Evaluation of the Interaction between Electromagnetic Waves and the Human Body Using Biological Tissue-Equivalent Phantom in HF and VHF bands

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1 Introduction

1.1 Background

In recent years, radio frequency (RF) devices, which are usually placed in the vicinity of the human body, are widely used. In addition, these devices are used in various situations, for example, talk (cellular phones and portable radio terminals), data communication (wireless LAN, personal data assistant (PDA)), broadcast, home electrical appliance (e.g. microwave oven, induction heating (IH) cooker), medical applications (e.g. body-centric wireless communications, hyperthermia), etc.

Until now, the effect of the human body on the characteristics of antenna of devices which are placed nearby the human body has been widely investigated [1]-[4]. For the future, the quantitative evaluation of the interaction between electromagnetic waves (EM) and the human body, which is radiated by the devices, is indispensable. Here, the “interactions” means two ways: an influence of the human body on the performance of a mobile terminal and an influence of EM waves at the human body. The devices close to the human body cause the variation in the input and radiation characteristics of the antenna. On the other hand, the influence of the EM waves at the human body is dependent on the frequency [5]. The EM waves mainly contribute the heat effect, which is generated by the absorption of the energy in the microwave region. The specific absorption rate (SAR) has been usually used for the primary dosimetric parameter of the EM waves exposure in the standards [6]-[13]. In high frequency (HF) band and below, EM waves contribute the stimulating effect.

Until now, various investigations have been mainly made on the interaction between the mainly cellular phones and the human body at ultra high frequency (UHF)
band. For the future, it is necessary to evaluate the interaction between the EM waves and human body in other situations (e.g. frequency, devices, etc.). Especially, recently, the variety of antennas for HF-VHF band application is widely investigated [49]-[52]. For example, portable radio terminals, logistics or individual authentication using RFID system, medical applications (e.g. body-centric wireless communications, hyperthermia, magnetic resonance imaging (MRI), etc.). Generally, the interactions should be estimated from numerical simulations and experimental evaluations. In the experimental evaluations, biological tissue-equivalent phantoms, which are adjusted to have the same electrical properties (relative permittivity and conductivity) of biological tissues, are usually used. Until now, various biological tissue-equivalent phantoms have been proposed for evaluating the interaction of the human body and the electromagnetic waves [14]-[40]. However, most of these studies have mainly focused on phantom for ultra high frequency (UHF) band because various investigations have been mainly made on interaction at UHF band such a cellular phone. In other words, recipes of phantom for HF and VHF bands are very few. This is because of measurement method of electrical properties of the phantom has not been established. Therefore, biological tissue-equivalent phantoms for HF and VHF bands are desired.

This dissertation focuses on the biological tissue-equivalent phantoms for HF and VHF bands. Especially, the author focused on phantom for MRI systems in this dissertation.
1.2 Purpose of this study

Three major purposes to this dissertation are as follows.

1) Investigation of electrical properties of the biological tissue-equivalent phantom.
2) Development of the biological tissue-equivalent phantom in HF-VHF bands.
3) Evaluation of the interaction between the EM waves and human body using the biological tissue-equivalent phantom.

1.3 Contents

This dissertation deals with the evaluation of the interaction between EM waves and the human body with phantom in HF and VHF bands. Figure 1-1 shows the constructions of this dissertation.

In chapter 2, dielectric theory and mechanisms are described at the beginning because human body and phantom are one of the dielectric substances. In addition, classification of the biological tissue-equivalent phantom is described.

In chapter 3, measurement method of electrical properties of the phantom in HF and VHF bands are described. Firstly, comparison of measurement method of electrical properties is described. Measurement setup and measurement principle in this study are described. Measurement method in this dissertation can be used to obtain electrical properties of the phantom in HF and VHF bands.

In chapter 4, firstly, electrical properties of the base compound of the phantom as a fundamental study are described. In addition, electrical properties of phantoms in HF and VHF bands are described.

In chapter 5, how to evaluate the interaction between the EM waves and human body using the biological tissue-equivalent phantom. Firstly, measurement method of
SAR evaluation for applying thermographic method is described. In addition, SAR evaluation using numerical phantom is described. Moreover, measurement method of magnetic field distribution inside the phantom is described. Finally, antenna design considering effect of human body is described.

Finally, conclusions of this dissertation are presented in Chapter 6.
Chapter 1
Introduction

Chapter 2
Biological tissue-equivalent phantom

Chapter 3
Measurement method of electrical properties

Chapter 4
Development of the biological tissue-equivalent phantom

Chapter 5
Evaluation of the interaction between EM waves and the human body using the phantom

Chapter 6
Conclusions

Fig. 1-1 Constructions of this dissertation.
2 Biological Tissue-Equivalent Phantom

2.1 Introduction

Human body and phantom are one of the dielectric. Therefore, in this chapter, dielectric theory and mechanisms are described at the beginning. The electrical properties that will be discussed here are permittivity. It is important to note that electrical properties are not constant. They can change with frequency, temperature, orientation, mixture, pressure, and molecular structure of the material. In addition, classification of the biological tissue-equivalent phantom is introduced.

2.2 Dielectric theory and mechanisms

2.2.1 Electrical properties

A material is classified as “dielectric” if it has the ability to store energy when an external electric field is applied (For example, [45]). If a direct current (DC) voltage source is placed across a parallel plate capacitor, more charge is stored when a dielectric material is between the plates than if no material (a vacuum) is between the plates. The dielectric material increases the storage capacity of the capacitor by neutralizing charges at the electrodes, which ordinarily would contribute to the external field. The capacitance with the dielectric material is related to dielectric constant. Parallel plate capacitor in the case of DC is shown Fig. 2-1. If a DC voltage source $V$ is placed across a parallel plate capacitor, more charge is stored when a dielectric material is between the plates than if no material (a vacuum) is between the plates. Where $C$ and $C_0$ are capacitance with and without dielectric, $\kappa' = \varepsilon_r$ is the real dielectric constant or permittivity, and $S$ and $d$ are the area of the capacitor plates and the distance between
them. $C$ and $C_0$ are given as following equations.

$$C_0 = \frac{S}{d} \quad (2-1)$$

$$C = C_0 \kappa' \quad (2-2)$$

![Parallel plate capacitor, DC case](image)

The dielectric material increases the storage capacity of the capacitor by neutralizing charges at the electrodes, which ordinarily would contribute to the external field. The capacitance of the dielectric material is related to the dielectric constant as indicated in the following equations.

$$I = I_c + I_i = V(j\omega C_0 \kappa' + G) \quad (2-3)$$

if

$$G = \omega C_0 \kappa'' \quad (2-4)$$

then

$$I = V(j\omega C_0)(\kappa' - j\kappa'') = V(j\omega C_0)\kappa \quad (2-5)$$

Parallel plate capacitor in the case of alternate current (AC) is shown Fig. 2-2. If an AC
sinusoidal voltage source \( V \) is placed across the same capacitor, the resulting current will be made up of a charging current \( I_c \) and a loss current \( I_l \) that is related to the dielectric constant. The losses in the material can be represented as a conductance \( G \) in parallel with a capacitor \( C \).

The complex dielectric constant \( \kappa \) consists of a real part \( \kappa' \) which represents the storage and an imaginary part \( \kappa'' \) which represents the loss. The following notations are used for the complex dielectric constant interchangeably

\[
\kappa = \kappa^* = \varepsilon_r = \varepsilon_r^*
\]  

(2-6)

![Parallel plate capacitor, AC case](image)

From the point of view of electromagnetic theory, the definition of electric displacement (electric flux density) \( D_f \) is:

\[
D_f = \varepsilon E
\]  

(2-7)

where \( \varepsilon = \varepsilon^* = \varepsilon_0 \varepsilon_r \) is the absolute permittivity (or permittivity), \( \varepsilon_r \) is the relative
permittivity, $\varepsilon_0 = 8.854187 \times 10^{12} \ F/m$ is the free space permittivity and $E$ is the electric field.

Permittivity describes the interaction of a material with an electric field $E$ and is a complex quantity.

$$\kappa = \frac{\varepsilon}{\varepsilon_0} = \varepsilon_r = \varepsilon' - j\varepsilon''$$  \hspace{1cm} (2-8)

Dielectric constant ($\kappa$) is equivalent to relative permittivity $\varepsilon_r$ or the absolute permittivity $\varepsilon$ relative to the permittivity of free space $\varepsilon_0$. The real part of permittivity $\varepsilon_r'$ is a measure of how much energy from an external electric field is stored in a material. The imaginary part of permittivity ($\varepsilon_r''$) is called the loss factor and is a measure of how dissipative or lossy a material is to an external electric field. The imaginary part of permittivity ($\varepsilon_r''$) is always greater than zero and is usually much smaller than ($\varepsilon_r'$). The loss factor includes the effects of both dielectric loss and conductivity.

When complex permittivity is drawn as a simple vector diagram the real and imaginary components are 90° out of phase. The vector sum forms an angle $\delta$ with the real axis ($\varepsilon_r'$). The relative “lossiness” of a material is the ratio of the energy lost to the energy stored.

The loss tangent or $\tan \delta$ is defined as the ratio of the imaginary part of the dielectric constant to the real part. $D$ denotes dissipation factor and $Q$ is quality factor. The loss tangent $\tan \delta$ is called tan delta, tangent loss or dissipation factor. Sometimes the term “quality factor or $Q$-factor” is used with respect to an electronic microwave material, which is the reciprocal of the loss tangent. For very low loss materials, since $\tan \delta \approx \delta$, the loss tangent can be expressed in angle units, milliradians or microradians.
2.2.2 Electrical properties of the human body

In this section, electrical property of the human body is described. Electrical properties (relative permittivity and conductivity) of the human body depend strongly on the tissue type and frequency of interest. Temperature blood or fluid perfusion, and individual differences are second-order effects that are normally not considered [46].

Figure 2-3 shows the electrical properties of two very extreme tissues in the body (muscle and fat) as a function of frequency. Muscle has very high salinity and high water content, and fat is a poor conductor (good insulator, lower conductivity). At low frequencies, the conductivity of the tissue dominates the behavior of the field ($\varepsilon'' = \sigma / \varepsilon_0 \omega$, and $\omega$ is small), and high frequencies, the relative permittivity $\varepsilon_r'$ tends to dominate. Table 2-1 shows the electrical properties of muscle and fat in Industrial Scientific Medical (ISM) bands [47], [48]. ISM bands are radio bands (portions of the radio spectrum) reserved internationally for the use of RF energy for industrial, scientific and medical purposes other than telecommunications and broadcasting. Examples of applications in these bands include radio-frequency process heating, microwave ovens, and medical diathermy machines.
Fig. 2-3  Electrical properties of two very extreme tissues in the body (muscle and fat).

Table 2-1  Electrical properties of muscle and fat in ISM bands.

<table>
<thead>
<tr>
<th>Frequency</th>
<th>Muscle</th>
<th>Fat</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$\varepsilon_r$</td>
<td>$\sigma$ [S/m]</td>
</tr>
<tr>
<td>13.56 MHz</td>
<td>132.1</td>
<td>0.656</td>
</tr>
<tr>
<td>27.12 MHz</td>
<td>93.7</td>
<td>0.680</td>
</tr>
<tr>
<td>40.68 MHz</td>
<td>81.1</td>
<td>0.695</td>
</tr>
<tr>
<td>2.45 GHz</td>
<td>53.6</td>
<td>1.810</td>
</tr>
<tr>
<td>5.80 GHz</td>
<td>49.0</td>
<td>5.201</td>
</tr>
</tbody>
</table>


2.2.3 Effect of EM waves’ exposure on the human body

The effect of the EM waves, which is radiated from RF devices, on the human body is dependent on the frequency as well known [5], [49]. The effect is divided broadly into two categories, which are the stimulus or heat effect. Figure 2-4 describes the threshold of the effect of \( E \)-field on the human body. As shown in Fig. 2-4 the threshold of the stimulus effect is lower than that of the heat effect below 10 kHz because the increase in the capacitance of the cell membrane is proportional to the increase in the frequency above 1 kHz. Therefore, generation of the stimulus effect by the \( E \)-field is larger than that easier that of these heat one below 10 kHz. Here, the stimulus effect, which is caused by the induced current, mainly generates around the skin. On the other hand, the threshold of the heat effect is larger than that of the stimulus one above 100 kHz. Therefore, generation of the heat effect by the effect is at \( E \)-field is easier than that of the stimulus one above 100 kHz. In addition, the boundary of the effect is at 10 to 100 kHz. As for frequencies from 10 MHz to 300 GHz, heating is the major effect of EM energy absorption, yet the temperature elevation more than 1-2 °C that able to stimulate adverse health effects such as heat exhaustion and heat stroke [50].

According to the ICNRP guidelines [6], based on the energy absorption properties in the human body, EM frequency spectrum can be categorized as follows:

1. From 100 kHz up to 20 MHz, the absorption in the human trunk decreases rapidly as decreasing frequency and significant absorption may occur in the neck and legs;
2. From 20 MHz up to 300 MHz, relatively high absorption can occur in the whole body, and to even higher values if partial body resonances are considered;
3. From 300 MHz up to several GHz, significant local, non uniform absorption
occurs;

The primary biological effects associated with the RF radiation are related to the thermal effects of the EM field. The human body absorbs the energy of EM waves as magnetic loss and dielectric loss. The amount of that absorption energy \( P_L \) [W] can be expressed by the following formula.

\[
P_L = P_{L(\text{Dielectric})} + P_{L(\text{Magnetic})} \tag{2-9}
\]

where \( P_{L(\text{Dielectric})} \) is the dielectric loss [W], and is given as following equation.

\[
P_{L(\text{Dielectric})} = \frac{1}{2} \int \sigma |E|^2 \, dv \tag{2-10}
\]

\[
\sigma = 2\pi f \epsilon_{\text{r}} \epsilon''
\]

where \( \sigma \) represents the electric conductivity [S/m], \( v \) is the volume [m\(^3\)], \( E \) is the electric field strength in the human body [V/m]. \( \epsilon'' \) is imaginary part of permittivity, respectively. In contrast, \( P_{L(\text{Magnetic})} \) is the magnetic loss [W], and is give by following equation.

\[
P_{L(\text{Magnetic})} = \frac{1}{2} \int \sigma_m |H|^2 \, dv \tag{2-12}
\]

\[
\sigma_m = 2\pi f \mu_{0} \mu''
\]

where \( \sigma_m \) is the magnetic conductivity [1/S\(\cdot\)m]. \( \mu_{0} \) is magnetic permeability of the vacuum: \( 4\pi \times 10^{-7} \) H/m, \( \mu'' \) is imaginary part of permeability, respectively. In HF-VHF bands (30-300 MHz), imaginary part of permeability is very nearly zero. This is why the second term in right-hand side of eq. 2-9 is ignored. The primary dosimetric term
that used to describe the EM energy absorption is called SAR. The SAR is the mass
normalized rate at which RF power is coupled to biological tissue and is typically
expressed in watts per kilogram [W/kg].

