 3 the brightness temperature T_B of the biological structure is reported as a function of frequency for direction of observation perpendicular to the layers, together with the relative contributions to T_B from skin, fat, and muscle. All curves exhibit a nonmonotonic behavior, so that radiometric operation at suitable frequencies can lead to selectively sensing the thermal state of the various layers of tissues. The above results agree with those obtained by the subroutine given by *Wilheit* [1978], provided that some minor errors in formulas for the vertical polarization case are removed from his program.

INVERSE PROBLEM

Weighting functions. As a result of the direct problem, equation (1) relates the brightness temperature of the biological structure to the thermodynamic temperature profile $T(z)$ within the tissues, and hence, by solving the inverse problem, the knowledge of $T_B(\hat{\mathbf{k}}, \omega)$ leads to the reconstruction of $T(z)$. The retrieval of the temperature profile is based on the

solution of a Fredholm integral equation of the first kind:

$$T_{Bp}(\alpha) = \int_{-L}^{0} W_p(\alpha, z)T(z)\, dz \qquad (2)$$

It is readily recognized that (2) is equation (1) rewritten in a convenient and compact form, in which variable α has the meaning either of angle of observation or of frequency and p selects the polarization. The lower limit of integration, $-L$, indicates the depth within the tissues beyond which the contribution to the brightness becomes negligible. The various parameters in (1) have been compounded with the dyadic Green's function components and with the imaginary part of the dielectric constant of the tissues to form the weighting function W. As is known, this kernel plays a fundamental role in the retrieval process, since it settles the degree of ill conditioning of the inverse problem. Smooth kernels give indication of possible instabilities in the retrievals, so that when this is the case, particular care must be taken in inverting (2).

Closed-form expressions of the weighting function have been obtained by solving the coherent radiative transfer problem for the biological structure being considered. For horizontal polarization the weighting function has the following expressions in the various tissue layers:

$$W_{hj}(z) = 4 \cos\theta \left(\frac{\omega}{c}\right)^3 \varepsilon_{r0}^{1/2}\, \varepsilon''_{rj} |a_j(z)|^2 \qquad j = 1, 2, 3 \qquad (3a)$$

where $j = 1$ refers to the skin ($-d < z < 0$), $j = 2$ to the fat ($-d_1 - d_2 < z < -d_1$), $j = 3$ to the muscle ($z < -d_1 - d_2$), and

$$a_1(z) = \frac{(y_1 + y_2)(y_2 + y_3)}{i\omega\mu_0 D} (e^{\kappa_2 d_2} + \Gamma_{12}\Gamma_{23} e^{-\kappa_2 d_2}) \cdot [e^{\kappa_1(z+d_1)} + \Gamma'_{12} e^{-\kappa_1(z+d_1)}]$$

$$a_2(z) = \frac{2y_1(y_2 + y_3)}{i\omega\mu_0 D} \cdot [e^{\kappa_2(z+d_1+d_2)} + \Gamma_{23} e^{-\kappa_2(z+d_1+d_2)}]$$

$$a_3(z) = \frac{4y_1 y_2}{i\omega\mu_0 D} e^{\kappa_3(z+d_1+d_2)}$$

$$D = (y_0 + y_1)(y_1 + y_2)(y_2 + y_3) \cdot (e^{\kappa_2 d_2} + \Gamma_{12}\Gamma_{23} e^{-\kappa_2 d_2})(e^{\kappa_1 d_1} + \Gamma_{01}\Gamma'_{12} e^{-\kappa_1 d_1})$$

$$k_j = \frac{\omega}{c} (\varepsilon_{r0} \sin^2\theta - \varepsilon_{rj})^{1/2} \qquad j = 1, 2, 3$$

$$y_j = \frac{\kappa_j}{i\omega\mu_0} \qquad j = 0, 1, 2, 3$$

where now subscript 0 refers to the medium in which observation takes place;

$$\Gamma_{12} = \frac{y_1 - y_2}{y_1 + y_2} \qquad \Gamma_{23} = \frac{y_2 - y_3}{y_2 + y_3}$$

