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Characterization of Ultrasonic Wave Propagation for Intra-Body Communication

Francois RIVET [a] Samuel REDOIS [a] and Yann DEVAL [a]
[a] *Universite de Bordeaux, Bordeaux, France*

Abstract. Intra-body area network will enable healthcare applications. Sensors and actuators are supposed to be interconnected thanks to wireless communications. But radio frequency (RF) are limited when intra body communications are concerned. This paper investigates ultrasonic waves as an alternative wireless carrier of information. Indeed, many studies has shown that water and biological environments are most suited to propagating ultrasonic waves. Our goal is to characterize how ultrasonic waves propagate in the human body for intra-body communications. We present trade off in terms of frequency and dimensions of the transmitter based on theory, simulations and experimental setup demonstration.

Keywords. Intra-Body Communications (IBC), ultrasonic waves, near field, far field

1. Introduction

Connected health is in full expansion. Indeed, large companies like Samsung consider introducing in the human body sensors and actuators to analyse in real time vitals within 10 years. Biomedical systems of implanted or wearable miniaturized device wirelessly interconnected into an intrabody area network could enable revolutionary healthcare and clinical applications. The main obstacle to enabling networked implantable device is posed by the dielectric nature of the human body, which is composed primarily (65%) of water, a medium through which radio frequency (RF) electromagnetic waves do not easily propagate. The human body is more suited to propagate ultrasonic waves. Indeed, the water is not at all compliant for radio frequency propagation over 100MHz but is compliant for low frequency acoustic waves up to 100kHz [1].

This paper aims at investigating Intra-Body Communication (IBC) by determining the most relevant transmitter in terms of frequency for an in-vivo telecommunication. Human body covers various level of heterogeneous media of propagation such as blood, bones, skin, fat. Our paper presents a theoretical model of a non-linear ultrasonic wave propagation. First, section II introduces mathematical tools required to understand theoretically a non-linear ultrasonic wave propagation. Then, section III details parameters required to simulate an ultrasonic wave propagation in biological medium with muscle as case of study. These results are also interpreted to define optimized characteristics of the transmitter. Finally, in section IV, we study experimentally a low frequency ultrasonic propagation in an absorbent medium (air) in order to validate our theoretical study.

2. Wave propagation in a biological medium

The goal of this section is to model a non-linear ultrasonic wave propagation in an absorbent medium. Ultrasounds are mechanical and elastic waves. Therefore, the wave propagation causes a medium particles displacement. This displacement can be parallel, or not, to the direction of propagation. We can introduce the notion of waves polarisation. In case of acoustic wave, two types of polarisation can be distinguished:

- Longitudinal bias or compression wave: this bias corresponds to a particles displacement along the propagation axis.
- Transverse polarisation or shear wave: this bias corresponds to a particles displacement perpendicular to the propagation axis.

In biological medium, the shear waves are highly attenuated (2.10^4 to 30.10^4 $dB.cm^{-1}$). That is why we only consider longitudinal waves. In the one dimensional case, compression wave attenuation is expressed as follows:

$$I_d(x) = I_0 e^{-0.23\beta f x} \tag{1}$$

where β is the absorption coefficient, I_0 the initial intensity and f the ultrasonic wave frequency. Attenuation coefficients vary widely for different media. The attenuation coefficients of common biological materials are listed in Tab. 1 [2]:

Material	**β (dB.cm^{-1}.MHz^{-1})**
Air	1.64
Blood	0.2
Fat	0.48
Muscle	1.09
Water	0.002
Bone	6.9

Table 1. Absorption coefficient in common biological materials

In order to solve and understand waves propagation equations, several parameters have to be defined: velocity potential φ and particle velocity vector **u**. In the three-dimensional case, these quantities are related by the equation 2 [3]:

$$\nabla\varphi = u \tag{2}$$

Therefore, the acoustic pressure $\mathbf{p_e}$ can be expressed as follows:

$$p_e = -i\omega\rho\varphi \tag{3}$$

where ω is the pulse wave and ρ the volumetric mass density.

