A Survey on Intrabody Communications for Body Area Network Applications

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(Review Paper)

***Abstract*—The rapid increase in healthcare demand has seen novel developments in health monitoring technologies, such as the body area networks (BAN) paradigm. BAN technology envisions a network of continuously operating sensors, which measure critical physical and physiological parameters e.g., mobility, heart rate, and glucose levels. Wireless connectivity in BAN technology is key to its success as it grants portability and flexibility to the user. While radio frequency (RF) wireless technology has been successfully deployed in most BAN implementations, they consume a lot of battery power, are susceptible to electromagnetic interference and have security issues. Intrabody communication (IBC) is an alternative wireless communication technology which uses the human body as the signal propagation medium. IBC has characteristics that could naturally address the issues with RF for BAN technology. This survey examines the on-going research in this area and highlights IBC core fundamentals, current mathematical models of the human body, IBC transceiver designs, and the remaining research challenges to be addressed. IBC has exciting prospects for making BAN technologies more practical in the future.**

***Index Terms*—Capacitive coupling, electrical properties, galvanic coupling, human body tissues, intrabody communication (IBC), modeling, radio frequency (RF), transceiver design, wireless communication.**

I. Introduction

THE current paradigm in healthcare is the notion of continuous remote patient monitoring using a network of wireless sensors. These healthcare sensor network systems [1], consisting of body area networks (BAN) and infrastructure area networks avoid the need for a manual self-administered health system and may enable users to take control of their health disorders in the future. BAN technology envisions miniaturized sensors worn or implanted on the body, continuously monitoring health parameters and acting to prevent the onset of critical health events. For example, diabetics now have access to an automatic insulin pump which monitors glucose levels and administers insulin when glucose levels are high. Similar technologies will one day result in devices which can minimize incidences of heart attack or stroke. They could prevent frequent hospital visits and save 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 will expand with an annual growth rate of over 180% and become a $4.4 billion industry in 2013 [2]. Statistics such as these indicate a rising demand for portable health monitoring devices e.g., BANs, which are currently undergoing tremendous research and development.

At the heart of BAN technology is the need for short-range data communications, both from sensors to a decision maker (usually a hub node) or from the hub node to an actuator, e.g., insulin delivery system [3]. Wireless data communications based on radio frequency (RF) have been successfully developed using popular protocols such as Bluetooth, Zigbee, and ANT [4]. A major drawback of wireless RF propagation for miniaturized medical portable monitoring devices is the high-power consumption which limits the practical duration of an operation. Most current research claim that Zigbee and ANT have a battery life of three years, but this is at a low operating data rate, e.g., 1 B transmitted per 5 min [4]. The IEEE 802.15.4 standard for low power Zigbee protocol indicates a transmission power output of 0 dBm (1 mW). Continuous operation at the maximum data rate of 250 kb/s generally consumes a normal Lithium ion battery in a matter of hours. It is evident that new approaches to ultralow power wireless technology are required for improving next generation BAN technology.

This review aims to compile the latest research on a new form of wireless communications which is fast gaining attention. Intrabody Communications (IBC) is a novel non-RF wireless data communication technique which uses the human body itself as transmission medium for electrical signals. The IBC technique is defined in the recently ratified IEEE 802.15.6 WBAN protocol. 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 kb/s [5] which makes this approach potentially appealing as a short-range communications alternative.

This paper is organized as follows: Section II briefly reviews the IEEE 802.15.6 physical layers (PHYs) particularly the background of the IBC technique which is primarily based on capacitive or galvanic coupling. Section III presents the effects

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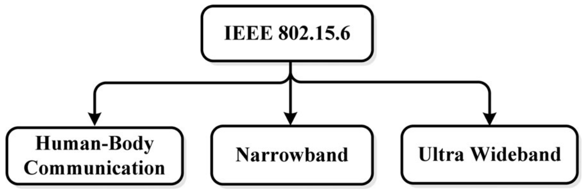

Fig. 1. PHYs of IEEE 802.15.6 standard.

of human body tissues dielectric properties on signal transmission through the body. The various human body communication (HBC) channel modeling methods reported in the literature and their characteristics are reviewed in Section IV. These are followed by the latest state of the art transceiver designs and concluding with the remaining research issues in this exciting new field.

