ULTRASONIC NETWORKING FOR E-HEALTH APPLICATIONS

G. ENRICO SANTAGATI AND TOMMASO MELODIA, STATE UNIVERSITY OF NEW YORK AT BUFFALO
LAURA GALLUCCIO AND SERGIO PALAZZO, UNIVERSITY OF CATANIA

ABSTRACT

Wirelessly networked systems of intra-body sensors and actuators could enable revolutionary applications at the intersection between biomedical science, networking, and control with a strong potential to advance medical treatment of major diseases of our times. Yet, most research to date has focused on communications *along the body surface* among devices interconnected through traditional electromagnetic radio-frequency (RF) carrier waves; while the underlying root challenge of enabling networked *intra-body* miniaturized sensors and actuators that communicate through body tissues is substantially unaddressed. The main obstacle to enabling this vision of networked implantable devices is posed by the physical nature of propagation in the human body. The human body is composed primarily (65 percent) of water, a medium through which RF electromagnetic waves do not easily propagate, even at relatively low frequencies. Therefore, in this article we take a different perspective and propose to investigate and study the use of ultrasonic waves to wirelessly internetwork intra-body devices. We discuss the fundamentals of ultrasonic propagation in tissues, and explore important tradeoffs, including the choice of a transmission frequency, transmission power, and transducer size. Then, we discuss future research challenges for ultrasonic networking of intra-body devices at the physical, medium access and network layers of the protocol stack.

INTRODUCTION

Body area networks (BAN), typically understood as networks of wearable wireless devices designed to enhance clinical applications, have received considerable attention in the last few years [1]. Even more intriguing is, however, the fascinating promise of a future where carefully engineered miniaturized biomedical devices implanted or ingested by humans are wirelessly internetworked to collect diagnostic information and to fine-tune medical treatments over extended periods of time.

Networked systems composed of *intra-body* sensors and actuators could enablerevolutionary applications. These include for example measuring the level of glucose in the blood of diabetic patients, and reactively controlling the administration of insulin through under-skin actuators; monitoring the digestive tract through pill-sized ingestible cameras or monitoring bone-growth for diabetes treatment. Furthermore, these technologies may enhance minimally-intrusive microsurgery, thus overcoming the limitations of currently available catheter technology.

Yet, most BAN research to date has focused on communications *along the body surface* using electromagnetic radio-frequency (RF) waves [2].

However, while application-specific implantable medical microsystems at the *sub – mm*^{3} scale with perpetual energy harvesting have been demonstrated [3], the underlying root challenge of enabling wireless networks of intra-body miniaturized sensors and actuators that communicate through body tissues is substantially unaddressed. The main obstacle to enabling this vision of networked implantable devices is posed by the dielectric nature of the human body, which is composed primarily (65 percent) of water, a medium through which RF electromagnetic waves notoriously do not easily propagate. In addition, the human body is known to distort and delay transmitted RF signals; and while the medical community is still divided on the real risks caused by exposure of individuals to RF radiations, high RF transmission powers are certainly undesirable because of serious health concerns. Accordingly, most research in this field has focused on developing techniques to reduce the radiated power to avoid overheating of tissues. Finally, there are significant electromagnetic compatibility concerns, since environmental interference (both malicious and unintended) may easily jam any form of RF communication.

These challenges cannot be overcome unless a major paradigm shift in networking through body tissues is made to address the limitations of RF propagation in the human body. Recent work has explored the potential of molecular and other bio-inspired communication techniques for low-rate communications at the micro and nano scale [4]. We take a different point of view and, for the first time, investigate the use of ultrasonic waves to wirelessly internetwork intra-body miniaturized devices. For this purpose, we discuss the fundamentals of ultrasonic propaga-

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tion in tissues, and explore important tradeoffs, including the choice of a transmission frequency, transmission power, and transducer size. In addition, we introduce a system architecture and discuss future research challenges for ultrasonic networking of intra-body devices at the physical, medium access and network layers of the protocol stack.

The rest of the article is organized as follows. We discuss ultrasonic waves and their interaction with human tissues. We discuss ultrasonic transmission. We discuss ultrasonic channel modeling. We outline the system design challenges of intra-body area networks at different layers of the protocol stack, and we illustrate two approaches for evaluation of ultrasonic intra-body area networks. Finally, we conclude the article.

Ultrasonic vs. Electromagnetic RF Communications

Potential advantages of using ultrasonic communications (UC) as compared to RF communications (RFC) can be summarized as follows:

- *Propagation*: Ultrasonic waves are subject to lower absorption as compared to electromagnetic waves, mainly because of the significant water content in human tissues. In [5, 6], attenuation values ranging from 20dB at 100 MHz to 60dB at 1GHz have been reported for distances less than 10 cm, which make RF-based communication inside the human body very difficult.
- *Health Concerns*: Ultrasounds have been successfully used for therapeutic and diagnostic purposes inside the human body since the 1960s with no known detrimental effects [7]. On the contrary, the medical community is still divided on the risks caused by exposure of human tissues to RF radiations.
- *Interference Management*: The radio spectrum is crowded. Therefore, environmental or malicious interference may potentially jam intra-body RF devices.

