An Eccentrically Coated Asymmetric Antenna Applicator for Intracavitary Hyperthermia Treatment of Cancer

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Abstract—In this paper a model based on transmission line theory is used to predict the behavior of an eccentrically coated asymmetric antenna applicator for use in intracavitary hyperthermia. Theoretical results for the heating rate (HR) of the applicators are compared to experimental results. The experimental results were obtained at City of Hope National Medical Center using four different 915-MHz applicators, each with a different antenna size and eccentricity of the coating.

A parameter \( \delta \) is defined where \( \delta \ll 1.0 \) is a thin wire approximation; \( \delta \) is primarily a function of the eccentricity of the coating, the antenna diameter, and the coating diameter. It is found that when \( \delta \leq 0.5 \), the theoretical model works well. In particular, it predicts the directivity due to the eccentricity of the coating. However, as this eccentricity is increased or as the antenna diameter is increased (\( \delta \geq 0.6 \)), the model no longer accurately predicts directivity. Thus, the model that can be used to predict the HR profiles for an eccentrically coated asymmetric antenna only when \( \delta \leq 0.5 \).

I. INTRODUCTION

In recent years, there has been increasing interest in the treatment of cancer using hyperthermia as an adjunct therapy—that is, the application of heat in combination with radiation or chemotherapy. One method of hyperthermia utilizes intracavitary applicators [1], [2]. This noninvasive technique involves the placement of an insulated microwave antenna in a natural body cavity and can be used to treat cancer of the rectum, colon, cervix, uterus, stomach, esophagus, and other body cavities.

During an ideal hyperthermia session only the tumor would be heated [3]. The design of a delivery system to approach this ideal is an ongoing research effort [4]-[6]. One method is to use a directive applicator to maximize heating in the tumor and minimize healthy tissue exposure.

In this paper a microwave antenna applicator with a homogeneous, eccentric Teflon coating is examined for use in hyperthermia (see Fig. 1). Transmission line theory is used to find the current on the antenna surface which is then used to calculate the electric field. The heating rate (HR) patterns, where the heating rate is the amount of temperature increase per unit time per unit power [14], are found indirectly from the electric field. First, the specific absorption rate (SAR) is used in a circuit analogy to approximate the induced temperature increase, \( \Delta T \), this accounts for any diffusion effects that may be present. The \( \Delta T \) profiles are then normalized by the exposure time and the applied power to obtain the HR profiles. These theoretical HR results were compared to experimental results obtained at City of Hope National Medical Center using four different 915-MHz applicators, each with a different antenna size and eccentricity of the coating.

A parameter \( \delta \) describing the uniformity of the current distribution around the antenna surface is defined [see (21)]. It is found that when \( \delta \leq 0.5 \), the theoretical model works well. In particular, it accurately predicts the directivity due to the eccentricity of the coating. However, as this eccentricity is increased or as the antenna diameter is increased, the model breaks down. In addition, the model does not adequately account for junction effects near the antenna feedpoint.
II. THEORY

The use of equivalent transmission line parameters to describe the behavior and performance of a concentrically insulated antenna is well established [7]-[9]. Work done by T. T. Wu, L. C. Shen, and R. W. P. King (henceforth referred to as WSK theory), demonstrated that the current on an eccentrically coated antenna can also be described by equivalent transmission line parameters [10], [11]. For the derivation of this result and the derivation of the electromagnetic fields for this applicator, the reader is referred to [10]-[12]. In this paper only the results from the application of WSK theory are presented.

The applicator is divided into three regions as shown in cross section in Fig. 1 and in profile in Fig. 2. Region 1 is the asymmetric antenna consisting of the conductor with radius \( a \) centered at \( x = D \); it is assumed to be a perfect conductor. Region 2 is a homogeneous dielectric layer with radius \( b \), permittivity \( \varepsilon_2 \), and wavenumber \( k_2 = \beta_2 \). Region 4 corresponds to the body tissue one is trying to heat. It is a lossy medium with permittivity \( \varepsilon_4 \), finite conductivity, \( \sigma_4 \), wavenumber \( k_4 = \beta_4 + i\omega\varepsilon_4 \varepsilon_0 \) (where \( \varepsilon_0 = \sqrt{-1} \)); it is homogenous and infinite in extent. A cylindrical coordinate system is used (see Fig. 2) where \( r = \sqrt{x^2 + y^2} \) and \( \theta = \arctan(y/x) \).

