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A Review of Implant Communication Technology in WBAN : Progresses and Challenges

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***Abstract*—Over the past six decades there has been tremendous progress made in the field of medical implant communications. A comprehensive review of the progress, current state-of-the-art and future direction is presented in this paper. Implanted Medical Devices (IMDs) are designed mainly for the purpose of diagnostic, therapeutic and assitive applications in heath-care, active living and sports technology. The primary target of implanted medical devices (IMDs) design revolve around reliable communications, sustainable power sources, high degree of miniaturisation while maintaining bio-compatibility to surrounding tissues adhering to the human safety limits set by appropriate guidelines. The role of internet of things (IoT) and intelligent data analysis in implant device networks as future research is presented. Lastly, in addition to reviewing the state-of-the art, a novel intuitive lower bound on implant size is presented.**

***Index Terms*—Medical implants, Intra-body communication, Body area network, Electromagnetic model, Implant power sources, miniaturisation.**

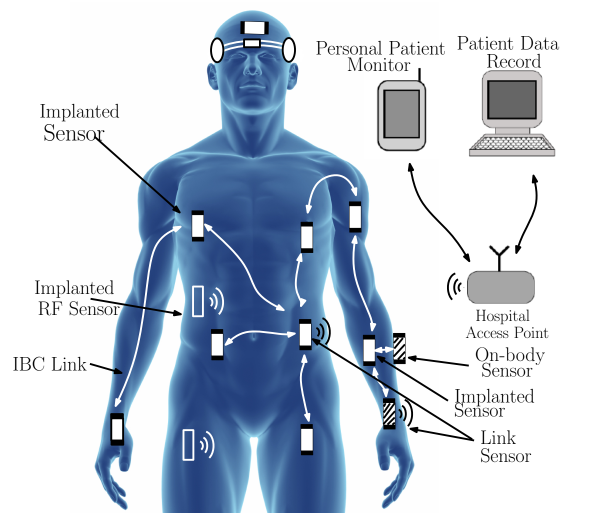

Fig. 1: Wireless Body area network. Part of the implanted sensors use RF communication to the external access point; others, use the the HBC mechanism to communicate to the link sensor

I. Introduction

SINCE the 1950's, research has sought to address the demand for long-term operation and low power communication for medical implants [1], [2]. Implants are now an integral part of the wireless body area network (WBAN) where different implanted or wearable devices are interconnected via implanted or wearable link sensor nodes as shown in Fig. 1. In the WBAN scenario, the defacto implant communication is one where the implanted medical device (IMD) communicates with a wearable data presentation device or a controller located outside the human body and vice versa. In fact, implants also communicate with other implants where an intuitive example is the case of an implanted glucose sensor with an insulin pump.

Unlike traditional through-the-air wireless radio frequency (RF) communication, implant communication uses living tissues as part of its transmission channel and hence faces extra challenges. Firstly, the human body is a hostile channel to high frequency electromagnetic signals. To understand the human body's influence to electromagnetic signals several researchers conducted experimental, analytical and simulation-based characterisations. The early work of Gabriel *et al* [3] characterised dielectric properties of different tissues as a function of frequency; this has enabled testing different hypotheses and theories as to how the tissues affect signals at different frequency.

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In addition, the invasive nature of implantation surgeries places a stringent miniaturisation requirement and sustainable powering regime for implants, especially for long term duration. As such, proper design and use of electronic medical implants need to take into account the specific application of the device. The data rate required dictates the bandwidth and even the implant communication mechanism. For example, subcutaneous glucose sensor transmitter is implanted just under the skin (depth of 4 mm) with low data rate (of less than 10 kbps [4]) and intermittent transmission while a cochlear implant requires a deeper implantation depth of 2.5 cm with a high data rate of up to 500 kbps in a continuous transmission mode. To this end, different communication mechanisms have been studied and employed; these include inductive coupling between implanted and surface mounted coils[5][6], antenna enabled radio frequency (RF) communication[7], and recently the galvanically and capacitively coupled intra-body communication (IBC)[8]. Other emerging technologies such as ultrasonic[9][10], optical[11–13] and molecular intrabody communications[14–16] are also being investigated.

Despite the progresses made by the research community, there are still challenges and avenues for improvement for reliable implementation. This paper reviews the progress and the challenges in implant telecommunications mechanisms for various biomedical applications with respect to the path loss introduced, sustained powering mechanisms, offered bandwidth and device miniaturisation trade-offs. The rest of the paper is presented as follows; biomedical applications of med-

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Application	Example	Range	Power consumption	frequency	size	data rate	Technology
Diagnostic	Glucose sensor [17]	4 cm	100 μW	38 MHz	9.5 mm × 7 mm × 8 mm	-	icDT
		4 cm	41 mW	2 MHz	9.5 mm × 7 mm × 8 mm	-	icPT
	Oximeter [18]	>1cm	-	1.2 GHz	8–10 mm diameter	-	icDT-usPT
	pH sensors [19]	-	-	915 MHz	-	-	RFID
	gastrointestinal imagery pill	-	-	-	length 18-25 mm diameter 6-12 mm	-	-
Therapeutic	pacemakers[20],	≤12.5 cm	-	-	25.9 mm length	-	-
					6.7 mm diameter	-	-
	nerve and muscle stimulator [21]	5–14 mm	100 mW	5–17 MHz	8 mm × 8 mm × 0.2 mm	4.8kbps	icPT Stimulation
		5–14 mm	35mW	5–17 MHz	8 mm × 8 mm × 0.2 mm	1.3 Mbps	icDT recording
	deep brain stimulator [22]	7 cm	132 μW	130 Hz	14.5mm diameter 7 cm length	-	-
Assistive Technology	Cochlear implants [23–25]	2.5 cm	574 μW	5, 12, 49 MHz	10–25 mm length	0.5–1 Mbps	Piezoelectric
	bionic vision implants [26], [27]	2 cm	-	5 MHz	9 mm × 9 mm × 2.5 mm	100 kbps	icDT and icPT
	brain computer interfaces	-	-	-	-	-	-
	for prosthetic limbs [28],	-	-	-	-	-	-
	athletic performance monitors	-	-	-	-	-	-

TABLE I: Summary of biomedical applications of IMDs focussing on communications. Here icDT = Inductively coupled data transfer, icPT = inductively coupled power transfer, usPT = ultrasonic power transfer

ical implants are reviewed in Section II. Implants as integral part of the WBAN architecture are discussed in Section III. Existing and emerging implant communication technologies are reviewed in Section IV. The major requirements of medical implants with respect to what has been achieved and the challenges posed are discussed in Section V. Finally, our conclusion is given in Section VI.

