Developing a Wireless Implantable Body Sensor Network in MICS Band

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***Abstract*—Through an integration of wireless communication and sensing technologies, the concept of a body sensor network (BSN) was initially proposed in the early decade with the aim to provide an essential technology for wearable, ambulatory, and pervasive health monitoring for elderly people and chronic patients. It has become a hot research area due to big opportunities as well as great challenges it presents. Though the idea of an implantable BSN was proposed in parallel with the on-body sensor network, the development in this area is relatively slow due to the complexity of human body, safety concerns, and some technological bottlenecks such as the design of ultralow-power implantable RF transceiver. This paper describes a new wireless implantable BSN that operates in medical implant communication service (MICS) frequency band. This system innovatively incorporates both sensing and actuation nodes to form a closed-control loop for physiological monitoring and drug delivery for critically ill patients. The sensing node, which is designed using system-on-chip technologies, takes advantage of the newly available ultralow-power Zarlink MICS transceiver for wireless data transmission. Finally, the specific absorption rate distribution of the proposed system was simulated to determine the *in vivo* electromagnetic field absorption and the power safety limits.**

***Index Terms*—Body sensor network (BSN), electromagnetic field (EMF) absorption, implantable biomedical devices, medical implant communications service, specific absorption rate (SAR).**

I. Introduction

RECENTLY, the body sensor network (BSN) technologies have been experiencing a rapid development. BSN is a fusion of sensing technologies and wireless communication technologies and has been hailed as a solution for long-term continuous physiological monitoring for chronic patients and elderly people for early detection of life-threatening abnormalities as well as for chronic condition management [1]–[5]. It facilitates the paradigm shift of healthcare from doctor centric to patient centric by providing home-based long-term monitoring that allows patients to engage in their normal daily life activities rather than having to stay in a clinical environment. The home-based monitoring also provides great opportunity for healthcare cost reduction.

Generally speaking, a BSN is a wireless network consisting of a series of miniaturized and low-power biosensors wholly or partially covering the body area. Those biosensors can be either on-body (wearable) or in-body (implantable). Several similar architectures have been proposed and implemented in last decade [1]–[3]. Within a BSN, the sensor node is generally required to be autonomous, i.e., be able to perform its own task or action and communicate with a network coordinator (a personal server) that can send the collected physiological information to a remote medical server for further analysis and diagnosis.

The sensor node design and the adopted wireless communication technology are two key technical issues for developing a BSN system. For most proposed on-body BSN, the system design is largely based on commercially available off-the-shelf technologies [3]–[8]. A typical BSN system was proposed in 2003 that used a TI MPS430 microcontroller as the core technology to acquire and process multiple physiological signals [4]. In addition, a 900 MHz RF transceiver module was adopted as a low power transmission solution. Later, with the advent of Chipcon 2.4 GHz IEEE 802.15.4/ZigBee RF transceiver, a modified BSN node was developed in 2005 [1], [3]. The open-source operating system, TinyOS [1], [6], [8], [9], was chosen for on-board processing, communication, and network synchronization tasks. Since then, the Telos platform, the so-called Telos Mote, which integrates an 8 MHz TI MSP430 microcontroller, a USB interface, a wireless ZigBee-compliant radio with an on-board antenna and with the TinyOS operating system installed, has been widely adopted to construct BSN systems [6], [8], [9].

ZigBee (IEEE 802.15.4) has been greeted as an ideal choice for industrial, commercial, and home automation sensor network because it provides low power consumption, cost effective (the single chip cost is about \$1 U.S. now) as well as standard-based wireless networking solutions. ZigBee can operate at 868 MHz, 915 MHz, and 2.4 GHz which are part of the industrial, scientific, and medical (ISM) RF bands. The UHF range carrier frequency ensures that it has sufficient bandwidth for physiological data transfer. The over-the-air data rate for ZigBee operating at 2.4 GHz is 250 kb/s per channel [10]. Its covering range can be either 10 or 75 m and it supports flexible network topologies such as mesh, star, and ad hoc

Manuscript received April 12, 2010; revised March 3, 2011; accepted April 21, 2011. Date of publication May 12, 2011; date of current version July 15, 2011. This work was supported by a Research Seed Grant from the Royal Melbourne Institute of Technology University, Melbourne, Vic., Australia.

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Digital Object Identifier 10.1109/TITB.2011.2153865

configurations. Based on this, ZigBee can be considered as the right choice for a wireless on-body sensor network system.