In the rest of this article, we will discuss the main communications and networking challenges to be addressed to enable ultrasonic networking in the human body.

Ultrasonic Propagation in Tissues

Ultrasonic waves originate from the propagation of mechanical vibrations of particles in an elastic medium at frequencies above the upper limit for human hearing, i.e., 20kHz. Acoustic propagation through a medium is governed by the acoustic wave equation (Helmholtz equation).

Acoustic waves are characterized through their physical parameters:

- *Amplitude* (A), i.e., the local displacement of particles from their rest position
- *Propagation speed* c, i.e., the rate at which the vibratory energy is transmitted in the direction of propagation
- *Pressure* (P) i.e., a measure of the compressions and rarefactions of the molecules in the medium through which sound waves propagate
- *Particle velocity* (u), i.e., the speed of particle oscillations around their rest position caused by wave propagation in a medium
- *Intensity* (I), i.e., the average energy carried over time by a wave per unit area normal to the direction of wave propagation

Health Concerns

It is important to understand the potential risks of radiating acoustic power through internal tissues. The most obvious effect is *heating*. A significant portion of the energy is absorbed and converted into heat when ultrasounds propagate. This could potentially lead to a temperature increase. As the wave intensity increases, the temperature rises and adverse biological effects may occur. However, no lethal effects have been observed for temperatures lower than 41°. Since heating is strictly caused by the wave intensity, pulsed transmissions with a low duty cycle can potentially reduce this effect of a factor proportional to the duration of the duty cycle.

Another phenomenon caused by ultrasound wave propagation is so-called *cavitation*, which denotes the behavior of gas bubbles within an acoustic field. Pressure variations of the ultrasound wave cause bubbles in the propagation medium to contract and expand.

Unfortunately, experimental data collected on the bio-effects of ultrasounds are frequently inconsistent and controversial. However, the medical experience of the last decades has demonstrated that ultrasounds are fundamentally *safe*, as long as acoustic power dissipation in tissue is less than $50\ J/cm^2$ [8]. Therefore, ultrasound communications in tissues at low transmission pressure levels, and consequently low transmission power levels, are not expected to cause any lethal bio-effects. More importantly, ultrasonic wave heat dissipation in tissues is minimal when compared to electromagnetic RF waves; we thus believe that ultrasonic communications can represent a feasible alternative to classical electromagnetic RF communications.

Attenuation

Two main mechanisms contribute to attenuation: absorption and scattering. Note that the phenomena responsible for ultrasound absorption in biological tissues (i.e., thermal conductance effects, chemical effects, viscous effects) are less than fully understood. In ultrasound propagation through absorbing media, the initial pressure, P_0 reduces to $P(d)$ at a distance d according to [9]

$$P(d) = P_0 e^{-\alpha d}, \quad (1)$$

where α (in [np · cm−1]) is a coefficient that captures all the effects that cause dissipation of energy from the ultrasound beam. Parameter α is a function of the carrier frequency as $\alpha = a f^b$, where f represents the carrier frequency and a (in [np m−1 MHz^{-b}]) and b are attenuation parameters characterizing the tissue. Parameter b is commonly assumed to be very close to one. For all practical purposes, a is a unique parameter characterizing the tissue. Typical experimental values of α, a, and b in different tissues are reported in Table 1.

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Tissue	a	a	b
Blood	0.095–0.13 @ 5 MHz	0.014–0.018	1.19–1.23
Heart	0.23 @ 1 MHz	0.23	1
Kidney	0.23 @ 2 MHz	0.115	1
Liver	0.17–0.57 @ 5MHz	0.041–0.07	0.9–1.3

Table 1. *Parameters characterizing body tissues.*

Reflections and Scattering

Whenever a sound beam encounters a boundary between two materials, some of the energy is reflected, thus reducing the amount of energy that actually passes through that boundary. The direction and magnitude of the reflected and refracted wave depend on the orientation of the boundary surface and on the different tissues. This phenomenon can be described by means of the Snell law for acoustics [10].

When an acoustic wave encounters an object that is relatively small with respect to the wavelength, or a tissue with an irregular surface, a phenomenon called *scattered reflection* occurs. Since the human body is composed of different organs and tissues, each of them with different sizes, densities and sound velocities, it can be modeled as an environment with a pervasive presence of reflectors and scatterers. Consequently, the received signal is obtained as the sum of numerous attenuated and delayed versions of the transmitted signal, which makes propagation in ultrasonic intra-body area networks deeply affected by multipath fading.

