A Study of Magnetic Resonance and Ultrasound based Through-the-body Communications

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***Abstract*—This paper explores the application of Magnetic Resonance Coupling (MRC) and Ultrasonic Coupling (USC) for through-the-body wireless communication (TBWC) for biomedical applications. Detailed (on-body/in-body) simulations using the Sim4Life package and real-world on-body experiments are used to evaluate signal propagation, power transfer efficiency, and path-loss characteristics through this complex media. The results indicate that although USC has a somewhat smaller attenuation at lower frequencies, MRC demonstrates a more consistent performance with frequency variation and a smaller difference between in-body and on-body scenarios. These results, coupled with the fact that the lowest attenuation occurs around 25-30 MHz, for which USC transducers are very difficult to design, suggests that MRC provides a desired technology for intrabody networks.**

***Index Terms*—Magnetic Resonance Coupling; Ultrasonic Coupling; intra-body network; Sim4Life, path-loss.**

I. Introduction

There is an increasing array of human assistive technologies that generally require sensing some vital parameters and accordingly applying or simply recommending some corrective action. The signal may not necessarily be captured in the same place where the action occurs, thereby requiring a small wireless network that can collect all signals efficiently without interference, fuse them together, and decide upon the appropriate action. Such networks are crucial for managing chronic conditions and may include some sensors attached to the skin while others are implanted.

Thus, through-the-body wireless communications (TBWC) become essential to operate these networks. Furthermore, since long-lasting batteries are usually too big for implants, a wireless power transfer (WPT) to the nodes is also required. For the most part, the WPT and communications face similar issues and have resulted in a significant amount of research in Human Body Communications (HBC) technologies.

The two most promising technologies include Magnetic Resonance Coupling (MRC) and Ultrasonic Coupling (USC). Our experiments in [1], [2] show that among the electromagnetic methods, MRC works much better than others and is very robust against movement, posture, clothing, person to person variations (e.g., build, weight, etc.). We have shown that USC also works quite well [3], but being mechanical in nature, it cannot operate at very high frequencies [4]. It also has difficult acoustic impedance matching issues [5]. Thus, this paper explores MRC in much more depth than USC.

Both MRC and USC have been explored in the context of both wearables (on-body) and medical uses (on/in-body) both for communication [6], [7] and WPT [8]. For example, Reference [7] considers ear to ear transmission with 2 centimeters air gap on each side, whereas our goal in this study is to consider a truly on-body/in-body with no air-gap as far as possible. The biomedical application of MRC and USC generally consider very short distances (a few mm to few centimeters). Reference [9] explores an USC powered microprobe for electrolyte ablation. Reference [5] develops a USC power receiver for medical devices and discusses MRC power transfer to medical devices. Reference [10] designs a sub-10-pJ/bit 5-Mb/s MRC transceiver.

There are currently very few detailed studies of MRC/USC propagation through the body at frequencies of a few to few tens of MHz range and distances of tens of centimeters. Much of the longer-range characterization models the body either separately for different types of tissues, or by using average dielectric properties, or in the context of very high frequencies. Reference [11], for example, explores both through COMSOL simulation of EM propagation through a homogeneous soft-tissue media and via some experiments on chicken breast. Reference [7] is typical of channel characterization efforts through measurements and simulations. It considers propagation over much higher frequency region (50Mhz to 2.4GHz) and uses average dielectric properties of human body, which is not appropriate at lower frequencies.

In this paper, detailed (on-body/in-body) simulations using the Sim4Life package and real-world on-body experiments are used to evaluate signal propagation, power transfer efficiency, and path-loss characteristics through this complex media. The results indicate that although USC has a somewhat smaller attenuation at lower frequencies, MRC demonstrates a more consistent performance with frequency variation and a smaller difference between in-body and on-body scenarios. The novelty of this paper lies in its comprehensive experimental and simulation-based analysis, revealing non-intuitive behaviors of MRC and USC within the human body, which have not been thoroughly investigated in previous studies.