II. Biomedical Applications of Implants

The practical application of electronic medical implants date back to the late 1950s where the first heart pacemaker was successfully implanted [20]. Ever since implants have been used in stimulation, sensory (readout) and in closed loop control settings (full implanted operation). The biomedical applications of IMDs can be classified into three broad categories as diagnostics, therapeutic and assistive devices.

Diagnostic implants measure vital health signs and include devices such as intra-cranial pressure monitors [29], [30], glucose sensors [31], deep brain activity sensors [32], oximeters [33], pH sensors [34], and gastrointestinal imagery pills [35]. The second category, therapeutic IMDs, have been used to treat some form of ailment via electromagnetic stimulation or targeted biochemical intervention according to a pre-calibrated stimulus or controlled closed loop feedback generated by another implanted sensory unit. These IMDs are used in applications such as pacemakers, nerve and muscle stimulator, deep brain stimulator [36], gastric defibrillators [37], targeted drug delivery systems [38]. Finally, assistive IMDs assist sick or even healthy people in improving anatomical and physiological functions. Some examples include cochlear implants [25], bionic vision implants [27], brain computer interfaces for prosthetic limbs [28], and athletic performance monitors. A summary of biomedical applications of implants is presented in Table I. These implants could be either surgically implanted, ingested as a capsule or injected into the particular region of the body. The later two types are minimally invasive. Injectable IMDs are being championed as the future of electronic implants as technology improves miniaturisation.

So far, IMDs are designed for singular applications where communication is restricted between the IMD and external monitoring station either on-body or indoors. However, IMDs could be integrated into a wireless network of implants for more holistic and efficient data transmission. As such, the network of implants can be envisioned as an integral part of internet of things (IoT) for mainly two applications. On one hand, critical medical information could be passed on to patients' physicians and/or next of kin for immediate medical intervention irrespective of where the patient is. On the other hand, diagnostic information from individual patients could be compiled and analysed over time to assist in medical research. However, the later should be done in such way that patient privacy is protected. Although some IMD designs consider privacy and secrecy of medical data, this calls for a rigorous inclusion of physical and application layers of the implant communication network which increases transmission overhead and reduces bandwidth efficiency.

III. Implants in the WBAN Architecture

A. WBAN and Implants

The Wireless Body Area Network (WBAN) is a subset of the metropolitan area network which is specific to communication around the human body. The general architecture of body area networks, as shown in Fig. 1, is that a link node wearable on the surface talks to and listens from the implanted and other surface mounted devices. It then combines and relays the signal to devices external to the body – mainly a monitoring or controlling device on the surface or a few meters away from the body. Another likely scenario is the

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possibility of two implants talking to each other; for example, a glucose sensor and an insulin pump. To reduce complexity and power consumption it is better to implement advanced security features at the link node rather than each individual implanted or on-body device.

B. Communication Modalities

Most of the IBC modalities considered in literature are based on the on-body (surface-to-surface) communication where both the transmitter and receiver are worn on the surface of the skin as shown in Fig. 2a. This modality, for example, enables ubiquitous communication for wire-free patient vital sign monitoring setups in hospitals. For implant communication as part of the WBAN architecture, we consider two modalities. The first modality is the implant-to-implant communication shown in Fig. 2d where both the transmitter and receiver are inside the human body. This modality can be used to communicate implants that operate in a closed loop control setting. Besides, implant to implant communication can also serve as information relaying mechanism to cover a long communication distance by chaining implants. The second assumes communication between a transceiver on the surface of the skin and an implant inside human body of given electrical characteristics as shown in Fig. 2b (implant-to- surface) and Fig. 2c (surface-to-implant). The implant-to-surface implants are used in diagnostic application where a sensed quantity is transmitted to outside the human body. On the other hand, the surface-to-implant modality can be used with therapeutic and assistive implants to pass stimulation or control signals from the outside to the implant. This modality is also extensively used in wireless powering techniques to transfer power from external sources.

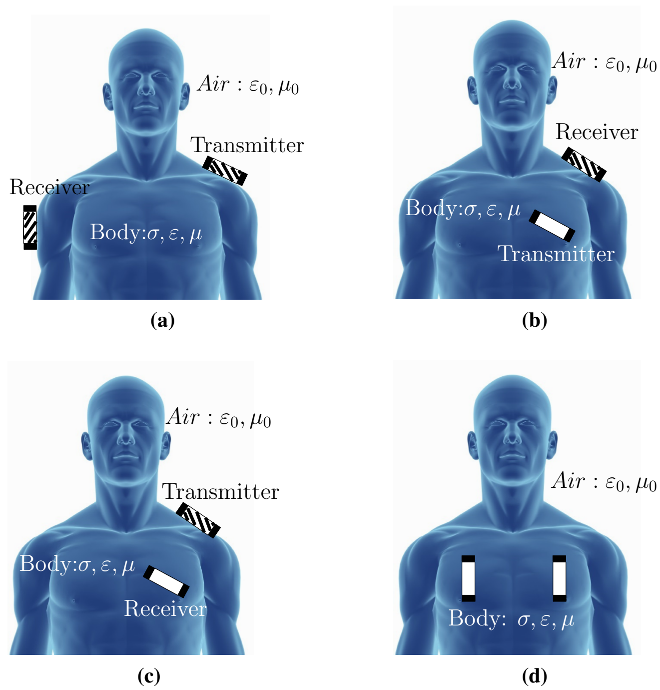

Fig. 2: Implant communication modalities : (a) Surface-to-Surface communication, (b) Implant-to-Surface communication, (c) Surface-to-Implant communication and (d) Implant-to-Implant communication.

IV. IMPLANT COMMUNICATION TECHNOLOGIES

In this section existing and emerging communication technologies are presented. Inductively coupled and antenna based RF technologies have been extensively used to enable communication between implanted and external devices. Other techniques such as intrabody communication that exploit the lossy dielectric nature of the human body, ultrasonic, optical and molecular techniques are emerging as alternative means for implant communication. A summary of implant communication technologies is given in Table II.

A. Antenna based Radio Frequency (RF) techniques

This technique is employed by IMDs where the transmitted signal is fed to an antenna that radiates RF electromagnetic signals through the human body to an external receiver and vice versa. The external device is wearable on the surface of the body or located some distance away from the human body. Following previous designs and proposals, the antenna based RF communications for implants have been standardised by the Federal Communications Commission (FCC) in 1999.

