Signal Path Loss Simulation of Human Arm for Galvanic Coupling Intra-body Communication

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Received: April 12, 2015; Accepted: January 29, 2016; Published: August 8, 2016

Abstract. Galvanic coupling intra-body communication involves the formation of a network between small terminals applied to the surface of the human body and bio-signal sensors embedded within the body. To enable the design of a communication device, it is important to fully understand the signal transmission loss characteristics of the human body, while developing a method that optimizes the transmission efficiency. This study analyzed the signal path loss during galvanic-coupling intra-body communication of a human arm through the application of a four-terminal circuit and a finite-element method (FEM) model, with special attention given to the return path. The effect of the interface circuit of an LC series-parallel circuit that injected the signal into the human body was also examined. Without the LC series-parallel circuit, the attenuation of the transmitted signal was minimized within a range of 2–5 MHz in the circuit model and 3–7 MHz in the FEM model. The addition of the LC series-parallel circuit improved the attenuation by 1.9–5.8 dB at the resonant frequency (2 MHz).

Keywords: Galvanic coupling, Intra-body communication, Circuit model, FEM model

1. Introduction

Intra-body communication (IBC) is a new signal transmission technology that uses the human body as a transmission medium. Studies on IBC began to attract attention after Zimmerman published the article "Personal area networks: Near-field intrabody communication" in 1996 [1]. Various studies have since been conducted, including those on IBC applications and its communication mechanisms [2]. IBC is thought to be one of the most promising communication technologies for constructing a body area network (BAN), and it is advantageous in terms of confidentiality and communication efficiency because the main transmission medium is the human body itself and its surrounding environment. In the first half of 2012, a new
BAN protocol (IEEE802.15.6) was formulated, and capacitive coupling IBC was standardized as its one system [3].

In general, there are two types of IBC, namely, capacitive coupling type and galvanic coupling type [4], as shown in Fig. 1. In both coupling types, the transmitter and receiver each use two electrodes. In capacitive coupling, one of the electrodes of the transceiver (signal electrode) is placed on the human body side, while the other electrode (ground electrode) is placed on the other side in a floating state (Fig. 1(a)). The human body is treated as a conductor. The electric field signal is induced in the human body from the signal electrode of the transmitter, and a portion of this induced signal is detected by the signal electrode of the receiver [1, 5]. Since the signal induced on the human body has a large component that returns to the ground electrode of the transmitter itself and a very large component that escapes to the earth ground, the electric field signal that arrives at the receiver is very small. The electric field component that escapes to the earth ground fluctuates depending on the neighborhood environment, and the signal transmission quality with this method is more affected by this external environment than by the human tissue. It is preferable to use a larger electrode in terms of the signal quality because the capacitive coupling between the electrodes of the transmitter and receiver becomes strong.

In galvanic coupling, which is the focus of this study, both electrodes of the transceiver are placed directly on the human body (Fig. 1(b)). An electric signal is differentially induced from the two electrodes of the transmitter and the attenuated signal that passes through the human body is differentially detected using the two electrodes of the receiver. In the galvanic coupling approach, the signal is confined within the human body. Therefore, it has been thought that the signal transmission quality with this method is less affected by the external environment. However, we believe that, in the galvanic coupling approach, a return path is needed, and in an actual communication state, it is realized by the electrical coupling between the ground electrodes of the transmitter and receiver (Fig. 1(b)).

Several groups have attempted to investigate the propagation mechanism in galvanic coupling IBC, and models based on equivalent electrical circuits [6, 7, 8], the finite-element method [9], and the theoretical electromagnetic principle [10] have been proposed. However, there are no models that consider the return path. In this paper, we propose two new models for galvanic coupling-type IBC that include the electrical coupling of the return path. The first model is a four-terminal circuit model based on a simplified equivalent circuit representation of the human arm. The second model is a 3D finite element method (FEM) model of the human arm. These models include circuit elements such as a transmitter and receiver, in addition to the body element. Using these models, we examined the signal transmission characteristics of galvanic coupling-type IBC, especially the effects of both the signal trans-
mission distance and electrical coupling amplitude. Moreover, we examined the effectiveness of an optimized interface circuit for signal transmission, which was proposed by Okamoto et al. [11].

