Challenges and Implications of Using Ultrasonic Communications in Intra-body Area Networks

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***Abstract*— Body area networks (BANs) promise to enable revolutionary biomedical applications by wirelessly interconnecting devices implanted or worn by humans. However, BAN wireless communications based on radio-frequency (RF) electromagnetic waves suffer from poor propagation of signals in body tissues, which leads to high levels of attenuation. In addition, in-body transmissions are constrained to be low-power to prevent overheating of tissues and consequent death of cells.**

To address the limitations of RF propagation in the human body, we propose a paradigm shift by exploring the use of ultrasonic waves as the physical medium to wirelessly interconnect in-body implanted devices. Acoustic waves are the transmission technology of choice for underwater communications, since they are known to propagate better than their RF counterpart in media composed mainly of water. Similarly, we envision that ultrasound (e.g., acoustic waves at non-audible frequencies) will provide support for communications in the human body, which is composed for 65% of water. In this paper, we first assess the feasibility of using ultrasonic communications in intra-body BANs, i.e., in-body networks where the devices are biomedical sensors that communicate with an actuator/gateway device located inside the body. We discuss the fundamentals of ultrasonic propagation in tissues, and explore important tradeoffs, including the choice of a transmission frequency, transmission power, bandwidth, 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.

I. Introduction

In[1] recent years, so-called *body area networks* (BAN) have attracted considerable interest thanks to the availability of miniaturized sophisticated bio-medical devices that can be implanted or worn by humans [4]. Consequently, numerous research projects proposed system architectures and/or service platforms for BANs [2], [3], [16]. In *intra-body area network* (IBAN) information is transmitted between in-body sensors and one or more actuators/gateway that can be connected to an external network, while *extra-body area network* are concerned with data delivered between the actuator/gateway and the external network.

Intra-body communications can be classified as *in-body* and *along-the-body* communications. In in-body communications, signals propagate through body tissues and losses are mainly caused by the absorption of power, which may result in heating of tissues. In along-the-body communications, devices are deployed on the body surface and signals may propagate along its surface.

Body area networks (BAN) promise to enable revolutionary biomedical applications by wirelessly interconnecting sensing and actuating devices implanted or worn by humans. As an example, sensors deployed inside the body of diabetic patients could measure the level of glucose and, consequently, an actuator implanted under the skin would adaptively administer the correct dose of insulin. Unfortunately, however, BAN wireless communications based on radio frequency (RF) electromagnetic waves suffer from poor propagation of signals in body tissues, which leads to high levels of attenuation. In addition, in-body transmissions are constrained to be low-power to prevent over-heating of tissues and consequent death of cells, and to guarantee continued operation of battery-powered implanted devices over long periods of time. Health concerns impose, in particular, that the amount of power should remain within some predefined safety levels [17].

To address the limitations of RF propagation in the human body, we propose a paradigm shift by exploring the use of ultrasonic waves to wirelessly interconnect in-body devices. Acoustic waves are the transmission technology of choice for underwater communications, since they are known to propagate better than their RF counterpart in media composed mainly of water. Indeed, acoustic waves, typically at frequencies between 0 and 100 kHz, have been used successfully for underwater communications since World War Two [1]. Low frequency waves incur relatively modest path loss, which enables long-range transmissions (over distances that can span kilometers) at relatively low transmission powers [1].

Similarly, we believe that ultrasound (i.e., acoustic waves at non-audible frequencies, above 20 kHz) has a strong potential to provide support for communications in the human body. In average, 65% of the human body consists of water, and blood is a fluid used to support organs-tissues communications, transport drugs, minerals, nutrients.

Accordingly, in this paper, we first assess the feasibility of using ultrasonic communications in intra-body BANs. We discuss the fundamentals of ultrasonic propagation in tissues, and explore important tradeoffs, including the choice of a transmission frequency, transmission power, bandwidth, and

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transducer size. In addition, we propose a system architecture and discuss future research challenges for ultrasonic networking of in-body devices at the physical, medium access and network layers of the protocol stack.