Ultrasonic Transmission

An ultrasonic transducer is a device capable of generating and receiving ultrasonic waves. It is essentially composed of an active element, a backing, and a wear plate. The *active element* is usually a piezoelectric material that converts electrical energy to mechanical energy and viceversa. The *backing* is a dense material used to absorb the energy radiated from the back face of the piezoelectric element. The *wear plate* protects the piezoelectric transducer element from environmental wear and corrosion.

Operating Frequency Selection

The acoustic radiation pattern of a transducer is a representation of the transducer sound pressure level as a function of the spatial angle and is related to parameters such as the carrier frequency, size and the shape of the vibrating surface. Furthermore, the acoustic radiation pattern is reciprocal, i.e., its shape is the same no matter if the transducer is working as a transmitter or as a receiver. Depending on the desired application, transducers can be designed to radiate sound according to different patterns, e.g., omnidirectional or not. For a transducer with a circular radiating surface, the narrowness of the beam pattern is an inverse function of the ratio of the diameter of the radiating surface and the wavelength at the operating frequency, D/λ. This relationship is described by the beam spread formula

$$\sin\left(\frac{\theta}{2}\right)=\frac{0.514c}{fD}, \quad (2)$$

where $\theta/2$ is half the angle between the –6dB points of the acoustic radiation pattern. The higher the frequency or the larger the transducer's diameter, the smaller the beam spread.

Since most sensing applications require highly directional transducers, operation at high frequencies is required to keep the transducer size small. Conversely, as discussed earlier, higher transmission frequencies lead to higher attenuation. Therefore, we need to operate at frequencies compatible with small devices, but at the same time limit the maximum attenuation. In Fig. 1, we depict the maximum carrier frequency with respect to distance, for different maximum attenuation values, with blood as the propagation medium, and with a limit on the maximum operating frequency of 1GHz. In Table 2, we summarize our results for a maximum 100 dB attenuation. Since attenuation increases as a function of frequency and distance, for a fixed value of attenuation we have an inverse relationship between frequency and distance. Therefore, for communication distances ranging from some μm up to a few mm (short range communications) frequencies higher than 1GHz can be considered. When distances are higher than 1mm but still lower than some cm (medium range communications) transmission frequencies should be decreased to approximately 100 MHz. For distances higher than a few cm (long range communications) the transmission frequency should not exceed 10 MHz. These results are independently confirmed in [11], where the authors, based on considerations on the physics of ultrasound propagation, observed that frequencies in the [10, 300] MHz range are suitable for short-range communications.

Energy Consumption

The acoustic energy radiated is tightly related to the (electrical) power absorption of the ultrasonic transducer in use. Considering a piezoelectric ceramic transducer, outside the region of resonance, this can be viewed (from the electrical point of view) as a parallel plate capacitor with capacitance C_0. Thus, the main source of power consumption comes from charging this capacitor [12]. Then, considering a capacitor with voltage supply V and charge/discharge repetition frequency f, the power consumption P_c can be expressed as $P_c = f C_0$.

We can then impose appropriate limits to the voltage supply to in turn limit the radiated acoustic power to safe levels. Considering as in [11] a maximum allowable power dissipation in tissue of 10^4 W/m^2, if we consider, for example, muscle tissue as the primary propagation medium, the maximum pressure magnitude P_{MAX} that can be radiated by the transducer is approximately 0.13MPa. Thus, we can derive the related transducer voltage input, corresponding to the maximum radiated pressure, through the constitutive equation of piezoelectric materials,

Communication range	Distance	Frequency Limit
Short Range	μm – mm	>1GHz
Medium Range	mm – cm	≃ 100 MHz
Long Range	> cm	≃ 10 MHz

