Abstract—Microwave ablation is based on localized heating of biological tissues, enabled by an electromagnetic field. Ablation antennas are commonly designed in a forward approach, where the shape and type of antenna are manipulated to generate elevated temperatures in the target region. However, little attention has been dedicated to designing antennas in an inverse approach to allow controlled synthesis of temperature profiles customized for the application, in a reconfigurable way. Also, existing designs are based on generating a particular electric field or SAR profiles, not accounting for discontinuities in the thermal conductivity of tissue. We propose an inverse multiphysics strategy for source design that involves optimizing the current distribution on the antenna to synthesize a desired temperature profile, while accounting for temperature diffusion. We apply this optimization procedure to a simulated test case in human liver and study the associated fields and temperature profiles, as well as the optimized current distributions required. Our results indicate and quantify the clear advantage of this multiphysics approach compared to electric field focusing in terms of meeting objectives in the thermal domain. We show how the optimized current distributions can be easily implemented in a practical design and integrated into treatment planning in a clinical setting.

Index Terms—Antenna optimization, microwave ablation (MWA), multiphysics, Optimization.

I. INTRODUCTION

ABLATION therapy is a medical treatment that involves rapid delivery of heat to a targeted region in tissue, inducing temperatures in excess of 50°C [1]. This can severely damage ("ablate") the targeted cells over a short span of time, in the order of minutes. Applications of ablation include treatment of malignant tumours [2], [3], cardiac arrhythmias [4] as well as inflammations due to osteoarthritis [5]. One method of delivering heat to the target is through electromagnetic energy. Microwave frequencies generated via interstitial antennas have been used clinically as agents of heat delivery [1], and the respective treatment method is known as microwave ablation (MWA).

One major challenge with ablation treatment is designing sources to selectively apply heat to the target region without damaging surrounding healthy tissue. Several existing ablation antenna designs attempt to achieve this confinement by focusing an electric field within the target [2], [6]. This assumes that a focused electric field will lead to a focused temperature profile. However, the relationship between an applied electric field $\mathbf{E}$ and the consequent temperature profile $T$ is generally more complex, especially when the local electrical and thermal properties of tissue are inhomogeneous. Thus, focusing or confining the electric field does not guarantee an optimal temperature profile for ablation of a thermally inhomogeneous target region.

Moreover, several existing MWA sources are designed via a forward approach, where a particular type of antenna is proposed to generate a particular temperature profile [4], [7]–[12]. One major limitation of these designs is their inability to generate controllable temperature profiles customized to the shape of the region being targeted. Also, these designs are not reconfigurable; each design can only be used to create a specific temperature profile.

Another heat-based therapeutic technique that has been used successfully in clinical applications is hyperthermia, where an antenna array placed outside the body is used to focus the electric field at a target within the body. This is an inverse technique based on optimization of the array to focus the electric field or specific absorption rate (SAR) at the target. In [13], an antenna array was proposed that allows subwavelength focusing and shaping of the magnetic field in the near-field of the array via source optimization. This technique can also be applied to shaping the electric field in the near-field. However, little attention has been given to applying this inverse approach of array optimization in designing interstitial antennas for MWA.

In this paper, we propose a new inverse approach to designing interstitial sources for ablation therapy to synthesize controlled temperature profiles. The generated temperature profile depends on the applied electric field, which depends on the current distribution on the antenna. Thus, in our approach, we optimize the current distribution on the antenna based on an thermal profile Fig. 1 outlines the methodology of our approach: Given a target to ablate, a corresponding target temperature profile, $T_{\text{target}}$, and target electric field profile, $\mathbf{E}_{\text{target}}$, are defined. The ablation antenna is modeled as an array of surface current elements placed inside the target region, such that the current distribution on the elements generates an electric field $\mathbf{E}$ and...
corresponding temperature profile $T$. The current element magnitudes and phases are then optimized such that $T$ best approximates $T_{\text{target}}$. Rather than manipulating the physical shape of the antenna, we use the properties of antenna arrays to, through their near fields, control the generated temperature profiles. This allows for the possibility of designing reconfigurable antennas that can generate customized temperature profiles for different applications. Reconfigurability is possible in existing external hyperthermia applicators, but has not been extended to interstitial ablation therapy.

