

Article

Minimum Representative Human Body Model Size Determination for Link Budget Calculation in Implanted Medical Devices

Iñaki Ortego-Isasa ^{1,2,*} , Ainhoa Rezola ² , Yue Gao ³ , Xiaodong Chen ⁴ and Daniel Valderas ^{2,5,*} 

¹ TECNALIA, Basque Research and Technology Alliance (BRTA), Mikeletegi Pasealekua 2, 20009 Donostia-San Sebastián, Spain

² Department of Electrical and Electronic Engineering, TECNUN, University of Navarra, Mikeletegi Pasealekua 48, 20009 Donostia-San Sebastián, Spain; argarciandia@tecnun.es

³ Institute for Communication Systems/5GIC/6GIC, School of Computer Science and Electronic Engineering, University of Surrey, Guildford GU2 7XH, UK; yue.gao@surrey.ac.uk

⁴ School of Electronic Engineering and Computer Science, Queen Mary University of London, Mile End Rd, Bethnal Green, London E1 4NS, UK; xiaodong.chen@qmul.ac.uk

⁵ CEIT, Basque Research and Technology Alliance (BRTA), Manuel Lardizabal 13, 20018 Donostia-San Sebastián, Spain

* Correspondence: inaki.ortego@tecnalia.com (I.O.-I.); dvalderas@tecnun.es (D.V.)

Abstract: In this work, the optimum homogeneous phantom size for an equivalent whole-body electromagnetic (EM) modeling is calculated. This will enable the simple characterization of plane wave EM attenuation and far-field link budgets in Active Medical Implant (AMI) applications in the core region of the body for Industrial, Scientific, Medical and MedRadio frequency bands. A computational analysis is done to determine the optimum size in which a minimum phantom size reliably represents a whole-body situation for the corresponding frequency of operation, saving computer and laboratory resources. After the definition of a converge criterion, the computed minimum phantom size for subcutaneous applications, 0–10 mm insertion depth, is $355 \times 160 \times 255 \text{ mm}^3$ for 402 MHz and 868 MHz and a cube with a side of 100 mm and 50 mm for 2.45 GHz and 5.8 GHz, respectively. For deep AMI applications, 10–50 mm insertion depth, the dimensions are $355 \times 260 \times 255 \text{ mm}^3$ for 402 MHz and 868 MHz, and a cube with a side of 200 mm and 150 mm for 2.45 GHz and 5.8 GHz, respectively. A significant reduction in both computational and manufacturing resources for phantom development is thereby achieved. The verification of the model is performed by field measurements in phantoms made by aqueous solutions with sugar.

Keywords: electromagnetic propagation in absorbing media; biomedical applications of electromagnetic radiation; biomedical computing; biomedical measurements; implantable biomedical devices



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1. Introduction

The fast technological developments in electronics, biomaterials and computer science constitute an unprecedented impulse towards the improvement of actual medical devices and the promotion of new ones [1]. These new devices will have the capability to perform in vivo diagnostic and therapeutic intervention, improving the quality of life for many patients [2].

In this framework, the study of interaction between Electromagnetic (EM) fields and biological tissues, commonly called bioelectromagnetics, has become a relevant research topic. Its applications cover a very wide area within the healthcare system: therapeutic use (e.g., regeneration of tissues), diagnostic purposes (Magnetic Resonance Imaging—MRI), cancer and psychological disturbances research, etc. Considerable research has been carried out on the biological effects which are caused by exposure to RF radiation [3–7].

Understanding the underlying interaction mechanisms caused by EM fields is fundamental for evaluating the possible impact on biological tissues and being able to design the

link with wirelessly operated implanted medical devices (AMI) for improving the quality of life of patients [2]. In this respect, Transparency Market Research (TMR) estimates that the global implantable medical devices market will expand at a 4.6% compound annual growth rate (CAGR) between 2019 and 2027 [8].

In this context, human bodies are modeled by EM phantoms. A phantom is a physical or numerical model with the same electrical characteristics as the human body to accurately reproduce the effect of the body on electromagnetic radiation and vice versa. These models can be classified according to different criteria: (1) nature (physical and numerical/computer models) (2) composition (homogeneous and heterogeneous) and (3) geometry (planar, spherical, cylindrical and more realistic models based on MRI images). The right choice among them depends on the accuracy required by the application.

The physical phantoms are classified as solid-dry, semi-solid and liquid phantoms. Whenever possible, the latter is the most convenient approach, as a very reasonable trade-off between complexity and accuracy is achieved. Liquid phantoms are selected in this paper, as they allow the movement of an antenna-probe inside them to monitor the electromagnetic dosimetry. Moreover, they have been extensively employed as testbeds for standard mobile phone compliance with widespread acceptance [9,10]. The impact of their inherent homogeneous nature has been extensively reported in [11,12]. When employed to characterize implanted antennas [13], they may be enough to test their performance and potentially simplify the measurement setup [14]. The slight detuning experimented by an implanted antenna in a homogeneous phantom [15] may well be made up by the simplicity of the model, avoiding unnecessary calculations. This is particularly true when the target is not the design of the implanted antenna but field measurements.

The liquid is usually water plus several additional elements in order to control the permittivity and conductivity of the liquid. There are different liquid tissue models using sugar, salt, oil-in-water emulsions or non-ionic surfactants. Most common recipes add sugar for low frequencies and Diacetyl, Di-ethylene Glycol Butyl Ether (DGBE, sometimes abbreviated “Glycol”) or Ethoxylated p-tert-octylphenol (Triton® X-100) for higher frequencies (above 1 GHz) in order to control the permittivity [10,16]. Several manufacturers also provide off-the-shelf tissue-simulating liquids [17,18]. In some countries, the handling of these products requires the use of protective gear and specialized ventilation [19]. Thus, for the sake of simplicity, the interest in this study will be focused on liquid, homogeneous, water solvent and sugar-based phantoms as they are cost effective, simple to fabricate and do not need any special protective protocol. Their 900 MHz frequency limit [10] will be reviewed.

In order to provide a comprehensive solution, simplicity of corresponding homogeneous numerical phantoms should be consistently provided. It is known that numerical computation of body phantoms demands a high number of resources. A key factor in this respect is the phantom electrical size. Dismissing non-representative regions of the phantom, where the actual field is negligible, would alleviate these required resources. This is particularly meaningful for implants inserted into the trunk (e.g., pacemakers [20], spinal cord stimulators [21], gastro-intestinal stimulators [22] and central venous catheters [23], among others) where there are no clear criteria to define the minimum representative size at commercial RF frequencies. Extracting the minimum homogeneous phantom size representative of the whole body would build on the simplicity of sugar-based homogeneous phantoms. This will be done in the present paper by studying the convergence of the electric field attenuation inside a phantom with increasing size. The analysis will be done in a simulation model, corroborated by anechoic chamber measurements for a specific case. The bands to be analyzed are set to Industrial Scientific Medical (ISM) bands $f = 868$ MHz, $f = 2.45$ GHz and 5.8 GHz; and MedRadio $f = 402$ MHz for AMI [24].

The paper is divided as follows: after the introduction, the second section deals with the employed materials and methods. In the third section, the computational analysis to determine the minimum required size of the phantom, the corresponding phantom

characterization and the measurements at the anechoic chamber that will validate the computer model are shown. The last section describes the conclusions.

2. Materials and Methods

In this section, the computational model that has been built for the analysis is firstly described. Secondly, the methodology to fabricate a liquid phantom with sugar and the characterization procedure to determine its dielectric properties are presented. This is necessary to validate the computational model. Lastly, the setup and procedure followed for the measurement of a plane wave attenuation in the fabricated physical phantom is shown.

2.1. Computational Modeling

The particular trunk phantom that was used was modeled in CST Microwave Studio using the Finite Integral Technique (FIT) solving method (Figure 1).