Table 2. *Frequency limits for A = 100 dB.*

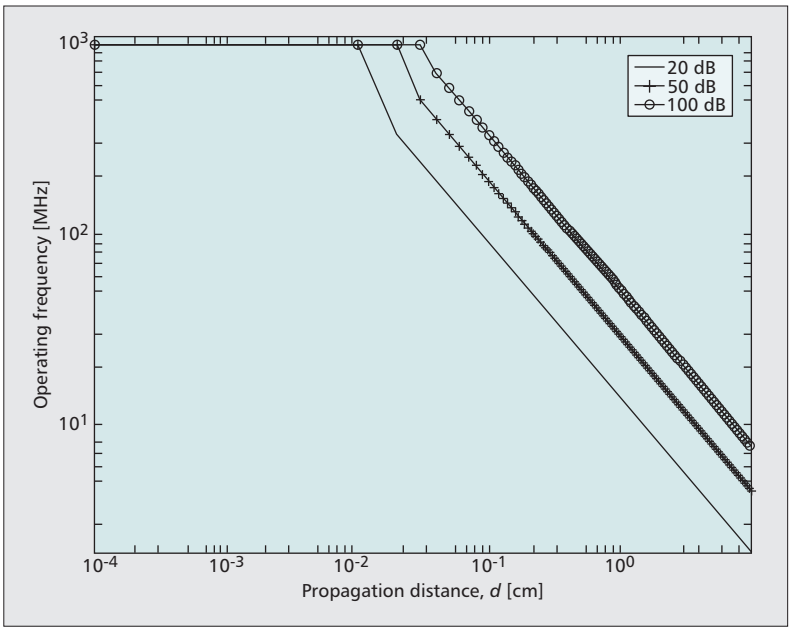

Figure 1. *Maximum carrier frequency in blood.*

which expresses the relationship among mechanical strain and electrical displacement for the piezoelectric element. One can accordingly derive the average power consumption of the transducer

$$\overline{P}_c = C_0 \, (g_{33} \, P_{MAX} \, t_h)^2 \, [\mathrm{W}]. \qquad (3)$$

where t_h is the thickness of the piezoelectric element, and g_{33} is a coefficient that relates the mechanical pressure with the electrical displacement.

Ultrasonic Channel Modeling

The first step towards the design of high-performance ultrasonic networked systems is to characterize in detail the ultrasonic propagation channel in tissues - for which, unfortunately, there is basically no literature available to date. Two major classes of channel models are typically considered, i.e.,

- *Statistical models*
- *Deterministic propagation models*

Statistical models are typically based on *measurement data*, while deterministic propagation models can be derived based on acoustic wave propagation theory (e.g., ray-tracing and full-wave electromagnetic numerical techniques [13]). In the following, we focus on the latter.

The propagation law of acoustic waves through biological tissues is governed by three coupled first-order equations, i.e., the *continuity equation*, the *force equation* and the *equation of state*, which represent relationships among acoustic pressure, P, acoustic particle velocity u, and medium density ρ, and can be rearranged to obtain the Helmhotz equation. A realistic channel model of ultrasonic signal propagation in human tissues that incorporates all the aspects discussed above including attenuation, scattering and multipath needs to satisfy the three first order equations simultaneously. The solution represents the acoustic field behavior in time, evaluated at each spatial coordinate of the propagation medium.

Traditionally, partial differential equations are solved using numerical methods such the finite-difference-method (FDM) based on a linear approximation of the field gradient and provides accurate results and spatial granularity at the expense of high computational complexity. A different, computationally more efficient approach is based on the pseudo-spectral and k-space methods. Basically, the pseudo-spectral (PS) method reduces computational complexity in the spatial domain by using Fourier series expansions and FFTs, while the k-space method operates in the time domain by using k-space propagator functions (instead of classical finite differences) to approximate temporal derivatives.

As a representative example of the approach discussed above, we modeled a section of the human arm, including bones, muscles, fat and skin. We considered an heterogeneous two-dimensional rectangular area of length 20cm and width 10cm, where density and sound velocity are distributed as shown in Fig. 2. The bone half-section has a 18mm width, the muscle half-section has a 22mm width, while fat and skin have a half-section of 7mm and 3mm width, respectively. We obtained the channel impulse response of the ultrasonic channel by releasing an ideal Dirac pulse in the left top corner of the muscle section. The receiver is located in the left bottom corner of the muscle section. Transmitter and receiver are located 20cm away from one another. We observe that the effect of the bone and tissues is to partially reflect and scatter the acoustic wave transmitted by the source. Figure 3 reports the resulting impulse response in this scenario. We observe that the effect of multipath and scattering is to introduce attenuated signal replicas spaced in time.

We can model the channel response as a complex-valued low-pass equivalent impulse response, as $h(\tau, t) = \Sigma_{k=1}^{K} \alpha_k(t) \delta(\tau - \tau_k) e^{j\theta_k(t)}$ where δ is the Dirac delta function, K is the number of resolvable multipath components, τ_k are the delays of the multipath components, α_k are the path amplitude values and θ_k are the path phase values. Based on this characterization, we can calculate the power delay profile (PDP), i.e., the squared magnitude of the impulse response and standard quantities that characterize the channel, i.e., the first and second central moment of the PDP, also known as