Although the Telos-based intelligent sensor node has gained its popularity in research communities, there exist great obstacles that prevent it from gaining wide acceptance from its intended end users. One major obstacle is the physical implementation of physiological sensors. The typical package dimension of Chipcon2420 and MPS430 is 7 mm × 7 mm and 15 mm × 15 mm, respectively [9], which suggests that the final size of a BSN node will be at least at the centimeter order. Moreover, the bulky on-board antenna is also an obstacle to provide wearability and implantability. TinyOs, as an embedding operating system, is designed for general wireless sensor networks not particularly for BSN. The TinyOs-based BSN node is a heavy weight architecture from which many available resources are rarely or even never used, for example, the USB connectivity from the Telos Mote is hardly used. The typical power consumption is in ten milliwatts order; therefore, the long-term continuous monitoring is still not feasible by using those approaches.

Nowadays, various implantable biomedical devices are playing more and more important roles for both diagnosis and therapeutic treatment by taking advantage of the rapid advance of very large scale integration (VLSI) and communication technologies. It is feasible to develop implantable body sensor networks (IBSNs) by adding network function to multiple standalone implantable devices. Such an IBSN system can eliminate many usability obstacles and provide a new solution to the end patients. Recently, an energy-efficient application-specific integrated circuit has been designed for a wireless BSN [11]. By adopting work-on-demand mode and realizing zero standby current, this design achieved ultralow power. Nevertheless, this design is not for implantable applications and the frequency band used for RF communication is the ISM band 915 MHz, not the dedicated 402 MHz medical implant communication service (MICS) band.

In this paper, we present a novel design of an IBSN operating in MICS band. The major design challenges including the choices of wireless communication, the powering mode, and the system-on-a-chip (SoC) network node design are discussed in detail. In addition, the in-body electromagnetic field (EMF) exposure caused by the implanted RF antenna was modeled and simulated using computer simulation technology (CST) microwave studio software. By taking the cutting edge ultralow-power SoC design and adopting the newly available Zarlink MICS transceiver, this system innovatively incorporates both sensing and actuation nodes to form a closed-control loop for physiological monitoring and drug delivery for critically ill patients. This system not only attains low power consumption but also achieves low *in vivo* EMF exposure.

II. IBSNs

A. Implantable Biomedical Devices

Many implantable biomedical devices have already been developed for various medical applications and have significantly improved the quality of life for many patients. Those medical implants not only relieve the inconvenience caused by the wearable devices, but also bring in their new functionalities. Almost, every aspect of human health can be monitored or regulated by an implanted device. In general, the medical implants can be classified into therapeutic and diagnostic categories. For diagnostic implants, there are implantable devices for *in vivo* physiological signal monitoring such as ECG, SvO2, blood pressure, blood glucose level, and neural activity recording [12]–[14] as well as devices for medical image capturing such as capsule endoscope [15]. The implantable therapeutic systems include implantable apparatus for microrobotic surgery, implantable drug delivery [16], [17], and various neuromuscular microstimulation systems. For implantable microstimulators alone, there are variety of possible therapeutic applications such as to restore the control of paralyzed limbs, enable bladder and bowel muscle control, maintain regular heart rhythm, and restore impaired vision. The range of biomedical devices and systems being implanted into the human body is increasing rapidly [18].

For implantable medical systems, power is the major concern in the system design. Battery alone is not the optimal choice for implantable devices due to its limited lifetime as well as its bulky size. In recent years, electromagnetic propagation through inductive coupling links has been commonly used to deliver power and information for implanted systems [19], [20]. The power transmission efficiency, the data transmission rate, and the coupling insensitivity are the major considerations in this kind of power telemetry technique.

An implantable medical device generally works as an isolated standalone device rather than as a connected and coordinated system. During many critical medical treatment procedures, comprehensive real-time body information is often required for medical decision making. For example, the implanted drug delivering device needs the feedback information from the targeting areas in order to release the right dosage at the right time and in the right place.

The majority of those human body wearable or implantable biomedical devices have wireless communication capabilities. With the fast expansion of clinical adoption of those technologies, the biological effects caused by EMF from implanted sources have become a concern. Though many pieces of research have been conducted for RF bioeffects from external sources such as mobile phones, few efforts have been spent on those internal sources.

B. Current Development Efforts on IBSN

Though the concept of a network of wireless microimplants was proposed almost the same time as the on-body sensor networks was proposed [9], [21], few actual design and implementation works have been published. On the other hand, more and more micro/nanoimplants have been surgically implanted or injected into the bodies of those patients suffering from diabetes, chronic hypertension, coronary heart disease, and neuromuscular dysfunction. Those multiple implants are often required to work in a synchronized or at least coordinated way to achieve their designated medical goals. Thus, it is meaningful to design a generic framework for IBSNs that are reconfigurable to suit particular application scenarios.

Among the available literature, a wireless microimplants network system was proposed by [22] for neuromuscular electrical stimulation. This functional electrical stimulation system was developed as a rehabilitation therapy for patients who had stroke and spinal cord injuries. This system can have up to 20 miniaturized (2-mm diameter × 16-mm long) implants that can be configured dynamically to support motor sensing and stimulation functions. The system adopts wireless telemetry for power transfer and bidirectional data communication. Dual carrier frequencies are used for forward and reverse data telemetry, that is 480 kHz for forward data transfer at a rate of 120 kb/s and 400 MHz for reverse date transfer at a rate of 480 kb/s.