$$\Gamma'_{12} = \frac{\Gamma_{12} + \Gamma_{23} e^{-2\kappa_2 d_2}}{1 + \Gamma_{12}\Gamma_{23} e^{-2\kappa_2 d_2}}$$

$$\varepsilon''_{rj} = \text{Im}\,[\varepsilon_{rj}]$$

For the vertical polarization the weighting functions are

$$W_{vj}(z) = 4 \cos\theta \varepsilon_{r0}^{-1/2} \frac{\omega}{c} \varepsilon''_{rj} \cdot \left[\left|\frac{da_j}{dz}\right|^2 + \left(\frac{\omega}{c}\right)^2 \varepsilon_{r0} \sin^2\theta\, |a_j(z)|^2\right] \qquad j = 1, 2, 3 \qquad (3b)$$

where the corresponding functions $a_j(z)$ are obtained from the previous set of equations by simply substituting μ_0 with $\varepsilon_0 \varepsilon_{rj}$.

The weighting functions for direction of observation perpendicular to the layers of tissues computed for three microwave frequencies are reported in Figure 4. Note how the effect of the interference patterns in the skin and in the fat layer produced by reflections at the interfaces between the different tissues and from the boundary of the structure becomes apparent as the frequency is increased. At the same time, the monotonically decreasing branch of the muscular weighting function becomes steeper, thus indicating that in general, the region contributing to the brightness becomes shallower when frequency increases. However, it is worth noting that with the exception of the deepest layers the intensity of the emerging radiation originating at a given depth of the muscle is not a monotonic function of frequency, in accordance to the global results already shown in Figure 3.

Retrieval algorithm. Equation (2) relates the unknown temperature profile $T(z)$ to the known function $T_B(\alpha)$. In practice, the temperature at selected locations in the tissues is to be determined from a suitable set of measured brightness temperatures. In the following discussion, values of the brightness temperature at several frequencies for a fixed direction of observation, perpendicular to the layers, are assumed to be known. The form of the kernel of (2) indicates that the problem is ill conditioned and that the solution is affected by numerical instability. A suitable inversion technique must therefore be used to circumvent such a difficulty.

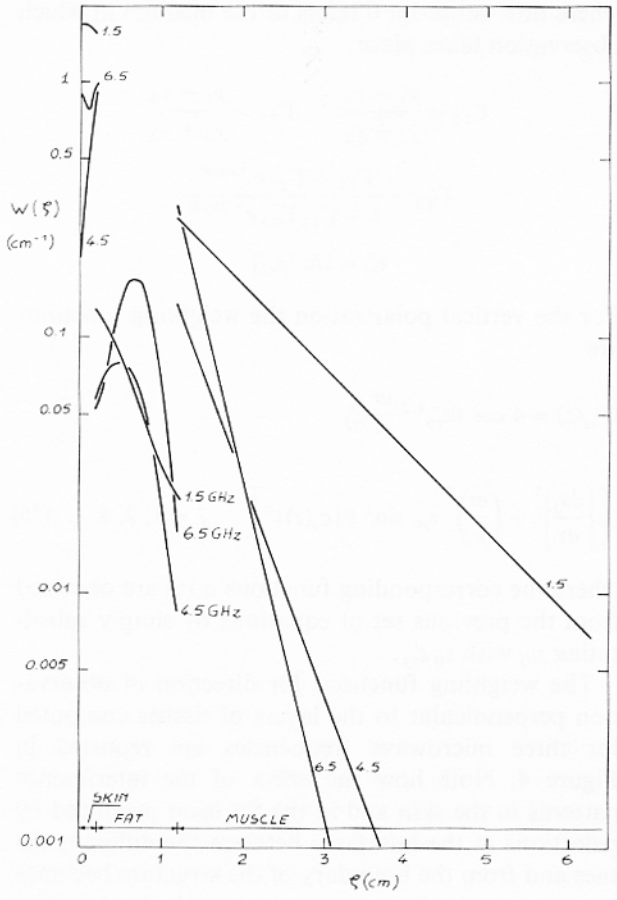

Fig. 4. Weighting functions in the biological tissues at three frequencies for direction of observation perpendicular to the layers; ζ is the depth in the tissues.