The rest of the paper is organized as follows. Section II discusses the magnetic and Ultrasonic Coupling basics. Section III discusses our detailed experimental setup. The systematic comparison of magnetic and ultrasonic coupling through the human body is discussed in section IV. The paper is concluded in section V.

II. MRC and USC Communications

A. Magnetic Resonance Coupling (MRC)

MRC involves resonant communication between two coils. Each coil is in series with a capacitor to create a LC tank that has resonance frequency of $f_r = 1/(2\pi\sqrt{LC})$, where L and C are the inductance of coil and the capacitance of the capacitor respectively. For maximum energy transfer efficiency, the transmitter and receiver sides must have matching resonance frequency. Furthermore, to avoid reflections, the two sides also need to have matching impedance (which is purely resistive at resonance frequency).

The TBWC and other applications of MRC, it is assumed that the technology works in the *near-field* regime, often under the name NFMI (near field magnetic induction). In contrast, the traditional RF communications such as BLE are considered as *far field*. The near-field assumption holds when the operating distance d is much smaller than the wavelength λ_r, or more precisely, $d < \lambda_r/2\pi$ [12]. For example, a popular operational frequency for NFMI is 13.56MHz, which is the standard frequency for RFID operation. At this frequency, the near-field limit is 3.5m in the air. In case of far field, the RF power is transferred through Friis equation for propagating EM waves, where electric and magnetic fields are coupled through Faraday's and Ampere's laws in Maxwell's equations. The concept of near-field is really an idealization and characterized by making some assumptions which effectively decouple the magnetic and electric fields. The resonant LC circuits in MRC produce a strong magnetic field and thus a characterization in terms of current flow in the receiver coil due to magnetic induction is an appropriate characterization. Under the near-field assumption, the received power can be easily related with the transmitted power in terms of various parameters [12]. This relationship suggests the following: The induced current increases linearly with the operating frequency and goes down very rapidly with distance h (as h^{-3}). Since the power is proportional to I_r^2, the induced power decays as h^{-6}, which is much faster than the decay for far-field (or RF) case, where the power goes down as h^{-2}. The very rapid decay of the induced power with distance makes the MI technology inherently a small range technology, and this effect is usually more limiting than the near field requirement of $h < \lambda_r/(2\pi)$. The area of transmit and receive coils (proportional to ρ_t^2 and ρ_r^2 respectively) and the number of turns (K_t and K_r) directly influence the mutual inductance and hence the induced current. Increasing the range requires bigger coils and more turns, both of which may be undesirable in applications where small size is required. The frequency and the transmit coil current directly increase the induced current, and hence the overall power consumption (bad) and the range (good).

B. MRC Transmission in Other Media

For transmission through materials other than air, it is important to consider their electrical and magnetic properties, which affect the speed of EM propagation ("speed of light"). The latter, denoted by c_r, is given by $c_r = 1/\sqrt{\epsilon \times \mu}$, where ϵ is the electrical permittivity and μ is the magnetic permeability of the media with ϵ_r and μ_r as relative values to the vacuum with permittivity ($\epsilon_0 = 8.85410^{-12}$), and permeability ($\mu_0 = 4\pi \times 10^{-7}$). Since $c_r = \lambda_r \times f$, then a high permittivity or permeability reduces both λ_r and c at a given frequency. For the human body $\mu_r \approx 1$ but ϵ_r varies tremendously from organ to organ [13], [14], with absolute values (including real and imaginary parts) ranging from tens to thousands. The imaginary part results from the conductivity of various organs, which also varies significantly. Furthermore, the permittivity is not constant but goes down with the operating frequency. Table I shows a sample of permittivity and conductivity at both 13.56 MHz and 25 MHz and one can see significant variations even within a single organ. Thus the near-field limit inside the body is much smaller than in air. For example, at 13.56MHz, the near field limit for muscle is only about 30 centimeters and at 25 MHz, the limit shrinks to 16.2 centimeters. Thus, the propagation inside the body would operate in the "mid-field" in many practical cases, which yields much more complex and somewhat nonintuitive results that we demonstrate this via measurement and simulation in this paper.

