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Antennas and Propagation for Implanted RFIDs for Pervasive Healthcare Applications

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***Abstract*—Radio Frequency Identification (RFID) is a growing technology, with the potential for reducing medical errors and improving the quality of healthcare in hospitals. The benefits include more secure and safe access in the healthcare environment (with the possibility, for example, to track patients, personnel and equipment), as well as providing the means to easily identify patients and their medications with low risk of error.**

In this paper, we present an overview of the challenges faced in antenna design, electromagnetic modelling and wave propagation for RFID implants. The performance of Ultra High Frequency (UHF) subcutaneous tag antennas were investigated numerically and validated with measurements. Furthermore, the wave propagation between an off-body reader and an implanted tag was analysed, in both free space and a scattered indoor environment. Results demonstrated that a passive tag solution allows a very limited communication range, due to the body losses, the electrically small size of the antenna and nulls in the radiation pattern. In comparison, a maximum communication range of 10 m was predicted as achievable for an active tag operating indoors with a limited power (-20 dBm).

I. Introduction

The development of new wireless technologies for healthcare applications has attracted a great deal of attention from both academic and industrial researchers, due to the aging population in the occidental world and the associated rise of healthcare costs and demand on hospital resources [1]. Among them, Radio Frequency Identification (RFID) technology offers promising solutions in building wireless mobile healthcare services ([2], [3]), such as personnel locating and tracking.

Keeping track of people in hospitals is important for a number of reasons, including the need for security, the ability to manage emergency scenarios and the efficient and prompt delivery of services.

Implanted RFID tags were used in the past for tagging animals; recent developments have opened up a new potential scenario, where RFID tags are implanted in the human body [4]. The use of implanted tags (rather than body-worn) reduces the risk of the tag being lost; in addition, it is invisible and ideal for non-cooperative subjects.

In the past, Low Frequency (LF) and High Frequency (HF) RFID systems were used for on-body and implanted applications. Although at these frequency bands the signal is not strongly attenuated from the presence of the body tissues, these systems are limited by their short range and a very low data rate. In Ultra High Frequency (UHF) and Microwave RFID systems, the signal is strongly attenuated inside the body tissues [5]–[7]; however, in such systems longer operational range and higher data rate is achievable [8].

Thanks to the fast developments in low-power electronics, it is now possible to utilise RFID with integrated sensing and signal processing capability [8]–[11], in addition to more traditional tagging operations. The RFID implanted tag can be then used in conjunction with implanted sensors monitoring physiological parameters (blood pressure, temperature, heartbeat, etc.), in order to create a RFID-based Personal (or Body) Area Network (PAN and BAN, respectively).

In [12], a global overview of implanted microsystems and their clinical application is presented; it is shown that, at the current stage, implanted devices can be almost everywhere inside the human body. To ensure adequate wireless connectivity of such implanted systems with external base stations (or other body-worn, or implanted, systems) is maintained, an accurate understanding of the radio propagation channel, including the effect of the antenna, is urgently needed [13]–[21].

Although antennas have been widely used in RF systems, the antenna efficiency, radiation pattern and input impedance will suffer when surrounded in a lossy environment, such as the human body [22]. In this paper, a review of body-centric RFID systems is presented. Furthermore, the challenges faced in antenna design for implanted applications are emphasized through the study of a miniaturized meandered Planar Inverted-F antenna (PIFA) deployed as a subcutaneous tag antenna. Simulations were performed using a three-layer digital phantom and validated with measurements. In addition, an investigation into the propagation characteristics was performed, examining the radio channel between an implanted RFID tag and an off-body reader, and including the effect of the indoor environment. To the authors' knowledge, this is the first time that implantable RFIDs and associated radio channels are systematically analysed and experimentally studied.

An overview of RFID implants is given in Section II, together with a discussion of possible pervasive medical applications. Section III presents a study of different subcutaneous meander-line PIFAs, optimized to work at 433, 868 and 2400 MHz, three commonly-used RFID frequencies. The antennas have been simulated using a three-layer model (composed of skin, fat and muscle tissues) and the Specific Absorption Rate (SAR) was calculated as a function of the input power. Section IV illustrates the development of a physical phantom; measurements were performed at 868 MHz and were found to be in good agreement with simulated data. Section V analyses the radio propagation between the implanted tag and the off-body reader, for both passive and active RFID systems; different indoor propagation scenarios were considered. Finally, Section VI draws some conclusions.

II. Background

A. Implantable RFIDs and Antennas

In 1998, an RFID tag was implanted for the first time inside a human body, enabling him to control doors, as well as activate lights [23].

RFID tags can be categorized as either passive or active. Passive tags operate without batteries and store information in read-only form; being battery-less, they offer longer lifetime. Active RFID tags contain an internal battery and can transmit data over longer range [4]. Currently, only passive RFID tags have been approved for implantation inside human subjects. VeriChip was the first to obtain approval from the US Food and Drug Administration (FDA) for its implantable passive RFID chips [24], in 2002.

The antenna plays an important role in sustaining the RFID performance and especially its reading range. For implantable applications, the dispersive nature of the human tissues in which the antenna is embedded must be accounted for. In addition, antenna efficiency, radiation pattern and input impedance will be affected [22].

Implantable antennas have already been proven to be useful for therapeutic purposes [25], where targeted cells are heated up due to the absorption of electromagnetic waves. Recently, there has been growing interest and research on designing implantable antennas, as part of an integrated sensor for diagnostic purposes, with the objective of continuously recording physiological data [26], [27]. Previous systems largely relied on an inductive link [26], which inherently has low data rate. The communication link between an UHF implantable antenna and a receiving antenna, as part of an external receiver, has the potential for high-speed, short-range access to the subject's state of health in real time [17], [19], [27].