The rest of the paper is organized as follows. In Section II we start off by describing the proposed system architecture. In Section III we perform a preliminary study on the feasibility of using ultrasonic communications with a focus on frequency selection, transceiver design and noise/attenuation characterization. In Section IV we discuss design issues on medium access in ultrasonic scenarios. In Section V we discuss design issues at the network layer and relate them to the application requirements. In Section VI we discuss health concerns raised by the propagation of ultrasound in the human body. Finally, in Section VII we conclude the paper.

II. System Architecture

We consider an IBAN architecture that consists of a set of biomedical wireless sensors deployed inside the human body. As an example, some of these sensors could be implanted into the wall of a patient's hearth through a catheter [5] or be inserted into the body by swallowing them up as pills [5].

Biomedical nodes communicate through acoustic links with a gateway/actuator node that can be either implanted in the body or outside, for example in a watch or in a PDA carried by the patient.

The gateway node can be connected to a remote medical center where the medical personnel can check the patient status and/or the patient data can be stored in a medical database for future use, or be remotely accessed by the patient's physician.

Also, without any interaction with the external network, the actuator/gateway node could inject into the human body specific drugs. As an example, in diabetic patients, sensors could monitor the level of glucose in the blood and send notifications back to the actuator node upon realizing that the level of glucose is too high. The actuator would then inject insulin, stored in a built-in reservoir and pump, into the blood.

A scheme of the system architecture is illustrated in Fig. 1. Typical requirements of such a system are:

- *High reliability.* Data should be delivered to the actuator/gateway node reliably since it is health-related and might trigger, if needed, appropriate actions at the medical center or at the actuator/gateway node itself.
- *Rapid delivery.* Data related to the patient status should be delivered rapidly so that appropriate actions can be undertaken promptly.
- *Reduced Emission Power.* Low-power transmissions are clearly desirable. Also, transmission schemes that avoid overheating specific body areas and instead distribute the load over larger body regions should be preferred.
- *Long Battery Lifetime.* Biomedical sensors should be implanted in the body for years, possibly for decades. Accordingly, mechanisms specifically designed to increase the network lifetime need to be incorporated at all layers of the protocol stack.

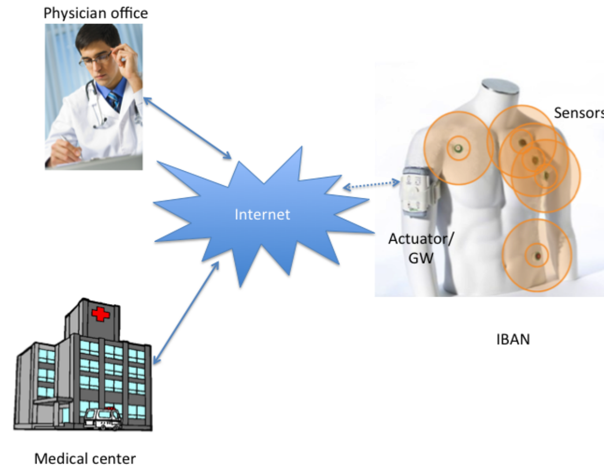

Fig. 1. System architecture

- *Small Form Factor.* Devices should be implanted under the skin or introduced in the body and remain there for decades; accordingly, these devices should be small in size and not cause any noise to the patient carrying them.

III. Ultrasonic Communications in Human Tissues

In this section, we briefly introduce the basics of ultrasound waves (III-A). Then, in Sections III-B, III-C and III-D we discuss the propagation characteristics of ultrasound waves and transducers. In Section III-E we discuss the effects of noise and in Section III-F we provide an analysis of the usable bandwidth.