Fig. 2-4  Threshold of the effect of the E-field in the human body [3].
2.3 Classification of the biological tissue-equivalent phantom

The interaction is generally estimated from numerical simulations and experimental evaluations. In these evaluations, biological tissue-equivalent phantoms are usually used [18]-[43]. In recent years, various types of phantom have been developed for evaluation of interaction. They are phantom for ultrasonic, phantom for X-ray, phantom for EM wave, and so on. In this dissertation, we focused on phantom for EM wave. The phantom for EM wave has the same electrical properties of human tissues. Literature [44] details for the types, characteristics, use of the phantoms. Here, the phantom can be classified into two major categories: the numerical phantom and experimental phantom. In this section, feature of each phantom were introduced.

2.3.1 Numerical phantom

With the rapid rise in computer performance, numerical calculation has become indispensable for radio frequency dosimetry. Voxel human models have therefore become one of the most important infrastructures for numerical calculation. Many numerical phantoms have been developed and used for numerical calculation. A numerical phantom is a mathematical representation of the external envelope of the human body shape together with the boundaries of the internal organs and tissues. The volumes that the organs define are filled with a medium that has the chemical elements that compose the tissues in the correct proportion and the tissue density.

The first numerical phantoms were the mathematical models. They were mathematical in the sense that equations for planes, spheres, cylinders, cones, ellipsoids and elliptical-cylinders (or parts and combinations thereof) were used to represent
internal organs. However, simple geometrical volumes described by equations do not accurately conform to the shape of real organs and do not fit alongside each other in the way human organs do. Furthermore, the constraints imposed by fitting defined shapes within an elliptical cylinder mean that their positioning is not anatomically realistic.

The rapid advancement and increased use of medical imaging technology, specifically computed tomography (CT) and magnetic resonance imaging (MRI), has provided high resolution cross-sectional digital images of internal anatomy. The pixel data from such images, when extended into the third dimension become cuboidal volume elements called voxels. The data may be used to create a three-dimensional digital representation of the shape, volume and composition of human organs. Each voxel is defined so that it contains a uniform medium and is assigned an index that identifies it as belonging to a particular organ or tissue. Various MRI or CT based head models have been reported.

The finite-difference time-domain (FDTD) method is currently the most widely accepted means for evaluate the interaction between EM waves and human body [51], [52]. This method adapts very well to the human models which are usually derived from MRI or computed tomography CT scans and offers great flexibility in modeling the heterogeneous structures of anatomical tissues and organs.
2.3.2 Experimental phantom

The biological tissue can be classified into two major categories: the high water content tissue such as muscle, brain and internal organs, and the low water content tissue such as fat and bone. The electrical properties of each tissue have been reported. Among them, the experimental phantom realizes almost the same relative permittivity and conductivity as those of tissue. In addition the experimental phantom can be classified into three major categories: liquid, solid, or semisolid. Feature of each phantom are as follows.

(1) Liquid phantom

Liquid phantoms are generally used for the SAR estimation using the $E$-field probe. Liquid phantoms are basically made of sugar, deionised water, salt, and cellulose for frequencies of less than 1GHz. For higher frequencies, mixture of deionised water and polyhydric alcohol, such as diethylene glycol monobuthyl ether and polyethylen glycol mono phenyl ether are recommended. While liquid phantom has been widely used so far, some intrinsic problems exist. First, a container (rigid shell) is necessary to contain and shape the material, and hence it is difficult to estimate directly the SAR distribution on the surface of the liquid phantom.

(2) Solid phantom

Solid phantoms are basically made of carbon materials, silicone rubber and ceramic raw materials. Solid phantom have been used to evaluate dosimetry as well as antenna characteristics. Solid phantoms can maintain their shapes by themselves. This means that a container such a shell is not required. While solid phantom has been widely used
so far, some intrinsic problems exist. First, it is difficult to arbitrary shape, and multi layers.

(3) Semisolid phantom

Two kind of semisolid phantoms based on agar have been used to evaluate dosimetry as well as antenna characteristics. One of these phantom is called the thickening agent phantom (TX-151 phantom) and the other is called the glycerin phantom. TX-151 is most often-used as a thickening agent. Each phantom has different features. Compared to glycerin phantom, the TX-151 phantom can simulate high-water content organs such as the muscle for a wider frequency range below 3 GHz. On the other hand, the glycerin phantom can reproduce both high- and low-water content organs and is superior to the TX-151 phantom in terms of preservation. Since the glycerin phantom provides a moisturizing action, there is less evaporation of water from the glycerin phantom than that from TX-151 phantom. Both phantoms can also maintain their shapes by themselves. This means that a container is not required. Another feature of these phantoms is that arbitrary shapes can be fabricated using a mold. The summarized features of these phantoms are:

- Easy adjustment of electrical properties
- Easy manufacturing arbitrary shape and multi layers
- Not necessary using a container (rigid shell)
- To maintain the shapes by themselves
- Cheap and popular ingredient
- Preservation period (TX-151: ~ 1 month, glycerin: ~ 6 month)
As mentioned above, semisolid phantom has many merits. Therefore, in this paper, the author focused on the semisolid phantom.

### 2.4 Summary in Chapter 2

Human body and phantom are one of the dielectric materials. Therefore, in chapter 2, dielectric theory and mechanisms are described at the beginning. In addition, effect of EM waves’ exposure on the human body was introduced. Finally, classification of the biological tissue-equivalent phantom was described.
3 Measurement method of electrical properties in HF-VHF bands [J-1], [J-2]

3.1 Introduction

Until now, various investigations have been undertaken mainly on the interaction between the cellular phones and the human body at UHF band. For the future, it is necessary to evaluate the interaction between the EM waves and human body in other situations (e.g. frequency, devices, etc.). Especially, in recent years, the variety of antennas for HF-VHF band application is widely investigated [14]-[17]. In addition, these antennas are used in various situations, for example, talk (e.g. portable radio terminals), logistics or individual authentication using RFID (Radio Frequency IDentification) system, medical applications (e.g. body-centric wireless communications, hyperthermia, etc.), wireless power transmission, and so on.

Generally, the interactions should be estimated from numerical simulations and experimental evaluations. In these evaluations, biological tissue-equivalent phantoms are usually used. Until now, various phantoms have been proposed for evaluating the interaction of the human body and the electromagnetic waves. However, most of these studies have mainly focused on phantom for UHF band. In other words, recipes of phantom for HF-VHF bands are very few. This is because of measurement method of electrical properties of the phantom has not been established. Therefore, in this chapter, measurement methods of electrical properties in HF-VHF bands are investigated before developing phantoms.
3.2 Comparison of electrical property measurement method

Many factors such as accuracy, convenience, and the material shape and form are important in selecting the most appropriate measurement technique [53]. Some of the significant factors to consider are summarized here:

- Frequency range
- Expected values of electrical properties
- Required measurement accuracy
- Material properties (i.e., homogeneous, isotropic)
- Form of material (i.e., liquid, powder, solid, sheet)
- Sample size restrictions
- Destructive or nondestructive
- Contacting or non-contacting
- Temperature
- Cost
3.2.1 Coaxial probe method

In relatively-high frequency region such as VHF band and above, the electrical properties of materials are most commonly measured with an open-ended coaxial probe. The typical measurement system using a coaxial probe method consists of a network analyzer, a coaxial probe and software. An external computer is needed in many cases to control the network analyzer through General Purpose Interface Bus (GP-IB).

The open-ended coaxial probe is a cut off section of transmission line. The material is measured by immersing the open-ended coaxial probe into a liquid or touching it to the flat face of a solid or powder material. The outer conductor of the coaxial probe is typically connected to a larger guard electrode (fringe). The large flange allows measurements of flat surfaced solid materials, in addition to liquids and semi-solids. The electric field lines of an applied probing signal (which are always perpendicular metal) are perpendicular to both the end of the center conductor and the fringe. Thus, they bend around the open end of the coaxial line as shown in Fig. 3-1. These electric field lines interact with the material being measured and produce a reflected field that returns down the coaxial transmission line. The characteristics of this reflected field (phase and magnitude) are a function of the material’s electrical properties. The probing signal is typically swept in frequency to determine the electrical properties as a function of frequency. The high frequency cap is due to the dimensions of the coaxial line and the requirement to not propagate higher order modes. Materials measurements at low frequencies can become less accurate due to the fact that network analyzers measure reflection coefficient $\Gamma$ directly. The accuracy is expressed as a certain percentage of $\Gamma$, both magnitude and phase. As shown eq. 3-1, the measured reflection coefficient $\Gamma$ corresponds to the complex permittivity of the measurement.
material:

\[ \varepsilon^* = \varepsilon' - j\varepsilon'' = \varepsilon(1 - j\tan\delta) \]  

(3-1)

Here, \( \varepsilon^* \) is the complex permittivity, \( \varepsilon' \) is the real part of complex permittivity (relative permittivity), \( \varepsilon'' \) is the imaginary part of the complex permittivity, and \( \tan\delta \) is the loss tangent.

In other words, there is bidirectional mapping of the complex permittivity and reflection coefficient. This illuminates how a polar display of reflection coefficient is transformed into values of relative permittivity and loss tangent. Figure 3-2 shows the mapping of the relative permittivity and loss tangent of the material into a polar reflection coefficient plot for two frequencies, tens of megahertz and hundreds of megahertz. The tens of megahertz plot depicts a random \( \Gamma \) vector and on the tip of the vector, there is a circle, representing the uncertainty (not to scale). At tens of megahertz, a wide range of dielectric constant is compressed into a small area of the polar plot. This implies that a small percentage uncertainty for measuring reflection coefficient now becomes a large percentage error for dielectric constant or loss tangent. The electrical properties error at tens of megahertz is much larger than the error at higher frequencies [54]. In other words, minimum frequency is limited at tens of megahertz to hundreds of megahertz.
Fig. 3-1 Open-ended coaxial cable probe used for measuring the electrical properties.

(a) Tens of megahertz

(b) Hundreds of megahertz

Fig. 3-2 The mapping of the relative permittivity and loss tangent into a polar reflection coefficient plot [54].
3.2.2 Two-terminal method

When using an impedance-measuring instrument to measure electrical properties, the two-terminal method is usually employed [53]. The two-terminal method involves sandwiching a material or liquid between two electrodes to form a capacitor. The measured capacitance is then used to calculate permittivity. In an actual test setup, two electrodes are configured with a test fixture sandwiching dielectric material. The impedance measuring instrument would measure vector components of capacitance and dissipation and a software program would calculate permittivity and loss tangent. In other words, two-terminal method does not measure reflection coefficient $\Gamma$. Thus, this measurement method is able to produce better results at lower frequencies than coaxial probe method. Usually, maximum frequency is limited at tens of megahertz.
3.3 Measurement setup and measurement principle in HF band

3.3.1 Measurement setup
The coaxial probe method using the network analyzer is commonly-used in measurement of electrical properties at higher frequencies such as VHF band and above [45]. However, electrical properties measurement at lower frequencies can become less accurate due to the fact that network analyzers measure reflection coefficient $\Gamma$ directly. Thus, in this dissertation, we explored other approaches in order to produce better results at lower frequencies. The electrical properties measurement setup is shown in Fig. 3-2. This system employs two-terminal method using the impedance analyzer (6530B, Wayne kerr Electronics, upper limit frequency: 30 MHz). Impedance analyzer is commonly-used to measure the material properties at lower frequencies. Impedance analyzer is connected to the personal computer (PC) by GP-IB. In this study, measured frequency range is 300 kHz to 30 MHz. Two-terminal method using the impedance analyzer does not measure reflection coefficient $\Gamma$. Thus, this system is able to produce better results at lower frequencies.
(a) Block diagram of setup for measurement of electrical properties.

(b) Measurementsystem.

Fig. 3-2 Measurement setup.
3.3.2 Measurement principle of the two-terminal method by impedance analyzer

Figure 3-3 shows the equivalent circuit of the two-terminal method [53]. The two-terminal method using the impedance analyzer involves sandwiching a phantom between two electrodes to form a capacitor. In order to obtain the electrical properties of the phantom, it is necessary to measure the impedance of the phantom. The impedance of the phantom is derived from the measured voltage and current values. The measured impedance of the phantom is converted to electrical properties by the following equations:

\[
\begin{align*}
\varepsilon' &= \frac{dC}{\varepsilon_0 S} \quad (3.1) \\
\varepsilon'' &= -\varepsilon' \left| \frac{Z\cos \theta}{Z\sin \theta} \right. \quad (3.2) \\
\sigma &= 2\pi f\varepsilon' \varepsilon_0 \quad (3.3)
\end{align*}
\]

Here, \(\varepsilon'\) is the real part of complex permittivity, \(d\) is the length of the sample [mm], \(C\) is the capacitance of the sample [F], \(\varepsilon_0\) is the permittivity in vacuum [F/m], \(S\) is the area of the electrode \([m^2]\), \(\varepsilon''\) is the imaginary part of the complex permittivity, \(|Z|\) is the absolute value of the impedance [\(\Omega\)], \(\theta\) is the phase difference between the measured voltage and current, \(\sigma\) is the conductivity [S/m], and \(f\) is the frequency [Hz].
Fig. 3-3   Equivalent circuit of two-terminal method.
3.3.3 Source of measurement error

The two-terminal method is the simplest method of connecting a sample but contains some error sources [53]. For example, lead inductances, lead resistances, and stray capacitance between two leads are added to the measurement result. Contact resistances between the test electrodes and the samples are also added to measured impedance.

a. Electrical properties by changing electrodes dimension

Here, electrical properties by changing electrodes dimension are described. Figure 3-4 shows dimension of electrode used for electrical properties measurement. In this dissertation, circular electrodes are adopted. In addition, deionized water (target value: $\varepsilon' = 80$, $\sigma$ is nearly zero) was used as a test sample. The electrode diameter is changed 9 mm, 16 mm and 28 mm. Here, deionized water is contained in acrylic pipe which has internal diameter equal to electrode diameter. Figure 3-5 shows acrylic pipe used for electrical properties measurement. In this dissertation, anisotropic electrical properties were not considered. In addition, electrode is 1mm in thickness. Figure 3-6 shows measured electrical properties by changing electrode diameter. Here, measured frequency range is 300 kHz to 30 MHz, and average measured value is shown. From these results, we confirmed that electrode of diameter in 26 mm is closest target value.

Figure 3-7 show variation in measured value of electrical properties by changing electrodes diameter at 13.56 MHz in one instance. 13.56 MHz is one of the ISM band in HF band. From Fig. 3-7, variability in measured value becomes smaller with increasing diameter of the electrode was confirmed. It is considered that human error (e.g. parallelism of electrodes) becomes fewer with increasing diameter of the electrode. From these results, in this dissertation, electrode which has 26 mm in diameter is adopted.
Fig. 3-4 Dimension of electrodes use for electrical properties measurement.