To optimize the current elements, we use a multiphysics approach that accounts for thermal inhomogeneities in tissue. We compare the results to the standard approach of simply generating strong and confined electric fields or SAR profiles, to confirm our claim that a purely field-based optimization procedure does not guarantee an optimal temperature distribution for a given target in biological tissue. Although it is intuitively apparent that a temperature-based multiphysics optimization approach would work better than a purely field- or SAR-based one, it is useful to quantify the differences between the two approaches and to ascertain whether the multiphysics approach is worth the added computational costs. We also present a proof of concept design of a potential antenna to synthesize the optimized current distributions obtained via the aforementioned process. The novelty of this paper is threefold.

1) We demonstrate, for the first time, how a MWA antenna can be optimized based on its intended thermal profile; note that so far, most ablation antennas in the literature are standard wire antennas, nearly omnidirectional in free space.

2) We show that this approach is meaningful, i.e., field focusing does not imply thermal focusing and that proper optimization of the antenna needs to be pursued with objectives in the thermal domain. Along the way, we clarify a long-standing confusion in the literature that the best way to create an ablation zone in the target is to focus the electric field around this area. We provide both field and thermal patterns to elucidate this point.

3) We provide a prototype for such an antenna that is simple to realize and reconfigure.

In Section II of this paper, we motivate the need for a multiphysics approach in designing ablation antennas. In Sections III and IV, we describe the theory behind the two optimization procedures considered here, and the geometry and simulation details of our test case. In Section V, we present the optimization results, including the optimized temperature profiles, current distributions, and a discussion and comparison of the two optimization approaches. In Section VI, we provide a physical implementation of the optimized currents, and discuss the potential integration of this design procedure in clinical treatment planning in Section VII.

II. MULTIPHYSICS OPTIMIZATION: MOTIVATION

To design antennas that accurately ablate a specific target region, it is important to quantify and understand the relation between the applied electric field and the consequent temperature profile, as well as the fundamental limits on the ability to control and shape these profiles. To this end, and to motivate our investigation into multiphysics methods, a study was conducted to analyze the characteristics of temperature diffusion as a function of the applied electric field profile.

The relation between an impressed external electric field $E$ and the consequent temperature profile $T$ in tissue is given by the Pennes Bioheat Equation [14] as

$$\rho C_p \frac{\partial T}{\partial t} = \frac{1}{2} \sigma |E|^2 + \nabla \cdot (k \nabla T) + W_b C_b (T_b - T)$$ (1)

where $\rho$, $C_p$, $k$, and $\sigma$ are the density, heat capacity, thermal conductivity, and electrical conductivity of tissue, and $W_b$ and $C_b$ are the perfusion rate and heat capacity of blood. These terms account for the cooling effect of blood perfusion through tissue via the last term on the right-hand side of (1). It is clear that the temperature profile depends on both, the applied electric field, and heat diffusion in the tissue, which in turn depends on the thermal conductivity. The purpose of this initial study was to, through simulations, understand and quantify the extent to which each of these factors contributes toward the temperature distribution.

In this study, a dipole antenna array with $5 \times 5$ elements placed near human liver tissue was simulated to focus electric

![Diagram](image_url)
fields within the tissue to varying degrees, and the resulting temperature profiles were computed using (1). The computations were carried out using the commercial finite-element method package, COMSOL Multiphysics (COMSOL Inc., Burlington, MA, USA). The excitations of the array elements were chosen to generate focused electric fields with varying main lobe beam widths.

The maximum diameter of the 50 °C isotherm at steady state is reported in Fig. 2 as a function of the maximum electric field beam width at −6 dB, for different thermal conductivities of tissue, in order to understand the contributions of deposited electric energy and heat diffusion to the obtained temperature profiles. Fig. 3 shows examples of a slice of the field and temperature profiles obtained for a particular value of thermal conductivity. Following two main conclusions can be drawn from these results.

1) There is a fundamental limit to which the temperature profile can be focused or confined, regardless of how focused the electric field is, for a given amount of energy deposited. This limit arises due to heat diffusion around the area where the electric field is confined, and must be taken into account when attempting to ablate a given target region.