Of the several inversion methods that have been proposed for the solution of radiometric problems, the Kalman filtering estimation algorithm has proven to be a fairly effective technique, from the point of view of both accuracy of retrievals and attainable spatial resolution [*Basili et al.*, 1981]. In addition, being a recursive technique, Kalman filtering is well suited for use in monitoring the temporal evolution of temperature profiles, as is required in the relevant case of hyperthermic treatments of tumoral tissues.

The emitting biological structure is regarded as a linear, stochastic dynamical system, and the brightness temperature T_B, which carries the information about its thermal state, is assumed to be linearly related to the thermal state vector $\mathbf{T}$:

$$\mathbf{WT} = \mathbf{T}_B + \mathbf{N} \tag{4}$$

where $\mathbf{T}_B$ is the vector of measurements, i.e., the brightness temperatures at the various frequencies; $\mathbf{W}$ is the observation matrix, i.e., the matrix of the discretized weighting functions; and $\mathbf{N}$ represents the noise in the measurements. The Kalman linear estimation procedure yields a minimum variance estimate of $\mathbf{T}$ from data $\mathbf{T}_B$:

$$\hat{\mathbf{T}} = \hat{\mathbf{T}}_p + \mathbf{K}(\mathbf{T}_B - \hat{\mathbf{T}}_{Bp}) \tag{5}$$

where now subscript p denotes an a priori expectation and the circumflex denotes the estimate. Vector $\hat{\mathbf{T}}_{Bp}$ contains the brightness temperatures corresponding to the a priori expected temperature profile $\hat{T}_p$. Kalman gain $\mathbf{K}$ is derived from the error covariance matrix $\mathbf{P}_p$ of the a priori expected profile $\hat{\mathbf{T}}_p$ and from the error covariance matrix $\mathbf{R}$ of the measurement vector $\mathbf{T}_B$, through the formula

$$\mathbf{K} = \mathbf{P}_p \mathbf{W}^T (\mathbf{W}\mathbf{P}_p \mathbf{W}^T + \mathbf{R})^{-1} \tag{6}$$

which settles an optimal balance between a priori estimate and measurement innovation, according to the precision of the former and the accuracy of the latter. Matrix $\mathbf{P}_p$, whose elements are the second moments of the random components of vectors $\mathbf{T}_i - \mathbf{T}_p$ (where $\mathbf{T}_i$ is an individual thermal profile), provides a statistical measure of the uncertainty in the a priori expected profile. Similarly, matrix $\mathbf{R}$, formed by the second moments of the random components of the measurement vectors $\mathbf{T}_B$, gives information on the error by which the measurements are affected. The possibility of propagating estimates in time, whenever required, is a significant feature of this approach. The evolution of the thermal state of the biological structure is described by

$$\mathbf{T}_{i+1} = \boldsymbol{\phi}_{i,i+1} \mathbf{T}_i + \mathbf{S}_i \tag{7}$$

where $\boldsymbol{\phi}_{i,i+1}$ is the matrix which expresses the deterministic transition between successive states, denoted by subscripts i and $i+1$, and $\mathbf{S}_i$ is the random component of the transformation. An analogous equation allows covariance $\mathbf{P}_{i+1}$ of the error affecting the state vector $\mathbf{T}_{i+1}$ to be calculated from previous covariance $\mathbf{P}_i$. The propagated temperature vector and covariance matrix are identified with the a priori expectation $\hat{\mathbf{T}}_p$ and covariance $\mathbf{P}_p$ to start a recursive procedure through which successive thermal profiles are estimated from a sequence of radiometric measurements. Information supplied by past measurements is therefore adequately exploitable.

Implementation of the filter. An a priori expected thermal profile is necessary to initiate the filter. Because of the obvious difficulty in obtaining suf-

ficiently ample statistics from in vivo measurements, a thermal model of human living tissues has been used [*Jain*, 1980]. This model takes into account the metabolic heat generation and the heat exchange effect with the blood flow, in addition to conduction effects and the heat exchange with the environment. Random fluctuations with Gaussian distributions have been superimposed on the average of the 10 parameters of the model to simulate the variable characteristics of the single individuals. About 1000 sets of parameters were needed to obtain stable statistics. The average temperature profile was assumed as the initial expected profile, while the error covariance matrix of this a priori estimate was determined by comparing the temperatures of each individual profile with the means calculated at the various depths within the tissues.

