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Tissue-Independent Implantable Antenna for In-Body Communications at 2.36 - 2.5 GHz

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Abstract—In this paper, a compact printed meandered folded dipole antenna with a volume of 114 mm^3 suitable for implantation in a range of different body tissue types with diverse electrical properties is presented for operation in the 2.36 - 2.4 GHz MBAN and 2.4 GHz ISM bands. Its performance was verified and compared against that of a wire dipole and slot loaded monopole antenna in an implant phantom testbed containing tissue equivalent liquids representing body tissues with high and low water content. The antenna was shown to maintain its return loss performance in the 2360 - 2400 MHz, 2400 - 2483.5 MHz and 2483.5 - 2500 MHz frequency bands with equivalent or better performance than a fundamental wire dipole despite having approximately half the physical length.

Index Terms—medical body area network, implantable medical devices, implantable antenna, body phantom

I. INTRODUCTION

THE use of Implantable Medical Devices (IMDs) has continued to rise in recent years due to the benefits they present in diagnosing and treating a wide range of medical conditions in a wide range of patients by providing real-time biotelemetric data. Recent examples of this include wireless implantable heart failure monitoring systems [1], brain-computer interface [2] and artificial bladder sphincter control devices [3].

Magnetic induction based techniques have been widely adapted for the majority of implant communication systems but the demand for external receiver location flexibility and greater bandwidth to facilitate more advanced functionality and security has seen the rapid introduction of active RF based implant communication systems. Initially, the Medical Implant Communication Service (MICS) band (402 - 405 MHz) was adopted but, with the demand for increased communication bandwidth, the Medical Body Area Network (MBAN) band (2360 MHz - 2400 MHz) was introduced.

There are many challenges associated with active RF implant communication systems, with one of the most important being power consumption. Many implant transceivers are battery powered which, once implanted into the patient, would require additional surgical procedures to replace. Therefore, creating an efficient communication link

is of utmost importance. Alongside intelligent communication protocols and low power circuit design, antenna radiation efficiency is a major factor to consider when implementing in-body implant to external receiver links.

Another critical requirement for implantable systems is that they are of minimal volume to allow safe, unobtrusive and comfortable implantation into a patient. Increasing the operating frequency of the system would reduce the size of the in-body antenna. However, propagation losses increase with frequency through biological tissue [4] but it can be argued that this can be offset with gains in antenna efficiency and available communication bandwidth at higher frequencies.

Furthermore, implant antennas are usually designed for operation in single tissue types with specific electrical properties [5]–[7]. The distribution and amount of these different tissues can vary greatly depending on specific patient attributes such as sex, age, weight, etc. An implant antenna's performance is strongly dependant on the tissue immediately surrounding it [8] with potential radiation pattern fragmentation, polarisation distortion, reduced radiation efficiency and changes in antenna input impedance [9].

Tissue within the human body can be categorised into two main groups, each with distinctive electrical properties. The first is high water content tissues, which have relatively high permittivities and conductivities such as muscle and vital organ tissue. The second is low water content tissues with low permittivities and conductivities such as bone and fats [10]. The conductivity of the tissue surrounding the implanted antenna can directly affect the bandwidth, radiation pattern and radiation efficiency. Likewise, as the permittivity of the surrounding tissue changes, so too will the wavelength within that tissue type, resulting in a resonant frequency shift of the implant antenna, which leads to unpredictable performance.

Therefore, in clinical applications an implant antenna designed to operate in one type of tissue in one patient may not perform sufficiently in another tissue type or patient.

In this paper we present for the first time a novel, compact, implantable antenna that maintains its return loss and radiation performance in a broad range of tissue types in the 2360 - 2400 MHz, 2400 - 2483.5 MHz and 2483.5 - 2500 MHz frequency bands. A multiple tissue equivalent liquid phantom testbed was used to verify the antennas performance in both a high and low water content tissue material. The efficiency of the new design was measured and compared to the performance of two other broadband implant antennas. The new antenna was shown to have good radiative properties in both types of tissues despite its relatively compact size and this was further

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verified through robust far field measurements emulating those that would be encountered in a clinical application. Section II outlines the new dynamic implant antenna environment concept with Section III detailing the design of the antenna and reference antennas. Section IV describes the measurement and simulation set up used to verify the antenna's performance with Section V detailing the measurement results. The paper concludes with a summary of the findings.