TABLE I: Relative Permittivity and Electrical Conductivity of Various Tissues at Different Frequencies [15]

Tissue type	**13.56MHz**		**25MHz**	
	Rel. Permit.	Elec. Cond.	Rel. Permit.	Elec. Cond.
Blood	210.6	1.12	133.1	1.15
Muscle	138.4	0.63	99.3	0.65
Skin (dry)	285.2	0.24	175.7	0.32
Fat (avg.)	25.4	0.06	18.6	0.06

C. Characteristics of Ultrasound Communications

USC is a very well-researched technology, and has been widely explored for both communications [16] and power transfer [11], [17] in clinical applications. The USC velocity in human tissue is around 1500 m/s. Thus at 1 MHz, the wavelength λ is only 1.5 mm, and even lower at higher frequencies. Ultrasound propagates via pressure waves from a transducer in form of a vibrating diaphragm interfaced with a piezoelectric crystal for electrical to ultrasound conversion on the transmit side. A reverse process occurs on the receive side. The intensity I of USC waves (in mW/cm^2) can be related to the pressure P, the density of the media ρ, and the speed of sound c as $I = P^2/(\rho c)$. FDA regulates this intensity to be 720 mW/cm^2 for most of the applications [18].

As with MRC, USC also shows the near-field and far-field effects. In the near-field region, the pressure waves are irregular but focused, and its extent, denoted by N is given by $N = D^2/(4\lambda)$. Thus, for a transducer diameter of 1 centimeters and $\lambda = 1.5mm$, $N = 16.7$mm. This is the range for which USC has typically been used in biomedical applications and thus will be expected to suffer very little attenuation. Beyond this range, into the far-field range, the pressure $P(d)$ at distance d drops rapidly as $P(d) = P_0 e^{-af^b d}$, where a and b are some positive constants and f is the frequency. Also, since USC waves are mechanical, they scatter at boundaries between two materials (e.g., soft tissue and bone), according to Snell's law. With many boundaries of irregular shape in the body, the net impact of the scattering is a rapid attenuation of the signal, which suggests a rather short tissue penetration range. However, our experimental and simulation results suggest that USC works quite well inside the body – in fact, even better than MRC.

USC performance in an on-body scenario may be enhanced by the phenomenon of Rayleigh surface acoustic waves [19]. Surface acoustic waves (SAW) travel along smooth surfaces and can cover significant distance without much attenuation; however, undulations in the surfaces of the order of a few wavelengths can disrupt them. For on-body applications, we have examined both scenarios, e.g., bare skin (typically quite smooth) and skin covered with clothing or other materials that have significant undulations. Inside the body, however, SAW should not play any significant role.

III. Experimental and Simulation Setups

A. Real Measurements

Consistent with our earlier work on the subject, we use flat circular coil that can be placed securely on the skin with an electric gel to make good contact (see Fig. 1). The coils were shielded magnetically on the top so that there is no signal leakage into the air. The transmit and receive coils were identical and placed on different parts of the volunteer's body. We used coils of several different sizes, although most of the results reported here were obtained using a 20 mm diameter case.

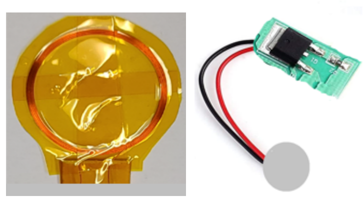


Fig. 1: (a) MRC Antenna (b) USC Antenna

The key novelty of the work reported in this paper is the experimental and simulation based study of MRC at different frequencies. A frequency change is nontrivial as it requires a change in inductance and capacitance of the antennas on both transmit and receive sides in a synchronized manner. A simple solution is to use mechanically variable inductors and capacitors (usually changed by turning a screw) and tune them manually. However, we found such variable inductors/capacitors to be rather unstable to the point of being unusable. The voltage controlled inductors and current controlled capacitors also have similar issues. Therefore, we built LC circuits with suitably chosen capacitors values along with change in the number of turns to change the inductance (Changing coil diameter requires different coils which were not available). Note that the resistors and capacitors come only in certain well-known sizes and have 1% or worse tolerance; therefore, achieving the precise resonance frequency or matching the transmitter and receiver is often a tricky and very time consuming process. We largely achieved this by trial and error.

