

Comprehensive Survey of Galvanic Coupling and Alternative Intra-body Communication Technologies

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Abstract—Fundamental understanding of the processes that affect the state and well being of the human body continues to be a challenging frontier. Recent technological developments point towards the possibility of implanting sensors inside the body, thus enabling in-situ data gathering and real-time actuation with the transmission of the sensed data to other implants as well to external computational clouds. This paper aims to survey and compare the existing methods for achieving intra-body communications (IBC) covering the methods of galvanic coupling (waveguide based), capacitive coupling (electric field based), ultrasound (acoustic based) and magnetic resonant coupling (magnetic field based). The performance of each method is evaluated in terms of its physical layer characteristics, including operating frequency range, attenuation effects within the body, channel modeling methods, power consumption, achievable data rates, communication distance and safety limits under prolonged signal exposure. Based on this metric-driven study, we identify specific scenarios where one or more of these different IBC methods are better suited. For the specific case of galvanic coupling, we provide select system designs, experimental results from a custom-designed testbed and target applications that are more appropriate for this method of IBC. To the best of our knowledge, this is the first work that encapsulates the state of the art and current research trends for several different types of IBC solutions, a topic that promises to revolutionize the future of healthcare.

I. INTRODUCTION

Healthcare today is mired in a web of costly preventive procedures available to a limited segment of society, complex insurance rules, and lifestyle choices, all of which impact timely diagnosis. In fact, healthcare expenditures in the United States have reached over \$3 trillion USD a year [1], signaling an urgent need to explore new forms of sensor technology that can autonomously identify and react to abnormal conditions in real-time.

Traditionally, medical tests are conducted on-site at a medical facility, which may require the use of wires and leads, leading to intrusive procedures and physical discomfort. Miniaturization and advancements in MEMS technology have given rise to sensors that can be placed unobtrusively in and around the human body. The resulting connected body area network (BAN) imposes stringent design constraints that go beyond those commonly assumed for classical over the air wireless sensor networks (WSNs), particularly with respect to power consumption, size, tissue heating and electromagnetic wave absorption. Understandably, BANs have resulted in new

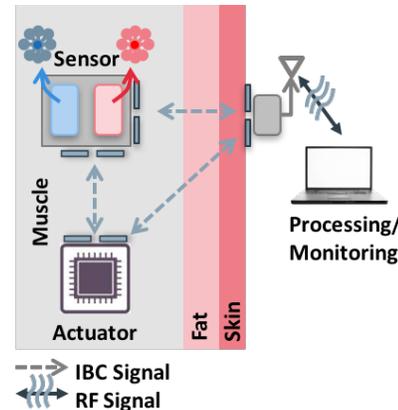


Fig. 1. Conceptual representation of an intra-body network

possibilities for preventative medical treatment by recognizing trends in medical data for rapid identification of treatment options or for sending emergency alerts for critical events [2].

• **Motivation for intra-body communication:** Solely limiting the placement of sensors on the surface of the body (i.e., the network architecture typically assumed within BANs) limits the types of biological information that can be obtained. Furthermore, it can potentially increase patient discomfort as the number of devices begins to scale and additional restrictions on movement come into play [3]. As a result, we advocate for a completely new paradigm where the sensors are implanted inside the body, with direct access to the specific target regions of interest. The implant-driven architecture, shown in Figure 1, is composed of small-scale sensors that gather biological information, actuators responsible for drug delivery functionality, and an on-body relay node for data aggregation and communication with external networks. Communication among implants can also be used to coordinate activities between different sub-sets of sensors, in terms of performing heterogeneous sensing tasks, local data aggregation and decision making by sharing computational loads.

• **Types of intra-body communication:** The most common form of an IBC link uses classical Radio Frequency (RF) in the form of narrowband (NB) or ultra-wideband (UWB) signals for implant-to-implant or implant-to-relay communication [3]. However, there are other non-traditional forms of wireless

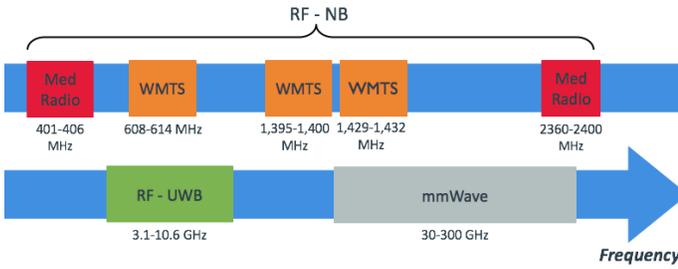


Fig. 2. Radio frequencies used by implantable medical devices

communication that have unique properties that may be more suited for IBC. The alternative methods are as follows: Ultrasound (US), widely known in the medical community for its application in imaging, is a viable option for connecting medical implants with the use of acoustic waves well above the 20 kHz range [4]. Capacitive Coupling (CC) enables the propagation of near electric fields around and through the human body, creating links that span the length of the medium [5]. Magnetic Resonant Coupling (RC), similar to inductive coupling techniques, employs loosely coupled coils wrapped around parts of the body to transmit and receive magnetic energy. This form of IBC takes advantage of the property of freely allowing the magnetic fields to flow through biological tissues [6]. Finally, Galvanic Coupling (GC), injects a weak current, on the order of 0.5 mA, into human tissue as the main signal that can be modulated to interconnect implants. Leveraging the dielectric properties of human tissue (e.g., conductivity, permittivity, etc.), this waveguide based approach confines the signals within the human body, while ensuring low signal absorption within human tissue [7].

• **Survey goals and organization:** The data captured in Figure 3, is comprised of the number of articles (including patents), from Google Scholar search results, that have referenced the terms *intra-body communication* and *intra-body networks*. As shown in Figure 3, there is a growing interest in this area of implantable sensors, which makes this survey of various IBC physical layer solutions timely. We compare and contrast several physical layer characteristics to provide: (i) comprehensive tables and figures with metrics from current literature that provide at-a-glance overview of the design constraints of IBC systems, (ii) different scenarios suited to specific IBC schemes, and (iii) a vision of the utility of GC as a robust IBC paradigm that gives best overall trade-off of in performance while addressing the needs of a larger application space. The remainder of this survey paper is organized as follows: Section II introduces important concepts related to intra-body communication and provides an overview of similar body communication surveys. Section III provides the detailed overview of all the IBC methods of interest in this study. Section IV introduces the various limits associated with operating within human tissues. Section V presents information regarding the current power consumption levels that have been measured in ASIC and FPGA implementations. Section VI

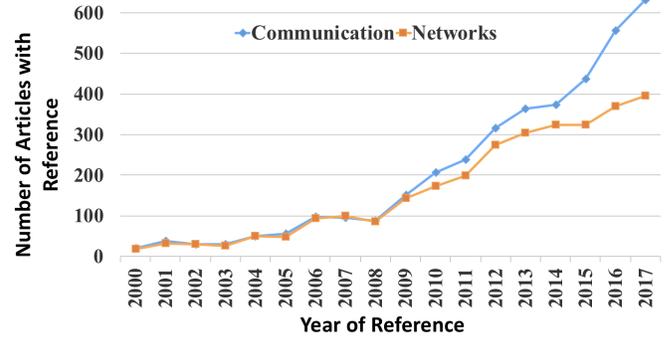


Fig. 3. Number of articles referencing intra-body communication and intra-body networks

details the various channel modeling methods used to study each IBC solution. Section VII provides a detailed comparison of each IBC technique based on the information given in earlier sections. Section VIII delves deeper into the galvanic coupling based approach for IBC, presenting elements on PHY layer system design. Section IX outlines potential applications that could benefit from utilizing GC-IBC. Section X addresses the research challenges associated with system design and application development for a GC-centric approach and finally, section XI concludes the paper.

II. RELATED WORK

Extensive work has been done to illuminate the possibility of utilizing RF and alternative methods for connecting devices within, on and around the human body. In this section, we highlight various surveys and specific contributions in the domain of intra-body communication.

A. Nano Communication Networks

Advancements in the field of nanotechnology have spawned revolutionary scenarios in which an alternative form intra-body communication can be realized. Nanonetworks, or Nano-communication networks, utilize device-to-device communication through interconnected nanomachines within the body. These nano-scale devices are responsible for identifying the presence of harmful agents and delivering drug-induced counter measures without the company of a physician [8]. These devices range in size from hundreds of nanometers to a few micrometers at most and are only bestowed with the capability to perform simple tasks such as computing, storing of data, sensing of biological markers and actuation. [9]. Several communication methods have been proposed for use in nanonetworks, but the most promising and widely studied is molecular communication, where molecules are used to encode, transmit and receive data. Communication via nanomachines function alongside the biological operation of the human body. More specifically, the molecule emission process, molecule diffusion process and receptor binding process, all describe the transmitter, channel and receiver functionality of a single link within a nanonetwork, respectively [10]. As a result, the work conducted in [8], [9],

[10] contribute several channel and system models that are based upon actual phenomena that occur in the biochemical realm within the body. Unfortunately, this research is still in an embryonic stage and although current analysis will be useful in designing a complete system, it has not yet yielded a physical implementation in which performance of a communication link can be quantified. Thus, we do not compare molecular intra-body communication with the other techniques evaluated within this survey.

B. Cognitive Radio for Intra-Body Networks

Cognitive radio (CR) is a highly utilized technology within the wireless network community. It's presence is geared towards addressing the spectrum scarcity problem by supporting techniques for dynamic spectrum access. It's functionality relies on the intelligent enabling of unlicensed secondary users to opportunistically access underutilized licensed frequency bands that are designated for primary users. In the context of Intra-body Networks (IBNs), cognitive radio plays an important role in alleviating the spectrum scarcity problem due to increased deployment of IBNs for medical monitoring and telemedicine related applications. Additionally, it is aimed at reducing interference with neighboring medical devices not necessarily equipped with wireless communication capability. The work conducted in [11] surveys various MAC protocol implementations for CR-IBNs, while [12] surveys a broad classification of CR-IBN contributions, including but not limited to, MAC protocols, spectrum sensing and system architectures. In both the aforementioned works, which specifically target the use of medical applications, it is assumed that sensor-to-relay and relay-to-base station data transfer both employ RF-based methods for intra-body communication. Utilizing RF-based IBC greatly supports the need for CR-IBNs, as the need for EMI mitigation and spectrum efficiency drastically increases when all devices typically operate in the Industrial, Scientific, and Medical radio (ISM) band and RF-bands specified for wireless medical telemetry, medical implants or wearables. Although these recent works focus on the use of CR-IBNs with RF-IBC, the use of Cognitive radio technology can be expanded to other IBC methods whose communication range can extend beyond the body. As more personal IBNs are deployed, body-to-body interference management becomes a problem that cognitive radio aims to solve. Additionally, IBN topologies will continue to leverage the use of an RF-based external gateway node for the transmission of data to remote monitoring centers. However, the operation of many CR related functions (e.g., channel sensing, resource allocation and spectrum mobility) depend on the performance of sophisticated signal processing and upper layer protocols [11]. Therefore, the research surveyed in this work will not focus on examining the use of CR-IBNs in great detail, as the focus is on simple point-to-point physical layer characteristics of each IBC method.