1) Medical Device Radiocommunications Service (MedRadio): In 1999 the medical implants communication system (MICS) was proposed by the FCC and later adopted by the European Telecommunications Standard Institute (ETSI) in 2002. The standard covers the communication between the implant and the controller, and implants within the same body using RF[7].

The MICS standard uses the 402 MHz – 405 MHz frequency band with a bandwidth of 300 kHz per channel. This bandwidth is shared for up-link and down-link as the implant operates in full duplex mode, i.e., the sum of the up-link and down-link bandwidth should be 300 kHz. The duplex setting is mandatory because the standard employs a Listen Before Talk (LBT) protocol to prevent the implant from transmitting without the controllers request. The MICS standard is strict in the sense that the power at the band edges needs to be -36 dBm where the maximum power is limited to -16 dBm (25 μW) of Equivalent Radiated Power (ERP). Expansions of the MICS spectrum, initiated by the ETSI in 2004 and by FCC in 2006, led to the inclusion the 401–402 MHz and the 405–406 MHz as wing bands for non-emergency reporting and monitoring applications. Thus, the revised standard released in 2009 was renamed as Medical Device Radiocommunications Service (MedRadio).

It is interesting to observe here that the small bandwidth (300 kHz) is sufficient to support implants like pacemakers that require small data rates; however, does not guarantee the high data rate future requirements of implants (video, audio or networked). Although the MICS band is unlicensed, it is already in use by Meteorological Aids Service for telemetry of weather by weather balloons. As a result, existing MICS implants use several interference mitigation techniques to minimise the impact of meteorological services. Some of the techniques use multiple error correction codes (ECC) and automatic repeat request (ARQ) to overcome impulsive interferences [39]. The proposed error control codes such as BCH code [40] and Reed-Solomon (RS) code [41] employ sophisticated algorithms such as viterbi decoding algorithm. Other mitigation techniques require frequency agile algorithms to choose channels with lowest noise [39]. Thus, MICS implants require a complex transceiver structure. For this reason, the use of MICS is mostly limited to indoors with long polling intervals.

According to ETSI, yet another bandwidth for wideband implant communications is the Industrial Scientific and Medical (ISM) band around 2.4 GHz. In fact, this band is shared by other services like WiFi and BlueTooth. The standard for this band is set to use a Frequency Hopping Spread Spectrum (FHSS) and Direct Sequence Spread Spectrum (DSSS) with a maximum EIRP of 100 μW. The allocated range of frequency for this service is 2.36 GHz – 2.4 GHz.

2) Miniature Antenna Design: For small implant sizes the resonant frequency of implanted antennas considered fall within the Ultra High Frequency (UHF) band especially from 400 MHz – 2.4 GHz. Different types of antennas have been designed for compact implementation of IMDs at 402 MHz and dual band of 402 MHz/2.4 GHz to comply with the the MedRadio implant communication standard. These include monopole antennas [22] of size $18\times16\times1$mm^3, dipole antennas[9,32,31] with size ranging from $6\times6\times1.5$mm^3 to $16.5\times15.7\times1.27$mm^3, Planar inverted-F antennas (PIFA) [37,34] with sizes $13.5\times15.8\times0.635$mm^3, patch antennas [23] of sizes $15\times15\times3.81$mm^3 and cavity slot antennas of size $1.6\times2.8\times4$mm^3. Several techniques have been used to miniaturise the sizes of these antennas. For example, most dipole and monopole antennas use spiral arms while others use inductive loading and ceramic substrates; some patch, slot and PIFA antennas use stacked and meandered structures.

Several other conformal and non-conformal antennas have been investigated including the human body itself as a lossy monopole antenna [42]. The main challenge with antenna enabled implant communication is that the human body tissues incur increasingly high path loss with frequency. Unlike the air-to-air channel, the lossy dielectric nature of human body (66-70% water) and variable tissue layers have been shown to drift the designed resonant frequency of the antennas in practice.

For an implant transmitting an electromagnetic field inside the human body, the field strength to penetrate through the transversal tissue layer is an important parameter. The depth at which the electric field is attenuated to $\frac{1}{e}$ of the original value is called skin depth where e is the base of the natural logarithm. The larger the skin depth is the deeper the implant can be installed. However, for MICS, apart from being shared by other popular services, penetration of electric field through the human tissue, i.e., skin depth is less than the corresponding value for lower frequency signals [1]; for example, it is 0.14 m at 20 MHz and 0.028 m at 402 MHz for muscle tissue. The quest for alternative communication mechanisms in and around the human body has led to investigation of other schemes such as the intrabody communication that use the human body as channel.

B. Inductively Coupled Data Transfer

Inductively coupled data communication between IMDs and wearable devices is achieved via mutual inductance between primary and secondary coils. The current injected into the primary coil (transmitter side) induces magnetic flux which in turn induces current in the secondary coil (receiver side) according to the coupling coefficient between the two as shown in Fig. 3. In most applications, a narrow band or single sinusoidal power source is used to continuously power the IMD and the IMD uses back telemetry to send sensory recording data back to the external device. This technique has been extensively used for applications that require short range implants such as muscle stimulators, retinal implants, cochlear implants and pacemakers [5], [6]. Different resonance frequencies have been used for inductive data transfer; for example, 1 MHz [43], 5 and 10 MHz [44], 24 MHz [45] and 49 MHz [23] for the advanced bionics cochlear implant.

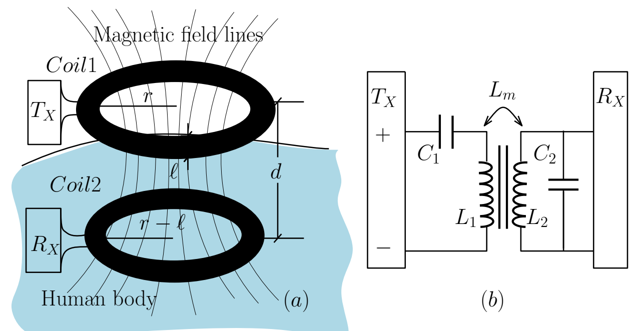

Fig. 3: Inductively coupled intrabody communication. (a) conceptual diagram and (b) A series parallel equivalent circuit

The challenges associated with inductively coupled systems is that it offers a small bandwidth (often in hundreds of kbps) when designed for efficient power transfer. To this end, several mitigation techniques have been proposed for wideband communications. These include separating the power and data link coil pairs in orthogonal dual-band arrangement to limit cross talk and using load shift keying (LSK) for data transmission [46–48]. Others use three coil pairs where one pair is used for power transfer and the other two orthogonal pairs are used for bidirectional offset quadrature phase-shift keying (OQPSK) modulation scheme to increase channel use and bandwidth [49], [50]. Recently, emerging solutions promise inductive coupling links with stacked multilayered coils with data rates in Gbits/s [51], [52]. Based on these, Burhan [53] proposed use of multi-layered graphene nano-coils to push the theoretical channel capacity in Tbits/s with operating frequencies in terahertz range. However there is a significant electromagnetic absorption by tissues at such high frequencies and feasibilities are yet to be tested.

