Ultrasonically Rechargeable Platforms for Closed-Loop Distributed Sensing and Actuation in the Human Body

Raffaele Guida Tommaso Melodia
Northeastern University, Boston, USA
{guidar, melodia}@ece.neu.edu

***Abstract*—In the near future, innovative medical therapies will be administered by means of intra-body wireless sensor networks of implantable medical devices (IMDs). However, realizing wireless networks with traditional implantable bio-sensors and actuators is challenging, because (i) they often rely on wired connections that are invasive and prone to infections; and (ii) they are powered by batteries that occupy most of the device volume and have a relatively short lifetime.**

This article reports on the design of a system of inter-connected implantable nodes that leverage ultrasonic wireless propagation to (i) be remotely recharged, removing the need for batteries; (ii) to create wireless communication links avoiding wires or radio-frequency (RF) connections that have poor performance when operating in tissues. We illustrate the design of the core building blocks to realize ultrasonically rechargeable medical sensors and actuators equipped with ultrasonic connectivity. We further demonstrate their use in a practical implementation of a sensing/actuation closed-loop system with distributed sensor nodes. We also develop and experimentally validate a mathematical model to predict the system performance.

***Index Terms*—UTET, ultrasonic communications, intra-body wireless sensor networks**

I. Introduction

In the near future, the administration of both traditional and innovative medical therapies will be entrusted to wireless networks of smart implantable medical devices (IMDs) [1], [2]. These will consist of multiple miniaturized and wirelessly interconnected devices implanted inside the human body. Each node will perform operations including sensing, processing, actuation, and data communication.

As of today, however, implantable biomedical devices are largely based on stand-alone systems whose reliability could be improved by fusing data coming from multiple heterogeneous sensors deployed at different locations inside the human body. Therapies could be adaptively adjusted based on these readings, thus increasing their efficacy [3]. However, intra-body wireless networks based on modern IMDs have, as of today, only been demonstrated as a proof of concept. Two fundamental challenges impede the technical realization and commercialization of wirelessly networked IMDs: (i) the lack of new, reliable, less invasive batteries to power the implants, and (ii) the absence of effective wireless technologies to reliably interconnect heterogeneous nodes *in vivo*. With regard to the first challenge, traditional medical systems are, in fact, powered by batteries that occupy most of the device volume and periodically need to be replaced through surgery with associated clinical costs and risks for the patient [4]. In addition, futuristic IMDs will implement new functionalities, e.g., advanced data processing, wireless communication and networking, real-time reprogrammability, among others, that will result in additional power drain and therefore require larger batteries. As for the communication limitations, the majority of IMDs currently available on the market do not have wireless intra-body connectivity. For example, many cardiac pacemakers are wired-connected to an implantable electrocardiogram (ECG) sensor and to leads that are prone to infections and dislodgment [5], [6].

Solutions based on electromagnetic (EM) waves have been demonstrated and applied. Radio frequency (RF) based communication is an alternative to wires, whereas transcutaneous energy transfer (TET) by means of electromagnetic fields (EMF) is suitable for remote powering of batteryless IMDs [7]. However, biological tissues significantly attenuate EM waves and distort RF signals, and EM compatibility and cybersecurity are two other major concerns relative RF communications for IMDs.

Ultrasonic propagation represents a promising alternative to EM fields and RF for both connectivity and TET (TET is specifically called ultrasonic transcutaneous energy transfer (UTET) in this case). The main advantage of using ultrasonic waves for implantable devices and intra-body networks is that they are characterized by a significantly lower absorption rate with respect to RF [2], which means that less energy is transferred from the wave to the tissues during the propagation. Consequently, ultrasonic waves are safer and more energy efficient than EM and RF counterparts. For this reason, the Food and Drug Administration (FDA) allows ultrasonic power intensity of two orders of magnitude higher than RF in human tissues exposed to the radiation ($720\,\mathrm{mW/cm^2}$ for ultrasounds and $10\,\mathrm{mW/cm^2}$ at RF).