II. DYNAMIC IMPLANT ANTENNA ENVIRONMENT REQUIREMENTS

Future implant communication systems require an implant antenna to operate in an unpredictable, dynamic in-body environment. This means it is required to operate effectively in any tissue type that it is implanted within without any prior knowledge of that tissue and it must be able to continue to be effective even if the tissue environment local to the implant antenna changes over time. It is desirable for a dynamic implant antenna to have an omnidirectional radiation pattern as a changing environment within the body could cause the implanted device to move, as any pattern misalignment will degrade the communication link. Polarisation is a less critical factor for implant antennas as polarisation diversity is easier to implement in on or off body receiver antennas where antenna volume is not a limiting factor.

A dynamic implant antenna must be able to maintain its return loss performance in both high and low water content tissues. Narrowband antennas are not optimal for this as resonant frequency and bandwidth changes caused by changing permittivity and conductivity of the surrounding tissue can cause degradation of return loss performance. Mismatch losses must be minimised as much as possible for implant antennas as a good Total Radiation Efficiency (TRE) is extremely important if an implanted antenna is to maintain an efficient, low power consuming communication link.

A Printed Inverted F Antenna (PIFA), presented in [11], designed to operate in body tissue with averaged electrical properties suffered a maximum detuning of 49 MHz and 73 MHz in male and female anatomical models respectively. A maximum impedance mismatch difference of 22.6 dB and 28.5 dB respectively in both simulation scenarios is reported. Although this is antenna dependant, it highlights the potential return loss performance degradation that can occur with varying implant antenna placement throughout the body and the difference that can occur between patients of different sexes. Work presented in [12] showed that a "stationary" helical implant antenna, placed in the single tissue type for which its return loss was optimised, also experienced return loss fluctuations caused by natural body functions. The antenna was used for a cardiovascular pressure sensor in the left ventricle of three live pigs and its return loss was monitored over time. Breathing, heart rate and arrhythmia resulting in movement of the heart and the environment immediately surrounding the antenna caused the return loss to vary. Also, the antenna came into contact with not only heart tissue, but also blood and papillary muscle which caused up to a 4 MHz detuning effect, despite the antenna being placed directly in a location with properties for which it was optimally designed.

Another scenario that could occur is that the body composition could change with time or implant migration may occur. For example, an implant sensor designed to operate in low water content tissue such as breast fat could come into contact with a much higher water content tissue, such as a tumour. As the tumour grows and comes into closer contact or even grows around the implant antenna, the implanted antenna's performance could be greatly affected. In [13] a large scale study of the dielectric properties of 155 normal, cancer and benign breast tissue samples from 0.5 to 20 GHz is presented. It showed that the dielectric properties between normal adipose dominated breast tissue and malignant breast tissue can range up to a 10:1 contrast. For an antenna designed to operate in adipose breast tissue, close proximity to a mass with this contrast in dielectric properties would almost certainly cause a large reduction in its return loss performance.

III. PFMD ANTENNA

One potential solution to overcome the challenges associated with a multi-tissue implant antenna is to design an antenna with a sufficiently wide impedance bandwidth so as to maintain an in-band return loss of -10 dB (VSWR=2). However, this can prove difficult as not only does the resonant frequency change with the surrounding tissues permittivity but so too does the antennas bandwidth with the changing conductivity. This may cause marginal impedance matches in the tissues with the most contrasting electrical properties. Another solution is to design an antenna with multiple resonances which would resonate as the properties of the tissues surrounding the antenna change between those of high water content tissues and low water content tissues, maintaining its in-band return loss performance. This multi-resonance method for tissue matching was implemented in this work.

The proposed antenna is a Printed Folded Meandered Dipole (PFMD), a concept previously articulated in numerical study only in [15]. The layout of the PFMD can be seen in Fig. 1(a) with its dimensions detailed in Table II. The new antenna has a compact volume of 114 mm³, comparable with other previously published implant antennas shown in Appendix A, all of which are only operational in a single tissue type.