Despite its popularity, inductive data transfer has not been standardised. As such, there is no noted inter-operability and integrability of devices based on inductive data transfer with

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Technology	Method of propagation	Physical Layer Modulation	Range	Frequency	Standard
Antenna based RF	Radio wave propagation	FHSS, DSSS, QPSK	2 m	401 - 406 MHz, 2.36-2.4 GHz	MedRadio / MICS
Inductively coupled data transfer	Magnetic field (Mutual Inductance)	Load Shift Keying (LSK), Amplitude modulation, OQPSK	a few cm (<4 cm)	1,2,5,10,24,49 MHz	none
IBC - body channel communications	Electric field	FSDT frequency selective spread codes are used to spread the digital signal and select the carrier frequency	body confined (<10 cm for implants)	Galvanic < 10 MHz, Capacitive 21 MHz	IEEE Std. 802.15.6
Molecular Communications	Action potential, Axon propagation, ionic diffusion through junction gap, Hormone diffusion through blood	inter-spike interval codes, discrete binary and frequency coding, neural spike pulse modulation, chemical concentration, hormones as modulated carriers	nm -μm	(0,3 kHz]	none
Optical Communications	infrared propagation	IR Laser diode, VCSEL	mm (mostly subcutaneous <4 mm)	700 nm - 1 mm (300 GHz - 430 THz)	none
Ultrasonic Communications	ultrasonic propagation	OOK, ASK	Several centimeters (≈10 cm)	1-3 MHz	none

TABLE II: Summary of implant communications technologies. Here, FHSS = Frequency Hopping Spread Spectrum, DSSS = Direct Sequence Spread Spectrum,QPSK = Quadrature Phase Shift Keying, OQPSK = Offset Quadrature Phase Shift Keying, FSDT = Frequency Selective Digital Transfer, IR = Infrared, VCSEL = Vertical–Cavity Surface–Emitting Laser, OOK = On – Off Keying , ASK = Amplitude Shift Keying.

other devices. Selection of operating frequency and physical layer architecture seems rather arbitrary.

C. Galvanic and Capacitive Coupled Intra-body Communications

In recent years, communication techniques that employ electric field communication (EFC) through human body as a channel have been proposed. These communication mechanisms are interchangeably referred to as human body communication (HBC), or body channel communication (BCC) or intra-body communication (IBC). In this technique, the human body is effectively a volume conductor. It exploits the lossy dielectric nature of the conductive tissue layers to induce a current, and hence a potential distribution as a result of the electric field caused by the current injected by the transmitter electrode(s). Frequencies ranging from a few hundreds of kilohertz to a few tens of megahertz are suitable choice for HBC. Such a low frequency signal is expected to penetrate deeper into the tissue layers, thus requiring less power be detected by a receiver on the surface of skin. Moreover, due to a non-conducting free space outside the skin surface, the signal is confined to within the human body.

The concept of IBC was pioneered by Zimmermann in his masters thesis in 1995 [54]. Since then, it has attracted a number of researchers mainly because it promises low power and inherently secured communication as the transmitted signal is confined within the body. Based on the way the current is coupled to the human body, we have two types of intra-body coupling mechanisms - capacitive coupling and galvanic coupling.

1) Capacitively Coupled IBC: In capacitive coupling, the signal is coupled into the human body via the signal electrode of the transmitter and receiver; a return path is established via the ground electrode through the path outside of the human body as shown in Fig. 4. This technique was originally proposed by Zimmermann [54]. Using this scheme, he successfully transmitted a 330 kHz signal with 3 V at 2.4 kbps digital transmission rate consuming $1.5\ mW$ of power.

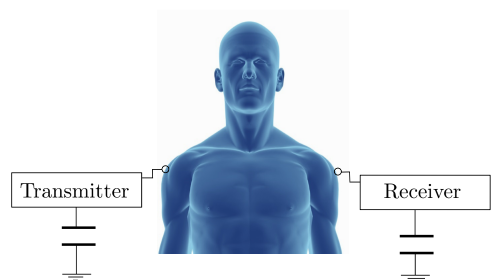

Fig. 4: Capacitively coupled intrabody communication

Later, M. Gray [55] developed the capacitive coupling further and designed a system with increased data rate capacity of 2 Mbps at carrier frequency of 100 kHz. He also showed that the noise in IBC is mainly due to circuit noise and interference from environmental EM sources. In 2004, Shinagawa *et al* [56] developed a capacitively coupled IBC system with a half-duplex transmission rate of 10 Mbps; this marked the first practical application of IBC. Then, in 2007, Fujii et al [57] developed a wearable ID key and used FDTD (Finite Difference Time Domain) simulation model for IBC; later that year, Cho *et al* [58] developed a distributed RC circuit model of the human body; they have validated that their model is consistent with human experimentation in the frequency range of 100 kHz – 150 MHz.

From all the studies conducted in capacitively coupled IBC, we can see that it is practically suited for surface-to-surface communication; it cannot be used for implant communication as the return path needs to be established outside the human body.

2) Galvanically Coupled IBC (gc-IBC): In galvanic coupling, both the signal and current return electrodes of the transmitter and the receiver are in contact with the human body to couple current differentially as shown in Fig. 5. Galvanic coupling IBC was first introduced by Handa in 1997 [59]. This IBC mechanism detects received signal differentially and has inherent common mode rejection capability. Compared to capacitively coupled IBC, the effect by environmental noise outside the body is negligible for galvanic coupled systems. The signal noise is mainly due to differential mode noise. As a result, the system proposed in [59] only used a supply current of 20μA with a power consumption of only 8 μW. Thus, galvanic coupled IBC is a preferred scheme for implant communication.