II. IEEE 802.15.6 WBAN

A. RF and Non-RF WBAN

Monitoring vital signs (e.g. heart-beat, body posture, blood pulse pressure, oxygen saturation [6]) is now performed through miniature sensors. They are lightweight and intelligent devices, capable of sensing physical phenomena, basic processing and wireless transmission. Wireless sensor motes are attached on or implanted in the body and develop short distance wireless networks around the human body, using RF transmission technology [7]. In the late 2011, the standardization of a new wireless body area network (WBAN) protocol, IEEE 802.15.6 by task group (TG6) was ratified. The IEEE 802.15.6 standard outlined three PHY schemes as shown in Fig. 1 [8]. Narrowband (NB) and Ultrawideband (UWB) PHY operation are based on RF propagation, while a new non-RF technique based on HBC was introduced. Multiple frequency bands of operation supported by NB PHY include the 402–405 MHz for implantable devices, three different frequency bands (863–956 MHz) for wearable applications, and finally 2360–2400 MHz for medical demands. Meanwhile, UWB PHY operates in the higher frequency regions, particularly the 3–5 GHz and the 6–10 GHz bands with channel bandwidth (f_{BW}) of 499.2 MHz [12]. Data rates of NB PHY range between 100 and 1000 kb/s and UWB PHY range from 395 kb/s–12.636 Mb/s in the mandatory mode. Unfortunately, these frequency bands are not suitable for HBC, due to high-signal attenuation through the human body as well as severe shadowing effects. Instead, the operating frequency band of HBC is centered at 21 MHz with scalable data rates of 164 –1312.5 kb/s [12].

HBC has been referred by other authors as body channel communications (BCC) [9] or intrabody communications (IBC). In this paper, we will use IBC to denote this communication method as the review will cover work outside the IEEE 802.15.6 standard. The IBC technique was touted to have the following desirable characteristics.

Security: The IBC system is a protected and private communication network which provides natural security and interference-free communication. The required operating frequency of IBC is much lower compared to RF systems. This means signals are confined to the person's proximity since reading data requires body contact [10]. There is no signal leakage through the skin in IBC method and environmental noise has less effect on communication. At higher frequencies (300 MHz to several gigahertz), the signal wavelength becomes comparable to the human body channel length and body radiates energy acting as an antenna (dipole antenna). Since transmitter and receiver contain small size electrodes (for example Neuroline electrodes active area is 54 mm^2) instead of antennas, the larger wavelength of the carrier signal compared to the electrode size results in interference-free IBC below 300 MHz [11].

TABLE I
COMPARISON OF IBC AND RF SPECIFICATIONS IN IEEE 802.15.6

	IBC	RF (NB , UWB)
Communication Medium	Human Body	Air
Frequency Band	Centered at 21 MHz (f_{BW} = 5.25 MHz)	Different bands (402MHz-10GHz)
Data Rate	< 2 Mb/s	< 13 Mb/s
Transmission Range	< 2 m	10 m
Signal Attenuation	Low	High (Body Shadowing)
On-Body Antenna	No	Yes
Energy Efficiency	High (High Conductivity of the human body)	Low (Air has low conductivity)

Energy consumption: The key issue with RF propagation in portable devices is that it consumes battery life quickly. For example Zigbee has the maximum data rate of 250 kb/s at 26.5 mW resulting in 106 nJ per received bit [9]. The energy consumption of UWB is 2.5 nJ/b when data rate was 16.7 Mb/s. Bae *et al*. [9] recently demonstrated that IBC consumes an order of magnitude less energy (0.24 nJ/b) at data rates up to 10 Mb/s which makes it an attractive communications method for WBAN applications.

Frequency reuse: IBC forms a short-range communication network inside and around human body and therefore allows the same frequency band to be reused by WBANs on other users with minimal interference. This property potentially allows future designs to focus on improving data rates, reducing power consumption, and integrating smaller form factors.

Table I (adapted from [12]–[15]) indicates the brief comparison between IBC and RF WBAN specifications.

B. 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 (electric field) and the galvanic coupling (waveguide). Fig. 2 illustrates schematically the two different types of IBC coupling. For both coupling types the transceiver needs two pair of electrodes. In capacitive coupling only one of the electrodes (signal electrode) of the transmitter side and receiver side is attached to the body, while the other electrode (ground electrode)

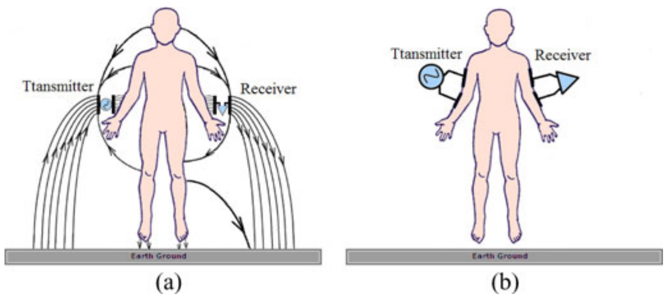

Fig. 2. (a) Capacitive-coupled IBC and (b) Galvanic-coupled IBC method.

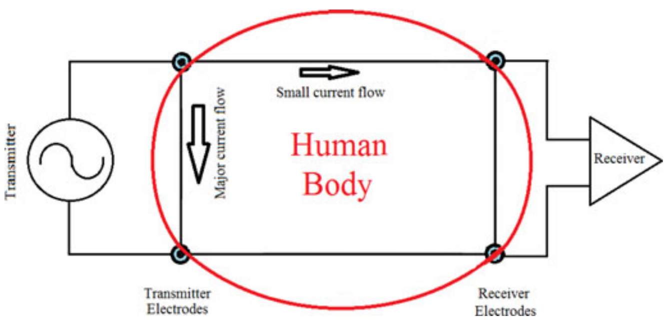

Fig. 3. Current flow establishes between transmitter and receiver electrodes in the galvanic method.