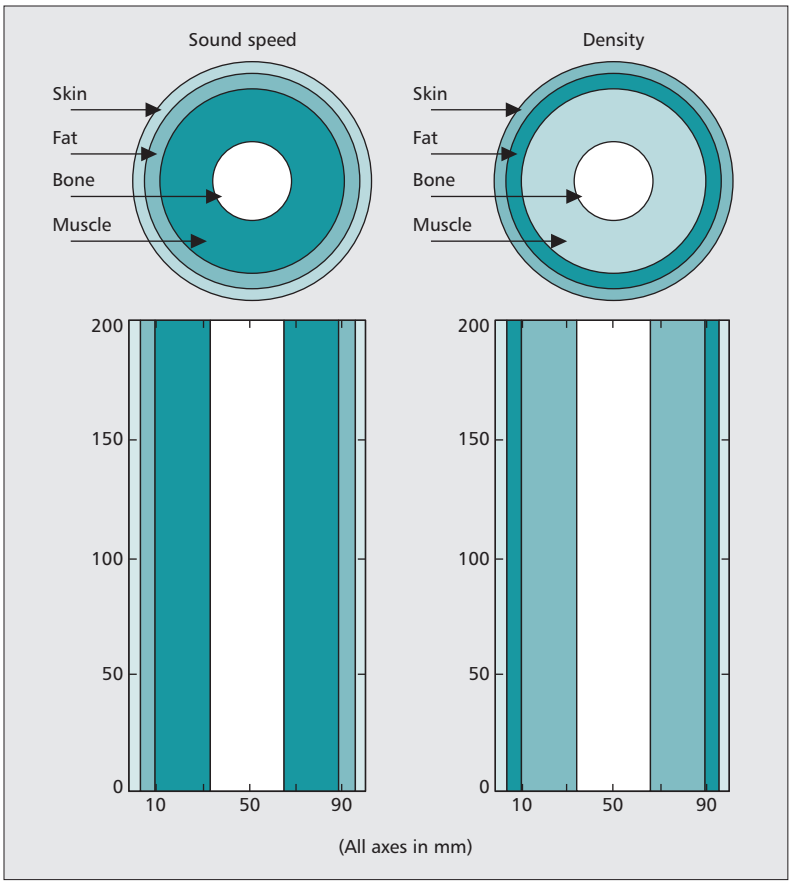

Figure 2. *Sound speed and density distributions in a simulated arm.*

the mean excess delay (τ_m) and the RMS delay spread (τ_{rms}). For example, for the arm channel simulated above, we obtained $\tau_m = 4.4353 \cdot 10^{-6}$ s and $\tau_{rms} = 2.3389 \cdot 10^{-6}$s. Since the coherence bandwidth of the channel is proportional to the inverse of τ_{rms}, *we should consider the above channel as frequency selective for signals of bandwidth above approximately* 85kHz.

System Design Challenges

As discussed in the previous sections, ultrasound propagation in tissues is deeply affected by multipath fading because of the inhomogeneity of the human body in terms of density and, consequently, sound velocity, and the pervasive presence of very small organs and particles. As discussed and observed earlier, reflectors and scatterers accurately model the obstacles encountered by signals propagating in the body. Therefore, numerous attenuated and delayed versions of the same transmitted signal reach the receiver, making detection and decoding a challenging operation. Moreover, the low speed of sound in tissues leads to high delays that have to be considered in system design.

Based on these observations and by taking into account the above described channel model as well as the energy concerns, in this section we discuss research challenges for ultrasonic networking of intra-body devices at the physical, medium access and network layers of the protocol stack.

Physical Layer Design

Physical layer transmission schemes need to be designed to achieve low-complexity and reliable communications against the effect of multipath reflections and scattering within the human body. It is also necessary to limit the thermal effects of radiations, which may be detrimental to human health. Broadly used carrier-modulated waveforms are strongly affected by multipath replicas that destructively overlap in time, and continuous waveform can also produce undesirable bio-effects. For these reasons, a pulse based low-duty-cycle transmission scheme appears to be a desirable solution.

Based on these considerations, we designed and proposed a new ultrasonic transmission and multiple access technique, which we refer to as Ultrasonic WideBand (UsWB) [14]. Ultrasonic WideBand is based on the idea of transmitting very short ultrasonic pulses with an adaptively controllable duty cycle following a pseudo-random adaptive time-hopping pattern, and with a spreading code of adaptive length superimposed. Impulsive transmission and spread-spectrum encoding combat the effects of multipath and scattering and introduce waveform diversity among interfering nodes.

Given the frequency-selective nature of the ultrasonic channel, a potential alternative to impulsive transmission is to rely on orthogonal frequency division multiplexing (OFDM) transmission techniques. Our current preference for impulsive transmissions is motivated by three core concerns. First, standard OFDM requires much more complex, digital processing-intensive transmitters and receivers, which may be less than fully appropriate for energy-constrained implantable devices. Second, impulsive transmissions for medical imaging and diagnostic are well-understood in the ultrasound community; and transceivers tested for impulsive transmissions are available off-the-shelf. Finally, low-duty-cycle schemes reduce the potential of undesirable bio-effects.

