An Overview of Medical Implant Antennas

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***Abstract*—Antenna miniaturization for integration with small implants for sensing, wireless powering, and communication is a multi-parameter design task. The implanted antenna performance is governed by the antenna size, operating frequency, antenna surrounding tissues, subsequent biological tissues, antenna encapsulation, the electronics and metal objects nearby the antenna, implant depth, and the electromagnetic radiation source defined by the antenna geometry. The antenna performance can be characterized by impedance matching, bandwidth, antenna near-field, far-field radiation pattern, efficiency and gain, and the specific absorption rate (SAR). This paper reviews the design steps for miniaturized implant antennas, the frequency dependency of biological tissues, and EM modeling and simulation tools to support the design. Examples of miniaturized implant antennas are provided for application in the cardiac, gastric, and brain.**

***Index Terms*—antenna, implant antennas, miniaturized antennas, wireless body area networks**

I. Introduction

The health status of a person can be passed to a cloud for preventive medicine by utilizing smart implants. The implant technology is still advancing with developments in biomedical sensors for in-situ tracking of biomarkers and with the improvements in the communication and powering strategies which is opening doors to real-time sensing and data communication with a standalone implant. Fig. 1 shows a conceptual drawing of a person with multiple implants and on-body hubs that interconnect the implants to a communication network. In this scenario, the implants can be remotely powered, and different wireless links can transmit the sensed information to an on-body aggregator. Different sensors have already been approved for human healthcare monitoring, such as cardiac pacemakers, implanted ECGs, cardiac accelerometer sensors, glucose sensors for diabetic patients, stomach bleeding sensors, gastric sensors, brain signal sensing for amputees, and brain-machine interface (BMI) devices. Among these sensors, some require continuous high-data rate sensing and transmissions, such as brain-implanted electrodes and wireless endoscopy capsules, and some require a moderate rate or low-rate sequential sensing. The antenna is the primary unit that plays an essential role in the power efficiency of the implant system in different applications such as wireless communication, wireless powering, and sensing. The implanted antenna can be used in the near-field, mid-range, or far-field scenarios for the applications above. The antenna design must be adapted to the use case. The typical design basis for antennas is the operating frequency, platform geometry and size, encapsulation, required bandwidth, specific absorption rate (SAR), and the antenna surrounding that affects the design protocol. Due to the variety of design parameters, there is no standard implant antenna design protocol that can be generalized for all applications. So far, many papers have been published to address specific implant needs with different approaches that mainly address antenna miniaturization by retaining the antenna radiation efficiency. Designing an implantable antenna with a tiny form factor for simple integration with implants and support for a wider bandwidth to prevent detuning owing to differences in human tissue properties is always difficult. In this paper, we summarize the concept design of miniaturized implant antennas by accounting for the intrinsic limitations governed by frequency allocations, biological tissue loss, application area, radiation safety, and encapsulation.

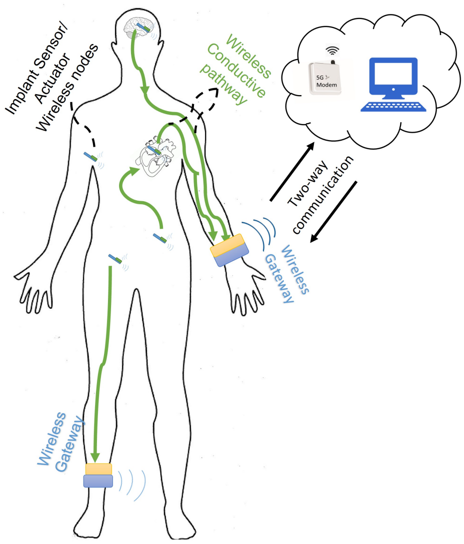


Fig. 1. Illustration of body implants with wireless connectivity to an on-body agregator for remote network connectivity.