Fig. 3-5 Acrylic pipes use for electrical properties measurement.
Fig. 3-6 Electrical properties of deionized water by changing electrode diameter.
Fig. 3-7 variation in measured value of electrical properties by changing electrodes
diameter at 13.56 MHz
b. Electrical properties by changing length of sample

Here, electrical properties by changing length of test sample are described. In this dissertation, electrode which has 26 mm in diameter was adopted. In addition, deionized water (target value: $\varepsilon' = 80$, $\sigma$ is nearly zelo) was used as a test sample. Here, deionized water is contained in acrylic pipe. Figure 3-8 shows acrylic pipe used for electrical properties measurement. The length of the test sample was changed 30 mm, 50 mm and 100 mm. First, electric fields between electrodes were investigated by FDTD calculation because electric fields between electrodes are defined as uniform in eq. (3-1). Fig. 3-9 shows FDTD calculation model. Fig. 3-10 shows component of electric field between electrodes in the $y$-$z$ plane at center of the electrode ($x=0$mm). From results, it was confirmed that vertical component to electrodes is dominant component. Figure 3-11 shows electric field distribution around electrodes in the $y$-$z$ plane at center of the electrode ($x=0$mm). Magnitude of electric field was normalized with the maximum value in the observation plane. From Fig. 3-11, the author confirmed that decrease electric field homogeneity between electrodes when increasing length of the test sample. In addition, standard deviation of electric field inside test sample in observation plane was shown in Table 3-1. Here, the levels of electric field were normalized with the maximum value in the test sample in each case. As a result, in the case of Fig. 3-11 (a) and (b) (length of the test sample are 30 mm and 50 mm, respectively), standard deviation of electric field in observation plane were 5.576E-3 and 1.778E-2. On the other hand, in the case of Fig. 3-11 (c) (length of the test sample was 100 mm), standard deviation of electric field in observation plane was 9.735E-2. Figure 3-12 shows measured electrical properties by changing length of the sample. From Fig. 3-12, the difference of electrical properties at 13.56 MHz between measured and target value was
slight in each case. In addition, difference of relative permittivity between 30mm and 100mm was about 5% despite relatively-inhomogeneous electric field between electrodes in the case of the longer sample. Therefore, it is considered that dominant measurement error sources is not inhomogeneous electric field between electrodes. For example, human error is thought as one of the dominant measurement error because it is difficult that an electrode to set accurately parallel to another electrode. From results of preliminary experiments, it was confirmed that electrical properties of the test sample was clearly effected by parallelism of electrodes. Especially, shorter sample was more likely to be affected than longer sample because rate of change in distance between electrodes was raised. Therefore, in order to remove above influence, the author adapts relatively-long sample (length of the sample more than 50 mm but less than 100 mm) in this dissertation.
Fig. 3-8 Acrylic pipes use for electrical properties measurement.

Length
100 mm 50 mm 30 mm

Water($\varepsilon_r=80, \sigma=0$)

Electrodes (Diameter: 26mm)

Length of sample: 30, 50 and 100mm

Fig. 3-9 Calculation model.
Fig. 3-10 Component of electric field.
Fig. 3-11 Electric field distribution.
Table 3-1 Standard deviation of electric field inside test sample.

<table>
<thead>
<tr>
<th>Length of sample [mm]</th>
<th>Average value of electric field inside the sample</th>
<th>Standard deviation of electric field inside test sample</th>
</tr>
</thead>
<tbody>
<tr>
<td>30</td>
<td>0.983</td>
<td>5.576E-03</td>
</tr>
<tr>
<td>50</td>
<td>0.946</td>
<td>1.778E-02</td>
</tr>
<tr>
<td>100</td>
<td>0.626</td>
<td>9.735E-02</td>
</tr>
</tbody>
</table>
Fig. 3-12 Electrical properties of deionized water by changing length of test sample.
3.4 Measurement method of electrical properties in VHF band

In recent years, the variety of antennas for VHF band application is also widely investigated [14]-[17]. For example, talk (e.g. portable radio terminals), logistics or individual authentication using RFID (Radio Frequency IDentification) system, medical applications (e.g. Magnetic Resonance Imaging (MRI), wireless power transmission to capsular endoscope), and so on. Therefore, phantom for VHF band is also required to evaluate the interaction between EM waves and human body.

In this section, measurement method of electrical properties in VHF band is introduced before adjustments to the electrical properties of the phantom.

Figure 3-13 shows the measurement setup of electrical properties. Electrical properties are measured by the coaxial probe which is connected to the network analyzer, as shown in Fig. 3-13. Figure 3-14 shows electrical properties of deionized water in VHF band. Here, measured frequency range is 30 MHz to 300 MHz. In order to confirm the validity of the measured result, measured electrical properties of deionized water was compared with target value of water in ref. [55]. Relative permittivity and conductivity of water in ref. [55] are 78 and almost exactly zero in VHF band, respectively. From this result, it is confirmed that measured result is agreement with the target value of water. Therefore, it is considered that coaxial probe method is one of the effective methods for electrical properties measurement in VHF band.
Fig. 3-13  Measurement setup of coaxial probe method.

Fig. 3-14  Measured electrical properties of deionized water in VHF band.
3.5 Summary in Chapter 3

In this chapter, comparison of electrical properties measurement method was introduced at the beginning. Next, measurement setup in HF band was described. In this dissertation, two-terminal method using the impedance analyzer was adapted. This is because of two-terminal method using impedance analyzer is able to produce better results at low frequencies such as HF band.

In addition, measurement principle of two-terminal method was introduced. Additionally, source of measurement error was investigated. In this dissertation, electrical properties by changing electrodes dimension and length of test sample were investigated. As a result, test sample which has longest diameter and electrode which has largest diameter were adopted because these were closest target value. In addition, the difference of electric properties between measured and target value was slight. From these results, validity of measured results was confirmed. At last, measurement method of electrical properties in VHF band is introduced. Here, coaxial probe method was chosen. In order to confirm validity of measured results in VHF band, measured result was compared with target value. As a result, the difference of electric properties between measured and target value was slight. Thus, validity of measured results in VHF band by coaxial probe method was confirmed.
4 Development of the biological tissue-equivalent phantom in HF-VHF bands [J-1], [J-2]

4.1 Introduction

In Chapter 3, measurement method of electrical properties was investigated. In addition, validity of measured results was confirmed. In this chapter the biological tissue-equivalent phantom for HF and VHF bands are described. Firstly, electrical properties of base compound of the phantom were explained. In addition, adjustments to the electrical properties of the phantoms are shown. Then, the composition of the phantom are shown.

4.2 Electrical properties of base compound of the phantom in HF band

Deionized water, glycerin and silicone emulsion were used as a base compound of the phantom. These materials have different electrical properties. In addition, electrical properties of these base compounds were not identified in HF band. Therefore, electrical properties of base compound of the phantom were measured, before adjustments to the electrical properties of the phantom. Figure 4-1 shows electrical properties of deionized water. Here, two-terminal method was adapted as a measurement method. In addition, temperatures of room and deionized water are 21.8 °C. As a result, the author confirmed that relative permittivity and conductivity of deionized water in the range of 300 kHz ~ 30 MHz were 80 and very nearly zero, respectively. Figure 4-2 shows electrical properties of glycerin. From Fig. 4-2, the author confirmed that relative permittivity of
glycerin was almost 43 in HF band. Moreover, conductivity of glycerin was increased with frequency. Figure 4-3 shows electrical properties of silicone emulsion. From results, the author confirmed that relative permittivity of silicone emulsion in HF band was almost 35. On the other hand, conductivity of silicone emulsion was 0.015 ~ 0.020 S/m in HF band.

Figure 4-4 shows comparison of electrical properties of base compounds. It is plotted the electrical properties of muscle and fat in HF band (8, 10, 13.56 and 21 MHz). Here, 8 MHz is one of the frequencies for hyperthermia. 10 and 21 MHz are one of the frequencies for body-centric wireless communications. 13.56 MHz is one of the frequencies in ISM band. From these results, it is confirmed that conductivity of base compounds were lower than biological tissues. Relative permittivity of deionized water was the highest of three. These results mean that it is necessary added substances to adjust electrical properties of phantom. On the one hand, relative permittivity of silicone emulsion was the lowest of the three. Therefore, silicone emulsion is adapted as a base compound of low water content tissue phantom.
Fig. 4-1 Electrical properties of the deionized water.

Fig. 4-2 Electrical properties of the glycerin.
Fig. 4-3 Electrical properties of silicone emulsion.
Fig. 4-4 Comparison of electrical properties (deionized water).
4.3 Adjustments of the electrical properties of the Biological tissue-equivalent phantom in HF band

In order to adjust electrical properties of the phantom, effect of added substances were investigated. In this dissertation, polyethylene powder, sodium chloride (NaCl), aluminum powder and cocking oil were selected as added substances. From results of the preliminary experiment, relative permittivity of the phantom is adjustable by changing amounts of polyethylene powder, aluminum powder and cocking oil. In addition, conductivity of the phantom is adjustable by changing amounts of polyethylene powder and NaCl.

4.3.1 High water content tissue

Usually, relative permittivity of high water content tissues (e.g. muscle, brain) are relatively high. Thus, in this study, deionized water, which has high relative permittivity, was selected as a base compound of the high water content tissue phantom.

Table 4-1 shows composition of the base phantom for high water content tissue. This phantom consists of deionized water and agar. Agar is used to maintain the shape of the phantom. Figure 4-5 shows electrical properties of the base phantom. As a result, electrical property of the base phantom was almost equal to deionized water.

Relative permittivity of high water content tissue such as a muscle is higher than deionized water in HF band. Thus, in order to adjust relative permittivity, aluminum powder is added to base phantom because it is assumed that the capacitance of the phantom is increased by the addition of aluminum powder. In addition, in order to adjust conductivity, NaCl is added. Table 4-2 shows amounts of added aluminum powder and NaCl. Figure 4-6 shows relationship between electrical properties of
phantom and amount of aluminum powder. From Fig. 4-5, it was confirmed that relative permittivity was increased with increase amount of aluminum powder. In addition, conductivities were nearly-unchanged by changing amount of aluminum powder. Figure 4-7 shows relationship between electrical properties of phantom and amount of NaCl. These result means, relative permittivity of phantoms were not changed by changing amount of NaCl. Additionally, conductivity of phantoms were increased with increase amount of NaCl.

Figures 4-8, 4-9, 4-10 and 4-11 show adjustment of electrical properties of the high content water tissue phantom at 8, 10, 13.56, 21 MHz, respectively. Here, the vertical and horizontal axes represent conductivity and relative permittivity of phantoms, respectively. The target values of high content water tissue were plotted as a black closed circle. Each symbol except black closed circle in the graph represents the electrical properties for each high content water tissue phantom, which are changed based on the amount of aluminum powder and NaCl. The vertical and horizontal dotted lines represent the same amount of aluminum powder and NaCl, respectively. The amount of NaCl and aluminum powder are changed to confirm the range of the achievable electrical properties of the phantoms. The amount of the other added substances are same as those in Table 4-1. In Fig. 4-6 and Fig. 4-7, electrical properties depend on the amounts of both NaCl and aluminum powder. The relative permittivity depends on the amount of both NaCl and aluminum powder while the conductivity only relies on NaCl. Difference of electrical properties between measured value and target value [47], [56] were shown in Table 4-3. From this result, difference of relative permittivity and conductivity are within 5 %, respectively. Tables 4-4, 4-5, 4-6 and 4-7 show range of achievable electrical properties of high content water tissues. In this
dissertation, time dependence of electrical properties of phantom is not considered because semi solid phantom can keep electrical properties for more than a month with wrapping.

Table 4-1  Composition of the base phantom.

<table>
<thead>
<tr>
<th>Material</th>
<th>Amount [g]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Deionized water</td>
<td>1036</td>
</tr>
<tr>
<td>Agar</td>
<td>22.5</td>
</tr>
</tbody>
</table>

Table 4-2  Amounts of added substances
(a) Aluminum powder

<table>
<thead>
<tr>
<th>Ratio of aluminum powder to base phantom [wt%]</th>
<th>Amount of aluminum powder [g]</th>
</tr>
</thead>
<tbody>
<tr>
<td>10 %</td>
<td>62.3</td>
</tr>
<tr>
<td>20 %</td>
<td>140.1</td>
</tr>
<tr>
<td>30 %</td>
<td>240.2</td>
</tr>
<tr>
<td>40 %</td>
<td>358.7</td>
</tr>
</tbody>
</table>
(b) NaCl

<table>
<thead>
<tr>
<th>Ratio of aluminum powder to base phantom [wt%]</th>
<th>Amount of aluminum powder [g]</th>
</tr>
</thead>
<tbody>
<tr>
<td>10 %</td>
<td>62.3</td>
</tr>
<tr>
<td>20 %</td>
<td>140.1</td>
</tr>
<tr>
<td>30 %</td>
<td>240.2</td>
</tr>
<tr>
<td>40 %</td>
<td>358.7</td>
</tr>
</tbody>
</table>

Table 4-3  Difference of electrical properties between measured value and target value.

<table>
<thead>
<tr>
<th></th>
<th>Relative permittivity</th>
<th>Conductivity [S/m]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Target value</td>
<td>128.3</td>
<td>0.63</td>
</tr>
<tr>
<td>Measured value</td>
<td>132.1</td>
<td>0.66</td>
</tr>
<tr>
<td>Error [%]</td>
<td>2.8</td>
<td>4.5</td>
</tr>
</tbody>
</table>
Fig. 4-5 Electrical properties of the base phantom.
Fig. 4-6 Relationship between electrical properties of phantom and amount of aluminum powder.
Fig. 4-7 Relationship between electrical properties of phantom and amount of NaCl.

(a) Relative permittivity

(b) Conductivity
Fig. 4-8  Adjustment of electrical properties of the high content water tissue phantom at 8 MHz.
Table 4-4  Achievable electrical properties of high content water tissues at 8 MHz.

<table>
<thead>
<tr>
<th>Tissue name</th>
<th>Relative permittivity</th>
<th>Conductivity [S/m]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 Gall ladder</td>
<td>99.0</td>
<td>0.902</td>
</tr>
<tr>
<td>2 Tendon</td>
<td>107.8</td>
<td>0.404</td>
</tr>
<tr>
<td>3 Aorta</td>
<td>114.3</td>
<td>0.341</td>
</tr>
<tr>
<td>4 Blood Vessel</td>
<td>114.3</td>
<td>0.341</td>
</tr>
<tr>
<td>5 Lung Inflated</td>
<td>151.1</td>
<td>0.217</td>
</tr>
<tr>
<td>6 Trachea</td>
<td>172.7</td>
<td>0.451</td>
</tr>
<tr>
<td>7 Spinal Chord</td>
<td>183.1</td>
<td>0.213</td>
</tr>
<tr>
<td>8 Brain White Matter</td>
<td>192.9</td>
<td>0.148</td>
</tr>
<tr>
<td>9 Gland</td>
<td>193.7</td>
<td>0.712</td>
</tr>
<tr>
<td>10 Lymph</td>
<td>193.7</td>
<td>0.712</td>
</tr>
<tr>
<td>11 Pancreas</td>
<td>193.7</td>
<td>0.712</td>
</tr>
<tr>
<td>12 Thymus</td>
<td>193.7</td>
<td>0.712</td>
</tr>
<tr>
<td>13 Thyroid</td>
<td>193.7</td>
<td>0.712</td>
</tr>
<tr>
<td>14 Muscle</td>
<td>203.0</td>
<td>0.608</td>
</tr>
<tr>
<td>15 Cervix</td>
<td>203.1</td>
<td>0.616</td>
</tr>
<tr>
<td>16 Dura</td>
<td>207.1</td>
<td>0.533</td>
</tr>
<tr>
<td>17 Lung Deflated</td>
<td>210.3</td>
<td>0.428</td>
</tr>
<tr>
<td>18 Cartilage</td>
<td>217.5</td>
<td>0.358</td>
</tr>
<tr>
<td>19 Lens</td>
<td>218.6</td>
<td>0.511</td>
</tr>
</tbody>
</table>
Fig. 4-9  Adjustment of electrical properties of the high content water tissue phantom at 10 MHz.
Table 4-5  Achievable electrical properties of high content water tissues at 10 MHz.