A. Fundamentals of Ultrasound Waves

Ultrasounds are mechanical pressure waves with a frequency above the upper limit for human hearing, i.e., $20\,\mathrm{kHz}$. Ultrasounds consist of mechanical vibrations of particles in a material. Even if each particle oscillates around its rest position, the vibration energy propagates as a wave traveling from particle to particle through the material. Acoustic waves in general, and ultrasonic waves in particular, can be characterized through their physical parameters as follows:

- The *frequency*, f, represents the number of complete oscillations per second for each particle. As stated above, ultrasound waves have a lower bound of $20\,\mathrm{kHz}$.
- The *pressure*, P, is a measure of the compressions and rarefactions of the molecules in the medium through which sound waves propagate, and is typically measured in Pascal $[\mathrm{N/m^2}]$.
- The *amplitude*, A, represents the displacement of particles from their rest position.
- The *propagation speed* c is the rate at which the vibratory energy is transmitted in the direction of propagation and increase with the rigidity of the material. In tissues, we consider an average sound speed of $1540\,\mathrm{m/s}$.
- The *intensity*, I, is the average energy carried over time by a wave per unit area normal to the direction of

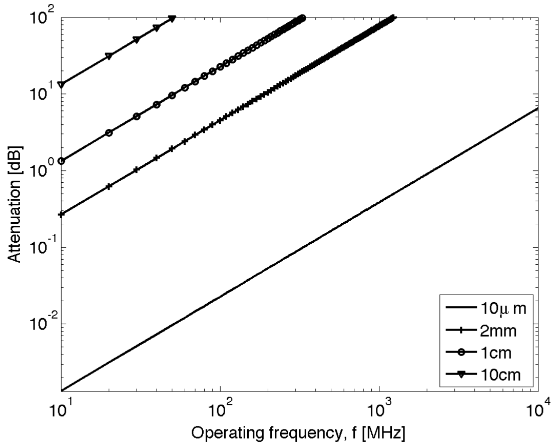

Fig. 2. Attenuation in the blood as a function of the operating frequency for different values of distance.

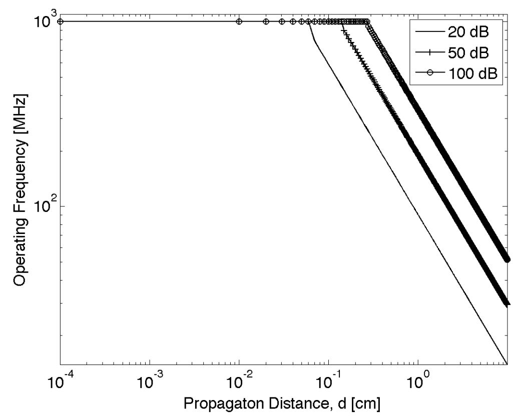

Fig. 3. Maximum Carrier Frequency in the blood.

propagation, and is usually measured in $\mathrm{mW/cm^2}$. It can be expressed as a quadratic function of pressure as

$$I = \frac{(P_{RMS})^2}{\rho c} \quad [W/m^2], \tag{1}$$

where P_{RMS} is the sound pressure root mean square (rms), ρ and c represent the density of the medium and the speed of sound in the medium, respectively.

B. Attenuation

When ultrasound waves propagate through an absorbing medium, the initial pressure, P_0 reduces at $P(d)$ at a distance d according to

$$P(d) = P_0 e^{-\alpha d}, \tag{2}$$

where α (in $[\mathrm{np} \cdot \mathrm{cm}^{-1}]$) is the amplitude attenuation coefficient that captures all the effects that cause a dissipation of energy from the ultrasound beam. This parameter α is a function of the carrier frequency as [6]

$$\alpha = af^b \tag{3}$$

where f represents the carrier frequency and a (in $[\mathrm{np\ m^{-1}\ MHz^{-b}}]$) and b are attenuation parameters characterizing the tissue. Typical experimental values of these parameters in different tissues are reported in Table I [7].

Based on this, we can compute and plot attenuation as

TABLE I
TISSUES PARAMETERS

Tissue	α	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

a function of propagation frequency for different mediums. Results for blood are reported in Fig. 2.