The size of the antenna is critical, as miniaturization is needed for implantation but may result in a less efficient antenna. Techniques to reduce the size of the conventional patch antenna have been widely used. For example, the PIFA structure has earned much attention recently by introducing a shorting pin from the radiator to the ground plane. The use of a ground plane offers the possibility to host electronics packaging. However, antennas developed with this technique often suffer from a narrower bandwidth. The addition of high permittivity substrates and superstrates are also often used for size reduction, as well as bandwidth improvement. Meandered-line antennas ([17], [19]) also present an attractive choice for reducing the tag size, by increasing the electrical length of the antenna. The aforementioned techniques have all been used together in [17] and [19], where a meandered line PIFA antenna has been developed at 402 MHz in the subcutaneous region. In [19], a dual-band PIFA for continuous glucose monitoring was also presented.

B. RFID Systems and Potential Applications

There are a number of interesting applications for RFID tags implanted in the human body. They can be summarised under the following categories [4]:

- control – generally related to access control and personnel tracking, with applications in commercial, private (e.g., anti-kidnap devices) and government (e.g., passports) schemes;
- care – any application related to medicine, health-care or well-being, including patient ID, storage and portability of medical data with the patient and monitoring physiological parameters for medical purposes;
- convenience – these are applications that make day-to-day life easier, including financial (e.g., credit/debit card replacement, travel-card substitute), ID and location-tracking (e.g., help with emergency services, roadside assistance) and ID for home-automation and access-control.

The control category underlies and enables the other applications. Simple ID-based applications have negligible bandwidth requirements and are limited only by the achievable range. Other applications may require the transfer of greater amounts of data (e.g., patient records), which implies greater bandwidth would be an advantage. This, in turn, suggests that RFID systems at low frequencies (LF and HF) may be inappropriate. The remainder of this section reviews published examples of patient-tracking systems, as the most prevalent application of RFID tags in the literature. Expansion of such networks for other applications (e.g., patient-record handling) does not fundamentally affect the underlying antenna, propagation and network issues exemplified by solutions focussing on patient tracking.

Modern hospitals are busy places, where hundreds may be present on any given day, comprised of staff (of all types), patients and visitors. Even the smallest clinics or general practitioner (GP) surgeries may see volumes numbered in the tens. Keeping track of all these people is important for a number of reasons, including the need for security within the environment, the ability to manage emergency scenarios (particularly those where evacuation may be required, such as fires and bomb threats) and the efficient and prompt delivery of services. Other resources, such as medicines, portable equipment (ranging from surgical kits to defibrillators and more) and consumables (e.g., syringes, needles, catheters, sterile gloves) must also be managed efficiently and securely. The necessity of locating, or tracking the movement of, these items and people was identified from as early as the 1980s [28], whilst the general concept dates to at least the late 1960s [29].

There are a number of technologies that may be employed to achieve this objective, including infra-red (IR), microwave/RF and ultrasonics. One of the factors involved in selecting the technology is the coverage required. Cheaper technologies, such as ultrasonics and IR, will not necessarily scale up to cover larger areas as effectively as the relatively expensive RF solutions, which may actually offer more functionality than IR and ultrasonics in any case. With the advent of miniaturised RFID tags, the cost of the RF technology has become more competitive.

Another issue to consider is the resolution required in the locating and tracking algorithms. For example, is it sufficient to locate a patient within a room, or is a more specific requirement necessary (e.g., to within 1 m)? Whilst this may be application-specific, a multi-function network — attractive from reasons of efficiency and cost — must be able to offer the minimum resolution for the most demanding application, potentially offering this to other applications "for free".

Radar and the Global Positioning System (GPS) are the two most common methods of locating and tracking "targets" in outdoor applications. Neither of these methods is particularly suited to the indoor environment. Here, a network approach is optimal, based on distributed sensors throughout the infrastructure. Most, if not all, of these sensors will be permanently fixed in one place and therefore have known locations, enabling the mapping of detected signals to location. The questions of coverage and resolution become crucial in determining the network configuration and capabilities and are, in fact, linked. For example, if a sensor has a maximum detection range of 1 m, complete coverage in a long corridor may require a sensor-spacing less than or equal to 2 m (depending on the application). The resolution is fixed by the sensor spacing, assuming the sensor uses a non-directional means of detecting the target (e.g., an omni-directional antenna). Enhancing this raw resolution requires advanced techniques in the sensor.

True RFID technology (i.e., conforming to the standards of either or both the International Organization for Standards and the industry-led Electronic Product Code body, EPCglobal Inc. [30]) has been commercially available for at least a decade. As might be expected, there are a number of papers in the literature relating RFID to location-monitoring and tracking applications. One of the earliest examples is [31], where a number of readers were combined with additional "reference tags" to determine the location of target tags. The use of reference tags, with known fixed locations, is an interesting feature, as it reduces the number of readers required to achieve a given location accuracy, as it allows a reader, or central processing node, to calibrate its measurements. Another feature of interest is the requirement to know the received signal strength (RSS). Many systems that calculate location require this information.

An even earlier system, SpotON, also made use of the received signal strength to determine location [32]. A commercially available tag/reader combination was used as the basis of the system. The architecture requires multiple readers with overlapping coverage, common to most implementations. Each reader provides the central processing server with a set of RSS measurements, mapping to the tags in its coverage field and corresponding to an approximate separation between reader and tag. The equation linking signal strength and separation was derived empirically; triangulation of received data is performed by the central server to convert the relative separations into locations.

A paper from 2004 suggests uniting RFID with floor pressure sensors [33]. The argument is that combining the complementary properties of each approach results in a better overall system. The pressure sensors will detect the presence of someone as he moves about; the RFID system identifies the person (ensuring security), as well as assisting the pressure sensors. For example, when many people are present in one location, the pressure sensor system may struggle to discriminate the presence of an extra person, whereas the RFID system, implemented with readers at doorways, enables a count to be kept. The resolution of this method is low, as it is essentially a presence-detection system; any tracking or location-determination is computed elsewhere.