C. Intra-Body Communication Surveys

The exploration of RF-based intra-body communication and the emanation of alternative methods, has spawned several examinations regarding how these techniques can be properly leveraged for wireless communication in the body. The work in [3] defines intra-body networks using a different nomenclature, but the functionality is the same. The authors survey scenarios in which body communication can be invasive or non-invasive, with communication occurring on, in or around the human body. The goal of this survey paper is to provide an overview IBN applications, compare and contrast communication standards that are applicable to intra-body communication (IEEE 802.15.4, IEEE 802.15.6, and Bluetooth Low Energy), and present various findings on current challenges, channel modeling methods, energy efficiency and coexistence. This work mentions the use of Galvanic Coupling, Capacitive Coupling, Ultrasound communication and Molecular communication, but primarily focuses on comparisons for RF-based methods (i.e., narrowband and ultra-wideband). The work in [13], surveys the on-going contributions to mathematical channel models (numerical and analytical) for human body communication for RF-Narrowband, RF-Ultra wideband, Galvanic Coupling and Capacitive Coupling. Additional content introduces comparisons of different IBC transceiver designs, and highlights the current research challenges that need to be addressed. The authors also conduct experiments to uncover additional insight in Capacitive Coupling channel behavior with respect to joint and limb movement. The work presented in the two surveys cited above, compare the largest range of IBC based techniques, while the work in [14], [15], [16] only provide surveyed content for only Capacitive Coupling, mmWave and RF-based methods, respectively. The collective contribution of all the aforementioned work provide great insight into the realm of different intra-body communication techniques. However, a single work does not exist in which data driven metrics are used to compare IBC systems with some form of real-world implementation. Additionally, the analysis of some IBC methods are not accounted for, while others are minimally explored. Therefore, we present this survey as a means to provide a comprehensive overview of the available IBC methods at the physical layer.

III. INTRA-BODY COMMUNICATION METHODS

A. Radio Frequency (RF)

We first discuss the salient features of narrowband (NB) and ultra-wideband (UWB) channels. Here, the RF transceivers, in the form of implanted sensors, communicate by emitting electromagnetic waves in different frequencies, as shown in Figure 4.

- **RF-Narrowband:** RF-NB utilizes bands known as the Wireless Medical Telemetry Service (WMTS) and the Medical Implant Communications Service (MICS) regulated by the Federal Communications Commission [17]. The bands allocated for WMTS are in the 608-614 MHz (i.e., channel 37 in the digital TV band), 1395-1400 MHz (lower-L) and 1429-1432 MHz (upper-L) range [18]. MICS uses the 401-406 MHz

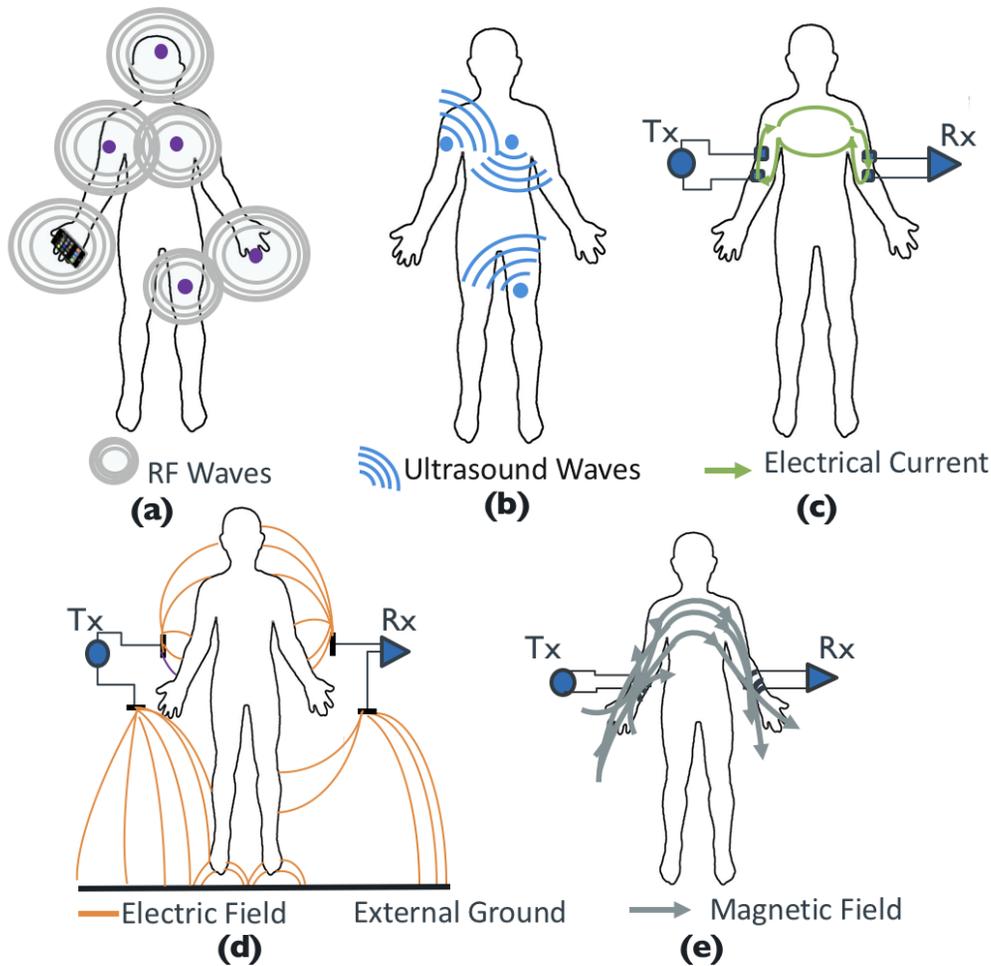


Fig. 4. Conceptual diagram of (a) radio frequency, (b) ultrasound, (c) galvanic coupling, (d) capacitive coupling and (e) resonant coupling IBC

range [19]. The main difference between the two bands is that WMTS is used for remote patient monitoring whereas the main purpose of the MICS band is on-body and embedded sensor communication. All the RF frequency bands used in IBN applications are depicted in Figure 2. MICS has since been renamed as MedRadio and more frequency bands have been added, such as the 2360-2400 MHz band specifically designated for Medical Body Area Networks (MBANs) [19]. More specifically, it is used for low power body sensors networks controlled by a hub device either on or in close proximity to the body. MICS band is used by implanted medical devices for diagnostic and therapeutic purposes, such as cardiac pacemakers, defibrillators as well as neuromuscular stimulators. The bandwidth of the channels range from 100 kHz to 6 MHz and can be used only by authorized health-care providers.

MICS, WMTS and mostly wireless BAN related applications use the IEEE 802.15.6 standard, defining the PHY and MAC layers for specific medical or non-medical applications [20]. The MICS and WMTS bands are preferred over the unlicensed ISM bands due to the reduced interference and the security provided by the regulatory protection of the former.

The FCC recognized this and allocated bands specifically for communication between medical devices, whether implanted or not [17].

RF-NB technology has been used in a variety of implantable medical devices, such as the Zarlink transceiver, which uses the MICS frequency range to communicate at data rates of 200-800 kbps in the body [17]. Additional experimental work in [21] showed data rates of up to 1 Mbps covering a distance of 20 cm in the body. Similar experimental work in the MICS frequency range showed an attenuation of 60-80 dB for 15-20cm links in the body [22].

- **RF-Ultra wideband:** The RF-UWB spectrum utilizes the 3.1-10.6 GHz range. Experimental results have shown that RF-UWB used for communication in the body can reach data rates up to 1 Mbps covering a distance of 12cm, when operating at a frequency of 4 GHz [23]. A consistent characteristic of RF-based IBC is the high signal attenuation levels. Through animal trials, [23] demonstrated that the signal attenuation is above 80 dB through tissues. Even with the high attenuation levels, RF-UWB and RF-NB technology have already been used within implantable pacemakers that wirelessly communicate important diagnostic information, such as cardiac rhythm

and device status (power level) to the physician [17]. The same technology has inspired researchers to develop implantable neuromodulation systems that communicate diagnostic information, such as the pressure of transporting fluids (e.g. blood). There are two major concerns that arise when considering the use of RF, both stemming from power consumption. The high attenuation requires transmission at greater power levels within the tissues compared to an equivalent coverage distance in free space. This not only poses concerns on tissue heating, but also for energy replenishment of the sensors, once they are implanted.

- **Millimeter wave:** Terrestrial fifth generation (5G) mobile networks aim to alleviate the issue of spectrum scarcity by utilizing the large available spectrum in the millimeter wave (mmWave) band, between 30 and 300 GHz. Promising to provide an enormous increase in communication capacity [15], emerging applications may allow device-to-device (D2D) inter-connectivity among wearable electronics that require throughput on the order of gigabits per second (e.g., virtual reality, augmented displays). However, some of the characteristics of mmWave communication (large bandwidth, reasonable isolation and dense deployment) must be balanced with the high propagation loss, need for directivity and sensitivity to blockage/interference. Studies from [24] indicate that although tissue heating can extend to deeper layers of the body, more than 90% of the transmitted electromagnetic power is absorbed within the epidermis and dermis layers of the skin. These characteristics suggest that mmWave communication is difficult to practically implement within implantable devices. However, there is significant opportunity for on-body scenarios. For example, [25] proposes an on-body mmWave channel that spans a link distance of up to 50 cm with associated maximum attenuation values of approximately 80 dB.

B. Ultrasound

Mechanical waves that propagate in frequencies above 20 KHz (the upper limit of human hearing) are also known as ultrasound waves [26]. Ultrasound has been used extensively for underwater communications due to efficient propagation through media composed of mostly water. For this reason, [4] advocates the use of ultrasound for IBC since the human body consists of 65 % water. Ultrasound has been used in medical applications where the transmission of image and telemetry data occurs from a device inside the body to receivers on the outside (e.g., capsule endoscopy). Ultrasound is preferable to RF in this example case of capsule endoscopy due to its higher power efficiency.