C. Why Is an IBSN System Needed?

An IBSN integrates BSN and implantable medical sensors and devices to form an in-body network for diagnosis and treatment purposes. Though a number of proposed BSN systems have already discussed the possibility of an IBSN, the development of IBSN is still at the beginning stage. Due to its invasive nature, the general applicability might not be able to match that of an on-body BSN. But the acceptance will be higher for some particular end user groups such as critically ill patients, army special force members, and patients with life-threatening chronic conditions. Because IBSN employs multiple in-body sensors that measure the body physiology directly without using skin surface electrodes, the electrode polarization impedance is greatly reduced. Hence, an IBSN provides a more accurate snapshot of body condition than a BSN does. When it is implanted temporarily during an emergency procedure, IBSN provides an ideal solution for episodic events monitoring that is crucial for live rescuing. Nevertheless, its most applicable area in the foreseeable future could be in the animal experimentation for new drug discovery and toxicology testing. By adopting IBSN systems, the animals under test will live in their natural physiological statuses; thus, more accurate pharmacological response will be observed.

D. Types of Possible IBSNs

The possible IBSN systems can be classified into different types based on powering modes and types of implanted sensor nodes. Based upon the chosen powering mode, they could be active type which has an implanted power source and/or charging system, such as a battery. A passive type is an IBSN without an implanted power source, but can be powered or excited by an external power transfer mechanism, such as an inductive coil, or use an implanted power scavenging system. Based on the types of nodes used, the IBSN systems can be classified into physiological sensing networks and controlling networks. The physiological sensing networks are constructed by only using sensor nodes that have a similar architecture to many proposed on-body sensor networks. The controlling networks have both sensor and actuation nodes that form a closed-control loop mechanism by using the biofeedback from the sensor nodes. By having such a system, the disease condition could be controlled autonomously with minimal patient awareness and intervention. This is especially useful for some severe chronic disease patients such as diabetic patients and hypertension patients who need to closely regulate their glucose level and blood pressure.

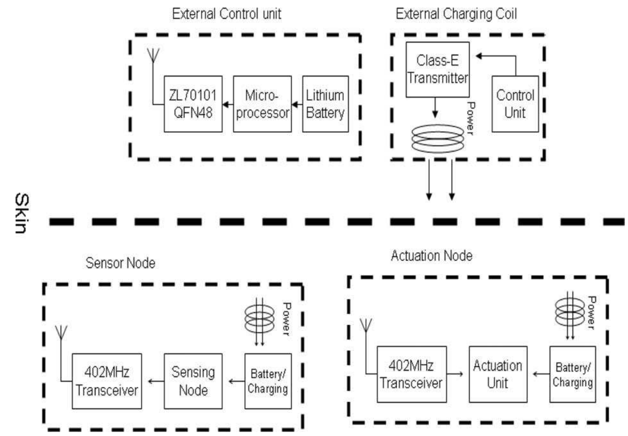


Fig. 1. System architecture of a biofeedback-based controlling IBSN. Zarlink ZL70101 402 MHz MICS transceiver is adopted in this design.

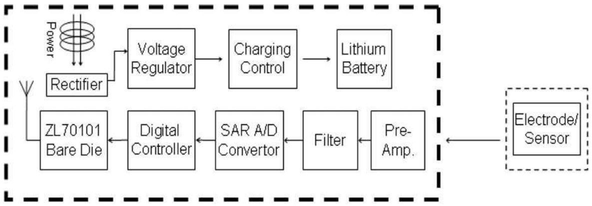
III. New Design of IBSN With Sensing and Actuation Nodes

A. System Architecture

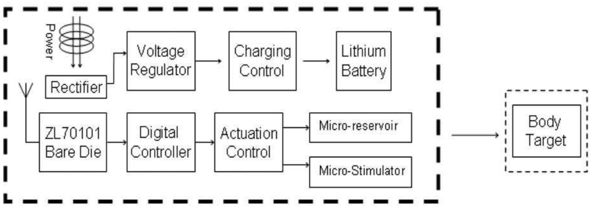
The design of an innovative IBSN system with both sensing and actuation nodes as a continuous monitoring platform for critically ill cardiovascular patients is presented here. The system consists of internal (in-body) and external (on-body) parts that can communicate each other using 402 MHz MICS band. Two types of sensing nodes, the ECG acquisition node and the blood pressure measurement node, are adopted. Both of these two nodes are SoC designed, which integrate electrode, analog front end, and inductive charging circuit. An implantable miniport with a wireless controllable valve can be connected to a system as the actuation node for drug delivery. The detected abnormal cardiac rhythms will activate the blood pressure measurement as well as trigger the status alarming and start the drug delivery. Fig. 1 shows the general system architecture.

