

Chapter 2

Literature Review

The term “Body Area Network” (BAN) was first introduced by Van Dam et al. [36] in 2001. BAN technology envisions miniaturized sensors worn [37] or implanted on the body, continuously monitoring health parameters and acting to prevent the onset of critical health events. For instance, diabetics currently have access to an automatic insulin pump which monitors glucose levels and administers insulin when glucose levels are high. Similar technologies are also leading toward the construction of devices which can minimize incidences of heart attack or stroke, thus reducing hospital visits and saving costs for both the individual patient and a nation’s healthcare system. According to a recent report from Parks Associates, the U.S. market for wireless home-based healthcare applications and services are expanding with an annual growth rate of over 180 % and becoming a \$4.4 billion industry in 2013 [38]. Such statistics indicate a rising demand for portable health monitoring devices, e.g. BANs, which are currently undergoing tremendous research and development.

This chapter aims to compile the latest research on a new form of wireless communications in BAN which is fast gaining attention. IBC is a novel non-RF wireless data communication technique which uses the human body itself as transmission medium for electrical signals. The inhibition of communication to the person’s proximity in IBC users prevents the energy from being dissipated into the surrounding environment, resulting in potentially lower power consumption. Research has shown that IBC is capable of low transmission power below 1 mW and data rates of more than 100 kbps [20] which makes this approach potentially appealing as a short range communications alternative.

This chapter is organized as follows: Sect. 2.1 briefly introduces the IBC technique which is primarily based on capacitive or galvanic coupling. Section 2.2 presents the effects of human body tissues dielectric properties on signal transmission through the body. The various human body communication channel modeling methods and their characteristics are reviewed in Sect. 2.3. The final section discusses the latest state of the art transceiver designs and implements.

2.1 IBC Specifications and Methods

In general, IBC can be classified into two basic coupling types (i.e. how the electrical signals are transmitted): capacitive coupling (near-electric field) and the galvanic coupling (Waveguide). Figure 2.1 illustrates schematically the two different types of IBC coupling. For both coupling types both transmitter and receiver needs two pair of electrodes each. In capacitive coupling only one of the electrodes, i.e. signal electrode, of the transmitter side and receiver side is attached to the body while the other electrode (ground electrode) is floating. In the galvanic coupling method both electrodes of transmitter and receiver side are attached to the human body. The theory of capacitive coupled IBC is established based on the capacitive coupling of the human body to its surrounding environment. The signal is generated between the body channel transceiver by making a current loop through the external ground. The signal electrode of the transmitter induces the electric field in to the human body. The induced electrical signal is controlled by an electrical potential and the body acts as a conductor with the ground as the return path. Data propagation by capacitive coupling through the human body was first proposed by Zimmerman as means of communications in a personal area network (PAN) [39].

On the other hand, galvanic coupling is achieved by coupling alternating current into the human body. It is controlled by an AC current flow and the body is considered as a transmission line (waveguide). In the galvanic coupled IBC an electrical signal is applied differentially between the two electrodes of the transmitter. Major propagation of the signal occurs between the two transmitter electrodes and a largely attenuated signal is received by the two receiver electrodes. Figure 2.2 shows pathways for a current flow between transmitter and receiver electrodes in the galvanic method. The small current also results in a differential signal between the electrodes

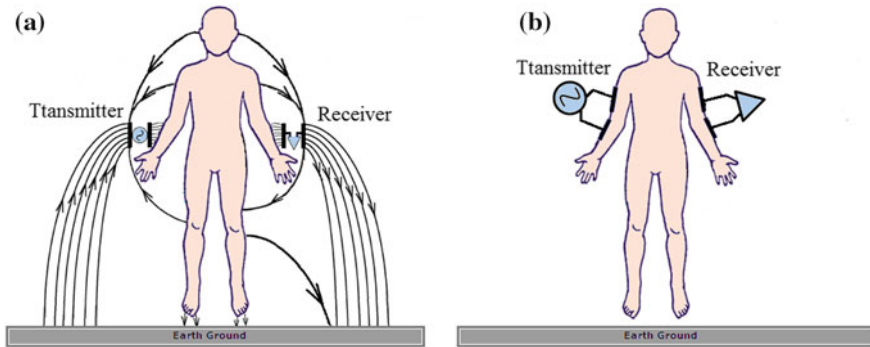


Fig. 2.1 **a** Capacitive coupled IBC. Lines of electric field surrounding the human body are shown in the figure. **b** Galvanic coupled IBC method

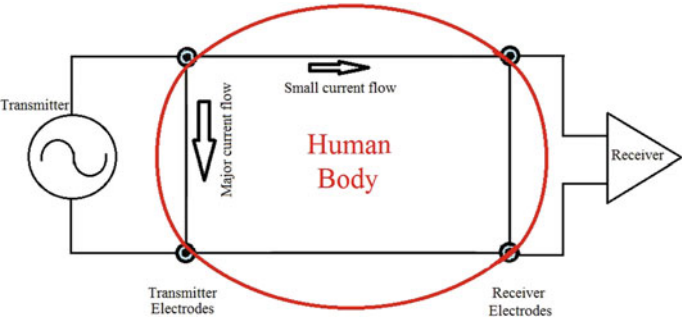


Fig. 2.2 Current flow establishes between transmitter and receiver electrodes in the galvanic method. The major current travels in short inter-electrode paths in both transmitter and receiver sides