TABLE I: Dielectric Properties of Simulated Tissues at 2.38 GHz [14]

Tissue Type	Relative Permittivity	Conductivity (S/m)
Stomach	62.27	2.15
Cerebrospinal Fluid	66.35	3.39
Small Intestine Lumen	52.82	1.69
Skin	38.09	1.43
SAT	10.84	0.26
Visceral Fat	5.29	0.10
Muscle	52.82	1.69
Lung (Inflated)	20.52	0.78
Lung (Deflated)	48.48	1.64
Liver	43.15	1.64
Large Intestine Lumen	52.82	1.69
Kidney	52.90	2.37
Heart Muscle	54.96	2.20
White Brain Matter	36.25	1.18
Grey Brain Matter	49.03	1.76
Cortical Bone	11.42	0.38
Cancellous Bone	18.63	0.78
Bladder Wall	18.04	0.67

Fig. 1: Antenna Geometries: (a) PFMD (b) SLM (c) Insulated Wire Dipole

TABLE II: Implant Antenna Dimensions

Antenna	Dimensions (mm)												
	$l1$	$l2$	$l3$	$l4$	$w1$	$w2$	$d1$	$d2$	$d3$	$d4$	$d5$	$h1$	$h2$
PFMD	16.7	15.7	5.375	6.5	0.7	0.5	0.5	0.4	0.5	0.375	0.5	0.635	1.27
SLM	20	30	17.6	-	0.5	-	1	16	-	-	-	0.635	1.92
Wire Dipole	32	13.5	-	-	0.92	2.5	2.75	0.5	-	-	-	-	-

The antenna element was printed on Rogers RT/duroid 6010LM ($\epsilon_r = 10.2$, $\sigma = 0.0014$ S/m) substrate with the same material used for the superstrate layer. A substrate and superstrate is required to insulate the metallic element from the surrounding biological tissues for biocompatibility reasons and also to reduce the near field coupling to the surrounding lossy tissue, thus increasing radiation efficiency [16].

A thin substrate layer of 0.635 mm was used to reduce the near field losses by containing more of the near field in the low loss substrate but not completely isolating it from the higher permittivity surrounding tissue so that there was a significant increase in the resonant frequency of the antenna. Although the substrate used in this study is not biocompatible, it does have dielectric properties similar to biocompatible alumina 99.5% ($\epsilon_r = 9.8$, $\sigma = 1.0904\text{e-}7$

S/m). The substrate's dielectric properties also vary little with temperature [17], ensuring operation at body temperature. The radiating dipole element was first meandered to increase the electrical length of the antenna and then folded to produce the dual resonances, overcoming the bandwidth narrowing effect of the meandering and producing a wide in-tissue bandwidth. The substrate layers were glued together using cyanoacrylate glue. Finite Difference Time Domain (FDTD) simulation software Sim4Life by Zurich MedTech was used to simulate the performance of the PFMD in a number of tissue types. The PFMD model was placed in the centre of a 110 x 110 x 110 mm cube model with the dielectric properties of each tissue shown in Table I. The return loss performance of the PFMD antenna for each tissue is presented in Fig. 2.

Fig. 2: Wideband PFMD S_{11} Performance in (a) Low Water Content Tissue Types and (b) High Water Content Tissues

The return loss plots (Fig. 2) show that the PFMD maintained an in-band return loss of -10 dB or less in all tissues simulated with the exception of visceral fat. In these figures, the two modes of the antenna can be clearly observed.

The higher resonance (shown in-band in Fig. 2(b)) can be seen to resonate in the high water content tissues such as the vital organ tissues with cerebrospinal fluid causing the largest resonant frequency shift to lower frequencies, which can be attributed to having the highest relative permittivity and conductivity of the tissue types tested.

The lower resonance (shown in-band in Fig. 2(a)) can also be seen to resonate in low water content tissues such as cortical bone and Subcutaneous Adipose Tissue (SAT). The largest resonant frequency shift in low water content tissues was caused by visceral fat due to its much lower ($\epsilon_r = 5.29$, $\sigma = 0.1$ S/m) electrical properties, causing the PFMD to maintain a less than -6 dB return loss performance. This is not a critical issue however as SAT makes up the majority of adipose tissue in the human body. Both these resonant modes can be seen in the simulated surface current distributions in Fig. 4. Alongside the two general categories of high and low water content tissues, a third category of tissue can be seen which has intermediate properties to these two such as inflated lung, cancellous bone and urinary bladder wall. Nevertheless, the PFMD maintains an excellent input impedance match in tissues from this intermediate category due to its broadband performance caused by the close spacing of the two resonances.