RADAR, a non-RFID solution using a WaveLAN system of base-stations to determine location, was discussed in [34]. WaveLAN (similar in objective to IEEE 802.11/Wi-Fi) hardware was used as it provided access to both RSS and signal-to-noise ratio (SNR) data. The RSS was used to determine location, when coupled with empirical measurements, signal propagation modelling, or both. Essentially, the system is calibrated for the given base-station locations and data corresponding to known locations and signal strengths stored. In "on-line" mode, the system compares the observed data set to the stored data set to determine the likely location of the target(s), using a nearest-neighbour signal space algorithm using the Euclidean distance method. The main advantage of this system is that it — potentially, at least — makes use of existing infrastructure (i.e., WLAN access points and routers). Its main disadvantage would be the need for empirical data, usually building-dependent.

Another patient tracking system using RFID was discussed in [35]. As in the examples discussed above, the patients all wore RFID tags that were detected using a network of readers in fixed locations. The issue of dynamic tracking was examined, with an emphasis on potential interference and its mitigation. Two sources were identified: when tags were in close proximity or when two "field generators" (readers) were in close proximity (which is always the case when using over-lapping regions for locating objects). A scheduling approach was taken to minimise the interference caused by readers operating simultaneously, called "Graphic colouring". This issue is essentially that of having multiple nodes trying to access the same channel; the solution is a time-domain multiple access (TDMA) anti-collision strategy. The fact that only one tag can respond at any point in time places a limit on the "refresh rate" of locations and suggests that distributed pre-processing of the data may assist in minimising any lag in the data accuracy.

A fairly complete solution for the hospital environment was presented in [36], where patients, charts and equipment all possess RFID tags. Each ward has an RFID reader at the doorway to monitor movement in and out, plus a networked computer with attached reader to enable the medical staff to obtain the necessary data for the specific tag. The interface is via a web browser; the Internet and web-based services are proposed, although security and privacy issues may encourage the use of a firewalled intranet and servers.

The inverse problem (that of a mobile reader using tags in fixed locations as navigational aids) has also been examined. One implementation, using RFID technology, is aimed at "first-responders" (e.g., fire-fighters), who must navigate their way around a normally unfamiliar building in difficult circumstances [37]. This system uses multiple tags per room, but is essentially a presence-based approach. One interesting point for consideration is the effect of the fire on the RF propagation characteristics (for instance, smoke and other substances in the air will increase attenuation, whilst thermal layers may form that can reflect or refract RF waves [37]). Another question is the deployment and maintenance of the RFID system (both the tag network and the database mapping tag ID to location), and how that information is made available to the emergency services when required.

These systems may, in principal, all be implemented using either wearable or implanted RFID tags. The propagation problem for both is similar, although implanted tags must cope with greater loss. The primary advantage of an implanted solution is in the security engendered by its physical location, in that a tag can be lost and remain unnoticed in a worn-tag solution, but this is impossible if implanted.

III. Subcutaneous Implantable Antenna

The challenge of designing an antenna for implantable RFID tags lies in understanding the impact of lossy human tissues and the trade-offs between size, types of antenna, efficiency and safety.

It is also essential to study the long-term exposure of the antenna to the biological environment. The implanted antenna must be bio-compatible over the long-term, and the choice of the bio-compatible material used to load the antenna and the surrounding tissues is critical in the design of implantable RFID antennas. In this section, we focus on the study of the sensitivity of antenna properties to the surrounding tissues, as well as the depth of implanted tags.

A. Three-Layer Phantom

Since the dielectric characteristics of biological tissues must be used in the optimization of the device operation, an accurate assessment and knowledge of the properties of biological tissue, and especially their electrical characteristics, is crucial. The dielectric properties of biological tissues have previously been determined by Gabriel *et al.* [38]. The electrical properties (relative permittivity, ϵ_r; and conductivity, σ) of three different biological tissues (skin, fat and muscle) have been listed at the frequencies of interest. As seen in Table I, muscle and skin tissues have higher conductivity and permittivity values than tissues with a lower water-content, such as fat.

TABLE I
Relative permittivity (ϵ_r) and conductivity (σ) of skin, fat and muscle at 433 MHz, 868 MHz and 2.4 GHz [38].

tissue	433 MHz		868 MHz		2.45 GHz	
	ϵ_r	σ (S·m^{-1})	ϵ_r	σ (S·m^{-1})	ϵ_r	σ (S·m^{-1})
skin	41.6	0.7	41.6	0.85	38.0	1.46
fat	5.6	0.04	5.5	0.05	5.3	0.1
muscle	56.9	0.8	55.1	0.93	52.7	1.74

A simplified three-layer tissue model, composed of skin, fat and muscle tissues, is considered (Fig. 1), with the antenna placed inside the fat tissue layer. The thickness of the skin (t_1), fat (t_2) and muscle (t_3) layers have been set to 2 mm, 10 mm and 25 mm, respectively; these have been chosen to be similar to average human body tissues. The impact of these tissues and their thicknesses on the antenna performance within the body is analysed in the next section. The implanted depth of the antenna is also studied as a function of the distance, d, from the radiating element to the top of the model. The following simulations results have been obtained using CST Microwave Studio™.

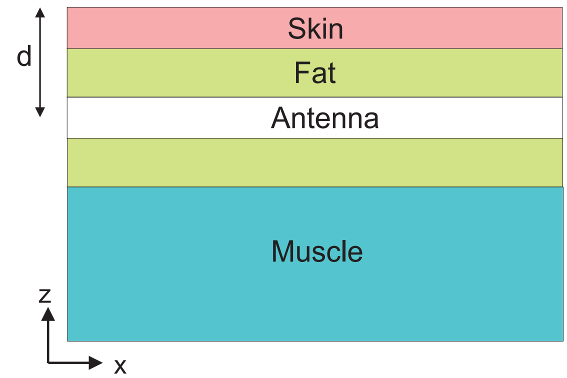
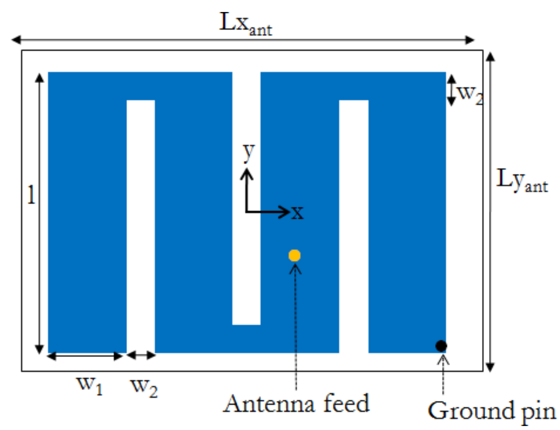