UsWB in a Nutshell — Consider, as in Fig. 4 a slotted time divided in chips of duration T_c, with chips organized in frames of duration $T_f = N_h \cdot T_c$, where N_h is the number of chips per frame. Each user transmits one pulse in one chip per frame, and determines in which chip to transmit based on its own pseudo-random time hopping sequence (THS). Since pulses may collide with a probability that depends on the frame size, N_h, we can superimpose an adaptive channel code to additionally reduce the effect of mutual interference from co-located devices. *By varying the coding rate, we can dynamically regulate the effective information rate to adapt to the characteristics of the ultrasonic channel and to the interference level*. Various channel coding solutions have been presented in [15] with different performance levels and computational complexity. We represent each information bit with *simple pseudo-orthogonal spreading codes of variable length*, N_s because of

- Their excellent, and well-understood multiple access performance
- Limited computational complexity
- Inherent resilience to multipath.

The resulting data rate is expressed as

$$R(N_h, N_s) = \frac{1}{N_s T_f} = \frac{1}{N_s N_h T_c}.$$

By regulating the TH frame length N_h, i.e., the average inter-pulse time, a user can adapt its transmission rate, and as a consequence modify the average radiated power and therefore the level of interference generated to other ongoing communications. Differently, regulating N_s, the number of chips, and then the number of pulses, representing one information bit, there is a tradeoff between robustness to multi-user interference (which increases with longer spreading codes), energy per bit consumption (that is increased by a factor N_s) and information rate.

Medium Access Control Layer Design

The design of a distributed, uncoordinated medium access control can enable multiple access and combat the effect of interference from multiple co-located and simultaneously transmitting devices, thus allowing users to share the communication medium fairly, guaranteeing maximum throughput and minimizing the protocol overhead. Unfortunately, the characteristics of the ultrasonic channel make the design of an efficient MAC protocol for in-body communications a challenging task. In particular the design of a medium access solution needs to take into account the following observations:

- *Frequency division*: Multipath fading strongly affects narrowband signals. Therefore, pure FDMA scheme does not seem to be a desirable solution.
- *Time division*: The high delays, caused by the low speed of sound in the medium, make implementation of an efficient TDMA scheme challenging, since any TDMA scheme would require long time guards to prevent collisions between adjacent time slots.
- *Carrier-sense*: Carrier-sense multiple access (CSMA) schemes would be strongly affected by the low propagation speed in tissues, since carrier sensing becomes inefficient with long propagation delays.
- *Code Division*: Since the different multipath copies of the signal of interest can be considered as disturbance, the use of spreading codes as in DSSS techniques may help the receiver discriminate among them, thus making DSSS signals immune to multipath fading.

The UsWB technique discussed above was designed based on the aforementioned observations. The use of a low-duty-cycle impulse-based transmission scheme together with a superimposed spreading code allows different transmitters to coexist on the same channel, and share the same available bandwidth, i.e., no frequency division is required. Dynamically adapting the time-hopping frame length and spreading code length allows to implement a MAC protocol that does not require mutual temporal exclusion between different transmitters. Moreover, the adaptation creates several tradeoffs among

- Resilience to multi-user interference and ultrasonic channel errors
- Achievable information
- Energy efficiency