Experimental data in [27] reveals that an ultrasound signal with Single Side-Band modulation (SSB) or Double Side-Band modulation (DSB) presents higher frequency and power efficiency than RF signals, through there is an associated trade-off in the low propagation speeds of mechanical waves. The low radiated power allows for a better safety margin, while providing the required image bandwidth for the application. Another phenomenon unique to ultrasound propagation in human tissue, known as cavitation, is the expansion and

contraction of gas bubbles due to the varying pressure of an acoustic field. This health effect must also be considered along with traditional safety limits (to be discussed later in Section III) placed on tissue heating.

A complete prototype for ultrasonic intra-body communication is presented in [28] that achieves a data rate of 90 kbps with a BER of 10^{-6} , and power consumption is at 36 mW. The data rates of 28.12 Mbps achieved in ultrasound appear promising, presently demonstrated through synthetic phantoms [29]. However, experimental data rate at a power of 40 μ W measured at 700 kbps, which is significantly lower [26]. Studies on ultrasounds for communication in the body has shown that attenuation is lower than that of RF communications. At a distance of 20 cm, when operating at 5 MHz, the attenuation is experimentally shown as 25 dB [30], much lower than that for RF. Overall, ultrasound is a promising communication method for IBNs, especially for applications that require high data rate transmission.

C. Capacitive Coupling

Coupling methods in general are based on the energy transfer between a set of transmitters and receivers to generate an electrical signal that propagates through the human body [38]. The electrical signal generated by coupling methods is low-frequency (under 200 MHz) and low-power (in the order of μ W) compared to traditional electromagnetic signals that go up to several GHz. For this reason, coupling methods have become a popular component of the on-going research on IBC since their low power and low frequency complies with safety considerations and decreases energy consumption [38], [13], [39], [40].

Capacitive coupling occurs when two circuits sharing the same electric field cause a flow of energy from one circuit to the other. In the case of intra-body capacitive coupling, the common electric field of the body and its environment causes an induced current flow from a transmitter to a receiver in the form of electrodes. One transmitting and one receiving electrode are attached (or remain in close proximity) to the body, while the other two are floating, acting as ground electrodes. The body acts as a conductor of the electric potential and the ground acts as a return path for the signal [13].

Capacitive coupling has been studied extensively to determine the data rate it can sustain, common levels of channel attenuation and expected levels of power consumption. Recent works on capacitive coupling at 60 MHz exhibits an attenuation of 20-25 dB while covering a distance up to 140 cm in the body [32]. At this operating frequency, the attenuation is lower than that of both RF IBC methods and ultrasound. Additionally, the data rates that have been calculated theoretically in [13], where promising results indicate a maximum rate of 2 Mbps for the 1-200 MHz range. Recent experimental works show a maximum data rate of 60 Mbps by employing a multi-level coded transmission scheme [34], while operating within a bandwidth extending from 40-80 MHz. Overall, capacitive coupling is a promising method for IBC, although its ability

TABLE I
OVERVIEW OF IBC METHODS AND THEIR CHARACTERISTICS

IBC Method	Max. Attenuation [dB]	Max. Link Distance [cm]	Operating Frequency	Data Rate
RF-NB	80 ^[22]	20 ^[22]	401-406 MHz	800 kbps ^[17]
RF-UWB	> 80 ^[23]	12 ^[23]	3.1-10.6 GHz	500 Mbps ^[31]
mmWave	80 ^[25]	50 ^[25]	30-300 GHz	Gbps Range ^[25]
US	100 ^[4]	10 ^[4]	1-100 MHz	28.12 Mbps ^[29]
CC	65 ^[32]	170 ^[33]	100 kHz-120 MHz	60 Mbps ^[34]
GC	65 ^[35]	15 ^[35]	100 kHz-10 MHz	1.56 Mbps ^[36]
RC	35 ^[37]	130 ^[37]	DC to 50 MHz	-

to significantly extend into the surrounding environment can have some impairing factors.

D. Galvanic Coupling

Similar to capacitive coupling, galvanic coupling is a method that uses the human body as a channel to propagate the electrical signal created by a pair of coupled electrodes. The difference between this method and CC, is that the alternating current is coupled inside the body instead of between the body and the environment. On both the transmission and reception side there are two electrodes; an applied voltage between the two transmission electrodes causes flow of an alternating current through the body to be measured differentially at the receiver electrodes. The current propagates due to the ion content of the human body. The two properties of the body that allow the propagation of GC signals are relative permittivity and electrical conductivity [13]. Those properties aid in the definition of channel models used in IBC research to simulate and study galvanic coupling.

Galvanic coupling is a promising IBC method because, similar to CC, it offers low-power and low-frequency signals. The transmission data rates appear to be lower than those of CC but at the same time, there is no need for a floating ground reference or the environment as a path for the signal to traverse [13]. GC signal frequency ranges from as low as 10 kHz [41] to 100 MHz [42] for most effective communication. Experimental work in [43] has shown data rates up to 1.23 Mbps when transmitting at just 200 kHz with attenuation levels typically at 50 dB when covering distances up to 15 cm [35].

Galvanic coupling can be used for communication between skin, muscle and fat tissue and its properties are affected by the tissue layer that is used as a medium as well as the location of the electrodes on the body. Experimental research in [44] has shown that GC communication exhibits less attenuation for muscle-to-muscle links and, in terms of location, the thorax seems to have the best transmission characteristics [40]. Overall, Galvanic Coupling is a safe and efficient method but it is relatively new thrust, with ongoing research on data rate, attenuation, optimal frequency, among others, in order

to achieve the most efficient communication throughout the human body.

E. Resonant Coupling

Resonant Coupling uses the properties of electromagnetic resonance to generate a magnetic field throughout the body. RC is employed to create a near-field wireless transmission of electrical energy between two coils wrapped around parts of the body, driving the field propagation. Its potential benefits arise from low power requirement.

The spectrum range most commonly used in RC research extends from DC up to 50 MHz yielding a maximum attenuation of only 8.1 dB for a 40 cm distance covered [37]. The results of RC are very promising but this area is still in a nascent stage, for e.g., there are no conclusive results for the data rate achieved by communicating through the body using RC. The wavelength of the magnetic field proposed in the work by Park in [37] is 2.3m, which can potentially interfere with the magnetic fields of other nearby devices. Also, the interference of the system from other magnetic fields, including electrical machinery, needs deeper investigation. To summarize the general physical layer characteristics for each IBC method, we present Table I. The data represented in this table is composed of metrics from different sources, for each IBC method, listing the maximum reported values for each metric that provides an indication of link performance. These metrics will play an important role in determining the IBC technologies that will ultimately enable the creation of a network where several applications can be actively executed. Based on application requirements such as data rate, latency, and required BER, the appropriate IBC scheme can be chosen. For example, in [45], applications involving biological signal monitoring (e.g., ECG, EMG, EEG) can require data rates in the range of 70 kbps to 600 kbps with a target BER of 10^{-6} , depending on the measurement configuration. Therefore, subsequent sections of this survey will extensively uncover additional characteristics to provide a thorough analysis of the many IBC techniques available in the research community.

IV. SAFETY CONSIDERATIONS FOR HUMAN TISSUE

The ways in which electromagnetic waves interact and affect human tissue have been widely studied. As a result, several standards such as the IEEE C95.1 standard [46] have been created that set explicit exposure limits in terms of field strength, power density and average time of exposure. Specifically, this standard discusses the limits of signal exposure within the body from the perspective of electrostimulation (in 3 kHz to 5 MHz) and tissue heating (in 100 kHz to 300 GHz). A transition region (0.1 to 5 MHz) set limits that account for both phenomena. Also within this region, tighter bounds are placed on continuous wave signal exposure that have adverse effects associated with heating, while short isolated pulses (with low duty cycles) have more restrictions on health effects related to electrostimulation [46]. The Specific Absorption Rate (SAR—Watt/kg) is a measure of the amount of energy absorbed within a given amount of mass. It is calculated from various field strength values and is used to derive a metric known as Maximum Permissible Exposure (MPE). This quantity describes the maximum rms, peak electric or magnetic field strengths, or power density to which a person may be exposed to without incurring adverse health effects (e.g., tissue heating). This value can be derived or estimated from induced electric field, SAR, or power density values. From the information provided within the IEEE C95.1 standard we calculate values for MPE based on the fundamental properties (wave type and operating frequency) of each IBC method in order to provide an adequate comparison. Table II presents the calculated MPE ($mWatt/cm^2$) values, in an uncontrolled environment, at a corresponding operating frequency for each IBC solution within this survey. This table highlights the differences between the MPE values associated with the IBC methods that employ electric (GC and CC), acoustic (US), magnetic (RC) and electromagnetic-based fields (RF) for intra-body communication. When comparing the various IBC methods, it is important to understand that ultrasonic waves do not undergo the same wave propagation phenomena as the other methods studied in this survey. Thus, it is expressed in terms of Intensity, a measure of the energy absorbed through the transfer of mechanical waves into a medium. The limits of ultrasound intensity are set by the Food and Drug Administration and the American Institute of Ultrasound in Medicine [26] and utilize the same units for quantifying MPE, allowing for a better comparison against the other non-acoustic based approaches in this study.

•**Performance Summary** Results indicate that the ultrasound intensity exhibits at least 360x increase from the highest MPE values represented in Table II. The work presented in [47], attributes this phenomena to the difference in how ultrasonic waves behave. Specifically, the speed of sound in tissue is significantly smaller in comparison to the propagation speed of electromagnetic waves. This difference in velocity yields vast differences between the documented levels of tissue heating. This supports the notion that ultrasound IBC creates the least amount of energy absorption when interacting with the human

TABLE II
MAXIMUM PERMISSIBLE EXPOSURE OF IBC METHODS

IBC Method	Operating Frequency	MPE (mW/cm^2)
RF-UWB	10.6 GHz	1
RF-NB	406 MHz	0.203
US	5 MHz	730
CC	200 MHz	0.2
GC	10 MHz	1.8
RC	50 MHz	2.02
mmWave	30 GHz	1

body medium. Consequently, heat dissipation in the tissue compared to the other approaches based on electromagnetic waves will also be less. The results presented in this and the remaining sections will be comprehensively compared in further detail within section VII, taking into account all of the information surveyed thus far.

V. POWER CONSUMPTION

Powering of biomedical implants enabled with wireless connectivity creates energy-related challenges, with a great emphasis on how to minimize the instances of battery replacement/recharge. As this technology continues to progress, ultra-low power transceiver design plays an intricate role in the ubiquitousness of intra-body networks. The design of these wireless communication systems varies greatly, depending upon which IBC technology is employed and the application that is meant to be utilized. Additionally, obtaining a proper comparison of power consumption from system-to-system is nontrivial. Several factors about the implementation must be compared (see table III) that provide a detailed understanding of which IBC solution has the potential to offer the most reasonable power budget for an implantable operation. This fact also gives way to variance among the power consumption values listed in this survey. Data represented in Figure 5 displays the Energy/bit Vs. Power Consumption values for each IBC method. This table shows a general trend of power consumption related metrics and is comprised of multiple values from several sources.