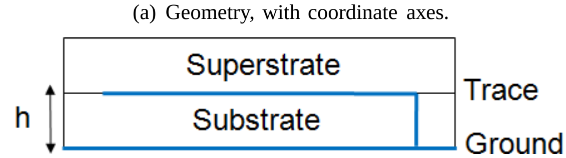
Fig. 1. Physical layout of the implantable antenna inside a three-layer tissue model, composed of skin, fat and muscle tissues.

B. Antenna Geometry

Fig. 2(a) shows the geometry of the meandered PIFA antenna considered in the present work. The radiating element is covered by a bio-compatible superstrate of dielectric constant $\varepsilon_r = 10.2$ and thickness of 1.25 mm, as seen in Fig. 2(b).



(a) Geometry, with coordinate axes.



(b) Cross-section.

Fig. 2. The meandered PIFA antenna considered [17].

The implantable antenna has been optimized to operate at 433, 868 and 2400 MHz, when placed inside tissue layers with the configuration presented in Fig. 1. Table II lists the dimensions of the antennas at the three RFID frequencies of interest.

When designing an RFID tag antenna, the designer must match the antenna input impedance to the highly capacitive impedance of the chip; in [8], the design of a set of wearable tag antennas has been proposed. To maximize the transferred power, such antennas present a highly reactive input impedance; however, it is shown that the chip impedance strongly varies from manufacturer to manufacturer; furthermore, it is impossible to have a single design optimized for every chip.

However, the aim of this work is numerical and experimental characterisation of implanted RFID antennas; therefore, the input impedance has been matched to 50Ω, without a loss of generality.

TABLE II
ANTENNA DIMENSIONS (MM).

	433 MHz	866 MHz	2.45 GHz
L_{xant}	24.0	19.5	5.6
L_{yant}	20.0	11.0	4.9
l	15.2	7.4	3.5
w_1	4.0	3.5	0.95
w_2	1.3	1.0	0.3

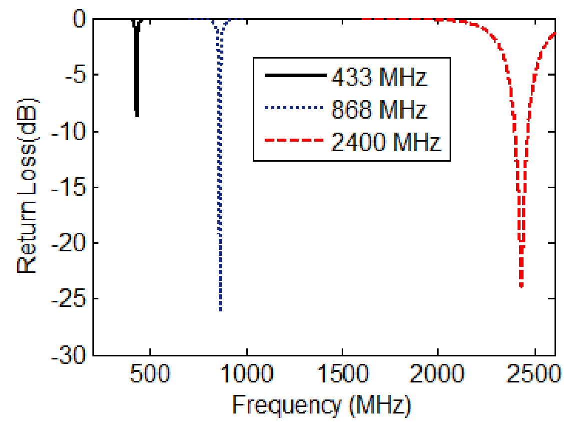


Fig. 3. Simulated return loss (dB) of the meandered PIFA at the three frequencies considered.

C. Simulation Results

Fig. 3 shows the simulated return loss of the optimized antennas listed in Table II. Good impedance-matching characteristics are observed at 868 MHz and 2400 MHz. However, although the antenna optimized at 433 MHz resonates at the desired frequency, a high return loss due to the impedance mismatch ($S_{11} > -10$ dB) is seen. Further optimization techniques can be used to improve the antenna performance [19].

The simulated gain G_{imp} is listed in Table III at the three frequencies of interest. It is evident that the antenna gain is directly linked with the effective volume of implanted RFID antennas. At 433 MHz, the antenna has the lowest gain, due to its small electrical length. It is also important to note that the body tissues are highly conductive (and, therefore, very lossy) at higher frequencies (Table I), which may lead to higher SAR values; hence, there are trade-offs between the RFID antenna size and human health and safety.

TABLE III
SIMULATED VALUE OF IMPLANTED GAIN G_{imp}.

Freq. (MHz)	G_{imp} (dB)
433	-28
868	-16
2400	-12

D. Specific Absorption Rate (SAR)

As previously mentioned, biological materials are lossy mediums; thus, they will absorb and dissipate as heat the energy of the electromagnetic (EM) waves impinging on their surfaces. The SAR criterion has been defined to measure the maximum rate at which energy is absorbed by the body, when exposed to electromagnetic fields.

The limits, which apply in general for mobile telephones and similar apparatus, are drawn directly from the applicable source documents: ANSI/IEEE C95.1 for the US and CNIRP for Europe and most of the rest of the world. Two limits are used: a lower value of 0.08 W$\cdot$kg^{-1}, for exposure averaged over the whole body, as well as a higher value (US: 1.6 W$\cdot$kg^{-1}; and Europe: 2 W$\cdot$kg^{-1}) applicable for local exposure to parts of the body. This partial-body SAR is averaged over a volume of tissue, defined as a tissue volume in the shape of a cube (1 g of tissue in the US standard, as opposed to 10 g for the European standard).

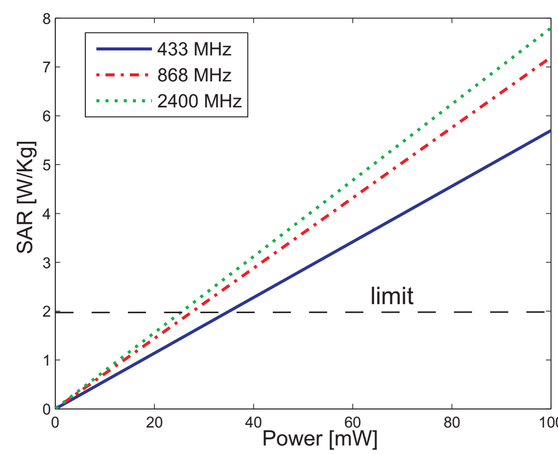


Fig. 4. SAR as a function of the input power, the European regulations states that the SAR has to be smaller than 2W/Kg.