Table 2.1 Comparison between characteristics of capacitive and galvanic coupling IBC methods

Capacitive coupling (Electric field)	Galvanic coupling (Waveguide)
The induced signal is controlled by an electrical potential (Applying static charged electrode)	The induced signal is controlled by a current flow (alternating currents over multiple electrodes)
Only signal electrodes of the transmitter and the receiver are attached to the body, while both ground electrodes float	A pair of transmitter and receiver electrodes is attached to the body
Ground is required as a reference	Ground is not required as a reference
The dominant signal transmission pathway is the environment [31]	The dominant signal transmission pathway is the body tissue
Higher transmission data rate and channel gain (Higher operation frequency compared to galvanic coupling) [21]	Lower transmission data rate [9]
The human body is modeled as a volume conductor (body is approximated as a single node [41])	The body is modeled as a waveguide for signal conduction [42]
Signal quality influenced by the environment around the body	Signal quality influenced by dielectric properties of human tissue
Interference from surrounding devices that could capacitively couple directly to the IBC device	Sensitive to the body location because of the dependence on inter-electrode distance and orientation along the body
Does not require direct contact with the human body. Needs only to be in the proximity	Direct contact with body tissue is necessary. Capable of both on-body and in-body (implanted) device communication [43]

of the receiver. In general, the ion content in the human body is the carrier of information in the galvanic coupling method. The principles of galvanic coupling IBC was first introduced by Oberle [40] while designing a single chip low power biomedical system. Table 2.1 briefly compares capacitive and galvanic coupling methods.

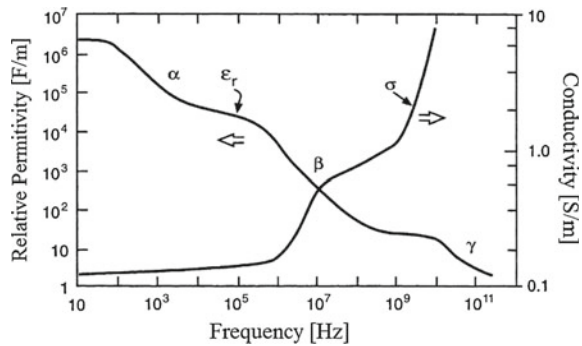
2.2 Electrical Properties of the Human Tissues

Propagation of galvanically or capacitively coupled signals through the human body in IBC is largely governed by human tissue electrical properties. The two major properties are relative permittivity (ϵ_r) and electrical conductivity (σ). The relative permittivity and electrical conductivity of a material are, respectively, the dipole and current densities induced in response to an applied electric field of unit amplitude [44]. Tissue types, the operation frequency range, temperature, intactness of cellular membranes, and tissue water content are some of the major factors which determine the tissue electrical properties in the human body.

The most comprehensive overview on human body electrical properties is presented by Gabriel et al. in 1996 [1]. Experiments were performed on human and animal tissue within the frequency range of 10 Hz to 20 GHz. During the experiments the temperature was fixed (37 °C) and it was assumed that tissue layers were homogenous. The electrical properties of a living tissue were measured through the interaction between electromagnetic radiation and tissue cells. Additional research revealed that dielectric properties of living tissue vary differently with frequency dispersion. Frequency dispersion mechanism was first introduced by Schwan [45] to characterize the electrical properties of biomaterials. The dispersion refers to the behavior of tissues at various frequency ranges; low frequency ranges, RF ranges, and gigahertz frequency ranges which are respectively referred to as Alpha, Beta, and Gamma dispersion. Figure 2.3 shows the dispersion of relative permittivity and specific conductivity of human body tissue. The major characteristics of biological tissues can be summarized as follows:

- Permittivity strongly declines, whereas conductivity increases within these frequency dispersions.
- The polarization of water molecules creates the gamma (γ) distribution in the gigahertz region (microwave frequencies). The gamma dispersion is not strong and it has minimal effect on the electrical properties of body tissues which carry protein bound water.

Fig. 2.3 Variation of human tissues electrical properties, relative permittivity and conductivity, against frequency. Three dispersion areas of α , β , and γ are characterized in the figure [48]



- The polarization of cellular membranes is an obstacle for an ion to flow in, or out of the cell and leads to the beta (β) dispersion. This region lies within hundreds of kilohertz to ten megahertz range. The polarization of protein is another contributing factor to the beta dispersion trend.
- The transport of ions across a biological membrane is related to the low frequency alpha (α) dispersion. The alpha dispersion can be found in frequency range between 1 Hz up to 100 kHz. An increase in tissue conductivity is rarely evident in the alpha dispersion and the permittivity shows a significant decrease [45].

The electrical properties of human tissue are a key element (feature) for designing an energy efficient, low noise, and cost effective IBC transceiver system achieved through the modeling of the human body transmission channel characteristics.

2.3 Modeling Methods of Body Tissues

The human body is modeled as a communication channel to investigate the propagation behavior of galvanically or capacitively coupled signals, and hence predict the transmitted data quality. Transmission characteristics of the body have been examined via modeling human body tissues. Although, there is encouraging progress in human body modeling, large discrepancies still exist between empirical results and model predictions. There are different methods for human body channel modeling, including electric equivalent circuit models which are based on parametric model of human tissues, numerical simulations such as finite element models (FEM) and finite difference time domain (FDTD) models. Transmission propagation models have been frequently used to guide RF transceiver designs [46], and tissue-specific models are expected to be applied similarly in the IBC field to future IBC transceiver design.