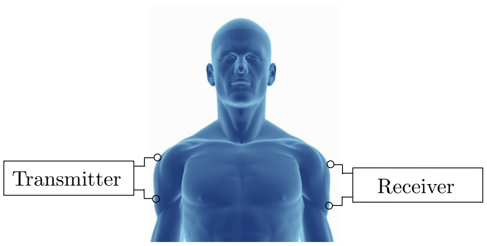

Fig. 5: Galvanically coupled intrabody communication

Lindsey et al [60] built a biomedical telemetry system to measure the tension in anterior cruciate ligament after surgery. They implanted a transmitter inside a cadaver and the receiver was on the surface (i.e., implant-to-surface setting). From their experiment setup, they reported best performance when a current of 3 mA is used at 37 kHz. Following Zimmermann [54], Hadana [59] and Lindsey [60], there has been several developments and researches on surface-to-surface intrabody communication system and communication models. However, not much has been done on practical implant communication based on intrabody communication techniques.

In 2005, Hachisuka et al [61] developed a two electrode and four electrode circuit model for the intra-body communication. From their experiments and model predictions, they concluded that the two electrode circuit model performed better than the four electrode model in the kHz and MHz frequency range. The two electrode model can be interpreted as the capacitive coupling and the four electrode model can be interpreted as the galvanic coupling IBC.

In 2011, Song et al [62] developed the four electrode circuit model and derived a transfer function for the galvanic coupled IBC. The model fits the measurement for the 100 kHz – 150 MHz well up to a scale factor. We note here that, most of the models are developed based on the surface-to-surface galvanic coupled setting. For implant communication, it is important to extend these models to include the transversal implant-to-surface communication channel. Wegmuller [63] first studied galvanic coupling for implant-to-implant communication using a simplified Finite Element Method (FEM) simulation model in Comsol Multiphysics software. However, limitations in the models did not explicitly capture the holistic essence of the channel. To this end, recently, we developed an analytical electromagnetic models that captures the variation of potential distribution and path loss as functions of variable number of tissue layers and thicknesses, implant size, depth of implantation transceiver properties in [8].

3) IEEE 802.15.6 Wireless Body Area Network (WBAN): The IEEE 802.15.6 work group was established in 2007 to standardise wireless communication in, on and near the human body - the wireless body area network (WBAN) which was released in 2012 [40]. The IEEE 802.15.6 WBAN standard has three layers the Ultra Wide Band (UWB), the Narrow Band (NB) and the Human body Communication (HBC) layers. Frequencies used in the existing MedRadio standard are included in the Narrow Band specifications of the standard. The 2.4 GHz ISM band is also included in the UWB specification. The standard specifies the HBC to be centered around 21 MHz and uses frequency selective digital transmission (FSDT). Here, frequency selective spread codes are used to spread the digital signal and select the carrier frequency.

Although, HBC is specified mainly for surface to surface communication, works such as [64] [8] and [65] have explored the use, especially, of galvanically coupled IBC for implant communications at frequencies ranging from 100 kHz to 10 MHz. The implant communication channel is very different from the surface-to-surface channel. While the surface to surface channel has a band pass gain characteristic where the gain picks in the region 20-60 MHz, the implant-to-surface communication channel has a low-pass gain characteristics that favours lower frequency transmission for minimal path loss.

Antenna-free miniature implementation, low attenuation and body-confined transmission features of galvanically coupled IBC makes it a good alternative for implant communication. To this end, this paper calls for inclusion and proper specification of the galvanically coupled IBC scheme for implant communication. With the advances in diagnostic and health monitoring sensors, the demand for high speed and long term communication continues to grow. Thus, a large number of people are expected to be aided by medical implants (in addition to the existing more than 25 millions of people with implants in the United States alone - estimated in the early 2000s). Hence, there will be a high risk of interference from implants in different people which was not clearly addressed by the MICS standard. The use of HBC will undoubtedly address the issue of interference in implants in two people as the functional communication is limited to within the body.

D. Ultrasonic Communications

Ultrasonic communication is enabled by mechanical waves propagating inside the human body as an elastic medium with frequencies above 20 kHz as shown in Fig. 6(a). Conversion between electrical and ultrasonic signals is achieved through ultrasonic transducers mainly piezoelectric transducers. In some reports a backscatterng modulation is used to effectively transmit data at a rate of 50 kbps at 1 MHz only consuming 184 μW [9] where on-off keying (OOK) and amplitude shift keying (ASK) modulations are employed. A more involved PHY and MAC protocol that enables a high speed data rate of 700 kbps only consuming 40 μW has been proposed in [10].

Although it suffers from high attenuation, this technique is emerging as a valid alternative for short range communications that can be coupled with piezoelectric power scavenging techniques or be powered externally for a backscattering

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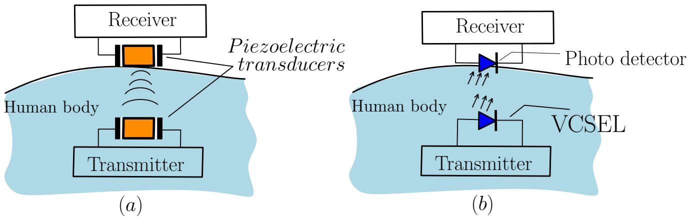

Fig. 6: (a) Ultrasonic Implant communication. (b) Optical implant communication using VCSEL.

transmission. However, standards have not been adopted to extend its application to integrate with other systems.

E. Optical Communications

Optical implant communications are enabled by optical propagation of infrared (IR) waves. An implanted transmitter couples electrical signal in to the channel (human body) by converting it into IR signal using a form of laser diode. Often a vertical-cavity-emitting laser (VCSEL) diode is used at the transmitter. When the IR signal is incident, part of it is reflected and the reminder is scattered or absorbed by the human body. IR absorbtion by the human tissues (especially by skin) is so high that effective communication is limited to millimeters. Thus, this communication technique is restricted to transcutaneous or subcutaneous implants. As shown in Fig. 6(b), a receiver on the surface of the skin employs a photo detector to convert the IR signal to electrical and proceed with the demodulation and detection of transmitted message. This technique is suitable for operation in the 700 nm - 1 mm (300 GHz-430 THz) band.

For example, Abita *et al* [11] used an 860 nm (348.596 GHz) carrier to transmit an RS-232 data of rate 115.2 kbps. Here, a photonic detector LED PDI-E804 is used over less than 24 mm communication distance where a porcine skin is used as a channel. Although the 24 mm range is an over-estimate for the subcutaneous implants which are often limited to less than 4 mm, the technique at nanometer wave could be used to achieve a much faster data rate up to 50 Mbps as presented in [12] (860 nm, 4 mW power consumption, 4 mm range, VCSEL Tx and PIN Si Photodiode Rx). In a relatively recent work, Mujeeb *et al* [13] proposed an all-optical (optical powering and data transfer) solution for subcutaneous implant communication of distance less than 4 mm. They used the "therapeutic window" of the spectrum which is the near infrared (NIR) band (from about 700 nm to 2500 nm) to obtain a CMOS based optimal optical power and data transfer since this band falls with in the silicon absorption band. As such, this scheme promises a great deal of miniaturisation with the advances in silicon based technology.