II. Frequency Allocations

Different national and international agencies, including the FCC, ETSI, and ITU, allocate frequencies for medical radio transmission to maintain the interference-free functioning of the radio bands. There are mainly three frequently used operating frequencies for implantable antennas, namely the medical implantable communication service (MICS, 402–405 MHz), the wireless medical telemetry service (WMTS, 1.395–1.4 GHz), and the industrial, scientific, and medical (ISM, 433–434 MHz, 868–868.6 MHz, 902–908 MHz, 2.4–2.48 GHz, 5.715–5.875 GHz). In addition, some regions have been authorized ultra-wideband (UWB, 3.1–10.6 GHz) for high-quality transmission. The frequency spectrum below 150 kHz is also allocated for inductively coupled implant medical devices that can transport a small amount of data when continuous data transmission is not required. Inductive coupling uses resonant frequencies of 1 MHz, 5 MHz, 10 MHz, 24 MHz, and 49 MHz, and is only applicable for implants near the surface ($<$ 40 mm), such as retinal and cochlear implants [1]. Human body communication (HBC), also referred to as intrabody communication (IBC) or body channel communication (BCC), uses frequencies below 60 MHz. IEEE 802.15.6 standard designates galvanic coupling as the human body communication with a center frequency of 21 MHz [2]. The conductivity of the tissues is the main factor that helps to establish the connection link. Antennas in the form of coils (for magnetic coupling), electrode patches (for conduction coupling), and patch geometries for radio frequency are the selections.

III. Effects of Tissues and Radiation Limits

Implant antenna design require a thorough knowledge of the biological tissues embedding the antenna, in which the mutual interactions affect the antenna characteristics. Gabriel [3] has provided precise measurements of different biological tissue properties ranging from KHz region to 20 GHz. The biological tissues indicate high conductivity with increasing frequency and show large permittivity depending on the tissues' water concentration. Fig. 2 shows the material properties of some selected biological samples. Electromagnetic properties of human body tissues are frequency-dependent; varying depending on the subject's age, the sample temperature, and water contents by 5%-20%. The electrical conductivity is the source of propagation loss, while the high permittivity shortens the effective wavelength that causes the antenna near-field concentrate to the antenna proximity which introduce further near-field loss in the surrounded lossy tissues. The biological tissues are nonmagnetic with unity permeability. Therefore, the type of electromagnetic excitation source, controlled by the antenna geometry, can significantly influence the antenna proximity loss, the SAR value, and the link performance.

SAR is a defined norm with a specified limit that permits using electromagnetic radiation devices for humans. The known effect of electromagnetics is the thermal effect. IEEE C95.1-1999 standard restricts the SAR averaged over any 1 g of tissue to be less than 1.6 W/kg, and the IEEE C95.1-2005 SAR averaged over any 10 g of the tissue to be less than 2 W/kg [4]. The point SAR would be a more indicative parameter for the small implants as the antenna surrounding tissue would be much smaller than the above weight definitions. Additionally, the accumulated thermal effects and SAR due to the antenna structure heating, integrated electronics heating, and near-field antenna heating should be considered in system design.

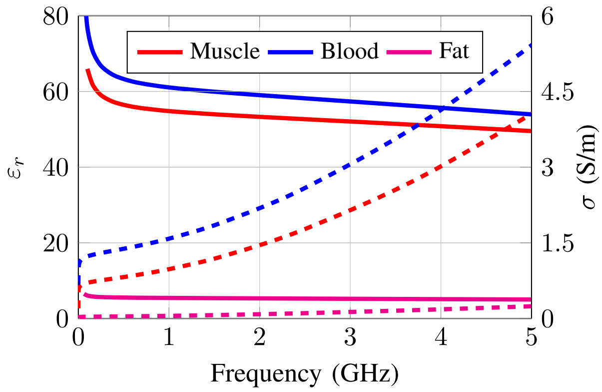


Fig. 2. Material properties of sample tissues. Solid and dashed lines represent relative permittivity and conductivity, respectively.