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.

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. Fig. 3 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 of the receiver. In general, the ion content in the human body is the carrier of information in the galvanic coupling method. Table II briefly compares capacitive and galvanic coupling methods.

III. Electrical Properties of the Human Body 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 [18]. 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 [19]. Experiments were performed on human and animal tissue within the frequency range of 10 Hz –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 [20] 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. Fig. 4 shows the dispersion of relative permittivity and specific conductivity of human body tissue. The major characteristics of biological tissues can be summarized as follows.

1) Permittivity strongly declines, whereas conductivity increases within these frequency dispersions.
2) 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.
3) 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.
4) 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 [20].

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.

IV. 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

TABLE II
COMPARISON BETWEEN CHARACTERISTICS OF CAPACITIVE AND GALVANIC COUPLING 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 [16].	The dominant signal transmission pathway is the body tissue.
Higher transmission date rate and channel gain (Higher operation frequency compared to galvanic coupling) [9].	Lower transmission date rate [43].
The human body is modeled as a perfect conductor (body is approximated as a single node [29]).	The body is modeled as a waveguide for signal conduction [55].
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 [17].

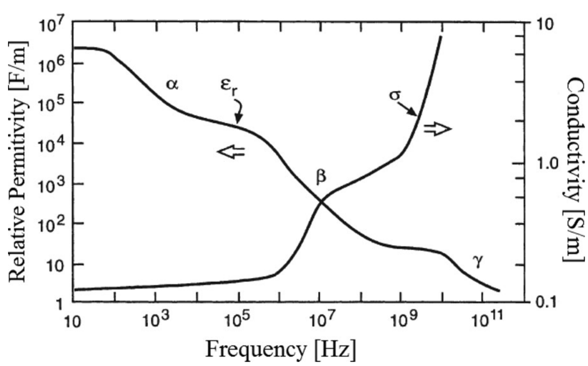

Fig. 4. Variation of human tissues electrical properties against frequency.

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 [21], and tissue-specific models are expected to be applied similarly in the IBC field to future IBC transceiver design.

A. 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 [22] presents the change of dielectric properties of a tissue over a broad frequency range

$$\varepsilon^*(\omega) = \varepsilon_\infty + \frac{\Delta\varepsilon_n}{1 + (j\omega\tau_n)^{(1-\alpha_n)}} \tag{1}$$

where $\varepsilon*$ is the complex dielectric constant, $\Delta\varepsilon_n$ 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

$$\varepsilon^*(\omega) = \varepsilon_\infty + \sum_n \frac{\Delta\varepsilon_n}{1 + (j\omega\tau_n)^{(1-\alpha_n)}} + \frac{\sigma_{\mathrm{i}}}{j\omega\varepsilon_0} \tag{2}$$

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

$$\sigma^* = j\omega\varepsilon_0\varepsilon^*,\ \mathrm{z}^* = \frac{1}{\sigma^*}. \tag{3}$$

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. 5. The simulation results are plotted based on (2) by [19].

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 leakages [23].
2) A more common method is presenting the circuit model by considering a "constant phase element" (CPE) with a complex valued impedance given by $Z_{\mathrm{CPE}^*} = A(j\omega)^{-n}$ 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 [24]. The model representation of resistive and capacitive elements appears to explain the