<table>
<thead>
<tr>
<th>Tissue name</th>
<th>Relative permittivity</th>
<th>Conductivity [S/m]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1       Lens Nucleus</td>
<td>98.4</td>
<td>0.222</td>
</tr>
<tr>
<td>2       Gall Bladder</td>
<td>98.8</td>
<td>0.903</td>
</tr>
<tr>
<td>3       Tendon</td>
<td>103.2</td>
<td>0.408</td>
</tr>
<tr>
<td>4       Lung (Inflated)</td>
<td>123.6</td>
<td>0.225</td>
</tr>
<tr>
<td>5       Trachea</td>
<td>146.1</td>
<td>0.459</td>
</tr>
<tr>
<td>6       Muscle (Parallel Fiber)</td>
<td>149.2</td>
<td>0.672</td>
</tr>
<tr>
<td>7       Nerve (Spinal chord)</td>
<td>155.1</td>
<td>0.223</td>
</tr>
<tr>
<td>8       Avg. Muscle</td>
<td>160.0</td>
<td>0.645</td>
</tr>
<tr>
<td>9       Thyroid Thymus</td>
<td>162.7</td>
<td>0.719</td>
</tr>
<tr>
<td>10      Muscle (Transverse Fiber)</td>
<td>170.7</td>
<td>0.617</td>
</tr>
<tr>
<td>11      White Matter</td>
<td>175.7</td>
<td>0.158</td>
</tr>
<tr>
<td>12      Lens Cortex</td>
<td>176.0</td>
<td>0.521</td>
</tr>
<tr>
<td>13      Cartilage</td>
<td>179.3</td>
<td>0.369</td>
</tr>
<tr>
<td>14      Lung (Deflated)</td>
<td>180.3</td>
<td>0.438</td>
</tr>
<tr>
<td>15      Dura</td>
<td>194.9</td>
<td>0.544</td>
</tr>
<tr>
<td>16      Eye Tissue (Sclera)</td>
<td>208.2</td>
<td>0.798</td>
</tr>
<tr>
<td>17      Tongue</td>
<td>208.2</td>
<td>0.568</td>
</tr>
</tbody>
</table>
Fig. 4-10  Adjustment of electrical properties of the high content water tissue phantom at 13.56 MHz.
Table 4-6  Achievable electrical properties of high content water tissues at 13.56 MHz.

<table>
<thead>
<tr>
<th>Tissue name</th>
<th>Relative permittivity</th>
<th>Conductivity [S/m]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1  Tendon</td>
<td>96.3</td>
<td>0.415</td>
</tr>
<tr>
<td>2  Gall Bladder</td>
<td>98.4</td>
<td>0.904</td>
</tr>
<tr>
<td>3  Trachea</td>
<td>118.4</td>
<td>0.471</td>
</tr>
<tr>
<td>4  Nerve (Spinal chord)</td>
<td>125.2</td>
<td>0.236</td>
</tr>
<tr>
<td>5  Muscle (Parallel Fiber)</td>
<td>125.7</td>
<td>0.683</td>
</tr>
<tr>
<td>6  Avg. Muscle</td>
<td>132.1</td>
<td>0.656</td>
</tr>
<tr>
<td>7  Thyroid Thymus</td>
<td>132.3</td>
<td>0.729</td>
</tr>
<tr>
<td>8  Lens Cortex</td>
<td>134.5</td>
<td>0.534</td>
</tr>
<tr>
<td>9  Muscle (Transverse Fiber)</td>
<td>138.4</td>
<td>0.628</td>
</tr>
<tr>
<td>10 Cartilage</td>
<td>140.8</td>
<td>0.383</td>
</tr>
<tr>
<td>11 Lung (Deflated)</td>
<td>148.7</td>
<td>0.452</td>
</tr>
<tr>
<td>12 White Matter</td>
<td>153.1</td>
<td>0.176</td>
</tr>
<tr>
<td>13 Eye Tissue (Sclera)</td>
<td>162.2</td>
<td>0.813</td>
</tr>
<tr>
<td>14 Tongue</td>
<td>162.2</td>
<td>0.582</td>
</tr>
<tr>
<td>15 Dura</td>
<td>174.8</td>
<td>0.565</td>
</tr>
<tr>
<td>16 Skin (Wet)</td>
<td>177.1</td>
<td>0.384</td>
</tr>
<tr>
<td>17 Liver</td>
<td>181.3</td>
<td>0.336</td>
</tr>
<tr>
<td>18 Stomach Esop Duodenum</td>
<td>189.3</td>
<td>0.802</td>
</tr>
<tr>
<td>19 Testis Prostate</td>
<td>190.4</td>
<td>0.801</td>
</tr>
<tr>
<td>20 Cornea</td>
<td>202.7</td>
<td>0.893</td>
</tr>
</tbody>
</table>
Fig. 4-11 Adjustment of electrical properties of the high content water tissue phantom at 21 MHz.
Table 4-7  Achievable electrical properties of high content water tissues at 21 MHz.

<table>
<thead>
<tr>
<th>Tissue name</th>
<th>Relative permittivity</th>
<th>Conductivity [S/m]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 Tendon</td>
<td>85.1</td>
<td>0.430</td>
</tr>
<tr>
<td>2 Trachea</td>
<td>91.4</td>
<td>0.486</td>
</tr>
<tr>
<td>3 Nerve (Spinal chord)</td>
<td>94.8</td>
<td>0.256</td>
</tr>
<tr>
<td>4 Gall Bladder</td>
<td>97.4</td>
<td>0.910</td>
</tr>
<tr>
<td>5 Lens Cortex</td>
<td>97.6</td>
<td>0.550</td>
</tr>
<tr>
<td>6 Muscle (Parallel Fiber)</td>
<td>101.6</td>
<td>0.698</td>
</tr>
<tr>
<td>7 Cartilage</td>
<td>104.5</td>
<td>0.402</td>
</tr>
<tr>
<td>8 Thyroid Thymus</td>
<td>104.6</td>
<td>0.743</td>
</tr>
<tr>
<td>9 Ave. Muscle</td>
<td>104.8</td>
<td>0.671</td>
</tr>
<tr>
<td>10 Muscle (Transverse Fiber)</td>
<td>107.9</td>
<td>0.644</td>
</tr>
<tr>
<td>11 Lung (Deflated)</td>
<td>116.8</td>
<td>0.472</td>
</tr>
<tr>
<td>12 Eye Tissue (Sclera)</td>
<td>120.5</td>
<td>0.832</td>
</tr>
<tr>
<td>13 Tongue</td>
<td>120.5</td>
<td>0.602</td>
</tr>
<tr>
<td>14 White Matter</td>
<td>122.5</td>
<td>0.206</td>
</tr>
<tr>
<td>15 Skin (Wet)</td>
<td>132.8</td>
<td>0.411</td>
</tr>
<tr>
<td>16 Liver</td>
<td>138.1</td>
<td>0.364</td>
</tr>
<tr>
<td>17 Stomach Esop Duodenum</td>
<td>138.2</td>
<td>0.825</td>
</tr>
<tr>
<td>18 Testis Prostate</td>
<td>139.4</td>
<td>0.825</td>
</tr>
<tr>
<td>19 Dura</td>
<td>141.3</td>
<td>0.604</td>
</tr>
<tr>
<td>20 Blood</td>
<td>149.3</td>
<td>1.144</td>
</tr>
<tr>
<td>21 Cornea</td>
<td>149.7</td>
<td>0.922</td>
</tr>
<tr>
<td>22 Ave. Brain</td>
<td>158.7</td>
<td>0.294</td>
</tr>
<tr>
<td>23 Colon (Large Intestine)</td>
<td>162.8</td>
<td>0.546</td>
</tr>
<tr>
<td>24 Uterus</td>
<td>173.0</td>
<td>0.827</td>
</tr>
<tr>
<td>25 Heart</td>
<td>182.9</td>
<td>0.564</td>
</tr>
</tbody>
</table>
4.3.2 Low water content tissue

Until now, various phantoms have been developed. These phantoms are mainly homogeneous and high water content tissues. However, the tissue structure of the human body is inhomogeneous. For example, in the HF band the composition of multi-layer generally has an impact on the heating performances in the phantoms [15], the impedance of the antenna close to the phantoms [57], and so forth. Therefore, it is necessary to develop the multi-layered phantoms in the HF band. In other words, require the high water content tissue equivalent phantom as well as the low water content tissue equivalent phantom.

Relative permittivity of low water content tissue such as a fat is relatively-low. Thus, in this study, silicone emulsion, which has low relative permittivity, was chosen as a base compound of the fat equivalent phantom because of relative permittivity of silicone emulsion is low. As an example, in this dissertation, the author focused on the phantom at 21 MHz, because 21 MHz is one of the frequencies in IEEE 802.15.6-2012 for body centric wireless communications.

Here, electrical properties of the fat phantom are adjusted in a similar way as 4.3.1. The composition of the fat equivalent phantom is shown in Table 4-8. The proposed fat equivalent phantom consists of glycerin, silicone emulsion, cooking oil, agar, polyethylene powder and sodium chloride. Fat has a relatively low relative permittivity because of the low water content of the tissue. Therefore, silicone emulsion which has low relative permittivity is chosen as a base compound of the fat phantom. The electrical property of the fat phantom is adjustable by changing the amount of polyethylene powder and sodium chloride [43]. In particular, relative permittivity can be changed by the amount of polyethylene powder. In addition, conductivity can be
changed by polyethylene powder and sodium chloride, respectively. Additionally, the differences of the electrical properties between the measured value and the target value [47] of the fat equivalent phantom are shown in Table 4-9. From the results, the differences of relative permittivity and conductivity are within 10 %, respectively. At last, the fabricated three-layered phantom is shown in Fig. 4-11. Composition of multi-layered phantom is shown Table 4-10 and difference of electrical properties between measured and target value at 21 MHz is shown Table 4-12. As a result, we confirmed that the electrical properties of the proposed phantom and multi-tissues (muscle, fat and skin) are almost the same.

<table>
<thead>
<tr>
<th>Material</th>
<th>Amount [g]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Agar</td>
<td>25.0</td>
</tr>
<tr>
<td>Sodium chloride</td>
<td>1.0</td>
</tr>
<tr>
<td>Glycerin</td>
<td>102.0</td>
</tr>
<tr>
<td>Silicone emulsion</td>
<td>306.0</td>
</tr>
<tr>
<td>Cooking oil</td>
<td>102.0</td>
</tr>
<tr>
<td>Polyethylene powder</td>
<td>200.0</td>
</tr>
</tbody>
</table>

Table 4-8  Composition of the fat equivalent phantom at 21 MHz.
Table 4-9  Error of electrical properties between fabricated fat phantom and target value.

<table>
<thead>
<tr>
<th></th>
<th>Relative permittivity</th>
<th>Conductivity [S/m]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Target value</td>
<td>9.5</td>
<td>0.032</td>
</tr>
<tr>
<td>Measured value</td>
<td>10.0</td>
<td>0.031</td>
</tr>
<tr>
<td>Error [%]</td>
<td>+ 5.26</td>
<td>− 3.13</td>
</tr>
</tbody>
</table>

Table 4-10  Compositions of proposed phantom and error of electrical properties between measured and target value.

<table>
<thead>
<tr>
<th>Material</th>
<th>Muscle</th>
<th>Skin</th>
<th>Fat</th>
</tr>
</thead>
<tbody>
<tr>
<td>Deionized Water</td>
<td>538.0</td>
<td>538.0</td>
<td>-</td>
</tr>
<tr>
<td>Agar</td>
<td>22.5</td>
<td>22.5</td>
<td>25.0</td>
</tr>
<tr>
<td>Aluminum powder</td>
<td>100.0</td>
<td>276.0</td>
<td>-</td>
</tr>
<tr>
<td>Sodium chloride</td>
<td>1.2</td>
<td>0.27</td>
<td>1.0</td>
</tr>
<tr>
<td>Glycerin</td>
<td>-</td>
<td>-</td>
<td>102.0</td>
</tr>
<tr>
<td>Silicone emulsion</td>
<td>-</td>
<td>-</td>
<td>306.0</td>
</tr>
<tr>
<td>Cooking oil</td>
<td>-</td>
<td>-</td>
<td>102.0</td>
</tr>
<tr>
<td>Polyethylene powder</td>
<td>-</td>
<td>-</td>
<td>200.0</td>
</tr>
</tbody>
</table>
Table 4-10 Error of electrical properties between measured and target value at 21 MHz.

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Relative permittivity</th>
<th>Conductivity [S/m]</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Muscle</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Target value</td>
<td>104.8</td>
<td>0.671</td>
</tr>
<tr>
<td>Measured value</td>
<td>106.2</td>
<td>0.650</td>
</tr>
<tr>
<td>Error [%]</td>
<td>+ 1.33</td>
<td>− 3.13</td>
</tr>
<tr>
<td><strong>Skin</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Target value</td>
<td>201.3</td>
<td>0.296</td>
</tr>
<tr>
<td>Measured value</td>
<td>204.0</td>
<td>0.316</td>
</tr>
<tr>
<td>Error [%]</td>
<td>+ 1.34</td>
<td>+ 6.76</td>
</tr>
<tr>
<td><strong>Fat</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Target value</td>
<td>9.5</td>
<td>0.032</td>
</tr>
<tr>
<td>Measured value</td>
<td>10.0</td>
<td>0.031</td>
</tr>
<tr>
<td>Error [%]</td>
<td>+ 5.26</td>
<td>− 3.13</td>
</tr>
</tbody>
</table>

Fig. 4-12 Fabricated multi-layered phantom.
4.4 Biological tissue-equivalent phantom for VHF band

In recent years, the variety of antennas for VHF band application is also widely investigated [14]-[17]. For example, talk (e.g. portable radio terminals), logistics or individual authentication using RFID (Radio Frequency IDentification) system, medical applications (e.g. Magnetic Resonance Imaging (MRI), wireless power transmission to capsular endoscope), and so on. Therefore, phantom for VHF band is also required to evaluate the interaction between EM waves and human body.