There are other challenges also brought about by using the antenna on-body. Human body has a small capacitance that must be accounted for in achieving the precise resonance frequency, although this appears to remain constant. Other issues concern the quality of antenna contact with the skin and the skin properties of the person. For example, skin moisture and resistance is a function of skin hydration, humidity, temperature, and the stress level. For example, it is well known stress increases skin conductance (formally known as electrodermal activity or EDA) due to action of sweat glands, although different parts of the body have different concentration of sweat glands and hence different amount of change in conductance [20]. Similarly, skin hydration changes the conductance. Thus, the path-loss results can easily vary by a few dB or more across experiments for the same distance and frequency on the same person.

For our experiments, we asked the volunteer to either sit in a chair or stand on the lab floor. Neither mattered since the lab floor is nonconducting. We also confirmed that different postures (e.g., seated on cushioned chair with feet off the ground) did not make any difference. We measured the communication performance in two ways. First, we connected the transmitter to a signal generator and receiver to vector network analyzer (VNA) to measure the received signal. There was no common ground connection between the signal generator and the VNA, as that would invalidate the results.

For real packet transmissions, we used a pair of USRP (Universal Software Radio Peripheral) N210 boards produced by Ettus Research [21]. The boards enable flexible implementation of software radio including various type of modulation schemes along with the ability to connect different types of antennas. It includes a Xilinx® Spartan® 3A-DSP 3400 FPGA, 100 MS/s dual ADC, 400 MS/s dual DAC and Gigabit Ethernet connectivity to stream data to and from host processors. A modular design allows the USRP N210 to operate from DC to 6 GHz. In our experiments, we used BPSK modulation to study packet delivery ratio through the body at different distances.

B. Simulation Based Evaluation

As discussed earlier, real experiments must be limited to on-body case, and even in that scenario, they are inconvenient to perform on a large scale. This gap can be filled with a comprehensive simulator that supports standard methods to solve Maxwell's equations in a complex environment. Equally important is the availability of highly detailed and realistic human phantom models. There are several open source packages summarized in [22]. Unfortunately, most do not come with human phantom models. Two packages that do include them are CST studio (open source) and Sim4Life (commercial) [23]. Sim4life supports very detailed "Virtual Population" (ViP) phantom models of human body.

For the modeling of the propagation of pressure waves through extremely non-homogeneous media such as tissue and bone, Sim4Life provides a full-wave Acoustics Solver called P-ACOUSTICS. This solver is based on the linear pressure wave equation and is tuned for heterogeneous, lossy materials. Different organs in the human body have very different mechanical properties as well, which result in varying levels of absorption, scattering, reflection, and refraction inside the body.

Realistic modeling of EM propagation through the human body requires accurate handling of surfaces with very different EM properties; therefore, numerical solutions using a fine 3D grid ("voxel") is necessary. The three main methods in this regard are [22]: Finite-differences-time domain (FDTD), Finite Element Method (FEM), and Method of Moments (MoM), or equivalently, Boundary Element Method (BEM). We have used Sim4Life FDTD method in our modeling using full Maxwell's equations (Without considering magnetostatic assumption) and thus the results should be valid regardless of the frequency. However, to avoid error accumulation due to finite differences, the voxels must be rather small in size.