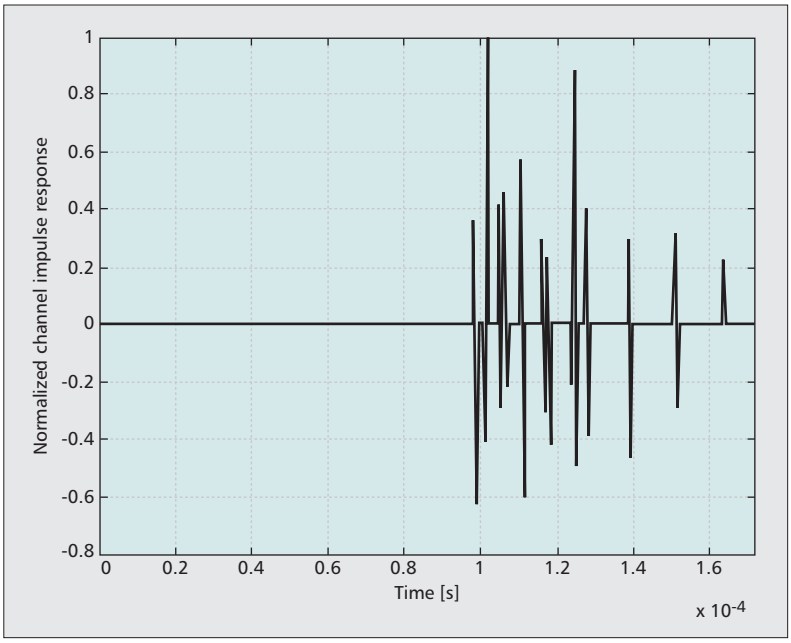

Figure 3. *Example of ultrasonic channel impulse response in the arm.*

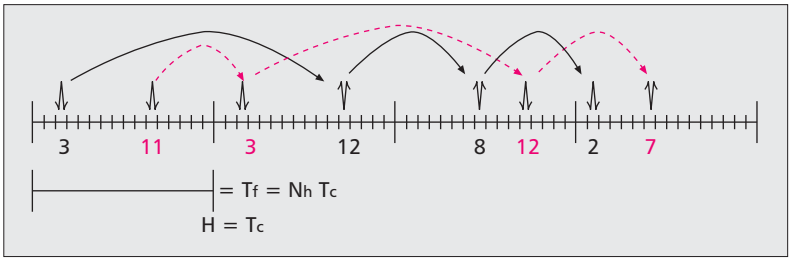

Figure 4. *Example of two ongoing transmissions using time hopping sequences* $TH_1 = \{3, 12, 8, 2\}$ *and* $TH_2 = \{11,3, 12, 7\}$ *and PPM-BPSK spreading codes* $SC_1 = \{1, –1, –1, 1\}$ *and* $SC_2 = \{1, 1, 1, –1\}$.

Thus, the goal of the UsWB is to optimally, distributively, and asynchronously regulate these tradeoffs to achieve different performance objectives through rate-maximizing and/or energy-minimizing adaptation strategies.

Network Layer Design

Data carrying health-related information needs to be delivered reliably and timely; at the same time, body tissues are sensitive to heating and it is imperative to reduce thermal stress and overheating. The main aspects that should be taken into account in designing network layer protocols are:

- *Attenuation*: To cope with the high attenuation introduced by the body channel, data forwarding should be redundant to guarantee that data reach the gateway node. However, redundancy may cause excessive overheating and energy consumption while batteries in biomedical sensors cannot be replaced.
- *Reliability*: To increase the probability that data are received at an actuator/gateway node, redundancy can be employed. This however can cause again increase in energy consumption and heating.
- *Single vs. Multihop Communications*: As discussed in the previous sections, due to the

Medical training phantoms (i.e., devices that substitute human subjects with acoustically accurate representations of anatomy) are interposed between transmitters and receivers to emulate propagation through tissues with high fidelity.

high attenuation in ultrasonic scenarios, data cannot travel over a single hop from the biomedical sensors. Instead, multihop transmissions are needed. This implies that no centralized routing solutions can be employed but distributed approaches are needed.

- *Area-aware Routing*: To avoid overheating of specific regions, routing should be designed to equalize the overhead traffic on network paths. Taking inspiration from the literature on QoS routing in ad hoc networks, the QoS parameter of interest could be identified with the energy consumption and paths where biomedical devices have higher residual energy or the traversed areas are less sensitive to increase in temperature can be selected for forwarding. Again, maintenance of these paths requires that nodes exchange some signaling information which, on the other hand, could increase the power consumption.

System Evaluation

In this section, we discuss two complementary approaches to evaluate intra-body ultrasonic communications: multi-scale simulation and experimental evaluation on a software-defined radio platform.

Multi-scale Simulator of Ultrasonic Networks

In our work [14], we have evaluated ultrasonic networked operations at three different levels, i.e.,

- At the wave level by modeling ultrasonic propagation through reflectors and scatterers
- At the bit level by simulating in detail the proposed ultrasonic transmission schemes
- At the packet level by simulating networked operations and distributed control and adaptation to evaluate metrics such as network throughput and packet drop rate

Each simulation level needs to leverage empirical models derived through simulation at the lower level.

The wave-level component of the simulator can be implemented in Matlab and can be based on existing acoustic propagation modeling tools (e.g., k-wave). This simulation step generates channel models to be fed to the bit-level component of the simulator. The latter can leverage well-developed communication libraries and toolboxes available in Matlab to simulate the performance of transmission schemes over ultrasonic channels. The output of this portion of the simulator can be empirical models (up to the second-order statistics) of the BER performance of a transmission scheme under varying channel and interference conditions, to be employed in packet-level network simulation studies. Finally, the packet-level portion of the simulator can be developed implementing a classical discrete-time and event-driven network simulator based on a high-level object-oriented programming language. Our current working version of the packet level simulator, is based on the Java programming language because of its modularity and simplicity. Major well-know simulators (i.e., ns-2 or ns-3) can be alternatively used.

Software-defined Testbed for Ultrasonic Networking

Experimental evaluation can also be used to validate simulation results. Our architecture for an ultrasonic testbed is based on USRP2/N210 software-defined radios interfaced with low-frequency (LFRX and LFTX) daughterboards, connected to ultrasonic transducers. The LFRX and LFTX USRP daughterboards operate from DC to 50 MHz, which includes the ranges of commercial transducers of interest to us. A switch board is required to use a single transducer as transmitter and receiver in time division duplex. Signal processing algorithms and protocols can be implemented using a framework that combines GNU radio and the freely available and open source FPGA code for USRP. Medical training phantoms (i.e., devices that substitute human subjects with acoustically accurate representations of anatomy) are interposed between transmitters and receivers to emulate propagation through tissues with high fidelity.