2.3.1 Human Tissues Parametric Model

To establish a proper parametric model of tissue properties, dielectric alteration of these properties is determined as a function of frequency. The *Cole–Cole* equation [47] presents the change of dielectric properties of a tissue over a broad frequency range:

$$\epsilon^*(\omega) = \epsilon_\infty + \frac{\Delta\epsilon_n}{1 + (j\omega\tau_n)^{(1-\alpha_n)}} \quad (2.1)$$

where ϵ^* is the complex dielectric constant, $\Delta\epsilon_n$ is the magnitude of the dispersion which is calculated from the difference between permittivity at static (ϵ_s) and infinite frequency (ϵ_∞), ω is the angular frequency, τ is the relaxation time constant which depends on physical processes such as ion effects, and α_n is distribution parameter

which is between 0 and 1. The decrease in the three separate dispersion areas, i.e. alpha, beta, and gamma, is determined by summation of the frequency dependent permittivity expressed by:

$$\varepsilon^*(\omega) = \varepsilon_\infty + \sum_n \frac{\Delta\varepsilon_n}{1 + (j\omega\tau_n)^{(1-\alpha_n)}} + \frac{\sigma_i}{j\omega\varepsilon_0} \quad (2.2)$$

where σ_i is static ionic conductivity. The dielectric performance of biological tissue is predicted by this summation through proper parameter selection for each tissue. The complex conductivity and the complex specific impedance of tissue are calculated by:

$$\sigma^* = j\omega\varepsilon_0\varepsilon^*, \quad z^* = \frac{1}{\sigma^*} \quad (2.3)$$

The relative permittivity and electrical conductivity of wet skin, fat, muscle, bone cortical, and bone marrow as a function of frequency is shown in Fig. 2.4. The simulation results are plotted based on Eq. (2.2) by [1].

Electrical properties of human body tissues can be modeled by equivalent electrical components such as resistors and capacitors. There are two types of circuit models:

1. RC elements connected in series could be employed to model single limbs and limb linkages [48].
2. A more common method is presenting the circuit model by considering a *constant phase element* (CPE) with a complex valued impedance given by $Z_{CPE}^* = A(j\omega)^{-n}$ where A is a constant and $n = \alpha$. This CPE impedance reduces to a simple resistance for $n = 0$ and to a capacitance reactance for $n = 1$. The physical meaning of the CPE is not clearly understood [49]. The model representation of resistive and capacitive elements appears to explain the empirical measurements well. However, determining the exact tissue components responsible for these properties is complicated by the non-homogeneous nature of tissue and randomly distributed cells sizes.

The circuit model of Eq. (2.2), when $\alpha = 0$, is depicted by a parallel combination of an ideal capacitor, a resistor and a CPE which is a series combination of a frequency-dependent capacitance and resistance [50] in Fig. 2.5. In fact, the *Cole–Cole* model is mainly applied to biological materials while other distributions like *Cole–Davidson* and *Havrila–Negami* are used for non-biological materials [48].

Electrical current flow through human tissue follows several pathways including intra-cellular, extra-cellular, and cellular membrane pathways. The high frequency current tends to pass easily through the tissues (higher conductivity as seen in Fig. 2.4). The intra-cellular fluid (ICF) is a liquid that circulates inside the cell and surrounded by cell membrane as well as extra-cellular fluid (ECF). The two lines in Fig. 2.6 show the low frequency (LF) current and high frequency (HF) current pathways [51]. Extra-cellular resistor and capacitor is modeled by R_e and C_e . Constituents of cell membrane and intra-cellular contribution are indicated by R_m and

