# A Multi-Cast Communication Scheme Using Weak Electrical Current for Intra-Body Networks


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## ABSTRACT

Implantable medical devices have paved the way for realizing intra-body networks (IBNs) that are capable of communicating information from within the body. Traditional forms of RF-based based communication face drawbacks in terms of high absorption within tissues and can potentially result in security and privacy issues, owing to the omnidirectional radiation. Instead, we propose the use of weak electrical current as a means of intra-body communication for implants within the human body and to a relay and gateway node located on the surface of the skin that facilitates remote patient monitoring. The main contributions of this paper are: (i) signal reflection and refraction analysis of electromagnetic waves through human tissue boundaries, and (ii) the design of a combined multi-hop and multi-cast communication scheme for communication between implants. Our results reveal that a multi-cast communication scheme can be achieved for IBNs with the appropriate selection of transmission parameters.



## Categories and Subject Descriptors

C.2.1 [**Computer-Communication Networks**]: Network Architecture and Design – *network communications, network topology, wireless communications.*

## General Terms

Performance, Design, Theory.

## Keywords

Intra-body communication, intra-body area networks, multi-cast communication, multi-hop communication, weak electrical current.


## 1. INTRODUCTION

Intra-body networks (IBNs) promise advanced medical procedures that involve controlled drug injection; continuous implant based medical monitoring, and personalized medicine [5]. The above mentioned applications can be accomplished by establishing a communication channel between sensors in the form of micro-scale implants (responsible for gathering biological data), and an external entity used to monitor and analyze incoming data.

The ability to process patient data in real time will result in not only offline evaluation of patient medical information, but also enable immediate emergency action based on irregular changes in a patient's body. In the current state of the art work, common methods for building IBN communication links rely on RF-based approaches in the form of Narrowband and Ultra-wideband (UWB) Communication, and the use of Ultrasonic waves to inter-connect wireless implants. However, traditional forms of RF-based wireless communications find limited use for such an application based on the following reasons: (i) limited penetration in human tissue (high attenuation), (ii) increased risk of tissue heating through signal absorption, and (iii) an increased need for secure transfer of patient-physician information [1]. Ultrasound as a means of intra-body communication also faces drawbacks in the form of: (i) severe multi-path fading and (ii) high delay (caused by slow propagation speeds), which in turn, may require the need for more complex system designs [2].

To mitigate the above-mentioned issues, extensive work has been done in [6] to model the human body as a communication channel under the influence of weak electrical signals to facilitate the creation of IBNs. This research involved the study of the propagation characteristics of using weak electrical signals within the human body to send data among sensors through a recently explored method known as Galvanic Coupling. This form of communication utilizes a pair of electrodes to directly couple weak electrical current (~ 1 mA) into the body, staying well below the permissible limit in the human body. Notable contributions of this method of communication are: the ability to lower energy consumption by several orders of magnitude and the provision of better internal security between communicating entities, when compared to traditional forms of RF-based communication [6]. The results of this work signify that the use of galvanic coupled electrical signals is a promising approach for future IBNs.

In this paper, we envision a similar network of implanted devices that communicate within the human body, between sensors (via weak electrical signals), and to the surface of the human body to a relay and gateway node, capable of sending data to remote monitoring personnel using over the air RF. We answer the following important questions in this paper: Is it possible for a single implant embedded within a tissue layer to simultaneously transmit data to a peer node within the same layer, as well as to an upstream node in a layer immediately above? We show that it is indeed possible, and we create the first multi-cast network for galvanic coupled IBNs, where selected nodes can receive data, based on application nodes. We achieve this form of communication by intelligently coupling the signal via the electrodes on the source sensor through appropriate selection of angle of transmission, frequency and amplitude. Our approach takes into account the electrical properties of the various tissue layers, and the signal reflection and refraction that occur, owing to these properties at the tissue boundaries.

In summary, our contributions include: (i) an analysis of signal reflection and refraction of electromagnetic waves through human tissue boundaries, and (ii) the design of a combined multi-hop and multi-cast communication scheme for communication between implants. The human body is modeled as a lossy dielectric medium (tissue conductivity > 0) with four tissue layers (bone, muscle, fat and skin). Using this model, we also investigate whether a single transmission that crosses multiple tissue boundaries (e.g., from bone to skin) at a given frequency performs better in comparison to multi-hop forwarding of data and the tissue-specific selection of transmission parameters that can enable multi-cast communication.