The accuracy of the measurements of the brightness temperatures at the various frequencies is limited by the instrumental noise. The noise equivalent temperature of the radiometric system has been assumed to be 0.5°K at all wavelengths of operation, and, in addition, a certain degree of correlation between successive measurements has been considered to evaluate the covariance matrix of the measurement error.

As has already been said, the determination of temperatures in the human body can be a valuable support in diagnostics and, on the other hand, is a need during hyperthermic treatments. In the diagnostic application, since the temperature in neoplastic tissues is higher than that of the surrounding normal tissues of the host, a localized increase of temperature must be detected. In hyperthermic applications the temperature in the tissues must be continuously monitored to ensure that the required heating rate in the tumor bulk is attained without concurrent damaging effects on the surrounding normal cells. This paper refers mainly to the first application, in which the recursiveness of the Kalman filter is not exploited, except for eventually improving the retrievals from single-measurement data sets. In case the evolution of temperatures is to be monitored, the transition matrix relating the temperature profile at a given time to the previous one must be determined. To this end the model already adopted to evaluate the a priori profile and the associated covariance matrix is augmented to include the time-varying power deposition rate by the electromagnetic field applied in the hyperthermic treatment. A solution of this transient thermal problem through a variational formulation allows the temporal evolution of the temperatures to be determined as the dynamical response of an infinite-dimension system [*Bardati*, 1981]. This approach is well suited, since it leads naturally to the introduction of the transition matrix, which can be evaluated in terms of thermal conductivities, densities, specific heats of tissues, and the cooling effect due to blood flow, as well as the eigenvalues of the pertinent self-adjoint boundary value problem. Once the transition matrix and the statistics of the random component $\mathbf{S}_i$ of (7) are known, the filter can be implemented recursively to monitor the transient thermal response of tissues undergoing electromagnetic heating in hyperthermia.

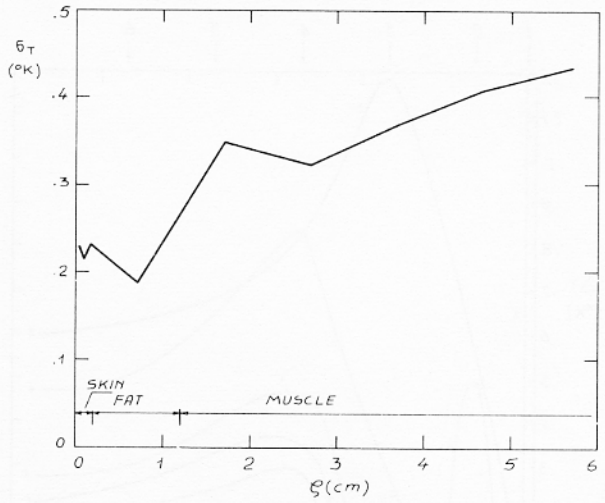

Fig. 5. Standard deviation of temperature retrieved from 18 brightness measurements; the noise equivalent temperature of the radiometer is assumed to be 0.5°K at all frequencies; ζ is the depth in the tissues.

Numerical results. The effectiveness of the Kalman filter in retrieving thermal profiles of layered biological structures from noisy radiometric data has been tested both for the case of normal tissues and when a localized temperature increase is present. Suitable statistics have been produced by randomizing the thermal model described above as well as by adding noise to the brightness temperatures at 18 different wavelengths, used as data sets. The selection of the 18 frequencies in the range 0.975–13 GHz attempted to minimize the interdependence of the corresponding kernels in (2), thus increasing the information content of the set of brightness temperatures. Figure 5 shows the standard deviation of the retrieved temperatures at the various depths for a normal profile. It can be noted that the accuracy of

