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A Review of Coaxial-Based Interstitial Antennas for Hepatic Microwave Ablation

John M. Bertram,1 Deshan Yang,2 Mark C. Converse,3 John G. Webster,1 & David M. Mahvi3

1Department of Biomedical Engineering; 2Department of Electrical and Computer Engineering; 3Department of Surgery, University of Wisconsin, Madison, WI 53792 USA

Address all correspondence to John G. Webster, Department of Biomedical Engineering, 1550 Engineering Drive, University of Wisconsin, Madison, WI 53706 USA; Fax: 608-265-9239; webster@engr.wisc.edu

ABSTRACT: Although surgical resection remains the gold standard for treatment of liver cancer, there is a growing need for alternative therapies. Microwave ablation (MWA) is an experimental procedure that has shown great promise for the treatment of unresectable tumors and exhibits many advantages over other alternatives to resection, such as radiofrequency ablation and cryoablation. However, the antennas used to deliver microwave power largely govern the effectiveness of MWA. Research has focused on coaxial-based interstitial antennas that can be classified as one of three types (dipole, slot, or monopole). Choked versions of these antennas have also been developed, which can produce localized power deposition in tissue and are ideal for the treatment of deep-seated hepatic tumors.

KEYWORDS: microwave ablation, hepatic, resection, coaxial, liver, liver cancer, antenna, probe, interstitial, hyperthermia

I. INTRODUCTION

Liver cancer is a dangerous condition that is caused by hepatocellular carcinoma (HCC) or colorectal metastases and has a mortality rate of 100% in untreated cases at five years.1 HCC, or primary liver cancer, accounts for ~ 4.1% of all cancers and is the leading cause of liver cancer, with ~ 437,000 new cases per year in 1990 and an estimated annual mortality of one million people worldwide.2–5 Colorectal carcinoma, the second leading cause of cancer death in men age 40–79 years, was diagnosed in 150,000 new cases and responsible for 57,000 deaths in the United States in 2002, with a majority occurring from liver metastases.5,6
Several treatment options exist for patients diagnosed with liver cancer. Surgical resection is currently and has long been considered the “gold standard” for treating the disease, as negative tumor margins can be achieved in ~ 80–90% of patients. However, only 5–10% of patients suffering from liver cancer can be treated by surgical resection due to the surgical risk associated with multiple tumors, poorly defined tumor margins, tumors in unresectable locations, and insufficient hepatic reserve. In addition, postoperative complication rates following resection can be as high as 46% in patients with cirrhosis.

These risks and the low number of resection candidates have resulted in an increased need for alternative therapies. The most frequently used alternative to surgical resection is radiofrequency (RF) ablation, a technique in which 150–200 W power at frequencies between 460 and 500 kHz is applied to probes inserted in cancerous tissue. This results in heating by current conduction and ionic agitation, which subsequently induce cell necrosis by causing denaturation of intracellular proteins and lysis of tumor cell membranes in regions above 45–50°C. RF ablation can be performed percutaneously, laparoscopically, or during surgery and can be controlled by applying constant power to the probes, maintaining a constant probe tip temperature, or keeping the system impedance below a certain level. Although lesion size can be influenced by several factors, including probe placement and the proximity of large blood vessels that act as a heat sink, the procedure has been shown to be a viable alternative to resection, with local recurrence occurring in only 10–20% of cases. Furthermore, chemotherapeutic agents, internally cooled probes, and interstitially infused saline have all been effective in increasing the lesion area produced by RF probes. The Pringle maneuver, a technique in which the hepatic artery and portal vein are occluded, has also been shown to increase lesion size. In addition to the treatment of liver tumors, RF ablation has also been widely used as a treatment for cardiac abnormalities such as supraventricular arrhythmias and ventricular tachycardia, and has gained international approval for the treatment of benign prostate hyperplasia and endometrial ablation.