A specific acoustic impedance z is defined as pressure $\mathbf{p_e}$ to particle velocity vector:

$$z = \frac{p_e}{u} \tag{4}$$

According to the vector representation, a phase difference between the pressure and the velocity potential is highlighted in Fig. 1 [4]:

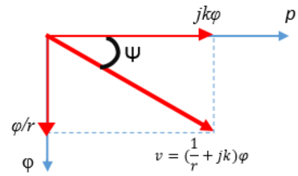

Figure 1. Vector representation of a spherical wave

where **k** is the propagation constant and **r** the distance from the source. According to the Fig. 1, the phase difference can be define by the relation:

$$\cos(\psi) = \sqrt{\frac{k.r}{1 + k^2.r^2}} \tag{5}$$

Consequently, based on the values of **k** and **r**, two fields can be identified: the far field and the near field (cf. Fig. 2).

2.1. *Near field*

This subsection details the near field modeling in case of ultrasonic wave propagation in muscle ($\beta = 1.09dB.cm^{-1}.MHz^{-1}$,$f = 10MHz$, $c = 1569m.s^{-1}$, $a = 1cm$, $\lambda = 157\mu m$, $A_0 = 1$, where c is the velocity in the medium, λ the wavelength, a the transmitter radius and A_0 the initial intensity). Simulations are processed thanks to Matlab. In the near field case, the product $k.r$ is much lower than 1. The equation 5 shows that pressure and velocity potential are in quadrature ($\cos(\psi) = 0$). Moreover, the acoustic impedance is purely complex. That is why the medium is very reactive and the wave amplitude fluctuates along the axis of the transmitter. This fluctuation can be define by the following equation 6:

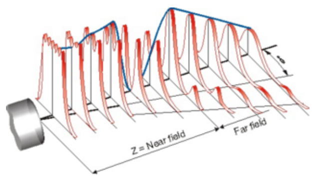

Figure 2. Near Field and Far Field

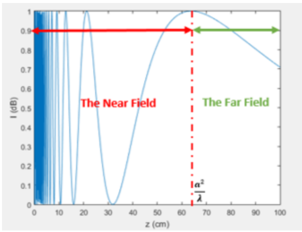

Figure 3. The axial intensity as a function of distance in the vicinity of a vibrating transmitter

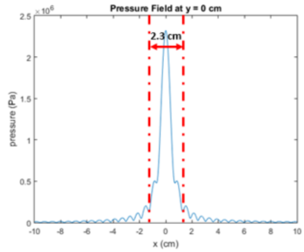

Figure 4. Modeling of side lobes of ultrasonic wave in the far field ($f = 10MHz$, $\beta = 1.09$, $a = 1cm$, $c = 1569m.s^{-1}$, $z = 50cm$)

$$I_d(z) = A_0 \sin\left(\frac{\pi a^2}{2z\lambda}\right)^2 \tag{6}$$

Simulation results of this function are provided in Fig. 3. Thanks to this figure, we can identify the near field and the far field. The location of the last maximum of this curve, $z = \frac{a^2}{\lambda}$, is usually taken as the limit between the near field and the far field. It will be the most suited trade-off in terms of distance to receive or transmit a signal because of its maximum of magnitude (here normalized at $1dB$).

2.2. *Far field*

In the far field, the product $k.r$ is much greater than 1. The pressure and velocity potential are in phase ($\cos(\psi) = 1$). Consequently, the acoustic impedance becomes real. The pressure and velocity potential decrease with a ratio of $\frac{1}{r}$. Fluctuations vanish and side lobes appear (cf Fig. 4). The propagation channel width is based on simulation results depicted in Fig. 4. Based on a 10MHz transmitter with $1cm$ radius, the propagation channel width is $2.3cm$.

3. Modeling of ultrasonic wave propagation in muscle

To model an ultrasonic wave propagation in the human body, we begin by considering a propagation in homogeneous and isotropic medium with the case of study of muscle. Like any medium, muscle can be characterized by different parameters : velocity, absorption coefficient and volumetric mass density. The transducer is defined by its radius, its form and its bend. All of these parameters are defined in Tab. 2.