In this paper, we propose the design of the core building blocks of a sensing/actuation ultrasonic closed-loop system. We present a general architecture that can be adapted to build batteryless bio-sensors and actuators that are: (i) efficiently powered via UTET method, (ii) equipped with an ultrasonic communication system. We present a practical

This material is based upon work supported in part by the National Science Foundation under Grants CAREER CNS-1253309 and CNS-1458019.

978-1-5386-3512-4/18/$31.00 ©2018 IEEE

implementation demonstrating how to realize an implantable closed-loop network with distributed wireless sensor nodes and a wireless actuator central node. All the nodes in the network are ultrasonically rechargeable and we showcase how the UTET method needs to be combined with an ultrasonic communication transceiver. We also define and validate through experiments a mathematical model to predict the system performance given the transmitted power and the energy storage capacity.

The remainder of this work is organized as follows: in Section II we report the related work, in Section III we propose our system, node architectures, and mathematical model. Section IV illustrates the hardware and Section V reports the system evaluation and model validation through experimental results. We conclude the paper in Section VI.

II. Related Work

A. Transcutaneous Energy Transfer

Several attempts to meet the power requirements of IMDs (as reported in Table I) through Transcutaneous Energy Transfer (TET) methods have been presented in the literature.

Table I: Powering requirements of different IMDs [8], [9]

IMD	Power
Neurostimulator	0.42–100 mW
Pacemaker	1–8 μW
ICD*	5–10 W
Endoscopy capsules	550–1200 mW

**implantable cardioverter-defibrillator*

Denisov et al. [10] conducted a study to compare ultrasonic and inductive wireless approaches to powering biomedical implants. They showed that, while inductive coupling is more effective for superficial implants (1 cm), UTET solutions work better for deep implants (10 cm). In [11] ultrasound is used to directly power an implanted load, and the system is able to transfer 100 mW of power to the implant. *In vitro* and *in vivo* studies have also been reported, for example in [12]. In this work, the recharging of a Li-ion battery by means of ultrasonic waves at a depth of 1–2 cm was demonstrated in vivo. Quantitative results show that a 4.1 V battery half depleted can be recharged with 300 mW in about 2 hours, achieving an average efficiency of 20%. In [13], an ultrasonic link at 700 kHz is used to demonstrate the remote charging of a 0.22 F supercapacitor in deeply implanted medical devices.

B. Data Connectivity

Table II: Biosensor sampling parameters [3]

Signal	Sampling rate [Hz]	Resolution (bits)	Data rate (bits/s)
ECG (per lead)	120–250	12	1,440–3,000
Temperature	0.2–2	12	2.4–24
Oximetry	60	12	720
Blood pressure	120	12	1,440
Respiratory rate	20	12	240
Heart rate	10	12	120

In Table II the most common biosensor parameters are listed. TET systems with a data transfer link have been discussed in some literature studies. In [14] an ultrasonic WPT scheme with a backward data link is illustrated. The data link is realized by varying the impedance of the receiver that, in this way, reflects part of the energy, that can be used to carry information back to the transmitter. Charthad et al. [15] reported an ultrasonically rechargeable implant with a hybrid bi-directional communication link. The system has ultrasonic downlink and ultra-wideband (UWB) RF uplink. In the European project ULTRAsponder [16], a system was developed that leverages ultrasonic wireless transmission for both energy transmission and communication from the IMD to the external unit. In ULTRAsponder, communication is based on the passive backscattering technique. The advantage of backscattering is that the implant does not need energy storage, but the range of applications is limited, and intra-body networking is not possible. In [17], an UTET rechargeable platform with ultrasonic connectivity is proposed. However, the system uses two different transducers, one for charging and one for communication and is not reprogrammable. The implantable device discussed in this paper, instead, includes a field programmable gate array (FPGA) that makes it reprogrammable and more versatile and it uses only one transducer.