The bare antenna is divided into three sections as shown in Fig. 3: the upper and lower sections and the junction. The asymmetric antenna is fed at \( z = 0 \) in the center of the gap of length 2\( g \). The upper section of the antenna has a fixed length of \( h_u - g \) while the lower section has a length of \( h_b + g \) determined by the insertion depth. The insertion depth, \( h_b + h_u \), is the overall length of the antenna placed in the tissue.

WSK theory [10] and expressions derived in [12] are used to find the electric field components in the lossy medium, region 4:

\[
E_z(r, \theta, z) = \frac{\mathcal{J}(z)k_2^2}{2\pi\omega\varepsilon_2(k_2b)} \left\{ H_0^1(k_2r) \over H_1^0(k_2b) \right\} + 2 \sum_{m=1}^{\infty} \left( {b_0 \over b} \right)^m \frac{H_m^1(k_2r)}{H_{m+1}^0(k_2b)} \cos(m\theta) \tag{1}
\]

\[
E_t(r, \theta, z) = F_{Dz} \left( F_{Dz} - \omega F_{Bz} \right) \tag{2}
\]

\[
E_r(r, \theta, z) = F_{Dz} \left( F_{Dz} - \omega F_{Br} \right) \tag{3}
\]

where

\[
I(z) = \begin{cases} 
V_0/Z_r \sin[k_2(h_u - z)] & \text{for } g < z \leq h_u \\
V_0/Z_t \sin[k_2(h_u - g)] & \text{for } |z| \leq g \\
V_0/Z_t \sin[k_2(h_b + g)] & \text{for } -h_b < z \geq -h_b \\
0 & \text{otherwise}
\end{cases} \tag{4}
\]

\[
Z_t = iZ_r \left( \tan[k_2(h_u - g)] + \tan[k_2(h_b - g)] \right) \tag{5}
\]

\[
Z_r = {k_t \Omega_a \over 2\pi\omega\varepsilon_2} \tag{6}
\]

\[
k_t^2 = k_t^2 \left[ 1 + {\Delta_0 \over k_b\Omega_a} \right] \tag{7}
\]

\[
\Omega_a = \cosh^{-1} \left( {a^2 + b^2 - D^2 \over 2ab} \right) \tag{8}
\]

\[
\Delta_0 = H_0^0(k_bD) \tag{9}
\]