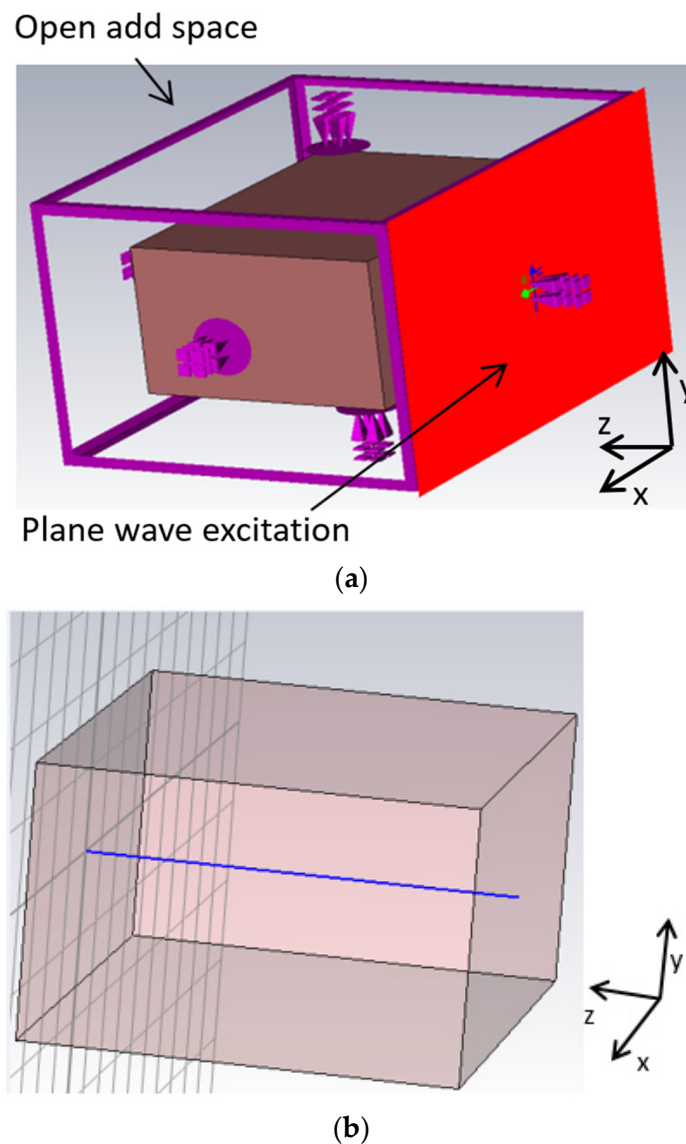


Figure 1. (a) Simulation setup showing the calculation domain (in the example with “open add space” boundary condition) and plane wave excitation. (b) Illustration of the direction along the z axis where the electric field is evaluated.

The dielectric and conductivity properties were finely tuned to the specific frequency of interest by a 4 Cole-Cole model based on the methodology explained in [25–27].

Plane wave excitation impinging upon the phantom along z direction was defined (Figure 1a). As will be shown, the phantom shell can be a rectangular prism different from a cube. In this case, once the field propagates inside the phantom, reflections from internal phantom/air interfaces will depend not only on the incident angle but also on its polarization [28]. Therefore, linear polarization variation (vertical or horizontal) was also included in the analysis. “Open add space” boundary conditions that mimic physical phantom construction were used.

E-field was computed along the z axis (Figure 1b). There was no receiving antenna in the model. It was assumed that a possible implanted antenna would not be electrically large, with a size according to the wavelength of operation. In this sense, it would be electrically much shorter than the phantom size that will be proposed. In order to compare with future measurements, a field normalization was done with respect to the transmitted field at the interface ($Z = 0$).

2.2. Phantom Fabrication and Characterization Procedure

As mentioned in the introduction, homogeneous aqueous phantoms with sugar were analyzed. In order to check the simulation results with a real model, different water- and sugar-based recipes have been fabricated for the mentioned frequency bands.

There are different measuring techniques for the characterization of the dielectric properties of materials [29]. In this research work, the 85070E high temperature dielectric probe [30] and E8362B PNA Network Analyzer [31] have been used (Figure 2). The main advantage of this method is its suitability for measuring liquids and its simplicity for broadband characterization of material properties [29].



Figure 2. Setup for measuring the dielectric properties of liquid phantoms. (a) PNA. (b) Dielectric probe. (c) Platform to move the sample. (d) Tank to store the phantom. (e) Heater and mixer to fabricate the phantom.

2.3. Characterization of Plane Wave Attenuation

In this subsection, the setup and the procedure that was followed for the characterization of electric field attenuation in the fabricated phantoms are shown. In Figure 3, the setup used in the anechoic chamber to measure the electromagnetic field inside the phantom is described.

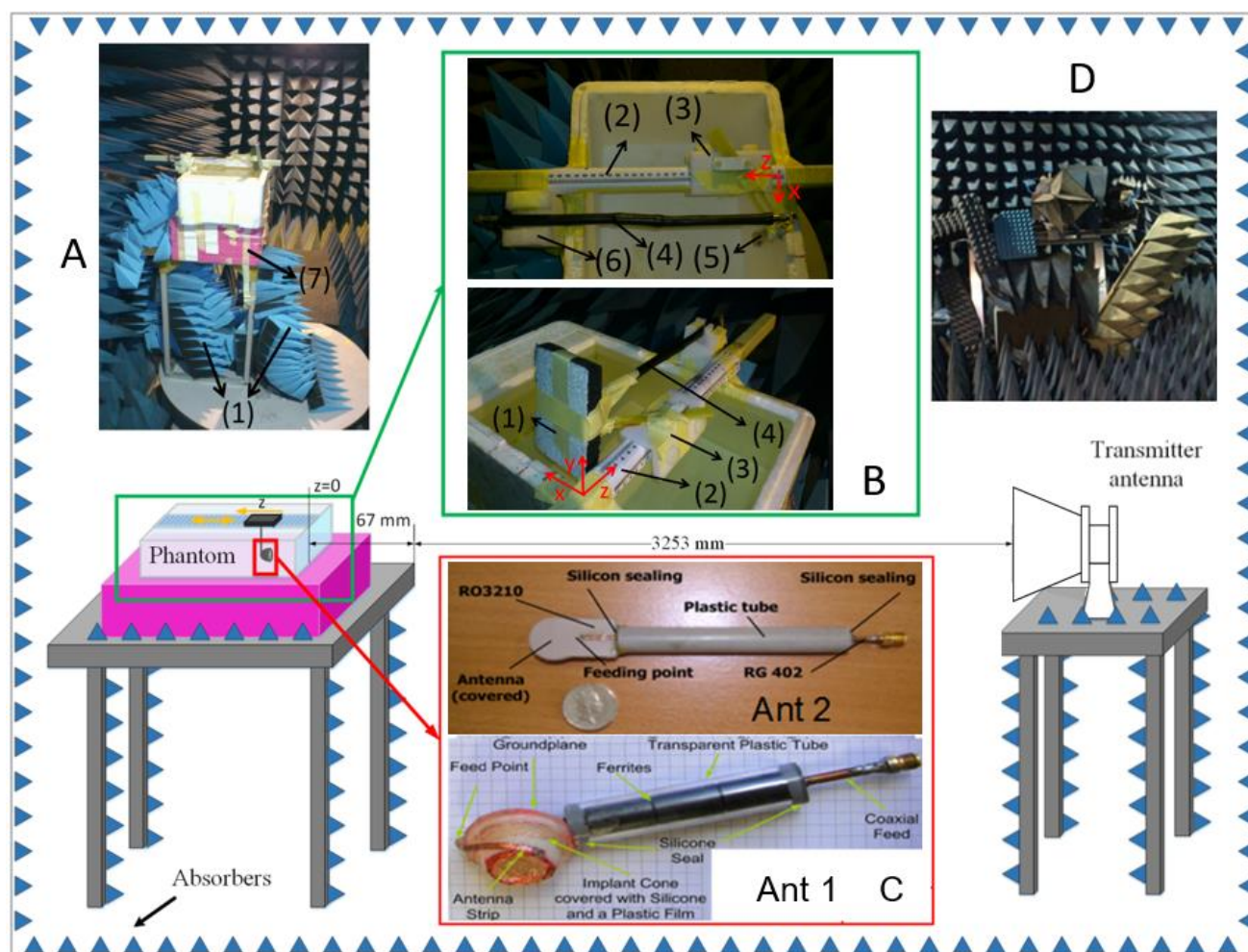


Figure 3. Setup used in Queen Mary University of London's anechoic chamber for the characterization of plane wave attenuation. (A,B) Test bed of the phantom with a probe antenna inside. (1) Absorbing panels, (2) ruler, (3) structure for moving the antenna, (4) feeding cable with ferrite-based absorbing material, (5) antenna, (6) backside foam structure, (7) polystyrene box to ensure that the transmitter antenna and the implanted antenna are at the same height. (C) The two implanted broadband antennas employed to cover the desired frequency ranges. (D) The transmitting horn antenna.