IV. Numerical Analysis of Implant Antennas

Electromagnetic (EM) characteristics of the antenna implanted in biological tissue can be analyzed using numerical EM methods of finite element method (FEM), finite difference time domain (FDTD), or using spherical dyadic Green's function (DGF) if the simulation geometry allows [5]. The tissues can be modeled as homogeneous phantoms, multilayer homogeneous phantoms or heterogeneous phantoms produced by computer tomography (CT), MRI, or sliced photography from Visible Human Project (VHP®) by integrating the electric characteristic data of human biological tissues. Whole-body human phantoms comprising around 50 different biological tissues (including skin, fat, and muscle) is available in different commercial EM tools, which also integrate the frequency-dependent tissue properties. SEMCAD®, CST MWS®, ANSYS HFSS®, and REMCOM XFDTD® are some useful EM tools for simulating biologically integrated antennas. Human phantoms with heterogeneous voxel data are supported in FDTD tools, while other non-voxel CAD models can be used with FEM simulations. Graphical processing unit (GPU) hardware is beneficial for time domain solvers and results in hardware acceleration. FEM tools are appropriate for narrowband, low frequency, and small antenna simulations and can consider the antenna detailed geometry using multilayer homogeneous phantoms. EM computations should also include the antenna coating material and thin encapsulation layers with precise meshing to correctly evaluate the near-field loss, SAR, and impedance features. A realistic heterogeneous phantom can also be used to simulate the antenna. Note that the voxel data has lower resolution (down to 1 mm^3 for Hugo model) hence the antenna meshing at the border of the voxel and the thin coating should be defined precisely to avoid simulation errors. Using a homogeneous phantoms allow the material interface to be defined precisely, which makes the simulation more accurate. Both heterogeneous and homogeneous phantoms could help correctly estimate the results.

For implantation, the antenna should either be designed and produced with biocompatible materials, or should be coated with biocompatible material to avoid any toxic reactions. The coating, namely the dielectric insulation, also avoids direct contact of the antenna with the biological medium which controls the SAR value and provides higher radiation efficiency. Silicone ($\varepsilon_r = 2.2$, $\tan\delta = 0.007$) [6], alumina ceramic ($\varepsilon_r = 9.4$) [7], polyethylene ($\varepsilon_r = 2.25$, $\tan\delta = 0.001$) [8] are some examples of biocompatible materials used for implant coating.

V. Implant Antenna Design Considerations

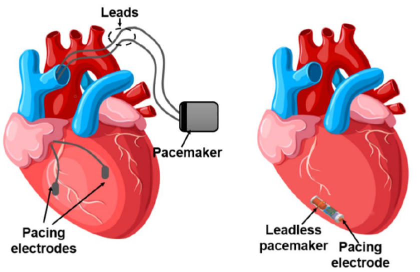
Small antenna design faces several challenges due to the difficulty in the matching circuit and the poor radiation efficiency imposed by the low radiation resistance resulting in large current and ohmic loss in the antenna structure or the antenna proximity. The theoretical limit for small antenna performance that was derived decades ago by Wheeler and Chu governs design trade-offs for size, bandwidth, and efficiency [9], [10]. Based on the study conducted by Sievenpiper et al. [11], the small antennas that have the highest performance in free space are those that involve the lowest permittivity, have an aspect ratio close to unity, and for which the fields fill the minimum size enclosing sphere with the greatest uniformity. All of these conclusions are valid in free space where the antenna is surrounded by a lossless medium of unit permittivity and permeability. An implant antenna is operated as a sub-system integrated with electronics, cables, batteries, etc., and is encapsulated in a metal or RF transparent casing. The antenna is surrounded by biological tissues with different electrical properties. The tissue-embedded miniature antenna design needs a new design perspective and constraints. For operational stability, the influence of the surrounding tissues on antenna impedance must be minimized. Additionally, considerations of the antenna SAR, embedded antenna radiation efficiency, far-field gain or near-field coupling, and the antenna structural loss to reduce device thermal effect should be taken into account. In tissues with high permittivity, the effective wavelength becomes shorter intensifying the antenna near-field. In this intense near-field, the conductivity of the tissue will cause more loss. On the other hand, the conductivity of the tissue contributes in favor of the antenna size, where the appropriate antenna configuration can extend the antenna's virtual size to the medium, which creates a size beyond the physical size of the antenna. Finally , due to the nonmagnetic nature of the tissues, the realization of magnetic antennas, such as closed current paths (e.g., loop antennas) with stronger magnetic near-field will reduce the SAR and alleviate the impedance changes with tissues [12]. A thorough analysis of the fundamental limits for implanted antennas using Hertzian electric and magnetic dipoles is conducted in [13]. Some losses due to the near-field coupling to the biological medium can be engineered by realizing electric or magnetic current sources, and some losses are unavoidable due to the propagation inside the tissues or impedance mismatch in the tissue-air interface. The upper bound for the power density that can reach free space from an implant depends on the frequency, the electrical properties of the biological tissue, the depth of the implant, the size of the implant encapsulation, and the type of the excitation source [14]-[16]. Also, for any given size of implant and depth, an optimum frequency governs the maximum power efficiency [12], [17]. Ideally, a magnetic antenna is the best choice for a biomedical implant scenario with limited near-field loss and lesser influence from the medium, however, realizing such an antenna with an electric source is not straight-forward. In [18] an electric loop antenna with a ferromagnetic core is analyzed and compared to a true magnetic dipole in the implant scenario where improvement of almost 5 orders of magnitude can be expected, with some realization challenges. Magnetoelectric antennas [19]-[21] are another solution that would excite the magnetic dipoles using mechanical vibrations and can be realized in the MHz and GHz range with potential use in biomedical applications.