The implantable system has a master–slave architecture. The nodes communicate with the external master control unit. Zarlink ZL70101 402 MHz MICS transceiver is world's first ultralow-power RF wireless chip designed specifically for implantable communication [23] at the MICS band It can be stacked on the sensing unit or the actuation unit to build up the network nodes. ZL70101 supports a typical raw data transmission rate of 200 to 800 kb/s which is sufficient to the needs of an IBSN with less than 100 nodes. Currently, the Zarlink transceiver is commercially available as an implantable-grade bare die.

Both acquisition node and actuation node are designed using analog SoC and MEMS (microelectromechanical systems) methodology and have amplifier, filter, and inductive charging circuit in common. For ECG node, it is a three-electrode biopotential acquisition system and for blood pressure node, a third-party surface-acoustic-wave device is adopted [13]. The acquired signals are amplified, filtered, digitized, and then sent



(a)



(b)

Fig. 2. (a) Architecture of a sensor node. The implanted lithium is charged by a transcutaneous power telemetry system. (b) Architecture of an actuation node. The implanted lithium is charged by a transcutaneous power telemetry system. The actuation unit can be either a microreservoir-based drug delivery system or a functional electrical stimulator.

out to an external receiver for recording by the implanted Zarlink ZL70101 402 MHz transceiver. For actuation nodes, the digital command data sent from an external master control unit are received and extracted and then fed into the control circuits to start a drug delivery process or a functional stimulation session. The block diagrams of the sensing node and the actuation node are shown in Fig. 2(a) and (b), respectively.

B. Battery and Transcutaneous Power Telemetry

Although many innovative energy scavenging approaches have been proposed [9], the associated technologies have not been made mature yet. Lithium-ion (LI) battery is still the best candidate for long-term active implantable biomedical systems. There are a few specifically designed reliable mini LI batteries commercially available, e.g., the latest Quallion LI battery has an outside diameter of 2.7 mm and a length of 13.0 mm. It has a capacity up to 10 mWh [24].

Though ZL70101 transceiver represents the latest design of low-power RF circuit, it requires 5 mA for continuous TX/RX and 1 mA for low-power mode which means that the Quallion LI battery will be exhausted in less than 10 h on working mode. Zarlink introduces an ultralow-power wakeup circuit that requires 250 nA current and operates on 2.5 GHz with the belief that the transceiver of an implant idles most of the time and activates with a large time interval (several hours or even weeks) [23]. Nevertheless, without a proper charging system, the implanted LI battery cannot last long enough to match the design requirement. The expected life span of an implanted system can vary from a few months to ten years. Thus, we incorporate a transcutaneous power telemetry to charge the implanted battery in this design. By using such a system, a lithium battery, such as the one from Quallion, can be charged regularly from outside the body. Therefore, the life span of the implanted battery can be extended substantially. The detailed charging system is described in Section IV.

IV. Key IBSN Design Issues

An IBSN system resides within a special and complicated environment, a human body. Many sound considerations for on-body sensor networks are not appropriate anymore. The design methodology and decision criteria need to adjust accordingly.

A. Wireless Communication for Medical Implants

Both MICS band and ISM band are widely used by various wearable and implantable medical systems. The MICS bands (range of 402–405 MHz) are the RF bands allocated by Federal Communications Commission for medical implant communication services while the ISM bands are assigned by International Telecommunication Union [25] for ISM applications. The centre frequencies of ISM bands range from 6.78 MHz to 245 GHz. The MICS band are mainly used by human and animal implantable devices while the ISM bands are used widely for a variety of applications. The popularly adopted RF identification (RFID) frequency (13.56 MHz), IEEE 802.15.1 (Bluetooth), and IEEE 802.15.4 (ZigBee) are all ISM bands.

ZigBee has been successful for wireless sensor networks as well as for on-body sensor networks. However, it is not an ideal candidate to support in-body (implantable) sensor networks due to the following reasons.

1) For implanted EMF sources, the surrounding tissues are near the radiation field range. The absorption of electromagnetic energy increases rapidly above 500 MHz due to the high losses caused by high water content in body tissues [26]. Thus, comparing with lower ISM bands or MICS band, it is not efficient to use 2.4 GHz ZigBee.
2) The 10 m covering range of ZigBee is a redundancy for a body area network. A much smaller covering radius ($<$2 m) is sufficient. The MICS band matches this requirement very well.
3) ZigBee supports 65,536 devices (nodes) per network through its 16-bit addressing mode; for most implantable applications, the required number of nodes is much smaller; thus, there is a great resource redundancy by using ZigBee's networking capability.
4) The typical ZigBee application is designed to have a single battery lifetime of one to two years (10–30 mA). However, the implantable devices may require the lifetime to be five years or even more.