The advantage of optical communication links in the human body is that they are least affected by channel interferences, but suffer from high path loss attenuations. Despite being a potential alternative, there is limited work on optical communication for implant communication and integrating it as part of the existing standards.

F. Molecular Nano-networks and Communications

Molecular communications (MC) refer to biological intrabody communications where the communication between nano-transmitters and nano-receivers is achieved by a combination of chemical and electrical signalling through the channel linking cellular transmitter and receiver. In living bodies, these nano transmitters and receivers (also referred to as nano-machines) are the basic functional units of nano networks that are able to convey simple information [66]. A single pathway in MC ranges in nm–μm distance with very small frequencies of 0–3 kHz [67]. These simple nano-machines form a large scale nano-networks where they exchange and cooperate to enable transmission of complex information over an extended distance, e.g., the nervous system capable of interconnecting extremities of the human anatomy. Investigation of MC is motivated to understand the state-of-art mainly for two reasons. These are, to develop bio-inspired nano-networks for applications such as artificial prosthetics and to detect ailments or develop therapeutic interventions such as treatment of neurological ailments and targeted drug delivery when integrated within living organs. Mechanisms of MC are radically different from the conventional signal transmission techniques. Most MCs are enabled by diffusion of molecules while others are enabled by microtubles as channels. In scenarios such as the nervous and cardiovascular systems a combination of electrical and chemical molecules are employed. Analog type communications are observed in cases where a continuous emission of molecules of varying concentration enables diffusion or variable interspike pulse widths enable electrochemical action potential propagation. Digital communication is observed in the discrete type of chemicals that bind with nanotrasmitter molecules or quantised release time of molecules. The three main nano-networks in human body are the nervous, cardiovascular and endocrine nanonetworks. These networks use a radically different communication paradigm than other traditions wireless communications.

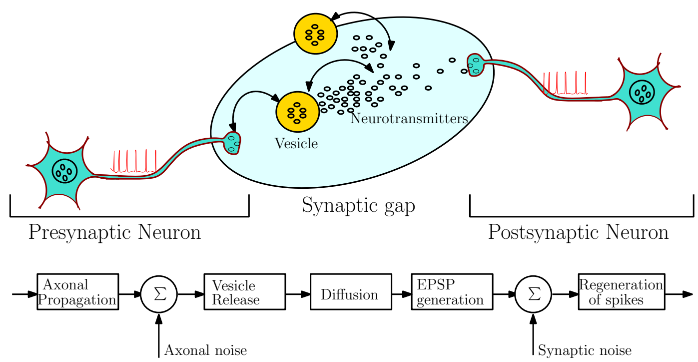

Fig. 7: An example molecular communication: Nervous nanonetwork channel [66].

The Nervous nanonetwork (NnN) is an ultralarge network of neurons where information is transmitted between different parts of the body [15]. The fundamental pathway in NnN is one that exists between two nanotransceiver neurons vis presynaptic neuron and postsynaptic neuron. There are two modes of communication to complete the path - Axonal propagation and synaptic propagation [68] as shown in Fig. 7.

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In axonal propagation action potential information is encoded as electrical impulses of variable width and the axon acts as a channel. At the interface membrane of the axon with the synaptic cleft, the propagated action potential induces calcium ions which bind with proteins released through the membrane. The synaptic transmission propagates the neurotransmitter chemicals via diffusion across the synaptic gap where the rate of diffusion is controlled by the concentration of the chemicals released by the axon (which in turn is proportional to the electrical impulses that propagated the action potential). At the post synaptic end, the diffused proteins are intercepted by special receptors which will extract the ions and inject them through the receiving neuron membrane - synaptic decoding. In the postsynaptic neuron the rate of change of concentration of the ions induces action potential to regenerate the transmitted impulses. This induces a chain that can extend over a long distance. The axonal propagation suffers from axonal noise due to random opening and closing of the ion channel while the synaptic channel suffers from Brownian motion and interference from thousands of neighbouring synapses.

The Cardiovascular nanonetwork is based on the spontaneous action potential created by the cardiac pacemaker cells that are propagated to the cardiomyocytes to create the beating of the heart. It uses connexons as channels in the gap junctions. The connexon is normally closed and opens when it receives the propagated action potential to yield ion transfer in to the gap where diffusion takes over the rest of the way. In Endocrine nanonetworks, hormones are the modulated carries of molecular information. It uses diffusion through the blood. The rate of blood flow is proportional to the data rate and noise is mainly due to Brownian motion of the hormones in the blood.

As communication networks, molecular communications have been modeled as single-input-single-output (SISO) [16] and multiple-input-single-output (MISO) [14] schemes to information theoretically model communications. However, analysis has been limited to simplistic modes and more realistic models with experimental validations are yet to be carried out for complete understanding of these communication techniques.

V. Required Features and challenges of Medical Implant Communications

A. Implant Powering and Power Consumption

The main requirement in implant transceiver design is the low power consumption and a sustained power supply system as the implant is embedded inside the human body. Following the MICS standard, several implant transceivers have been designed and implemented. The most popular MICS implant transceiver is by Bradley [41] of Zarlink Semiconductors. This MICS transceiver consumes less than 5 mA current in the active mode and about 250 nA in the sleep mode at a supply voltage of 2.1–3.5 V; it consumes an average power of 11.5 mW and has a receiver sensitivity of 20 μ V r.m.s. at 400 MHz for a 200 kbps transmission. Microsemi Corporation [69] has also commercialised transceiver, for various MICS telemetry, that has similar features as given in [41]. In 2009 Cho *et al* [70] developed a dual MICS/HBC transceiver consuming a total of 10 mW (2.3 mW for HBC and 8.5 mW for MICS) at a data rate of 50 kbps at the transmitter and 200 kbps at the receiver. Although this transceiver uses less power for MICS than Bradley's [41], the receiver sensitivity is higher at 35 μV_{rms}. The MICS transceiver that consumes the lowest power was developed by Pandey *et al* [71]. At 400 MHz the transceiver consumes 90 μW with an output (transmitted) power of 20 μW (less than the maximum 25 μW set by MICS) with a 200 kbps data rate. The transmitter has an active area less than 200 μ m $\times$ 200 μ m.