2) The 50 °C isotherm diameter is inversely related to the thermal conductivity. This can be understood as follows: although a lower thermal conductivity implies less heat diffusion, this reduced diffusion causes an accumulation of heat near the source, thus raising the overall temperature around the source. Given that biological tissue may be inhomogeneous, and that tumours can exhibit thermal conductivities ranging from 0.3 to 0.7 W · m⁻¹ · K⁻¹ [15], [16], the significant relation between temperature and thermal conductivity must be taken into account in designing antennas for ablation.

The observations made previously clearly demonstrate the complex relation between applied fields and consequent temperature profiles, and indicate the necessity for taking these complexities into account by using a multiphysics approach in designing MWA antennas.

### III. MWA Antenna Optimization: Standard and Multiphysics Approach

The inverse method of designing MWA antennas presented here is based on the idea that, to synthesize a particular temperature profile, an appropriate electric field profile must be generated, by synthesizing an appropriate source current distribution. The current distribution on an ablation antenna can be thought of as a linear array of discrete current elements with magnitudes $I_i$ and phases $\beta_i$, where $i$ is an index to each element. Our aim is to optimize $I_i$ and $\beta_i$ such that they produce an electric field profile $E$, and consequently, a temperature

![Fig. 2. Temperature confinement as a function of electric field beam width and thermal conductivity.](image1)

![Fig. 3. (a) Electric field (dB) and (b) corresponding temperature (°C) profile on a 2-D slice through the domain, for a particular value of thermal conductivity.](image2)
profile $T$, that best approximates a given target temperature profile, $T_{\text{target}}$.

As mentioned previously, several existing ablation antennas are designed with the aim of generating a particular electric field or SAR pattern. In this paper, we consider a multiphysics approach that accounts for discontinuous thermal conductivities. In optimizing the current elements, we compare two optimization approaches: an electric field-based approach where the elements are optimized such that the SAR is confined to within the target (SAR-based optimization), and a multiphysics approach based on synthesizing $T_{\text{target}}$ directly ($T$-based optimization).

A. SAR-Based Optimization

In this approach, a locally high electric field and thus SAR is generated within the target, such that energy deposition is minimized outside the target. The SAR is defined as follows:

$$\text{SAR} = \frac{\sigma |E|^2}{2\rho}.$$  \hspace{1cm} (2)

Since we assume no prior information about the actual field intensities required to induce ablative temperatures, the optimization is performed in two steps.

1) First, we consider normalized SAR only, and attempt to shape the pattern such that the $-6$-dB SAR isoline lies within the target region, and decays below $-10$ dB near the target boundary. Based on these constraints, we define a normalized target SAR distribution, $SAR_{\text{norm}}$, $f^{(i)}$ and $\beta^{(i)}$ are then optimized by minimizing the following objective function:

$$f_E = \sum_{m,n\in K}(SAR_{\text{norm}}^{m,n} - SAR_{\text{norm}}^{m,n}_{\text{target}})^2$$  \hspace{1cm} (3)

where $K$ is the set of indices over which the optimization is performed and $SAR_{\text{norm}}^{m,n}$ is the normalized SAR distribution.

2) At this point, an appropriate scaling of the optimized current magnitudes is required to induce ablative temperatures within the target. Thus, we introduce a scaling factor $b$ that is multiplied with the optimized current magnitudes. The scaling factor itself is optimized such that the resulting temperature at the boundary of the target approaches $50^\circ C$. This is achieved by minimizing the following cost function:

$$f_B = \sum_{m,n\in B}(T^{m,n} - 50)^2$$  \hspace{1cm} (4)

where $B$ is the boundary of the target region. The updated optimized source current magnitudes are $b f^{(i)}$, and the optimized phases are still $\beta^{(i)}$.

B. Multiphysics Approach: Temperature-Based Optimization

In this approach, a target temperature profile, $T_{\text{target}}$, is defined such that the temperature is above $50^\circ C$ within the target region, close to $50^\circ C$ at the target boundary, and decays to normal body temperature ($37^\circ C$) elsewhere. The goal is to optimize the source currents to generate a temperature profile that approaches $T_{\text{target}}$. To achieve this, $f^{(i)}$ and $\beta^{(i)}$ are optimized by minimizing the following objective function:

$$f_T = \sum_{m,n\in K}(T^{m,n} - T_{\text{target}}^{m,n})^2$$  \hspace{1cm} (5)

where $K$ is the set of indices over which the optimization is performed. No assumptions or conditions are placed on the corresponding electric field required to generate $T_{\text{target}}$. This method involves computing first the field profile, and then, the consequent temperature profile via (1) at each iteration of the optimization.