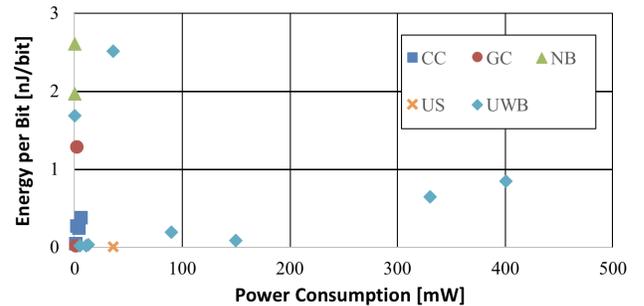


Fig. 5. Energy/bit vs power consumption

TABLE III
COMPARISON OF POWER CONSUMPTION ACROSS IBC METHODS

IBC Method	Power Consumption	Energy/bit [nJ/bit]	Supply Voltage [V]	Modulation Scheme	Operating Frequency	Implementation
RF-NB ^[48]	481 μ W	2.60	0.45	OOK/FSK	402-405 MHz	ASIC
RF-UWB ^[49]	835 μ W	1.67	1.20	FSK	3.328-4.608 GHz	ASIC
US ^[50]	36 mW	70	3.30	PPM	700 kHz	FPGA+MCU
GC ^[36]	2 mW (Tx only)	1.28	3.30	PPM	65 MHz	FPGA
CC ^[45]	4.4 mW	0.24	1.00	FSK	40-120 MHz	ASIC

Table III provides a comparison of the power consumption along with other related metrics, and each row represents data summarized from a specific system implementation in published literature. As of late, implementations have gone beyond classical RF while galvanic and capacitive coupling, as well as ultrasound have begun to emerge. Solutions involving resonant coupling are still in the early stages with limited information on power consumption. While power consumption analysis for mmWave based IBC communication is absent, we expect large antenna arrays needed for beamforming and sampling of signals in the GHz range may contribute towards high power consumption for mmWave communication [51].

A. RF-Ultra wideband

There have been many works related to low power design of UWB systems for body area/intra-body network applications. The work presented in [49], describes a low-power frequency modulated ultra-wideband transmitter developed in 130 nm CMOS technology. The transmitter operates within the frequency range of approximately 3.3 to 4.6 GHz, and consumes 835 μ W of power from a 1.2V supply while achieving an energy efficiency of 1.67 nJ/bit. In [52], a UWB transceiver designed for high data rate for human body communication is developed in 65 nm CMOS technology, and designed for operation between 7.25 and 9.5 GHz. It utilizes OOK modulation with a spectrum efficient frequency hopping technique. Results from this transceiver show that it consumes a total of 13.3 mW from a 1 volt supply, with an energy efficiency of 26.6 pJ/bit.

B. RF-Narrowband

Narrowband transceivers that operate in the MICS band have also been widely studied and developed in ASIC platforms. In [48], combined modulation schemes of OOK (for wake-up and data communication) and FSK (for always-on communication) are implemented in .18 μ m CMOS technology. The wake-up mechanism with OOK modulation is done to reduce power consumption levels, while FSK operation grants greater noise immunity and BER. Power consumption totals in the amount of 481 μ W, with an energy efficiency of 2.6nJ/bit (OOK) and 1 nJ/bit (FSK). The work done in [53], designed specifically for implantable devices, is also constructed in 0.18 μ m CMOS technology and is supplied with .7 Volts for operation. It is specifically configured to mitigate

interference, while employing FSK based modulation. This system is capable of an energy efficiency of 1.96 nJ/bit, while consuming 890 μ W of power.

C. Ultrasound

For ultrasound implant-to-implant connectivity, a hardware and software architecture is presented in [50]. The miniaturized system implementation demonstrates the capability of achieving a total power consumption of 36 mW for both transmit and receive operations. It is comprised of a dual FPGA (responsible for PHY layer functionality) and MCU (data processing and upper-layer protocols) design for simplistic reconfiguration in a small form factor. Utilizing an ultrasonic ultra wide-band pulse position modulation (PPM) technique, performance as low as 70 nJ/bit, with an operating supply of 3.3 V, can be realized.

D. Capacitive Coupling

Capacitive coupling transceiver design for low power communication has been represented on FPGA platforms [54], but have recently shifted towards the ASIC domain, indicating the growth in attractiveness for alternative IBC solutions. One such example is a crystal-less double FSK transceiver and a collection of other FSK based systems that are designed and studied within [45], respectively. This system was implemented in .18 μ m CMOS technology and operated with a 1 volt supply. The transceiver consumes a total of 5.4 mW with an energy efficiency of 3.2 nJ/bit. A more recent study on front end design for a capacitive coupling receiver, detailed in [55] is developed in 65 nm CMOS technology and is designed to be suitable for data rates on the order of kbps, operated with a supply of 1.2 volts, with an active power consumption of approximately 250 μ W.

E. Galvanic Coupling

For GC based systems, the most recent works have been implemented on platforms with an FPGA at its core [56]. The work proposed in [36] also consists of an FPGA-based system and implements a pulse position modulation (PPM) scheme, similar to an impulse radio ultra-wideband technique. Results demonstrate that the transmitter system power consumption is 2.0 mW with a 3.3 volt supply, and an energy efficiency of approximately 1.28 nJ/bit. This design also achieves a data rate for GC-IBC in the range of 1.56 Mb/s. Having energy

efficiency as low as 1.28 nJ/bit provides the potential for enhanced solutions where portable biomedical applications can be applied.

•**Performance Summary** Initial observations of the power consumption values listed in table III indicate that RF-based IBC methods provide better levels to that of Capacitive Coupling (CC), for ASIC implementations. However, the ASIC implementation of the CC system achieves significantly lower levels of energy/bit. Additionally, the Galvanic Coupling system, despite being implemented on a less power efficient FPGA, also achieves similar levels in the realm of energy/bit. These results confirm the improved link efficiency, leading to longer device lifetimes, when transmitting within or on human tissues for non RF-based methods. These results also support the data represented in Figure 5, where the majority of non-RF implementations achieve lower energy per bit values compared to their counterparts.

VI. CHANNEL MODELING METHODS

Signals emitted by each intra-body communication technology undergo different channel-induced transformations while propagating on or within the human body. In this section, we provide context into the common channel modeling techniques employed for uncovering signal propagation behavior in human tissues. One representation of this, Table IV, depicts the occurrence of several types of channel modeling methods with respect to each IBC technology. We also present Table V, in order to compare channel characteristics and provide the necessary data for designing transceiver hardware to mitigate various channel impairments. The data for each IBC originates from a particular source, while the propagation speed values are calculated based how the speed of light and sound change in a human body medium. We next discuss techniques under the classification of Analytical, Empirical, Numerical, Statistical, and Hybrid (Analytical + Numerical) methods, for the human body channel under the influence of each IBC method.

A. RF-Ultra wideband

RF-ultra wideband channel for the human body medium in [58] is modeled analytically by finding closed form expressions via Maxwell's Equations inside a homogeneous medium. This method captures similar results to the work done in the numerical domain, such as [71], where the dielectric nature of human tissue is considered. In this work, Finite Difference Time Domain analysis is done to gain a better understanding of the intra-body communication from a node embedded in the chest to an external base-station like device. The characterization work presented in [23] evaluates and validates the performance of liquid based phantom (similar to muscle tissue) trials by conducting experimental modeling on a live animal (pig) where nodes are surgically implanted. The overall analysis of channel behavior is consistent among the other works presented in this subsection. Other results indicate that multipath fading is present, but due to high path loss beyond a certain range, its effect presents little to no impact on the performance of signal propagation. To exemplify this, the

work done in [23] indicates that the relatively short multipath delay spread can be approximated using a two-ray propagation model. Additionally, blood and muscle tissues contribute to higher path loss when compared to bone and fat layers within the body.

B. RF-narrowband

Channel modeling RF-narrowband is conducted via hybrid, empirical and numerical based methods. In [21] a path loss model for multi-layered homogeneous tissue layers within a heterogeneous body is presented. Finite Difference Time Domain approach is used in conjunction with equations based on Friis transmission using while accounting for the dielectric nature in tissues. Empirically, work has been done in [66] to study the path loss and channel characteristics for nodes implanted within a single tissue layer. A ray-tracing based approach to channel modeling was performed on a homogeneous tissue phantom, also with representative dielectric properties. The Finite Difference Time Domain numerical approach is presented in [22], where the RF antenna is also modeled as a part of the channel, for a simplified homogeneous 3-layered tissue structure. Channel behavior uncovered from the hybrid modeling method, details that log normal shadowing is present in the channel and that an increase in tissue conductivity yields degradation of the path loss (similar to RF-UWB).

C. mmWave

In crowded environments such as train cars or airline cabins, human bodies become significant sources of blockage in the mmWave frequencies. Self-blocking can occur where the user's body location results in restricted access to the local network in addition to causing unwanted interference for other wearable networks. A possible network architecture assumes the user's smartphone to function as a gateway hub, and the neighboring wearable networks associated with different users are likely uncoordinated [72]. Thus, techniques involving stochastic geometry are used to model and analyze the performance of these networks with a finite number of interferers in a bounded network region. These models incorporate different path-loss and small-scale fading parameters, depending on the state of the link.

Current models for on or near the human body operation describe the propagation characteristics as quasi-optical. [25] presents an empirical and statistical approach at modeling the mmWave on-body channel. From the empirical model, the separation of short-term and long-term fading, occurs by experimentally employing a temporal averaging window of the order of 0.11 seconds. The data gathered from the body-worn setup is also used to facilitate statistical analysis, uncovering that the short-term fading best fits the Cauchy-Lorenz distribution. In [59] an analytical approach, validated through numerical simulation is used to uncover insight about the overall system performance (e.g., spectral efficiency, SINR coverage probability, blockage probability) in the presence of multiple interfering agents. In this study, the Nakagami distribution is assumed and the human body itself is modeled

TABLE IV
COMMONLY STUDIED CHANNEL MODELING METHODS

Modeling Method	RF-UWB	RF-NB	mmWave	US	GC	CC	RC
Statistical			[25]	[57][30]			
Analytical	[58]		[59]	[60]	[44][61][62]	[63]	[64][65]
Empirical	[23][66]	[66]	[25]	[57][26]	[61][67][62][40]	[54][68][69][70]	[37]
Numerical	[71]	[22]	[72]	[60]	[44][61]	[70][35]	[6][37]
Hybrid		[21]				[63][32]	

as a source of link blockage. Despite the recent work related to on-body mmWave channel modeling, significant work is needed to quantify signal transmission characteristics and incorporate the results into practical systems design.