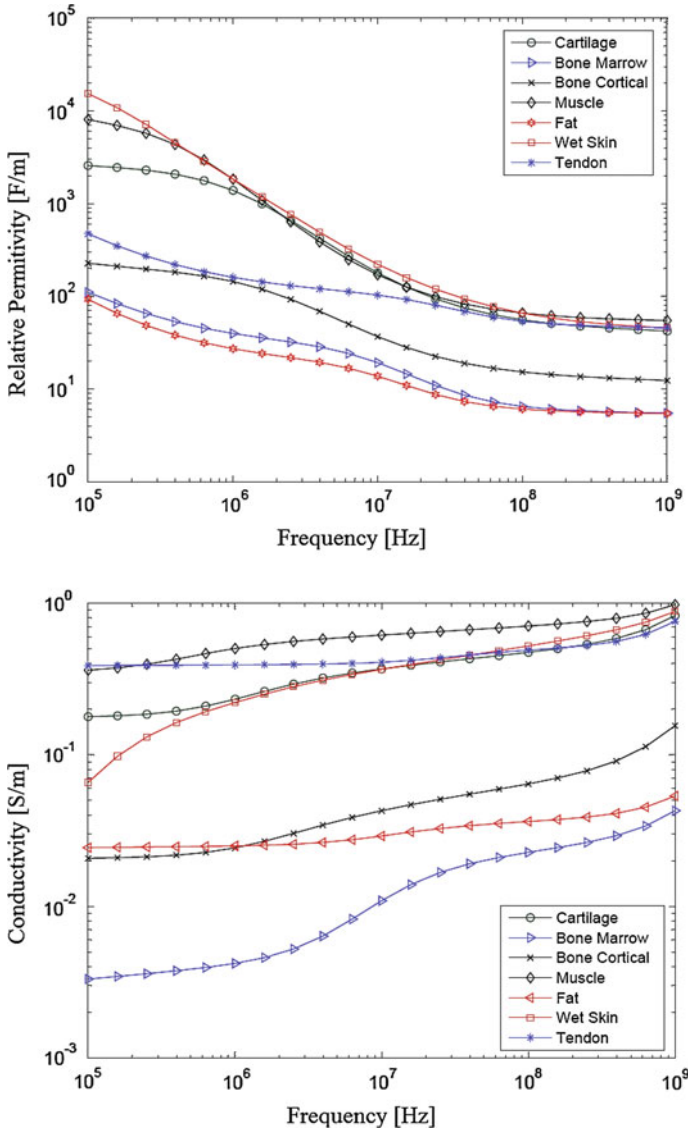


Fig. 2.4 Relative permittivity and conductivity of human body tissues at different frequencies are plotted based on Gabriel et al. research findings

C_m as well as R_i and C_i respectively. Kanai et al. in 1987 [52] has presented the complex and simplified circuit model of human tissues seen in Fig. 2.7. The resistors and capacitors in the proposed model represent physiological effects including blood circulation, metabolism of tissues, and electrolytic concentration of intra- and extra-cellular fluids within the body [52].

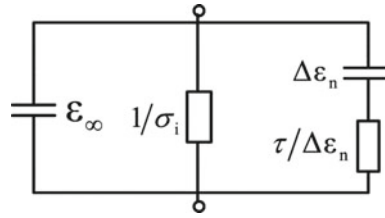


Fig. 2.5 Cole-Cole equation equivalent circuit model for a single time constant

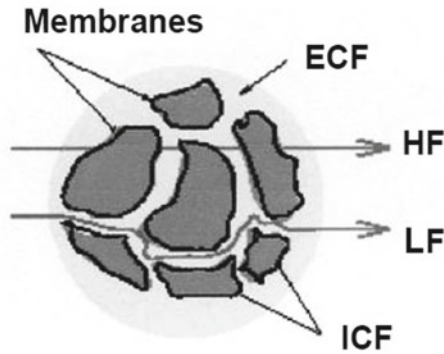


Fig. 2.6 Current flow in human body tissues [50]

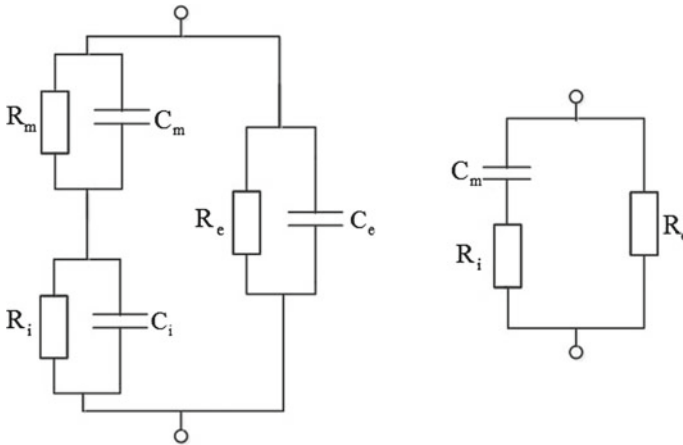


Fig. 2.7 Complex (*left*) and simplified (*right*) equivalent circuit model of human tissues. The equivalent components of extra-cellular, intra-cellular, and cell membrane are respectively specified by the indexes *e*, *i*, and *m* [51]

2.3.2 Body Channel Circuit Model

Obtaining the transfer function of the IBC system is the first step for construction of the circuit model. Zimmerman [39] proposed the first circuit model of body communication channel. The model consists of four significant transversal and longitudinal impedances in between transmitter and receiver electrodes. The inter-electrode impedances between the electrodes of the transmitter and receiver were ignored in impedance calculations of his model.

Capacitive coupling frequency characteristics of the human body are identified through a developed RC model in [41]. The body is considered as a single node due to the large impedance of the return path. However, this approximation is not true at high frequency. In this model, human body parts are presented as three cylindrical models which are divided into RC unit blocks each. To measure the R and C quantities the values from Gabriel's research [1] and Zimmerman's circuit model [39] were used. Xu et al. [46] proposed a capacitive coupling channel circuit in which they took the body shielding effects into account. Recent examinations [53] also presented skin propagation circuit models obtained from the electro physiological properties of skin. However, the authors did not verify the proposed model through empirical measurements. The comparisons of the model were confirmed only through achieved outcomes from other research data which were performed under different experimental conditions.