Another alternative to resection is cryoablation, a technique in which liquid-nitrogen-cooled probes are used to induce cell necrosis through intracellular and extracellular freezing in tissue. Compared to RF ablation, cryoablation can create larger lesions and is more effective in treating tumors in multiple lobes, with similar recurrence rates of 10–15%. It also does not cause vessel walls to lose their elasticity or completely denature proteins, which subsequently allows for nerve regeneration. However, lesions formed by cryoablation are also significantly affected by blood flow when hepatic inflow occlusion is not performed and the procedure can lead to complications, such as hemorrhage, bile collection and biliary fistula, liver abscess, renal failure, and the onset of cardiac arrhythmias. In addition, cryoablation has been shown to cause multiple systematic organ failure in rare cases as a result of cytokine release, which has severely reduced its clinical use.
Because of the limitations and complications posed by these procedures, a new alternative to resection known as microwave ablation (MWA) is being developed for the treatment of liver tumors. The technique is similar to RF ablation, as it uses a microwave generator to heat tissue percutaneously or during surgery to induce cell necrosis above 50°C. However unlike RF ablation, heating is performed through the use of thin interstitial antennas that radiate electromagnetic power at microwave frequencies, usually 915 MHz or 2.45 GHz. Heating is governed by the oscillation of polar molecules in the induced electromagnetic field, which results in faster tissue heating and subsequently allows for shorter treatment times. This heating is determined mainly by power deposition in tissue, often expressed as specific absorption rate (SAR), but is also dependent on both the dielectric and thermal properties of the tissue being ablated.

Unlike other treatments for liver cancer, MWA has been shown experimentally to be capable of forming large lesions in the presence of blood perfusion, which makes it ideal for the treatment of vascularized tissue. In addition, MWA systems are not restricted by tissue charring, which makes them capable of heating tissue to higher temperatures than those achieved during RF ablation. Although MWA has not yet been approved for commercial use in the United States, a recently published international study in which MWA was performed on patients with HCC showed one-, two-, and three-year survival rates of 96%, 83%, and 73%, respectively. In contrast, a recent study that examined the survival rates of patients with recurrent HCC treated by RF ablation showed lower respective one-, two-, and three-year survival rates of 82%, 72%, and 54%. The many perceived advantages of MWA have driven researchers to develop innovative antennas to effectively treat deep-seated, nonresectable hepatic tumors. Although different antennas have been proposed for MWA, including the helical antenna, etc., research has primarily focused on thin, coaxial-feedline-based interstitial antennas. With recently proposed improvements, such as the cap-choke design, etc., these antennas are minimally invasive, capable of delivering a large amount of electromagnetic power in more localized patterns, and are more suitable for various MWA operations than other antenna designs. These coaxial-based antennas can usually be classified as one of three types (dipole, slot, or monopole) based on their physical features and radiative properties. This paper presents an overview of each of these types, as well as a review of current, more advanced interstitial antennas.

II. DIPOLE ANTENNAS

Figure 1(a) shows the schematic for one of the original types of antennas used in MWA and microwave hyperthermia applications, the coaxial-fed interstitial dipole antenna. This antenna is usually constructed from thin, semirigid coaxial cable, with its design focusing on three regions. The first region is often referred to as the gap or junction of the antenna, which acts as the effective source of
electromagnetic wave propagation. This region is usually designed to be much less than a wavelength so that the gap can be approximated as an infinitesimal dipole. The other regions of the antenna are often referred to as the extensor and insertion regions, and are represented in Fig. 1(a) by the metal segment of length \( h_A \) at the distal end of the antenna and the insertion depth \( h_B \) of the antenna in tissue, respectively. Effective dipole length \( L \) is equal to \( h_A + h_B \), as gap length can usually be assumed to be negligible.\(^{33}\) For ablation or hyperthermia treatments, the antenna is normally placed in a catheter to allow easy insertion and removal from the tissue, provide mechanical stability, and improve the effective coupling to tissue.

Figure 1(b) depicts the near-field analysis model for an insulated coaxial-fed interstitial dipole antenna.\(^{34}\) In this model, the current distribution of the insulated dipole antenna in a lossy medium, such as liver, can be written as

\[
I(z') = I(0) \frac{\sin[\beta_L (h - z')] \sin(\beta_L h)}{\sin(\beta_L h)} \quad [\text{mA}]
\]

where \( I(0) \) is equal to the product of excitation voltage \( V^e_0 \) and antenna admittance \( Y = 1/Z \), \( h \) is the half-length of the antenna, and \( \beta_L \) is the complex wave number, assuming time dependence for all fields. This current relation is used to determine the nonzero axial and radial components of the electric field, which can be subsequently used to calculate specific absorption rate (SAR), a critical metric for analyzing the performance of interstitial antennas. SAR represents the amount of time average power deposited per unit mass of tissue (W/kg) at any position \( \rho \). It can be expressed mathematically as

\[
\text{SAR}(\rho) = \frac{\sigma}{2\rho} |\bar{E}(\rho)|^2 \quad [\text{W/kg}]
\]