The rest of this paper is organized as follows. Section 2 describes an optimized interface circuit that utilized an LC series–parallel circuit. Sections 3 and 4 describe the four-terminal circuit model and circuit-coupled FEM model of the human arm, respectively. Section 5 describes numerical simulations and in vivo measurements, followed by a discussion of the results in Section 6. Section 7 summarizes the conclusions of this paper.

![Figure 1: Signal transmission methods of IBC: (a) capacitive coupling type and (b) galvanic coupling type.](image)

### 2. Interface Circuit Utilizing LC Series–Parallel Circuit

In order to optimize the transmission efficiency, we employed an LC series–parallel circuit that injected a signal into the human body as an interface circuit of the transmitter. This LC circuit, which was proposed by Okamoto et al. [11], increases the signal voltage to be injected using electrical resonance. Figure 2 shows the LC series–parallel circuit and its equivalent circuit. A secondary coil of the transformer is connected in parallel with series-connected capacitors. A junction node between the capacitors in the LC series–parallel circuit is connected to the human arm through electrodes. When the capacitance of the capacitor components of the human body is much smaller than the capacitance of an external capacitor $C_2$, the resonant frequency $F_r$ is determined by the inductor $L$ and the external capacitors $C_1$ and $C_2$, and it is represented as follows:

$$F_r = \frac{1}{2\pi L} \frac{C_1 \cdot C_2}{C_1 + C_2} \quad (1)$$
In addition, the quality factor $Q$ depends on the conductivity of the human body. The LC series–parallel circuit has the additional functions of preventing electric shock and band limiting the transmitted signal.

![Figure 2: LC series–parallel circuit and its equivalent circuit: (a) LC series–parallel circuit connected to human arm and (b) equivalent circuit.](image)

### 3. Four-Terminal Circuit Model

#### 3.1. Model outline

In this study, the frequency range selected for the investigation of the transmission medium in the galvanic coupling IBC was 100 kHz to 10 MHz. This region can be modeled using static circuit models. The signal transmission of the galvanic coupling IBC can be described using the four-terminal circuit model shown in Fig. 3. In Fig. 3, $Z_i$, $Z_o$, $Z_{t1}$, $Z_{t2}$, $Z_{b1}$, and $Z_{b2}$ are impedances of the human arm, the values of which were determined by the application of the improved Song’s method [7]. The impedances of $Z_i$ and $Z_o$ are treated as the input and output impedances of the human arm, respectively. $Z_{t1}$ and $Z_{t2}$ are the transverse impedances of the transmission path, while $Z_{b1}$ and $Z_{b2}$ are the cross impedances of the transmission path. The coupling impedance between the electrode and skin is represented as $Z_c$. The value of $Z_c$ used in our simulations was determined based on the measured results of Wegmuller et al. [12].

The output resistor of the transmitter and the input impedance of the receiver are represented as $R_o$ and $Z_{cc}$, respectively. In this study, the effect of the interface circuit of the LC series–parallel circuit that injects the signal into the human body was also examined. $Z_h$, $Z_{c1}$, and $Z_{c2}$ are the impedances of the LC series–parallel circuit of the transmitter. When incorporating the interface circuit in the circuit model, the receiver was assumed to have an LC parallel tuned circuit. $Z_{res}$ is the impedance of this LC parallel circuit of the receiver. The current signal from the transmitter electrode flows through the human arm and reaches the receiver electrode. After that, a portion of it returns to the transmitter electrode through the parasitic return path. Therefore, the model includes the parasitic return path, the impedance of which is represented as $Z_{cc}$. 