E. Impact of implant depth and tissue thickness

The effects of implant depth and thickness of the surrounding tissues on the performance of the antenna have been studied. The simulation is performed at 868 MHz and the trend should also apply at other frequencies. The implanted depth, d, of the antenna is varied, in order to analyse its impact on antenna resonant frequency detuning and impedance mismatch, as seen in Fig. 5.

A small change in the resonant frequency of the implant is observed. Based on Fig. 5, an optimum depth, d_{opt}, for the implant can be chosen; it is found to lie between 5 mm and 9 mm. Outside this range, the detuning of the antenna becomes significant.

The effect of the different tissue thicknesses is now investigated. Figs. 6(a) and 6(b)show the return loss of the antenna

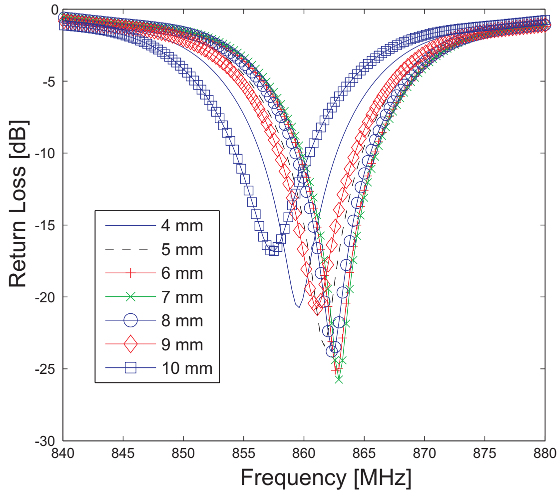
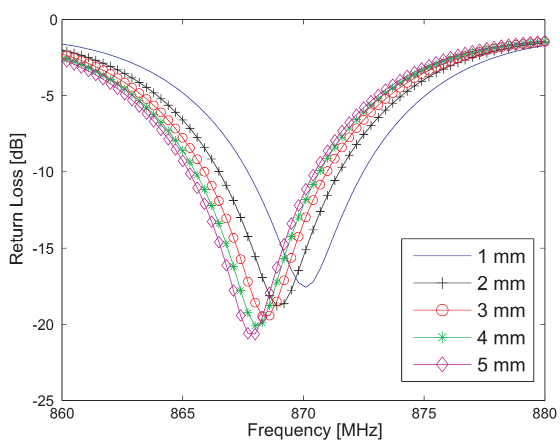
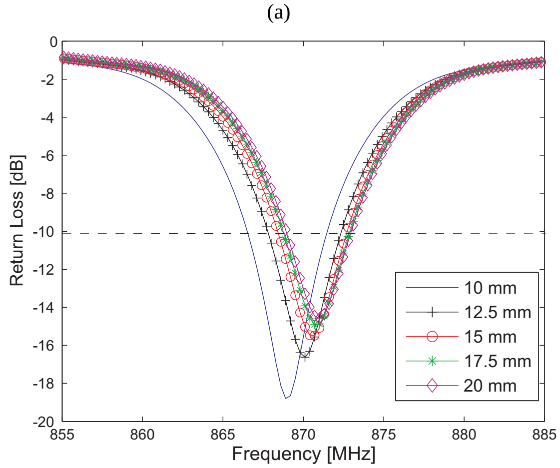


Fig. 5. Return loss as a function of the implant depth.

for different thicknesses of skin tissue (t_1) and fat tissue (t_2), respectively. The depth of the implanted antenna remains constant. It is found that the antenna impedance mismatch is weakly dependent on the skin tissue thickness and good impedance matching is seen for all fat thicknesses.



(a)



(b)

Fig. 6. Return loss changing the thickness of the tissues: (a) skin; (b) fat.

IV. Three-Layer Phantom Fabrication and Measurements

In order to validate the simulation measurements, the implanted antenna optimised for 868 MHz has been fabricated and embedded inside dispersive phantom materials. As the electrical properties found in the human body vary greatly with frequency, it is difficult to find ready-made materials with equivalent properties. Various recipes for tissue-equivalent dielectric liquids have been proposed, based on the values of the dielectric characteristics of the biological tissues [19], [39].

Phantom models representative of skin, fat and muscle layers have been fabricated, in order to simulate the tissue properties at 868 MHz. The compositions of the fabricated phantoms are listed in Table IV. In order to realize a three-layer phantom, the fabrication of a gel-mimicking phantom is necessary. For this purpose, 10 g of gelatine is also added into 100 ml of de-ionized water for the skin and muscle phantoms.

TABLE IV
Composition in weight (%) of fabricated phantom for skin, fat and muscle tissues at 868 MHz.

	Skin (%)	Fat (%)	Muscle (%)
De-ionized Water	50	2.9	59.5
NaCl	-	0.1	0.5
Sugar	50	-	40.0
Vegetable Oil	-	30.0	-
Flour	-	67.0	-

The measured relative permittivities and conductivities of the fabricated phantoms are listed in Table V and compared to their expected values [38].

TABLE V
Relative permittivity ϵ_r and conductivity σ of skin, fat and muscle at 868 MHz [38].

	Values from [38]		Dielectric Probe	
Tissue	ϵ_r	σ (S·m^{-1})	ϵ_r	σ (S·m^{-1})
Skin	41.6	0.85	38.72	0.77
Fat	5.5	0.05	4.9	0.04
Muscle	55.1	0.93	53.0	0.92

A picture of the fabricated antenna optimised for 868 MHz is shown in Fig. 7. Fig. 8 shows the antenna implanted inside the fabricated three-layer phantom. Fig. 9 shows the measured return loss of the fabricated antenna embedded inside the three-layer phantom; the antenna is well matched at the frequency of interest.