Conclusions

We discussed ultrasonic networking principles for intra-body implantable devices that go beyond traditional "along-the-body-surface" BAN communications. Accordingly, we discussed fundamental aspects of ultrasonic propagation in tissues, and explored system tradeoffs, including the choice of a transmission frequency, transmission power, and transducer size. We also discussed limitations of RF electromagnetic communications in tissues, thus motivating the choice of ultrasonic waves. Furthermore, we outlined future research challenges for ultrasonic networking of intra-body devices at the physical, medium access, and network layers of the protocol stack. The research discussed in this work may thus pave the way for a new communication paradigm that has the potential to enable new applications for medical treatment of major diseases at the intersection between biomedical science, networking and control.

Acknowledgments

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Biographies

G. Enrico Santagati (santagat@buffalo.edu) is a Ph.D. student in the Department of Electrical Engineering at the State University of New York (SUNY) at Buffalo. He is currently working in the Wireless Networks and Embedded Systems Laboratory under the guidance of Professor Tommaso Melodia. He received his B.S. and M.S. in Telecommunication Engineering from the University of Catania (Italy) in 2010 and 2012, respectively. His current research interests are in ultrasonic intra-body area networks and software defined radios.

Tommaso Melodia (tmelodia@buffalo.edu) is an Associate Professor with the Department of Electrical Engineering at the State University of New York (SUNY) at Buffalo, where he directs the Wireless Networks and Embedded Systems Laboratory. He received his Ph.D. in Electrical and Computer Engineering from the Georgia Institute of Technology in 2007. He had previously received his "Laurea" and Doctorate degrees in Telecommunications Engineering from the University of Rome "La Sapienza", Rome, Italy, in 2001 and 2005, respectively. He serves in the Editorial Boards of IEEE Transactions on Wireless Communications, IEEE Transactions on Mobile Computing, and Computer Networks (Elsevier). He received a National Science Foundation CAREER award, and coauthored a paper that was recognized as the Fast Breaking Paper in the field of Computer Science by Thomson ISI Essential Science Indicators and a paper that received an Elsevier Top Cited Paper Award. His current research interests are in modeling, optimization, and experimental evaluation of wireless networks, with applications to cognitive and cooperative networking, ultrasonic intra-body area networks, multimedia sensor networks, and underwater networks.

Laura Galluccio (laura.galluccio@dieei.unict.it) received her laurea degree in Electrical Engineering from University of Catania, Italy, in 2001. In March 2005 she got her Ph.D. in Electrical, Computer and Telecommunications Engineering at the same university. Since 2002 she is also at the Italian National Consortium of Telecommunications (CNIT), where she worked as a Research Fellow within the VICOM (Virtual Immersive Communications) and the SATNEX Projects. Since November 2010 she is Assistant Professor at University of Catania. From May to July 2005 she has been a Visiting Scholar at the COMET Group, Columbia University, NY under the guidance of Andrew T. Campbell. She is the Guest Editor for a Special Issue of the Wireless Communications Magazine on Wireless communications at the nanoscale and serves in the editorial boards of Wiley Wireless Communications and Mobile Computing and Elsevier Ad Hoc Networks, as well as in the technical committee of several major conferences sponsored by ACM and IEEE. Her research interests include ad hoc and sensor networks, protocols and algorithms for wireless networks, network performance analysis and nanonetworks.

Sergio Palazzo (sergio.palazzo@dieei.unict.it) received his degree in electrical engineering from the University of Catania, Italy, in 1977. Since 1987, he has been with the University of Catania, where is a Professor of Telecommunications Networks. In 1994, he spent the summer at the International Computer Science Institute (ICSI), Berkeley, as a Senior Visitor. He is a recipient of the 2003 Visiting Erskine Fellowship by the University of Canterbury, Christchurch, New Zealand. Since 1992, he has been serving on the Technical Program Committee of INFOCOM, the IEEE Conference on Computer Communications. He has been the General Chair of some ACM conferences, including MobiHoc 2006 and MobiOpp 2010, and currently is a member of the MobiHoc Steering Committee. He has also been the TPC Co-Chair of the IFIP Networking 2011 conference. He currently serves the Editorial Board of the journal Ad Hoc Networks. In the recent past, he also was an Editor of IEEE Wireless Communications Magazine (formerly IEEE Personal Communications Magazine), IEEE/ACM Transactions on Networking, IEEE Transactions on Mobile Computing, Computer Networks, and Wireless Communications and Mobile Computing. His current research interests include mobile systems, wireless and satellite IP networks, and protocols for the next generation of the Internet.

The research discussed in this work may thus pave the way for a new communication paradigm that has the potential to enable new applications for medical treatment of major diseases at the intersection between biomedical science, networking and control.