The high and low water content resonant modes can be seen in the simulated surface current distributions in Fig. 4. The resonant frequencies of each mode can be adjusted by varying the length of the folded Arms A and B as shown in Fig. 5. With respect to Arms A and B having length equal to l_2 shown in Fig. 1(a) and equal substrate area, the resonant frequency of the lower resonance can be increased by reducing the length of Arm A and the resonant frequency of the upper resonance can be increased by reducing the length of Arm B. Input impedance matching is achieved by varying the spacing between tracks with bandwidth predominantly controlled by

track width.

To be able to compare the performance of the PFMD accurately, two other fundamental but somewhat larger implant antennas were developed. These were also designed to operate in the 2360 - 2500 MHz frequency range. To allow a valid comparison, both of these antennas also exhibit a wideband return loss performance and were designed to primarily operate in a high water content tissue such as muscle. No minimisation techniques were used in their design to allow comparison of the PFMD's performance with more fundamental type antennas.

A Slot Loaded Monopole (SLM) antenna was designed (Fig. 1(b)), which exhibits a simulated wideband performance in high water content media. The SLM is composed of three substrate layers of the relatively high permittivity, low loss Rogers RT/Duroid 6010LM. The top and bottom substrate layers are used to insulate the microstrip line and ground planes respectively from the surrounding tissue. The length of the microstrip line with respect to the ground plane slot was used to produce a wideband impedance match in muscle tissue as shown in Fig. 11(a). It is worth noting that the SLM has a significantly larger volume than the PFMD with a volume of 1143 mm^3 as shown in Table II.

As the PFMD is a dipole derivative design, it was logical to test its performance against a fundamental wire dipole designed for implantation in human tissue as shown in Fig. 1(c). The metallic dipole arms and central spacing were insulated in the biocompatible elastomer Silastic MDX4-4210 ($\epsilon_r = 3.3$, $\sigma = 0.01$ S/m). The wire dipole exhibits a strong simulated resonance when implanted in muscle as shown in Figure 11(b).

It is known that the cable can affect the accuracy of antenna measurements [18]. Currents induced on the outer sleeve of the cable can affect the overall radiating structure of the system, leading to possible errors in the estimation of antenna characteristics. This problem is further compounded when characterising implantable electrically small antennas. The radiative properties of balanced antennas without baluns fed by coaxial cables are particularly vulnerable to this [19].

Fig. 3: Manufactured Antennas: (a) Wire Dipole (b) SLM (Top View) (c) SLM (Bottom View) (d) PFMD

Fig. 4: Simulated PFMD Current Distribution Implanted in Muscle and SAT

Using ferrite chokes around the feeding coaxial cable was proposed as one solution to this problem but it is inefficient and could potentially affect the AUTs performance as well [20]. The solution that was chosen for the wire dipole and PFMD implant antennas was to place a small 0603 case style, 2.3 - 2.7 GHz chip balun at the input terminals of both antennas.

A u.fl coaxial port connector was then placed at the input of all three antennas as it offers a more compact

Fig. 5: PFMD S_{11} Variation With Varying Top and Bottom Arm Length

solution compared to other connectors such as SMA while still facilitating calibration at the antenna input terminals. The dipole antenna was also placed on a Rogers RT/Duroid 6010LM substrate to facilitate integration with the balun and u.fl circuit. A layer of hot melt glue ($\epsilon_r = 2.25$, $\sigma = 0.0005$ S/m) no more than 4 mm thick was used to insulate the u.fl connector and balun chip from the surrounding tissue and also to improve the mechanical strength of the connection. No coatings other than those indicated in Fig. 1 were added to the AUTs. All manufactured boards are shown in Fig. 3.

IV. SIMULATION AND MEASUREMENT SETUP

In this section, the methodology and rationale to robustly and rigorously characterise and validate the performance of implantable antennas and systems is described. Three key experimental measurement scenarios are adopted which use a suitable phantom test-bed, to address potential uncertainty:

- 1) Implant antenna total radiation efficiency measurements in a reverberation chamber [21].
- 2) Implant antenna far-field $|S_{21}|$ measurements.
- 3) Implant antenna stand-alone transmitter measurements (received power).

A. Implant Phantom Tissue Test Bed

To investigate the performance of all AUTs in different tissue types, a suitable human tissue representative phantom testbed was developed. Previously, implant antenna performance had been validated using phantom test beds which do not represent the human body sufficiently enough to guarantee measured performance when implanted in a live human patient. During the development stage of an implant antenna's design, it is essential that a high quality human tissue representative phantom test bed is used to allow accurate *in-vitro* analysis of the antennas performance. Ideally, the phantom's electrical and physical properties would resemble those of a human patient as closely as possible. Many tissue phantom liquids come in solid and gel forms [22], [23], but for implant antenna characterisation these are not ideal. To facilitate implantation into the tissue mimicking material, it is best if the tissue material is of low viscosity.