III. System Model

In this section the general model of an intra-body ultrasonic closed-loop network with UTET rechargeable nodes is described. This section then reports an architectural building blocks model adaptable to the design of both sensors and actuators. A mathematical model to predict the system performance and efficiency is finally derived.

A. Network Model

We designed a platform used as the basic building block to realize sensor and actuator nodes that exploit ultrasound (i) to be remotely powered by external charging units via UTET, (ii) to communicate with ultrasound through body organs and tissues to form an intra-body wireless network. The nodes allow to realize closed-loop feedback systems in which actuator nodes deliver a certain therapy or perform some operations based on the processing of the physiological data received from multiple, either heterogeneous or identical, sensor nodes implanted in other locations in the body.

In Figure 1 the model of a possible intra-body system configuration of sensors and actuators ultrasonically interconnected is illustrated. All nodes in the network include some from of electronic storage, e.g., rechargeable batteries, or supercapacitors that can be energized via UTET. The energy capacity required is application-dependent, and the sensor nodes need a smaller amount of energy as their operations are limited to sensing and sampling, and transmission to the central node. In the scenario in Figure 1 the actuator also acts as a central node, hence it has processing capabilities and larger storage to provide enough energy for actuation, processing, and communication.

B. System Operating Principle

The operation of the closed-loop sensing/actuation with UTET can be divided into three phases. In practice, two switches (see Figure 2) enable transitions from one phase to the other. The three phases are:

1) *UTET:* the nodes are wirelessly recharged by transferring power to them from an external charging station via ultrasonic waves.
2) *Sensing and data transmission*: the sensor nodes measure a certain biomarker or physiological parameter and transmit the digitized data to a central node over the wireless ultrasonic link.
3) *Processing and actuation*: the central node that has received data from multiple sensors processes the measurements and takes a decision on the action to perform. The actuator can be located on the central node itself, or on a separated platform that is wirelessly controlled by the central node.

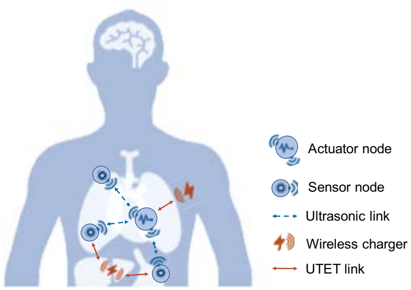

Figure 1: Model of an intra-body network of wirelessly rechargeable IMDs.

C. Architecture of the Nodes

The architectural model of the basic sensor or actuator node (Figure 2) is innovative as it combines an UTET technology with an ultrasonic connectivity module on the same system. The core building blocks of an UTET platform are reported in [13]. However, we enhance the basic UTET system with the following improvements: (i) by embedding basic components to interface it with sensors, actuators, and with the ultrasonic communication Internet of Medical Things (IoMT) platform described in [18]; (ii) by providing the system with the ability to switch from wireless charging to data processing and data communication; (iii) by realizing a more energy efficient platform, as it includes lumped components to match the receiving transducer to the load, larger energy capacity with similar form factor, and a low dropout (LDO) regulator to manage the stored energy.

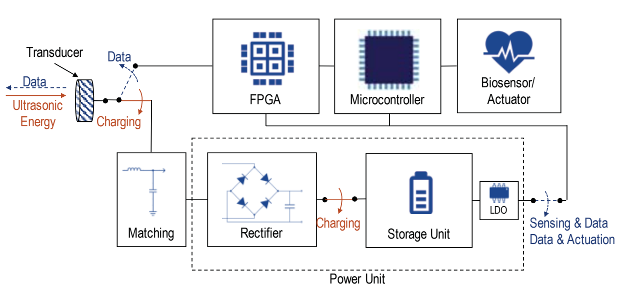

Figure 2: Block diagram of the actuator/sensor nodes.