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3.2. Formulation of input–output characteristic

The output voltage is calculated by applying the mesh current law to the circuit and solving the linear equations with seven unknown currents $i_1, i_2, ..., i_7$ (Fig. 3) [8]. The seven linear equations for a single frequency are shown in (2) in a matrix form.

$$
\begin{bmatrix}
V_i \\
0 \\
0 \\
0 \\
0 \\
0 \\
0 \\
0 \\
0
\end{bmatrix}
= 
\begin{bmatrix}
Ro + Zc1 + Zc2 + Zl & -Zc2 & 0 & 0 \\
0 & 2Zc + Zc2 + Zl & -Zi & -Zi \\
0 & 0 & Zb2 + Zi + Zt1 & Zb2 + Zi \\
0 & 0 & Zb2 + Zi & Zb1 + Zb2 + Zi + Zo \\
0 & 0 & 0 & 0 & -Zo \\
0 & 0 & -Zc & -Zb2 & -Zb2 - Zo \\
0 & 0 & 0 & -Zb2 & -Zb2 - Zo \\
Zc + Zc_{ceo} + Zo & -Zc_{ceo} & Zc + Zc_{ceo} + Zc_{reso} & Zb2 + Zo + Zt2 & i_1 \\
-Zc_{ceo} & Zb2 & Zb2 + Zo + Zt2 & i_2 \\
0 & 0 & 0 & 0 & i_3 \\
0 & -Zb2 & 0 & 0 & i_4 \\
0 & 0 & -Zb2 & 0 & i_5 \\
-Zo & -Zb2 & -Zb2 - Zo & 0 & i_6 \\
-Zb2 & 0 & Zb2 + Zo + Zt2 & 0 & i_7
\end{bmatrix}
$$

where $V_i$ is the output voltage of the transmitter, and $Z_{c_{ceo}}$ denotes a series connection between $Z_c$ and the parallel connection of $Z_{ceo}$ and $Z_{reso}$.

$$Z_{c_{ceo reso}} = Z_c + \frac{Z_{ceo}Z_{reso}}{Z_{ceo} + Z_{reso}}$$

The equations can also be represented in a matrix multiplication form,

$$V = ZI$$
where $\mathbf{V}$ is the voltage column vector, $\mathbf{I}$ is the mesh current column vector, and $\mathbf{Z}$ is the $7 \times 7$ impedance matrix. Therefore, the currents can be solved by multiplying the impedance inverse matrix by the voltage vector:

$$\mathbf{I} = \mathbf{Z}^{-1}\mathbf{V} \quad (5)$$

so that the output voltage $V_o$ can be calculated as

$$V_o = (i_5 - i_6) \times \frac{Z_{ceo}Z_{reso}}{Z_{ceo} + Z_{reso}} \quad (6)$$

The signal attenuation can be evaluated according to the following equation:

$$H_c = -20\log_{10}\left(\frac{V_o}{V_i}\right) \ \text{dB} \quad (7)$$

In addition, we evaluated the input voltage $V_h$ for the human arm using the next formula:

$$V_h = |(i_1 - i_2) \times Z_{c2}| \quad (8)$$

### 3.3. Modeling of human arm

We assumed that the human arm was composed of concentric layers of skin, fat, muscle, cortical bone, and bone marrow, in the same manner as Song’s method [7], as shown in Fig. 4(a). The thicknesses of these different layers for a human arm model with a radius of 50 mm are 1.5 (skin), 8.5 (fat), 27.5 (muscle), 6 (cortical bone), and 6.5 mm (radius, bone marrow). The dielectric properties of the human tissues can be expressed as the conductivity and relative permittivity. These values are determined using the website calculation tool Calculation of the Dielectric Properties of Body Tissues [13] and are shown in Table 1, where $\sigma$ is the conductivity, and $\varepsilon_r$ represents the relative permittivity.

#### 3.3.1 Impedances of $Z_t$ and $Z_b$

The transverse impedance $Z_t$ depends on the length of the signal transmission path, dielectric properties of the human tissues, and geometric parameters of the human body. It also depends on the electrode pair separation and electrode area. We assume that it is composed of the arm impedance $Z_{td}$, which needs to be modified according to the electrode pair separation, along with the skin impedance underneath the electrode $Z_s$. The arm impedance $Z_{td}$ can be expressed as the parallel connection of the impedances corresponding to the five layers

$$Z_{td} = \frac{L}{\sum_{l=1}^{5} \sigma_{lf} S_l + j \omega \varepsilon_0 \sum_{l=1}^{5} \varepsilon_{rlf} S_l} \quad (9)$$

where $L$ is the length of the signal transmission path. $S_l$ is the cross-sectional area of the $l$th layer. $\sigma_{lf}$ and $\varepsilon_{rlf}$ are the conductivity and relative permittivity corresponding to the different layers and signal frequencies, respectively. The skin impedance $Z_s$ is calculated as follows:
\[ Z_s = \frac{w_1}{(\sigma_{lf} + j\omega\varepsilon_0\varepsilon_{r_{lf}})\pi r^2} \]  