D. Ultrasound

The work in [57] uses an ultrasonic testbed to empirically measure the channel. This FPGA-based prototype, with a kidney phantom as the medium for propagation, is supplemented with a statistical model to characterize the behavior of interference. The model uses an approach dubbed as the M-sampling method, where the interference at the receiver is characterized by taking multiple sample sets at various instances of time within pre-specified intervals. In this work, the channel behavior takes on the form of the generalized Nakagami probability distribution. In order to design proper analytical models that represent ultrasound propagation in various media, [60], applies acoustic wave equations (Helmholtz equations) in order to characterize physical parameters such as pressure of sound waves, amplitude, propagation speed, wave intensity, etc. Numerical methods are further employed to satisfy the need to represent acoustic field behavior in time and space throughout the human body. The work in [60] utilizes what is known as the pseudo-spectral (PS) approach that exploits Fourier series expansions, FFTs and the k-space method (an approximation method to obtain temporal derivatives) to perform operations in the spatial and time domain, respectively. Results from the previously mentioned works, in addition to the research presented in [30], show that the general ultrasonic human body channel undergoes small scale fading and slow propagation speeds, resulting in large multi-path delay spreads. This phenomenon is due to the inhomogeneity of tissue in terms of density and sound velocity, where numerous reflectors and scattering surfaces exist. The channel characteristics of ultrasound inhibit the use traditional passband modulation or continuous wave transmission due to their poor performance in multipath channels. Thus, systems that use pulse based low duty cycled communication or OFDM are examples of desirable solutions. However, OFDM requires high system complexity and yields a peak-to-average power ratio that requires inefficient linear power amplifiers, a less ideal solution where strict power budgets exist.

E. Capacitive Coupling

Capacitive coupling channel characterization uses circuit models to drive analytical expressions [63] and different subjects to test human body variation during empirical modeling [54]. The work in [63] suggests that numerical models do not accurately account for the electrode-tissue contact impedance, an important factor used in estimation of signal propagation loss. Thus, the authors utilize a hybrid channel model, where the combined approach of an analytical electrostatic circuit model and numerical simulation of 3D human body geometries under electromagnetic fields is utilized. Each approach accounts for the frequency dependent dielectric properties of human tissue, and reasonable tissue dimensions. This Finite Element Modeling based approach is used to study path loss under different body positions and environments while validating the data from other literature. Capacitive coupling signal propagation characteristics can vary under different environmental conditions, human body posture, and if external interferers are present. Channel gain versus frequency plots provide evidence that frequency selectivity is present within the CC channel, indicating the possibility of multipath fading. However, no additional information is provided (e.g., phase variation, channel impulse response, etc.), making it difficult to assess the severity of the fading. Additionally, capacitive coupling based communication places tight requirements on the design of ground and signal electrodes and shows high variability in path loss measurement with different system configurations [63].

F. Galvanic Coupling

Similar to the methods done with CC, [44] proposes an analytical model in the form of a tissue equivalent circuit channel for galvanic coupling. The three-dimensional multi-layered structure accounts for various changes in physiological factors such as, frequency dependent dielectric properties, hydration levels and tissue thickness dimension. Validation of the analytical model was performed using ANSYS HFSS, enabling full-wave electromagnetic simulations, numerically analyzing the electric field distribution at user defined locations within the tissues using finite element analysis (FEA). The work in [67] designs a correlative channel sounding experiment to empirically uncover the channel impulse and frequency response of transmission and reception within the skin and muscle layers of a porcine tissue specimen. Other

TABLE V
CHANNEL CHARACTERISTICS COMPARISON OF IBC METHODS

IBC Method	Channel Modeling Method	Noise Sources	RMS Delay Spread [s]	Propagation Speed [m/s]	Channel Characteristics
RF-NB ^[21]	Analytical	Thermal Noise Ext. Interference	$\sim 10^{-9}$	$\sim 10^7$	Log normal Shadowing
RF-UWB ^[66]	Empirical	Thermal Noise Ext. Interference	$\sim 10^{-9}$	$\sim 10^7$	Two-Ray Propagation
mmWave ^{[16][25]}	Statistical	Thermal Noise Ext. Interference	$\sim 10^{-9}$	$\sim 10^8$	Cauchy-Lorenz Fading
US ^[57]	Statistical	Thermal Noise	$\sim 10^{-6}$	$\sim 10^3$	Nakagami Fading
GC ^[67]	Empirical	Thermal Noise	$\sim 10^{-9}$	$\sim 10^7$	AWGN
CC ^[54]	Empirical	Thermal Noise Ext. Interference	$\sim 10^{-9}$	$\sim 10^7$	Frequency Selective
RC ^[37]	Empirical	Ext. Interference	-	$\sim 10^7$	Frequency Selective

channel characterization work within [62] and [61] used similar methods at numerical, analytical and empirical modeling to capture the channel behavior, where FEM, tissue equivalent circuits, and porcine tissue or human forearms, were used respectively. Galvanic coupling also adds an inherent benefit of additional security and interference resilience. GC-signals are confined within the body cannot be intercepted unless one is in direct contact with the medium. The results from GC channel modeling indicate that the biological channel behavior exhibits no multi-path fading, and thus, can be considered equivalent to an AWGN channel [67].

G. Resonant Coupling

Recent work in the area of magnetic resonant coupling modeling of the body channel is mostly in the numerical and empirical domains. The channel studies in [37] and [6] use Finite Element Modeling (via Ansys HFSS) and Finite Difference Time Domain numerical analysis, respectively. These numerical approaches take advantage of mesh models that account for the frequency-dependent dielectric properties and geometry of human tissue. Results indicate that path loss of RC is very low within human tissue, due to low values of tissue magnetic permeability. This concept shows that magnetic fields can freely move throughout the human body. However, channel gain characteristics presented in [37] indicate frequency selectivity that is highly dependent on node placement and body posture. The limited channel modeling data for RC allows for partial completion of the field entries in Table V.

•**Performance Summary** Despite the vast array of methods at one's disposal to characterize channel behavior, the majority of works invoke the use of a computational based method, supplemented with an empirical validation. Additionally, it is

most convenient to classify the channel behavior with terminology frequently expressed in the wireless communication community, as it provides great insight into the complexity of the transceiver design. Table V indicates that Galvanic coupling and RF-UWB/NB offer the most simplistic channel characteristics, while methods such as Ultrasound have more complex channels.

VII. COMPREHENSIVE COMPARISON OF IBC METHODS

We compare the IBC methods in terms of attenuation, link distance, achievable data rates, power consumption, channel impairments, tissue safety, among other parameters, using Figures 6 and 7. The data within these figures represent a range of values surveyed from multiple theoretical and experimental findings throughout literature.

Although RF-UWB and RF-NB waves have been studied extensively, there are several issues that limit their deployment. RF-UWB requires high energy consumption of 2.6 nJ/bit [36]. In comparison, coupling methods have energy consumption as low as 0.24 nJ/bit. RF operates in high frequencies, 401-406 MHz for NB, the Industrial, Scientific and Medical radio ISM band (2.54 GHz) and 3.1-10.6 GHz for UWB. The health risks related to the high frequency of RF waves in the body are a concern due to tissue-heating caused by the high absorption of the wave-energy [4]. Hence, the MPE of RF methods is lower than that of ultrasound and other coupling methods. In addition, security intrusions and susceptibility to interception of information transmitted using RF through the body become possible due to the fact that the waves propagate both in-body and off-body, into the environment [4].

The emerging mmWave technology presents challenges when it comes to application on WBANs. In order to fully realize the design constraints for wearables, accurate channel

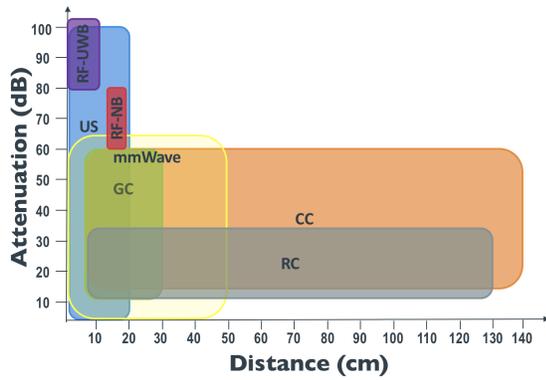


Fig. 6. Comparison of attenuation values of all IBC methods in terms of distance covered by communication link

models are needed. Models that account for the impact of reflections within the finite bounded region and mitigate the complexity involved in boundary value problems relating to Maxwell’s equations, will be a foundational step [25]. At higher frequencies, energy absorption is increasingly confined to the surface layers of the skin. At frequencies beyond 60 GHz, surface waves play no significant role in propagating over human skin and the generated fields are largely attributed to space waves. Penetration depths at the 60 GHz range are in the order of millimeters, and thus, only the dielectric properties related to the skin need to be considered for modeling scenarios where person-to-person body variability needs to be studied [25].

Resonant coupling is the least studied method out of all IBC technologies with extensive research underway to fully understand the channel characterization metrics. Though RC is safe for the human body in terms of tissue heating, the interference of the magnetic field created around the body with other magnetic fields in the environment is still an issue that may affect the communication link. Capacitive Coupling, though effective at providing long range IBC and sufficient data rates, does not solve some of the issues with RF and RC-IBC. As mentioned in section III-C, CC depends on the ground reference to create a capacitive link that extends outside the human body, thereby increasing its interference domain [35]. This level of susceptibility makes capacitive coupling less preferable for IBC, especially as more users are equipped with their own IBN/BANs.

Although high data rates systems have been achieved using ultrasound in the body, practical rates are in the high kbps range [28]. Ultrasound waves with low duty cycles can cover sufficiently long distances in the body without exposing tissues to high temperature. Ultrasound-UWB techniques mitigate the channel impairments and medium access control algorithms built on top of its physical layer solve the interference problem for ultrasound in the presence of multiple users. [4].

Galvanic coupling offers moderate transmission distances and lower data rate compared to methods such as CC, US and RF-NB and can safely operate with relatively high limits within the human body. Its true advantages are low attenua-

	Power Consumption	Propagation Distance	Transceiver Complexity	Interference Susceptibility	Data Rate	Tissue Safety	Security
RF-NB	●○○	●●○	●●○	●●○	●○○	●○○	●○○
RF-UWB	●○○	●○○	●●○	●●○	●●●	●●○	●●○
US	●●○	●○○	●●○	●○○	●●●	●●●	●●●
GC	○○○	●●○	○○○	○○○	○○○	●●●	●●●
CC	○○○	●●●	●●○	●●●	●●○	●●○	●○○
RC	○○○	●●○	○○○	●●○	○○○	●●●	●○○
mmWave	●●●	●●○	●●●	●●●	●●●	●○○	●○○

Fig. 7. Qualitative comparison of design constraints for all IBC methods

tion and full confinement of signals inside the human body, offering more security and interference-free communication. The simplistic GC-IBC communication channel allows for low complexity in transceiver design, resulting in lower power consumption. Though the data rates are lower compared to other IBC methods, they are still suitable for many telemedicine applications [45].