Despite this, development of low power implants and sustainable implant powering is still an open problem which has been reviewed in depth by Bazaka *et al* [72]. Single use or rechargeable batteries have been the common power source for implanted medical devices. For example, Medtronic has developed a small pacemaker a size of vitamin pill, Micra [73] powered by an estimated average 12-year battery life. However, in the case of single use batteries, expensive replacement surgeries are required. To address this challenge, IMDs with rechargeable batteries have been designed to be charged by external power sources. These batteries constitute most of the implant size and impede implant miniaturisation in addition to the risk of adverse bio-compatibility effects[20]. Thus, batteryless powering techniques have been adopted to replace batteries.

Batteryless techniques can be broadly classified into two; the first being the use of the surface to implant transmitter which transmit a coded power in the form of electromagnetic field to an implant, and the second is the use of electro- and bio- chemical reactions in the body to generate power. With respect to the former, inductively coupled power transfer have been extensively used to power IMDs [74], [75] such as pacemakers, bionic vision[27] and cochlear implants. Here,the primary coil associated with the external power source couples the time harmonic power signal to the secondary coil associated with the IMD via a mutual inductance. External RF electromagnetic sources radiating coded power signals have been used power IMDs where the IMD modulate the message and relay the signal back via backscattering similar to RFIDs[76], [77]. Bio-battery systems using glucose oxidation has been reported to generate a power of 3.4 $\mu W\ cm^{-2}$ – 180 $\mu W\ cm^{-2}$ [72]. Other systems using glucose bio-fuel cells were shown to deliver power density up to 1.3 mW cm^{-2} with an open circuit voltage of 0.95V; by combining two of these cells in series the authors were able to generate 3.25 mW at 1.2 V and an open circuit voltage of 1.8 V. Moreover, electrical power scavenging mechanisms within the body such as human body movement[78], [79], piezoelectric [80], bioelectrical[81] and other biochemical reactions [82], [83] to power IMDs have been investigated. The potential of these techniques render batteryless power source as the future of implant powering.

B. Miniaturisation and Lower-bound on Implant size

As a foreign object, an implant introduces discomfort to surrounding tissues in addition to requiring invasive surgeries. Thus, light weight and small implant sizes are key features for minimal invasiveness. The size is dependent on electronics, antenna required and battery size. Advancement of the circuit technology ensures a great deal of miniaturisation. The use of batteries is being replaced by betteryless powering regimes. Thus, research on miniaturising implants is focused on antenna sizes. Although low frequency signals experience low path loss, radio wave based RF transmissions are forced to move into the ISM band for miniature micro strip antennas [84], [85]. Helical antenna structures (that can be compacted in small volumes) with circular polarisation have also been used in pill sized ingested implants [86], [87]. The geometry of the antenna determines the frequency of operation and hence does not leave much room for flexibility. In fact, the most reported challenge with microstrip patch antennas at a very high frequency is that a small error in the cut dimensions results in a considerable shift in resonance frequency. Despite this, implant sizes up to 1.4 mm × 0.905 cm × 0.945 cm have been designed to operate at 2.4 GHz [88]. Although antennas in GHz frequency partially address the issue of implant size, the human body channel incurs a path loss of 50–60 dB in a commination distance of 4 cm [8]. Thus, it is worth considering the low frequency IBC techniques with a promise of a better trade-off (i.e., lower path loss and smaller size); this issue of implant size have not been addressed adequately by other researchers.

In general, for galvanically coupled IBC, the larger the intra-electrode distance is the smaller the pathloss will be. Hence, intra-electrode distance can be assumed to be the maximum dimension of the implant. Callejón *et al* [89] developed a simulation model for galvanic coupling IBC as a four port network using a finite element approach to analyse the electric field distribution and current density. Their model and experiments based on human arm on a surface-to-surface setting showed that path loss decreases as the intra-electrode distance increases. However, this study considered only a few electrode spacing in the surface-to-surface setting and hence needs further work to explain the effect of electrode spacing on implant communication. A trade-off between electrode spacing (intra-electrode) and communication distance (thereby path loss) is an important investigation. To this end, in our previous work [8], we presented analytical and simulation models of IBC for implants that explicitly showed the effect of electrode spacing and tissue layers in pathloss as a function of the communication distance and frequency. Supported by validation experiments, it was shown that galvanically coupled IBC is a feasible means for implant communication. Although smaller electrode spacing incurs larger path loss, the limit up to which this spacing could be reduced (for a required receiver sensitivity) is an important target to consider for implant miniaturisation.

We propose here an intuitive lower bound on the implant size for galvanically coupled IBC. We modeled the galvanic coupled transmitter electrodes as linear dipole antenna sitting inside a lossy dielectric as shown in Fig. 2a - 2b. The receiver electrodes are also modeled as linear dipole sitting inside the same dielectric and on an air-dielectric boundary to simulate the implant-implant and implant-surface communications respectively.

Considering the implant-to-implant communication scenario given in Fig. 2d. Any electromagnetic communication is governed by the set of Maxwell's equation given by

$$\nabla \times \overrightarrow{E} = -\frac{\partial \overrightarrow{B}}{\partial t}, \tag{1}$$

$$\nabla \times \overrightarrow{H} = \overrightarrow{J} + \frac{\partial \overrightarrow{D}}{\partial t}, \tag{2}$$

$$\nabla \cdot \overrightarrow{B} = 0 \text{ and} \tag{3}$$

$$\nabla \cdot \overrightarrow{D} = \rho \tag{4}$$

where $\overrightarrow{J}$ and ρ, respectively, are volume current and charge densities that exist in the medium (human body) as time-varying sources. The electric field flux D and the magnetic field density B can be defined in terms of the electric field E and the magnetic field strength H respectively as

$$\overrightarrow{D} = \epsilon \overrightarrow{E}, \text{ and} \tag{5}$$

$$\overrightarrow{B} = \mu \overrightarrow{H} \tag{6}$$

where ϵ is the permittivity and μ is permeability of the human body as a dielectric material. The relative permittivity (ϵ_r) and conductivity (σ) values as functions of frequency are given in [90]. For muscle tissue, the typical relative permittivity at 21 MHz is $\epsilon_r = 110$. Here $\epsilon = \epsilon_r \epsilon_0$ where $\epsilon_0 = 8.854 \times 10^{-12}$ and $\mu = 4\pi \times 10^{-7}$. Since the tissue is lossy, i.e., has non-zero conductivity value, we use the complex permittivity $\epsilon' = \epsilon + j\frac{\sigma'}{\omega}$ instead of just the real value to account for the loss. Here, σ is the conductivity of the muscle tissue which is typically 0.6426 at a frequency of f=21 MHz. The variable ω is the angular frequency given as $\omega = 2\pi f$.