The transmitter horn antenna (model 3164-03 Dual Polarized Horn from ETS Lindgren [32]) (Figure 3D) allowed both horizontal and vertical polarization without the need of rotating the antenna physically. The receiver antenna 1 (ant1) was designed to work at 402 MHz (MedRadio band) and ISM 2.45 GHz and the antenna 2 (ant2) was designed to work at ISM 868 MHz [23,33]. Both antennas showed good matching at 5.8 GHz. The antennas in the described setup were used as field probes when inserted in the phantom. It has been corroborated by simulations that their normalized radiation patterns were kept mostly constant with penetration into the phantom (see Appendix A).

As can be seen, to ensure that no reflections were produced near the phantom, the supporting table was surrounded with absorbing materials (Figure 3A, (1)). The impact of the feeding cable on the measurement was reduced by several means: firstly by reducing the cross section exposed to the plane wave on the z axis (see axes in Figure 1), secondly by covering the feeding cable with a ferrite-based absorber (Figure 3B, (4)) and thirdly by placing a foam-based absorber in front of it (Figure 3B, (1)). It was very important that the plane wave impinged upon the cable cross section and not transversally. This guaranteed that the disturbance in the measurement results was kept under a certain limit. To ensure that cable position was correct, a foam structure was placed along the backside of the phantom (Figure 3B, (6)). The ruler (Figure 3B, (2)) and the structure to move the antenna

along (Figure 3B, (3)) were used to measure the insertion depth of the antenna into the phantom in the direction of the plane wave radiation (z axis). They were not metallic.

With this setup, ant1 and ant2 were alternatively placed within the phantom at a certain distance from the front interface ($Z = 0$) and the S_{21} was measured with respect to the transmitting antenna located at 3.32 m (Figure 3). Implanted and transmitting antennas were placed at the same height. A procedure was followed to measure the S_{21} parameter for both polarizations for each position, while progressively introducing ant1 and ant2 in the liquid phantom every 2 mm (for 2.45 GHz and 5.8 GHz) and every 3 mm (for 402 MHz and 868 MHz) in the z direction. A change in polarization was made in the transmitter side. Normalization was performed with respect to the measurement on the interface of the phantom. This was done to de-embed the transmitted field magnitude (different with operational frequencies) and the effect of the probe antennas for comparison with the model. In all the measurements, the S_{11} parameter was checked to ensure that the antenna was matched and that the obtained results were, consequently, reliable.

3. Results

3.1. Computational Analysis to Determine the Optimum Phantom Size

The minimum phantom size that reliably represents a whole body for implant communication and EM attenuation is addressed in this subsection according to the model described in Section 2.1. An iterative increase of its size was performed until field convergence.

In order to set the direction for the size increment, a general scenario was analyzed. The electric field distribution in vertical polarization when a 1 V/m (peak) plane wave impinges upon one side of a $355 \times 160 \times 255 \text{ mm}^3$ phantom with muscle theoretical properties [25–27] is shown in Figure 4. It illustrates with the same scale the reflection at the boundaries (outside and inside) and the penetration into the phantom at the different commercial frequencies present in this study.

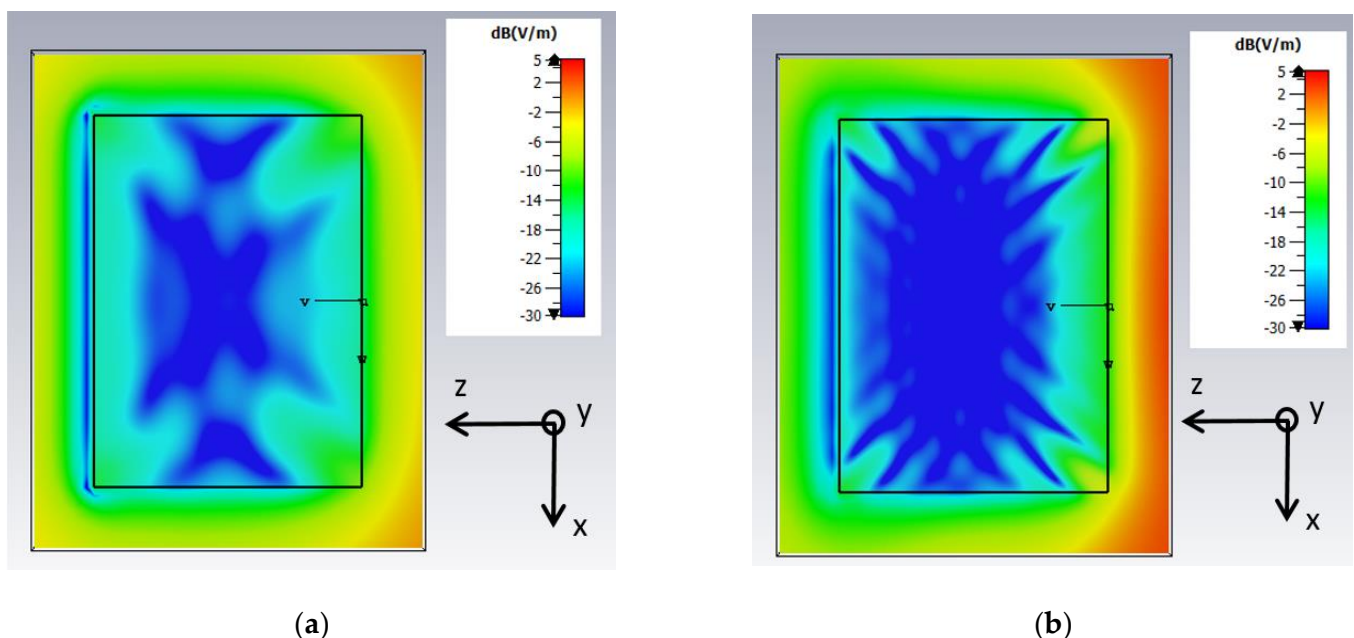


Figure 4. Cont.

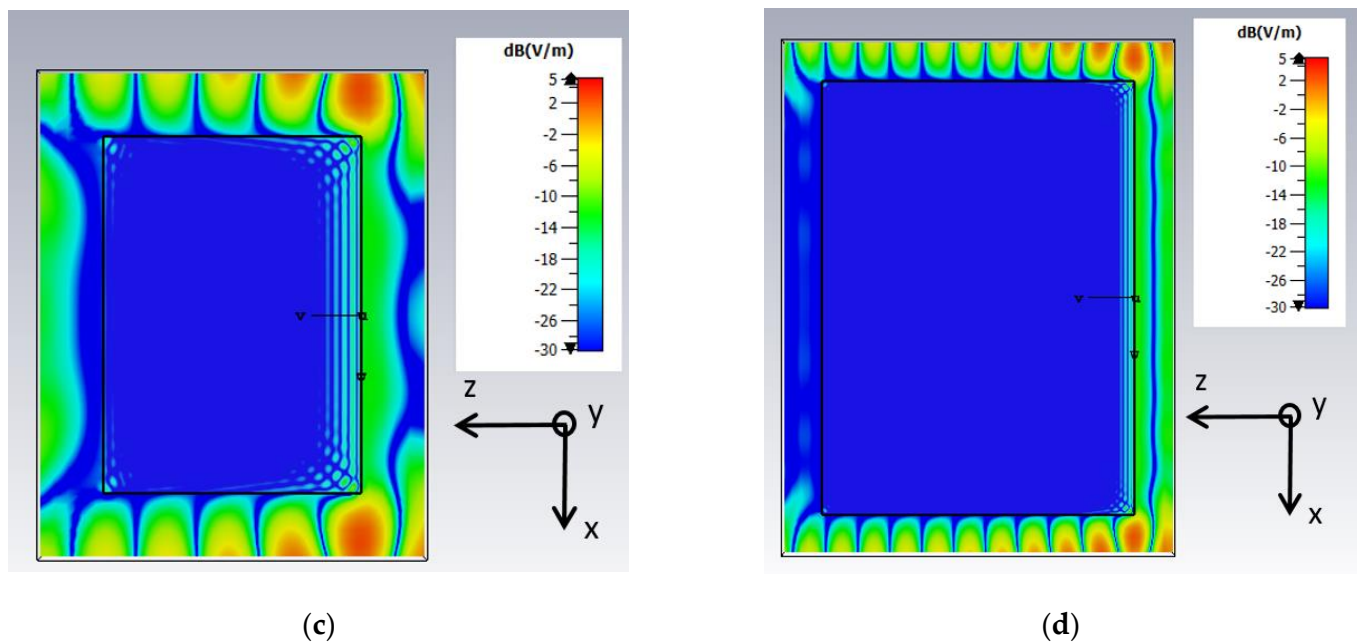


Figure 4. Top view ($355 \text{ mm} \times 255 \text{ mm}$) of simulated electric field distribution of a $355 \text{ mm} \times 160 \text{ mm} \times 255 \text{ mm}$ trunk phantom model: (a) $f = 402 \text{ MHz}$, (b) $f = 868 \text{ MHz}$, (c) $f = 2.45 \text{ GHz}$ and (d) $f = 5.8 \text{ GHz}$.