Regarding the wireless charging, the implantable architecture includes a piezoelectric transducer, a power management unit, consisting of a rectifier circuit, the storage unit, and a low dropout (LDO) regulator to limit the voltage delivered to the load (sensor, actuator, or processing circuitry).

The main building components of the implantable ultrasonic transceiver are an FPGA and a microcontroller (MCU) that enable communication and data processing operations (as reported in [18]) on the data received from other nodes or from a sensor directly connected to the board.

D. Efficiency Models

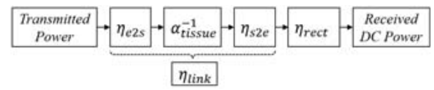

Figure 3: Block diagram of the charging efficiency model.

We evaluate the system performance using the two following definitions of efficiency.

- The *charging efficiency*, defined in (1), is the ratio between the energy accumulated into the battery or into the supercapacitor on the implant side (E_{stor}) and the total energy transmitted to recharge the storage component (E_{tx})

$$\eta_{ch} = \frac{E_{stor}}{E_{tx}}. \tag{1}$$

- The *operating efficiency* is defined in (2) as the ratio between the time (t_{on}) that a node can operate continuously with the energy received via UTET and the time needed to transmit that amount of energy to the implant (t_{ch})

$$\eta_{op} = \frac{t_{on}}{t_{ch}}. \tag{2}$$

t_{on} depends on the electrical characteristics of the load (for the IoMT PCB power requirements see [18]). We developed a model to predict the charging efficiency and the charging duration given the transmitted power and the energy storage capacity E_{stor} at the fixed implant depth of $5\ \mathrm{cm}$. To derive the amount of transmitted energy E_{tx}, we observe that the system is affected by some major sources of inefficiency that are taken into account with the variable η_{sys} (overall system efficiency). E_{tx} is, then, given by the product of the transmit power and t_{ch}. t_{ch}, in turn, is the ratio between the energy that the supercapacitor can store (E_{stor}) and the received DC power (after the rectifier). To obtain the received power from the transmit power (which is known), the last one has to be multiplied by η_{sys} to account for the inefficiency of the wireless link and of the rectifier. η_{sys} is defined as

$$\eta_{sys} = \eta_{link} \times \eta_{rect}, \tag{3}$$

where η_{rect} is the AC-to-DC conversion efficiency of the rectifier and η_{link} is expressed by the ratio

$$\eta_{link} = \frac{\eta_{e2s} \times \eta_{s2e}}{\alpha_{tissue}}, \tag{4}$$

where η_{e2s} and η_{s2e} are the electrical-to-acoustic and acoustic-to-electrical transduction efficiency of the piezoelectric converters, respectively, and α_{tissue} is the attenuation in human tissue. This model does not account for the losses at the interface between the transducers and the tissue and for the equivalent series resistance (ESR) of the supercapacitor. However, we minimized the acoustic mismatch by using a coupling gel at the transducers interfaces.

Table III: PCBs components details

Circuit	Component	Value/Model	#
Power Unit			
Rectifier	Diode	BAT-54A	2
	Smoothing cap.	0.01 µF	1
Storage circuit	2-input prioritizer	LTC4419	2
	Supercapacitor	330 mF (ESR* 75Ω)	1
Voltage regulation	LDO	TPS727	1
Communication Unit			
Processing/communication	Microcontroller	Freescale FRDM-KL03	1
Communication	FPGA	Lattice iCE40 Ultra	1

**Equivalent series resistance*

IV. Node Prototypes

In this section we briefly describe the hardware prototypes used to realize the ultrasonic sensor/actuator nodes according to the architecture in Figure 2. Each node is composed of two printed circuit boards (PCBs) illustrated in Figure 4.