The container that was chosen to house the tissue equivalent liquid can be seen in Fig. 6 with dimensions 200 mm x 100 mm x 400 mm. It was designed to have a thickness that might approach the thickness of the largest tissue mass that would be found in the average human body when implanted in the centre of the phantom (eg. 50mm of muscle or SAT) in its Y axis [24]. The X and Z axis are sufficiently large enough that the predominant signal will propagate through the thinner Y axis as the tissue liquid losses will be significantly less on this path. The shape of the phantom container is also suggestive of the shape of the human torso, approximating the boundary conditions that would be encountered by an outward propagating wavefront. The phantom is hollow to allow it to be filled with the chosen tissue liquid with the phantom itself made out of Nylon 66 ($\epsilon_r = 3.4$, $\sigma = 0.04$ S/m) with a wall thickness of 2 mm [24].

Fig. 6: Phantom Measurement Setup

Fig. 7: Phantom with *in-vitro* AUT in Reverberation Chamber

During measurement, the AUT was connected to the measurement device coaxial cable via a 30 cm long u.fl pigtail with the SMA connection external to the phantom. The cable and AUT was connected to a 3.9 mm diameter nylon rod, lowered into the tissue liquid through the top phantom seal. The depth of the nylon rod in the liquid was then adjusted to place the antenna in the centre of the phantom. The u.fl cable was attached to the nylon rod using small cable ties. This set up can be seen in Figure 6 and would allow repeatable measurements and fast interchange between AUTs during measurements.

To test the dual resonance operation of the PFMD, two tissue types were selected which give a large deviation in antenna performance when implanted in that tissue type. Muscle was chosen for the high water content tissue and non-infiltrated fat (SAT) tissue was chosen for the low water content tissue. A muscle liquid previously developed by the

(a)

(b)

Fig. 8: (a) Tissue Material Measured Permittivity vs. Ideal Permittivity (b) Tissue Material Measured Conductivity vs. Ideal Conductivity

authors with accurate muscle properties in the frequency bands of interest was used for the high water content tissue measurements [24]. A SAT tissue liquid was then developed which has electrical properties almost identical to physical SAT in the measured frequency band of 400 MHz to 8000 MHz [25].

The permittivities and conductivities of both tissue liquids can be seen below in Figures 8(a) and 8(b) compared against ideal tissue properties found in the IT'IS tissue material parameter database [14]. The electrical properties of the tissues were measured using a commercially available, high accuracy DAK 3.5 dielectric probe kit by Speag. As the dielectric properties of both tissue liquids will vary with temperature, the measurement environments were maintained at an ambient temperature of 21.5 °C with the measurements shown in Figures 8(a) and 8(b) taken at this temperature.

B. Implant Antenna TRE Measurements in a Reverberation Chamber

An extremely important performance metric for any antenna is its Radiation Efficiency (RE) and TRE (the RE multiplied

by the impedance mismatch loss of the antenna). These metrics describe how much energy is effectively radiated from the antenna system and this is especially important for implanted device antennas where power consumption is especially important for such marginal links. The radiation efficiency of an implanted antenna is effectively dictated by the dimensions, electrical properties and geometry of the lossy media surrounding it. Therefore, the performance of an antenna is directly related to the geometry of the body it is implanted in and cannot be compared fairly against the performance of the same antenna in a different body. For example, even a slight change of electrical properties or geometry between two bodies can cause significantly contrasting efficiency measurements.

To rigorously compare the relative efficiency of each AUT the following testbed parameters were kept constant: electrical properties of the surrounding media, the geometry and dimensions of the surrounding media, position and orientation of AUT within that media, temperature and measurement equipment used. In this measurement, each AUT is placed in the same position in the centre of the phantom and its relative RE and TRE is then measured for the phantom filled with both muscle and SAT tissue liquids.

The efficiency measurements were carried out in a reverberation chamber manufactured by Bluetest.se with a Rohde and Schwarz ZVB-8 Vector Network Analyser (VNA) [21]. The efficiency of each AUT within a model of the phantom testbed was also determined by simulation for both tissue liquid types using Sim4Life.