## 2. SIGNAL ANALYSIS

When electromagnetic (EM) waves move through a lossy dielectric medium, they are partially reflected, partially transmitted, and even refracted. Reflection, transmission and refraction of a signal within human tissue vary depending on the type of tissue, as each layer exhibits different conduction properties. For communication between implants, the portion of the signal that crosses from one tissue boundary to the next (with the least reflection), has an impact on the signal strength at the receiver, and consequently, on the bit error rate (BER). Thus, studying the reflection and refraction properties used in physics and electrodynamics allow us to analytically verify the reflection levels, and how much the signal bends (refracts) as it crosses from one tissue layer to another.

The overall aim of this study in the context of a multi-cast scheme is as follows (also depicted in Figure 1): We will use the signal transmission from the source node (with refraction incorporated) to reach the next hop implant (e.g., $s_1$) in the subsequent tissue layer. The reflected signal will be used to reach another implant node (e.g., $s_2$) in the same tissue layer. Thus, the same source can transmit data to both $s_1$ and $s_2$ at the same time. Note that the actual propagation path of the signal is highly directional, and precisely guiding the signal to the locations of $s_1$ and $s_2$ is essential for successful multi-cast transmissions.

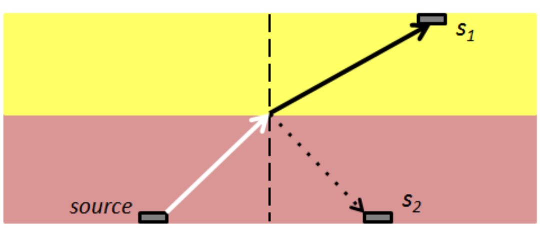


**Figure 1. Example diagram of a multi-cast communication scheme.**

### 2.1 Amplitude Reflection Coefficient

The amplitude reflection coefficient (1) for normally incident signals depends heavily upon the angle of incidence of the wave, tissue conductivity, and relative permittivity and permeability of the tissue. Let the variables $n_1$ and $n_2$ represent the refractive index (the speed of light in vacuum divided by the speed of light in the medium) for the current tissue layer (e.g., muscle) and immediately adjacent tissue layer (e.g., fat). The refractive index describes how much the signal bends when entering a certain medium. For example, a medium with a refractive index of 1.5, equates to the speed of light traveling 1.5 times slower in that particular medium [4]. In a lossy dielectric, the index of refraction relates to the tissue conductivity $(\sigma)$, relative permeability $(\mu_r)$ and complex relative permittivity $(\epsilon_r^*)$ shown in (2), but simplifies to (3), since human tissue is considered a non-magnetic medium (unit value of relative permeability).

$$\Gamma_{\mathrm{N}} = -\left(\frac{n_2 - n_1}{n_2 + n_1}\right) \tag{1}$$

$$n = \sqrt{\epsilon_r^* \mu_r} \tag{2}$$

$$n = \sqrt{\epsilon_r^*}, \;\; \epsilon_r^* = \epsilon_r - \frac{j\sigma}{\omega}, \tag{3}$$

Figure 2 (a) exemplifies a scenario for calculating the values for the amplitude reflection coefficient for normally incident waves. The mediums of interest are muscle $(n_1)$ and fat $(n_2)$, and the incident wave travels from within the muscle to the fat tissue layer, a higher refractive index to a lower refractive index. Note that for normally incident waves, the angle of reflection ($\theta_r$ – dotted black arrow), refraction ($\theta_R -$ solid black arrow), and incidence($\theta_I$), are all equal to the value of zero degrees.

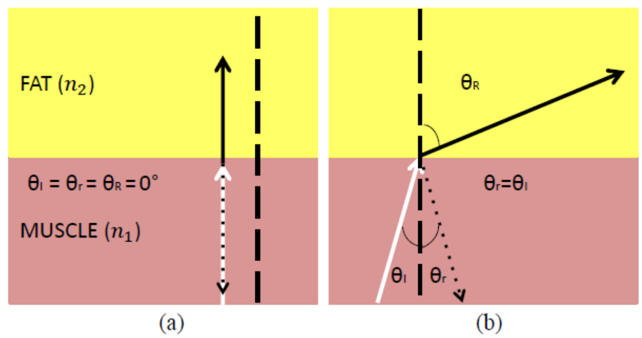


**Figure 2. Tissue boundary diagram for (a) normally incident reflection and refraction and (b) obliquely incident reflection and refraction.**

Figure 3 depicts the trend of the absolute value of the amplitude reflection coefficient for various combinations of tissue layer boundaries, as frequency increases within the desired range (100 kHz to 1 MHz) for the use of weak electrical current for intra-body communication [3]. The curves represented in figure 3 are obtained by taking the absolute value of the normally incident reflection coefficient $|\Gamma_N|$, found in (1). The frequency dependent nature of the dielectric properties of human tissues has a direct effect on the amount of reflection that occurs at each level of human tissue boundaries. Furthermore, results from this graph aid in the selection of the best operating frequency to conduct communication within the human body. It can be observed that for some situations, it may be more advantageous to use higher or lower frequencies, where reflection is less, for signals propagating from one layer to another. For example, the frequency of 100 kHz in the fat-to-skin boundary yields more desirable levels of reflection, and the frequency of 1 MHz is preferred for the bone-to-muscle boundary.