We designed the energy management PCB prototype (on the left in Figure 4) and used the IoMT system with ultrasonic connectivity (identified with nr. 6 in Figure 4) reported in [18]. The PCB board implementing the receiver of the UTET method was designed using off-the-shelf electrical components. Core components are listed in Table III with their values. The sensing nodes include an implantable MPX2300DT1 pressure sensor (nr. 4 in Figure 4), and both the central and sensing nodes are powered by a $0.33\,\mathrm{mF}$ (5.5 V) supercapacitor (nr. 5 in Figure 4). We chose a supercapacitor instead of a battery given its longer lifetime, better cyclability (number of recharges), and because it is easier to recharge.

Figure 4: Building components of the sensor nodes. *1) Full wave rectifier, 2) LTC4419 prioritizer, 3) LDO, 4) MPX2300DT1 pressure sensor, 5)* $330\,\mathrm{mF}$ *supercapacitor, 6) IoMT communication board, 7) FRDM-KL03 microcontroller, 8) iCE40 Ultra FPGA*

The IoMT communication unit includes two main components, a Freescale Kinetis KL03 ultra-low-power microcontroller and a Lattice Semiconductor iCE40 Ultra FPGA (nr. 7 and nr. 8 in Figure 4, respectively). The first manages ADC data acquisition and processing operations, whereas the FPGA mainly implements the physical and MAC layers of the ultrasonic wideband (UsWB) communication scheme [18].

V. Experimental Results

In this section we report experimental results obtained using the testbed in Figure 5 and compare them with the analytical values calculated with the model discussed in Section III. Each node uses one ultrasonic transducer (American Piezo) both for energy transfer and communications. The transducers are thin disks of $9.5\,\mathrm{mm}$ diameter and are designed to work at a central frequency of $700\,\mathrm{kHz}$ with $200\,\mathrm{kHz}$ of bandwidth. Their transduction efficiency was measured to be equal to 54.3%. To simulate the propagation of ultrasonic waves through human tissues, the nodes were placed at the opposite sides of an upper arm phantom (Blue Phantom) at a distance of $5\,\mathrm{cm}$. We assumed the attenuation through the phantom to be $\alpha_{tissue} = 1\mathrm{dB/cm/MHz}$, which at $700\,\mathrm{kHz}$ and $5\,\mathrm{cm}$ of distance equals to dividing the power (in [W]) by a factor of 5.12. The $1\,\mathrm{dB/cm/MHz}$ is justified since it is an intermediate value between the attenuation in soft tissues and the attenuation across fibers in muscles. The phantom mainly mimics muscle fibers but it also contains some vessels that reduce the attenuation. The rectifier efficiency varies with the load, and its average measured value is $\eta_{rect} = 40\%$ (value used in the analytical model).

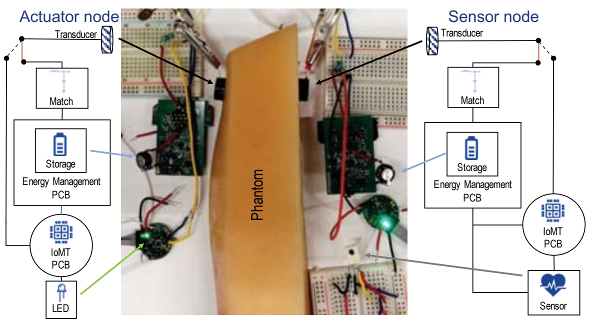

Figure 5: System testbed. On the left of the phantom the actuator prototype, on the right one of the two sensors.

A. Charging Efficiency

During the charging phase, the external charging station, composed of a waveform generator connected to a transducer through a power amplifier, transmits the energy to the implants wirelessly. The amount of acoustic power intensity, is kept below the FDA regulation limit ($720\,\mathrm{mW/cm^2}$) for all the experiments. The transmitted sound power P_s can be derived from the values of the transmitted electrical power P_{el} in Figure 6 or Figure 8 as: $P_s = P_{el} \times \eta_{e2a}/A_t$, where A_t is the surface of the transducer.