where \( \sigma \) is tissue conductivity (S/m), \( \rho \) is tissue density (kg/m\(^3\)), and \( |\bar{E}(\rho)| \) is the magnitude of the local electric field vector at \( \rho \).\(^{35}\) As the goal of hepatic MWA is to provide highly controlled heating in the region encompassing the tumor, SAR distributions that are nonlocalized near the antenna tip reduce treatment effectiveness. These nonlocalized distributions are usually asymmetric and extend along the entire insertion region of the antenna. They are due to the propagation of the wave along the outside of the applicator back toward the feed point. The outside of the outer conductor, a layer of Teflon\(^{\text{TM}}\) and then the conductive tissue can be thought of as a lossy wave guide allowing propagation back along the feed line. This is often referred to as backward heating and causes detrimental tissue heating along the antenna insertion region. In addition to reducing lesion size, backward heating can cause excess patient discomfort, induce necrosis in normal liver tissue, and lead to surgical complications. It is desirable to design a coaxial antenna with SAR distribution maximized near the antenna tip and
minimized at its extension to the insertion region, in order to reduce the backward heating.

Experiments and mathematical models have shown that lesions produced by the interstitial coaxial-based dipole antenna are highly dependent on insertion

---

**FIGURE 1.** (A) Basic structure of a coaxial-fed dipole antenna (from Ref. 32.) and (B) analysis model for an insulated dipole in a lossy medium (from Ref. 34).
depth.\textsuperscript{32,36,37} Figure 2 clearly shows this relationship, which can be explained from transmission line theory. The outer conductor, surrounding dielectric of the catheter, and conductive tissue can be thought of as a lossy transmission line. Then, using the knowledge that the input impedance of a transmission line is equal to

\[ Z_{in} = jZ_0 \tan(\beta h + j\Theta_h) \quad [\Omega] \]

where \( Z_0 \) is the characteristic impedance of the segment, \( \beta \) is the wave number, and \( h \) is segment length with the terminal function \( \Theta_h = 0 \) for an open-ended segment and \( \Theta_h = -j\pi/2 \) for a short-circuited segment, the impedance of the segment above and below the gap/slot can be determined. From this expression, the total impedance of a coaxial-fed interstitial dipole antenna can be written as the sum of the individual input impedances of segments A and B

\[ Z_d = Z_{inA} + Z_{inB} \quad [\Omega] \]

where \( Z_{inA} \) and \( Z_{inB} \) correspond to the extensor region \( h_A \) and insertion region \( h_B \) shown in Fig. 1 and \( Z_d \) is the impedance seen at the gap.\textsuperscript{32} A symmetric dipole (\( Z_{inA} = Z_{inB} \)) with segment lengths \( h_A = h_B = \lambda_{eff}/4 \) yields excellent matching to the 50 \( \Omega \) feed line and good power transfer,\textsuperscript{38} where the effective wavelength in tissue \( \lambda_{eff} \) was calculated using

\[ \lambda_{eff} = \frac{c}{f\sqrt{\varepsilon_r}} \quad [\text{m}] \]

where \( \varepsilon_r \) is the relative permittivity of liver tissue, \( c \) is the speed of light (m/s) in free space, and \( f \) is the operating frequency (Hz) of the microwave source.\textsuperscript{39} Since current flows from the inner to outer conductor using the tissue as a conducting medium, both the current and SAR distributions will be highly dependent on insertion depth.

Insertion depth has also been shown to affect the frequency dependent reflection coefficient of interstitial dipole antennas.\textsuperscript{40} This quantity is often used to evaluate antenna performance and can be expressed logarithmically as

\[ \Gamma(f) = 10 \cdot \log_{10} \left( \frac{P_r(f)}{P_{in}} \right) \quad [\text{dB}] \]

where \( P_r \) and \( P_{in} \) are reflected and input power (W), respectively. Low reflection coefficients indicate good matching and excellent power deposition in tissue. In contrast, high reflection coefficients are often caused by an impedance mismatch between the antenna and tissue, which prevents power from entering the tissue. Since an antenna can be treated as the load on the end of the coaxial transmission feed line, a mismatch between this load and the 50 \( \Omega \) line leads to power being
FIGURE 2. Normalized specific absorption rate (SAR) patterns for a conventional interstitial dipole antenna at insertion depths of (A) 75 mm, (B) 95 mm, (C) 115 mm. Measurements were performed in a brain tissue phantom (adapted from Ref. 32).
reflected back to the generator and a standing wave in the feed line. This standing wave causes additional heating (due to greater currents causing resistive losses in the nonideal conductor, as well as losses in the nonideal dielectric) in the feed line. This higher temperature can, due to the thin outer conductor, transfer heat to the surrounding tissue or damage the feed line itself, resulting in antenna failure. Whenever possible, antennas should be operated at the frequency where their reflection coefficient is minimal. This is often defined as the resonant frequency of the antenna.