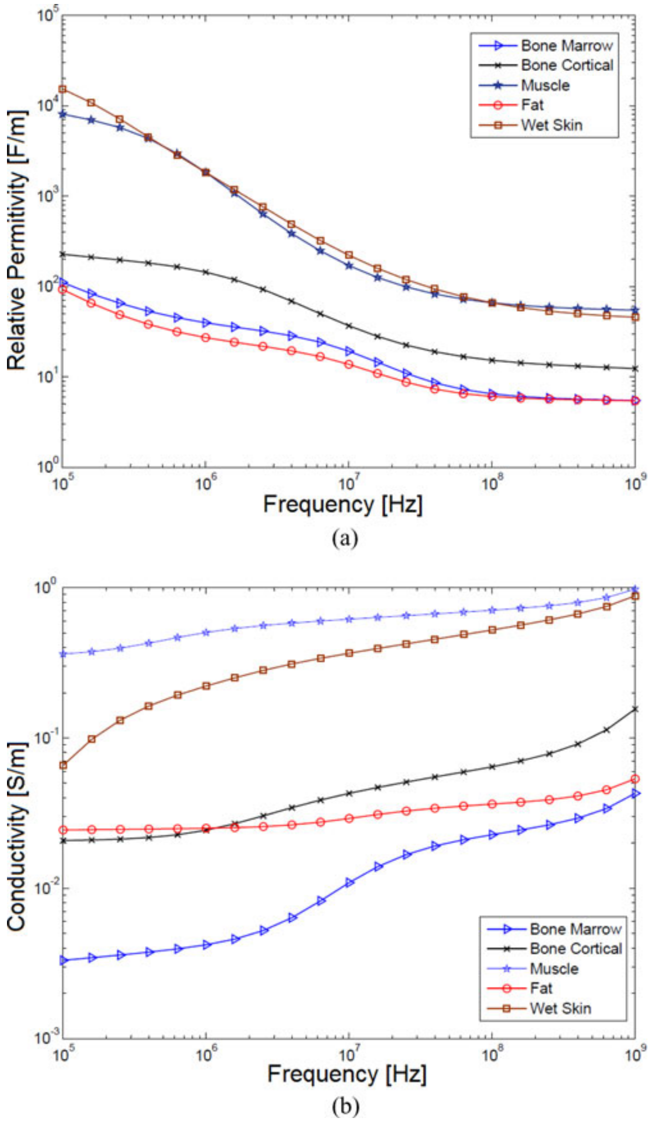

Fig. 5. (a) Relative permittivity of human body tissues at different frequencies. Fig. 5. (b) Conductivity of human body tissues at different frequencies.

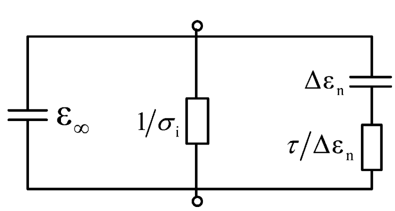

Fig. 6. Cole–Cole equation equivalent circuit model for a single time constant.

empirical measurements well. However, determining the exact tissue components responsible for these properties is complicated by the nonhomogeneous nature of tissue and randomly distributed cells sizes.

The circuit model of (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 [25] in Fig. 6. In fact, the Cole–Cole model is mainly applied to biological materials while other distributions

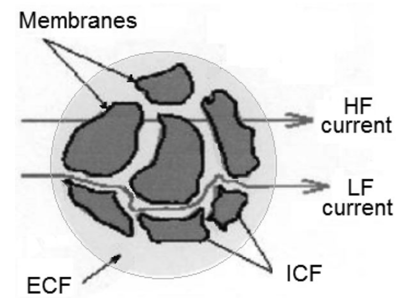

Fig. 7. Current flow in human body tissues [26].

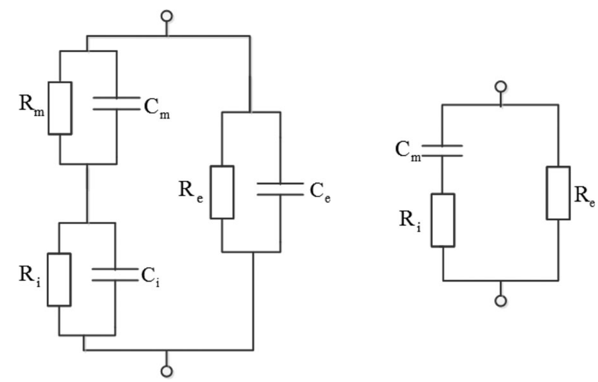

Fig. 8. Complex and simplified equivalent circuit model of human tissues.

like Cole–Davidson and Havrilak–Negami are used for nonbiological materials [23].

Electrical current flow through human tissue follows several pathways including intracellular, extra-cellular, and cellular membrane pathways. The high frequency current tends to pass easily through the tissues [higher conductivity as seen in Fig. 5(b)]. The intracellular fluid is a liquid that circulates inside the cell and surrounded by cell membrane as well as extracellular fluid). The two lines in Fig. 7 show the low frequency current and high frequency current pathways.

Extra-cellular resistor and capacitor is modeled by $\mathrm{R_e}$ and $\mathrm{C_e}$. Constituents of cell membrane and intracellular contribution are indicated by $\mathrm{R_m}$ and $\mathrm{C_m}$ as well as $\mathrm{R_i}$ and $\mathrm{C_i}$, respectively. Kanai *et al.* in [27] has presented the complex and simplified circuit model of human tissues seen in Fig. 8. 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 [27].

B. Body Channel Circuit Model

Obtaining the transfer function of the IBC is the first step for construction of the circuit model. Zimmerman [28] proposed the first circuit model of body communication channel. The model consists of four significant transversal and longitudinal impedances among transmitter and receiver electrodes. The interelectrode impedances between the electrodes of the transmitter and receiver were ignored in impedance calculations of his model.