VI. Implant Antenna Examples

Compared to their conventional counterparts, miniaturized antennas for implants have more limitations. They are dictated by the implant geometry, antenna placement inside the body, encapsulation material, implanted tissue, implant depth, available size or volume, device longevity, and tissue growth over the implant. The antenna miniaturization in free space has been explicitly discussed in the literature [22]. Miniaturization is generally achieved by changing an antenna's electrical and physical properties based on the topology and material. Commonly used methods include utilization of fractals, meander lines, engineered ground planes, reactive loads, slow-wave structures, high dielectric constant substrates, and metamaterials. The implant antenna design follows the same miniaturization process as antenna miniaturization in free space, but specific care should be given to the aforementioned limitations. In the following sections below, we cover some recent implant antennas designed and implemented for cardiac, brain, and gut applications.

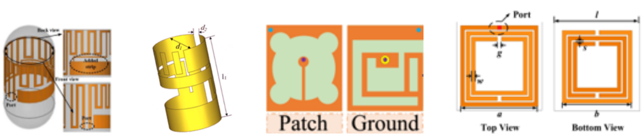
A. Antennas for Cardiac Pacemakers

Pacemaker devices are used to cure cardiac arrhythmia. Conventional pacemakers (CP) use wires to send the stimulation therapy signals to the cardiac chambers, leadless pacemakers (LP) are recent autonomous wireless devices that make the treatment as seen in Fig. 3. Both systems need communication antennas for programming and monitoring purposes. CPs use magnetic coupling to communicate with on-body readers for telemetry and programming. The new classes of CPs have Bluetooth connections to report their status to the patient hub. LPs are deep inside the heart (about $8-10$ cm) from the surface. They require a connection to a reader for programming and event reporting. Also, the dual chamber leadless pacemakers need communication in between, for synchronous operations where the communication in the intra-cardiac chambers is essential. Due to the power constraints, link efficiency is critical for LPs. Antenna design for pacemakers must satisfy the required size, encapsulation, medium, and integration. In [23] a meandered-shaped dipole antenna with 37.8% bandwidth is suggested. In this design, irrespective of the bulky volume of 329 mm^3, the small gain indicates a short telemetry range as seen in Fig. 4-a. In [24], an implantable antenna operating in the MedRadio band with a wider bandwidth of 400 MHz was reported as shown in Fig. 4-b. The gain was -23.19 dBi, but the antenna has a larger profile of 30 mm $\times$ 16 mm $\times$ 0.5 mm and the antenna geometry cannot fit in an LP. An ultra-miniaturized loop-shaped radiator antenna (6 mm $\times$ 7 mm) with ultra-wide bandwidth (790–3830 MHz) for avoiding detuning due to tissue growth has been designed for LP using different cuts in the radiator and ground plane that made the antenna configuration compact [4] as seen in Fig. 4-c. A broadband miniaturized implantable antenna (12 mm $\times$ 12 mm) is designed for integration in a conventional subcutaneous pacemaker by introducing split resonant rings, which can cover from 272 MHz to 1504 MHz (MICS, ISM), and the relative bandwidth is 138.7% [25]. Fig. 4-d shows the proposed antenna which is suggested for communication and wireless powering using an external metasurface with the coupling of -35 dBm at 25 mm distance in 430 MHz.