As the main model in Sim4Life, the Duke, an MRI-based full-body model obtained from a virtual population, was used. The model was segmented into 75 anatomical body tissues/organs, with a resolution of 1.5 × 1.5 × 1.5 mm^3 throughout the body. The height and weight of the model were 1.77 m and 70.2 kg, respectively. Since a FEM solver is capable of operating at various frequencies, we obtained the path loss distribution around the MRC/USC system with the human body model using the magneto FEM vector-potential and ultrasonic FEM solver. For USC, we used a parabolic diaphragm with specific focal point.

IV. Results and Discussion

A. Experimental MRC and USC Results

Figs. 2a and 2b show the results for in-body and on-body experiments. For USC, we used a 20mm diameter disc transducer whose thickness must decrease with the frequency. To enable in-body experiments, we have also included experiments on grocery-bought chickens (multiple of them jammed together for longer in-body distances). Fig. 2a shows the RSSI (received signal strength indicator) and Fig. 2b shows the PDR (packet delivery ratio) for both MRC and USC at 8 MHz frequency. It appears that USC shows some advantage over MRC in terms of attenuation and hence the range. It is also seen that the in-body case provides somewhat better performance in all cases than on-body, possibly due to loss at the antenna-skin interface. The range is seen to be a bit more than 30 centimeters in this case. It is seen that at this frequency, USC performs somewhat better than MRC.

B. Comparison of Measurement and Simulation Results

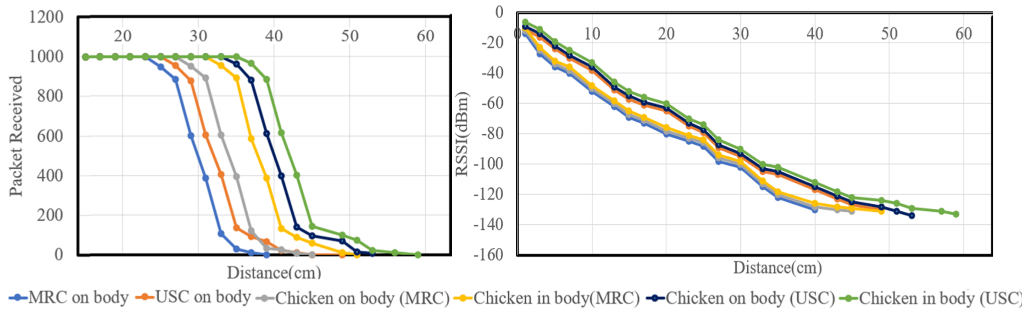
We next ran Sim4Life for several conditions similar to real measurements. Note that the Duke model in Sim4Life does not directly represent the characteristics of any of the volunteers and thus achieving the same path length through the body and the same positions on the body is not possible. Nevertheless, our extensive measurements in [2] indicate that the differences should not be significant. Fig. 3 shows the comparison for a wide range of frequencies from 3 MHz to 50 MHz. It can be seen that in most cases, there is a very good match between the two. In all cases, the measured loss is larger than the simulated one which could be explained by the fact that the simulated situation is ideal (there is no issue of skin-contact quality, field leakage outside the body, or even the minor loss of signal between the measuring equipment and the measured signal on the body). Accounting for this, will perhaps show an even better match. We found such a match in all the other (unreported) cases as well. The somewhat variable difference between measured and simulated values can be attributed to various sources of variations including: (a) body-type of the volunteer vs. that of the phantom, and (b) difficulties and practical issues in setting the precise values of L and C in each case, as mentioned in section III-A (c) unmodeled issues in the simulations.