The average return loss of each AUT was measured in the reverberation chamber using the set up shown in Figure 7. To ensure that the efficiencies that were being measured were above the noise floor of the measurement system, the AUT was replaced with a $50\ \Omega$ load and measured in the reverberation chamber to determine the noise floor of each measurement. This ensured that the relatively low antenna efficiencies were valid and above the noise floor of the measurement system.

C. Implant Antenna Far Field $|S_{21}|$ Measurements

S_{21} performance within an anechoic environment was also measured for each AUT. This is important as it can be used to develop a link budget for a realistic implant communication link as S_{21} measures the forward gain from an antenna connected to port 1 of the VNA (in this case the in-phantom antenna) and another antenna connected to port 2 (a dual polarized, wideband horn antenna manufactured by Flann Microwave was connected to port 2). As the dominant signal radiating from the phantom will be from the shortest path to the phantom surface from the implanted AUT through the lossy liquid, the front face of the phantom was placed facing the receive horn in an anechoic chamber. As the whole phantom itself can be considered as the radiator, the Fraunhofer distance from the front face of the phantom was calculated to be 2.54 m with the horn placed at 2.55 m from the front face of the phantom. Each antenna was then placed within the phantom at the same centric point with the receive horn focused on that point and $|S_{21}|$ was then recorded. A picture of this setup can be seen in Figure 10.

Fig. 9: PFMD Standalone Transmitter Unit

D. Implant Antenna Standalone Transmitter Measurements

To remove cable effects of spurious cable radiation and attempt to isolate the radiation performance of each AUT, a number of small, standalone, battery powered transmitter boards were developed. This allowed investigation of how each AUT would perform in a real world implant communication system as proximity to the device's circuitry and battery unit can affect an antenna's performance [26].

The transmitter is composed of a 2.25 - 2.5 GHz Voltage Controlled Oscillator (VCO) with a measured -0.3 dBm output. The PCB and antenna layouts are printed on the same Rogers RT/duroid 6010LM substrate and the board is powered by a single small form factor 20 mAh lithium polymer battery. To insulate the circuit from the tissue liquid, the electronic components (including the battery) were encased in Silastic. The antenna element was insulated from the liquid using the substrate and superstrate only. The board can then be switched on and off via reed switch, allowing activation without disturbing or damaging the silastic coating. The VCO could then be activated and placed in the centre of the phantom using the same method as the previous two measurements. Received power from the implanted transmitter was measured by a Tektronix RSA3408A real-time spectrum analyser connected to the receive horn at the same distance as the $|S_{21}|$ measurement.

Fig. 10: Anechoic Chamber Measurement Setup

Fig. 11: Return Loss Performance of AUTs Simulated and Measured in Both Muscle and SAT Tissue Equivalent Liquids: (a) SLM (b) Wire Dipole (c) PFMD

V. MEASUREMENT RESULTS

A. Wideband Return Loss Performance

Figs. 11(a) - 11(c) show the measured return loss performance of the AUTs in both muscle and SAT liquid for a frequency range of 800 - 3000 MHz vs. the simulated $|S_{11}|$ found during the antenna design process. For the PFMD shown in 11(c), the antenna maintained an excellent match with $|S_{11}| < -17$ dB in the band of interest in both SAT and muscle liquid. Variations from the simulated results can be attributed to a number of factors such as manufacturing irregularities (varying substrate/silastic thickness, solder roughness), difficulty in simulating the exact properties of the glue layer and the presence of the balun at the input of the dipole and PFMD antennas. The balun itself has a frequency dependant return loss which would account for why the lower resonance of the PFMD in muscle liquid is "filtered" out.

The SLM exhibits a strong broadband response with an in-band $|S_{11}| < -17$ dB in muscle liquid but this falls to an average $|S_{11}|$ of -6.0 dB in SAT liquid. With the increasing comparative wavelength in SAT, the resonance of the SLM has shifted up in frequency producing a poor match in the band of interest. The silastic insulated wire dipole exhibited a broadband match in muscle and SAT. This is due to its relatively thick coating of silastic which helps isolate it from the surrounding tissue liquid. This means that it can still remain in band as it suffers a smaller resonant frequency shift than the other two AUTs and it is still sufficiently broadband to stay matched. The downside of this however is that its largest dimension of 32 mm is not reduced by the presence of the relatively high permittivity tissues which could be an issue when designing an implantable device. Efforts to minimize this may reduce the bandwidth of the dipole and make it vulnerable to tissue dependant resonant frequency shifts.