Solving Maxwell's equation for the setting given, the electric and magnetic field strength are

$$\overrightarrow{H} = \frac{j\beta Il}{4\pi r}\Big(1 + \frac{1}{j\beta r}\Big)\sin\theta e^{-j\beta r} a_\phi \tag{7}$$

$$\begin{aligned}\overrightarrow{E} &= \frac{\eta Il}{2\pi^2 r}\Big(1 + \frac{1}{j\beta r}\Big)\cos\theta e^{j\beta r} a_r + \\ &\frac{j\beta Il\eta}{4\pi r}\Big(1 + \frac{1}{j\beta r} - \frac{1}{\beta^2 r^2}\Big)\sin\theta e^{-j\beta r} a_\theta\end{aligned} \tag{8}$$

where, β is the wave number given by $\beta = \omega\sqrt{(\epsilon'\mu)}$, η is the characteristic impedance of the medium given by $\eta = \sqrt{(\frac{\mu}{\epsilon'})}$, I is the current injected into the transmitting electrode, l is the intra-electrode distance, r is the inter-electrode distance (distance between transmitter and receiver) and θ is the angle the electrode makes with respect to the vertical axis.

The power contained in an electromagnetic field per unit area is given by the Poynting vector as

$$\overrightarrow{S} = \langle \overrightarrow{E} \times \overrightarrow{H}^* \rangle. \tag{9}$$

Hence the total power contained by the electromagnetic field can be calculated as

$$P = \oint \overrightarrow{S} \cdot ds. \tag{10}$$

Evaluating the power for the maximum power transfer scenario (i.e., matched impedance in the receiver circuit and receiver

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electrode), the power the receiver detects at a distance r is bounded by

$$P \leq \Re\Big\{\frac{3\eta I^2 l^2}{64\pi r^2}\Big(1 - \frac{1}{\beta^2 r^2} + \frac{1}{\beta^4 r^4}\Big)\Big\} \tag{11}$$

Finally, the lower bound on the intra-electrode distance is given by

$$l \geq \Re\left\{\sqrt{\frac{64\pi r^2 P}{3\eta I^2\Big(1 - \frac{1}{\beta^2 r^2} + \frac{1}{\beta^4 r^4}\Big)}}\right\} \tag{12}$$

Combining the requirements of MICS and the IEEE 802.15.6 standards, the receiver sensitivity can be set to -92.5 dBm of power. Assuming an input current of 1mA and material properties of the muscle tissue, the lower bound for different frequencies is graphed in Fig. 8. In fact, the bound can be computed for different implantation scenarios by considering appropriate tissue characteristics and frequency of operation for analysis purposes.

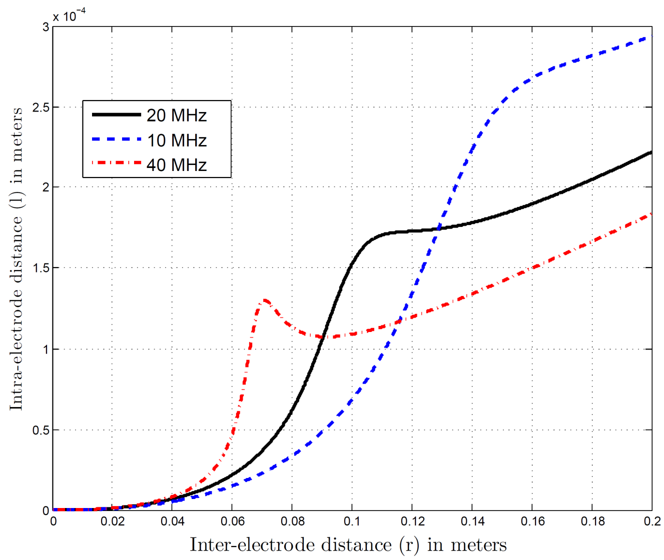

Fig. 8: The lower bound on inter-electrode spacing as a function of communication distance.

This lower bound is tight in the sense that it can be achieved with equality. This shows that the implant designs can further be miniaturised to a great extent. For example, at 1 mA transmitter current and -92.5 dBm receiver sensitivity we can achieve up to a minimum of 70 μm electrode spacing on a 10 MHz signal for a communication distance of 10 cm. A distance of 10 cm can be considered larger than most practical depths to embed medical implants. Besides, the lower bound shows that lower frequency signals prevail in the near field galvanic coupling setup; while higher frequencies dominate the far-field due to the onset of RF propagation.

C. Human safety and Bio-compatibility

In addition to design challenges for reliable implant communication, the emanating electromagnetic signal should not exceed the limits for human safety. Moreover, the IMD should be bio-compatible to the surrounding tissue (i.e., inert to its surrounding). With regards to human safety, the International Commission for Non-Ionising Radiation (ICNIRP) developed a guideline with regards to maximum electric filed, current density, specific absorption rate (SAR) exposures for the general public and occupational settings [91]. To ensure bio-compatibility, encasing the IMD inside steel containers, polymers and/or other superstrates inert to tissues nearby [92], [93]. However latter two are relatively short-lived compared to the desired life span of the IMD. Tissue growth around the device (especially the active region of sensing units) would degrade IMD efficiency over time. Besides, stimulation electrodes (especially in the case of cortical and deep brain implants) have been shown to cause tissue scaring over prolonged usage. To this end, electrode material and geometries that minimise such scarring have been investigated with promising results [94], [95].

VI. Conclusion

In this paper we have reviewed progresses and challenges in communication and applications of implanted medical devices. We have also presented possible future directions. Electronic medical implants have come a long way since their first application as a pacemaker in late 1950s. Today, IMDs are used in several diagnostic, therapeutic and assitive technologies in health-care and professional sports. Uncoordinated progresses have been standardised as MICS in 1999 which later was upgraded to MedRadio services in 2009. Recently, the WBAN standard was also released in 2012 to include non radiating body coupled communications that use the human body as a channel at lower frequencies than considered in MedRadio. These standards have aided the research community and industries like Medtronic and Zarlink to design and develop IMDs with compatible communications circuitry. Although, coordinated effort led to advancement in IMD technology, there are still challenges the research community is targeting to address. These include, reliable and sustainable power sources, implant miniaturisation, bio-compatibility and human safety. With further advancements and the soaring demand for IMDs, implants per user are envisioned to be interconnected via a wireless network for reliable data transmission to and from the patient to physicians and next of kin even far away. Besides, advancements in the big data analysis can be exploited by compiling data from various patients over time to aid in medical research.

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