Thus, 402 MHz MICS band is used in this design to replace the popularly used 2.4 GHz ZigBee. The MICS band has been already popularly used by many standalone implantable sensors such as glucose monitor, blood pressure monitor, and bionic eyes. MICS band has a maximum bandwidth of 300 kHz and a typical coverage range of 2 m [27]. In order to ensure the safety, the maximum power is limited to 25 μW equivalent radiated power (ERP) in air. The ERP is defined as the transmitted signal power measured on external surface of human body (skin surface in many cases).

B. Inductive Coupling for Transcutaneous Power Charging

Inductive coupling is one traditional but still popularly adopted wireless communication technique for sending

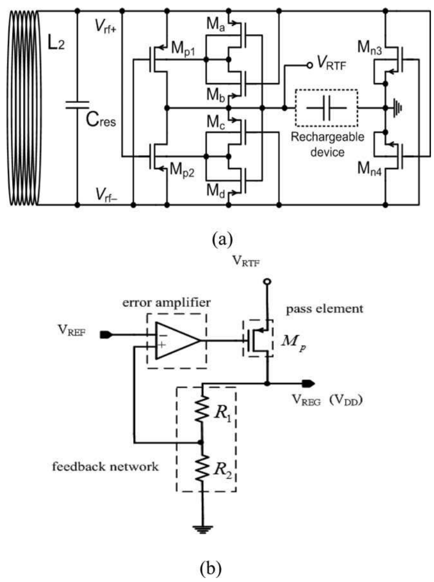


Fig. 3. (a) Self-driven bridge rectifier. (b) LDO regulator.

external commands to implants or retrieving acquired and stored data out of the body. This technique has been used in various implants that do not require high data transmission rate such as implantable microstimulator and implantable RFID transponder [28], [29]. Another popular use of inductive coupling technique is the transcutaneous wireless power transfer for an implanted system [30], [31]. Most transcutaneous telemetry systems use carrier frequency within either medium frequency (MF) or high frequency (HF) bands, such as 13.56 [29], 20, and 2 MHz [28].

The inductive transcutaneous charging system adopted in this design is similar to the one developed and implemented by one author in [32]. This charging system uses a Class-E amplifier with ASK modulations and a carrier frequency of 2 MHz to supply sufficient power for a low-coupling implantable SoC nodes and to charge the implanted battery. The relative lower frequencies used for power telemetry is of advantage to reduce the EMF energy loss due to the body tissue absorption which could generate hazardous thermal effects. A high-efficiency inductance coupling Class-E power amplifier has been commonly used for both wireless communication and power transmission [33], [34]. In order to achieve high-efficiency power transmission, there are many analysis methods for the Class-E amplifiers. Based on a load network synthesized to have transient response, it can maximize power efficiency even if the active device switching times are substantial fractions of the ac cycle [35].

As shown in Fig. 2, the proposed inductive wireless powering system contains a receiver coil, a rectifier, a voltage regulator, and a demodulation unit. The integrated full-wave rectifiers are comprised of diode-connected devices commonly used in RFID and implantable systems [36], [37]. Fig. 3(a) shows a self-driven synchronous rectifier that is adopted in this design. The voltage regulator receives the output voltage V_{rtf} from the rectifier and regulates the voltage to a stable and precise dc voltage source V_{reg} to recharge the battery or to power the implanted system.

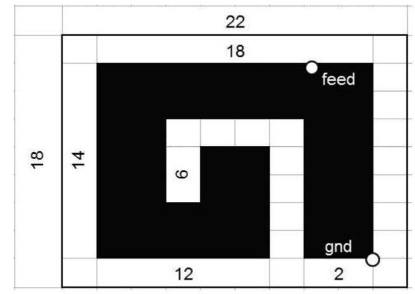


Fig. 4. Microstrip antenna working in an MICS band. The unit is in millimeters.

TABLE I
ANTENNA STRUCTURE FROM THE SIDE VIEW

	Thickness	Material	Dielectric Constant
Superstrate	4 mm	Rogers RO3210	10.2
Substrate	0.5 mm	Rogers RO3210	10.2
Ground	5 mm	Copper	N/A

Fig. 3(b) shows the chosen low dropout (LDO) regulator, which consists of an MOS pass element, an error amplifier, and one set of feedback network [38], [39].

C. RF Communication and MICS Antenna Design

Many active implantable medical systems, which can create a communication session from inside of a body, use RF-based communications for bidirectional data and command transfer. As mentioned earlier, the MICS band is chosen for wireless data communication in our design. The ZL70100 chip is a highly integrated MICS transceiver that requires a matched external antenna. There are a couple of key aspects to be considered when designing an implantable antenna. First, the antenna needs to be compact and able to integrate with the implantable devices easily. Conformability is another factor that needs to be considered in the design. Finally, an ability to use multiband is sometimes desirable for implantable devices. Based on those design constraints, a microstrip antenna optimized to work in an MICS band is designed to match the Zarlink chip for wireless data communication. A microstrip antenna normally consists of a ground plane, a substrate, and a microstrip line/patch. A superstrate is sometimes added to protect the patch/microstrip line from the undesirables. Besides, a superstrate can affect the resonant frequency of the antenna. In the case of an implantable antenna, a superstrate is highly essential to prevent the patch to be in contact with the human tissue. The design of the antenna was done using a CST microwave studio. Fig. 4 shows the designed MICS antenna working in an MICS band and Table I lists the parameters of the antenna structure from the side view.