•**Performance Summary** We observe that no specific IBC is solely superior to the other options for all performance metrics. Each IBC technique offers its own unique set of trade-offs that influence specific application choices. For example, in Figure 6, use cases that require short link distances (>10 cm), and on-skin propagation of data, can essentially leverage any of the techniques studied in this survey. However, if the above mentioned scenario is required to transmit at extremely high data rates (exemplified in Figure 7), design choices utilizing CC, US and RF-UWB may be the most suitable options. Based on the cumulative information presented in Figures 6 and 7, we delve deeper into the investigation of Galvanic Coupling as an enabling method for IBC. We select this method for its performance in the areas of power consumption, tissue safety, security, and transceiver complexity. Subsequent sections in this survey represent specific use cases in which we exploit the benefit of the GC-channel and its fundamental properties. Thus, we present selected GC-systems and applications in Sec. VIII and Sec. IX, respectively.

VIII. GALVANIC COUPLING SYSTEM DESIGN

Creating sustainable IBC-based applications requires careful design of the physical layer link. This section explains the setup of a unidirectional communication link through a synthetic human tissue phantom using the GC-concept [56] and the transceiver architecture needed for its implementation. Towards this end, we use the hardware support package of the Communications System Toolbox in MATLAB to demonstrate the effect of various modulation schemes on a prototype testbed.

We use the knowledge of the analytical channel behavior to aid in the design of experimental testbeds that utilize GC communication. Thus, prior to setting up the testbed, we examine the equations in literature that are used to calculate the gain and phase shift of the attenuated galvanic coupling signals. As mentioned in III, Galvanic Coupling uses two transmitting and two receiving electrodes for intra-body communication. A

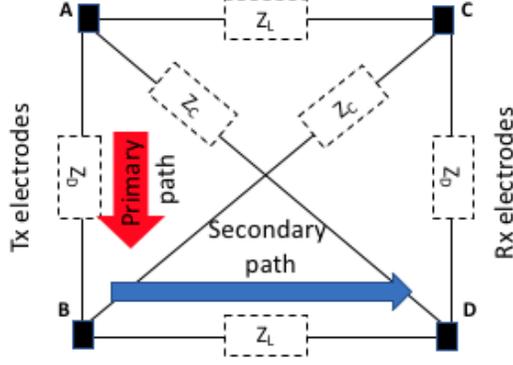


Fig. 8. Tissue equivalent circuit of one layer and flow of electric current in Galvanic Coupling

visual representation of the signal propagations in the body can be seen in Figure 8. The secondary flow of current is the one utilized for intra-body communication, in which the potential difference at the receiver electrodes is measured. In order to calculate the gain and phase offset between the transmitted and received signal, we require knowledge of the propagation path in the form of a tissue equivalent model such as the one in [44]. The model presents the impedance of the tissue path in the form of an admittance matrix A . The admittance matrix is created based on the electrode placement, distances and dielectric properties of the tissue. The matrix contains impedance values for each possible path, from any node on the tissue equivalent circuit to the others. For the example in Figure 8, the admittance matrix is:

$$A = \begin{bmatrix} A_{tot} & -\frac{1}{Z_D} & -\frac{1}{Z_L} & -\frac{1}{Z_C} \\ -\frac{1}{Z_D} & A_{tot} & -\frac{1}{Z_C} & -\frac{1}{Z_L} \\ -\frac{1}{Z_L} & -\frac{1}{Z_C} & A_{tot} & -\frac{1}{Z_D} \\ -\frac{1}{Z_C} & -\frac{1}{Z_L} & -\frac{1}{Z_D} & A_{tot} \end{bmatrix} \quad (1)$$

where $A_{tot} = \sum_{i \in \{C, L, D\}} \frac{1}{Z_i}$. The summation of the current in nodes A, B, C and D on the circuit from Figure 8 are calculated as follows:

$$\vec{I} = A \cdot \vec{V} \quad (2)$$

\vec{V} is the voltage vector with the voltage values at each node and \vec{I} is the current through each node. As an example, the current equation for node A is the following:

$$I_A = \frac{V_A - V_B}{Z_D} + \frac{V_A - V_C}{Z_L} + \frac{V_A - V_D}{Z_C} \quad (3)$$

Knowing the voltage at each point of the path, the following equation is used to calculate the attenuation of the signal in the path,

$$G = 20 \cdot \log_{10} \left| \frac{V_o}{V_i} \right| \quad (4)$$

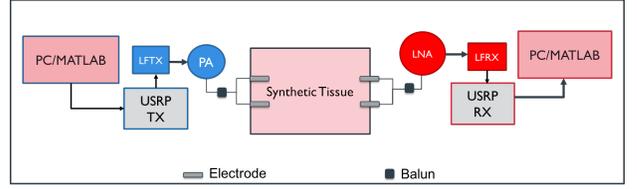


Fig. 9. Block diagram of GC-IBC testbed

where V_o is the voltage at the selected output (between nodes C and D) and V_i is the voltage across the transmitting electrodes (between nodes A and B). The gain depends on several factors such as the distance between the transmitting and receiving electrodes, the inter-electrode distance of the transmitter and receiver, electrode orientation and more. Additionally, the phase shift of the signal across the path is calculated below:

$$Phase = \arctan \left(\frac{Im(V_o)}{Re(V_o)} \right) \quad (5)$$

A. GC-IBC Experimental Testbed Architecture

The USRP N210 by Ettus ResearchTM is used as the Software Defined Radio platform to emulate an implanted sensor. As shown in the GC-IBC testbed block diagram in Figure 9 and the physical representation of the testbed in Figure 10, we use USRPs to form a communication link, which are in turn connected to host laptops.

The communication system employs PSK-family of modulation schemes and accommodates data rates up to 200 kbps. Elements of the communication system include, but are not limited to, bit generation, preamble insertion, raised cosine filtering and baseband modulation on the transmitter side (USRP Tx). In the receiver chain, components such as automatic gain control, phase and frequency offset compensation, and baseband demodulation exist. Within the USRP, low frequency daughter boards, the LFTX and LFRX, are used for the transmitter and receiver, respectively. The daughter boards operate within the frequency range of DC to 30 MHz, well within the permissible frequency band used for GC-based communication. However, the LFTX and LFRX have almost no internal gain, thus requiring signal amplification. We use external amplifiers (provided by MiniCircuits[®]) to account for the lossy nature of the channel and the limited hardware capability. Integrated with the Tx is the ZFL-500+, an SMA connector based Power Amplifier (PA) that has a maximum power output of 9 dBm. On the Rx side, a Low Noise Amplifier (LNA), the ZFL-1000LN+ is used, consisting of a noise figure of 2.9 dB.

The human body equivalent channel (see Figure 10) is provided by SynDaver LabsTM. This synthetic tissue is composed of salt, water and fiber to approximate the dielectric properties of actual human tissue. Bridging the connection between the channel and USRP, balun circuits (Schaffner IT239) isolate the common ground return paths of the transmitter and receiver, and perform single-ended to differential signal conversion.

TABLE VI
GC-IBC PHY LAYER PERFORMANCE COMPARISON

System Architecture	Occupied Bandwidth	Minimum Transmit Power	Maximum Bit rate	Modulation Order
BFSK	621.7 kHz	-15 dBm	200 kbps	2
BPM	587 kHz	-21 dBm	200 kbps	2

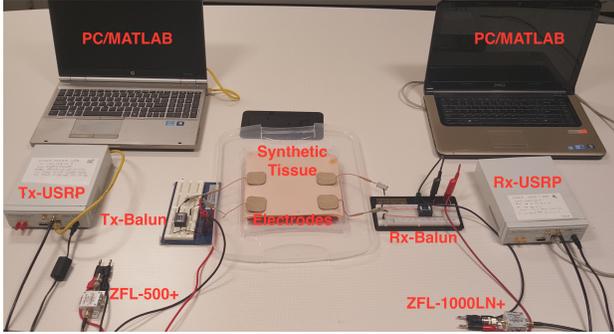


Fig. 10. Physical setup of GC-IBC testbed

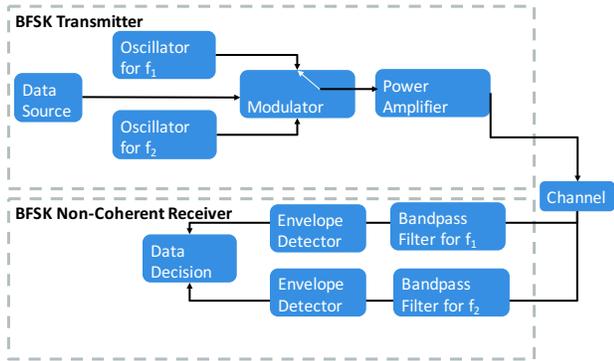


Fig. 11. Block diagram of the binary fsk transmitter and receiver

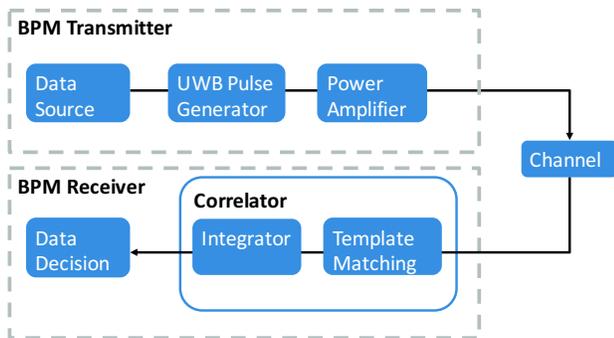


Fig. 12. Block diagram of bi-phase modulation transmitter and receiver

The electric current injection for signal transmission is made possible via electrodes provided by TENSPros.

B. GC-IBC PHY Layer Design

The GC human body channel is approximated as AWGN for a narrowband channel based on the previous channel sounding

studies in [67].