It was observed that, at 402 MHz and at 868 MHz, internal reflections became significant and the dimensions of the phantom should be increased to obtain a representative size of the whole body. However, the top view in Figure 4 is already considered a limit for the cross section of a representative human torso. Therefore, the geometrical dimension to be increased to pursue convergence will be done in the y direction (height). The chosen step was 100 mm. On the other hand, for the upper frequencies, at 2.45 GHz and at 5.8 GHz, the convergence was expected to happen before reaching this condition, and a cubical increment with a step of 50 mm will be performed in the three dimensions at the same time, as long as 355 mm and 255 mm limits are not overcome for x and z directions. By truncating at a certain size defined by a convergence criterion, computational and experimental resources will be saved.

Figures 5 and 6 represent the simulated normalized electric field amplitudes with insertion distance at the four frequencies with tissue theoretical characteristics [25–27]. In all cases, the incident field was along the z direction, as mentioned. The difference between horizontal and vertical polarization was negligible due to the cubical symmetry of the phantom for the higher frequency cases (Figure 6), as expected.

Convergence with increasing size was then analyzed. The criterion established to define the optimum size of the phantom was by setting a maximum 2 dB difference threshold limit at any point of the region of interest between two consecutive plots. For the lower frequency cases, the worst polarization case was considered. The 2 dB criterion was separately applied in two regions for the same plots (Figures 5 and 6) for subcutaneous (0–10 mm insertion depth) and deep (10–50 mm insertion depth) AMI operation. For example, for a subcutaneous implant, the 2 dB criterion could be met by a 100 mm^3 size phantom, whereas the same frequency may well need a more extensive phantom to accomplish the same criterion up to a 50 mm insertion length (deep implant). The obtained dimensions for each frequency and cases are summarized in Tables 1 and 2. A ratio of field attenuation with insertion distance is also provided. As expected, it increases with frequency.

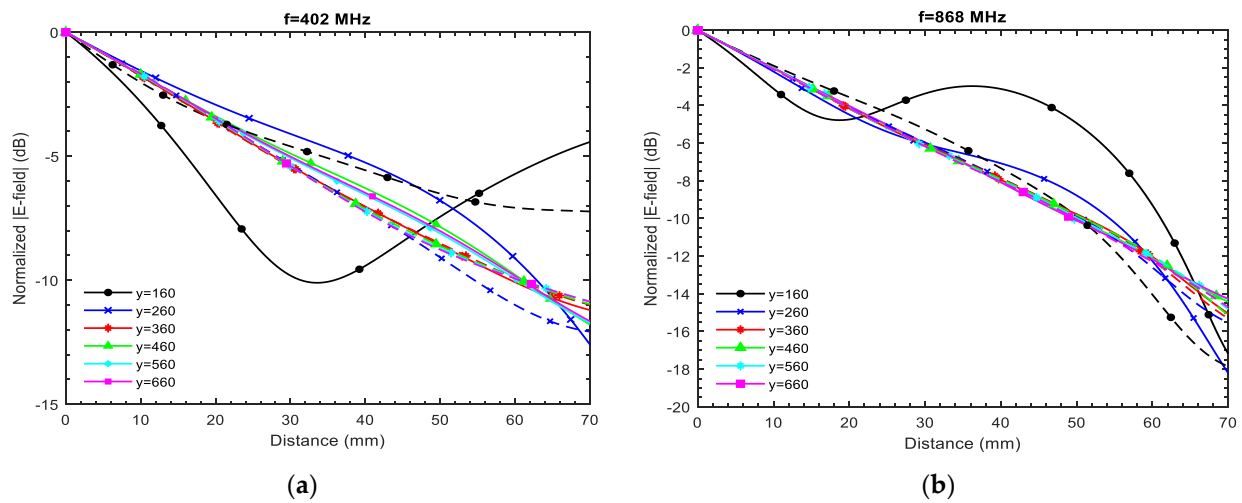


Figure 5. Normalized electric field vs. insertion distance (z axis) computation using CST for a phantom dimension of (355 × Y × 255) mm. H polarization (continuous line) and V polarization (dashed line): (a) f = 402 MHz and (b) f = 868 MHz.

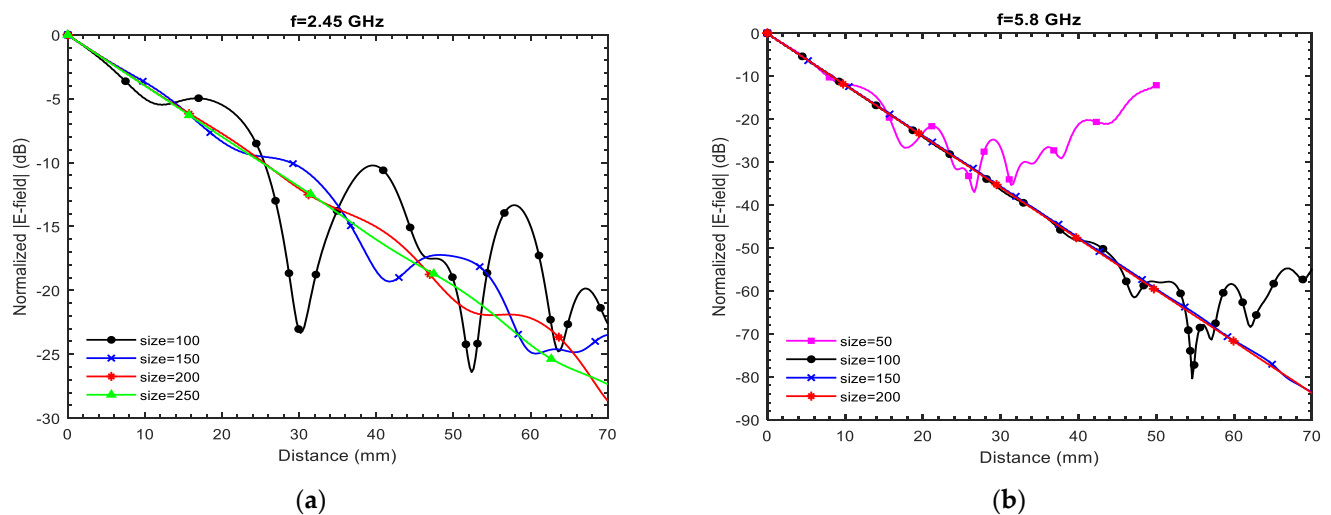


Figure 6. Normalized electric field vs. insertion distance (z axis) computation using CST for a phantom dimension of (size mm × size mm × size mm). (a) f = 2.45 GHz and (b) f = 5.8 GHz.

Table 1. Minimum phantom size that reliably represents an entire body for implant communication at a 10 mm to 50 mm insertion depth and corresponding absorption rate.

Freq [GHz]	Size [mm]			Absorption [dB/mm]
	x	y	z	
0.402	355	260	255	0.16
0.868	355	260	255	0.2
2.45	200	200	200	0.4
5.8	150	150	150	1.2

The size reductions with respect to a full body model estimated as 65 dm³ [34,35] are summarized in Table 3. It can be observed that important savings of manufacturing resources for phantom implementation was achieved in all cases and this reduction increased with frequency. A minimum volume reduction of 49.8% at 402 MHz for deep implants and up to a maximum of 99.8% at 5.8 GHz for subcutaneous systems with respect to the entire body was obtained.

Table 2. Minimum phantom size that reliably represents an entire body for implant communication and EM attenuation for subcutaneous applications (0–10 mm insertion depth). The average absorption rate is the same as for deep implants.

Freq [GHz]	Size [mm]		
	x	y	z
0.402	355	160	255
0.868	355	160	255
2.45	100	100	100
5.8	50	50	50

Table 3. Volume needed for the phantom for each case and reduction respect to whole body reference. Subcutaneous (S), Deep implant (D).