In Fig. 4, the dark line is the microstrip line that is made of copper material with the thickness of 0.1 mm. The feed and ground pins were also made of copper material with the cylinder diameter of 0.5 mm. With the ground pin included in the design, this antenna can be categorized as a planar inverted-F antenna (PIFA).

D. SoC-Based Sensor Node

As Jovanov pointed out in [8], the realization of a miniature and lightweight sensor node posed one of the most challenging tasks for designers. This problem becomes more critical while developing an IBSN. Apparently, the popularly proposed Telos Mote-based sensor node is not appropriate for such applications. It is necessary to adopt SoC [40] or system-in-package approaches to design miniaturized sensor nodes. SoC is a mainstream technology in which the whole circuit, including all RF and analog/digital mixed-signal components, was integrated on a single piece of silicon. Because of the highly integrated and single-chip characteristics of SoC, biomedical devices receive the most benefits from SoC technology especially when they are intended to be implantable. However, a few theoretical and technical problems lie ahead, such as low-power VLSI design, power telemetry, microantenna design, etc. While many current research efforts are focused on solving those problems, it is reasonable to expect an explosive growth of SoC applied to biomicrosystem in the near future.

Fig. 2(a) illustrates the designed sensing node which is composed of sensing electrodes, a preamplifier, an OTA-C filter, a successive-approximation A/D converter (SA ADC), a controlling unit with modulation function, and a Zarlink MICS transmitter. As the ASK modulation is characterized by simplicity and low power consumption, the digital output of the SA ADC will be modulated by the ASK scheme and transmitted to the external receiver for data analysis and recording by using a Zarlink MICS transmitter. The ZL70701 bare die is adopted to minimize the size. An ASK modulator developed in our previous work [41] is adopted in this design. A carrier frequency is generated from a ring oscillator and a divider. Then, through an AND logical gate, digital data from the SA ADC are modulated and sent to the MICS transmitter.

V. In-Body EMF Exposure and Safety Consideration

A. Body EMF Exposure Safety Guidelines

Recent years have witnessed the unprecedented increase of human exposure to EMF due to the widespread use of electrical appliances and mobile telecommunication devices. Many national and international organizations have enacted safety guidelines based on recognized scientific research outcomes. The IEEE C95.1-2005 [42] standard specifies the safety levels of human RF exposure from 2 kHz to 300 GHz. The International Commission on nonionizing radiation protection (ICNIRP) also published its guidelines for limiting exposure to time-varying EMF up to 300 GHz [43]. The appropriate physical parameters used to specify the exposure restrictions are frequency dependent. Given the fact that most implantable medical systems have a wireless communication frequency between 100 kHz and 10 GHz, the specific absorption rate (SAR) and current density (for frequency up to 10 MHz) are taken on.

ICNIRP guidelines define basic restriction as well as reference levels for general public exposure and occupational exposure, respectively. For a frequency range from 100 kHz to 10 GHz, the basic exposure restrictions recommended by ICNIRP for general public are 0.08 W/kg for whole body average SAR and 2 W/kg for localized SAR (head and trunk), while the basic restrictions for occupational exposures are 0.4 W/kg for whole average SAR and 10 W/kg for localized SAR. For frequency from 100 kHz to 10 MHz, restrictions on current density are set to f/100 mA m-2 for occupational exposure and f/500 mA m-2 for general public exposure.

Since 2005, the IEEE C95.1 standard has been harmonized with ICNIRP guidelines with respect to SAR limits in the frequency range from 100 kHz to 3 GHz. The IEEE C95.1-2005 uses the maximum permissible exposure (MPE) to restrict the external fields and induced current quantities such as electrical field strength, magnetic field strength, and power density for both controlled and uncontrolled environments. The MPE restrictions can be derived from SAR restrictions. Based on IEEE C95.1-2005, the human exposures to implanted medical systems are considered to be partial body exposures in an uncontrolled environment. The IEEE also provides detailed recommendations for numerical computation of spatial-averaged SAR [43]. The occupational exposure restriction (10 W/kg for localized SAR) is generally followed during the design process due to the consideration that medical implants are a part of the medical procedures controlled by medical professionals.