 Particularly, recently, the use of higher frequency (128, 170, 200, 300 MHz and so on) antenna and the power increase for pulsing EM waves are being to obtain high quality images and to shorten the acquisition time in MRI fields. Therefore, the SAR inside the body during the RF pulse radiation is necessary to be evaluated. In addition, as for designing the antenna used for MRI system, the reduction of the SAR in the human body is particularly important to be considered. From these reasons, the author focused on the phantom at 128 MHz, because 128 MHz is one of the frequencies in MRI system.
4.4.1 Example of a phantom for VHF band

The composition of the muscle equivalent phantom at 128 MHz is shown in Table 4-11 as an example. In addition, fabricated muscle phantom at 128 MHz is shown in Fig. 4-15. Here, electrical properties of the muscle phantom is adjusted in a similar way as 4.3.1. The proposed muscle equivalent phantom at 128 MHz consists of deionized water, glycerin, agar, sodium chloride and sodium dehydroacetate. Deionized water and glycerin are used as base compound of phantom. Agar is used to make the phantom hard enough to maintain the shape. Sodium chloride is used to control the values of conductivity. Sodium dehydroacetate is used primarily as a preservative substance. Relative permittivity can be changed by water-to-glycerin ratio. In addition, conductivity can be changed by amount of sodium chloride. Fig. 4-16 shows electrical properties of proposed phantom. From results, the differences between the target values and the measured electrical properties are 0.7 % and 2.9 % for relative permittivity and conductivity, respectively.
Table 4-11 Composition of muscle phantom at 128 MHz.

<table>
<thead>
<tr>
<th>Material</th>
<th>Amount [g]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Deionized Water</td>
<td>3920</td>
</tr>
<tr>
<td>Glycerin</td>
<td>3675</td>
</tr>
<tr>
<td>Ager</td>
<td>350</td>
</tr>
<tr>
<td>Sodium Chloride</td>
<td>126</td>
</tr>
<tr>
<td>Sodium Dehydroacetate</td>
<td>14</td>
</tr>
</tbody>
</table>

Fig. 4-15 Fabricated phantom for 128 MHz.
Fig. 4-16  Electrical properties of fabricated phantom.
4.5 Summary in Chapter 4

In this chapter the biological tissue-equivalent phantom for HF and VHF bands were described.

In section 4-2, electrical properties of base compound of the phantom were investigated. In this dissertation, deionized water, glycerin and silicone emulsion were used as a base compound of the phantom. As a result, it is confirmed that conductivity of these base compounds were lower than biological tissues. Relative permittivity of deionized water was the highest of three.

In section 4-3, the electrical properties of the phantoms in HF band were adjusted. In the case of high water content tissues, water was chosen as a base compound of the phantom. In addition, the electrical properties are adjustable by changing the amount of aluminum powder and sodium chloride. In particular, relative permittivity can be changed by the amount of aluminum powder because it is assumed that the capacitance of the phantom is increased by the addition of aluminum powder. Meanwhile, conductivity can be changed by aluminum powder and sodium chloride, respectively. In the case of low water content tissue such as a fat, silicone emulsion which has low relative permittivity is chosen as a base compound of the phantom. The electrical property of the fat phantom is adjustable by changing the amount of polyethylene powder and sodium chloride. In particular, relative permittivity can be changed by the amount of polyethylene powder. In addition, conductivity can be changed by polyethylene powder and sodium chloride, respectively. At last, composition of phantom for VHF band is presented.
5 Evaluation of the interaction between EM waves and the human body using the phantom

5.1 Introduction

Various investigations have been undertaken mainly on the interaction between the cellular phones and the human body at UHF band. For the future, it is necessary to evaluate the interaction between the EM waves and human body in other situations (e.g. frequency, devices, etc.). Especially, recently, the variety of antennas for HF-VHF bands application is widely investigated [18]-[21]. For example, portable radio terminals, logistics or individual authentication using RFID system, medical applications (e.g. body-centric wireless communications, hyperthermia, magnetic resonance imaging (MRI), etc.). Generally, the interactions should be estimated from numerical simulations and experimental evaluations. In these evaluations, phantoms are usually used [18]-[43]. Therefore, in this chapter, the author investigated evaluation of the interaction between EM waves and the human body using numerical and experimental phantoms. As an example, the author focused on magnetic resonance imaging (MRI) system in this dissertation.
5.1.1 MRI system

MRI is being widely used to obtain clear images inside the human body, especially of high water content tissues. The fundamental principle of MR imaging is to receive nuclear magnetic resonance (NMR) signals induced by irradiating the human body placed inside a strong static magnetic field with electromagnetic (EM) wave pulses, which is placed inside a strong static magnetic field. The MRI system is composed of various elements including a RF coil, which plays an essential role in imaging [58]. Several types of RF coils, such as birdcage coils, saddle coils, surface coils, TEM coils etc. have been developed and used depending on the body part to be imaged [66]-[69]. In this dissertation, a birdcage coil is employed as one of the most fundamental RF coils for MRI system. In addition, the birdcage coil is often used as both the transmission of RF pulse and reception of NMR signal. It is necessary to use an EM wave in certain specified frequency to emit the NMR signal. Here, frequency of EM fields, \( f \) [Hz] is given by the following equation (Lamor equation).

\[
    f = \gamma \cdot B
\]  

(5-1)

where \( f \) is the Lamor precession or resonant frequency of the nucleus [Hz], \( \gamma \) is the gyromagnetic ratio of nuclear [Hz/T]. In the case of water composed of hydrogen atoms, \( \gamma = 42.6 \) MHz/T. Additionally, \( B \) is the magnitude of the static magnetic flux density [T].

In this dissertation, the author describes magnetic flux density as magnetic field for reasons of expediency. Thus, frequency to emit NMR signal is risen in directly proportional to strength of the static magnetic field.
5.2 Measurement method of SAR evaluation for applying thermographic method

In recent years, various types of implantable medical devices have been developed. For example, cardiac pacemaker, stent, artificial joint, and so on. In addition, opportunities for MRI examinations of patients are increased, and there are many patients with implanted metallic medical devices. The materials and shapes of implantable medical devices such as carotid artery stents (CAS) has become increasingly diverse, making it more likely that subjects’ implanted medical devices will affect MR imaging, potentially causing temperature increases in unexpected regions. Despite this, guidelines for evaluating the effects of such cases on the human body are still unclear. Therefore, in order to consider the safety of patients with implanted medical devices during MR imaging, it is necessary to comprehend the detailed SAR distribution within the human body. While many existing studies have addressed cases involving pacemakers, none have dealt with carotid stents [63]. In Japan, 10,000 carotid endarterectomies (CEA) and carotid artery stents are carried out each year.

By the way, modern MRI systems employ high-frequency and high-powered radio frequency (RF) pulses. Given this, it is necessary to evaluate the SAR of the irradiated human body. Several studies have been performed the SAR measurement employing the actual RF coil. These studies only measured several points of temperature rise inside the phantom to evaluate the SAR [64]-[68]. Especially, the cross section of the SAR distribution inside had not been measured. According to the International Electrotechnical Commission (IEC) standards, there are two methods that
can measure the total absorption energy of human body model. One is “calorimetric method”. The other is “pulse energy method” [69]. However, these methods cannot be used to evaluate detailed local SAR distribution inside the human body. In order to investigate the factor that high SAR produces, the SAR measurement by using actual imaging coil is one of effective method. Moreover, the measurement technique is useful and required when employing conventional RF coils with unknown inner structure and also include active elements such as transistors, diode, etc., which are difficult to investigate in the numerical calculation.

This dissertation intended to establish measurement technique to estimate detailed SAR distribution inside the phantom which was generated by the conventional birdcage coil for 3T MRI systems. Here, the author evaluates the SAR distribution within the head, using carotid stents as an example of implanted medical devices. The characteristics of the coil and the SAR distribution inside the head phantom were measured by using this coil. Then, the validity of the measured results was confirmed in comparison with calculation results by FDTD calculations.

5.2.1 Thermographic method

Thermographic method is one of the SAR evaluation methods based on temperature measurement [30]-[32]. This method is basically independent on the frequency of irradiated EM wave since measured temperature rise convert to the EM energy absorption. In addition, the feature of thermographic method is possible to measure a two-dimensional SAR on the surface or arbitrary cross section of the phantom. Particularly, it is important to know the SAR distribution near the surface because the EM energy is mainly absorbed around the surface by skin-effect.
The measurement method was as following steps. First, the phantom is exposed to the EM waves radiated from the coil. Then, temperature rise on the arbitrary cross section or surface of the phantom is observed by use of the infrared camera. If the cross section is observed, the phantom is needed to cut at position for obtaining observation plane before exposure. In the thermographic method, the sufficient temperature rise (above 2~3 K) inside the phantom during short irradiation time is needed for evaluating the precision SAR distribution.

In order to minimize the effect on the heat diffusion in the phantom, the exposure time must be as shot as possible within a range that makes the temperature distribution observable. If the heat diffusion is negligibly small, the SAR is proportional to the temperature rise inside the phantom. The SAR at arbitrary point can be estimated by the following equation:

\[
SAR = c \frac{\Delta T}{\Delta t}
\]

where, \( c \) is the specific heat of the phantom [J/kg•K], \( \Delta T \) is the temperature rise [K] at the point, and \( \Delta t \) is the exposure duration [s].

5.2.2 Evaluation models

Figure 5-1 shows SAR evaluation model for birdcage coil loaded with head model. From the Lamor equation, operating frequency of the coil is approximately 128 MHz. This coil consists of two endrings and sixteen elements, whose width are 15 mm and 10 mm, respectively. The diameter and the length of the birdcage coil are 300 mm and 330 mm, which is sufficiently large for the head model to be set inside it. In addition, the coil is enclosed in a RF shield. The diameter and the length of the RF shield are 400 mm
and 440 mm, respectively. All these are composed of metallic sheets. A cylindrical biological tissue-equivalent phantom is used as head model. Table 5-1 shows composition of the phantom. This phantom is 180 mm in diameter, and 250 mm in length and is made of a uniform medium of a muscle tissue. The target value of electrical properties of the phantom are \( \varepsilon_r = 63.9 \) and \( \sigma = 0.74 \text{ S/m} \), respectively at 128 MHz. Here, the differences between the target values and the measured electrical properties are 0.3 % and 1.1 % for \( \varepsilon_r \) and \( \sigma \), respectively. Thus, there is no influence on the birdcage coil performances with such differences. The specific heat of the phantom used in the measurement was 3180 J/kg•K. Moreover, in order to control the resonance frequency to approximately 128 MHz, a total of thirty two capacitors were attached to the both endrings of the coil. The coil used for the measurement was CBU-3HBC-001 manufactured by the Takashima Corporation Ltd., Tokyo, Japan.

Figure 5-2 illustrates the experiment setup by using the birdcage coil. As for measurement setup, the procedure is almost same as noted above. First, the phantom was divided into top and bottom halves to acquire the observation plane. In order to capture the SAR distribution of the head model with a carotid stent in the coil, carotid stent was inserted into a bottom phantom. Figure 5-3 (a) shows the implanted position of the carotid stent. The implanted position is 10 mm from the body surface and 195 mm from the top. Figure 5-3 (b) shows the simplified carotid stent model of 10 mm × 10 mm × 40 mm. Because the size of the stent is much smaller than the wavelength of a 128 MHz RF wave, a square tube structure. Next the whole phantom was set in the center of the coil. The sinusoidal wave was divided equally at the splitter with difference of 90 degree output phase. After the irradiation, the phantom was split into half immediately placed in front of the thermographic camera to image the temperature
rise on the observation plane. Here, incident power and duration were 100 W and 180 s, respectively. The irradiation frequency was determined by placing the phantom inside the coil and then identifying the frequency corresponding to the best impedance matching. Figure 5-4 shows input characteristic of the birdcage coil.

![Diagram showing birdcage coil with feeding points](image)

**Fig. 5-1** SAR evaluation model for birdcage coil loaded with head model.

<table>
<thead>
<tr>
<th>Material</th>
<th>Amount [g]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Deionized Water</td>
<td>3920</td>
</tr>
<tr>
<td>Glycerin</td>
<td>3675</td>
</tr>
<tr>
<td>Ager</td>
<td>350</td>
</tr>
<tr>
<td>Sodium Chloride</td>
<td>126</td>
</tr>
<tr>
<td>Sodium Dehydroacetate</td>
<td>14</td>
</tr>
</tbody>
</table>

**Table 5-1** Composition of fabricated brain phantom.
Experiment setup by using the birdcage coil.

(a) Implanted position of the carotid stent.

(b) Structure of the carotid stent model.

Fig. 5-3 Implanted position and structure of the stent.
5.2.3 SAR distributions

To remove the influence of temperature diffusion, correction of relatively high SAR value point in the x-y plane was carried out as an example. Inside the phantom, the observation point is located at $x = 56$ mm, $y = 56$ mm, $z = 135$ mm. To begin with, the fiber optic thermosensor was inserted into the observation point and the temperature transition throughout the exposure was measured. The observation point is illustrated by cross mark in Fig. 5-6. Next, an approximate curve was calculated from the temperature transition values by using the following equation:

$$f(t) = a(1 - e^{-bt})$$  \hspace{1cm} (5-2)

where, $t$ is the exposure time. $a$ and $b$ are constant values. After calculating the amount of $a$ and $b$ from the measured temperature rise up to 180 s, a line of slope $f'(0)$ which can neglect the temperature diffusion was obtained. For example, temperature rise of the correction position is shown in Fig. 5-6. At last, from this figure, the correction
coefficient $A = f'(0) \times 180 \text{ s} / \Delta T_m$ was led and multiplied with the SAR values measured at the observation points. Here, $f'(0) \times 180 \text{ s}$ is the temperature rise after the correction and $\Delta T_m$ is the temperature rise obtained in the measurement. The correction for the SAR value at the observation point is listed in Table 5-2. It is observed from Table 5-2 that the measured SAR values got closer to the calculated results at observation point by using the correction method. Furthermore, the difference observed in the correction position decreased from 42% to 2%. Figure 5-7 (a) shows the SAR distributions in the x-y plane (at $z = 0 \text{ mm}$). Here, the SAR values are normalized by 1 W input power. As a result, peak SAR was observed nearby stent. It is considered that E-field is concentrated edge of the stent model. In addition, it is observed that the SAR distributions obtained by measurement after correction and FDTD calculation are good agreement either after remove the influence of temperature diffusion.

![Fig. 5-5 Correction for temperature diffusion.](image-url)
Table 5-2  Correction of SAR.