Fig. 12: (a) RE in Muscle Liquid Phantom (b) TRE in Muscle Liquid Phantom (c) RE in SAT Liquid Phantom (d) TRE in SAT Liquid Phantom

B. Antenna Efficiency Performance

Figs. 12(a)-12(d) show the measured and simulated REs and TREs of the AUTs in both types of tissue liquid. Table III shows the band averaged (2.36 - 2.4 GHz) RE and TRE values of all AUTs in both types of liquid. It shows that the SLM is the most efficient in both tissue liquids due to its significantly larger volume in comparison to the other two AUTs. It has the highest RE in both scenarios but its TRE in SAT liquid is slightly less than the wire dipole, due to the mismatch losses encountered in the SAT liquid. The PFMD has a 1 dB higher RE and TRE than the wire dipole in muscle liquid despite having approximately half the maximum electrical length of the wire dipole. The wire dipole however has a higher RE and TRE in SAT than the PFMD. This can be attributed to the wavelength in SAT being significantly larger than in muscle, with the TRE gains produced by the substrate and improved match of the PFMD unable to compete with the simply larger

electrical length of the dipole, which is a major factor in determining the radiation efficiency of an antenna [27].

The RE and TRE simulation results for all AUTs in SAT liquid strongly agree with the measured results, with a maximum deviation of only 0.32 dB. The maximum deviation window was from 0.33 dB to 3.68 dB in muscle liquid. The PFMD RE simulation had the greatest deviation in measured results in comparison to simulation, with the Dipole RE simulation having the least deviation. This difference between measured and simulated efficiency value in muscle liquid may be attributed to positioning errors within the phantom during measurement. As the muscle liquid is a high loss medium, even a slight positioning error closer to the surface of the phantom from the centre can cause a significant increase in efficiency, which can be seen in the difference between measured and simulated results.

TABLE III: Band Averaged Radiation and Total Radiation Efficiencies in Both Tissue Liquids

Tissue	Antenna	Measurement (dB)		Simulation (dB)	
		Avg. RE	Avg. TRE	Avg. RE	Avg. TRE
Muscle	PFMD	-32.8	-32.9	-36.2	-36.6
	SLM	-32.2	-32.2	-34.4	-34.5
	Wire Dipole	-33.7	-33.9	-34.1	-34.6
SAT	PFMD	-18.1	-18.1	-17.8	-17.8
	SLM	-15.6	-16.2	-15.4	-16.5
	Wire Dipole	-16.4	-16.5	-16.4	-16.5

C. Antenna $|S_{21}|$ Performance

Fig. 13: (a) AUT S_{21} Performance in Muscle Liquid Phantom (b) AUT S_{21} Performance in SAT Liquid Phantom

Figs. 13(a) and 13(b) show the $|S_{21}|$ plots for all AUTs in both SAT and muscle liquid filled phantoms in an anechoic environment. The polarisation of the receive horn was also alternated between horizontal and vertical polarisations. As can be seen from Fig. 13(a), the SLM has the highest band averaged $|S_{21}|$ value in muscle of -74.3 dB when in co-polarisation (vertically orientated) which was to be expected due to its higher RE. The PFMD has an $|S_{21}|$ value of -79.5 dB, approximately the same as the wire dipoles value

of -79.6 dB. This agrees with the efficiency measurements as the PFMD also has a similar RE as the wire dipole. Figure 13(b) shows the AUTs $|S_{21}|$ performances in SAT with the SLM again having the highest band averaged $|S_{21}|$ value of -58.4 dB. Again, the PFMD and wire dipole antennas have very similar $|S_{21}|$ traces although the PFMD has a higher band averaged $|S_{21}|$ values of -63.3 dB compared with the wire dipole's -64.4 dB. Although the wire dipole has the higher peak $|S_{21}|$, this drops off with frequency through the measured bandwidth which is also apparent in the RE measurements.