Emitter		**Medium**	
shape	circular	**c**	$1569m.s^{-1}$
radius	$6.5mm$	α	$1.09dB.cm^{-1}.MHz^{-1}$
bend	0	ρ	$1.184kg.m^{-3}$

Table 2. Characteristics of the transmitter and muscular medium

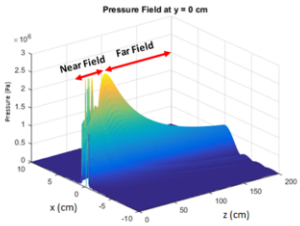

Figure 5. Modeling of ultrasonic wave propagation in muscle ($f = 10MHz$, $radius = 1cm$)

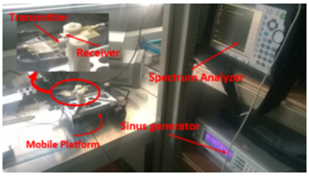

Figure 6. Setup measurement platform for Ultrasonic propagation analysis

Most modeling software suffer from errors and mismatching in the near field and especially at the transducer interface. The Michigan State University has developed a software, FOCUS, which performs calculation very quickly with the same accuracy than the other softwares [5]. After setting transducer and medium parameters, computer simulation results are exhibited in Fig. 5.

As detailed in section II, near field and far field are identified in simulation results. Consequently, FOCUS allows us to identify easily near field and far field in muscle. In far field, the pressure decreases exponentially according to the equation 1. The simulation results also depict the presence of side lobes. These side lobes widen according to an angle δ. This angle is defined by equation 7. Thus, the splay angle increases when frequency increases.

$$\delta = asin(\frac{2.44\lambda}{a}) \tag{7}$$

In order to limit thermal and mechanical constraints, selected transmitter frequency are between $5MHz$ and $10MHz$. At such frequencies, divergence angle is very low and simulations results show that the mechanical and thermal constraint indexes are in accordance with the relevant legislation. The choice of these frequencies is a trade-off between biological effects and propagation wave behaviour in biological medium.

4. Measurements of ultrasonic wave propagation

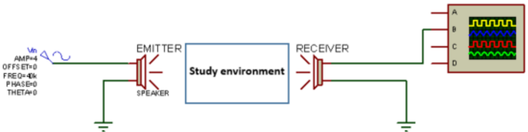

Figure 7. Electronic circuit used for ultrasonic wave propagation analysis

In order to exhibit near field and far field, an experimental protocol has been established. The transmitter is placed in a fixed position. The receiver is placed on a mobile platform, which can be adjusted in the X and Z directions. This platform can move by step of 1 μm. The demonstration setup is displayed in Fig. 6. First measurements are

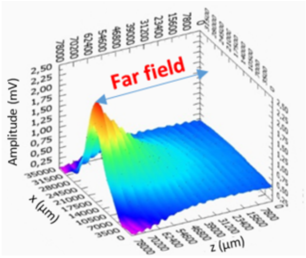

Figure 8. Experimental measurements of an ultrasonic wave propagation in air ($f = 40kHz$, $radius = 6.5mm$)

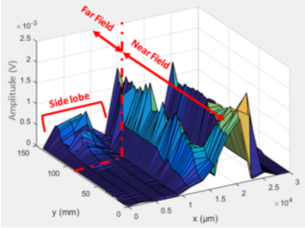

Figure 9. Experimental measurements of an ultrasonic wave propagation in muscle ($f = 40kHz$, $radius = 6.5mm$)

performed through air to validate the correct operability of the setup. The $40kHz$ transmitter used is a KPUS-40 T-16T and the receiver is a KPUS-40 T-16R. The frequency of $40kHz$ was chosen to highlight easily the exponentially attenuation of the pressure. Indeed, according to the equation 1, attenuation decreases when the frequency decreases.

Measurements are plotted in Fig. 8. They have been performed with a step of 500 μm. These measurements confirm results obtained during the theoretical study.

In case of ultrasonic wave propagation in muscle, the near field and the far field can be experimentally observed at $40kHz$ frequency. Measurements are plotted in Fig. 9. These measurements show the border between the two fields. The presence of side lobes in the far field is also exhibited.

5. Conclusion

Ultrasonic uses for IBC has a great potential. In this paper, the ultrasonic wave propagation in biological medium has been studied, theorized and simulated. By characterizing a biological and absorbent medium (muscle), a simulation has been done to determine the optimal transmitter range frequency, the optimal receiver position and the propagation channel width. This simulation was experimentaly demonstrated by a dedicated electronic setup using wave propagation in air. Perspectives are to perform advanced measurements within heterogeneous biological medium to approximate and validate the feasibility of ultrasonic wave propagation in the human body.

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