(10)

where \( w_1 \) and \( r \) are the thickness of the skin layer and radius of the circular electrode, respectively. Using (9) and (10), the transverse impedance \( Z_t \) can be calculated by

\[ Z_t = KZ_{td} + 2Z_s \]  

(11)

where \( K \) denotes a constant determined by the electrode pair separation. In this paper, \( K \) is assumed to be equal to six. The transverse impedance \( Z_t \) can be calculated in the same manner. The transmission path of the cross impedance is assumed to be \( L_b \), and the impedance corresponding to the arm is calculated as

\[ Z_{bd} = \frac{L_b}{\sum_{i=1}^{5} \sigma_{lf} S_i + j \omega\varepsilon_0 \sum_{i=1}^{5} \varepsilon_{r_{lf}} S_i} \]  

(12)

The cross impedance \( Z_b \) is therefore found as follows:

\[ Z_b = KZ_{bd} + 2Z_s \]  

(13)

3.3.2 Impedances of \( Z_i \) and \( Z_o \)

The current corresponding to \( Z_i \) and \( Z_o \) flows between the transmitting electrodes or receiving electrodes. Since the cortical bone and bone marrow are out of the current flow path, we assumed that their contribution to the flow path was very small. The current was assumed to flow only through the skin, fat, and muscle layers, which formed a parallel connection, as shown in Fig. 4(a). For simplicity, the shape of each layer is represented as plate-like (Fig. 4(b)). The values of \( Z_i \) and \( Z_o \) can be determined by using the following equation:

\[ Z_i = Z_o = \frac{L_e}{\sum_{i=1}^{5} \sigma_{lf} w_l (2r) + j \omega\varepsilon_0 \sum_{i=1}^{5} \varepsilon_{r_{lf}} w_l (2r)} \]  

(14)

where \( L_e \) and \( w_l \) are the length of the electrode pair separation and thickness of the \( l \)th layer, respectively.

Figure 4: Cross section of human arm model at electrode location and shapes of skin, fat, and muscle layers for calculating values of \( Z_i \) and \( Z_o \).
4. Circuit-Coupled FEM Model

4.1 Model outline

We developed a multi-layer FEM model of the human arm, which incorporates several circuit elements and the parasitic return path, using COMSOL 4.3b [14]. The human arm is modeled as a cylinder with a 50 mm radius and a length of 500 mm. It consists of five concentric and homogeneous layers relevant to the signal transmission of the galvanic coupling IBC: skin, fat, muscle, cortical bone, and bone marrow, positioned as shown in Fig. 5(a). The thicknesses of these are the same as those of the circuit model. Different tissue layers are characterized

<table>
<thead>
<tr>
<th>Freq. (MHz)</th>
<th>Skin (Wet)</th>
<th>Fat (Wet)</th>
<th>Muscle (Wet)</th>
<th>Cortical Bone</th>
<th>Bone Marrow</th>
</tr>
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<tr>
<td>0.1</td>
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<td>2.4E-02</td>
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<table>
<thead>
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<th>Freq. (MHz)</th>
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<th>Fat</th>
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<th>Cortical Bone</th>
<th>Bone Marrow</th>
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<td>3.7E+01</td>
</tr>
</tbody>
</table>

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by the conductivity and relative permittivity, and these parameters were set utilizing the values listed in Table 1. The electrodes are modeled as circular plates with a conductivity of $60 \times 10^6$ S/m and relative permittivity of one and are placed on the skin. The input–output element terms of the circuit are modeled as LCR circuits. The parasitic return path is usually considered to be a capacitive path between the ground electrodes of the transmitter and receiver. In this study, a capacitor was added between the ground electrodes of the circuit-coupled FEM model. The terms $R_0$, $Z_1$, $Z_{c1}$, $Z_{c2}$, $Z_{ces}$, and $Z_{cc}$ are related to the above-mentioned elements and parasitic return path. They are the same as those of the circuit model. The coupling impedance between the electrode and skin is modeled using the contact impedance, which is one of the functions of COMSOL 4.3b, not as the circuit element $Z_c$.