C. Optimization Method Comparison

The optimization methods are compared on the basis of two sets of metrics:

1) Confinement of the $50^\circ C$ Contour: To quantify how closely the $50^\circ C$ contour matches the target boundary, we define three comparison metrics: The relative error norm $f_1$, the Euclidean norm $f_2$, and the infinity norm $f_\infty$ with respect to the achieved and target temperatures at the target boundary:

$$f_1 = \sum_{m,n\in B}|T^{m,n} - T_{\text{target}}^{m,n}|$$  \hspace{1cm} (6a)

$$f_2 = \sum_{m,n\in B}(T^{m,n} - T_{\text{target}}^{m,n})^2$$  \hspace{1cm} (6b)

$$f_\infty = \max_{m,n\in B}|T^{m,n} - T_{\text{target}}^{m,n}|$$  \hspace{1cm} (6c)

where $B$ refers to the target boundary. Lower values of $f_1$, $f_2$, and $f_\infty$ indicate better optimization results.

2) Temperature Gradient at Target Boundary: To quantify the rate at which $T$ drops below ablative levels outside the target, so as to minimize damage to surrounding healthy tissue, we define $g_B$ as the $x$-component of the gradient of the optimized temperature profile, evaluated at a single point on the target boundary. We also compare $g_{\text{max}}$, the maximum of the $x$-component of the gradient of the optimized temperature profile on a line along $\hat{x}$ in the center of the target.

$$g_B = \nabla T.\hat{x} |_{(x_i,z_i)}$$  \hspace{1cm} (7a)

$$g_{\text{max}} = \max_{\text{line}} (\nabla T.\hat{x})$$  \hspace{1cm} (7b)

where $(x_i, z_i) \in B$ and $B$ is the set of points corresponding to the target boundary.

IV. OPTIMIZATION AND COMPUTATION DETAILS

We used COMSOL Multiphysics (COMSOL Inc., Burlington, MA, USA), to solve the forward problem of computing electric fields and temperature profiles, and interfaced this with MATLAB (The MathWorks Inc., Natick, MA, USA) for the optimization procedure and for visualization of results. The electric fields generated by COMSOL were verified by comparisons with homemade finite-differences in frequency-domain (FDFD) code. In both optimization approaches, we use the sequential quadratic programming (SQP) algorithm, which is a standard iterative algorithm for nonlinear optimization problems. A Lagrangian is defined based on the objective function and constraints, and a search direction is computed based on
a subproblem incorporating the Lagrangian [20]. The in-built MATLAB implementation of this algorithm is used here. A frequency of 2.45 GHz was used in all simulations; this is one of the clinical standards for MWA [2].

Electric fields are computed in the frequency domain, and temperatures are computed at steady state. It was found that steady state in the thermal domain is achieved at \( t \approx 4 \text{ min} \), and most ablation treatments last at least that long. Thus, computing temperatures at steady state is justified for the purposes of source optimization. Although tissue properties exhibit nonlinear changes at high temperatures, especially above \( 100^\circ \text{C} \), these effects are not considered in this paper since regions with such high temperatures are relatively small and confined only to the core of the target. Thus, the accuracy of the results obtained should be sufficient for the purposes of demonstrating the inverse design procedure and for comparing optimization methods.

### A. Sources

The tip of the ablation probe is divided into six independent surface current source elements. Each surface current element is a cylindrical surface of radius \( a = 1 \text{ mm} \) and height 2 mm, and all six elements are arranged in a linear array as shown in Fig. 4. The current elements are not included as part of the domains in which fields and temperatures are computed. The surface current density of the \( i \)-th element, \( J_s^{(i)} \), is defined as

\[
J_s^{(i)} = \frac{I^{(i)}}{2\pi a} e^{j \beta^{(i)}} \hat{z}
\]

where \( I^{(i)} \) and \( \beta^{(i)} \) are the magnitudes and phases of each current element as defined earlier. The current elements are encapsulated by an outer layer of teflon (PTFE), a biocompatible material commonly used as the outer sheath in MWA antennas [8], [9].