Specifically, the analysis in this section is performed with respect to Binary Frequency Shift Keying and Binary Phase Modulation (an ultra wideband modulation scheme) whose block diagrams are represented in Figure 11 and Figure 12, respectively. Similar experimental platforms utilizing GC-communication systems have been explored in [43], [13], [7]. The work in [43] employs Differential binary phase-shift-keying (DBPSK) modulation for its resiliency to amplitude variations and low hardware complexity. The motivation for UWB-based architecture stems from the work conducted in [13] and [7]. Both works follow the specifications outlined in the outlined in IEEE 802.15.6 standard for Wireless Body Area Networks. The former makes strides on the physical layer, while the latter focuses on a multiple access implementation. The physical layer design depicts a carrier-free, pulse position modulation (PPM) transmitter built upon an FPGA platform. These studies have shown promising results for GC communication system design

- **Binary Frequency Shift Keying:** We consider non-coherent Binary frequency shift keying (BFSK) due to its high power efficiency and ease of implementation. The number of independent frequency oscillators needed for this type modulation is proportional to the number of bits allocated per symbol, and therefore it consumes more power as the number of bits increases. For non-coherent (signal detection without the need for a voltage control oscillator) configurations of FSK, the complexity and the power consumption of the receiver can be reduced. In addition to this, FSK modulations have a constant envelope, allowing systems to utilize efficient non-linear power amplifiers. We restrict our current implementation of FSK to a modulation order of 2 ($M = 2$), as higher values of M in FSK are not considered due to increased system complexity and bandwidth usage [73]. The transmitter consists of a standard two-oscillator setup, where the modulator toggles the carrier frequency based on the data to be sent. The receiver architecture filters and detects the signal directly at the carrier frequency values, eliminating the need to translate the signal to baseband.

- **Bi-Phase Modulation (UWB):** The 3dB fractional bandwidth of the GC-channel exceeds 25 percent, classifying it as ultra-wideband, as specified by the FCC. Thus, we exploit the benefits of UWB systems (e.g., high data rate and low system complexity) and those afforded by nature of galvanic coupling. One example of a typical signal that can be considered for UWB transmission is the Gaussian pulse and its more commonly used higher order derivatives. Each

pulse occupies an ultra wide spectral mask in the frequency domain. To produce the desired spectrum allocation, the order of the Gaussian derivative and pulse width of the time domain signal can be altered. This type of transmission also does not require the use of additional carrier modulation [74], and is considered a baseband approach. Bi-phase modulation uses UWB pulses to represent bits by altering their polarity. We chose Bi-phase modulation (BPM) due to its simplistic design and its ability to be less prone to channel distortion (the difference between the positive and negative pulse amplitudes are twice the amplitude of an individual pulse [75]). One can detect Bi-phase modulated pulses using either an energy detection or template matching demodulation in the receiver. In this work, we implement the latter, using the first derivative of the Gaussian pulse and a $5 \mu\text{s}$ pulse duration to evaluate its performance against the aforementioned BFSK approach.

C. Performance Comparison

We simulated each of the above architectures within the MATLAB environment, using the stored channel impulse response obtained from the empirical channel characterization work conducted in [67]. Performance metrics are listed in table VI for a target BER of 10^{-4} . To compare each type, the symbol rate was fixed to a value of 200 kbps. Results indicate that the UWB-BPM system offers an improvement in terms of power efficiency and bandwidth efficiency, as it is able to achieve the same target BER while occupying less spectrum, and transmitting with less power. These characteristics reveal that the BPM transceiver system has the better potential to operate as the PHY layer for integration into the example applications presented in the previous section.

IX. APPLICATIONS

We present a series of application scenarios using different characteristic features of GC-IBNs, with different operational requirements.

A. Body-Generated Password

Securing personal information is a growing concern due to an increase in password protected mobile devices/services at our disposal. Malicious attempts to crack these secure passwords allow access to sensitive information, thereby compromising the entire system. Biometric authentication techniques exist to provide enhanced forms of security (e.g., to the fingerprint ID on smartphones), but are also being challenged by unique ways of exploitation. Recalling that the galvanic coupling IBC technology is energy efficient and robust against external interference, we envision applying the GC technology to specific contributions in the area of password replacement.

The propagation characteristics of galvanic coupling (signal transmission confined within the body) and unique biological markers generated by the body (e.g., electrocardiogram signals), can be combined to create an alternative method of securing our personal devices. The generation of GC-signals directly within human tissue would yield communication links that are immune to vulnerabilities such as eavesdropping

and spoofing. The desired use case, illustrated in Figure 13, requires the subject to simply wear an on-body device, preferably a wrist-worn band. Upon direct contact, the unique biological identifier is recorded, processed and communicated from within the body to a receiver located on the device of interest. Galvanic coupling, illustrated into the example listed above, provides a better understanding of how the IBC method can be applied to combat real-world, challenging problems.

B. Wireless Neural Stimulation and Monitoring

Electric medicine, or the use of therapeutic electric stimulation in health care applications, is a rapidly growing field. Deep brain stimulation for Parkinson's disease, cochlear and retinal implants, epidermal stimulation for spinal cord injuries, and vagus nerve stimulation for epileptic seizures are just a few recent applications of electric medicine. Neural stimulation and monitoring can greatly benefit from technological improvements in discovery platforms and clinical applications: biocompatibility/mechanics, invasiveness, resolution, durability and power demands. The use of intra-body communication can directly address some of those improvements in a system utilizing neural stimulation using a combination of high specificity electrodes communicating directly with the nerves in the brain and with a relay node that collects information to be delivered outside the body.

Designing and implementing the neural stimulation and communication system presented in Figure 14 can resolve some of the technical issues involved in electrical medicine. We propose a minimally invasive approach, since the implanted electrodes will not need frequent changes due to the low power consumption of galvanic coupled communication systems. The relay node can be implantable or placed on the surface of the body, depending on the level of invasiveness desired in each use case. Furthermore, the use of an array of electrodes and beamforming approach designed specifically for use within the brain channel can increase the resolution of data communicated to and from the relay node to the nerves. For the intra-body communication link between the electrodes and the relay node, [76], devises a distributed beamforming approach that allows coordinated transmissions from the implants/electrodes to the on-skin relay node as well as target remote sections of the brain tissues without direct electrode contact. Additionally, the same electrode elements can serve a multitude of purposes; communication, sensing and neural stimulation. In order to examine the behavior of the brain tissue as a channel for galvanic coupling, the typical skin-fat-muscle-bone model needs to be designed, and tailored to mimic the various layers within the brain. The other forms of galvanic coupling channel modeling methods mentioned in this study can also be used to characterize the human brain medium. Overall, using galvanic coupling as a communication link within a system that already utilizes electrodes demonstrates the unlimited capabilities to improve applications associated with electric medicine.

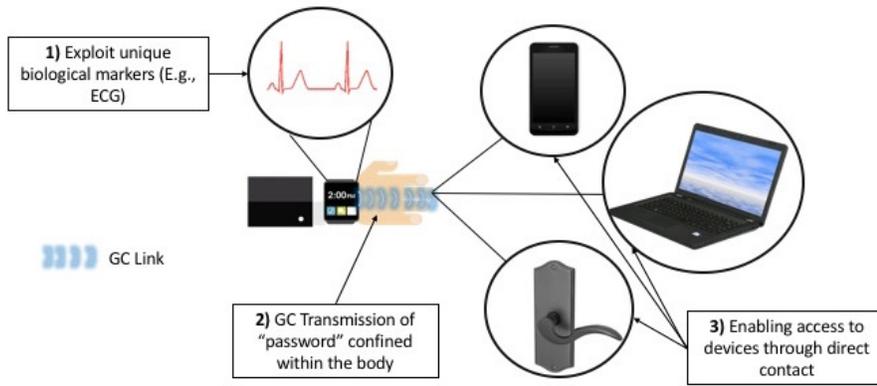


Fig. 13. Conceptual diagram of body-generated password using galvanic coupling

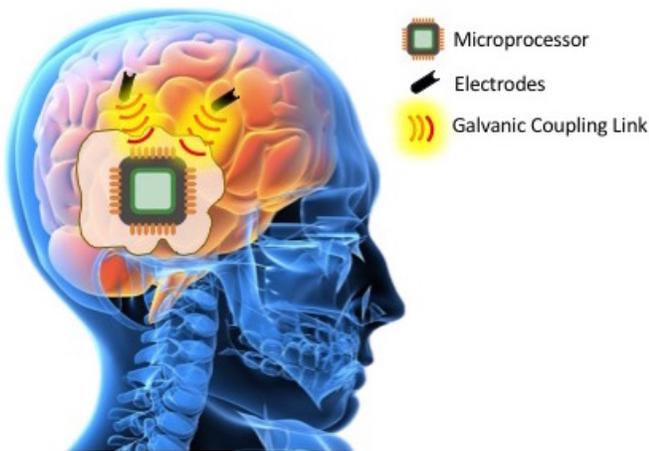


Fig. 14. Conceptual diagram of neural stimulation and monitoring system using galvanic coupling and radio frequency

C. Implantable Continuous Blood Glucose Management System

Intra-body Communication also has the potential to revolutionize treatment mechanisms for traditional diseases such as diabetes. Diabetes is a common health issue that has affected more than 29.1 million Americans [77]. Diabetes is caused by lack of insulin secretion in body (Type 1) or the body not being able to properly use the secreted insulin (Type 2).

The glucose concentration in a diabetic patient can be maintained close to the non-diabetic range by injecting the right amount of insulin into the body whenever required. The typical diabetic treatment option is stressful and tiresome [78], [79]; the patient tests the blood glucose level periodically with a glucose meter using the blood samples obtained from finger pricks. Current glucose levels are used to estimate the right dosage of insulin to be injected. This method of insulin calculation is cumbersome, inhibiting positive user experiences. Even if correct calculations are made, the chances of hyperglycemia (excessive glucose) and hypoglycemia (diminished glucose) persists in most people, especially with Type 1 diabetes.

The continuous glucose monitor (CGM) [80] has a sensor that captures the real-time blood glucose levels. The sensor then transmits the reading to the patient's mobile device, thus freeing the patient from frequent finger pricking. However, the patient still has to calculate the correct dosage of insulin for a given blood glucose value and operate the insulin pump periodically.

Automatic insulin pumps calculate the insulin quantity autonomously, thus avoiding human error. In the bionic pancreas, the insulin pump runs an application in the patient's mobile device to make decisions on the amount of insulin to be injected, based on the CGM values transferred to the device by the sensor [81]. Though beneficial, this technique forces the patient to always carry a mobile device as both the CGM and insulin pump communicate via the device. In this scenario, a mobile phone turning off from battery depletion would be catastrophic.