In the galvanic coupling circuit model, tissue impedance can be described by the *Cole–Cole* equation. Hachisuka et al. [54] designed an electric circuit model of galvanic coupled IBC for the first time. They also presented a new two terminal (electrodes) circuit model, where only two of the four electrodes were attached to the body (capacitive coupling). In [54], they suggested four terminals circuit model and used six impedances between transmitter and receiver electrodes. The results indicated that the two terminal electrode structures had a gain of 20 dB greater than the four terminal electrode structures. Four terminals circuit model with six body tissue impedances and four electrode-skin coupling impedances (Z_{ES}) were proposed by Wegmueller et al. [9] (see Fig. 2.8). The contact conditions between the electrodes and human body represented by the impedance (Z_{ES}) influence signal coupling and attenuation as well as the transceiver power consumption [55]. The value of Z_{ES} is represented by a series of three impedances consisting of electrode impedance, interface impedance (gel impedance), and skin layer impedances which include epidermis and subdermal impedances. According to Besio et al. [56], good contact between electrode and skin surface is established when $Z_{ES} < 10 \text{ k}\Omega$. The output resistance of the transmitter and the input resistance of the receiver were considered by Song et al. in a four terminal circuit model [57]. In the mentioned models, the geometry of the human body is approximated as a homogeneous solid volume. More detailed components such as joints were not considered in the circuit model. A limitation of this technique is that model complexity increases dramatically with the number of body tissue layers.

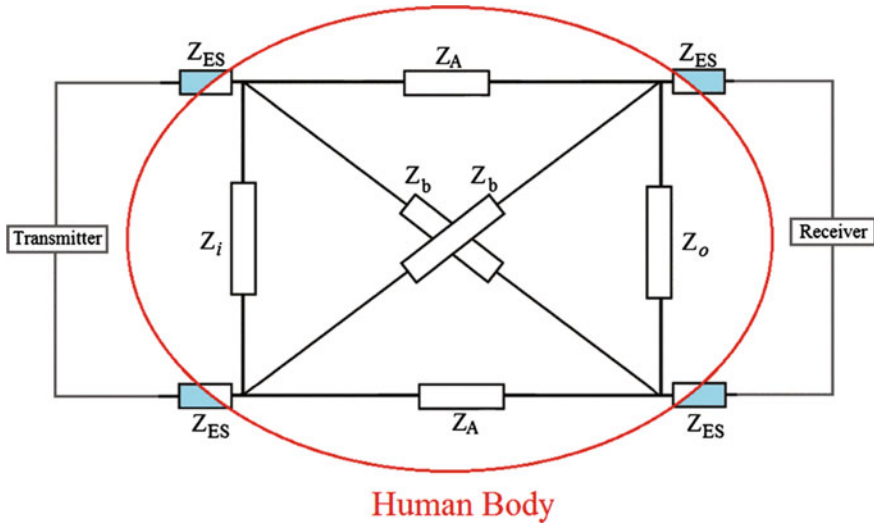


Fig. 2.8 Four terminals (electrodes) human arm circuit model during galvanic coupling IBC technique

2.3.3 Finite Element Model (FEM)

FEM is a technique which could model individual body tissues. It is based on the numerical solutions of partial differential equations and integrals. Electrical behavior of the human body is simulated through a physical model using this technique. The FEM is intended for better investigation of human anatomy effects on signal transmission in the IBC method. It is also able to reconstruct the potential distributions caused by induced current into the human tissue.

Xu et al. [58] have utilized FEM to investigate capacitive coupled IBC for the first time. The environment around the human body was separated into three different regions: near-field region, transmission region, and far-field region. Arm, chest, abdomen, and leg formed the four parts of the model. The results of FEM simulation showed that the presence of capacitive return path in capacitive coupled IBC plays a pivotal role in determining the characteristics of the body communication channel such as channel loss. It was also indicated that the return path is mainly coupled through the external ground during the capacitive IBC method.

The signal attenuation through the body in galvanic coupling IBC was investigated using FEM [59]. Authors explored the bioimpedance, electric field, current density, and the influence of channel length and the inter-electrode distance on signal propagation through the body in this work. They employed the default meshing option provided by COMSOL Multiphysics to model the arm geometry of the human body. While the model simulation results showed that an increase in the channel length led to higher signal attenuation, increasing the inter-electrode distance had a con-

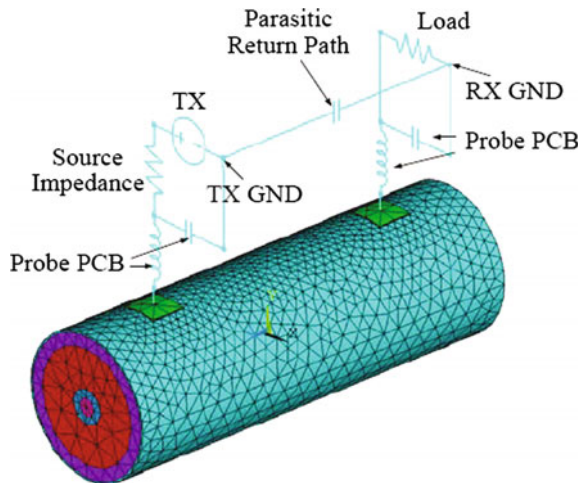
siderable decrement of about 1.0 dB/cm in attenuation. In general, there was good agreement between simulation and empirical results over the investigated frequency range of 1.0 k–100 MHz.

However, a major drawback with this method is the large size and the complexity of the human body which can significantly reduce the simulation accuracy [57]. Among different methods for channel modeling, the equivalent circuit model contains a few number of elements. However, the FEM is created from arbitrary number of nodes (more than thousands) to construct only one mesh. According to Wegmueller [6], to model the body channel using the FEM, mesh sizes were between 150000 and 200000 elements. Therefore, employing the circuit model could reduce the complexity as well as the analysis duration.