(a) (b)

Fig. 3. Illustration of cardiac pacemaker with subcutaneous can and wireless link, b) leadless cardiac pacemaker with wireless link [26].

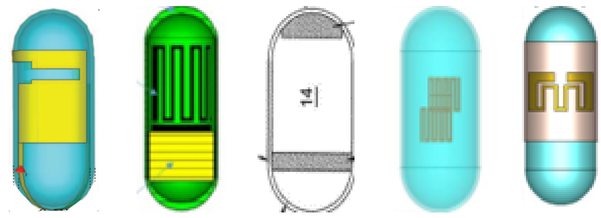


(a) (b) (c) (d)

Fig. 4. Sample designs of miniaturized pacemaker antennas.

B. Antennas for Wireless Capsule Endoscopy

Wireless capsule endoscopy (WCE) is used to visualize the gastric tract for medical diagnosis. It is a swallowable capsule with an embedded camera. The capsule moves through the gastrointestinal (GI) with different tissues around the capsule and captures images, which are transmitted to a receiver outside the patient's body. Usually, the capsule is battery powered and the emitted RF power is limited, in which the antenna SAR is not a big concern. Due to the implant depth and the available frequency allocations, the ISM band at 433 MHz would be a preference. Fig. 5 shows some sample designs for WCE. A conformal wideband loop antenna is introduced in [27], where the outer shell of the antenna is used as part of the antenna, and the other electronics are integrated inside the shell to gain space and achieve higher radiation efficiency. The proximity to the lossy tissues adds to the antenna bandwidth. Note that polarization mismatch should be compensated either on the receiver antenna or the capsule antenna, or both sides. The antenna might take three different polarizations, in which an array of antennas on the body with polarization and space diversity is needed. In [28], a meander-shaped reconfigurable conformal antenna is optimized for WCE in which the antenna is a narrowband design with loaded capacitive coupling. The antenna is used as a backscatter reflector to stream the camera data. The narrowband nature of the antenna increases the antenna efficiency but makes it sensitive to environmental changes. In [29], a wideband electrode antenna with insulation to the medium is developed to use the capsule medium as part of the antenna and extend the antenna's virtual size for higher efficiency video transmission. In [30], three orthogonal current paths are realized with a meander-shaped geometry at 2.4 GHz ISM band to mitigate the orientation problem. In [8], a wideband conformal antenna is proposed to operate at 650 MHz to 3600 MHz with less sensitivity to the environment. The antenna forms a meander loop that is loaded with medium loss. Galvanic or conductive communication is also a power-efficient option for WCE video transmission [31] that needs direct contact of electrodes with tissues. In a different approach, [32] utilizes the antenna de-tuning for capsule localization.



(a) (b) (c) (d) (e)

Fig. 5. WCE antennas (from left to right [27], [28], [29], [30], and [8])

C. Brain Implant Antennas

Wideband analysis of plane wave electromagnetic propagation in the human head [33] and a mini-horn antenna to analyze the wave penetration in ISM and UWB band for brain application [34] have been reported. The antenna is used either for wireless powering or communications. In [35]-[37] wireless link to a 1 $\times$ 1 $\times$ 1 mm^3 implant loop using regular, segmented, and tilted transmit loops is analyzed for a depth of about 16 mm, where limiting the SAR on the brain cortex with different external antennas is considered. Overall, the simulated maximum link power efficiency was -24.5 dB at 400 MHz with 54.5 μW delivered to the implant IC at 106.5 mV. The addition of a magneto-dielectric core in the implant improved these values to -22.4 dB, 110.5 μW, and 137.0 mV at 250 MHz. In [6], a small triple-band implantable planar inverted-F antenna (PIFA) resonating at MICS ($401-406$ MHz), and ISM ($902-928$ MHz and $2400-2483.5$ MHz) for wireless brain implants is introduced to improve the antenna gain using multilayer tissues. Recent research fields include miniaturized antennas for BMI for two-way communication for sensing and stimulation and wideband antennas for high data rate brain connectivity.

VII. Conclusion

The procedure for designing miniaturized implant antennas for biomedical implants is described. The design requirements are assessed, and sample design examples are provided. The implant antenna efficiency is a factor of the antenna near-field that is confined in the proximity of implant material and tissues, the operating frequency, and the implant depth.

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