### B. Simulation Domain

A cylindrical 3-D inhomogeneous domain was used for all simulations, as shown in Fig. 5. The cylinder has a radius of 3.7 cm, and a total height of 5.6 cm. The target region to be ablated is a tumour in human liver, which is approximately a sphere of radius 1 cm. The electrical and thermal material parameters used for all simulations are given in Table I [5], [21]. The domain is terminated on all sides with a perfectly matched layer [22].

Since a 3-D domain is used, the visualization of data is done on a slice-by-slice basis. The electric fields and temperature profiles are interpolated into a 3-D grid of \( 150 \times 150 \times 150 \) cells. In other words, we consider 150 slices of \( 150 \times 150 \) cells.

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**Table I**

<table>
<thead>
<tr>
<th>Material</th>
<th>( \varepsilon_r )</th>
<th>( \mu_r )</th>
<th>( \sigma ) (S/m)</th>
<th>( k_B ) (W/m/K)</th>
<th>( C ) (J/kg/K)</th>
<th>( \rho ) (kg/m(^3))</th>
</tr>
</thead>
<tbody>
<tr>
<td>Liver</td>
<td>43.0</td>
<td>1</td>
<td>1.69</td>
<td>0.52</td>
<td>3540</td>
<td>1079</td>
</tr>
<tr>
<td>Tumour</td>
<td>54.9</td>
<td>1</td>
<td>1.99</td>
<td>0.54</td>
<td>3540</td>
<td>1079</td>
</tr>
<tr>
<td>Teflon Layer</td>
<td>2.1</td>
<td>1</td>
<td>( 1 \times 10^{-25} )</td>
<td>0.27</td>
<td>1143</td>
<td>1806</td>
</tr>
<tr>
<td>Blood</td>
<td>3617</td>
<td></td>
<td></td>
<td></td>
<td>3617</td>
<td>1050</td>
</tr>
</tbody>
</table>
Fig. 6 shows in the $xy$ slice of target profiles for (a) E-based and (c) $T$-based optimization. Right: points over which optimization is performed (in black) for selected $xy$ slices for (b) E-based and (d) $T$-based optimization. Red curves represent target boundary.

in the $xz$-plane. The size of each grid cell is approximately $0.35 \times 0.35 \times 0.37$ mm. This is visually depicted in Fig. 5.

C. Target Profiles

The target field profile for E-based optimization, $E_{\text{target}}$, and the target temperature profile for $T$-based optimization, $T_{\text{target}}$, are defined over the entire domain. However, optimization is only performed over a set of coordinates in the vicinity of the target boundary, described as follows.

1) E-Based Optimization: We define $E_{\text{target}}$ such that it is normalized, and has a value of 1 within the target, and 0 outside. Since a discontinuous field profile is unrealistic in practice, we apply a Gaussian filter of size 41 and a standard deviation of ten cells. This yields a smooth $E_{\text{target}}$ with the $-10$-dB contour near the target boundary. Fig. 6 shows $E_{\text{target}}$ and the points over which optimization is performed. These points are picked concentric to the target, both inside and outside of it.

2) $T$-Based Optimization: In the ideal case, the optimized temperature profile would be at or above 50°C within the target, and would decay to 37°C just outside. The rate at which temperature decays outside the target is a function of diffusion as well as energy deposition via the electric field. In the ideal case, there would be no electromagnetic energy deposition outside the target, and so the temperature profile at and outside the target boundary would be dictated purely by diffusion. Thus, $T_{\text{target}}$ is defined such that it has a value of 50°C at the target boundary, and decays to 37°C outside due to diffusion only, in accordance with (1) with $|E|$ set to 0. Fig. 6 shows $T_{\text{target}}$ and the points over which optimization is performed. These points are picked concentric to the target, at the boundary and outside of it.

V. MWA ANTENNA OPTIMIZATION: RESULTS AND COMPARISONS

The results are presented and discussed here from three points of view: the confinement of the 50°C contour to the target boundary, the $T$ gradients achieved at the target boundary, and the optimized current distributions obtained via each optimization method.

A. Confinement of the 50°C Contour

Fig. 7 is a 3-D depiction of the optimized temperature profiles for selected slices on the $xz$-plane, for both optimization methods. In each case, the 50°C contour is well confined to the boundary of the target at all slices. However, in the $T$-based case the 50°C contour is closer to the desired profile. Table II lists the comparison metrics $f_1$, $f_2$, and $f_\infty$ for each method. As expected, the $T$-based multiphysics method performs better than the purely SAR-based method. In terms of the relative error, Euclidean norm, and infinity norm, the $T$-based method provides a 39.3%, 70.1%, and 67.9% improvement over the SAR-based method.