The x-z plane far-field radiation pattern of the fabricated antenna, implanted inside the phantom tissues, has been measured at 868 MHz, and it is plotted in Fig. 10. The peak gain of the implanted antenna is -17 dB at 868 MHz.

V. Propagation Between an Implanted RFID Tag and an Off-Body Reader.

Characterising the radio propagation between the tag and the reader is an important step in understanding the operational range of implanted UHF-RFID systems. The complexity of human body tissues, the shape of the body and the human-subject-specific behaviour of implanted antennas, make it difficult to derive simple propagation models.

Fig. 7. Picture of the fabricated antenna (substrate with radiating element dismounted from the superstrate) on Rogers R03010 (ϵ_r= 10.2, h = 1.27mm) placed next to a British 20 pence coin.

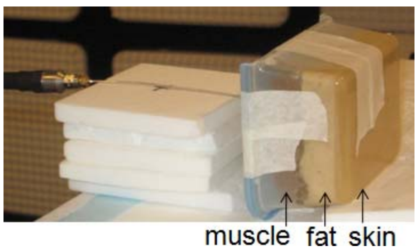


Fig. 8. Picture of the fabricated three layered phantom.

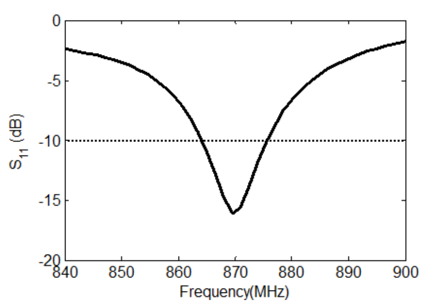


Fig. 9. Measured return loss of the fabricated antenna implanted inside the three-layer phantom. The antenna is well matched at 868 MHz.

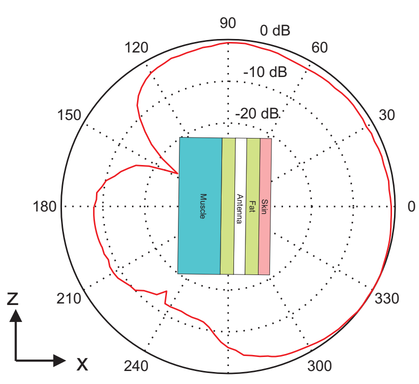


Fig. 10. Measured Radiation pattern of the fabricated antenna implanted inside the three layered phantom, on the $x - z$ plane.

For the case of propagation between an implanted transmitter and an off-body receiver, the communication is within the far-field region and the received power can be calculated using the following formula [40], [41]:

$$P_r[dBm] = P_t[dBm] + G_t[dB] - PL[dB] + G_r[dB] \quad (1)$$

where P_t is the transmitted power, G_t is the gain of the implanted antenna, G_r is the gain of the receiving antenna and PL is the path loss of the communication link. In this case, the gains of transmitter and receiver and the path loss of the radio channel are three independent variables; it is possible to find their product (or sum the log-magnitudes, as in Equation 1) to obtain the received power.

As highlighted in Section IV, the value of G_t is angle-dependent. The path loss of the radio channel is strongly dependent on the scenario considered. For propagation within a short range and in the absence of obstacles, the radio propagation can be approximated by free space conditions and the path loss can be calculated with the Friis formula:

$$PL[dB] = 20 \log_{10}\left(\frac{4\pi d}{\lambda}\right) \quad (2)$$

where λ is the free-space wavelength, and d is the distance between transmitter and receiver.

Due to reflections, diffraction and scattering of electromagnetic waves in the indoor environment, the transmitted signal reaches the receiver via more than one path. The path loss will, therefore, no longer have a d^2 variation. The most common path-loss model for indoor environment is the log-distance model [42]:

$$PL_{dB}(d) = 10\, n \log(d/d_0) + PL(d_0) + \chi_\sigma \quad (3)$$

where d_0 is a reference distance, chosen to be 1 m. $PL(d_0)$ is the path loss value at the distance d_0 and is assumed to be equal to the free space path loss. This assumption is justified from the fact that, at 1 m distance, free space and indoor path loss have approximately the same value.

The parameter n (known as the path loss exponent) indicates how fast the received power decays with distance ($n = 2$ in free space). Its value is usually obtained through experiment and is strongly environment-dependent. Values less than two were found for corridors and in-room propagation with line-of-sight (LOS) between transmitter and receiver; for densely furnished rooms, where the propagation includes non-line-of-sight (NLOS) and diffraction, values for the path-loss exponent were found to be around $n = 3$, or even higher. The parameter χ_σ is the shadowing factor; it is a zero-mean, normally-distributed statistical variable (with standard deviation σ), which takes into account of the deviation of the measured data from the modelled path loss.

A set of measurements were performed in the body-centric wireless lab [43] at Queen Mary, University of London (see Fig. 11), to characterise the propagation within a room for LOS between transmitter and receiver; the path loss exponent was found to be $n = 1.3$. Several propagation scenarios were considered in this work, and they are listed in Table VI.

TABLE VI
INDOOR PROPAGATION SCENARIOS USED IN THE STUDY OF PROPAGATION BETWEEN AN IMPLANTED RFID TAG AND AN OFF-BODY READER.

Scenario	Path loss exponent, n
Propagation in the body-centric sensor lab with LOS	1.30
Propagation within a room considering the presence of obstacles and losses from furniture [44]	3.00
Propagation within a floor of a building [42]	2.76
Propagation through one floor of a building [42]	4.19
Propagation through two floors of a building [42]	5.04
Propagation through three floors of a building [42]	5.22

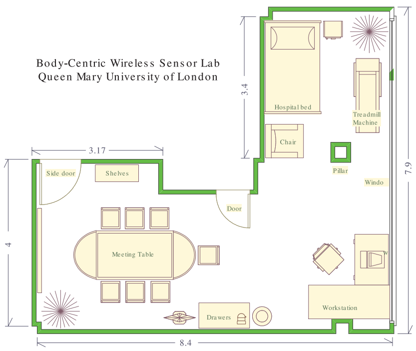


Fig. 11. Floor plan of the Body-centric wireless lab, at Queen Mary, University of London. The lab includes mock hospital suite.