\[
\varepsilon_0 = \left( {1 \over 2D} \right) \left\{ (b^2 + D^2 - a^2) - (b^2 + D^2 - a^2) - 4b^2D^2 \right\} \tag{10}
\]
and
\[
F_{Er} = \frac{\mu_0}{2\pi \omega} \sum_{m=1}^\infty \left[ \sum_{b=1}^\infty \left( \frac{x_0}{b} \right)^m \left( \frac{-H_0^{(1)}(k_rb)}{H_0^{(1)}(k_rb)} + 2 \sum_{n=1}^\infty \left( \frac{x_0}{b} \right)^n \frac{mH_r^{(1)}(k_rb) - H_{r+1}^{(1)}(k_rb)}{k_rbH_{r+1}^{(1)}(k_rb)} \cos(m\theta) \right) \right]
\]
\[
F_{Er} = \frac{-2k_0^2}{2\pi \omega} \sum_{m=1}^\infty \left[ \sum_{b=1}^\infty \left( \frac{x_0}{b} \right)^m \frac{H_0^{(1)}(k_rb)}{H_0^{(1)}(k_rb)} \sin(m\theta) \right]
\]
\[
F_{gb} = \frac{\mu_0}{\pi \omega b} \sum_{m=1}^\infty \left[ \sum_{b=1}^\infty \left( \frac{x_0}{b} \right)^m \frac{mH_r^{(1)}(k_rb) - H_{r+1}^{(1)}(k_rb)}{k_rbH_{r+1}^{(1)}(k_rb)} \sin(m\theta) \right]
\]
\[
F_{gb} = \frac{-2k_0^2}{2\pi \omega b} \sum_{m=1}^\infty \left[ \sum_{b=1}^\infty \left( \frac{x_0}{b} \right)^m \frac{H_0^{(1)}(k_rb)}{H_0^{(1)}(k_rb)} \sin(m\theta) \right]
\]
\[
F_{Dist} = F_{EM} + F_{Eg} + F_{s} + F_{h} + F_{hs}
\]
\[
F_{gap} = \frac{V_0}{2Z(k_2 - k_1)} \left\{ \begin{array}{ll}
{k_2 \cos [k_2(h_b - g)] + u_k \sin [k_2(h_b - g)]} & {e^{ik_2(z - g)}} \\
{-k_2 e^{ik_2(h_b - g)} e^{ik_2(z - g)}} & {z > -g} \\
{-k_2 \cos [k_2(h_b - g)] + u_k \sin [k_2(h_b - g)]} & {e^{-ik_2(z + g)}} \\
{k_2 e^{ik_2(h_b - g)} e^{-ik_2(z + g)}} & {z > -g}
\end{array} \right. 
\]
\[
F_{ag} = \frac{V_0}{2Z(k_2 - k_1)} \left\{ \begin{array}{ll}
e^{ik_2(z - h_a)} & {z > h_a} \\
e^{-ik_2(z + h_a)} & {z < h_a}
\end{array} \right. 
\]
\[
F_{ha} = \frac{V_0 k_1}{2Z(k_2 - k_1)} \left\{ \begin{array}{ll}
e^{ik_1(z + h_a)} & {z > h_a} \\
e^{-ik_1(z - h_a)} & {z < h_a}
\end{array} \right. 
\]
\[
F_{hb} = \frac{V_0 k_1}{2Z(k_2 - k_1)} \left\{ \begin{array}{ll}
e^{ik_1(z + h_b)} & {z > h_b} \\
e^{-ik_1(z - h_b)} & {z < h_b}
\end{array} \right. 
\]

where \( \omega \) is the frequency in radians per second, \( H_r^{(1)} \) are Hankel functions of the first kind [13], the time dependence is \( \exp(-i\omega t) \), and \( x = x_0 \) is the position of an equivalent line current used in the solution for the fields.

\( V_0 \) is the voltage in the gap region calculated from transmission line theory using the net power applied to the applicator.

WSK theory uses a thin wire approximation to solve for the current on the antenna surface. The assumption is that there are only uniform, axially-directed currents on the conductor. This is equivalent to

\[
\delta = \left| \frac{\Delta_0}{k_b b \Omega_n} \right| \ll 1.
\]

The actual quantity of interest to clinicians is the heating rate (HR) in the tissue [14]. To obtain the theoretical HR profiles the specific absorption rate (SAR) must first be calculated. The SAR is related to the electric field in the lossy medium by [15]

\[
SAR = \frac{\sigma_4}{2\rho} |\vec{E}_4|^2 \left( \frac{W}{kg} \right)
\]

where \( \vec{E}_4 = \vec{r}E_{x4} + \hat{\theta}E_{\theta4} + \hat{z}E_{z4} \) is the electric field in region 4 and \( \rho \) is the tissue density. The vectors \( \vec{r}, \hat{\theta}, \) and \( \hat{z} \) are unit vectors in the cylindrical coordinate system.