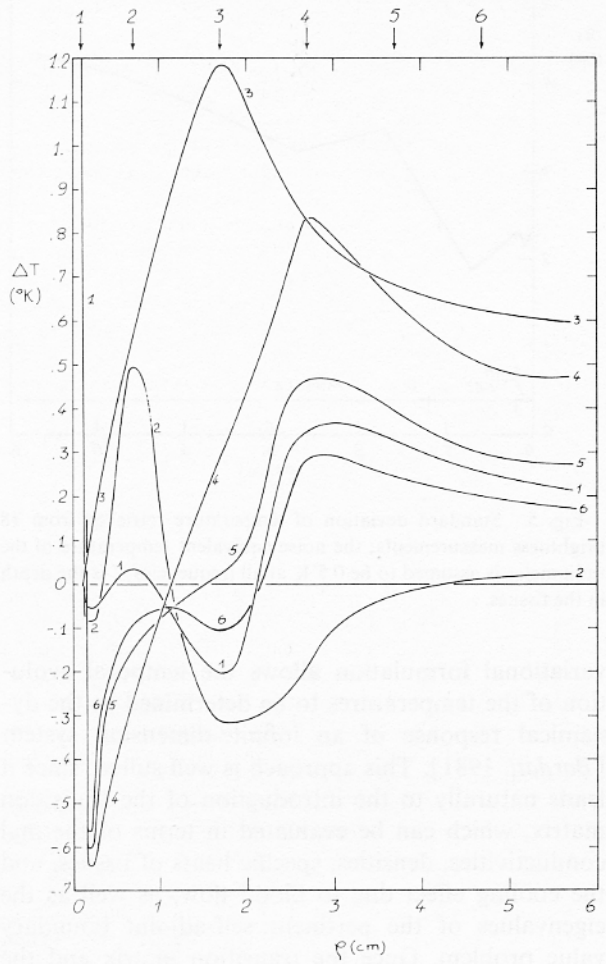

Fig. 6. Average excess temperature retrieved from 18 brightness measurements for varying position of a thermal anomaly $\Delta T_0 = 2.2°K$; noise equivalent temperature of radiometer is 0.5°K; ζ is the depth in the tissues. Curves 1 to 6 refer to the six locations of the thermal anomaly indicated by the arrows in the upper part of the figure, i.e., 0.2, 0.7, 1.7, 2.7, 3.7, and 4.7 cm, respectively.

retrievals tends to decrease at increasing depths, as does the emerging power which carries the useful information.

To assess the algorithm's capability of detecting thermal anomalies, the temperature at a single depth has been increased by 2.2°K, leaving the normal profile unchanged at the other locations, as can be the case for a localized neoplastic alteration of tissues. The averaged excess temperatures retrieved by Kalman-filtering 1000 noisy brightness data sets are reported in Figure 6. The thermal anomaly appears to be detected and correctly located except when deep muscular tissues are involved. However, although in this case the exact position of the anomaly is not recovered, the whole muscle appears to be warmer than in the normal situation.

CONCLUSIONS

The problem of determining the internal temperature distribution in the human body by radiometric means is examined. Microwave emission from a plane parallel three-layer model of living tissues has been evaluated, and the corresponding weighting functions have been determined. The numerical solution of the inverse problem by the Kalman filtering algorithm has then been investigated with the intent of gaining information on the temperature retrieval accuracy of the algorithm. Noninvasive measure of temperature within the human body by radiometric techniques is a difficult task for several reasons. Electromagnetic modeling is complicated by inhomogeneities and complex shapes of structures. Processing of measured brightness data for temperature retrievals has to surmount numerical instabilities, which amplify the errors introduced both by the measuring instrumentation and by the random deviations of single individuals from the ensemble average. The radiometric instrumentation exhibits involved characteristics because of the heavy requirements it has to meet.

Nevertheless, the general relevance of the results which can be obtained in this field seems to justify a first attempt to solve quantitatively the problem of human microwave thermography.

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Fernando Bardati and Domenico Solimini, Dipartimento di Ingegneria Elettronica, Facoltà di Ingegneria, II Università di Roma (Tor Vergata), Via O. Raimondo, 00173 Rome, Italy.