A. Radio propagation of Passive RFID system

In passive and semi-passive (i.e., where the tag is powered for other functions, but acts as a passive tag for communications) systems, the tag needs to be powered from the reader. The most important system parameter is the maximum distance at which the RFID reader can detect the tag. This range is defined with respect to a certain read/write rate (percentage of successful reads/writes) [40].

In passive RFID systems, the tag receives energy from the reader's interrogation signal and it responds by sending a modulated signal. The reading range can be limited by the power required for activation of the tag, or by the sensitivity of the reader. The typical range of the reader sensitivity is between -60 dBm and -90 dBm, while the tag sensitivity ranges from -10 dBm to -15 dBm, depending on tag chip vendors [45].

The reader illuminates the tag and some power is back-scattered by the tag, dependent on the radar cross-section (RCS) of the tag [46], [47]. Considering a free space propagation model (justified from the short range of passive implanted RFID systems), the power back-scattered by the tag and received by the reader can be calculated as:

$$P_r = P_t\, G_t^2\, G_{imp}^2 \left(\frac{\lambda}{4\pi r}\right)^4 k \tag{4}$$

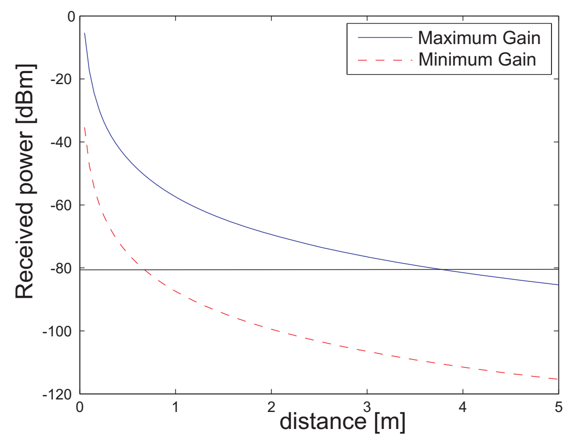
where k indicates the mismatch between tag chip and antenna. For this work, the antenna has been matched to the 50 Ω impedance of the coaxial cable used in the measurements. If the antenna is connected to a real chip, k assumes values ranging from 0.1 to 0.5, depending on chip vendors. The design can be modified in order to achieve an inductive input impedance [8] and improve the mis-match factor, but this is outside the scope of this work; thus, only an ideal case ($k = 1$) was considered.

The power received by the tag can be calculated as:

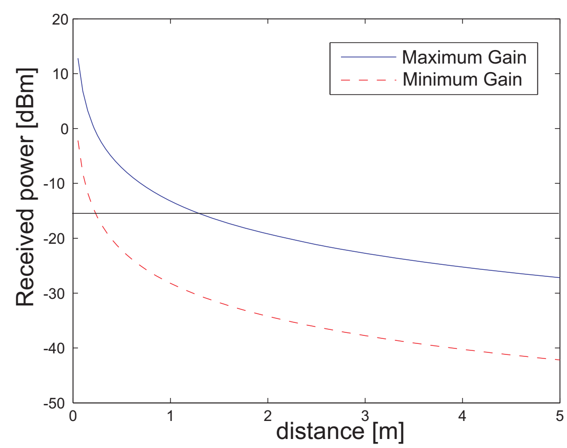
$$P_r = P_t\, G_t\, G_{imp} \left(\frac{\lambda}{4\pi r}\right)^2 \tag{5}$$

In [48], the European Telecommunications Standards Institute (ETSI) states that the maximum allowable effective radiated power (ERP) from the reader is 2 W, or:

$$P_t\, G_t \leq 2\mathrm{W} \tag{6}$$



(a) Power back-scattered from the tag



(b) Power received from the tag

Fig. 12. Propagation of the wave in passive systems; the communication range can be limited by the sensitivity of the reader, or by the activation power of the tag.

Fig. 12 shows the power back-scattered and received from the tag. The solid line is the best-case scenario, which means that the RFID reader is in the direction of maximum gain of the implanted antenna, while the dashed line indicates the case where the reader is in direction of minimum radiation. As explained in the introduction, the radiation pattern of body-worn and implanted antennas is strongly modified by the presence of the human body and presents a significant front-to-back ratio.

The maximum distance is low (1.3 m in the best-case scenario) and it is limited by the power of activation of the tag (Fig. 12(b)). RFID systems of this type could be used to activate an implanted sensor (assuming continuous monitoring is not needed) with a portable reader and retrieve real-time information with a high data-rate. Observing the very limited range in the worst-case scenario (dashed line), it is possible to conclude that a single-tag-single-reader system is not an optimum solution. The adoption of a double reader or double tag (e.g., one on each upper arm) system would overcome the poor radiation towards the body. A passive system of this type could be used, for example, for monitoring people entering or exiting a room; in this case, information about the subject entering the room could be quickly transferred to the reader. The advantage of a double-tag system would be cost: tags are intended to be cheap, even disposable, whereas readers are intended for longer-term use. However, a double-tag systems will only work if both tags are kept up-to-date with the same information, increasing system complexity; in addition, there may be greater objections by subjects to having more than one tag implanted. Thus, it is expected that double-reader systems will more common in the future; research will focus on optimising such systems, including the position of the tag in the body (from a system perspective, not merely the EM performance).

B. Radio propagation of Active RFID systems

Active RFID tags are self-powered and do not need to be in the range of the reader to be activated. They are used for long-range communications, but they are more expensive and, in general, bigger than passive tags. For implanted applications, the safety of the subject needs to be considered, and it has to be guaranteed that the is not greater than 2 $\mathrm{W{\cdot}kg}^{-1}$. Fig. 4 shows the SAR as a function of the power supplied to the antenna, calculated by applying an averaging mass of 10 g. The maximum input power for the tag antenna is around 20 mW (13 dBm).