2.3.4 Circuit-Coupled FEM Model

Circuit-coupled FEM provides useful insights into the capacitive coupled IBC. It allows modeling of IBC channel components with different abstraction levels. The human forearm was modeled by a multi-layer FEM and the parasitic effects of probe PCBs (printed circuit board) were modeled by LC circuits (Fig. 2.9, [58]). Since frequency-dependent features of the parasitic return path make it difficult to simulate, the authors used a simplified capacitor model for the parasitic return path [58]. The circuit-coupled FEM revealed that large distance between transmitter and receiver electrodes led to higher signal attenuation. The variation of distance between transmitter and receiver was thought to be mainly caused by the size of the parasitic capacitor return path.

Fig. 2.9 Human forearm model is simulated using the circuit-coupled FEM method. The parasitic return path is represented by mean of a capacitor element [57]



2.3.5 *FDTD Model*

Finite difference time domain (FDTD) model is a computational modeling technique used in the field of electromagnetism. It examines the distribution of the electric field inside and outside the complex geometries such as human body. Since the model is merely operating in the presence of electromagnetic fields in the simulation area, it is considered as the state-of-the-art method for investigation of signal behavior in IBC technology.

The arm was modeled by FDTD calculation model through the capacitive coupling method by Fujii et al. [60]. Authors believed that a simple homogeneous calculation model was sufficient for the capacitive coupled IBC. It was concluded that the ground electrode in the transmitter side was necessary to strengthen the electric field around the limb. However, FDTD is a time consuming process for constructing a model in the lower frequency range and more suitable for higher frequency ranges (several hundred megahertz [61]) which is outside our current range of interest.

2.3.6 *Theoretical Electromagnetic Model*

Theoretical models of human body channels can be developed by solving Maxwell's equations and specific boundary conditions. Maxwell's equations explain the coupling between electromagnetic signals around the body and body itself through a set of complete electric field equations. Recently, a theoretical model of the capacitive IBC using Maxwell's equations was proposed by Bae et al. [62] to predict the properties of the electrical wave travelling on the surface of the body. The complete equation of electric field around the body consists of three electric field components; the quasi-static near-field, induction-field radiation, and the surface wave far-field, were considered to obtain the general IBC model. The results from both measurement and proposed theoretical model indicated that increase in channel length led to channel path loss enhancement. Their model was empirically verified for operating frequencies of 0.1–100 MHz and channel lengths of up to 1.3 m.

Chen et al. [63] proposed a galvanic coupling electromagnetic model to predict the effects of the body channel loss opposed to body tissue (muscle) thickness in both IBC and RF techniques. The transmitter and receiver were supposed as implantable and wearable sensors, respectively. When the implanted sensors were 60 mm deep from the body surface, simulation results using COMSOL predicted 35 dB and 50 dB signal attenuation for IBC and RF techniques, respectively. This difference increased to 20 dB for a transmitter-receiver distance of 80 mm. The results also demonstrated that galvanic coupling IBC is more power efficient than RF for deeper implants.

2.4 IBC Transceiver Design

In communication systems design several challenges need to be addressed. Channel characteristics are the main challenge for an ideal communication system design. Typically, transmitter, communication channel, and receiver comprise three fundamental stages of any communication system. The transmitter is composed of several sub systems: an analog-to-digital converter (ADC) an encoder, and a modulator. Likewise, the receiver may include a demodulator, a decoder, and a digital-to-analog converter (DAC). A communication channel refers to a physical transmission path which allows the propagation of the signal. It determines the technique to be used in real communications. The block diagram of a general IBC transceiver system is depicted in Fig. 2.10.

A communication channel functions relatively like a filter that attenuates the signal and causes transmission signal loss and distortion. The channel distance affects the signal attenuation, where larger distances result in more attenuation. Furthermore, frequency dependent gain characteristics and multipath effects cause transmission wave shape distortion. These phenomena necessitate deeper understanding of the transmission medium to design more effective IBC transceivers [64].

The maximum efficiency of a transceiver is determined by its compatibility within the network [65]. Examples of transceiver design parameters are: data rate (number of bits per second), sensitivity which is the minimum signal power required to receive data correctly, transmitter output power, communication interface, carrier operation frequency, or the range of signal that can be sent and received, measurement resolution, and maximum transmission distance. Measurement resolution determines the smallest digital resolution while maximum transmission distance is the largest possible distance of the transmitter and receiver. Extra factors affecting the choice of transceivers involve power source, supply voltage, supply current, transmitter inputs, receiver inputs, and RF connector types. Uniquely designed transceivers will reduce the complexity and will offer the ability to fully integrate the whole system into a more compact form [43].