To calculate the initial rate of heating, HR profiles are often computed from the SAR using a conversion factor of 4186 (J/kcal), and \( C_p \), the specific heat in muscle phantom [14].

\[
HR = \frac{\text{SAR}}{4186C_pP_{\text{net}}} \left( \frac{^\circ{\text{C}}}{\text{Min. W}} \right).
\]

where \( P_{\text{net}} \) is the applied power. This equation for the HR makes use of \( P_{\text{net}} \) and differs from the definition in [14] where the SAR is specified in per unit Watt input. Equation (23) is accurate for exposure times when diffusion is negligible. However, for long exposure times diffusion cannot be ignored, and an alternate definition of the heating rate is used [14]

\[
HR = \frac{\Delta T}{t_mP_{\text{net}}} \left( \frac{^\circ{\text{C}}}{\text{Min. W}} \right)
\]

where \( \Delta T \) is the induced rise in temperature and \( t_m \) is the exposure time.
To obtain an accurate theoretical $\Delta T$ profile, a circuit analogy was used to model the heat transfer process including diffusion [12], [16], [17]. It used the SAR profile as an input into the model and calculated the resulting $\Delta T$ profile using a network simulation routine, SPICE [18] to find the transient response.

### III. Experiment and Results

Four different applicators were made using semi-rigid coaxial cable (Micro-COAX Inc., Collegeville, PA, part numbers UT-34M and UT-85) and Teflon sheaths. These four applicators, referred to as Cases 1, 2, 3, and 4, were used to examine the effect of varying geometries and eccentricities (see Table I). Case 1 had the least eccentricity, smallest antenna radius, and a thin wire parameter $\delta = 0.442$. Case 2 was similar to Case 1 but had a greater eccentricity and $\delta = 0.490$. In Case 3 the antenna diameter was approximately twice that of Cases 1 and 2, and $\delta = 0.678$. Finally, in Case 4 the antenna was the same diameter as in Case 3 but the sheath was slightly smaller; it had the greatest eccentricity and the largest thin wire parameter $\delta = 1.03$. Note that $\delta$ is a function of the antenna radius $a$, the eccentricity $D$, and the sheath radius $b$ [see (8), (9) and (21)]. Because of the assumption in WSK theory that $\delta << 1$, it was expected that the results would be most accurate for Case 1 and least accurate for Case 4. This prediction was confirmed by the following results. It was found that when $\delta \leq 0.5$, the theory works quite well. However, for $\delta \geq 0.6$ the theory is no longer accurate.

The upper section, $h_a - g$, of each antenna was a quarter wavelength long ($\beta h_a = (\pi / 2)$) at 915 MHz. The length of the lower section, $h_a - g$, was measured at the time of each experiment by the insertion depth. Wavenumbers and the physical parameters for muscle phantom are listed in Table II [15], [19], [20].

Each antenna was made by removing a small section (length $= 2g$) of the outer conductor of the coaxial cable together with the insulation beneath. The outer conductor of the $h_a - g$ section was then soldered to the exposed inner conductor at the gap to form the upper section of the antenna. The lower section of the antenna consisted of the outer conductor, starting from the gap and continuing toward the microwave generator (see Fig. 3). Prior to insertion in the Teflon sheath the antenna was encased in Mylar heatshrink tubing and then coated with Vaseline. The heatshrink tubing was used to provide the antenna with rigidity and strength. The Vaseline was used for lubrication and to minimize any air pockets between the antenna and sheath. The heatshrink tubing was of negligible thickness (0.0004 cm) and was ignored in the theoretical model. At 915-MHz Vaseline has the same dielectric constant as Teflon and was treated as such in the model [19].

A phantom model was created by filling a rectangular acrylic box 30 cm wide, 30 cm long, and 7.6 cm deep with muscle phantom [21]. An applicator was positioned with the gap region centered on the surface of the phantom and covered with more muscle phantom so that $h_a - g = 3a / 4$. A MCL Inc. 915-MHz microwave source provided power. A Narda 3020A directional coupler and two Hewlett Packard 486A power meters with 8481A power sensors were used to measure the forward and reflected power at a sampling rate of approximately 12 samples per second. After exposure a UTI-9000 infrared camera system was used to take infrared pictures, or thermograms. These images were then processed on an HP-3200/320 minicomputer to filter out noise in the thermal images and to obtain line scans from the images. A before-exposure thermogram was then subtracted from an after-exposure thermogram to obtain the induced temperature profile, $\Delta T$. Each profile was normalized by the exposure time, $t_{ex}$, and the net power, $P_{net}$ (see Table III), to obtain an HR profile.