<table>
<thead>
<tr>
<th></th>
<th>SAR [W/ kg]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Calculated SAR</td>
<td>0.59</td>
</tr>
<tr>
<td>value</td>
<td></td>
</tr>
<tr>
<td>Before correction</td>
<td>0.34</td>
</tr>
<tr>
<td>After correction</td>
<td>0.60</td>
</tr>
</tbody>
</table>

Fig. 5-6  Temperature observation point.
Fig. 5-7   SAR distributions.
5.3 SAR evaluation of human with metal implant during MR imaging [J-3]

In the section 5-2, detailed SAR distribution inside the homogeneous phantom with medical device is investigated. However, actual human body is not homogeneous. Therefore, it is necessary to realistic evaluation using heterogeneity phantom. In this section, the author evaluates the SAR distribution of realistic leg model with artificial knee joint as an example of implanted medical devices during 3T MRI imaging.

5.3.1 Numerical calculation model and condition

(1) Structure of the birdcage coil

Figure 5-8 shows the birdcage coil with RF shield. This coil consists of two end-rings (widths are 16mm) and sixteen equally spaced elements (widths are 10mm). Two circular rings are attached on the top and bottom of each element. Diameter and length of birdcage coil are 300mm and 330mm, respectively. The dimension of the coil was determined by reference [61], [63], and [70]. In addition, diameter and length of RF shield are 400mm and 440mm, respectively. The birdcage coil and RF shield are composed of perfect electric conductors which do not consider the thickness of sheet.

Figure 5-9 shows the enlarged view near to the feeding point of the model. As for the highpass birdcage coil, the lumped capacitors were loaded into the end-ring at midpoints between the elements. The values of capacitors to resonate birdcage coil at 128 MHz were 23.5–26.7pF. Additionally, the phase difference between two feeding points is 90 degree.
Fig. 5-8  Structure of the birdcage coil.

Fig. 5-9  Enlarged view around the feeding point.
(2) Realistic human model and artificial knee joint model

In this dissertation, realistic high-resolution human model of Japanese adult male was employed. The realistic high-resolution human model of Japanese adult male for EM dosimetry has been developed by the NICT (National Institute of Information and Communications technology) [27]. This model has average height and weight. The resolution of this model is $2 \text{ mm} \times 2 \text{ mm} \times 2 \text{ mm}$, and segmented into 51 different tissues and organs. Here, the author use only leg part of realistic high-resolution human model because of most part of the EM field is concentrated inside the coil. Figure 5-10 show realistic human model used in this study and calculated region. In this study, length of leg is 400mm.

Figure 5-11 shows structure of artificial knee joint. This artificial knee joint model is primarily consisted of metal parts (femoral component and tidal component) and dielectric parts (plastic spacer and patellar component). Figure 5-12 shows artificial knee joint model used in this study. This artificial knee joint model was designed based on three-dimensional CAD data file of actual artificial knee joint. Here, metal parts were modeled as a PEC. In addition, artificial knee joint model is integrated with leg model. Figure 5-13 shows human model who is implanted the artificial knee joint. Here, coordinate origin is central location of the artificial knee joint.

In order to incorporate the leg with artificial knee joint model into FDTD method, their electrical properties were required. The electrical properties corresponding to the tissues and organs of human models were mostly taken from the 4-Cole-Cole analysis as reported by Gabriel [48].
Fig. 5-10  Realistic whole-body voxel model of Japanese man.

Fig. 5-11  Structure of artificial knee joint.
(a) Enlarged view of artificial knee joint.

(b) Dimension of artificial knee joint.

Fig. 5-12  Artificial knee joint model.
Fig. 5-13 Human model who is implanted the artificial knee joint.

(2) Numerical calculation method and conditions

In the numerical calculation, the electric field in and around the coil is calculated by the FDTD method. The parameters used in FDTD calculations are listed in Table 5-3. In the numerical calculation, the leg model is placed in the center of the coil. The size of the cell is $\Delta x = \Delta y = \Delta z = 2.0\text{mm}$ and calculation space is $470 \times 470 \times 920$ cell. The absorbing boundary condition is 8 layers of Perfectly Matched Layer (PML) and time step is 3.84 ps. In addition, the steady state analysis is performed by enforcing a continuous sinusoidal wave of the electric field into the feeding gap to calculate the SAR distribution of the model. Moreover, the coordinate origin is set at the center of the birdcage coil. The SAR can be the following equation:

$$SAR = \frac{\sigma}{\rho} E^2 \quad [\text{W/kg}] \quad (5-3)$$
where $\sigma$ is the conductivity of the tissue [S/m], $\rho$ is the density of the tissue [kg/m$^3$], and $E$ is the RMS (root mean square) value of electric field in the tissue [V/m]. The SAR value is proportional to the square of the electric field generated inside the human body. Since the SAR is amount of absorbed EM energy in lossy medium per unit mass, it is equivalent to the heating source induced by the electric field in the human tissue. In addition, it has been confirmed that the result of numerical calculation corresponded with the measured result by employing a tissue-equivalent phantom.

Figure 5-14 shows the input impedance of the birdcage coil with leg model. As for impedance calculation, a derivative Gaussian pulse is exited at feeding point at the position $\phi=0^\circ$. This is useful for FDTD calculations where a loop structure exists. Here, impedance-matching of the birdcage coil is not considered because MRI system usually has matching circuit. As a result, the resonance frequency was confirmed to be approximately 128 MHz. In other words, birdcage coil which is used in this study can use 3 T MRI systems.
Table 5-3  Calculation conditions.

<table>
<thead>
<tr>
<th>Cell size [mm]</th>
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</table>

Fig. 5-14  Input impedance of the birdcage coil.
5.3.2 Calculated results

(1) Comparison of SAR between realistic and simplified leg model

In order to examine the effects of heterogeneity, homogeneous model should be considered for comparison. Figure 5-15 shows the simplified leg model. This model is 116 mm in diameter, 400 mm in length and is made of a uniform medium of the muscle. Here, electrical properties of the muscle tissue are \( \varepsilon_r = 63.9 \) and \( \sigma = 0.74 \) S/m, and density is 1040 kg/m\(^3\).

Figure 5-16 shows calculated SAR distributions inside leg model by use of birdcage coil for 3 T MRI systems. Here, the SAR value is normalized by the maximum value in the realistic model. In addition, observation planes are \( xy \) plane at \( z = 50 \) mm and \( xz \) plane at \( y = 24 \) mm including maximum SAR value. From Fig. 5-16 (a), relative high SAR value was observed around surface of the body. On the other hand, From Fig. 5-16 (b), relative high SAR value was observed around deep body. This is due to conductivity of muscle is higher than conductivity of other tissues such as skin or fat. Figure 5-17 show peak SAR in the calculation region. In the actual MRI system, various pulse sequences are employed according to the imaging region and its type. However, in this study, the SARs in the human models were calculated by exiting continuous sinusoidal wave at feeding point, because the input power of each pulse sequence was obscure. Here, in order to simplify conversion of input power, the SAR values are normalized by 1 W input power to the birdcage coil for all cases. From results of Fig. 5-17, peak SAR of each model are 28.7 W/kg (in the realistic model) and 6.4 W/kg (in the simplified model). In other words, the author confirmed that peak SAR in the realistic model is 4.5 times higher than peak SAR in the simplified model.

The spatial average SAR over 1g- or 10g- mass (1g- or 10g averaged SAR) is
adopted in order to evaluate the near-field exposure from the RF devices by the guidelines and the regulations on an international basis. In addition, in order to protect the human body during performing MRI, IEC has established a standard to limit maximum energy absorption (SAR) within the human subjects. In this study, the 10g averaged SAR is calculated using averaging technique of IEEE [71]. Figure 5-18 shows peak 10g averaged SAR. Here, the 10g averaged SARs were normalized by 1 W input power. From results of Fig. 5-18, peak 10g averaged SAR of each model are 2.09 W/kg (in the realistic model) and 1.26 W/kg (in the simplified model). In other words, the author confirmed that peak 10g averaged SAR in the realistic model is 1.6 times higher than peak 10g averaged SAR in the simplified model. From these results, it has confirmed that the realistic model is indispensable to accurate evaluate the SAR in human body.

Fig. 5-15  Simplified leg model.
Fig. 5-16  SAR distributions.
Fig. 5-17  Peak SAR of the simplified model and the realistic model.

Fig. 5-18  Peak 10g averaged SAR of the simplified model and the realistic model.
(2) SAR distribution in the realistic leg model

Here, the author describes influence of the artificial knee joint on SAR distribution. Figure 5-19 show SAR distributions inside the realistic leg model with or without the artificial knee joint. Here, input power to the birdcage coil was equivalent in all cases. Here, the SAR value is normalized by the maximum value in realistic leg with artificial knee joint model. In addition, observation planes including maximum SAR value are $xy$ plane at $z = 50$ mm and $xz$ plane at $y = 24$ mm. From these results, it was confirmed that SAR distributions of each case show different trends. In the case of the leg without artificial knee joint, peak SAR value was observed around surface of the body ($x = 56$ mm, $y = 22$ mm, $z = 66$ mm). On the other hand, in the case of the leg with artificial knee joint, peak SAR value was observed around surface of the body ($x = 38$ mm, $y = 22$ mm, $z = 48$ mm).

Figure 5-20 shows the peak SAR value. Here, the SAR values are normalized by 1 W input power to the birdcage. From results of Fig. 5-20, peak SAR of each model are 28.7 W/kg (with artificial knee joint) and 3.84 W/kg (without artificial knee joint). In other words, the author confirmed that peak SAR in the realistic leg with artificial knee joint model is 7.46 times higher than peak SAR in the without artificial knee joint model. This is due to the concentrated electric field at edge of metal parts of the artificial knee joint.

Figure 5-21 shows the peak 10g averaged SAR value. Here, the SAR values are normalized by 1 W input power. From results of Fig. 5-21, peak 10g averaged SAR of each model are 2.09 W/kg (with artificial knee joint) and 0.66 W/kg (without artificial knee joint). In other words, the author confirmed that peak 10g averaged SAR in the
realistic leg with artificial knee joint model is 3 times higher than the without artificial knee joint model. Subsequently, 10g average SARs were compared to restriction value of the IEC 60601-2-33. Here, in the case of leg, restriction value is 20 W/kg. As a result, it was confirmed that calculated peak 10g averaged SAR were smaller than restriction value of the IEC regulation.
Fig. 5-19  SAR distributions.
Fig. 5-20  Peak SAR of the realistic model.

- Without artificial knee joint: 3.84 W/kg
- With artificial knee joint: 28.7 W/kg

Fig. 5-21  Peak 10g averaged SAR of the realistic model.

- Without artificial knee joint: 0.66 W/kg
- With artificial knee joint: 2.09 W/kg
5.4 Measurement method of magnetic field distribution inside the phantom with metal implant

Recently, the use of higher frequency antenna and the power increase for pulsing electromagnetic wave are being addressed to obtain high quality images and to shorten the acquisition time. Magnetic field inhomogeneity due to shortened wavelength at higher frequency is a major cause of MRI nonuniformity in human body. As for designing the coil used for MRI system, the magnetic field homogeneity in the human body is particularly important to be considered. In addition, in recent years, patient who is implanted metallic medical devices have a greater opportunity to MRI examination. However, effects of implanted metallic medical devices on magnetic field inside the human body are still unclear. Therefore, magnetic field distribution inside the human body during MR Imaging is necessary to be evaluated. In this section, the author investigates each component of the magnetic field inside the human body. Then, the validity of the measured results was confirmed in comparison with calculation result by FDTD method.
5.4.1 Measurement method and measurement model

Magnetic fields are picked up using loops or wire, and in turn measuring the induced voltage across the ends of the wire. In this study, the shielded loop antenna with diameter of 5 mm was used as the magnetic field probe. This shielded loop antenna made from semirigid coaxial cable. At the tip of the shielded loop antenna, gap is constructed to generate the received signal voltage. A spectrum analyzer and power amplifier are also used. Experimental setup is shown in Fig. 5-22. Dimensions and position of the birdcage coil and the phantom are the same in Section 5-2. Here, gel phantom is used as a top side of the phantom because gel phantom can insert shielded loop antenna. Therefore, we can measure magnetic field inside the phantom. Gel phantom is filled in dielectric shell ($\varepsilon_r=2.1$) and thickness of dielectric shell is 3mm. The composition of the gel phantom for measurement of magnetic field distribution is listed in Table 5-4. At 128 MHz, the differences between the target values of electrical properties of the gel phantom and the measured electrical properties are less or equal 10 % for relative permittivity and conductivity, respectively.
5.4.2 Magnetic field distributions inside the phantom

Figure 5-23 shows magnetic field distribution on the observation line. Then, $H_x$ and $H_y$ were measured by orienting the probe antenna along the x-axis at $y = 0$ mm in Fig. 5-7. In addition, the $H_z$ component is ignored because its level is very low as compared to $H_x$ and $H_y$. The levels are normalized with the maximum value as 1.0. As a result,
maximum level of magnetic field is observed at $x = 75$ mm. This is due to influence of the stent. In addition, good agreement was observed between the measured and numerical calculated results of the magnetic field level. The difference between measurement and calculation was less than 10 %.

Fig. 5-23  Magnetic field distribution on the observation line.
5.5 Antenna design for MRI systems considering effect of human body [J-4]

For the future, the quantitative evaluation of the interaction between EM wave and the human body, which is radiated by the devices, is indispensible. In MRI field, RF coils close to the human body cause the variation in the input characteristics of the RF coil and magnetic field distribution inside the human body. Therefore, it is necessary to evaluate an influence of the human body on the performance of the RF coil nearby the human body. For the reason noted above, in this section, antenna design for MRI systems considering effect of human body is performed. Here, as one of the example, a birdcage coil for 4 T MRI systems with no lumped circuit elements was proposed.

5.5.1 Purpose of this section

Recently, MRI system which operates up to 3 T is being used in clinical practice in Japan [70]. In order to achieve the requirement of obtaining more high-quality images and short imaging time, devices, which utilize high magnetic field (>3 T) and high power EM wave pulses, have been developed [72], [73]. However, some problems caused by high magnetic field and high power EM wave pulses. For instance, the increase of static magnetic field strength causes a rise in the frequency of EM pulses. In such a case, small variability of capacitances used in the birdcage coil, which is one of the most-used volume coils and is loaded a lot of uniform capacitances, will not be ignored. In addition, the use of high power EM pulses will induce breakdown of the capacitances. The improvement of the above problems would increase costs for manufacturing of the coil. It is predicted that these problems are going to become increasingly serious problems when high magnetic field (>3 T) is utilized with
devices for the MRI system. Hence, if we can develop the birdcage coil for MRI system with no lumped circuit elements such as capacitance, it is extremely useful. For this reason, we have developed birdcage coil for MRI system with no lumped circuit element [74]. This birdcage coil has circular conductors which are two different diameters. These conductors were constructed on both surfaces of dielectric cylinder. In this way, this birdcage coil can generate capacitance between circular conductors which are two different diameters. Due to this, birdcage coil does not need any capacitors as lumped circuit elements. However, if conductors on dielectric cylinder are out of position during process of manufacture, characteristics of this birdcage coil become depleted because capacitance changes between some conductors on both surfaces of dielectric cylinder. In other words, it is difficult to fabricate this birdcage coil. Therefore, simply-structured birdcage coil with no lumped circuit elements to reduce manufacturing error is desired. In this section, the author proposed the simply-structured birdcage coil with no lumped circuit elements to reduce manufacturing error. To be more specific, conductors of the birdcage coil with no lumped circuit elements were constructed on single side surface of dielectric cylinder. It is considered that proposed birdcage coil can be manufactured by technique of making a single side printed circuit board.
5.5.2 Numerical calculation model and condition

(1) Head model

Figure 5-24 shows the head model used in the numerical calculation and the experiment. This model is 180 mm in diameter, 250 mm in length and is made of a uniform medium of an average brain tissue [62]. The composition of the phantom for experiment is listed in Table 5-5. The target values of electrical properties of the phantom are $\varepsilon_r = 58.2$ and $\sigma = 0.49$ S/m at 170 MHz [48]. At 170 MHz, the differences between the target values and the measured electrical properties are 3.4 % and 4.1 % for relative permittivity $\varepsilon_r$ and conductivity $\sigma$, respectively. Thus, there is no influence on the birdcage coil performances with such differences.