D. Antenna Standalone Transmitter Performance

TABLE IV: Transmitter Calculation Values (dB)

Tissue	Measured Tx Pwr (dBm)	Cable Loss (dB)	Measured $ S_{21} $ (dB)	Calculated Rx Pwr (dB)	Measured Rx Pwr (dB)
Muscle	-0.3	-2.7	-81	-84	-82.7
SAT	-0.3	-2.7	-64.7	-67.7	-67.5

The implant transmitter measurements were used to verify that the PFMD performance stated in the radiation efficiency and $|S_{21}|$ measurements in Sections V-B and V-C are not heavily reliant on spurious cable radiation. The power received from the in-situ PFMD transmitter unit was calculated using

$$P_r[dB] = P_t[dBm] + |S_{21}|[dB] + L_c[dB] \quad (1)$$

with P_r the calculated received power from the transmitter unit at the spectrum analyser, P_t the transmitter output power and $|S_{21}|$ the value found in Section V-C for that tissue equivalent liquid at that transmitter frequency. The cable loss, L_c , between the receive horn antenna and spectrum analyser was measured independently using a VNA.

Table IV shows the values used to calculate the power received from the in-situ PFMD transmitter and the measured received power value. The calculated and measured values are in excellent agreement with the measured received power being only 1.27 dB and 0.2 dB lower than the calculated values in muscle and SAT liquid respectively, showing that the antenna is radiating with little or no cable effects. It also shows that the performance of an implantable AUT can be estimated accurately using the implant antenna testbed described in Section IV. This further shows that a radiation chamber can be used to validate the efficiency performance of implanted antennas in both high and low water content tissue, as having at least 30 cm of high water content tissue liquid surrounding the coaxial feed cable can sufficiently dampen surface currents so that their effect can become negligible.

VI. CONCLUSION

A novel antenna is presented and was shown to maintain an excellent return loss performance in both a high and low water content tissue emulating liquid, both with vastly contrasting electrical properties. A robust and repeatable implant antenna testbed methodology was proposed which measured implanted antenna radiation efficiency and total radiation efficiency along

with $|S_{21}|$ performance to properly verify its performance with comparison to other basic implant antennas. The PFMD was shown to have better efficiency than a fundamental wire dipole in muscle with only a slightly lower efficiency in SAT despite its much smaller physical length. It was also shown to perform better in a real world communication link scenario than the wire dipole through $|S_{21}|$ measurements in an anechoic, far field environment. Finally, the efficiency and $|S_{21}|$ measurements were verified through the measurement of received power from an implanted standalone transmitter unit, proving that the previous measurements are not heavily reliant on cable radiation for the efficiency and $|S_{21}|$ values observed. It also demonstrated that the PFMD can be integrated with a battery and transmitter circuitry, showing that the proposed antenna is a promising candidate for implantable systems in dynamic environments.

VII. ACKNOWLEDGEMENT

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VIII. APPENDIX A

TABLE V: Table of A Selection of Implant Antennas in Literature Ranked By Volume

Implantation Tissue	Volume (mm ³)	Dielectric Material	Frequency (MHz)	Ref
Skin	31.5	RT/duroid 6010	402 - 405	[28]
Skin	32.7	Alumina	2400 - 2480	[29]
			402 - 405	
Skin	52.5	RT/duroid 6010	433.1 - 434.8	[30]
			868 - 868.6	
			902 - 928	
Skin	67.8	Rogers 3010	2400 - 2480	[31]
Range of Tissues	114	RT/duroid 6010	2360 - 2400	Proposed Antenna
Skin	121.92	Rogers 3210	402 - 405	[32]
Skin	127	Rogers 3010	2400 - 2480	[33]
Skin	139.7	Rogers 3010	402 - 405	[34]
Skin	149.6	Rogers 3210	402 - 405	[35]
	179		402 - 405	
Skin	186.3	Rogers 3010	2400 - 2480	[36]
Skin	190.5	Rogers 3210	402 - 405	[37]
Skin	193.2	FR4	402 - 405	[38]
			402 - 405	
Skin	203.6	Rogers 3210	433 - 435	[39]
			402 - 405	
Muscle	254	Rogers 3210	433 - 435	[40]
			2400 - 2480	
Vitreous Humor	273.6	Rogers 3210	402 - 405	[41]
Muscle	791	Rogers 3210	402 - 405	[42]
Skin	823	RT/duroid 6010	402 - 405	[43]
2/3 Muscle			402 - 405	
Skin	1265.6	Rogers 3210	2400 - 2480	[7]
2/3 Muscle	3457.4	Macor	402 - 405	[6]
Skin	10240	Rogers 3210	402 - 405	[44]

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