Many implantable system designers follow both the IEEE C95.1 and ICNIRP guidelines to address the EMF exposure safety issues. However, those guidelines are based on scientific research such as epidemiological studies, residential cancer studies, laboratory studies, and cellular and animal studies that mostly use external EMF sources [42], [44]. Though some investigations on *in vitro* and *in vivo* effects have been performed, few experiments were done using internal sources. Moreover, those guidelines mainly consider the thermal effects of EMF, i.e., the increase of tissue temperature due to the EMF exposure. The reported athermal interactions were not taken into account during the guideline formation with the belief that there are no sufficient and consistent evidence to support the biological significance of athermal effects [42]. Nevertheless, it is worth to point out that medical implants reside inside human body for a very long term (many years) and have a close spatial proximity to many key organs. The current research on EMF interaction with biomacromolecules such as protein and DNA is far from sufficient. Thus, particular caution shall be taken while designing implantable medical systems based on those published guidelines.

B. EMF Exposure Experiment From Internal Sources and Human Body Modeling: A Brief Review

It is commonly known that the human body absorbs energy from electromagnetic waves generated by any transmission circuits. Thus, the electrical properties of human tissue have to be considered while designing any implantable biomedical devices. The human body has a complicated shape and is of heterogeneous nature due to a variety of tissues and organs that have different permittivity, conductivity, and magnetic permeability. Different types of body tissue have different electrical properties. For instance, at 150 MHz, the blood has the conductivity of

1.644 S and the relative permittivity of 65.2, while the conductivity for fat is only 0.035 S and the relative permittivity is 5.79 [26]. With the rapid advancement of wireless communication technologies, many research efforts have been spent on human body EMF interaction with various EMFs and radiators [45]–[52]. Due to the limited condition imposed on human body and the difficulty to measure reference levels *in vivo*, few studies have been carried out on EMFs from implanted sources. As an alternative, computational human body models are widely used to investigate the body EMF interactions as an indirect approach.

There are three basic human body dielectric models, namely simple homogenous model, semisegmented model, and full-segmented model. In many practical cases, those models are used together to form a composite body model to reflect the actual case. The homogeneous body model is the most popularly used one due to its computational simplicity [48], [53]. In this model, the surrounding and adjacent tissues have homogeneous dielectric properties and the body shape can be determined using animation software.

In [49], RF absorption of a 418 MHz radio telemeter semi-implanted in the human vagina is compared with simulation results through human EMF absorption modeling. A composite human model was used in this investigation with three semisegmented models within 100, 200 and 300 mm varying tissue depths from the source, respectively, and one fully homogeneous 5-mm model (5-mm depth of skin) constructed as three layers (skin, fat, and muscle). Based on the assumption that the source-surrounded tissues are responsible for most of the bulk absorption loss, a full segment human body mode was not used in this study. The comparison results found that the radiation pattern and field strength do not vary much in different segment depths which in turn justify the original assumption made on the composite human model.

Another investigation on the electromagnetic radiation inside human body (inside intestine) was done in [26]. The frequency range under investigation is between 150 MHz and 1.2 GHz. Comparing to the method utilized in [49], the human modeling in this investigation was also constructed using the finite-difference time-domain method based on the data from the visible human project but simulated using full-segment body mode. Both near-field and far-field performances are taken into consideration.

C. SAR Estimation Using a Three-Layered Arm Model

Because the chosen sensing nodes in this proposed IBSN are for ECG and blood pressure measurements, the implantation location are on limbs, an arm model consisting of three layers (skin, fat, and muscle) is proposed (see Fig. 5). Each layer of this human arm model has its own electrical properties and they are frequency dependent. The two most important electrical properties that will influence the simulation results are dielectric constant ε_r and conductivity σ. Table II shows the electrical properties at 402 MHz of each material that compose a human arm model [54].

For SAR calculation, a set of 3-D rectangular layers were created (see Fig. 6) to mimic the three layers of the human arm. The reason of using these layers instead of the whole spherical arm model is to make the simulation more efficient, while the attributes are still the same for all corresponding layers.

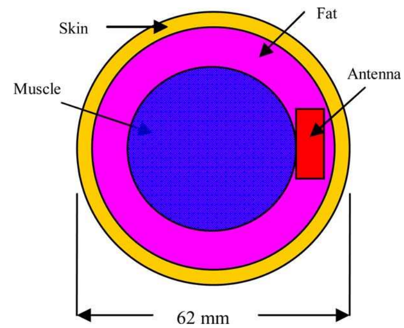


Fig. 5. Spherical arm model used for the simulation.