![Fig. 5-24 Head model.](image)

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(2) Structure of the birdcage coil
The RF coil employed in this section is a birdcage coil for 4 T MRI system. From the eq. (5-1), operating frequency of the coil is approximately 170 MHz. Figure 5-25(a) shows structure of the proposed birdcage coil, birdcage coil with no lumped circuit element in previous study [74] is shown Fig. 5-25(b) and the conventional birdcage coil is shown Fig. 5-25(c) for comparison, respectively. Figure 5-26 shows sectional view of two birdcage coils. The birdcage coil consists of two circular conductors and some conductors which make a connection with circular conductors. The former is called end-ring and the latter is called element. Here, both coils consist of sixteen elements.

Moreover, in order to generate a homogenous magnetic field inside the birdcage coil and control the resonance frequency of the conventional birdcage coil to approximately 170 MHz, a total of thirty two capacitors were attached to the both end-rings of the conventional birdcage coil such as Fig. 5-25(c). In this section, all capacitors have 14.4pF based on an analysis of Birdcage Builder [75] which is computer program provided by Pennsylvania State University. The Birdcage Builder can be used to calculate the current pattern on a conducting surface of element location, and the necessary capacitances of the birdcage coil of a given geometry (diameter of birdcage coil, number of elements, required resonant frequency, and so on) by performing lumped circuit analysis using virtual ground assumptions.

Figure 5-25(a) is the proposed birdcage coil without the use of lumped circuit elements. In order to control the resonance frequency of the proposed birdcage coil, gaps are used instead some capacitors such as Fig. 5-25(a). Therefore, this birdcage coil does not need the capacitor that is the lumped circuit elements. Structure of the proposed birdcage coil is simpler than the coil in reference [74] because conductor of reference [74] was constructed on both surfaces of dielectric cylinder. Here, end-rings
and sixteen elements were constructed on the dielectric pipe ($\varepsilon_r=2.1$), whose pipe wall thickness is 3 mm.

In this section, two coils and RF shield are same in size such as Figs. 5-26(a) and (b). The diameter and the length of both coils are 300 mm and 330 mm each, which is sufficiently large for the head model to be set inside. Additionally, both coils are enclosed in a cylindrical RF shield. The diameter and the length of the RF shield are 400 mm and 440 mm, respectively. All these are composed of metallic sheets. Dimensions of coil and the cylindrical RF shield were determined by references [58], [68], and [69]. In addition, two feeding points were employed in these calculations and the phase difference between them is 90 degrees. This excitation method is called “quadrature excitation” and can generate uniform magnetic field inside the birdcage coil by using circular polarized field.
Fig. 5-25 Birdcage coil.

(b) Previous birdcage coil with no lumped circuit element. (c) Conventional birdcage coil.

Unit: mm

(a) Proposed birdcage coil. (b) Conventional birdcage coil.

Fig. 5-26 Sectional view of birdcage coils.
(3). Numerical calculation method and conditions

In the numerical calculation, the input impedance $|Z|$ at the feeding point of the birdcage coil was calculated by the FDTD method [47], [48], and the magnetic field inside the coil was calculated for comparison with the measurements. The parameters used in FDTD calculations are listed in Table 5-6.

Figure 5-27(a) shows the numerical calculation model of the birdcage coil and the head model. The head model is placed in the center of the coil. Figure 5-27(b) shows enlarged view of around the feeding point of proposed birdcage coil. Feeding points are modeled such a Fig. 5-27(b). For the calculation of the input impedance, a derivative Gaussian pulse was excited with 50-Ω internal resistance at the feeding gap at the position $\phi=0^\circ$. In contrast, a continuous sinusoidal wave was input at the feeding gaps at the positions $\phi=0^\circ$ and $\phi=90^\circ$ to calculate magnetic field distribution. In order to reduce computational resources, the non-uniform grid FDTD algorithm was adopted. The absorbing boundary condition is 8 layers of Perfectly Matched Layer (PML).

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Fig. 5-27 The numerical calculation model consisting of the birdcage coil and the head model (Coordinate origin is center of the coil).
5.5.3 Input impedance of the proposed RF coil

In this section, in order to adjust the resonant frequency, widths of gaps between end-rings \( g \) (see Fig.5-25(a)) are changed. Other parameters (e.g. overlap-length between three end-rings, distance between element and center end-ring, and so on) are maintained constant. Figure 5-28 shows relationships between resonant frequency and \( g \). In this dissertation, impedance-matching of the birdcage coil is not considered because MRI system usually has matching circuit. Therefore, absolute value of impedance is shown in this paper because it is only necessary to adjust the resonant frequency. As a result, the resonance frequency was confirmed to be approximately 170MHz, when widths of gaps are 0.2mm. In other words, absolute value of impedance has local maximum value at approximately 170MHz that is operating frequency of 4 T MRI system. Thus, it is considered that proposed coil when widths of gaps are 0.2mm and conventional coil which are loaded with capacitance of 14.4 pF are equivalent.

In order to compare the measured and calculated input impedances of the proposed birdcage coil containing the head model, when widths of gaps are 0.2mm are shown in Fig. 5-29. The continuous line is the calculated result and the broken line is the measured result. The input impedance was calculated at the feeding point at \( \phi=0^\circ \), same as the measured. Here, the measuring instrument was N5230C PNA-L network analyzer by the Agilent Technologies, Santa Clara, California.

As a result, good agreement between the measured and calculated results is seen, except for slight shifts in resonant frequency and absolute value of impedance. These shifts attributed to fabrication error in, such as, the dimension of gaps.
Fig. 5-28 Input impedance.

Fig. 5-29 The measured and calculated input impedance of the proposed coil.
5.5.4 Magnetic field distribution

It is considered that magnetic field homogeneity in the center of the coil is important because the coil is placed in such a way that target organ is positioned roughly in the center of the coil. In addition, the direction of rotation of the nuclear is radial direction of the coil (i.e. $x$-$y$ plane). Thus, it is necessary to generate rotate magnetic field in $x$-$y$ plane. For these reasons, this paper describes the results of magnetic field distributions in $x$-$y$ plane.

(1). Calculated magnetic field distribution

Figure 5-30(a) shows the magnetic field distribution of the proposed coil containing the head model in the $x$-$z$ plane at center of the coil ($z = 0$mm). In addition, the magnetic field distribution of the conventional coil containing the head model is shown in Fig. 5-30(b) for comparison. Here, the magnetic field levels of proposed coil and conventional coil were normalized with equal input power. From these results, it was confirmed that both coils generate uniform magnetic field inside the head model.

Figure 5-31 shows the profiles of magnetic field. The observation line is $x$ axis at $y = 0$ mm in Fig. 5-30. The continuous line is the result of proposed coil and the broken line is the result of conventional coil. As a result, it was confirmed that the proposed coil has an equivalent uniformity to the conventional birdcage coil. Furthermore, each component ($H_x$, $H_y$ and $H_z$) of magnetic field distributions inside the proposed coil containing the head model is shown in Fig. 5-32. From these results, we confirmed that levels of $H_z$ component, which is parallel to the axial direction of the coil, was small in comparison with the levels of $H_x$ and $H_y$ components. Thus, the spatial phase difference of 90 degrees caused by quadrature excitation between $H_x$ and $H_y$ components of
magnetic field distributions were observed. These distributions were observed for a similar result in conventional birdcage coil and it is non-unique characteristics of proposed coil. From these results, it is suggested that the proposed birdcage coil can use for MR imaging.

Additionally, in order to compare the magnetic field uniformity inside the head, average variation of the magnetic field inside a head model were calculated by applying the following equation [65]. The average variation is defined as

\[
\delta_{\text{ave}} = \frac{1}{SH_{\text{ave}}} \int_{S} |H - H_{\text{ave}}| \, ds
\]

where \( S \) denotes the circular region [m\(^2\)], and \( H_{\text{ave}} \) denotes average value of \( H \) over the region. In this case, \( S \) is cross section of the phantom. In addition, range of \( H_{\text{ave}} \) as defined from 0 to 1. Table 5-7 summarizes the average variations of magnetic field in the head model for the proposed coil and the conventional coil. The average variation of the proposed coil is approximately \( \delta_{\text{ave}} = 0.147 \), the average variation of the conventional coil is the \( \delta_{\text{ave}} = 0.158 \). As a result, average variation of the conventional coil is slightly larger than the average variation of the proposed coil. However, difference of the average variation between the conventional coil and proposed coil was negligible small in terms of practice. Accordingly, it is considered that the proposed birdcage coil has an equivalent homogeneity of magnetic field to the conventional birdcage coil.
Fig. 5-30  Magnetic field distribution.

Fig. 5-31  Profiles of magnetic field.
Fig. 5-32  Magnetic field distribution.

Table 5-7  Comparison of the average variation $\delta_{\text{ave}}$.

<table>
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<th>$\delta_{\text{ave}}$</th>
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<tbody>
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<td>Proposed coil</td>
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</tr>
<tr>
<td>Conventional coil</td>
<td>0.158</td>
</tr>
</tbody>
</table>
(2). Measurement of the magnetic field distribution

Here, the author describes the measurement of the magnetic field distribution inside the human head model and thereby validates the calculated results. Figure 5-33 shows the experimental setup, where the arrangement of the birdcage coil and human head model is the same as that in Fig. 5-27. Here, gel phantom is used as a human head model. Gel phantom is filled in dielectric pipe ($\varepsilon_r=2.1$) and thickness of dielectric pipe is 3mm. At 170 MHz, the differences between the target values of electrical properties of the gel phantom and the measured electrical constants are less or equal 10 % for relative permittivity $\varepsilon_r$ and conductivity $\sigma$, respectively. In addition, the signal was excited at the position $\phi=0^\circ$ in Fig. 5-27. In order to feed the proposed birdcage coil, a signal generator (8657B Hewlett Packard, Palo Alto, California) and a power amplifier (LA100UF-CE Kalmus, Bothell, Washington) were used. The shielded loop probe was used as the magnetic field probe. Diameter of a shielded loop probe is 5mm. Then, $H_x$ and $H_y$ were measured by orienting the probe antenna along the $x$-axis at $y = 0$ mm in
Fig. 5-30. Here, the $H_z$ component is omitted because its level is very low as compared to $H_x$ and $H_y$. A spectrum analyzer (Agilent E4403B) was used to measure the received signal level. The scanning interval on the line was 10 mm. Figure 5-34 compares the measured and calculated magnetic field distribution on the observation line ($y = z = 0$ mm, $-70 \leq x \leq 70$ mm). The levels are normalized with the maximum value on the observation lines in both cases. The continuous line is the calculated result and square symbols are the measured result. As a result, good agreement was observed between the measured and numerical calculated results of the magnetic field level in the range of $|x| < 50$ mm.

Fig. 5-34  The measured and calculated magnetic field on the observation line.
5.6 Summary in Chapter 5

This chapter performed the fundamental investigation in order to evaluate interactions between EM waves and human body using both numerical and experimental phantoms.

In section 5.2, the evaluation method for understanding the detail SAR distribution inside the phantom has been described. At present, there are two types of standard SAR measurement method. However, both methods can know only average SAR. Recently, the materials and shapes of implantable medical devices have become increasingly diverse, making it more likely that subjects’ implanted medical devices will affect MR imaging, potentially causing temperature increases in unexpected regions. Despite this, guidelines for evaluating the effects of such cases on the human body are still unclear. Therefore, it is necessary to comprehend the detailed SAR distribution within the human body. In this section, the thermographic method applied to the RF coil was investigated. The SAR evaluation model for birdcage coil was investigated. The validation of this model was confirmed in comparison with FDTD calculation model. As a result, tendency of the SAR distributions agreed well between the measurement and the calculation.

In section 5.3, SAR evaluation using realistic numerical phantom is introduced. In order to examine the effects of heterogeneity, SAR value of realistic model was compared with simplified homogeneous phantom. As a result, it was confirmed that the SAR value of the realistic model is higher than simplified model. Therefore, the realistic model is indispensable to accurate evaluate the SAR in human body.

In section 5.4, the evaluation method for understanding the magnetic field distribution inside the experimental phantom has been described. In order to measure
the magnitude of magnetic field inside the head phantom, gel phantom was used. This is because of gel phantom can insert magnetic field probe, easily. Evaluation of magnetic field distribution model was almost the same model as Section 5.2. The validation of this model was confirmed in comparison with FDTD calculation model. As a result, tendency of the magnetic field distributions agreed well between the measurement and the FDTD calculation.

In section 5.5, in order to evaluate an influence of the human body on the performance of the antenna nearby the human body, antenna design considering effect of human body is introduced. As one of the example, a birdcage coil for 4 T MRI systems with no lumped circuit elements was proposed and it was improved structure in previous study [74]. In addition, it is confirmed that the proposed birdcage coil has an equivalent homogeneity of magnetic field to the conventional birdcage coil.
6 Conclusions

In recent years, RF devices, which are usually placed in the vicinity of the human body, are widely used. In addition, these devices are used in various situations. The effect of the human body on the characteristics of antenna of devices which are placed nearby the human body has been widely investigated. For the future, the quantitative evaluation of the interaction between EM waves and the human body, which is radiated by the devices, is indispensable.

Until now, various investigations have been mainly made on the interaction between the mainly cellular phones and the human body in UHF band. For the future, it is necessary to evaluate the interaction between the EM waves and human body in other situations (e.g. frequency, devices, etc.). Especially, recently, the variety of antennas for HF-VHF bands application is widely investigated. Generally, the interactions should be estimated from numerical simulations and experimental evaluations. Until now, various phantoms have been proposed for evaluating the interaction of the human body and the EM waves. However, most of these studies have mainly focused on phantom for UHF band. In other words, recipes of phantom for HF-VHF bands are very few. Therefore, biological tissue-equivalent phantoms for HF and VHF bands are desired.

In chapter 2, dielectric theory and mechanisms are introduced at the beginning because human body and phantom are one of the dielectric substances. This chapter also explained the classification of the biological tissue-equivalent phantom.