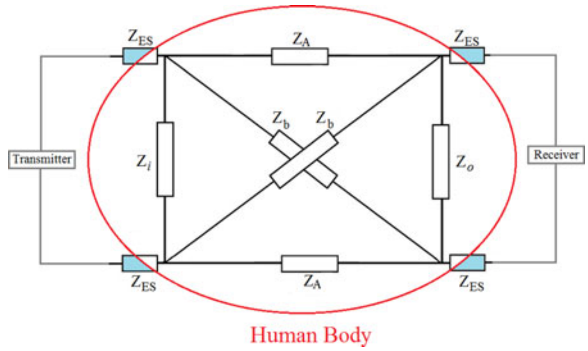

Fig. 9. Four terminals (electrodes) human arm circuit model.

In [29], capacitive coupling frequency characteristics of the human body are identified through a developed RC model. 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 [19] and Zimmerman's circuit model [28] were used. Xu *et al*. [21] proposed a capacitive coupling channel circuit in which they took the body shielding effects into account. Recent examinations [30] 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*. [31] 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 [31], 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*. [5] (see Fig. 9). 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 [32]. 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*. [33], good contact between electrode and skin surface is established when $Z_{\mathrm{ES}} < 10\ \mathrm{k}\Omega$. 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 [34]. 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 number of body tissues layers.

C. Finite-Element Model (FEM)

The FEM is a technique which could model individual body tissues. It is based on the numerical solutions partial differential equations, and integrals. Electrical behavior of 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 IBC method. It is also able to reconstruct the potential distributions caused by induced current into the human tissue. More developed opportunities are also provided by this method for simulating body geometries.

Xu *et al*. [21] 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 signal attenuation through the body in galvanic coupling IBC was investigated using FEM [43]. Authors explored the influence of the distance between transmitter and receiver electrodes, sensitivity to resistivity changes of selected tissues such as skin, and wet and dry skin on signal propagation through the body in this study. They employed the commercial package EMAG from ANSYS to model the arm of human body. While the model simulation results showed that an increase in the transmitter electrodes size led to lower signal attenuation, changing the size of receiver had a small effect of less than 1 dB in attenuation. In general, there was good agreement between simulation and empirical results over the investigated frequency range of 10–1000 kHz.

However, a major drawback with this method is the large size and complexity of the human body which can significantly reduce the simulation accuracy [34].

D. Circuit-Coupled FEM

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 multilayer FEM and the parasitic effects of probe PCBs (printed circuit board) were modeled by *LC* circuits (Fig. 10). 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 [35]. 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.

E. FDTD Model

The FDTD model is a computational modeling technique used in the field of electromagnetism. It examines the distribution of

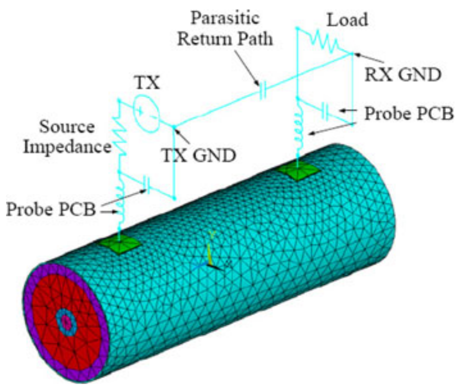

Fig. 10. Human forearm circuit-coupled FEM model [35].

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.* [36]. 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 [37]) which is outside our current range of interest.

F. 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 was proposed by Bae *et al.* [38]. Three electric field components, the quasi-static near field, induction-field radiation, and the surface wave field were considered in a 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.* [39] 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 encouraging demonstrate that galvanic coupling IBC is more power efficient than RF for deeper implants.

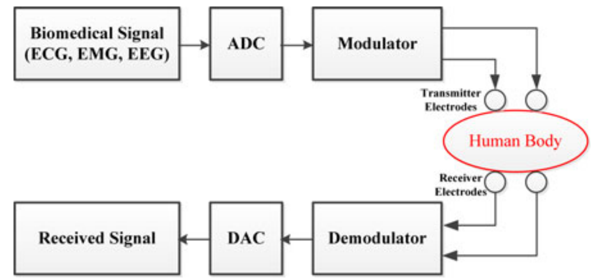

Fig. 11. Simplified block diagram of the IBC transceiver.

V. 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 subsystems: 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 a real communication. The block diagram of a general IBC transceiver system is depicted in Fig. 11.

A communication channel functions relatively like a filter that attenuates the signal and causes transmission signal loss. 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 [40].

The maximum efficiency of a transceiver is determined by its compatibility within the network [41]. 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 [17].

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 [42]. 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 [43]. 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 body [44].

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 quick survey of some recent transceiver designs is highlighted.