To eliminate the dependence of insertion depth on interstitial dipole antennas, a modified dipole was developed\(^{32}\) that added a \(\pi/2\) transformer \((\lambda_{\text{eff}} / 4)\) to the

![Diagram of coaxial-fed slot antenna](image)

**FIGURE 3.** (A) Basic structure of coaxial-fed slot antenna (adapted from Ref. 42) and (B) analysis model (from Ref. 43).
insertion region $h_R$. This transformer is often referred to as a choke and will be discussed in more detail later in this paper.

III. SLOT ANTENNAS

Figure 3 shows the general structure of one of the most popular designs for hepatic MWA and microwave hyperthermia, the coaxial-fed interstitial slot antenna. Physical construction of the slot antenna is straightforward. The antenna is fabricated from thin, semirigid coaxial cable in which a small ring slot of width $W_d$ is cut through the outer conductor close to the short-circuited distal tip of the antenna to allow electromagnetic wave propagation into the tissue. This width is usually chosen to be much smaller than a wavelength, which allows the source to be replaced by a narrow strip of magnetic current using equivalence in analytical models. As with coaxial-fed interstitial dipoles, the slot antenna is usually placed in a low permittivity catheter ($\varepsilon_r = 2.6$) to provide physical protection during insertion. Also the catheter tends to improve the electromagnetic coupling (which leads to more efficient power deposition in the tissue) with surrounding tissue. An antenna without a catheter can be used, but one loses the physical protection of the catheter.

The current distribution for slot antennas used in MWA can be determined by first recalling that the inner and outer conductors of the coaxial cable are short-circuited at the distal tip, with the radius of the inner conductor assumed to be much less than that of the outer conductor. This allows for the assumption of a uniform circumferential surface current with a line current along the antenna.

![FIGURE 4](image_url) Different coaxial-fed slot antenna configurations. (A) No layer between antenna and catheter (type A). (B) Thin air layer (type B). (C) Thin dielectric layer (type C). (D) Dielectric load near tip (type D). All units in millimeters (from Ref. 45).
Current distribution along the antenna can be subsequently determined using Method of Moments (MoM) using a series of expansion functions,
\cite{43,44} or calculated using transmission line theory.\cite{38} This current distribution can be extended to yield the radiative and power deposition patterns of the antenna.\cite{43} In a recent study that examined the effects of insertion depth and catheter thickness on the current distributions of coaxial-fed interstitial slot antennas,\cite{44} increasing the insertion depth was found to dramatically affect the current distribution, making it highly asymmetric. This indicates that power deposition is also highly asymmetric and that SAR patterns produced by these antennas become more nonlocalized with increasing insertion depth. Increases in catheter thickness were shown to decrease peak current and yield a more uniform current distribution. Again, this corresponds to decreased power localization and nonlocalized SAR patterns.

Figure 4 shows several slot antenna configurations that were examined to improve power localization.\cite{45} In these configurations, the dielectric material between a slot antenna of radius 1.8 mm and catheter was modified to alter electromagnetic wave propagation. Figure 5 shows the axial SAR patterns in tissue obtained by modeling each of these configurations. These results indicate that power localization can be improved by using a dielectric load of high permittivity.
ity (\(\varepsilon_r = 20\)) near the applicator tip. In contrast, a dielectric load with low permittivity, such as air (\(\varepsilon_r = 1\)), was found to have the opposite effect and effectively resulted in an almost uniform SAR distribution along the axial length of the antenna.

Although the basic structure of the slot antenna appears very similar to that of the interstitial dipole shown in Fig. 1, important differences exist between the two designs. Unlike the dipole, the physical structure of the coaxial cable is preserved for the entire length of the slot antenna, except for the ring slot and short-circuited tip. The presence of a dielectric slot rather than a gap region also makes the design more mechanically stable and easier to construct. Another key advantage is that multiple slots can be used,\textsuperscript{43,45,46} which allows for designs with improved power localization.