In chapter 3, measurement methods of electrical properties in HF and VHF bands were investigated before developing phantoms. In the case of HF band, two-terminal method using the impedance analyzer was adapted. Additionally, source of
measurement error was investigated. In this dissertation, electrical properties by changing electrodes dimension and length of test sample were investigated. As a result, test sample which has longest diameter and electrode which has largest diameter were adopted because these were closest target value. Therefore, it is considered that two-terminal method is one of the effective methods for electrical properties measurement in HF band. In the case of VHF band, coaxial probe method was adapted. In order to confirm the validity of the measured result, measured electrical properties of deionized water was compared with target value of water. As a result, it is confirmed that measured result is agreement with the target value of water. Therefore, it is considered that coaxial probe method is one of the effective methods for electrical properties measurement in VHF band.

In chapter 4, the biological tissue-equivalent phantom was investigated. Firstly, electrical properties of base compound of the phantom were investigated. Next, the electrical properties of the phantoms for HF band were adjusted. In the case of high water content tissues, water was chosen as a base compound of the phantom. In addition, the electrical properties are adjustable by changing the amount of aluminum powder and sodium chloride. Meanwhile, we can confirmed that conductivity can be changed by aluminum powder and sodium chloride, respectively. In the case of low water content tissue such as a fat, silicone emulsion which has low relative permittivity is chosen as a base compound of the phantom. The electrical property of the fat phantom is adjustable by changing the amount of polyethylene powder and sodium chloride In addition, conductivity can be changed by polyethylene powder and sodium chloride, respectively. Moreover, fabicated multi-layered phantom for HF band was shown. Finally, phantom for VHF band is also investigated.
In chapter 5, in order to evaluate interactions between EM waves and human body using both numerical and experimental phantoms were performed. At first, experimental SAR evaluation was performed. In this dissertation, in order to estimate detailed SAR distribution, thermographic method was applied as a SAR measurement technique. SARs were estimated by using the actual RF coil. The validation of this model was confirmed in comparison with FDTD calculations model. As a result, tendency of the SAR and magnetic field distributions agreed well between the measurement and the calculation. Therefore, it is considered that thermographic method is useful to estimate detailed SAR distribution inside the phantom. Secondly, in order to examine the effects of heterogeneity, SAR evaluation using realistic numerical phantom by the FDTD calculation is performed. SAR value of realistic model was compared with simplified homogeneous phantom. As a result, it was confirmed that the SAR value of the realistic model is higher than simplified model. Therefore, the realistic model is indispensable to accurately evaluate the SAR in human body.

Next, the magnetic field measurement technique was estimated using the gel phantom. The validation was confirmed in comparison with FDTD calculations model. As a result, tendency of the magnetic field distributions agreed well between the measurement and the calculation. Therefore, gel phantom is useful to measure magnetic field inside the phantom. Finally, in order to evaluate an influence of the human body on the performance of the antenna nearby the human body, antenna design considering effect of human body is introduced. As one of the example, a birdcage coil for 4 T MRI systems with no lumped circuit elements was proposed and it was improved structure in previous study.

At last, from these results, it is considered that proposed phantoms for HF-VHF
bands are effective for evaluate the interaction in various situations. For example, proposed phantom is useful in antenna design considering effect of human body for body centric wireless communications or RFID and SAR evaluation for hyperthermina or MRI. Particularly, it is considered that multi-layered phantom is useful to accurately evaluate the interaction between EM waves and human body. In addition, the author will be glad if this study is any contribution the investigation of evaluation of the interaction between EM waves and the human body.
Publication list

Journal papers


Other paper

International conferences


References


Experimental Techniques,” P1528, 200X.


[28] A. Drossos, V. Santomaa, and N. Kuster, “The dependence of electromagnetic energy absorption upon human head tissue composition in the frequency range


[37] K. Ito, K. Furuya, Y. Okano, L. Hamada, “Development and the characteristics


[50] American Conference of Government Industrial Hygienists (ACGIH), Threshold limit values chemical substances and physical agents and biological exposure indices (Cincinnati OH), 1996.


[75] Chin, C. L., Collins, C. M., Li, S., Dardzinski, B. J., & Smith, M. B.

Appendix

FDTD method

The finite-difference time-domain (FDTD) method is one of the numerical calculation techniques, which solve the Maxwell's equations of differential from in the time domain. The FDTD method was proposed by K.S. Yee [51]. At present, the FDTD method is widely employed for solving the electromagnetic problems such as improvement of structure of the antenna, research on the interaction between the electromagnetic waves and the human body, and so on. The finite difference approximation of the FDTD method is written as follows [52]:

\[
E_x^n(i + \frac{1}{2}, j, k) = C_{ex}(i + \frac{1}{2}, j, k)E_x^{n-1}(i + \frac{1}{2}, j, k) \\
+ C_{exy}(i + \frac{1}{2}, j, k) \left( H_z^{n-\frac{1}{2}}(i + \frac{1}{2}, j + \frac{1}{2}, k) - H_z^{n-\frac{1}{2}}(i + \frac{1}{2}, j - \frac{1}{2}, k) \right) \tag{A.1-1}
\]

\[
- C_{exz}(i + \frac{1}{2}, j, k) \left( H_y^{n-\frac{1}{2}}(i, j + \frac{1}{2}, k) - H_y^{n-\frac{1}{2}}(i, j, k - \frac{1}{2}) \right)
\]

\[
E_y^n(i, j + \frac{1}{2}, k) = C_{oy}(i, j + \frac{1}{2}, k)E_y^{n-1}(i, j + \frac{1}{2}, k) \\
- C_{oyx}(i, j + \frac{1}{2}, k) \left( H_z^{n-\frac{1}{2}}(i + \frac{1}{2}, j + \frac{1}{2}, k) - H_z^{n-\frac{1}{2}}(i - \frac{1}{2}, j + \frac{1}{2}, k) \right) \tag{A.1-2}
\]

\[
+ C_{oyz}(i, j + \frac{1}{2}, k) \left( H_x^{n-\frac{1}{2}}(i, j, k + \frac{1}{2}) - H_x^{n-\frac{1}{2}}(i, j, k - \frac{1}{2}) \right)
\]
\[ E_z^n(i, j, k + \frac{1}{2}) = C_{ex}(i, j, k + \frac{1}{2}) E_{y}^{n-1}(i, j, k + \frac{1}{2}) \]
\[ - C_{exx}(i, j, k + \frac{1}{2}) \left\{ H_{y}^{n-\frac{1}{2}}(i + \frac{1}{2}, j, k + \frac{1}{2}) - H_{y}^{n-\frac{1}{2}}(i - \frac{1}{2}, j, k + \frac{1}{2}) \right\} \]
\[ + C_{exy}(i, j, k + \frac{1}{2}) \left\{ H_{x}^{n-\frac{1}{2}}(i, j + \frac{1}{2}, k + \frac{1}{2}) - H_{x}^{n-\frac{1}{2}}(i, j - \frac{1}{2}, k + \frac{1}{2}) \right\} \] (A.1-3)

where,
\[
C_{ex}(i + \frac{1}{2}, j, k) = \frac{\sigma(i + \frac{1}{2}, j, k) \Delta t}{1 - \frac{2 \sigma(i + \frac{1}{2}, j, k)}{\sigma(i + \frac{1}{2}, j, k) \Delta t}} \] (A.1-4)
\[
C_{exx}(i + \frac{1}{2}, j, k) = \frac{\Delta t / \sigma(i + \frac{1}{2}, j, k)}{1 + \frac{1}{\Delta z}} \] (A.1-5)
\[
C_{exy}(i + \frac{1}{2}, j, k) = \frac{\Delta t / \sigma(i + \frac{1}{2}, j, k)}{1 + \frac{1}{\Delta y}} \] (A.1-6)
\[ C_{\psi}(i, j + \frac{1}{2}, k) = \frac{\sigma(i, j + \frac{1}{2}, k) \Delta t}{1 - \frac{2\epsilon(i, j + \frac{1}{2}, k)}{\sigma(i, j + \frac{1}{2}, k) \Delta t}} \]  
\[ 1 + \frac{2\epsilon(i, j + \frac{1}{2}, k)}{\sigma(i, j + \frac{1}{2}, k) \Delta t} \]  
(A.1-7)

\[ C_{\psi_e}(i, j + \frac{1}{2}, k) = \frac{\Delta t / \epsilon(i, j + \frac{1}{2}, k)}{\sigma(i, j + \frac{1}{2}, k) \Delta t} \frac{1}{1 + \frac{2\epsilon(i, j + \frac{1}{2}, k)}{\sigma(i, j + \frac{1}{2}, k) \Delta t}} \]  
(A.1-8)

\[ C_{\psi_x}(i, j + \frac{1}{2}, k) = \frac{\Delta t / \epsilon(i, j + \frac{1}{2}, k)}{\sigma(i, j + \frac{1}{2}, k) \Delta t} \frac{1}{1 + \frac{2\epsilon(i, j + \frac{1}{2}, k)}{\sigma(i, j + \frac{1}{2}, k) \Delta t}} \]  
(A.1-9)

\[ C_{e_x}(i, j, k + \frac{1}{2}) = \frac{\sigma(i, j, k + \frac{1}{2}) \Delta t}{1 - \frac{2\epsilon(i, j, k + \frac{1}{2})}{\sigma(i, j, k + \frac{1}{2}) \Delta t}} \]  
\[ 1 + \frac{2\epsilon(i, j, k + \frac{1}{2})}{\sigma(i, j, k + \frac{1}{2}) \Delta t} \]  
(A.1-9)

\[ C_{e_x}(i, j, k + \frac{1}{2}) = \frac{\Delta t / \epsilon(i, j, k + \frac{1}{2})}{\sigma(i, j, k + \frac{1}{2}) \Delta t} \frac{1}{1 + \frac{2\epsilon(i, j, k + \frac{1}{2})}{\sigma(i, j, k + \frac{1}{2}) \Delta t}} \]  
(A.1-10)
\[ C_{xy}(i, j, k + \frac{1}{2}) = \frac{\Delta t / \varepsilon(i, j, k + \frac{1}{2})}{\sigma(i, j, k + \frac{1}{2}) \Delta y} \frac{1}{1 + \frac{\Delta t}{2 \varepsilon(i, j, k + \frac{1}{2})}} \]  

\[ H_x^{\frac{1}{2}}(i, j + \frac{1}{2}, k + \frac{1}{2}) = H_x^{\frac{1}{2}}(i, j + \frac{1}{2}, k + \frac{1}{2}) \]

\[ - C_{hxy}(i, j + \frac{1}{2}, k + \frac{1}{2}) \left\{ E_x^n(i, j, k + \frac{1}{2}) - E_x^n(i, j + 1, k + \frac{1}{2}) \right\} \]  

\[ + C_{hxy}(i, j + \frac{1}{2}, k + \frac{1}{2}) \left\{ E_y^n(i, j + \frac{1}{2}, k) - E_y^n(i, j, k + \frac{1}{2}) \right\} \]

\[ H_y^{\frac{1}{2}}(i + \frac{1}{2}, j, k + \frac{1}{2}) = H_y^{\frac{1}{2}}(i + \frac{1}{2}, j, k + \frac{1}{2}) \]

\[ - C_{hxy}(i + \frac{1}{2}, j + \frac{1}{2}, k + \frac{1}{2}) \left\{ E_y^n(i + \frac{1}{2}, j, k + 1) - E_y^n(i + \frac{1}{2}, j, k) \right\} \]  

\[ + C_{hxy}(i + \frac{1}{2}, j + \frac{1}{2}, k + \frac{1}{2}) \left\{ E_x^n(i + \frac{1}{2}, j, k) - E_x^n(i, j, k + \frac{1}{2}) \right\} \]

\[ H_z^{\frac{1}{2}}(i + \frac{1}{2}, j + \frac{1}{2}, k) = H_z^{\frac{1}{2}}(i + \frac{1}{2}, j + \frac{1}{2}, k) \]

\[ - C_{hxy}(i + \frac{1}{2}, j + \frac{1}{2}, k + \frac{1}{2}) \left\{ E_y^n(i + 1, j, k + \frac{1}{2}) - E_y^n(i + \frac{1}{2}, j, k) \right\} \]  

\[ + C_{hxy}(i + \frac{1}{2}, j + \frac{1}{2}, k + \frac{1}{2}) \left\{ E_x^n(i + \frac{1}{2}, j + 1, k) - E_x^n(i + \frac{1}{2}, j, k) \right\} \]
where,

\[ C_{x,y} (i, j + \frac{1}{2}, k + \frac{1}{2}) = \frac{\Delta t}{\mu(i, j + \frac{1}{2}^2, k + \frac{1}{2}) \Delta y} \]  (A.1-14)

\[ C_{x,z} (i, j + \frac{1}{2}, k + \frac{1}{2}) = \frac{\Delta t}{\mu(i, j + \frac{1}{2}^2, k + \frac{1}{2}) \Delta z} \]  (A.1-15)

\[ C_{y,z} (i + \frac{1}{2}, j, k + \frac{1}{2}) = \frac{\Delta t}{\mu(i + \frac{1}{2}, j^2, k + \frac{1}{2}) \Delta z} \]  (A.1-16)

\[ C_{y,x} (i + \frac{1}{2}, j, k + \frac{1}{2}) = \frac{\Delta t}{\mu(i + \frac{1}{2}, j^2, k + \frac{1}{2}) \Delta x} \]  (A.1-17)

\[ C_{z,x} (i + \frac{1}{2}, j, k + \frac{1}{2}) = \frac{\Delta t}{\mu(i + \frac{1}{2}, j^2, k + \frac{1}{2}) \Delta x} \]  (A.1-18)

\[ C_{z,y} (i + \frac{1}{2}, j, k + \frac{1}{2}) = \frac{\Delta t}{\mu(i + \frac{1}{2}, j^2, k + \frac{1}{2}) \Delta y} \]  (A.1-19)

Here, for a continuous function of space and time \( f(x, y, z, t) \), its discretized form at \( n \)th time step can be written as \( f^n_{ijk} = f(i\Delta x, j\Delta y, k\Delta z, n\Delta t) \), where \( \Delta x, \Delta y \) and \( \Delta z \) are cell size in the finite difference representation and \( \Delta t \) is the incremental time step.

For the explicit finite difference scheme to yield a stable solution, the time step used for the updating must be limited according to the Courant Friedrich Levy (CFL) criterion. For the FDTD formulation of Maxwell's equations on a staggered grid, this
criterion reads

\[ \Delta t \leq \frac{1}{c \sqrt{\frac{1}{\Delta x^2} + \frac{1}{\Delta y^2} + \frac{1}{\Delta z^2}}} \]  \hspace{1cm} (A.1-20)

where, \( c \) the speed of light. From eq. A.1-20 it is clear that the time step is directly related to the cell size. In addition, the time step must be chosen for the smallest cell in the mesh. The cell size therefore has a significant important on the computational requirements of a simulation [76].
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