The results are shown in Figs. 4–9. In all the figures the solid line corresponds to the experimental result while the dashed line is the theoretical result. The feedpoint is located at $z = 0$, positive values of $z$ correspond to the upper length of the antenna $(h_a - g)$, and negative values of $z$ correspond to the lower portion $(h_a - g)$. All the data are calculated assuming observation points adjacent to the outer surface, $r = b$, of the applicator.

Each figure corresponds to a different experiment, each with a different net input power and exposure time. The theoretical results for the $\Delta T$ profiles were obtained by
TABLE III

<table>
<thead>
<tr>
<th>Case</th>
<th>θ</th>
<th>t (sec)</th>
<th>$P_{in}$ (W)</th>
<th>$P_f$ (W)</th>
<th>$P_r$ (W)</th>
<th>$\Delta T_x$ °C</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.0°</td>
<td>46.47</td>
<td>63.03</td>
<td>83.58</td>
<td>20.55</td>
<td>3.0°</td>
</tr>
<tr>
<td></td>
<td>110.0°</td>
<td>61.00</td>
<td>64.32</td>
<td>83.08</td>
<td>18.76</td>
<td>2.0°</td>
</tr>
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<td>0.0°</td>
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<td>109.83</td>
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<td>15.71</td>
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<tr>
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<tr>
<td></td>
<td>150.0°</td>
<td>20.63</td>
<td>121.76</td>
<td>139.37</td>
<td>17.61</td>
<td>none</td>
</tr>
</tbody>
</table>

Table III: Exposure Times, Power Levels, and Resistive Heating Components for Cases 1 and 3, where $t$ is the exposure time, $P_{in}$ is the net power, $P_f$ is the forward power, $P_r$ is the reflected power, and $\Delta T_x$ is the amount of resistive heating subtracted from the overall experiment $\Delta T$ profile before normalization and conversion to HR.

Fig. 4. Comparison of theoretical (dashed line) and experimental (solid line) heating rate (HR) profiles for Case 1 at θ = 0° and $r = \delta = 5.0$ mm. The dipole feedpoint is located at $z = 0$, $\delta = 0.442$.

Fig. 5. Same as Fig. 4 but θ = 110°.

Fig. 6. Same as Fig. 4 but θ = 180°.

Fig. 7. Comparison of theoretical (dashed line) and experimental HR profiles Case 3 at θ = 0°, $\delta = 0.678$.

Fig. 8. Same as Fig. 7 but θ = 70°.

assuming the same net input powers and exposure times used in the experiments (see Table III). In order to quantify the directivity of a particular applicator it was necessary to normalize both theoretical and experimental results to the same net input energy for the θ = 0° result. These results were then converted to HR's using the input power, the exposure time for the θ = 0° case, and (24). The HR's were greatest for θ = 0° and tapered off as θ
went to ±180°, where it was the least. For each applicator three different observation angles were used: one where the insulation between the muscle and antenna surface was the thinnest (θ = 0°), one where the insulation was the thickest (θ ≈ 180°), and another approximately half-way between the previous two angles (θ = 90°).

For Cases 1 and 2 the power levels for the coaxial cable used (Micro-COAX Inc., UT-34M) were near the manufacturer's maximum rated level and exposure times were long (see Table III). This ensured clarity and detail in the thermal images, but it also gave rise to a resistive heating component, ΔT_R, due to cable-loss heat diffusing into the muscle phantom. This resistive heating was quantified using the manufacturer's cable loss (60 dB/100 ft) and the forward and reflected power in each of the experiments (see Table III) in the circuit analogy. The ΔT_R's were subtracted from the experimental ΔT's before conversion to HR's. In Cases 3 and 4 resistive heat conducted from the antenna surface to the muscle phantom was small because of the shorter exposure times so no offset was needed.