A consistently good match provides us some assurance on both the simulations done via Sim4Life (and corresponding ViP body models) and the accuracy of our real experiments with volunteers. Note that the experiments can be affected due to parasitic capacitances, unknown ground paths, measurement errors, etc. The most interesting result from Fig. 3 is the "sweet-spot" for the frequency even in a rather limited range of 2-50MHz. In the figure, the path-loss is minimum around 25MHz and increases on both sides. The analytical model for magnetic resonance propagation through the air cannot explain such a behavior. We believe that this is a result of frequency dependent permittivity and conductivity as illustrated in Table I. The frequency at which the minimum occurs depends on several parameters as discussed later, and should not be construed as fixed. Even more significant, the actual values of path-loss very much depend on the Q-factor of the coils. In Fig. 3, the Q-factor is quite high (210 at 25 MHz), which is difficult to maintain for long term operation as discussed in section II-A. If we bring Q down to 100 or lower, the path-loss would likely increase by another 10dB. Another issue is that the Q-factor will change with frequency because of practical difficulties in maintaining the same L/C ratio. However, the experiments do suggest that the behavior remains intact.

Next we report results from some simulations, performed using both USC and MRC for 3, 5 and 8 MHz frequencies. We chose three different parts of the body, On-Body (OB) and In-Body (IB): 1. Right-lower Calf (RLC) to Left-Lower Calf (LLC), 2. Right-upper Calf (RUC) to Right-upper Calf (LUC), and 3. Right Waist (RW) to Left Waist (LW). The results are shown in Table II. As it can be seen, the path loss for MRC is less than the path loss for USC in this frequency. Next, we compare the simulation vs. measurement results for MRC and USC at 3, 5, and 8 MHz frequencies. Table III shows the comparative path-loss while at a distance of 20 centimeters. These results again show a good tracking of simulation vs. measured results for both MRC and USC.

It is also seen that in this frequency range, USC performs better than MRC and the path-loss goes down with frequency. For USC, this behavior is contrary to the equations discussed in section II-C, where the pressure at a given distance supposed to go down exponentially with the frequency. For MRC, the range increases with the frequency, since higher frequency implies higher energy. However, the in-air behavior of power (going down as sixth power of distance) clearly does not hold. A higher water content in the body might be helping ultrasonic waves to achieve more extended range. However, further exploration of physics is necessary to substantiate this claim.

In our experiments we used diaphragm USC transducers where the diaphragm thickness goes down with frequency. We



(a) (b)

Fig. 2: In-body and On-body MRC and USC (a) RSSI Comparison at 8MHz (b) PDR Comparison at 8MHz

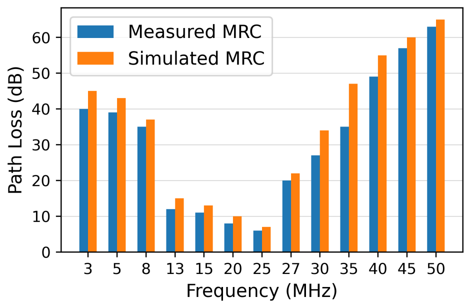


Fig. 3: MRC measured & simulated pathloss vs. freq.

TABLE II: In-body and On-body Simulations at 13.56 MHz

TX position	RX position	Dist. (cm)	USC PL (dB)		MRC PL (dB)	
			IB	OB	IB	OB
RLC	LLC	15cm	12	14	9	11
RUC	LUC	20cm	14	15	12	13
RW	LW	36cm	17	21	12	14

were unable to get transducers with operating frequencies > 8 Mhz and the results for those were obtained via simulation as discussed in the next section.

C. Further Simulation Studies of MRC

Having obtained some confidence in the simulation results, we conducted more extensive simulations, largely focusing on MRC. In particular, we explored the potential difference between in-body vs. on-body placement of the antenna. Fig. 4a shows the comparison of in-body vs. on-body results. It is seen that the MRC path loss is higher for frequencies up to about 10 MHz, and beyond that the difference between the two is very small; in fact, at the typical MRC frequency of 13.56MHz, MRC has slightly lower loss, and at higher frequencies the difference is negligible. Recall that the optimal frequency for MRC is closer to 25 MHz, but USC transducers at that frequency become quite expensive. It is also seen that the difference between on-body and in-body scenarios is rather small, with on-body showing a slightly higher path-loss.