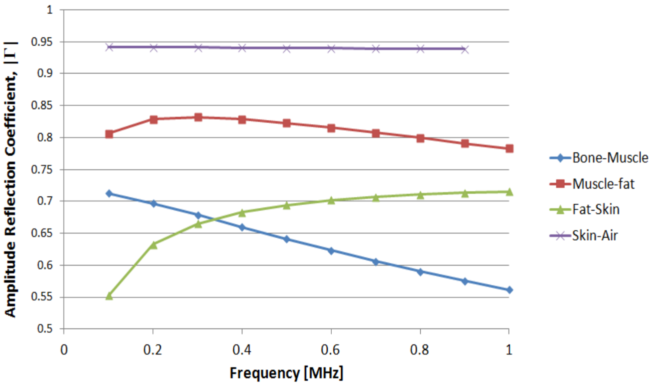


**Figure 3. Normally incident reflection coefficient vs frequency.**

When electromagnetic (EM) waves interact in a lossy dielectric medium at an oblique angle, polarization occurs, splitting the electric field and magnetic field into separate components. These components are distinguished and given their names based on their orientation in relation to the plain of incidence. The p-wave has its electric field (E-field) oriented parallel (p) to the plane of incidence (i.e., the incident, refracted and reflected signals lie in the plane of incidence), and the s-wave has its E-field perpendicular (s) to the plane of incidence [3]. The result of this behavior of EM waves yields two unique reflection coefficient values for each type of polarization. Equation (4), which shows the parallel polarization reflection coefficient, differs from (1), by showing a dependence on the angle of incidence and angle of refraction.

$$\Gamma_{\boldsymbol{p}} = \left(\frac{\boldsymbol{n_1}\cos\theta_{\mathrm{R}} - \boldsymbol{n_2}\cos\theta_{\mathrm{I}}}{\boldsymbol{n_1}\cos\theta_{\mathrm{R}} + \boldsymbol{n_2}\cos\theta_{\mathrm{I}}}\right) \tag{4}$$

The angle of reflection, as governed by the Law of Reflection, will be equal to the angle of incidence, at the interface normal to the surface where reflection takes place. For the sake of this work, all the analysis done with respect obliquely incident EM wave reflection will assume parallel polarization, and adhere to the laws governing the angle of reflection.

However, there exists a special case for parallel polarization, in which no reflection occurs, and the signal is entirely transmitted from one tissue layer to the other. This transpires when the angle of incidence is equal to what is known as the Brewster Angle. For angles of incidence that approach the Brewster angle in lossy dielectrics, the Brewster Angle loses its complete meaning, but the reflection coefficient values are still significantly less in comparison to other choices for the angle of incidence [6]. The Brewster angle for each of the tissue boundary reflection scenarios can be shown by the sudden drop in reflection for a particular value of incidence angle.

Figure 4 shows an example of the behavior of the reflection coefficient for obliquely incident waves for two values of the frequency range of interest, 100 kHz and 1 MHz, for varying values of the angle of incidence. By rewriting (5), we obtain the values of $\theta_R$ for each frequency dependent refractive index and for fixed values of $\theta_I$. This information allows us to plot the absolute values of obliquely incident reflection coefficients $|\Gamma_p|$, and observe the behavior their behavior with respect to different values of the angle of incidence. In figure 4, we observe the effects of the Critical angle (if $n_2 < n_1$, incidence angles exceeding the critical angle will be subject to the phenomenon of total internal reflection) and the Brewster angle taking place. For the boundary conditions of skin-to-air and muscle-to-fat, we observe total internal reflection, yielding a reflection coefficient of 1 for significantly small angles of incidence.