4.2 Numerical implementation

When the wavelength of the electromagnetic field in biological tissues is much larger than the tissue dimensions, the inductive effect and wave propagation can be neglected [15]. Therefore, the electric and magnetic fields are decoupled [9]. For our application, only the electric field is of interest. Maxwell’s equations can be simplified using the continuity equation and constitutive relations, leading to the following equation for the quasi-static electric field:

$$-\nabla \cdot \{(\sigma + j\omega\varepsilon_{0}\varepsilon_r)\nabla v\} = 0 \quad (15)$$

where $\omega$, $\varepsilon_0$, and $v$ are the field frequency, vacuum permittivity, and electric potential, respectively. It is known that the electric field $E$ can be expressed in terms of the electric potential $v$ as follows:

$$E = -\nabla v \quad (16)$$

and the current density distribution $J$ is given by

$$J = \sigma E \quad (17)$$

On the skin surface, a zero current flux is considered, and the Neumann boundary condition is employed (18), where $\mathbf{n}$ is the surface normal. On all the interior layer boundaries, the continuity of the current flux is applied (19):

$$\mathbf{n} \cdot J = 0 \quad (18)$$

$$\mathbf{n} \cdot (J_1 - J_2) = 0 \quad (19)$$

The geometry of the human arm cylinder was meshed using tetrahedral and triangular elements (Fig. 5(b)). A finer discretization was used around the electrodes. The mesh included around 60,000 elements.
Figure 5: Circuit-coupled FEM model of galvanic coupling: (a) simplified cylinder model of human arm with five layers, including circuit elements, and (b) FEM mesh of cylinder.

Figure 6(a) shows a typical example of the surface voltage distribution. The signal electrode (red disk) of the transmitter applies a signal voltage to the human arm, while its ground electrode acts as a reference with zero potential. The potential of the ground electrode of the receiver was low because it was connected to that of the transmitter through the capacitive path. Figure 6(b) and (c) shows the current density distributions in an axial and a transversal cut. It can be seen that the current in the transversal direction passes through the cross-section close to the electrodes, not through the entire cross-section uniformly; the current flowing between the electrodes of the transmitter passes relatively close to the surface of the human tissue.
Figure 6: Typical examples of surface voltage distribution and current density distribution in axial and transversal cuts. The simulation conditions included a frequency of 2 MHz and the connection of the resonant circuits.
5. Simulation and measurement Results

In this simulation, the influence to the signal attenuation on the signal transmission distance and parasitic return path amplitude was examined. The effectiveness of the interface circuit of the LC series–parallel circuit was also verified. Table 2 lists the remaining parameters used in both models for this simulation.

Table 2: Simulation Papameters. The underlined values are defaults.

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<th>Name</th>
<th>Value</th>
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</tr>
<tr>
<td>Electrode radius</td>
<td>5</td>
</tr>
<tr>
<td>Electrode interval</td>
<td>30</td>
</tr>
<tr>
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</tr>
<tr>
<td></td>
<td>30 [pF]</td>
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<tr>
<td>Contact impedance (FEM)</td>
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<tr>
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<td>382 [nF/m²]</td>
</tr>
<tr>
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</tr>
<tr>
<td>Input impedance of the receiver Zceo</td>
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</tr>
<tr>
<td></td>
<td>13 [pF]</td>
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<tr>
<td>Parasitic return path Zcc</td>
<td>100, 200, 300 [pF]</td>
</tr>
<tr>
<td>Impedances of the LC series–parallel Zl</td>
<td>100 [μH]</td>
</tr>
<tr>
<td>Impedances of the LC series–parallel Zc1</td>
<td>81(circuit)/105(FEM) [pF]</td>
</tr>
<tr>
<td>Impedances of the LC series–parallel Zc2</td>
<td>81(circuit)/105(FEM) [pF]</td>
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<tr>
<td></td>
<td>100 [μH]</td>
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</table>