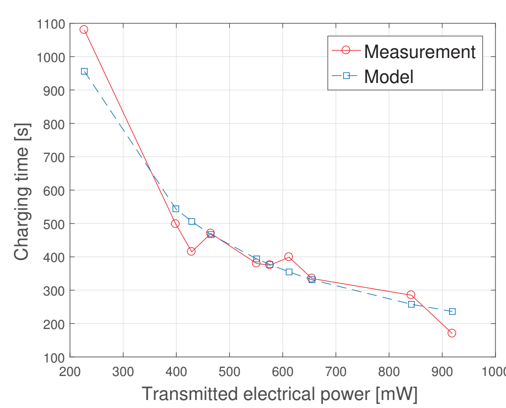

Figure 6: Duration of the charging phase vs. transmitted electrical power

In Figure 6 the charging times calculated with the mathematical model are compared to the measurements conducted with the the testbed. The measured charging efficiency η_{ch}, reported in Figure 7 varies between 2.04% and 3.20% around the calculated value of 2.3%.

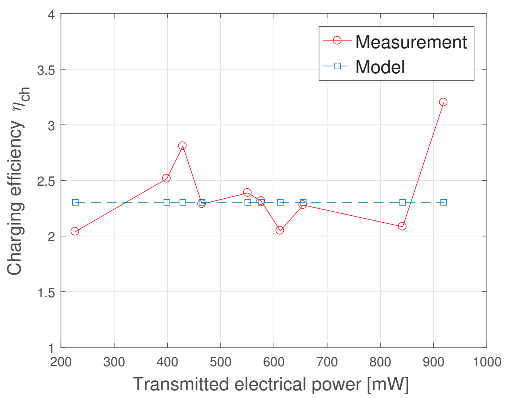

Figure 7: System charging efficiency as defined in (1)

B. Operating Efficiency

During the two phases following the charging, the sensor nodes take a measurement with the pressure sensor and send it to the receiving station. The latter processes the received data and performs a specific actuation operation accordingly. To showcase how the actuator performs different actions based on different results of the processed data, the node changes the blinking frequency of an LED on the IoMT board (Figure 5). In Figure 8 the values of the operating efficiency η_{op} for the sensing/transmitting and receiving/actuating phases are reported.

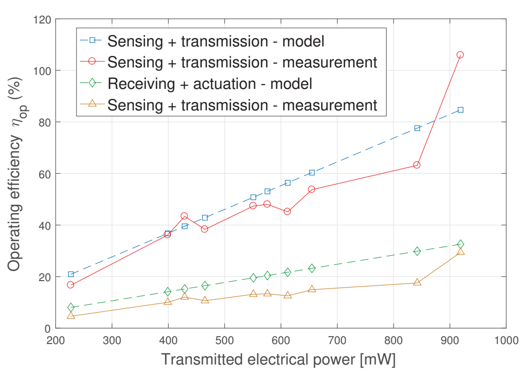

Figure 8: System operating efficiency as defined in (2)

VI. Conclusions

Next generation implantable medical devices (IMDs) will embed new, often energy-hungry functions. Therefore, traditional batteries will not be able to support futuristic IMDs and innovative powering methods will be needed. We presented the design and evaluation of a basic platform to build ultrasonically rechargeable medical implants provided with ultrasonic connectivity. A test bed consisting of a closed-loop system of nodes remotely powered via ultrasonic waves complying with the FDA exposure limits was presented. We developed a mathematical model to predict the system performance that showed an average charging efficiency of 2.3% and operating efficiency as high as 84.68%. We validated the mathematical model through testbed experiments, showing that the predicted values accurately match the measurements.

References

[1] F. Dressler and S. Fischer, "Connecting in-body nano communication with body area networks: challenges and opportunities of the internet of nano things," *Nano Communication Networks*, vol. 6, no. 2, pp. 29–38, 2015.

[2] L. Galluccio, T. Melodia, S. Palazzo, and G. E. Santagati, "Challenges and implications of using ultrasonic communications in intra-body area networks," in *Wireless On-demand Network Systems and Services (WONS), 2012 9th Annual Conference on*, pp. 182–189, IEEE, 2012.