### IV. MONOPOLE ANTENNAS

Another antenna that has been widely used in ablation is the coaxial-fed interstitial monopole antenna, which like the dipole and slot antennas, can be easily constructed from thin, semirigid coaxial cable. Although these antennas are most frequently used for the treatment of cardiac arrhythmias,\textsuperscript{47–50} their small size additionally makes them well-suited for hepatic MWA.

Figure 6 shows the several common variations of the monopole antenna.\textsuperscript{48} The most basic of these, the open-tip monopole (OTM), is characterized by an elongated inner conductor that is radially surrounded by dielectric material and open-ended at the distal tip. Another variation, the dielectric-tip monopole (DTM), differs from the OTM in that dielectric material surrounds the elongated

![FIGURE 6. The three basic classifications of monopole antennas (adapted from Ref. 48).](image-url)
inner conductor both radially and at the antenna tip. The final variation, the metal-tip monopole (MTM), uses a metal cap at the distal end of an extended dielectric-covered inner conductor to provide increased electric contact with tissue.

For all three variations, excellent power deposition occurs if the length of elongated conductor is $\lambda_{\text{eff}} / 4$, where $\lambda_{\text{eff}}$ represents the previously described

![Diagram](image)

**FIGURE 7.** Comparison of coaxial-based open-tip monopole (OTM) and metal-tip monopole (MTM) antennas. (A) 2D Normalized SAR profile. (B) return loss. The OTM is represented by the solid line; the MTM is represented by the dashed line. The metal cap shifts the resonant frequency of the MTM antenna (from Ref. 41).
effective wavelength in tissue.\textsuperscript{41} However, a computational and experimental analysis of these designs has shown that the MTM antenna is capable of yielding the greatest power deposition at the tip of the antenna.\textsuperscript{48} Figure 7A shows this relation, which can be observed from direct comparison of the normalized SAR contours of monopole antennas. Figure 7B shows that in addition to providing increased electric contact, the metal cap of the interstitial MTM also shifts the antenna’s resonant frequency.\textsuperscript{41} Despite these advantages, note that like other monopole antennas, the interstitial MTM’s axial SAR distribution is relatively uniform. As described previously, this is indicative of backward current flow.

\textbf{FIGURE 8.} Cap-choke antenna for microwave ablation, designed for operation at 2.45 GHz: (A) Basic structure and (B) comparison of simulated and measured normalized axial SAR at different radial distances (from Ref. 57).
V. CHOKED ANTENNAS

Studies have shown that one practical and effective solution to the backward heating problem that affects interstitial antennas is to electrically connect a thin metallic choke, usually \( \lambda_{\text{eff}} / 4 \) in length, to the antenna’s outer conductor to block axial current flow and localize power deposition near the distal tip of the antenna.\(^{51,52} \) As a result, properly designed choked antennas are capable of achieving highly localized SAR patterns that are less dependent on insertion depth, although such antennas are usually more invasive due to their slightly increased diameter. Chokes have also been found to aid in impedance matching and tissue coupling during MWA.\(^{53,54} \)

Current research efforts that have focused on antenna design for hepatic MWA have widely incorporated the choke into the design of their interstitial antennas. Figure 8A shows one of these designs, the cap-choke slot antenna.\(^{54-57} \) In addition to its effectively implemented choke, this innovative design uses an annular cap that is short-circuited across the inner and outer conductors of the coaxial cable and extends radially from the antenna to increase capacitance and improve radiation from the tip of the antenna. Although the cap-choke is not as minimally invasive as other antennas, it provides excellent localization of power at the distal end of the antenna. Figure 8B shows that the axial SAR pattern of this antenna is definitely more localized than those of coaxial-fed slot antennas, which were presented in Fig. 5. In addition, Fig. 8B indicates that lesions produced using this antenna are much more independent of insertion depth, a conclusion that we have confirmed through independent experimental findings. Although the antenna has been designed for operation at both 915 MHz and 2.45 GHz, 915 MHz is usually preferred since it can yield a deeper power deposition.\(^{54} \)

Figure 9 shows a recent modification of the cap-choke slot antenna, in which the choke was extended over an added second slot to produce more uniform SAR near the applicator tip in a brain equivalent phantom.\(^{58,59} \) In addition, the diameter of the antenna was reduced to make it less invasive. Through numerical techniques, it was demonstrated that this modified cap-choke slot antenna is capable of ablating tumors up to 2 cm in radius. Arrays of three antennas were also considered and found to be advantageous in cases where a single antenna was unable to treat the target region.\(^{59} \)