Figs. 4–6 show the results for Case 1 where δ = 0.442. In Fig. 4 (θ = 0°) there is rough agreement between theory and experiment. Discrepancies occur at the junction and along the lower length of the antenna. These discrepancies are due to junction effects and experimental error. The junction effects add an extra peak in the HR profile at z = 0. Differences along the lower half were caused by misplacement of the muscle phantom on top of the applicator. Muscle phantom was positioned beyond the end of the Teflon sheath affecting the location of the equivalent open circuit—that is, where the current vanished on the lower end of the antenna. Instead of being approximately at the end of the sheath, as in all other experiments, the open circuit occurred further along the coaxial cable. To compensate for the misplacement, the lower length of the antenna, h_s, was increased in the model. In Figs. 5 and 6 where θ = 110° and 180°, respectively, there is good agreement between the theoretical and experimental results. Junction effects are not noticeable because of the greater thickness of insulation between the antenna surface and the muscle phantom. Note also that the peak HR decreases as θ increases, demonstrating the directivity of the applicator. In Case 2 (δ = 0.490) the results (not shown) were similar with slightly greater discrepancies between theoretical and experimental results [12].

Figs. 7 through 9 show the results for Case 3, for which δ = 0.678. It is clear from the results that the assumption of only δ-directed currents degenerates. At all three observation points there are major discrepancies between the theoretical and experimental results. The experimental HR is much higher than predicted by theory. Junction effects are also prevalent at all observed angles while in Cases 1 and 2 these effects were negligible near θ = 90° and θ = 180°. In Case 4 (δ = 1.03) there was even stronger disagreement between theory and experiment [12].

The results above show that near θ = 0°, where the insulation was thinnest, the fields are most sensitive to junction effects and to the current distribution and are less sensitive as θ is increased toward 180°. Junction effects are more noticeable as the eccentricity increases and as the antenna diameter increases relative to the sheath diameter. When the theory starts to break down, it is mainly in the θ dependence on the amplitude of the HR distributions. The amplitude is incorrect for all angles, but the general shape of a heating pattern is still correct.

IV. CONCLUSIONS

An eccentrically coated asymmetric antenna applicator was examined for use in microwave intracavitary hyperthermia at a frequency of 915 MHz. Four different applicators were designed based on transmission line theory. Each applicator had a different conductor size or eccentricity. Semi-rigid coaxial cable was used to make the antennas and Teflon was used for the insulating sheaths. Muscle phantom experiments were performed with each applicator at three observations angles: one where the insulation between the antenna and muscle phantom was the thinnest (θ = 0°), one where the insulation was the thickest (θ ≈ 180°), and another approximately half way between the previous two angles (θ = 90°). Temperature profiles (ΔT) were measured using infrared thermography and then converted to heating rate profiles (HR). The results predicted by the theory were compared to experimental results to determine the accuracy of the model in predicting the HR profiles.

In the theoretical model it is assumed that there are only axially directed currents—a thin wire approximation. This is equivalent to the condition δ << 1 where δ is primarily a function of the antenna diameter and eccentricity. Results show that WSK theory is accurate when δ < 0.5, but as δ is increased beyond 0.6, the assumption that there are only axially directed currents breaks down. When this occurs, the theory predicts a significantly lower HR when compared to experimental results; however, it correctly predicts the axial profile shape. Other discrepancies be-
between theory and experiment are attributed to junction effects near the antenna feedpoint, which are not accounted for in the model. The model could be improved by including these junction effects along with the effects of nonaxial currents on the conductor surface. Finally, the model accurately predicts the directivity of the applicator due to its eccentricity for $\delta \leq 0.5$.