Since most sensing applications require highly directional transducers, one needs to operate at high frequencies to keep the transducer size small. Conversely, as shown in Fig. 2, higher transmission frequencies lead to higher attenuation. Therefore, for effective ultrasonic communications *we need to operate at frequencies compatible with small devices, but at the same time limit the maximum attenuation*. In Fig. 3, we depict the maximum carrier frequency with respect to the distance, for different maximum attenuation values, assuming blood as the propagation medium, and limiting the maximum operating frequency to 1 GHz. In Table II, our results for a $100\,\mathrm{dB}$ attenuation are summarized. Since the attenuation increases as a function of frequency and distance, for a fixed value of attenuation we get an inverse relationship between frequency and distance.

For communication distances that range from some $\mu\mathrm{m}$ up to a few mm (what we refer to as short range communications) frequencies higher than 1 GHz are allowed. When distances are higher than $1\ \mathrm{mm}$ but still lower than some cm, i.e., for medium range communications, transmission frequencies should be decreased to approximately $100\,\mathrm{MHz}$. For distances higher than a few cm, i.e., for long range communications, the transmission frequency should not exceed $10\,\mathrm{MHz}$.

TABLE II
FREQUENCY LIMITS FOR A=100 dB

Communication range	Distance	Frequency Limit
Short Range	μm - mm	$> 1GHz$
Medium Range	mm - cm	$\simeq 100MHz$
Long Range	>cm	$\simeq 10MHz$

C. Ultrasonic Transducers

To accurately model the ultrasonic communication channel in tissues we need to understand how propagation parameters are affected by the characteristics of the transmitting and receiving devices. An ultrasonic transducer is a device capable of generating and receiving ultrasonic waves. It is essentially made up of an active element, a backing, and a wear plate. The *active element* is usually a piezoelectric material that

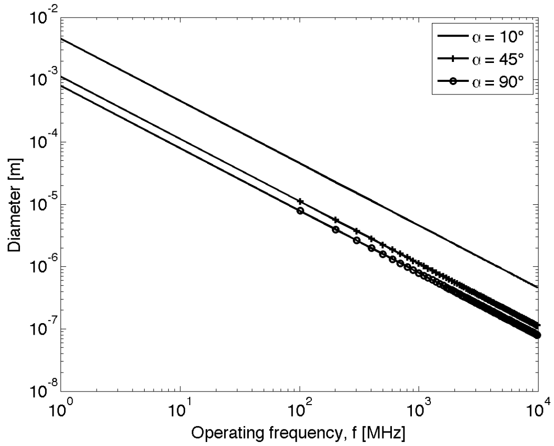

Fig. 4. Minimum transducer diameter as a function of the operating frequency.

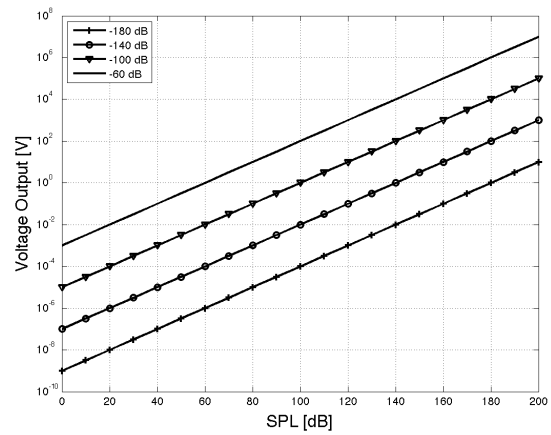

Fig. 5. Voltage Output vs. SPL Input, assuming different sensitivity values.

converts electrical energy to ultrasonic energy and viceversa. The *backing* is a very dense material used to absorb the energy radiated from the back face of the piezoelectric element. The *wear plate* allows to protect the piezoelectric transducer element from the environment and is generally selected to protect against wear and corrosion.

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 shape of the vibrating surface. Furthermore, the acoustic radiation pattern is reciprocal, i.e., its shape is the same whether 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 in very narrow beams. For a transducer with a circular radiating surface, the narrowness of the beam pattern is a function of the ratio of the diameter of the radiating surface and the wavelength at the operating frequency, D/λ.