A battery powered transmitter and receiver unit was developed in the first IBC system prototype personal area network (PAN) transceiver [45]. 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. The DSSS modulates the carrier signal with pseudonoise (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 and 500 kHz and a suitable data rate was considered to be 2.4 kb/s. 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.* [46]. 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 kb/s. 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 BAN such as those using portable electroencephalogram (EEG) 192-channel recordings [43], need data rates of almost hundreds of kb/s, 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 Mb/s) capacitive coupled IBC transceiver based on wideband signaling (WBS) communication method was introduced in [47]. The WBS scheme directly transmits binary digital signals through wideband pulse signals. Lin *et al.* [44] 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 Mb/s were reported for an 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). Very recently, Xu *et al.* [48] 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 10 Mb/s 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.* [9] was able to fulfill the requirements of the WBAN standard, such as network coexistence and quality of service (QoS) scalability, at the 10 Mb/s 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 [51] 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.* [31], [37] 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 kb/s, and 10.7 MHz FM were developed for digital and analog transmission, respectively. A suitable carrier frequency range between 10 and 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.* [52], [53], 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 kb/s 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 was used to provide interfaces between analog front end and digital communication link. However, the achieved data rate of transceiver was low for some

TABLE III
SUMMARY AND COMPARISON OF CURRENTLY REPORTED IBC TRANSCEIVERS

Author	Year	Coupling Method	Coupling Amplitude	Carrier Frequency	Modulation Technique	Data Rate	Overall Power Consumption
Zimmerman	1995	Capacitive	30 V	330 kHz	OOK	2.4 kb/s	400 mW
Partridge	2001	Capacitive	22 V	160 kHz	FSK	38.4 kb/s	Not reported
Hachisuka	2005	Capacitive	1 V	10.7 MHz	FSK	9.6 kb/s	Not reported
S. J. Song	2007	Capacitive	1 V	1 ~ 200 MHz	Not reported	2 Mb/s	2.4 mW
Y. T. Lin	2011	Capacitive	0.5 V	200 MHz	OOK	2 Mb/s	4.535 mW
R. Xu	2012	Capacitive	0.4 V	20 ~ 100 MHz	BPSK/QPSK	10 Mb/s	Not reported
J. Bae	2012	Capacitive	1 V	40 ~ 120 MHz	Double-FSK	10 Mb/s	4.4 mW
Oberle	2002	Galvanic	4 mA	60 kHz	CPFSK	4.8 kb/s	Not reported
Wegmuller	2007	Galvanic	1 mA	256 kHz	BPSK	64 kb/s	726 mW

biomedical applications 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 [54]. 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 Mb/s for IBC and 200 kb/s for the MICS transceiver. Since the common front-end circuits including amplifier and mixer were 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 III.

In 2009, Al-Ashmouny *et al*. [49] applied the IBC method using brain tissue as a communication medium calling it intrabrain 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 transmission power was less than 650 pJ/bit. Similarly, recent research on wireless intrabrain communication has also transmitted data through a rat brain [50]. Authors designed the small CMOS chip (550 μm $\times$ 700 μm) to successfully transmit a 50 MHz AM modulated signal using a 3.3 V supply voltage.

VI. FUTURE CHALLENGES OF IBC

IBC is a relatively new communications technology, and research is still on going to address the many challenges and issues in the field. In most communications systems, higher frequency implies a high data rate which is problematic in IBC due to an antenna and heating effects to the body. The IEEE 802.15.6 standard for WBAN postulates the maximum data rate of IBC as 1312.5 kb/s in the 21 MHz frequency band. Previous research and our results have shown that minimum signal attenuation was achieved between 80–90 MHz [56]. In the capacitive method, the carrier wavelength can approach the size of the human body when frequencies go above 300 MHz. The body behaves as an antenna and signals suffer significant channel variations. On the other hand, frequencies below 100 kHz are vulnerable to all kinds of electromagnetic interference [55]. The optimum range for the capacitive method appears to, therefore, be between these two frequencies. This is promising because further research could potentially extend the carrier frequency range in capacitive coupled IBC and improve data rates. In the galvanic coupling method, the frequency range is mostly governed by the dielectric properties of human tissues (relative permittivity and conductivity). According to Gabriel *et al*. [19], 100 kHz and 10 MHz is a suitable frequency range in which the permittivity and conductivity of human body tissues is almost constant. The reduced bandwidth compared to the capacitive method implies a lower data throughput. However, the reliance of the ground capacitive coupling to complete the current loop means that the capacitive technique is more susceptible to noise interference compared to the galvanic method. The research focus should be on improved data throughput within these constraints, for example exploring and adapting well-known approaches in an RF design, such as coding and compression.

Since various postures are encountered during human body movement, the IBC system should be adaptable to body motion. Although motion effects in RF communication systems have been reported, body motion in IBC systems has not been considered widely. We have shown recently *in vivo* experimental results toward understanding particular body movement effects on signal attenuation (S21) in capacitive coupling IBC systems [56]. Measurements on body limb joint have suggested that signal transmission will be affected by body movement. The measurements were carried out in the middle of an electronic laboratory with some measurement equipment, computers, and furniture. Three subjects volunteered to participate in this study (experimental setup and details in [56]). In all experiments, the transmitter electrodes were 134 and 146 cm above the ground level in female and male subjects, respectively (see Fig. 12). During measurements, the ground electrodes of transmitter and receiver were floating and perpendicularly fixed to a distance of 2 cm from the arm. The sweep signal frequency range was from 300 kHz to 200 MHz with a 50 Ω input impedance of the network analyzer. The measurements were repeated three times and the average attenuation was recorded. Short measurement cables (50 cm compared to 90 cm which was used in [58]) were employed between body and baluns, the effect of any changes in cable positions due to different joint angle was negligible.