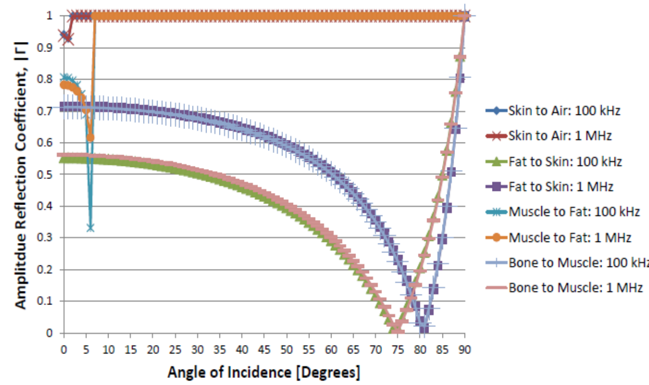


**Figure 4. Obliquely incident reflection coefficient vs angle of incidence.**

## 2.2 Angle of Refraction

Refraction occurs in a lossy dielectric medium under the influence of EM waves due to the change in speed a wave undergoes when moving from one medium to another. An increase in speed tends to make the signal refract away from the line normal to the surface of the two mediums (Figure 2 (b), where the angle of refraction is greater than the angle of incidence), while a decrease tends to cause bending towards the normal (angle of incidence greater than angle of refraction). The angle of refraction can once more be solved from (5), where $\boldsymbol{\theta_R}$ and $\boldsymbol{\theta_I}$ represent the angle of refraction and angle incidence respectively.

$$n_1 \sin\theta_{\mathrm{I}} = n_2 \sin\theta_{\mathrm{R}} \tag{5}$$

Figure 5 (a-d) plots the angle of refraction versus the operating frequency, for various values of the angle of incidence, subject to the critical angle. The results from these figures help us determine the proper receiver placement for the implant at the next hop tissue layer, while still taking into account possible levels of reflection.

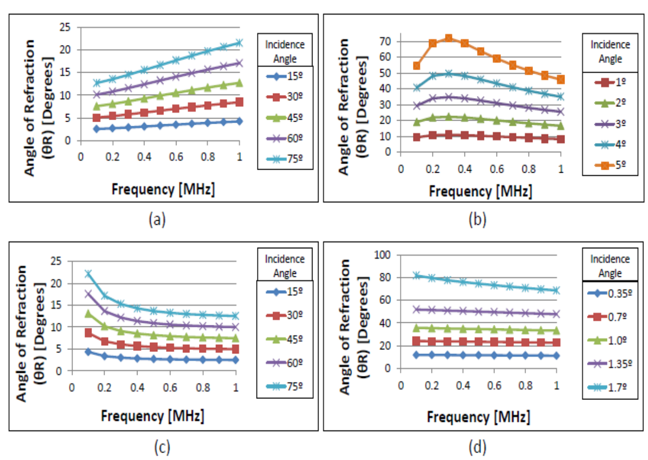


**Figure 5. Angle of refraction vs frequency**
**(a) bone-muscle (b) muscle-fat (c) fat-skin (d) skin-air.**

## 3. MULTI-CAST AND MULTI-HOP COMMUNICATION DESIGN

After careful study of the behavior of EM wave propagation in human tissues modeled as lossy dielectrics, we develop a strategy for the placement of next hop nodes and enable multi-cast communication. Combining the two approaches can yield a scenario where one can communicate from the inner most tissue layer, to a relay node on the skin surface, with a minimum number of hops, by exploiting the refraction and reflection properties of each human tissue layer.

By choosing the appropriate transmission power and frequency, part of the signal can permeate through the tissue boundary to an intended receiver (e.g., from muscle to fat), while the controlled reflected component returns back to a second receiver implant (e.g., also embedded in the muscle). Therefore, two different implants can potentially receive the same transmitted data, though at the loss of signal amplitude. Using the phenomenon of refraction, a sensor can be optimally aligned to receive the dominant component of the signal propagating into the direction of the next hop implant.

Although it is possible to send one signal from the bone layer to propagate to the relay node on the skin surface, several parameters have to be taken into account to ensure successful reception, with a reasonable BER. One must not choose a particular angle of incidence that could influence unwanted levels of reflection, and at the same time, yield an angle of refraction that could possibly cause total internal reflection at a subsequent tissue layer. Thus, a good tradeoff must be achieved between the angle of incidence, angle of refraction and the reflection coefficient values, in order to design a good multi-cast and multi-hop communication scheme.

One particular multi-cast case is presented in Figure 6, where the evaluation of reflection and refraction is provided for an operating frequency of 300 kHz, on a block of tissue 400 mm in length. The name designation for each tissue boundary is the following: Skin-Air (SA), Fat-Skin (FS), Muscle-Fat (MF) and Bone-Muscle (BM). First, the initial hop from bone to muscle is initiated by node *t*. The source of the multi-cast, *s*, then forwards the information to $s_1$ (both in muscle) through reflection, and to $s_2$ (in skin), through refraction. The reflected signal transmission resulting from communication between *s* and $s_2$, sends data along another path, to $s_3$ (in fat). Finally, the implant $s_2$ becomes the final hop source and transmits to the relay node *r* placed on the skin surface.