This relationship can be explained by considering the beam spread formulas defined as [8]

$$sin\left(\frac{\alpha}{2}\right) = \frac{0.514c}{fD}, \tag{4}$$

where $\frac{\alpha}{2}$ is half the angle spread between the -6 dB points.

The higher the frequency or the larger the transducer's diameter, the smaller the beam spread.

In Fig. 4, we show the minimum diameter of the transducer for different values of α as a function of the operating frequency. When a transducer is working as a transmitter, it is usually characterized in terms of pressure level. In particular, the sound pressure level $[\mathrm{in\,dB\,referred\,to}\,\mu\mathrm{Pa}]^2$ is defined as:

$$SPL = 20log\frac{P_{RMS}}{P_{REF}}, \tag{5}$$

where P_{RMS} [μPa] is the sound pressure rms and P_{REF} is the reference pressure, equal to $1\mu\mathrm{Pa}$ in underwater environments.

[2]In the following of this paper "referred to" will be denoted as re.

This pressure value can be converted to intensity through eq. (1), assuming appropriate values for density and speed of sound in tissues [7].

The main parameter characterizing a transducer used as a receiver of acoustic signals is the *sensitivity* (Sen) $[\mathrm{dB\,re\,V}/\mu\mathrm{Pa}]$, which can be defined as the voltage output found at the receiver when a specific value of SPL is given, i.e.,

$$Sen = 20log(\frac{V_{SPL}}{1\ V/\mu Pa}), \tag{6}$$

where V_{SPL} is the output voltage given by a specific SPL value. In Fig. 5, the I/O relationship for transducers with different sensitivity values are reported. These plots show the output in Volts, given an SPL value in dB re μPa, for different sensistivities expressed in $\mathrm{dB\,re\,Volt}/\mu\mathrm{Pa}$.

D. 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 difference between the different tissues. This mirror-like reflection can be expressed using the Snell law for acoustics, [9].

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. To describe scattered ultrasonic waves we need to model the statistics of the signal in a very general environment. Therefore, the model must have parameters that take in account tissue characteristics and are simple to compute. A model that satisfies these requirements is based on the Nakagami distribution [10].

If we consider a transmitted signal $s(t)$, the received signal due to scattering effect, assuming a noiseless channel and non-selective slow fading, can be expressed as

$$r(t) = \Re\{\rho e^{j\phi} s(t)\}. \tag{7}$$

To characterize the statistical behavior of the received signal, we assume the phase shift ϕ to be uniformly distributed in $[0, 2\pi]$ and the magnitude ρ to be Nakagami-distributed. Therefore, the probability density functions of these random variable are expressed as

$$f(\phi) = \frac{1}{2\pi} rect_{2\pi}(\phi) \tag{8}$$

$$f(\rho) = \frac{2m^m \rho^{2m-1}}{\Gamma(m)\Omega^m} e^{(-\frac{m}{\Omega}\rho^2)} U(\rho) \tag{9}$$

where m is the Nakagami parameter, Ω is a scaling parameter, $U(\cdot)$ is the unit-step function, $\Gamma(\cdot)$ is the gamma function and $rect_{2\pi}(\cdot)$ is the rectangular function of duration 2π.
Varying the value of the Nakagami parameter, we obtain different distributions such as the Gaussian distribution ($m = 0.5$), the Rayleigh distribution ($m = 1$) and the Rician distribution ($m > 1$). Therefore, the Nakagami amplitude distribution can describe different scenarios simply varying the value of m.
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 IBANs deeply affected by multipath fading.