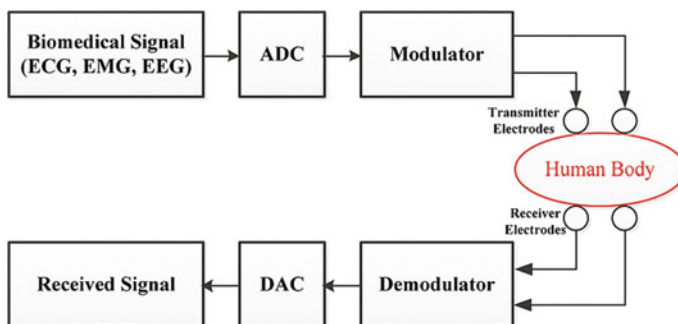


Fig. 2.10 Simplified block diagram of the IBC transceiver system

While a distinctive feature of IBC is to design efficient hardware transceivers, the hardware complexity of units including size, power, and cost must be minimized [66]. However, there is no definite principle to obtain the best electronic design of an IBC system. Power consumption, data rate, carrier frequency, and modulation method are the main concerns [9]. Therefore, the IBC transceiver should be simple, have full integration ability, consume low power, and have the ability to transmit at low power. Since power hungry transceiver nodes need large batteries, power sources like solar cells appear to be suitable for IBC. At the same time, low voltage operation is required for IBC due to health and safety reasons e.g. direct contact with the human body [67].

Several IBC transceiver designs have been proposed based on capacitive and galvanic coupling approaches. However, no acceptable standard has been established to implement an optimal design of a full intra-body transceiver system in terms of carrier frequency, modulation scheme, data rate, and power consumption. In the following, a survey of some recent transceiver designs is highlighted.

A battery powered transmitter and receiver unit was developed in the first IBC system prototype PAN (personal area network) transceiver [68]. Results indicated that the best received signal magnitude could be obtained when the PAN devices were placed on the feet close to the physical ground. Two kinds of digital modulation, on-off keying (OOK) and direct sequence spread spectrum (DSSS) techniques were examined in the scheme. In OOK, the existence of the carrier produces a binary one, while binary zero is represented by switching off the carrier. DSSS modulates the carrier signal with pseudo-noise (PN) sequences which are widely used in digital communications. The whole PN sequence is transmitted when a message bit is one, while it is inverted when a message bit is zero. To extract the message at the receiver, the transmitter and receiver PN sequence must be synchronized which is the greatest challenge of a DSSS system. Unsurprisingly, the OOK was found to be more effective and easier to implement for the PAN transceiver. The optimum carrier frequency range was determined between 100 kHz to 500 kHz and a suitable data rate was considered to be 2.4 kbps. However, the data rate was not practical for sensors sending high streams of data. The transceiver circuit of capacitive coupling IBC system was designed and implemented by Partridge et al. [69]. They extended Zimmerman's original IBC system by adding filters and amplifiers to the transceiver circuit. Two different microcontrollers were used in the circuit to generate and modulate the input digital signal. In addition, they utilized a frequency-shift keying (FSK) detector in the proposed circuit achieving a data rate of 38.4 kbps. The changes of frequency when message signal switch from zero to one or vice versa in FSK scheme leads to transmission of a large amount of data in a narrow bandwidth. Since complex body area networks such as those using portable electroencephalogram (EEG) 192-channel recordings [9], need data rates of almost hundreds of kbps, the achieved data rate was barely sufficient for continuous monitoring. To compensate the low data rate of an IBC transceiver in such a network, a larger array of sensors is required for increased redundancy which leads to increased power consumption at the same time. The low power consumption and high data rate (2.0 Mbps) capacitive coupled IBC transceiver based on wideband signaling (WBS) communication method was introduced

in [70]. The WBS scheme directly transmits binary digital signals through wideband pulse signals. Lin et al. [67] implemented the small size system on a chip (SOC) IBC system for biomedical applications for the first time. The authors believed that the suitable carrier frequency for IBC was 200 MHz and data rates up to 2 Mbps were reported for OOK modulation. Moreover, they proposed a new generation of transceiver which required less than 0.5 V making it suitable for use with solar cell power. Unfortunately, this system had a short channel length or transmission distance (from subject's wrist to forearm). Recently, Xu et al. [22] inspected the systematic features of the electric field IBC (EF-IBC) channel containing attenuation, noise, and distortion by making use of battery powered transceiver board. Two modulation methods of binary phase shift keying (BPSK) and quadrature PSK (QPSK) was examined and a high speed 10Mbps EF-IBC links was created. A microcontroller unit was used in transceiver board to generate the baseband in-phase and quadrature digital signals and also controlled the digital direct synthesizer (DDS). Various measurements with different received signal powers and symbol rates indicated that the channel noise was white Gaussian. It was also concluded that both the signal-dependent distortion and the signal independent noise influenced the channel signal-to-noise ratio (SNR). Considering the bit error rate (BER) as a modulation scheme metric, it was revealed that QPSK has the highest transmission power at both 10^{-3} and 10^{-6} BER. A recent suggested capacitive coupling IBC transceiver by Bae et al. [21] was able to fulfill the requirements of the WBAN standard, such as network coexistence and quality of service (QoS) scalability, at the 10 Mbps data rate in the operating frequency range of 40–120 MHz. This energy efficient transceiver was combined with contact impedance sensing (CIS), a ring oscillator as well as the double-FSK modulation scheme which decreased the power consumption of the transmitter and receiver to 2.0 and 2.4 mW, respectively.