Galvanic coupling can provide an improvement to the current system limitations. Currently, the CGM is an inconvenient add-on to the patient's body: the sensor has a needle inserted subcutaneously with an on-skin transmitter, a set-up that is prone to infection and requires to be replaced every 2-3 days. Apart from showing a large patch on body, the patient can experience itching and bruising at the site of insertion. This inconvenience to the patient is expected to last throughout the entire usage of this system. To reduce this hardship by half, [82] and [83] propose implanting the sensor along with the transmitter subcutaneously, avoiding any on-surface sensing component. An implanted sensor enables accurate more recent glucose values and the sensors are replaceable. They can be implanted with minimally invasive techniques, as shown in Figure 15. Here, the sensor communicates from inside the body to a receiver embedded in the pump attached on the skin. For such intra-body communication, the RF-based techniques that are most suitable for the over the air communication, prove to be unsafe owing to high amounts of RF signal absorption by tissues. As an alternative, GC uses less energy for communication than other peer techniques. A significant enabling feature is a multi-parameter closed-loop diabetes management system that uses the observations of several other

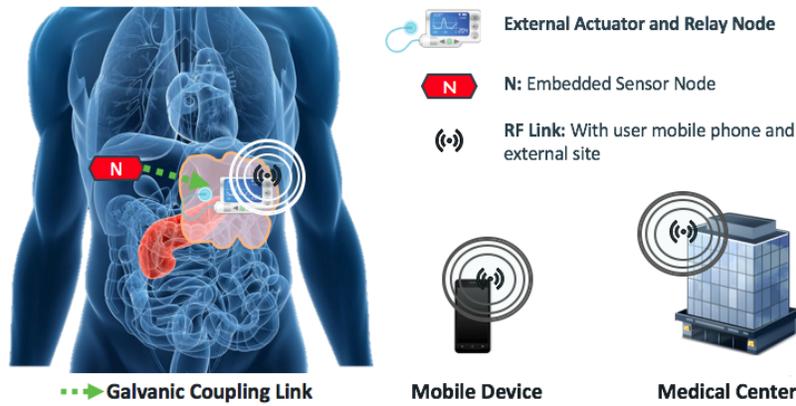


Fig. 15. Conceptual diagram of implantable continuous blood glucose management system using galvanic coupling

physiological parameters influencing blood glucose levels. Other factors such as the activity level, sleep level, food intake, perspiration rate, heart rate, estimations of food intake, varying pH levels, body temperature and ambient temperature are a few of the human body variables that can cause a fluctuation of glucose levels. The above mentioned observations can be made possible either with implanted or on-surface sensors and can be networked towards an adaptive control system to autonomously adapt the data reporting rate of the sensors and direct the action of the insulin delivery actuators. With the help of machine learning techniques and behavior predicting algorithms, the possibility of hyperglycemia or hypoglycemia can be completely avoided enabling diabetic patients to lead a normal life.

D. Wireless Energy Harvesting for Implanted Sensor Networks

The emergence of the implanted sensor technology yields similar challenges when compared to traditional Wireless Sensor Networks. Thus, size, cost and energy are still pertinent to the operational success of such a network. To solve the problem of frequent energy replenishment and battery replacement, we envision the use various forms of wireless charging (e.g., RF-energy harvesting, Magnetic Resonance Charging). Rf-energy harvesting consists of off-the-body, pre-placed Energy Transmitters, responsible for steering ambient RF energy to the implanted nodes. The example topology, illustrated in Figure 16, depicts a wireless charging scenario employing Magnetic Resonance for energy transfer. It operates by wirelessly transferring energy to the implanted sensors through an on-body, battery powered-relay node with a charging coil. Such a charging method extends the lifetime of wireless sensors by reducing the charge drawn from the battery or either eliminating the entire dependence on battery power. Here, the battery powered relay/sink is tasked with aggregating data from all the embedded sensors within the body and communicating this information to a base station. To aid in extending the lifespan of the internal nodes, GC can be utilized for energy savings in communicating information and during front end

hardware usage, by taking advantage of its energy efficient links and simplistic transceiver design, respectively. During node operation, each dual Magnetic Resonance Charging and Galvanic Coupling enabled sensor will maintain an internal energy meter (measured energy level over time). Whenever the energy level drops below a certain threshold (sensor operation dependent), the node must request energy from the relay/sink node. Utilizing intra-body networks in this capacity aims to minimize, or even almost negate the power constraints in an intra-body sensor networks and thus vastly improve the efficiency and user experience of those whom the sensor networks will be built upon.

X. RESEARCH CHALLENGES - GALVANIC COUPLING

GC-based communications has immense potential. However, there are a number of key challenges to that must be addressed related to directional transmission and its effect on safety limitations, system design and the coordination of multiple communicating entities. We also mention related works that present possible solutions.

A. Beamforming in the body

Beamforming within the human body provides high signal-to-noise ratio (SNR) at reception locations and significantly reduced contention between other nodes within the interference domain. However, this approach remains an open challenge due to (i) advancements in far-field beamforming being unsuitable for in-body scenarios involving the low frequency and short range communication of GC, and (ii) the strict safety limits placed on the human tissue in terms of power exposure. As more sources transmit constructively to a given area, an increase in the local tissue temperature can occur, requiring power levels to be monitored and dynamically adjusted. The work in [76] empirically analyzes the constructive and destructive combination of simultaneous GC transmissions through a human tissue phantom, motivating the adoption of beamforming within heterogeneous tissues. The main contribution in this work is to assign a set of complex weights to phase, beam direction and power exposure

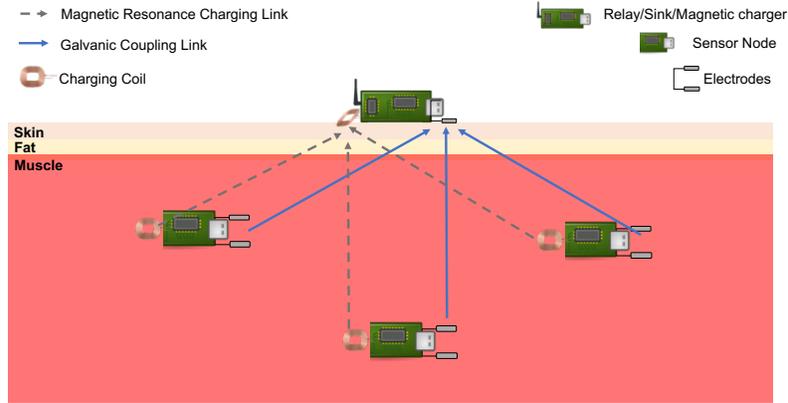


Fig. 16. Conceptual diagram of implantable sensor network with dual wireless charging and galvanic coupling IBC

levels to improve the directional emission of the near-field communication towards an on-body relay node, while reducing propagation in undesired paths.

B. Communication System Design

Although the muscle layer exhibits the least amount of path loss (15 dB for a 10 cm link) in GC based communication, scenarios involving data transfer from the skin to muscle display path loss variation up to 35 dB for the same link distance and operating frequency [56]. This tissue layer path loss variance becomes an extremely important challenge when designing an end-to-end communication system, where an accurate representation of all the losses and gains must be accounted for in the link budget. By assessing the link budget, a system can be properly designed without incurring extra cost (e.g., complexity, power consumption). A proper link budget analysis is crucial to maintain signal levels that are sufficiently above the noise floor, to ensure proper link coverage is met and that the desired bit error rates are within the limits specified by the application scenario. To accurately determine the link budget for galvanic coupled intra-body communication, proper estimation of path loss varying factors, such as tissue thickness, hydration levels and the tissue layer communication scenario (e.g., muscle-to-muscle, muscle-to-skin) present a unique challenge. Current efforts at modeling the GC human body channel attempt to make these parameters as tunable as possible [84]. However, the channel characteristics also change when transmitting and receiving nodes are placed in different portions of the human body (i.e., brain, arm, abdomen, etc.), possibly warranting an entirely different system. Thus, generalized estimates of path losses and gains for any given body cross-section is an open challenge.

C. Medium Access Control

The IEEE 802.15.6 standard, designed for wireless body area networks for implant and on-body communication, specifies the use of contention-based medium access control (MAC). This form of communication introduces the possibility of collisions, backoff and packet loss, events that are known to

incur significant energy loss due to retransmission and channel sensing. The strict energy requirements of embedded body sensor networks, creates a challenge for such infrastructures for long-term, sustainable operation. To mitigate these drawbacks, work in the GC research community [76] and [85], presents beamforming and multi-cast communication schemes that limit the number of concurrent transmissions that may take place within a specific region of the human body. This area still presents an open challenge, as the need for alternative forms of MAC still need to be evaluated and their performance compared with traditional methods. In addition to this, a large subset of IBN applications will involve low-information networking and energy-constrained event monitoring, where low rates and sparse traffic will be generated during the occurrence of abnormal physiological events or in specified update intervals. These characteristics can drastically limit the utility of schedule-based and contention-based multiple access techniques. As a result, the tradeoff problem between latency and energy consumption is exacerbated. The integration of intelligent algorithms and systems similar to wake-up radios, presents a new challenge for GC based IBC, as multi-node topologies continue to be explored [61].

D. Wireless Energy Harvesting

Energy harvesting techniques have been explored previously in the context of Body Area Networks. One example, introduced in [86], focuses on the presence of a thermal energy harvesting technique. The process involves extracting energy from human body, sources of energy that can include vibrational energy, thermal energy, energy from limb movement, passive motion, etc. However, utilizing Wireless Energy Harvesting for GC-IBC, introduces additional design challenges. In order to operate with battery-less implantable nodes, an accurately understanding of how much energy is consumed for every operational scenario is required. Thus, each portion of the node communication that contributes to energy loss must be modeled. A node is responsible for: (i) transmission and reception of data between itself and other implanted sensors (sensor-sensor communication), (ii) transmission of data to

the relay/sink node (sensor-relay communication), and (iii) reception of data from the sink (relay-sensor communication). Additional considerations need to be taken for the PHY and MAC design, which play an integral part in the energy efficiency of the link. Selection of the most energy efficient modulation scheme for the type of data sent throughout the network is an important design consideration, in addition to the selection of the best Medium Access Protocol for the IBN that reduces the energy consumption. All of the aforementioned design considerations should ultimately be factored into an application dependent threshold-based energy model that intelligently decides when to query for energy from energy emitting device. Additional design considerations should include the adaptation of the best energy saving strategy while having data to send and awaiting a charge from the relay, which may be involved in energy transfer to another node [87].

XI. CONCLUSION

This survey has thoroughly explored the key benefits, performance measures and challenges of number of IBC methods. As a summary, coupling methods are preferable to RF-methods for communication in the body due to a combination of high safety factors, low power and attenuation levels. We delve deep into the type of applications that may benefit from galvanic coupling, as it provides a significant trade off between data rate, power, attenuation, low interference and system design, as opposed to other peer technologies. Intra-body communication is a field of growing, continuous research. Many existing aspects of terrestrial wireless technologies can be scaled and adapted for IBC, while newer techniques like mmWave and beamforming can quickly become transformative if used in conjunction with the right coupling method for IBC.

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