E. Noise Sources

No previous study, to the best of our knowledge, has investigated in detail noise sources in tissue propagation. In underwater communications, the ambient noise level is given by the intensity of the ambient noise, expressed in dB re μPa, and is usually expressed as a sum of different components. Since we are focusing on ultrasonic frequencies, we focus on thermal noise produced by the agitation of molecules in tissue. As in [11], the power spectral density (PSD) of the thermal noise obtained for an ideal medium using classical statistical mechanics can be adopted with good approximation for a real medium such as body tissues. Assuming different densities and sound speed for the medium [7], we can derive an expression for the noise PSD in dB re μPa per Hz as

$$N(f) = -15 + 20log(f), \tag{10}$$

with f expressed in kHz. Based on (10), we can express the system SNR as

$$SNR(d, f) = \frac{P/A(d, f)}{N(f)\Delta f}, \tag{11}$$

where P is the transmitted power, $N(f)$ is the PSD of the noise, assumed to be equal to the PSD of the thermal noise, and $A(d, f)$ is the attenuation experienced by the transmitted pressure signal. From eq. (2), $A(d, f)$ can be expressed as

$$A(d, f) = e^{\alpha d}. \tag{12}$$

To estimate the impact of frequency and distance on the SNR, we can focus on the expression of $(A(d, f)N(f))^{-1}$ only. This quantity is shown in Fig. 6 as a function of the transmission frequency, assuming blood as the propagation medium and considering different propagation distances. Since noise and attenuation are monotonically increasing functions of the operating frequency, $(A(d, f)N(f))^{-1}$ monotonically decreases with increasing frequency. Furthermore, since attenuation is also a function of the propagation distance, our results show that $(A(d, f)N(f))^{-1}$ decreases faster as the link distance increases.

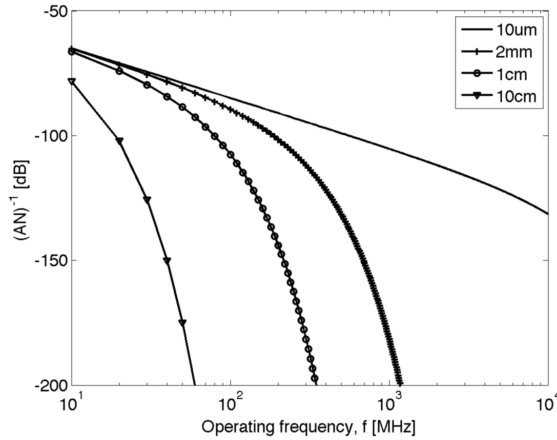

Fig. 6. $(A(d, f)N(f))^{-1}$ when the propagation medium is blood.

F. Bandwidth Evaluation

Based on this discussion, we can make some preliminary considerations on the bandwidth available for ultrasound communications in tissues. As in [12], we maximize the channel capacity using an optimal energy allocation for a fixed transmission power.
When we maximize the capacity with respect to the power spectral density of the signal denoted as $S_d(f)$, assuming a finite transmission power P, we obtain an optimal power distribution, and the solution satisfies the water-filling principle:

$$S_d(f) + A(d, f)N(d, f) = K_d, \tag{13}$$

where K_d is a constant depending on the value of P, which is fixed according to the desired SNR value. This result can be interpreted as follows. To achieve maximum capacity through an optimal energy allocation, the total PSD on the channel's band has to be flat and equal to K_d.
The maximum capacity obtained can be expressed as in [12]

$$C(d) = \int_{B(d)} log_2 \left[1 + \frac{K_d}{A(d, f)N(f)}\right] df, \tag{14}$$

where $B(d)$ is the bandwidth that varies depending on the distance d. Assuming now a desired SNR equal to 20 dB, we can compute the desired bandwidth and the maximum capacity using the numerical procedure reported in [12]. In Fig. 7 bandwidth and capacity are shown as a function of the propagation distance. In this plot, we assume $1\,\mathrm{GHz}$, $100\,\mathrm{MHz}$ and $10\,\mathrm{MHz}$ as operating frequencies for the short, medium

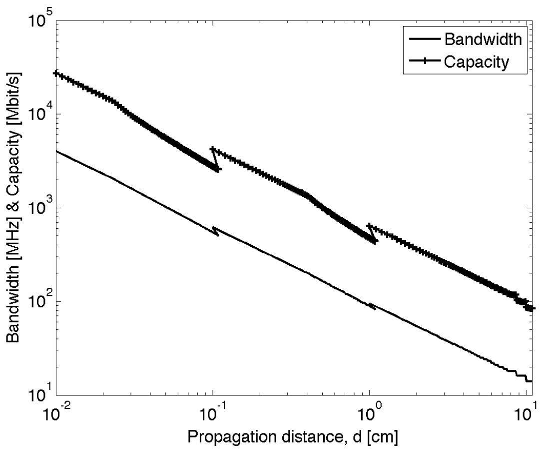

Fig. 7. Bandwidth [Mbit/s] and Capacity [MHz] vs. propagation distance.

and long range communications, respectively, according to our results in Section III-B.