[3] G.-Z. Yang, *Body Sensor Networks*. Springer, 2014.

[4] A. Kim, M. Ochoa, R. Rahimi, and B. Ziaie, "New and emerging energy sources for implantable wireless microdevices," *Access, IEEE*, vol. 3, pp. 89–98, 2015.

[5] C.-P. Lau, C.-W. Siu, and H.-F. Tse, "Future of implantable devices for cardiac rhythm management," *Circulation*, vol. 129, no. 7, pp. 811–822, 2014.

[6] F. Zanon, E. Baracca, *et al.*, "Multipoint pacing by a left ventricular quadripolar lead improves the acute hemodynamic response to crt compared with conventional biventricular pacing at any site," *Heart Rhythm*, vol. 12, no. 5, pp. 975–981, 2015.

[7] K. N. Bocan and E. Sejdić, "Adaptive transcutaneous power transfer to implantable devices: A state of the art review," *Sensors*, vol. 16, no. 3, p. 393, 2016.

[8] A. B. Amar, A. B. Kouki, and H. Cao, "Power Approaches for Implantable Medical Devices," *Sensors*, vol. 15, no. 11, pp. 28889–28914, 2015.

[9] W. Chen, G. Yan, S. He, Q. Ke, Z. Wang, H. Liu, and P. Jiang, "Wireless powered capsule endoscopy for colon diagnosis and treatment," *Physiological measurement*, vol. 34, no. 11, p. 1545, 2013.

[10] A. Denisov and E. Yeatman, "Ultrasonic vs. inductive power delivery for miniature biomedical implants," in *Body Sensor Networks (BSN), 2010 International Conference on*, pp. 84–89, IEEE, 2010.

[11] S. Ozeri, D. Shmilovitz, S. Singer, and C.-C. Wang, "Ultrasonic transcutaneous energy transfer using a continuous wave 650khz gaussian shaded transmitter," *Ultrasonics*, vol. 50, no. 7, pp. 666–674, 2010.

[12] L. Radziemski and I. R. S. Makin, "In vivo demonstration of ultrasound power delivery to charge implanted medical devices via acute and survival porcine studies," *Ultrasonics*, vol. 64, pp. 1–9, 2016.

[13] R. Guida, G. E. Santagati, and T. Melodia, "A 700 kHz ultrasonic link for wireless powering of implantable medical devices," in *SENSORS, 2016 IEEE*, pp. 1–3, IEEE, 2016.

[14] S. Ozeri and D. Shmilovitz, "Simultaneous backward data transmission and power harvesting in an ultrasonic transcutaneous energy transfer link employing acoustically dependent electric impedance modulation," *Ultrasonics*, vol. 54, no. 7, pp. 1929–1937, 2014.

[15] J. Charthad, M. Weber, T. C. Chang, M. Saadat, and A. Arbabian, "A mm-sized implantable device with ultrasonic energy transfer and rf data uplink for high-power applications," in *Proc. of IEEE Custom Integrated Circuits Conference (CICC),e*, pp. 1–4, Sept 2014.

[16] ULTRAsponder, "Invivo ultrasonic transponder system for biomedical applications invivo ultrasonic transponder system," 2008.

[17] F. Mazzilli, M. Peisino, R. Mitouassiwou, *et al.*, "In-vitro platform to study ultrasound as source for wireless energy transfer and communication for implanted medical devices," in *Engineering in Medicine and Biology Society (EMBC), 2010 Annual International Conference of the IEEE*, pp. 3751–3754, IEEE, 2010.

[18] G. E. Santagati and T. Melodia, "An Implantable Low-Power Ultrasonic Platform for the Internet of Medical Things," in *Proc. of IEEE Conference on Computer Communications (INFOCOM)*, (Atlanta, USA), May 2017.