Although the basic coaxial-fed, interstitial MTM antenna has been found to yield more uniform power deposition in the tip region compared with other monopole configurations, it is difficult to impedance match and results in significant reflected power.\(^{60} \) To circumvent this problem, Chiu et al.\(^{60} \) developed a novel choked MTM design commonly referred to as the expanded-tip wire (ETW) antenna. Figure 10A shows the design of the ETW antenna, which can be expanded by both length and width dynamically to change antenna impedance. Figure 10B shows the normalized SAR pattern of the ETW antenna, which indi-
cates that the lesions produced by this antenna are quite localized. Results also show a strong correlation between the antenna dimensions and frequency-dependent reflection coefficient.

Longo et al.\textsuperscript{53} recently developed another innovative monopole design, an OTM with adjustable choke that allows for the real-time adjustment of antenna

\begin{figure}[h]
\centering
\includegraphics[width=0.8\textwidth]{figure9.png}
\caption{Modified cap-choke antenna for microwave ablation for operation at 2.45 GHz: (A) Basic structure and (B) SAR profile (from Ref. 59).}
\end{figure}
impedance for better matching. This antenna uses a modified biopsy needle to guide the antenna into tissue while simultaneously serving as an adjustable choke. This was implemented by soldering a copper collar to the outer conductor of the coaxial cable, which makes sliding contact with the biopsy needle. As a result, this antenna is less invasive than earlier monopole designs and has a total applicator diameter of just over 2 mm, where antenna dimensions were chosen based on the operational frequency of 2.45 GHz. The reflection coefficient of the antenna was also measured over a frequency range of 1.5–3 GHz, and did not exceed −10 dB between 1.7 and 3 GHz. These measurements were done on an antenna with a fixed choke length of $\lambda_{\text{eff}} / 4$, although this was found to have a minimal effect on performance.

Nevels et al. performed a detailed examination of choked monopoles that compared the normalized SAR patterns of choked OTM and MTM coaxial-fed, interstitial antennas. Figures 11A and 11B indicate that the choked MTM is slightly more localized. In addition, it was shown that while reversing the aper

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**FIGURE 10.** Extended-Tip Wire (ETW) antenna for microwave ablation: (A) Basic schematic and (B) Normalized SAR pattern for the antenna (from Ref. 60).
ture direction of a choke can cause a large increase in the antenna’s resonant frequency, it does not have a significant effect on SAR. The effects of replacing the dielectric surrounding the elongated inner conductor of the MTM with one of higher relative permittivity \( \varepsilon_r = 12 \) were also examined, in which relative permittivity of the new dielectric was chosen based on a first-order approximation of a quarter-wave transformer.

**FIGURE 11.** Comparison of normalized SAR profiles for choked monopole antennas: (A) OTM, (B) MTM, and (C) MTM with dielectric plug (from Ref. 41).
where $\varepsilon_1$ is the dielectric of the coaxial cable, $\varepsilon_2$ is the replaced dielectric, and $\varepsilon_3$ is the tissue permittivity. Replacement of the dielectric was found to provide a better impedance match between the antenna and tissue, due to the shortened wavelength in the dielectric plug. Figure 11C shows that this also resulted in slightly increased radial power deposition and slightly decreased power deposition at the antenna tip. The dielectric plug also served to increase the antenna’s bandwidth from 0.8 to 1.2 GHz.

**FIGURE 12.** Triaxial choked interstitial dipole antenna for localized power deposition: (A) Axially symmetric model and (B) measured SAR distribution. All units in millimeters (from Ref. 40).
Coaxial-fed interstitial dipole antennas have also been modified to include chokes. Figure 12 shows the design of a triaxial choked dipole designed by Schaller et al.,\textsuperscript{40} that as with other choked antennas, is capable of producing a highly localized SAR distribution. A choke length of $\lambda_{\text{eff}} / 4$ significantly reduced the dependence of SAR on insertion depth, although increased insertion depths were still found to decrease the resonant frequency of the antenna. In a subsequent study, Wu and Lu,\textsuperscript{61} found that by shortening the extensor region $h_A$ of the triaxial choked dipole, the SAR pattern could be shifted toward the distal tip of the antenna. Optimal power deposition was found to occur for $h_A / h_B < 0.25$, where $h_B$ is insertion depth, due to increased charge density near the antenna tip. Another choked dipole developed by Saito et al.\textsuperscript{62} used heat-shrink tubing as a dielectric between the coaxial cable and choke to provide better power localization than earlier interstitial dipole antennas.