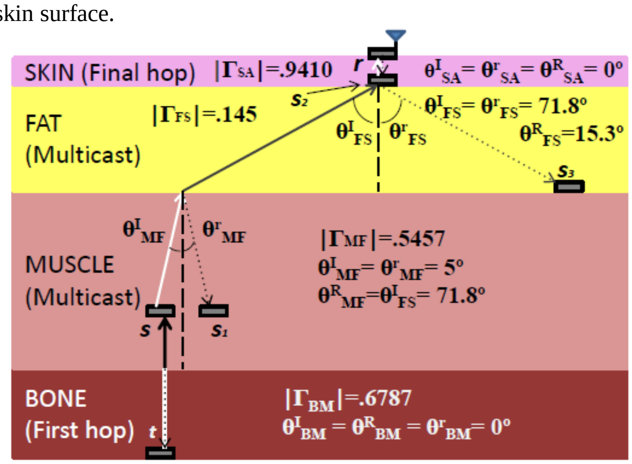


**Figure 6. Multi-cast and multi-hop communication scheme.**

## 4. PERFORMANCE EVALUATION

We show multi-cast communication for a maximum of 1 to 3 nodes, and a maximum of two hops (Bone-Muscle & Skin-Air). The specific electrical properties of the various tissue layers and the dimensions used in the calculations are shown in Table 1. The presence of large values of reflection (subject to the other parameters mentioned) limit the use of sending one signal from the bone layer to traverse through all the tissue layers, and to the relay on the surface. Even though larger angles of incidence in the bone layer can cause fewer reflections, they all lead to total internal reflection taking place within the muscle layer, when approaching the fat tissue layer boundary; thus, validating our approach for the first hop of data forwarding.

**Table 1. Electrical properties and dimensions of tissue layers**

| Tissue Layer (thickness - mm) | Conductivity (S/m) | Relative Permittivity (F/m) | Refractive Index |
|---|---|---|---|
| Skin (1) | 0.001934 | 1090.5 | 33.02 |
| Fat (9) | 0.02469 | 44.075 | 6.64 |
| Muscle (25) | 0.40693 | 5226.9 | 72.3 |
| Bone (20) | 0.021407 | 191.44 | 13.84 |

Taking advantage of the dielectric properties of the muscle, we can perform a multi-cast transmission with 1 node (within the muscle), that enables communication with 3 other nodes (1 also within the muscle, 1 within the fat and 1 within the skin). Table 2 summarizes the angles of incidence, reflection (equal to angle of incidence), refraction, as well as linear distances of the implants $s_1$, $s_2$, $s_3$, and *r* from the source *s*. This pathway serves as the intermediate flow of data between the first and last hop. Lastly, if we attempt to use the angle of incidence for the Skin-Air interface as the value of the angle of refraction from the Fat-Skin interface, then total internal reflection occurs, and the relay cannot be reached. Therefore, a separate and final hop must exist, forwarding the data along a normally incident path to the on-skin relay node.

**Table 2. Summary of sensor location and angles of incidence and refraction**

| Implant | Angle of Incidence/ Reflection (degrees) | Angle of Refraction (degrees) | Linear Distance from [Source, Relay] (mm) |
|---|---|---|---|
| S | 5 | 71.8 | [0, 37.77] |
| S1 | 5 | N/A | [2.63, 35.84] |
| S2 | 0 | 0 | [37.12, 1] |
| S3 | 71.8 | N/A | [58.04, 29.14] |

This information can be used to make informed decisions on the placement of source nodes and the selection of their transmission parameters, to effectively reach critical sensor points identified by medical personnel.

## 5. CONCLUSION

In this paper, a multi-cast and multi-hop communication scheme for intra-body communication is designed. Modeling the human body as a lossy dielectric medium, we proposed a method of communicating from deep within human tissue (e.g., bone) to a relay and gateway node located on the surface of the skin. In doing so, we have identified that the best multi-hop scenario exists within the muscle layer of human tissue and that next hop implants need to be positioned between layers that exhibit large amounts of reflection at tissue boundaries. The studies presented in this paper will be used in future work on examining beam forming techniques, the use of smart antennas, and state of the art algorithms for antenna switching in the context of IBNs.

## 6. ACKNOWLEDGEMENTS

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