The principles of galvanic coupling were first introduced by Oberle [40] while designing a single chip low power biomedical system. The continuous phase frequency shift keying (CPFSK) modulation scheme was used in his implementation. A waveguide measurement approach was first performed by Hachisuka et al. [54, 61] to confirm the possibility of IBC design and to analyze the transmission characteristics of the human body. The various electrode sizes and materials as well as an ideal carrier frequency range were specifically taken into account. Transmission devices using 10.7 MHz FSK at 9.6 kbps, and 10.7 MHz FM were developed for digital and analog transmission respectively. A suitable carrier frequency range of between 10 to 50 MHz was considered. However, the proposed transceiver board in this study suffered from low data communication rate, low integration level, and large form factor. In further experiments on the galvanic approach by Wegmueller et al. [71, 72], the human body characteristics were taken into account and models were generated for static position of human body. In practical measurements, two FSK and BPSK digital modulation types were applied with a 128 and 255 kbps data transfer rate respectively. A battery powered transceiver was also employed in order to isolate the sensor units from other power line. A field-programmable gate array (FPGA) was used to provide interfaces between analog front-end and digital communication link. However, the achieved data rate of transceiver was low for some biomedical appli-

Table 2.2 Summary and comparison of currently reported IBC transceivers

	Coupling method	Carrier frequency	Modulation technique	Data rate	Power consumption
[68]	Capacitive	330 kHz	OOK	2.4 kbps	400 mW
[69]	Capacitive	160 kHz	FSK	38.4 kbps	Not reported
[61]	Capacitive	10.7 MHz	FSK	9.6 kbps	Not reported
[70]	Capacitive	1.0–200 MHz	Not reported	2.0 Mbps	2.4 mW
[67]	Capacitive	200 MHz	OOK	2.0 Mbps	4.535 mW
[22]	Capacitive	20–100 MHz	PSK	10 Mbps	Not reported
[21]	Capacitive	40–120 MHz	Double-FSK	10 Mbps	4.4 mW
[40]	Galvanic	60 kHz	CPFSK	4.8 kbps	Not reported
[9]	Galvanic	256 kHz	BPSK	64 kbps	726 mW

cations such as medical implant communications service (MICS). An IBC/MICS dual modes transceiver was designed and fabricated by Cho et al. to communicate with both on-body and implanted sensors [73]. The operation frequency range was considered 30 to 70 MHz and 402 to 405 MHz for IBC and MICS, respectively. The achieved data rate of system was 5 Mbps for IBC and 200 kbps for MICS transceiver. Since the common front-end circuits including amplifier and mixer was employed in both IBC and MICS receiver, therefore the total power consumption of system was reduced to 10.8 mW. A list of some current transceivers based on both capacitive and galvanic coupling IBC is shown in Table 2.2.

In 2009, Al-Ashmouny et al. [74] applied the IBC method using brain tissue as a communication medium calling it intra-brain communication. Two miniaturized communication chips were designed and implanted in a rat brain. BFSK modulation technique was used in the proposed system with frequencies of 100 to 400 kHz. While the distance between electrodes was 15 mm, the energy consumption was less than 650 pJ per transmitted bit. Similarly recent research on wireless intra-brain communication has also transmitted data through a rat brain [75]. Authors designed the small CMOS chip ($550\text{ }\mu\text{m} \times 700\text{ }\mu\text{m}$) to successfully transmit a 50 MHz AM modulated signal using a 3.3 V supply voltage.

2.5 IBC Challenges

Studies in the literature have investigated the IBC technique from different perspectives. These aspects include diverse methods to measure the IBC characteristics, various experimental condition, different kinds of modeling, and several methods to design the efficient IBC system. However, despite the considerable and divergent amount of works on IBC and despite the fact that IBC has been ratified in the WBAN standard, there are still challenges to be addressed. For instance, the effect of IBC user motion on transmission quality is one of the main issues which need to be deal

with. The influence of static and dynamic body situations is an influencing element since the IBC system needs to continuously keep track of the patients' or athletes' vital signs.

At the same time, the increasing data rate for low frequency carriers is needed in order to develop an energy efficient IBC system. IBC standard postulates the maximum data rate of 1321.5 kbps in the 21 MHz however, the results fulfills only the static body condition during the capacitive coupled IBC. This study investigates the effects of body motions during galvanic and capacitive IBC methods within a wider frequency range (up to 200 MHz) compared to previous researches. It also pursues to achieve a suitable frequency range for IBC during body movement.

We examine a digital baseband signal traveling through the body and propose a new energy efficient IBC system applying the galvanic coupling IBC method.

2.6 Summary

The present chapter provided a deep survey of research in IBC method. IBC is a short range non-RF wireless communication technique specified by the IEEE 802.15.6 using the human body as a transmission medium. IBC is a recent and novel communication technique and research studies in this field are in progress so as to address the many challenges and issues. In this chapter, we reviewed the current IBC coupling methods, various IBC models, and latest transceiver designs. As it stands, the IBC technique potentially offers a more power efficient and naturally secure short range communication method for body sensor networks, compared to wireless RF.

In the next chapters, IBC characterizations based on empirical measurement will be investigated comprehensively at the higher frequency range compared to past studies. Since there was a dearth of empirical studies on the signal propagation through dynamic body, we aim to examine the body motion effects on IBC through the next chapter. Moreover, new body channel model will be proposed considering the limb joints effect in this dissertation. Finally, a new energy efficient IBC transceiver will be designed and implemented on FPGA. The transceiver adapts PPM modulation with high data throughput for WBAN applications.

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Biomedical Applications

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2017, XXIII, 108 p. 77 illus., 69 illus. in color., Hardcover

ISBN: 978-981-10-2823-6