B. Temperature Gradient at Target Boundary

Fig. 8 shows a surface plot of the optimized $T$ profiles for each method for one $xz$ slice at $y = 0$ cm. It is clear that the temperature falls off with a much sharper gradient in the $T$-based case than in the SAR-based case, and thus, is less likely to cause damage to surrounding healthy tissue. Sharper $T$ gradients allow for more precise targeting. Table II lists the comparison metrics $g_B$ and $g_{\text{max}}$ to quantify these gradients. At the boundary of the target, along the x direction, $T$-based optimization provides a 45.5% sharper gradient. The maximum slope along the x direction is 2.75 times steeper for the $T$-based case.

Fig. 9 shows the normalized SAR profile in decibel for the same slice as aforementioned. It is interesting to note that the electric field is more strongly confined within the target in the $T$-based case than the SAR-based case. This contributes to the fact that sharper $T$ gradients are achieved in the former, since the $T$ profile outside the target is influenced more by diffusion of heat than deposition of energy via the electric field. This is a direct result of the target profiles that were chosen in each optimization method. In fact, the importance of heat diffusion is part of the reason that thermal inhomogeneities in tissue play an important role in the temperature distribution, and must be taken into account in addition to the electric field distribution.

C. Optimized Currents

Fig. 10 shows the optimized current source magnitudes (top row) and corresponding phases (bottom row) for each element, for each optimization method. The optimized phases are fairly constant along the antenna, and the current magnitudes are realistic, as shown in the following section, where their realization is discussed. In terms of computational cost, the SAR-based method took approximately 6 h to converge, while the $T$-based approach took approximately 8 h. The added cost is justified by the significant advantages provided by the $T$-based method.

VI. PHYSICAL IMPLEMENTATION OF OPTIMIZED CURRENT ELEMENTS

In this section, we present a possible implementation of the optimized current distributions obtained in the previous section.
This design is a proof of concept demonstrating that ablation antennas can be designed in an inverse way by synthesizing desired current distributions.

Since the optimized surface current magnitudes have a central peak and a smooth decay on each side, we can implement the distributions via a single cylindrical dipole antenna with a varying radius along its axis. Since the optimized phases are nearly constant, we need only deal with the current magnitudes. The cylinder can be divided into six sections, each section corresponding to one of the sources in the optimization procedure described previously. Based on the continuity of the normal component of the current density at an interface, a simple method to vary the current magnitude is to modulate the cross-sectional area of the antenna. Hence, the radius of each section is chosen to be proportional to the surface current magnitude required in that section. Note that since this is a proof of concept rather than a ready-to-be-manufactured design, the feeding network of the antenna is not studied here. However, a feeding network can quite easily be taken into account by incorporating it in the electric field solver during the optimization process, such that the optimized current distribution already accounts for any coupling effects due to feed lines.

The antenna was simulated inside the same tissue structure that was used for the optimization procedure, to enable direct comparison of the resulting temperature profiles. The antenna cylinder radii are chosen such that the resulting current in each section corresponds to the \( T \)-based optimized currents obtained earlier. The length of each section is equal to the length of each source that was used for optimization. Copper was used as the antenna material. The feed point was placed in the middle of the largest section, with an 250-V voltage source, which provides the correct scaling of currents to match the target currents. The surface currents are sampled at the midpoint of each cylindrical section for comparison with the target currents that we are trying to synthesize.

Fig. 11 shows the structure that was simulated and the resulting surface current distribution along its length. The distribution is a fairly accurate replication of the optimized currents desired in each section. Figs. 12 and 13 show the temperature profile generated by this design. It is clear that the temperature profile is in good agreement with that generated via \( T \)-based optimization of six sources as in Figs. 7(b) and 8(b).

This proof of concept demonstrates that current distributions obtained via optimization can be implemented physically in principle, to generate controllable temperature profiles. The presented design is fairly straightforward, reconfigurable (cylinders of different radii can be assembled in any configuration), and resembles commercial antennas while achieving targeted ablation in a more intelligent and controllable way.