TABLE II
ELECTRICAL PROPERTIES OF EACH MATERIAL THAT COMPOSE HUMAN ARM

	Thickness (mm)	Relative permittivity ε_r	Conductivity σ (S/m)	ρ (kg/m^3)
Skin	2.3	46.741	0.68892	1100
Fat	9	5.5789	0.041151	916
Muscle	20	57.112	0.79682	1041

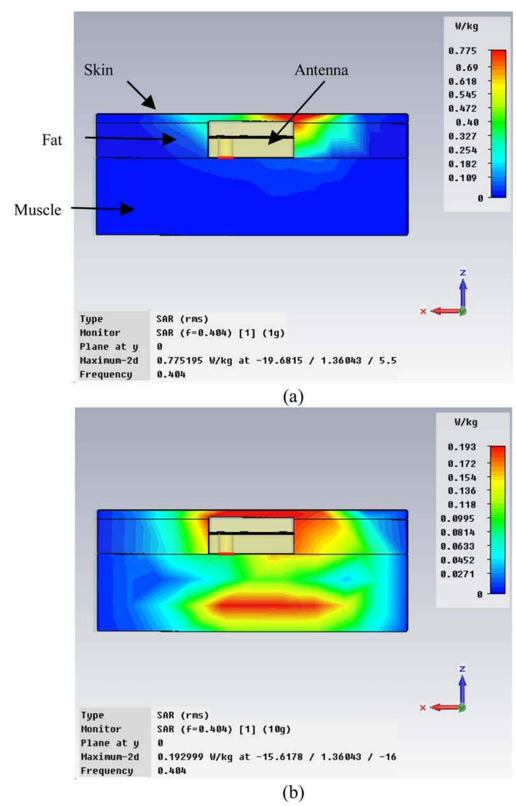


Fig. 6. (a) Average 1 g SAR and (b) average 10 g SAR distribution patterns from the side of the model. The different parts from the arm model are labeled in (a).

Two SAR values, the average 10 g SAR value and the 1 g SAR value, were investigated following the CST simulations, which are based on finite integration technique. These two parameters are very important because they are the standards used

TABLE III
SAR VALUES OF THE HUMAN ARM MODEL IN FOUR DIFFERENT POWER LEVELS

Power	10 g Average SAR (W/kg)	1 g Average SAR (W/kg)
1 W	19.3	121
100 mW	1.93	12.1
10 mW	0.193	1.21
1 mW	0.01933	0.121
Limit	2	1.6

in Europe and U.S., respectively, for determining the safety level of electromagnetic radiation exposure. For each parameter, four different power source levels were applied (from 1 mW to 1 W logarithmically) to understand what limit of power level is acceptable in terms of SAR values. After performing the simulations, it was learnt that as the source power was increased, the maximum SAR value also increased. This phenomenon occurred on both 10 and 1 g average SAR calculation. With the safety level stands at 2 W/kg (10 g average SAR) and 1.6 W/kg (1 g average SAR), the 100 mW and 1 W power sources cannot be considered due to their overlimit absorption values (see Table III). Fig. 6(a) and (b) shows the results of EMF simulation inside the human arm with the designed PIFA antenna implanted between the fat and muscle layers using CST microwave studio software at 402 MHz MICS band at the power level of 10 mW. As can be seen from Fig. 6 (a) and (b), there are two distinct patterns for different measurements. For the 1 g measurement, the high absorption rate occurred mostly at the top side of the model, which is mostly the skin layer, while for the 10g, it occurred almost evenly throughout the model even though the maximum value is smaller. In relation to Fig. 5, the results indicate that for extended period of time, the skin layer will absorb more energy relative to the other layers. The energy absorbed by the skin will eventually translate into heat. These results were generated with the power level of 10 mW at the input port. For different power levels, readers refer to Table III.

VI. CONCLUSION

The IBSN is a further development of the popular body area sensor network. It provides a new technological platform for vital physiology monitoring, illness treatment, and rehabilitation therapy. An IBSN system can also be implanted temporarily during emergent surgical procedures for episodic events monitoring. Moreover, it can play a great role for new drug design and testing while being implanted in testing animal bodies.

The development of a bioimplantable body sensor network is still at its inception stage. Many developed technologies for on-body sensor networks are not suitable for their in-body counterparts. The design obstacles that need to be addressed while designing and implementing the IBSN include in-body RF transmission, device miniaturization, low-power transcutaneous telemetry, low-power circuit design, and system integration and packaging. In order to optimally design the implantable electronic medical devices, the radiation characteristics of implanted EMF sources need to be fully understood. In the meantime, in order to ensure the safety guards set by international organizations such as ICNIRP, the SAR limits also need to be abided.

We present an innovative design of an IBSN, containing both acquisition nodes and actuation nodes to form a biofeedback control loop for critical disease monitoring and treatment. To our best knowledge, this is the first systematically reported design on such endeavor. This system incorporates a commercial available MICS transceiver, an SoC sensor node, and a wireless transcutaneous power charging system. This system not only realizes the ultra-low-power requirement but also abides EMF safety recommendation set by ICNIRP. Currently, the SoC ECG sensor and the transcutaneous charging circuit have been successfully developed [41], [55], [56]. The future implementation work will focus on integrating the individual modules to complete the system and start the testing.

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