Figure 7 shows the signal attenuation values when the signal transmission distance was 10 and 30 cm. As shown in Fig. 7(a), the signal attenuation of the four-terminal circuit model decreased as the signal frequency increased from 100 kHz to 5 MHz, and then increased slightly as the signal frequency increased from 5 to 10 MHz. As depicted in Fig. 7(b), the signal attenuation of the circuit-coupled FEM model had a similar tendency, although the
peak frequency was around 3 MHz. Moreover, in both results, an increase in the transmission distance from 10 to 30 cm caused an increase in the signal attenuation in a range greater than around 1 MHz. Fig. 8 shows the signal attenuation values when the capacitance of the parasitic return path was 100, 200, and 300 pF. The signal attenuation of both models increased, especially in the lower frequency range, as the capacitance decreased, and the peak frequency at which the signal attenuation was minimum moved to a higher frequency. Figure 9 shows the frequency characteristic of the signal attenuation and the input voltage into the human arm with and without the resonant circuits. As shown in Fig. 9(a) and (b), the signal attenuation was the smallest at the resonant frequency of 2 MHz, and the signal attenuation values were $-11.8$ dB in the circuit model and $-6.4$ dB in the FEM model. Without the resonant circuits, the values of the signal attenuation at 2 MHz were $-13.7$ dB in the circuit model and $-12.2$ dB in the FEM model, as shown in Fig. 7. The attenuation was improved by 1.9 and 5.8 dB, respectively. In addition, the input voltage applied to the human arm became 1.1- and 1.7-times higher, respectively, as depicted in Fig. 9(c).

(a)                                    (b)

Figure 7: Frequency characteristics of signal attenuation when signal transmission distance is 10 and 30 cm: (a) circuit model and (b) FEM model.

(a)                                    (b)

Figure 8: Frequency characteristics of signal attenuation when capacitance of parasitic return path is 100, 200, and 300 pF: (a) circuit model and (b) FEM model.
In order to validate the simulation results, we measured signal path loss of the human arm. Figure 10 shows the measurement setup that consists of a battery-powered notebook PC, a DDS function generator, and an Anritsu MS8608A spectrum analyzer (input impedance: 50Ω). The electrode distances were 10 or 30 cm, and the electrode pair separation was set to 3 cm. The function generator provides sinusoidal signals with a power of 3 dBm (50 Ω load). The frequency range selected for the investigation of signal attenuation characteristics was 100 kHz to 10 MHz, and the resonant frequency of the LC series-parallel circuit was set at 2 MHz. The signal path loss was measured in two directions: from arm to wrist, and from wrist to arm. Fig. 11 shows the signal attenuation values measured with and without resonant circuits. In both directions, the signal attenuation measured without the resonant circuits was smaller in the frequency band of several megahertz. Further, the signal transmission distance does not seem to largely affect the signal path loss. The signal attenuation measured with the resonant circuits was smallest at the resonant frequency of 2 MHz. The attenuation was improved approximately by 2 to 4 dB.
6. Discussion