Using the appropriate operating frequencies as identified in Table II, the bandwidth and capacity decrease almost linearly in log-scale as distance increases. This shows that high data rates can be reached in case of long distance communications, which provides a strong motivation for seeking to achieve robust and reliable communications through high-bandwidth signaling techniques such as ultra wide band (UWB) and/or code division multiple access (CDMA).

IV. Medium Access Control Layer Design Issues

As discussed in the previous section, 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 in Section III-D, 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.

An appropriate medium access control scheme has to allow users to share the communication medium fairly while limiting interference, guaranteeing maximum throughput and minimizing the protocol overhead. Unfortunately, the characteristics of the ultrasonic channel, as illustrated in Section III, make the design of an efficient MAC protocol for in-body communications a challenging task. In particular the main aspects that should be taken into account in designing medium access solutions are:

- *Frequency division*. As discussed in [1], multipath fading strongly affects any narrowband signals. Therefore, an 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 long propagation delays in tissues, since carrier sensing becomes inefficient when long delays are considered and even introducing some handshaking mechanisms such as RTS/CTS might have negative effects on the network throughput.
- *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 resistant to multipath fading.

As observed in Section III-F, considering different ranges of communications, the available bandwidth obtained on the channel is virtually always greater than the center frequency used. Therefore, the ultra wide band (UWB) technique could offer a promising solution.

UWB, in its time-hopping impulse radio variant (TH-IR-UWB), consists of transmitting information in very short pulses, which result in a very large bandwidth [13].

Numerous are the potential benefits of using the TH-UWB technique. First, since the duration of the UWB pulse is very short (in the order of hundreds of picoseconds in RF communications), reflecting and scattering do not overlap in time. This means that UWB is multipath-fading resistant, while, as discussed in Section III-D, multipath is one of the major concerns in ultrasonic channels. Finally, as will be discussed in Section VI, pulsed transmissions with low duty cycles reduce detrimental thermal and mechanical effects, thus drastically reducing the probability of negative bioeffects caused by heating or cavitation.

To counteract the effects of multipath fading and long propagation delays, integrated physical and medium access solutions should be considered. A very promising approach for medium access, as mentioned above, is direct sequence spread spectrum (DSSS). DSSS consists of using spreading codes to protect the signal of interest from interference.

The spreading code is a binary sequence with a rate higher than the information signal rate. The DSSS technique consists in multiplying the information signal by the spreading code before transmission, and consequently spreading the signal frequency spectrum into a much wider band. At the receiver side, using the same spreading code of the transmitter, the signal spectrum can be de-spread, and the information signal reconstructed, therefore reducing the effects of interference.

By distinguishing multipath arrivals, which are non-overlapping in UWB or uncorrelated in DSSS, it is possible to combine signal replicas caused by multipath with a time diversity receiver. Moreover, the two solutions presented above - combined or not - naturally enable multiple access to the channel by concurrently transmitting devices, thus allowing different devices to communicate simultaneously without interfering with each other.

Since the pulse duration in UWB is extremely short, multiple access schemes based on time-hopping have been proposed. The time-hopping technique allows users to transmit with a pseudo-random time-schedule, and consequently allows them to share the available frequency band with a reasonable level of robustness to collisions.

In DSSS communications, code division multiple access schemes are readily obtained using orthogonal or pseudo-orthogonal sets of spreading codes. By associating a spreading code to each communication and de-spreading the received signals with the appropriate codes, users can concurrently transmit on the same portion of the spectrum.