Freq [GHz]	Case Subcutaneous (S) Deep Implant (D)	dm ³	Volume Reduction with Respect to References [34,35] in. [%]
0.402	S	14.484	77.7
0.402	D	32.589	49.8
0.868	S	14.484	77.7
0.868	D	23.536	63.8
2.45	S	1	98.5
2.45	D	8	87.7
5.8	S	0.125	99.8
5.8	D	3.375	94.8

3.2. Measurements and Model Validation

Once the analysis had been done in the simulation model, simulated results were corroborated by anechoic chamber measurements for a specific case, following the methods described in Sections 2.2 and 2.3.

3.2.1. Phantom Dielectric Material Characterization

The results of the characterization of dielectric properties of the muscle homogeneous phantom are shown in this subsection. A study on the influence of sugar concentration in water's dielectric constant and conductivity has been carried out to fabricate a liquid that mimics muscle tissue. In Figure 7, the obtained values for different sugar concentrations are shown. Once the target real permittivity was achieved, mostly between 40% and 50% concentrations at low frequencies, it could be observed that beyond 1.6 GHz, the conductivity was higher than the one of the muscle. The opposite happened below this limit.

In this manner, liquid sugar-based phantoms are conducted for each of the studied frequency bands, where real permittivity matching is prioritized (Table 4).

Table 4. Fabricated muscle phantom properties for the frequencies of interest.

Frequency (GHz)	Target Properties [25–27]		Measured Phantom Properties		Phantom Material
	ϵ_r	σ	ϵ_r	σ	Sugar (%)
0.402	57.1	0.79	57	0.4	51
0.868	55.1	0.93	54	0.75	51
2.45	52.7	1.74	52.5	2.5	41
5.8	48.5	4.96	48.8	7.5	29

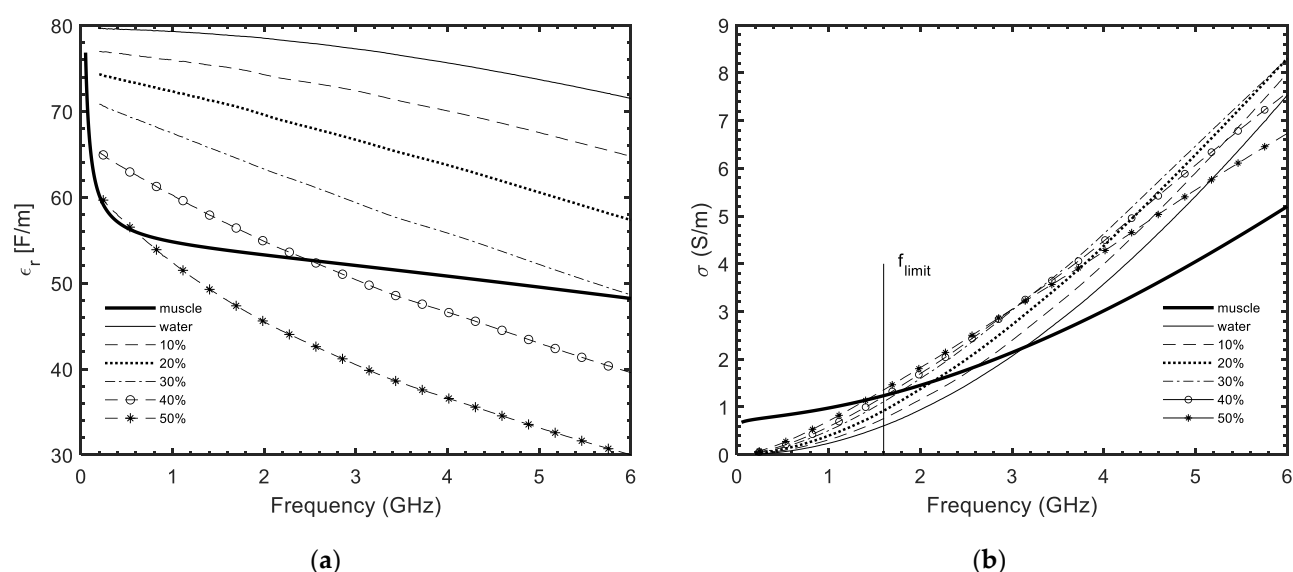


Figure 7. (a) Measured real part of permittivity (black lines) and (b) conductivity (grey lines) for different sugar concentration liquids and muscle theoretical values [25–27].

As expected, higher/lower conductivity values were obtained above/below the aforementioned limit of 1.6 GHz. However, for simplicity, they were considered as references for computer model validation as long as the numerical model imported these consistent properties for comparison. Therefore, emphasis was given to the coherence of the numerical-experimental validation. Once the results are verified, theoretical permittivity values corresponding to muscle properties can be adopted without loss of generality.

3.2.2. Computer Model Validation

A 355 mm × 160 mm × 255 mm polystyrene tank was filled with the corresponding tissue mimicking liquids of Table 4 as a reference design. At lower frequencies (402 MHz, 868 MHz), the chosen height y (160 mm) was below the minimum threshold defined by Table 1 for deep implants. Oscillations coming from reflections on the boundaries due to the electrical small phantom size should be expected. This phenomenon should be captured by the probe antennas inserted in the physical phantom. If the phantom is electrically large with dimensions mainly above the thresholds dictated by Table 1, as the present case for the upper frequencies 2.45 GHz and 5.8 GHz, the tendency should be decreasingly linear with minimum oscillations, regardless of the type of antenna.

In Figure 8, the measured results for the characterization of plane wave attenuation for the proposed phantom size with the liquids corresponding to Table 4 and the setup described in Section 2.3 are shown for both antennas and both polarizations for each position. Matching of the antennas was corroborated in each situation (Figure 9), disregarding antenna 2 at 402 MHz in Figure 8a, as expected.

The performances of the antennas at 2.45 GHz (Figure 8c) and 5.8 GHz (Figure 8d) followed the same trend in all cases. In each case, the difference between the antennas was due to the gain variety and polarization sensitivity. For distances larger than 60 mm at 2.45 GHz and 20 mm at 5.8 GHz, the measured S_{21} parameter was below -110 dB and became erratic in at least one polarization, so this region was neglected as noise. The four plots mainly overlapped for each frequency (2.4 GHz, 5.8 GHz) once the values were normalized by their respective maximum (Figure 10). This demonstrated that the antenna and its polarization did not have an impact on the normalized expected results for these high frequencies and that the phantom was representative of the whole body.