As shown in Figs. 7 and 8, although the peak frequencies differed to some extent, the results with the two models had similar tendencies: the signal attenuation in the range greater than 1 MHz was increased by as much as 1 dB when the signal transmission distance was increased. In addition, the signal attenuation was relatively greater, by approximately 9 dB, especially in the lower frequency range, when the capacitance of the parasitic return path was decreased. These results indicate that the influence of the parasitic return path was greater than that of the signal transmission distance. It is said that galvanic-coupling IBC is resistant to variations in the outer environment because the majority of the transferred signal is thought to flow inside the human body. However, in reality, it seems to be affected by the environment. Thus, it is necessary to design a communication device that is resistant to changes in the environment around the human body. As for the signal frequency employed, in view of the attenuation, we think it would better to use a value of several megahertz or more.
As shown in Fig. 9, although the values of the signal attenuation and input voltage at 2 MHz differed to some extent, the results with the two models also had similar tendencies: the signal attenuation was at its minimum at 2 MHz, and it increased dramatically at a position away from 2 MHz, with the maximum input signal found at 2 MHz. Adding the resonant interface circuits into the transceiver and receiver circuits led to 1.1- and 1.7-fold increases in the input voltage, and a signal attenuation improvement of 1.9–5.8 dB was confirmed in this simulation. If the resonant frequency could be set up well, since an unnecessary frequency component would also be simultaneously removed, it is thought that this would be an effective method. In this simulation, the resonant frequency was set at 2 MHz. If inductor $L$ is 100 $\mu$H and external capacitor $C_1$ equals $C_2$, then $C_1$ and $C_2$ are ideally calculated to be 128 pF using (1). The simulation was first conducted using these values, but the resonant frequency did not become 2 MHz, but was below 2 MHz. Therefore, using trial and error, we reset the values to 81 pF in the circuit model and 105 pF in the FEM model (TABLE 2), respectively, in order to reach 2 MHz. The discrepancy was thought to be due to the capacitor component of the human body. When the capacitor component of the human body is much smaller than the capacitance of capacitor $C_2$, equation (1) can be more effectively applied [11]. By taking the above into account, we want to examine the appropriate setting of the resonance frequency in detail in the future.

A similar tendency can be observed in both simulation and measurement results. Without the resonant circuits, the simulated and measured signal path loss was smaller in the frequency band of several megahertz, and it was not influenced by the signal transmission distance to a large extent. With the resonant circuits, the signal path loss was the smallest at the resonant frequency, and the attenuation was improved by several decibels. The external capacitors $C_1$ and $C_2$ used in the measurement were set at 66 pF. These results suggest that the two models have relatively high precision for the design of galvanic coupling IBC. In a subsequent study, we are planning to measure the signal path loss for several conditions, and estimate the values of the parasitic return path.

Other studies using a four-terminal circuit model calculated the human arm impedances [6, 7, 8]. The impedances of the human arm in our four-terminal circuit model were determined using equations (9)–(14), which were derived by modifying Song’s method [7]. The constant $K$ in equations (11) and (13) was set at six because the current in the transversal direction passes mainly through approximately one-sixth of the cross-sectional area, as shown in Fig. 6(b). In addition, as shown in Fig. 6(c), since the current between the transmitting electrodes or receiving electrodes flows mainly through the skin, fat, and muscle layers along a path the width of the electrode, we calculated the impedances $Z_i$ and $Z_o$ by approximating a three-layer plate with a width equal to the electrode diameter and a length equal to the ele-
trode interval. The absolute values of impedances $Z_t, Z_b$, and $Z_i = Z_o$ were 350, 360, and 375 $\Omega$, respectively, when the electrode distance was 10 cm, and the signal frequency was 2 MHz.

It is thought that capacitive coupling exists in the combination of all electrodes. However, because ground electrodes are connected to the ground elements in the transceivers, whose areas are normally quite large relative to that of the electrodes, it is postulated that the capacitive coupling between the ground electrodes is much larger than those of other combinations. This study therefore considers only the capacitive coupling between the ground electrodes of the transceivers. Precedence for this approach has been established by R. Xu et al [16]. Also, the capacitance value of the return path was set to be relatively large because not only the ground electrodes but also the ground elements of the transceivers were considered.

7. Conclusions

This study investigated the signal path loss of galvanic coupling intra-body communication in the frequency range of 100 kHz to 10 MHz using two models of a human arm, with special attention given to the return path and an optimization interface circuit. One model was a four-terminal circuit model, and we proposed an improved method of calculating the impedances of the human tissues. The other model was an FEM model, and it was characterized by being mixed with circuit elements. An LC series–parallel circuit was adopted as the optimization interface. A very close agreement between the results of the two models was obtained, which is believed to mean that the two models have relatively high precision for the design of galvanic coupling IBC. The simulation results revealed that it is necessary to take into account the capacitance of the parasitic return path when designing IBC transceivers, and an optimization interface circuit is effective at improving the signal path loss and denoising.

References


http://niremf.ifac.cnr.it/tissprop/.