A combined use of DSSS and UWB techniques, referred to as DS-UWB, may also offer a promising solution for medium access in ultrasonic networks. Each user, through a different spreading code, transmits information bits as a sequence of short pulses that depend on the spreading code itself. Since codes are uncorrelated, the transmitted information can be reconstructed by de-spreading the received signals.

V. Network Layer Design Issues

Networking issues in IBAN are tightly associated to the nature of the information being transported. In fact, data are related to health status and therefore should be delivered reliably and timely; at the same time, body tissues are sensitive to heating and it is imperative to reduce thermal stress and over-heating. The main aspects that should be taken into account in designing networking protocols are:

- *Attenuation.* To cope with the high attenuation introduced by the body channel, data forwarding should be redundant so that, eventually, data reach the gateway node. However, redundancy may cause excessive overheating and energy consumption while batteries in biomedical sensors cannot be replaced and devices should be long lasting.
- *Reliability.* To increase the probability that data are received at the 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 high attenuation in ultrasonic scenarios, data cannot travel over a single hop from the biomedical sensors. Instead, multi hop 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. As an example, in [14] a thermal aware routing algorithm (TARA) is discussed. In this work, data is routed away from areas characterized by high temperature. Evolutions of this methodology include schemes where a map of nodes' temperature is created so that data forwarding can be obtained by selecting minimum-temperature routes [15]. However, a disadvantage of such schemes is that overhead associated to this exchange of information could stress the IBAN significantly. Taking inspiration from the literature on QoS routing in ad hoc networks [18], 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.

VI. Health Concerns

Whenever any form of energy is introduced into the body, it is important to understand what risks might result from applying energy to internal tissues. In this section we briefly summarize the main effects of the propagation of ultrasonic waves in human tissues [19].

The most obvious effect is *heating*. Since a significant portion of the energy is absorbed and converted into heat during ultrasound propagation, this could potentially lead to a temperature increase. As the wave intensity is increased, temperature rises and if the temperature becomes higher than $38.5°C$, adverse biological effects may occur. However, no lethal effects have been observed for temperatures lower than $41°C$. Since heating is strictly caused by the wave intensity, a pulsed transmission with a low duty cycle can potentially reduce this effect of a factor proportional to the duration of the duty cycle.

Another important effect caused by ultrasound wave propagation is the, so-called, *cavitation* which denotes the behavior of gas bubbles within an acoustic field. Pressure variations of the ultrasound wave cause bubbles present in the propagation medium to contract and expand. For large pressure variations, the bubble size drastically increases, reaching an expansion peak when pressure is minimum and then collapsing when pressure reaches its peak. During this process the internal pressure and temperature in the bubble can reach high values causing serious biological effects and damaging the objects located in closest proximity. It can be shown that cavitation is a frequency-dependent phenomenon. Since higher frequencies lead to shorter pressure oscillations, the time for bubble expansion is restricted and the cavitation effect tends to disappear. Pulsed transmissions may also reduce the cavitation effects. In fact, during the off period the bubbles assume again their initial sizes without imploding. However, these destructive effects are not seen for low pulse intensity values.

Unfortunately, data collected on bioeffects of ultrasounds are frequently inconsistent and controversial. However, based on the medical ultrasound experiences of the last decades, no dangerous bioeffects have been observed as long as the energy provided to the tissues is less than 50 J/cm^2, [20]. Therefore, ultrasound communications in tissues at low transmission pressure levels, and consequently low transmission power levels, are not expected to cause any lethal bioeffects. We thus believe that ultrasonic communications can represent a feasible alternative to classical electromagnetic RF communications.

VII. Conclusions

We assessed the feasibility of using ultrasonic communications in intra-body BANs. We discussed the fundamentals of ultrasonic propagation in tissues, and explored important system tradeoffs, including the choice of a transmission frequency, transmission power, bandwidth, and transducer size. We discussed a system architecture and outlined future research challenges for ultrasonic networking of in-body devices at the physical, medium access and network layers of the protocol stack. Significant work is needed at the physical, medium access and network layers to make ultrasonic communications feasible in realistic network scenarios.

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