VI. ADDITIONAL DESIGNS

Although choked interstitial antennas have become popular for localized power deposition, other designs have been developed that can achieve similar performance without the need for a choke. Figure 13 shows a recently designed double-
slot antenna by Saito et al.\textsuperscript{63} that is capable of achieving much greater localization than single-slot antennas and is not affected by insertion depth. Figure 14 shows a normalized axial SAR comparison of these two antennas and indicates that backward heating could be further reduced using additional slots. In an earlier study by Saito et al.\textsuperscript{42}, two standard coaxial-fed slot antennas were used in a tip-split array. Figure 15 shows this configuration, which produced an approximately triangular SAR pattern that yielded more heating than a single-slot antenna, where power deposition was explained largely by the overlap of surface currents from each antenna. Another design shown to be capable of improving localization is the sleeved slot antenna,\textsuperscript{64,65} in which the sleeve consists of a thin ring of metal several effective wavelengths in length that is placed around the antenna proximal to the slot. Another similar design is the sleeve antenna, which also consists of a floating sleeve whose ideal length is approximately 1/2 wavelength in tissue.\textsuperscript{66} These designs are similar to a choked antenna, but are fundamentally unique as the sleeve is not electrically connected to the antenna. Instead, a dielectric, such as Teflon, is used to hold the sleeve in place.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{figure14.png}
\caption{Normalized axial SAR distributions for (A) single-slot antenna and (B) double-slot antenna. Air-tissue interface corresponds to \( z = 0 \); \( D_i \) corresponds to insertion depth. All units are in millimeters (from Ref. 63).}
\end{figure}
Because of current interest in minimally invasive surgical procedures, Brace et al. developed a triaxial OTM antenna that is less invasive than choked antennas. Although this antenna is very similar in construction to the choked monopole antenna developed by Longo et al., it appears that the outer conductor of the triaxial OTM is not connected to the biopsy needle. To minimize the reflection coefficient at the operational frequency of 2.45 GHz, the effective source of the antenna should be placed \((2n - 1)\lambda_{\text{eff}} / 4\) from the tip of the biopsy needle, where \(n\) is an integer. Although this antenna is highly resonant, mathematical modeling and experimental tests of the design in ex vivo bovine liver tissue show normalized SAR is dependent on insertion depth and comparable to that of a basic slot antenna.

![Diagram of Coaxial Slot Tip-Split Array Applicator](image)

**FIGURE 15.** Coaxial slot tip-split array applicator: (A) Basic structure, (B) calculated SAR profile, and (C) axial profile along SAR observation line (from Ref. 42).
VII. CONCLUSIONS

In the last ten years, great progress has been made in the development of less invasive interstitial antennas for microwave ablation that are capable of producing highly localized patterns of electromagnetic power deposition in tissue. This has been accomplished largely through the introduction of choked antennas, although some researchers have been able to implement alternative designs, such as those using multiple slots, which can achieve similar levels of performance. Although it has not been discussed in significant detail in this paper, research has also shown that arrays of multiple antennas can overcome many of the performance limitations of a single antenna and have been proven to be effective in trial studies for the treatment of abnormally large or geometrically unique tumors.

However, one problem that still remains is that the evaluation of current antennas is still largely rooted in electromagnetic performance metrics, such as localized power deposition. A recent study that used a prototype choked interstitial dipole from Vivant Medical in porcine liver in vivo found that thermal lesion length (4.4 cm) was much larger than thermal lesion diameter (1.5–2.1 cm). This indicates that tissue thermal properties also play a significant role in lesion formation under standard treatment durations and are critical to the development of accurate models for MWA. As such, there is an increasing need for more advanced combined multiphysics models, which would allow researchers to design more advanced antennas using transient solutions of the Pennes bioheat equation. Such models also require a complete understanding of the complex, temperature-dependent interactions between the tissue’s dielectric properties and thermal properties, which can be affected by factors such as tissue water content, latent heat loss, water vaporization and diffusion, and vapor condensation.

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REFERENCES


COAXIAL-BASED INTERSTITIAL ANTENNAS FOR HEPATIC MICROWAVE ABLATION