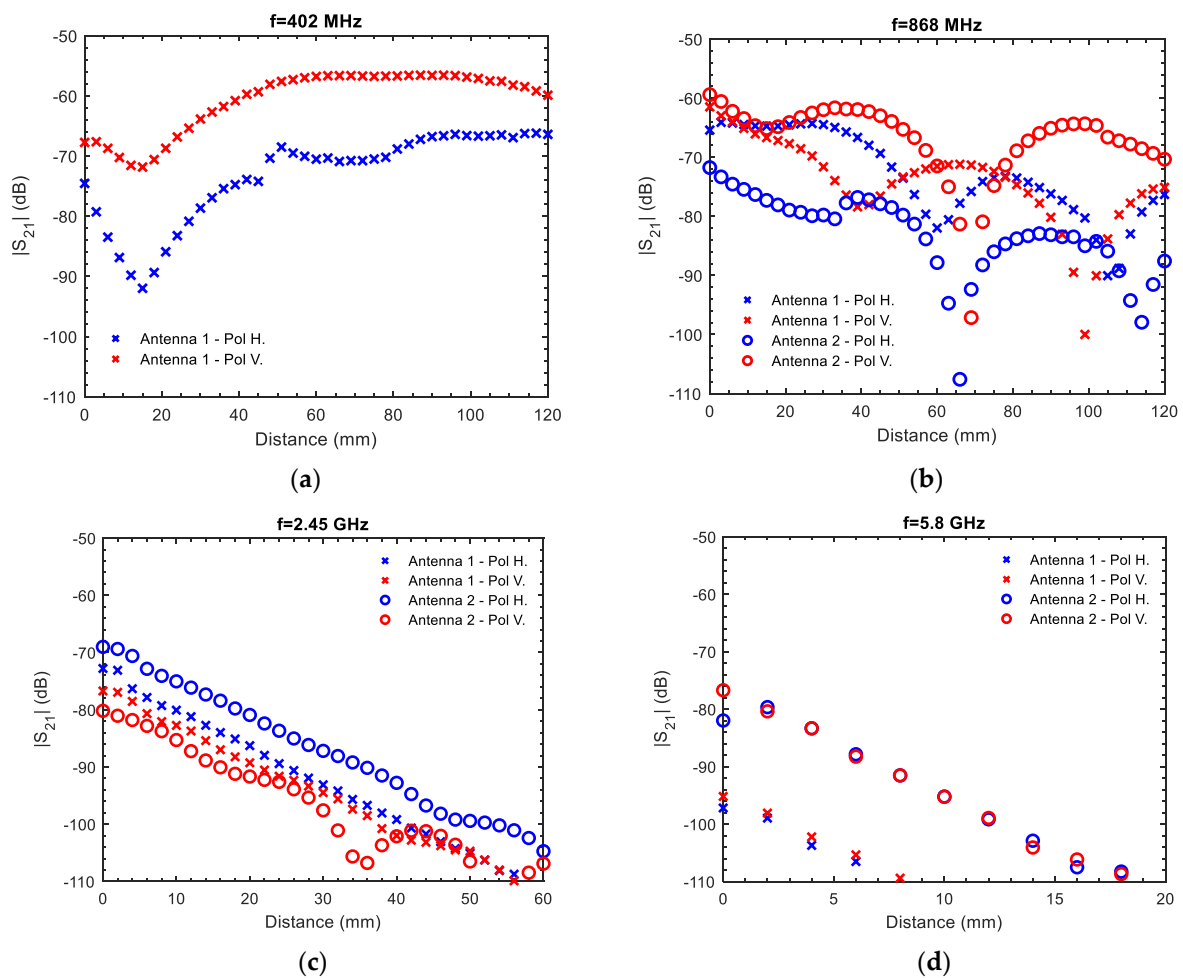


Figure 8. S_{21} dB vs. distance for the different antennas and polarizations: (a) $f = 402$ MHz, (b) $f = 868$ MHz, (c) $f = 2.45$ GHz and (d) $f = 5.8$ GHz.

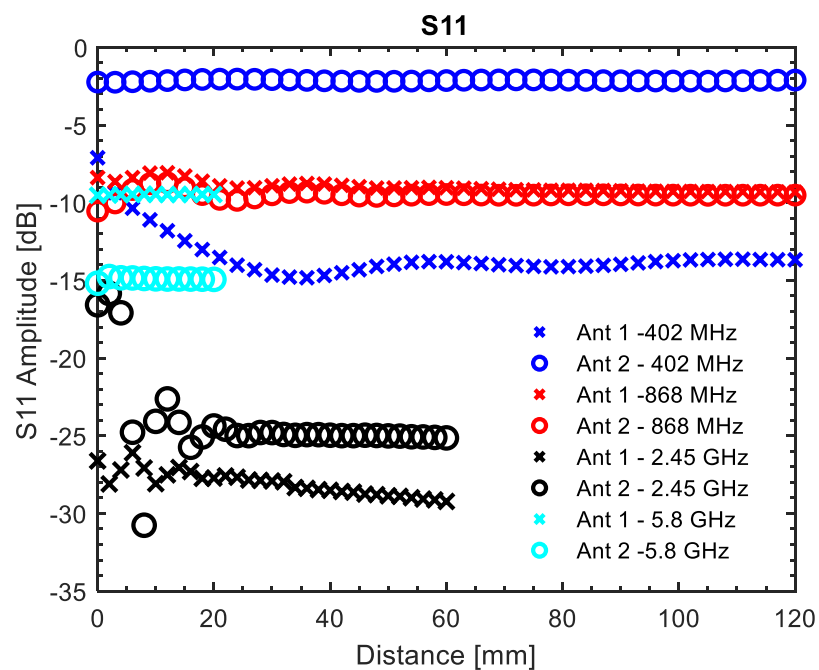


Figure 9. S_{11} dB vs. distance for each antenna at different frequencies.

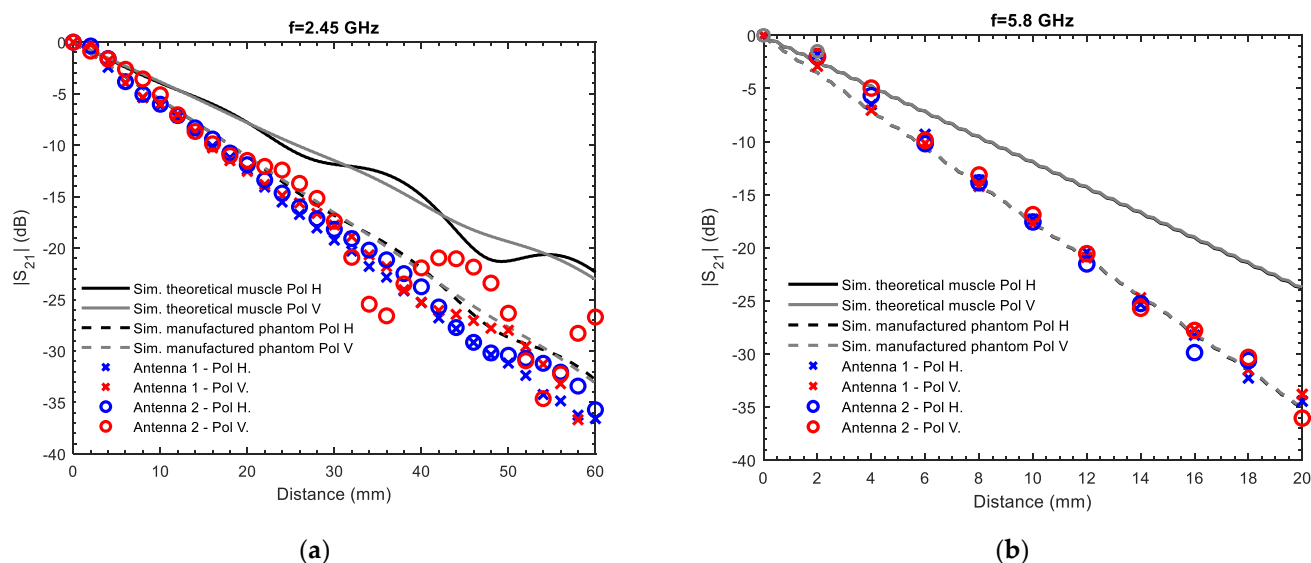


Figure 10. Normalized measurements vs. simulation (using theoretical and manufactured muscle dielectric properties). (a) $f = 2.45$ GHz and (b) $f = 5.8$ GHz.

A good agreement between the measurements and simulations can be observed when the properties that were used in the simulation were the ones of the fabricated phantom (Figure 10, markers and dashed lines, respectively). When real (theoretical) [25–27] properties were imported in the model, as in the previous section, a different slope with respect to the measured results was observed (Figure 10, solid lines), as expected. The difference between the real (theoretical) and the phantom’s conductivities explains the variability of the plane wave attenuation.

On the other hand, at 402 MHz and 868 MHz, the tendency was not consistent. The dips were a sign of internal reflections, which were not negligible for lower frequencies as the phantom exhibited a small electrical size. The internal reflection phenomena at the phantom boundaries made the radiation pattern variability between antenna 1 and 2 have an impact on the results. This is consistent with a phantom size smaller than the indicated by Table 1. Moreover, the anechoic chamber was prepared to work efficiently for frequencies above 1 GHz.

According to Section 2, in the computational model, ideal probes were used inside the phantom to monitor the field. The equivalence “antenna in the measurement–probe in the model” was only fulfilled for the 2.45 GHz and 5.8 GHz frequencies where the phantom was above the dimensions stated by Table 1 for deep implants: convergence was achieved and reflections were kept at a low level. In Appendix B, simulations showing the effect of using “open add space” boundary conditions vs. using an unlimited phantom in x and y axes (“periodic” boundary conditions) are shown to reinforce this idea.

The results certify that the methodology and the simulation tool were working properly with the aforementioned assumptions. Consequently, the computer model was representative of a whole body once the theoretical dielectric values [25–27] were employed in the analysis (as in Section 3.1) and it allowed to extract conclusions about the significant size of the phantom depending on the operational frequency.

4. Conclusions

In this work, a methodology to calculate the optimum homogeneous phantom size for plane wave electromagnetic (EM) attenuation characterization in the human torso is defined. It is applicable for far-field link budgets calculations in Active Medical Implants (AMI) at Industrial Scientific Medical (ISM) and MedRadio frequency bands.

Increasing the size of the defined numerical phantom until field convergence is achieved poses a convenient strategy to truncate the dimensions that represent the whole body.

By applying a 2 dB convergence criterion, a phantom size of 355 mm × 160 mm × 255 mm for 402 MHz and 868 MHz frequency bands and 100 mm³ and 50 mm³ cubic phantoms for 2.45 GHz and 5.8 GHz frequency bands, respectively, are recommended for subcutaneous applications (0–10 mm insertion).