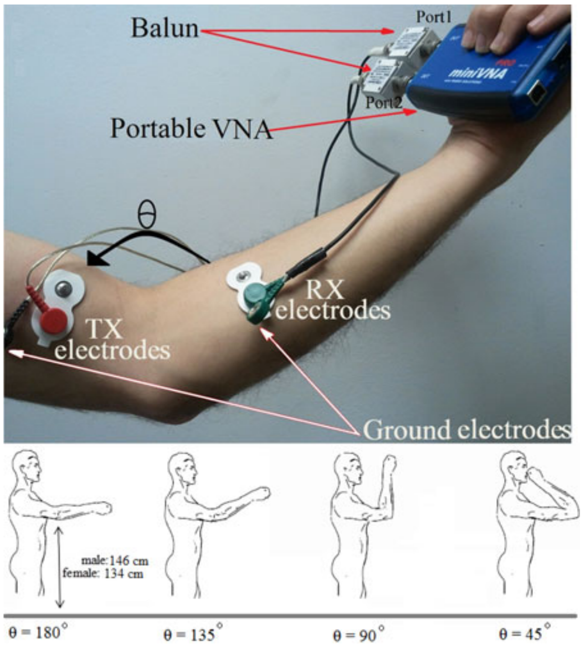

Fig. 12. Measurement setup; θ is angle between upper and lower arm, receiver and transmitter are attached to lower and upper left arm, respectively.

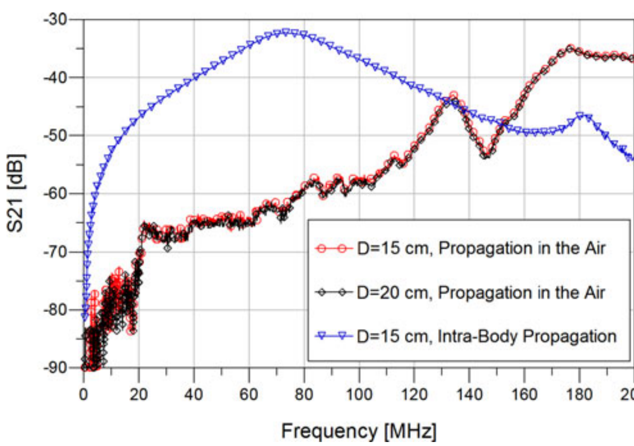

Fig. 13. Comparison of signal attenuation through the human body and air.

Fig. 13 indicates the signal attenuation between electrodes when coupled with and without the human body (air alone). Signal attenuation through air was higher below 137 MHz compared to the human body, demonstrating that the human body channel had less loss. In addition, separation distance between free electrodes appeared not to affect the signal attenuation characteristics. Above 137 MHz, the body appears to radiate the signals as shown by the increasing attenuation.

Our measurement results also showed that elbow joint flexion and extension lead to changes in signal attenuation in the capacitive coupling method (see Figs. 12 and 14). The minimum attenuation was 20.64 and 24.81 dB for the fix distance of 15 cm between transmitter and receiver electrodes and the joint angle of 45° and 180°, respectively [56]. The S21 variation during body movement occurs due to joint angle variation [57], influence of coupling between electrodes and different parts of body,

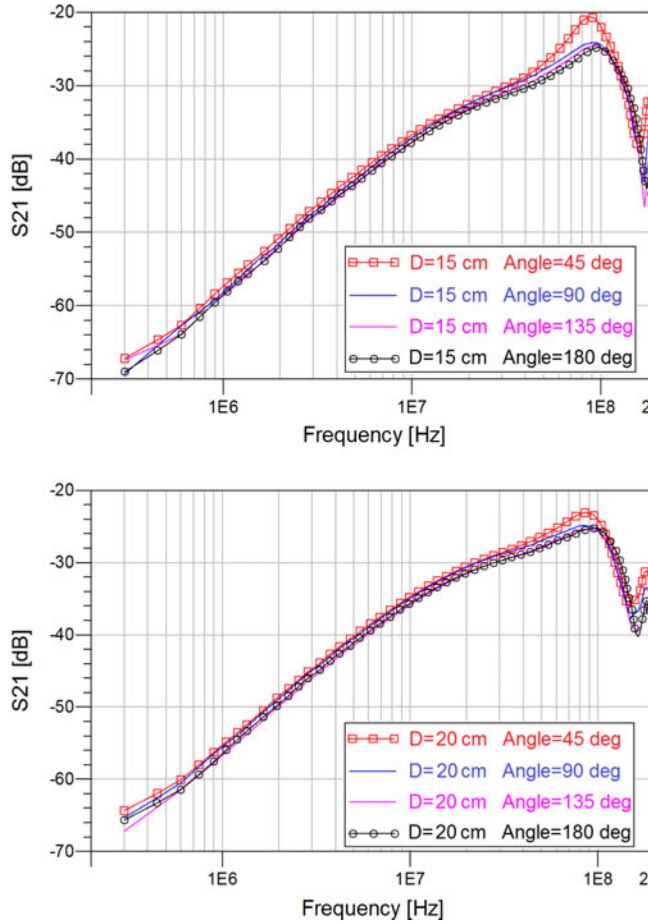

Fig. 14. Variation of signal attenuation against input signal frequencies.