### Table II

**Relative, Euclidean and Infinity Norm Errors for Comparing Optimization Methods**

<table>
<thead>
<tr>
<th></th>
<th>S.A.R.-Based Optimization</th>
<th>( T )-Based Optimization</th>
</tr>
</thead>
<tbody>
<tr>
<td>( f_1 )</td>
<td>( 4.96 \times 10^4 )</td>
<td>( 3.01 \times 10^4 )</td>
</tr>
<tr>
<td>( f_2 )</td>
<td>( 2.30 \times 10^5 )</td>
<td>( 6.87 \times 10^5 )</td>
</tr>
<tr>
<td>( f_\infty )</td>
<td>18.2</td>
<td>5.85</td>
</tr>
<tr>
<td>( g_0 )</td>
<td>30.8 , ^\circ \text{C/cm}</td>
<td>44.8 , ^\circ \text{C/cm}</td>
</tr>
</tbody>
</table>
| \( g_{\text{max}} \) | \( 8.06 \times 10^3 \, ^\circ \text{C/cm} \) | \( 2.22 \times 10^4 \, ^\circ \text{C/cm} \)

From a clinical perspective, an ablation antenna with reconfigurable current element magnitudes and phases can be incorporated into treatment planning as follows.

1) A stack of images of the target region can be obtained via MRI, PET, etc., and the different tissues and structures are identified by contouring techniques.
2) The electrical and thermal properties of the tissues and structures are obtained as documented in the literature.
3) The image stack is taken as the geometry to be used in the optimization procedure described previously, with the current sources positioned appropriately inside the target region.
4) A target temperature profile is created with respect to the image stack, such that ablative temperatures are achieved within the target.

### VII. CLINICAL INTEGRATION

From a clinical perspective, an ablation antenna with reconfigurable current element magnitudes and phases can be incorporated into treatment planning as follows.

1) A stack of images of the target region can be obtained via MRI, PET, etc., and the different tissues and structures are identified by contouring techniques.
2) The electrical and thermal properties of the tissues and structures are obtained as documented in the literature.
3) The image stack is taken as the geometry to be used in the optimization procedure described previously, with the current sources positioned appropriately inside the target region.
4) A target temperature profile is created with respect to the image stack, such that ablative temperatures are achieved within the target.
Fig. 8. Surface plot of $T$ for an $xz$-slice at $y \approx 0$ cm. (a) SAR-based optimization, and (b) $T$-based optimization. Target region boundary shown in red. Empty region in the center represents position of source array.

Fig. 9. Surface plot of normalized $|E|$ (in dB) for an $xz$-slice at $y \approx 0$ cm. (a) SAR-based optimization, and (b) $T$-based optimization. Target region boundary shown in red. Empty region in the center represents position of source array.

Fig. 10. Top row: optimized current source element magnitudes. (a) SAR-based optimization. (b) $T$-based optimization. Bottom row: optimized current source element phases. (d) SAR-based optimization. (e) $T$-based optimization.
5) The current sources are optimized using the multiphysics techniques described previously.

6) Once the optimal current magnitudes and phases are obtained, the current elements on the antenna are configured or programmed accordingly (depending on the physical design of the antenna); the antenna is then ready for clinical use.

Reconfigurable ablation antennas designed in this way would allow for more predictable, controllable and customizable ablation of targets while reducing damage to surrounding healthy tissue. These antennas can also be more versatile in their applications.

VIII. CONCLUSION

We have presented an inverse approach to designing antennas for interstitial MWA via multiphysics optimization. Desired temperature profiles are synthesized by optimizing an array of current source elements. It was shown that a multiphysics approach is necessary to account for thermal inhomogeneities in tissue when optimizing the design of MWA antennas. A purely electric field-based optimization method was compared quantitatively to a multiphysics temperature-based method in a simulated 3-D test case with a target in human liver. It was found that the multiphysics method provides a significant improvement over the electric field-based method in confining ablative temperatures to within the target, and allows for significantly steeper temperature gradients, which could lead to reduced damage to surrounding healthy tissue. A proof of concept design was presented to show that the optimized current distributions can be physically implemented in a simple way. This optimization-based source design approach can be incorporated quite easily into treatment planning in a clinical setting, and allows for more controllable ablation for multiple applications.

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