The power received from the reader is calculated adopting (1). As we are considering a longer distance than in the passive case, the effect of the surrounding environment must be considered. The path loss is thus calculated with the log-distance model (3), as explained above.

As shown in Fig. 14, communication through different floors is impossible, because the power decays very rapidly; however, for propagation within the same floor, a communication range of 20 m is achievable with an input power of 0 dBm. A range of 40 m is achievable when applying a power of 10 dBm.

This system could be used for non-continuous operations. For example, if a patient has an implanted sensor and a critical status is revealed, an alarm can be generated and sent to the reader.

Fig. 13 shows results for the power received by the reader in various scenarios. It is clear that it is possible to cover a range of 10 m, for indoor propagation within our laboratory, with LOS between transmitter and receiver and a low transmit power (-20 dBm). These results show the feasibility of active implanted RFID for localisation and tracking purposes. In these calculations, the maximum gain value has been used. In multipath propagation scenarios, the system is less affected by the nulls in the tag antenna radiation pattern; however, a solution using two readers on opposite sides of the room (or the floor) increases the useful range. These active systems could be used for security purposes; for example, if the reader does not receive the signal from the tag, it means that the subject is no longer present in the observed region (i.e., the room or floor).

VI. Conclusions

Radio Frequency Identification (RFID) is a growing technology with the potential for reducing medical errors and improving the quality of healthcare in hospitals. The benefits include more secure and safe access in the healthcare environment, with the possibility, for example, to track patients, personnel and equipment, as well as to identify patients and their medications, reducing the risk of error in treatment.

In this paper we presented an overview of the challenges faced in antenna design, electromagnetic modelling and wave propagation, for the applications of RFID implants. The performance of a Ultra High Frequency (UHF) subcutaneous tag antenna was investigated numerically and validated with measurements results. Furthermore, the wave propagation between off-body reader and implanted tag was analysed in free space and in a scattered environment. Results demonstrated that, due to the body losses, the electrically-small size of the antenna and the directional radiation pattern, a passive tag solution allows a very limited communication range. If the tag is powered (active tag) with a limited power (-20 dBm), a maximum communication range of 10 m was calculated for propagation within a room.

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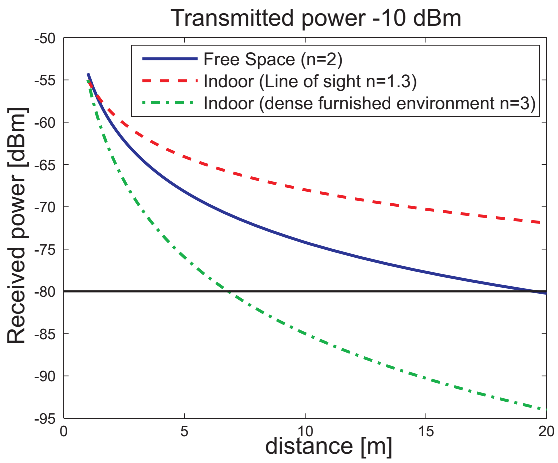
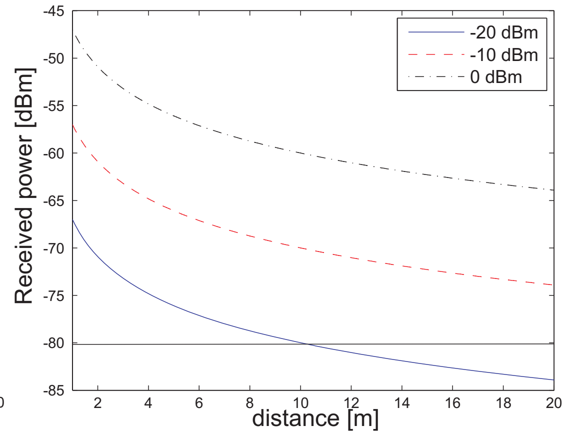
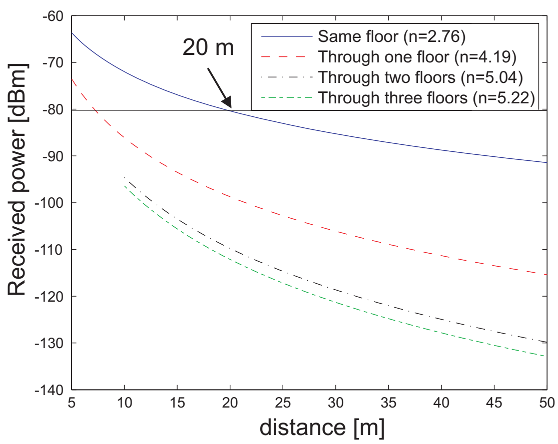


(a) Received power in indoor environment

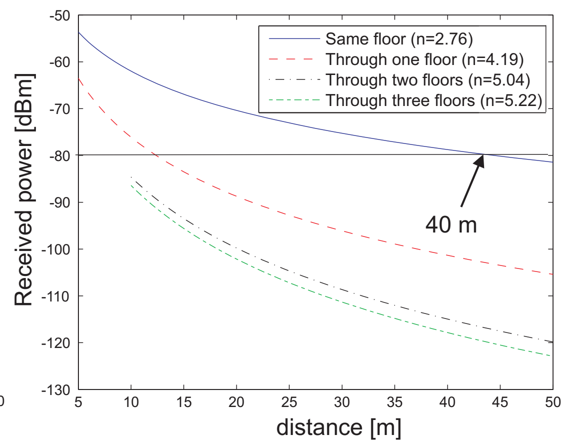


(b) Power received as a function of the distance in our lab ($n = 1.3$).

Fig. 13. Power received from the reader considering propagation within a room.



(a) Input Power 0 dBm



(b) Input Power 10 dBm

Fig. 14. Received power from the reader for propagation in a building.

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