Likewise, for deep implanted applications (10–50 mm insertion), a size of 355 mm × 260 mm × 255 mm for 402 MHz and 868 MHz and a cube of 200 mm³ and 150 mm³ for 2.45 GHz and 5.8 GHz frequency bands, respectively, are recommended.

Very significant reductions of the higher frequencies are obtained with respect to a full body model: between 49.8% and 77.7% for the lower frequency cases and between 87.7% and 99.8% for the higher frequency cases. Thus, solid foundations to reduce the computational and manufacturing resources to implement representative EM phantoms are provided.

The measured field results by two implanted broadband probe antennas into simplified water and sugar liquid phantoms showed virtually the same values when normalized by the maximum field at the phantom/air interface as soon as the phantom size is above the specified limits.

These measured results were compared with the model predictions and certified that they are independent from the implanted antenna as long as the latter was not electrically large.

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Conflicts of Interest: The authors declare no conflict of interest.

Appendix A

In this appendix, the simulations that illustrate that the two antenna's normalized radiation patterns are kept mostly constant with penetration into the phantom along short distances are shown. The dimensions of the phantom are the ones that were taken for verification purposes in measurements (355 mm × 160 mm × 255 mm). Its properties are the ones of the measured phantom (Table 4). The antennas are placed in the center of the phantom according to Figure A1.

Simulated normalized gain patterns at each particular frequency with insertion distance variation are shown in Figures A2 and A3 to corroborate their stability. The $\theta = 180^\circ$ direction indicates the direction pointing at the air–phantom interface according to the coordinate system of Figure 1. Axis x is labeled on the patterns.

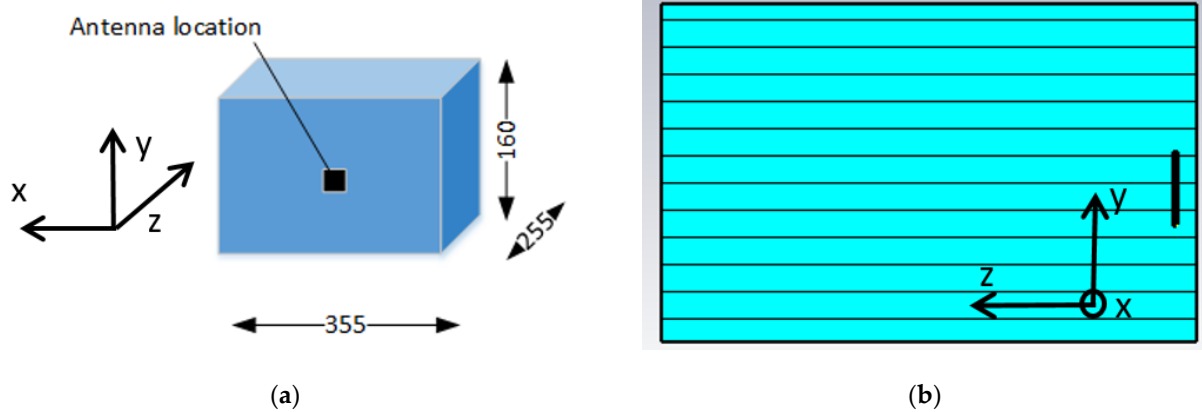


Figure A1. Placement of the antenna in the phantom. (a) Perspective and (b) side view. Units in mm.

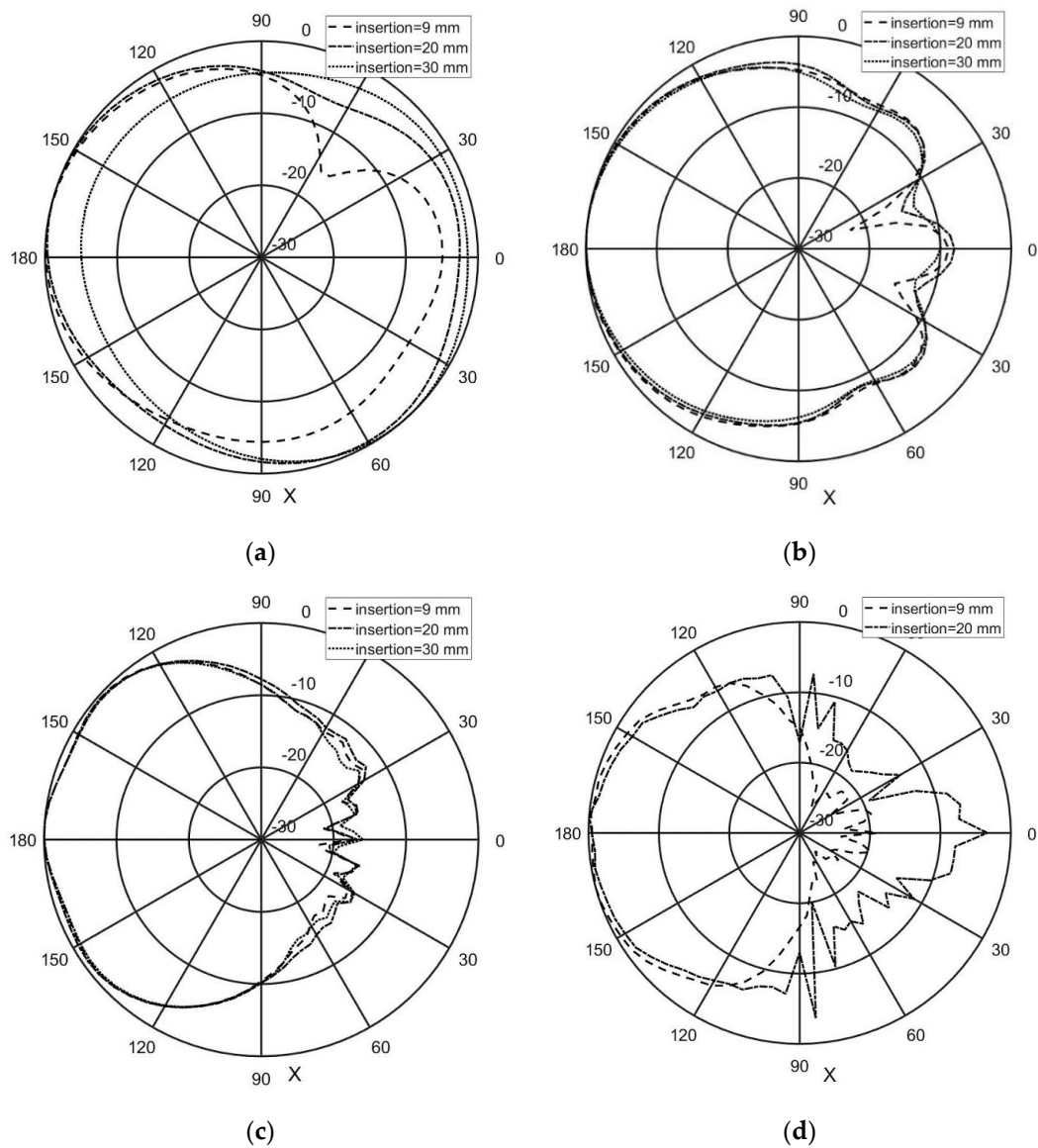


Figure A2. Normalized gain pattern with insertion distance of antenna 1 on the horizontal plane: (a) $f = 402$ MHz, (b) $f = 868$ MHz, (c) $f = 2.45$ GHz and (d) $f = 5.8$ GHz (only two insertion distances are shown as the noise is dominant above 20 mm at this frequency).