and distance between transmitter and receiver electrodes from external ground (was constant in 45°, 90°, and 135°). To investigate the effect of coupling between electrodes and other parts of body, the subject was asked to stand and extend the left arm first out to the side and then out to the front. At each position, arm distance from the external ground was 146 cm and the joint angle (90°) was constant. The distance between electrodes was fixed to 20 cm. Fig. 15 indicates the results of body coupling effects in different arm position which was less than 1.0 dB. These observations suggest that further human body modeling which incorporates motion as a parameter is required for future IBC transceiver designs.

Finally, the technical hardware requirements, such as transmitter and receiver specification, need to be considered to satisfy the IEEE 802.15.6 standard for WBAN [12] in future IBC transceiver designs. Based on the WBAN standard, a transmit spectrum mask is defined to minimize possible interference with other devices in the operating frequency band. Though the transmitted signal could pass through a transmit filter to satisfy the standard, the power consumption enhancement due to this additional component remains a key issue in developing optimal transmitters. The receiver sensitivity on the other hand is another concern to achieve the maximum system data rate. According to the standard [12], the minimum receiver sensitivity level should be −88 dBm for a maximum data throughput of 1312.5 kb/s

TABLE IV
SUMMARY OF CHALLENGES AND FUTURE WORK IN THE IBC TECHNIQUE

Challenges	Future Work
1. The frequency dispersions of human tissue limits maximum carrier frequencies and data rate.	1. New techniques to improve data rates in low frequency ranges required.
2. The coupling impedance between the electrodes and the human skin (Z_{ES}) is difficult to determine, but greatly affects the transmission quality.	2. New electrode-body impedance matching schemes required to maximize power transfer.
3. Channel models incorporating human movement required. In addition, the effects of all component devices in the measurement setup should be considered.	3. Theoretical models to understand the body channel characteristics and constraints. Human channel modeling will be necessary while the body is in motion.
4. Effect of long term use on health.	4. Studies on tissue effects for IBC use in the long term, e.g. Cancer.
5. Transceiver specifications based on WBAN standard (low power consumption, transmit mask, receiver sensitivity, and signal modulation and coding) difficult to fulfill.	5. The specification of WBAN needs to be considered in future IBC transceiver design.
6. Networking issues such as security for information, QoS, inter-relationship in different environments, and mobility.	6. New transceiver design based on high security, low power consumption, enhanced data rate, and small form factor for ease of use compared to current IBC systems.

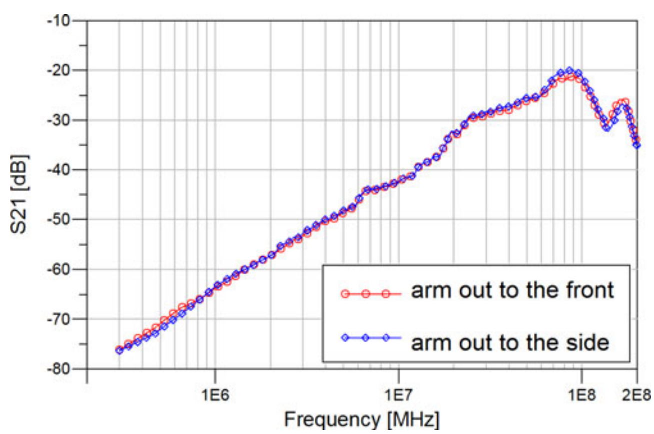

Fig. 15. Effect of coupling between electrodes and other parts of body in left arm "out to the front" (near to body) and "out to the side" (far from body) positions.

in the 21 MHz band. To improve the sensitivity of the receiver system, a transmit filter should be employed. The ultimate goal of future IBC transceiver designs, therefore, is to improve data rates and power consumption while meeting the IEEE 802.15.6 standard requirement. Table IV shows a summary of challenges and future work which should be considered in the IBC area.

VII. CONCLUSION

In the present review, research in intrabody communication (IBC) was surveyed. IBC is a new short-range non-RF wireless communication technique specified by the IEEE 802.15.6 using the human body as a transmission medium. 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. Though ratified in a standard, there are still remaining challenges such as the effect of user motion on transmission quality, increasing data rates for low frequency carriers and effect of long-term use on health which requires further advances before this technology matures.

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