REFERENCES


Charles W. Manry, Jr. (S’92) was born in Houston, TX, in 1965. He received the B.S. and M.S. degrees in electrical engineering from Washington State University, Pullman, in 1988 and 1990, respectively. He is currently working towards his doctoral degree in electrical engineering at Washington State University. His interests are in biomedical engineering with emphasis on ultrasonic imaging of tissues.

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Shira Lynn Broschat (M’89) received the B.S. degree in 1982, the M.S. degree in 1985, and the Ph.D. degree in 1988, all in electrical engineering from the University of Washington, Seattle. From 1983 to 1985 she was a Research Associate for the Bioelectromagnetics Research Laboratory at the University of Washington, where she did work on microwave hyperthermia treatment of cancer. From 1985 to 1989 she was a Research Associate at the University of Washington’s Applied Physics Laboratory, where she was involved in research on wave scattering from rough surfaces. In February 1989 she joined the faculty of the School of Electrical Engineering and Computer Science at Washington State University, Pullman. Her current research interests are in rough surface scattering, ultrasound mammography, radar scattering from icy moons, high-resolution remote sensing of the ocean, and engineering education.

Dr. Broschat was the recipient of a Young Investigator Award from the Office of Naval Research in 1989 and a Presidential Young Investigator Award from the National Science Foundation in 1990.

Chung-Kwang Chou (S’72-M’75-SM’86-F’89) was born in Chung-King, China on May 11, 1947. He received the B.S. degree from the National Taiwan University in 1968, the M.S. degree from Washington University, St. Louis, MO, in 1971, and the Ph.D. degree from the University of Washington, Seattle, in 1975, all in electrical engineering.

During his graduate study at the University of Washington, he had extensive training in both electromagnetics and physiology. He spent a year as a NIH postdoctoral fellow in the Regional Primate Research center and the Department of Physiology and Biophysics at the University of Washington. He was Research Associate Professor in the Center for Bioengineering and the Department of Rehabilitation Medicine, as well as Associate Director of the Bioelectromagnetics Research Laboratory until August 1985, engaged in teaching and research in electromagnetic dosimetry, exposure systems, biological effects of microwave exposure, and RF hyperthermia for cancer treatment. He is now the head of the Biomedical Engineering Section and the Director of the Radiation Research at the City of Hope National Medical Center, Duarte, CA. His main re-
search there is in cancer hyperthermia. A consultant for the NCRP's Scientific Committee 53 on the biological effects and exposure criteria for radio frequency electromagnetic fields, he has also served on the ANSI Subcommittee C95.4, now the IEEE SCC28, since 1978. He was the Chapter Chairman of IEEE's Seattle Section on Antennas and Propagation/Microwave Theory and Technique in 1981-1982. He was on the Board of Directors of the Bioelectromagnetics Society and is now the Associate Editor of the Journal of Bioelectromagnetics. He is the Chairman of the Microwave and RF Subcommittee of the Committee on Man and Radiation, of the IEEE US Activities.

Dr. Chou received the Special Award for the Decade of the 1970s for contributions in medical and biological research in 1982, and he received the Outstanding Paper Award, both from the International Microwave Power Institute in 1985. He is a member of BEMS, AAAS, IMPI, the Radiation Research Society, North American Hyperthermia Group, Tau Beta Pi, and Sigma Xi.

John A. McDougall (M'86) received the B.A. degree in chemistry in 1953 and the B.S. degree in electrical engineering in 1965, both from the University of Washington, Seattle. From 1953 to 1962, he served in the USAF as a Squadron Officer and Pilot, and it was there that his interest changed from Chemistry to Electronics. In 1966, he became a Charter Member of what would become Bill Guy's Bioelectromagnetics Group at the University of Washington and remained there until 1985. During this period, he was instrumental in developing phantom modeling materials and thermographic techniques. In September 1985, he became a Senior Research Associate in the Department of Radiation Research of the City of Hope National Medical Center. His work here has centered on antenna development for intracavitary and interstitial hyperthermia.