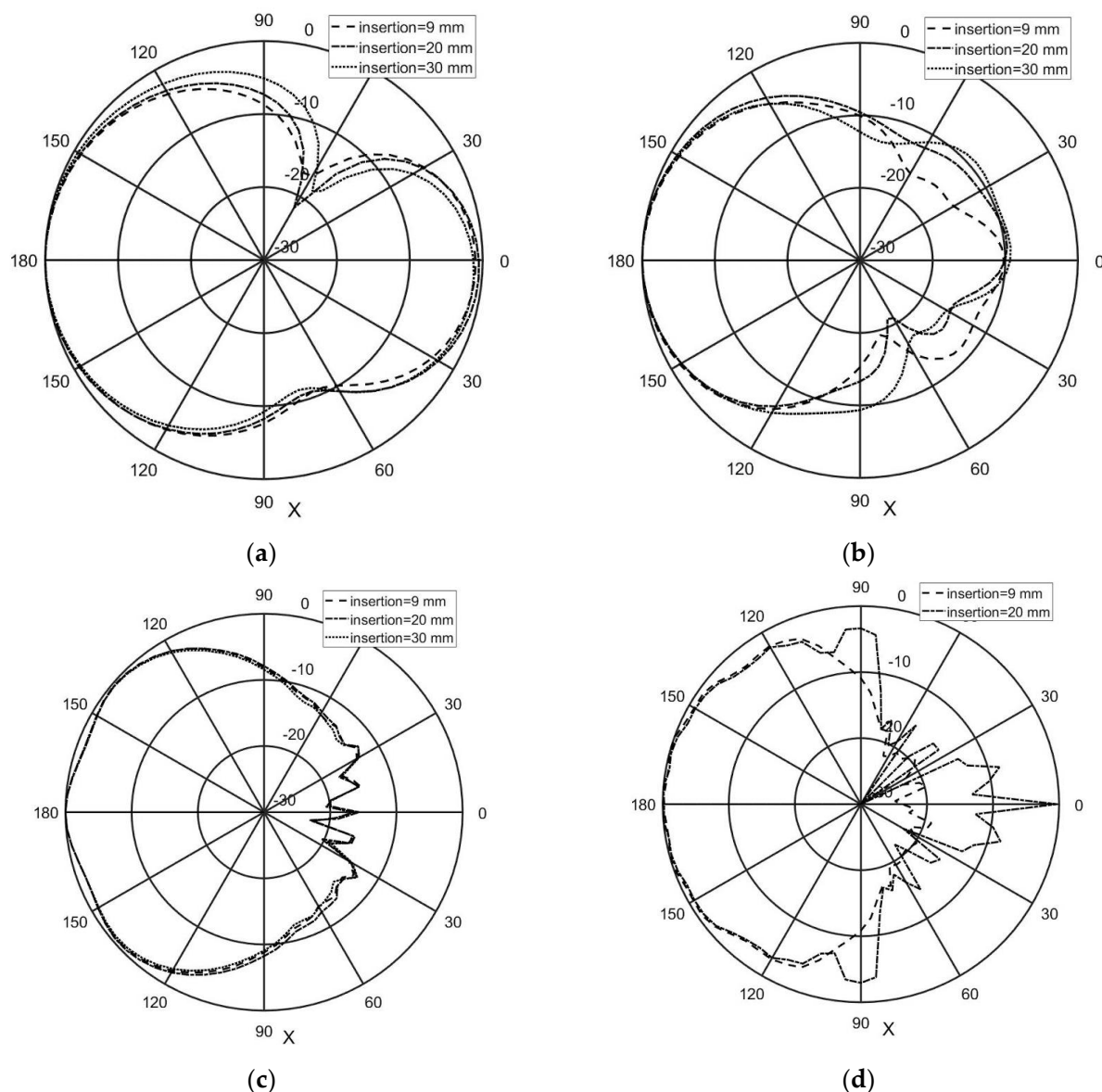


Figure A3. Normalized gain pattern with insertion distance of antenna 2 on the horizontal plane: (a) $f = 402$ MHz, (b) $f = 868$ MHz, (c) $f = 2.45$ GHz and (d) $f = 5.8$ GHz (only two insertion distances are shown as the noise is dominant above 20 mm at this frequency).

As can be observed, the normalized radiation patterns are kept mostly constant with penetration into the phantom along short distances.

Appendix B

In this appendix, the simulations that show the effect of using “open add space” boundary conditions vs. using an unlimited phantom in x and y axes (“periodic” boundary conditions) are shown (Figure A4). The 3D model takes $355 \text{ mm} \times 160 \text{ mm} \times 255 \text{ mm}$ as the theoretical properties [25–27].

For the high frequency cases (2.45 GHz and 5.8 GHz), the effect for the studied phantom ($355 \text{ mm} \times 160 \text{ mm} \times 255 \text{ mm}$) was negligible because the phantom was electrically large and, consequently, using open add space boundaries or using a phantom infinite in x and y directions gave the same results. However, the effect was very relevant for the low frequency cases (402 MHz and 868 MHz) where important changes were observed: when employing periodic boundary conditions (equivalent to infinite body), the field amplitude

decreases linearly and oscillations are leveled. This is clearly because the phantom is not electrically large enough to ignore the reflections due to the boundaries. The model was conceived to always employ “open add space” boundary conditions in a situation where it is electrically large. This is done because it was geared to suggest a physical phantom of the same dimensions.

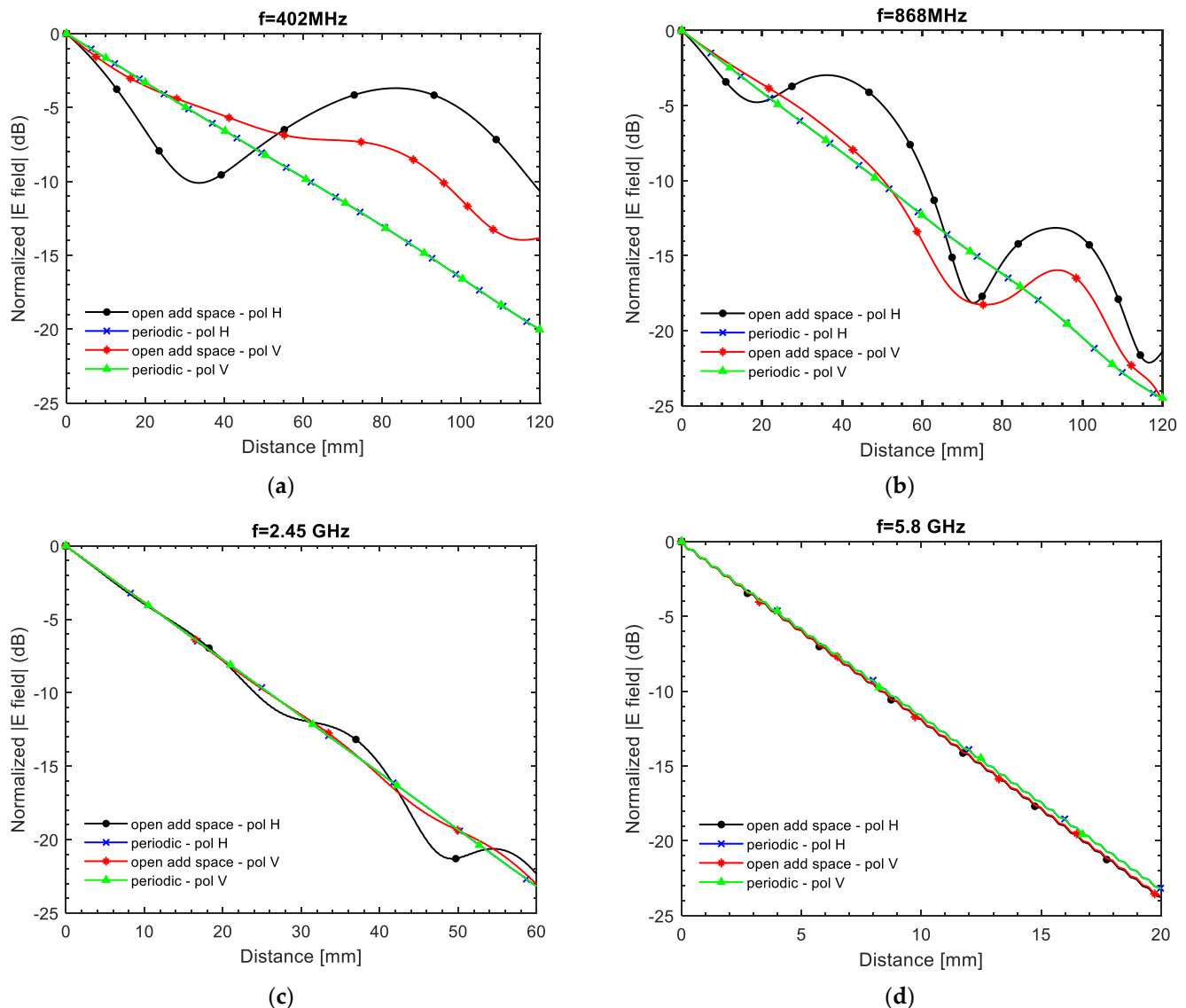


Figure A4. Normalized E-field with open add space vs. periodic boundaries for both polarizations and different frequencies: (a) $f = 402\text{ MHz}$, (b) $f = 868\text{ MHz}$, (c) $f = 2.45\text{ GHz}$ and (d) $f = 5.8\text{ GHz}$.

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