We further studied the behavior of MRC with frequency when the number of turns of the coil are changed. Fig. 4b shows the behavior of 20mm coil for 3, 5, and 7 turns. The results are generally expected – the path-loss decreases with the number of turns. One thing not obvious is slight shift to the left with increasing number of turns. This behavior has been reported in [24] and is again not expected in a homogeneous media like air.

TABLE III: Simulated vs. measured path-loss at 20 cm distance

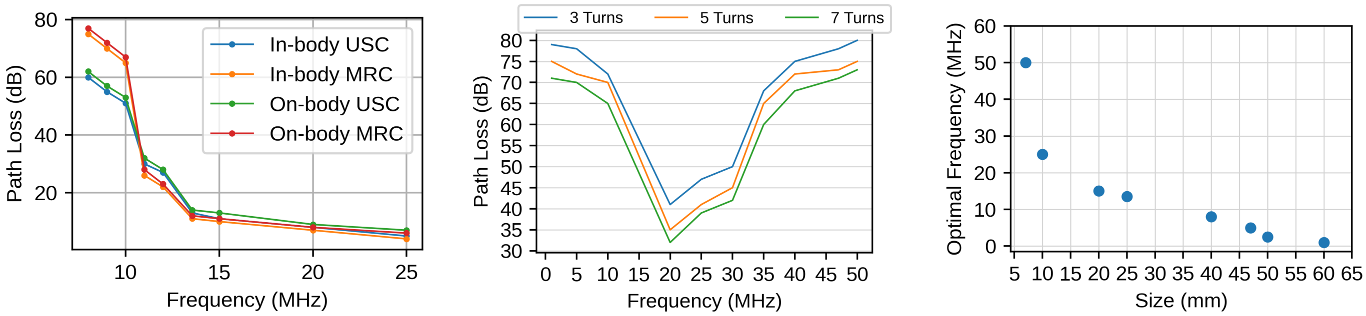
Freq(MHz)	Sim(USC)	Meas(USC)	Sim(MRC)	Meas(MRC)
8	32	35	39	37
5	35	39	38	43
3	37	40	39	45

Given the "sweet-spot" behavior of path-loss vs. frequency, we decided to examine optimal frequency as a function of coil size for MRC. The result is shown in Fig. 4c. It is seen that there is a consistent decrease in optimal frequency as the coil size increases. Please note that these results were obtained by trial-and-error, since there is no equation to indicate the optimal operating point, therefore, the results are approximate. Further approximation errors can be expected due to variation in the Q-factor. To keep the same Q-factor, the ratio of L to C would need to be maintained the same; however, this is difficult to do as explained in section III-A.

Our exploration suggests that the signal propagation behavior through the human body is quite different than in a homogeneous nonconducting media like air. The results are also markedly different than are often observed with idealized modeling of body with average dielectric parameters or by considering tissues of only one type. Our results are, however, consistent with those reported in [24] and [25]. Other researchers have also reported variations in channel gain as a function of frequency for both MRC and USC [7], [11]. However, the complexity of the body and different frequency ranges, dielectric property assumptions, etc. make a direct comparison difficult.

V. Conclusions

In this paper we have examined in detail the characterization of magnetic resonance and ultrasound based communications through the human body, both via direct on-body measurements and via simulations using the Sim4Life package and its ViP phantom human body models. We show a close match between measured and simulated results in spite of many limitations of the measurements, which seems to validate our measurements.



(a) (b) (c)

Fig. 4: (a) MRC and USC simulation result for in-body and on-body cases (b) Simulated Path loss vs. frequency for different # of turns in coil (c) Optimal frequency of operation vs. coil diameter for 7-turn coil

The most significant result from these measurements is that the human body does not behave at all like a homogeneous, non-conducting media like air, and thus the mathematical equations expressing the behavior in homogeneous media cannot be used to characterize the behavior through the body. The simulations and measurements provide many non-intuitive results that are difficult to explain and somewhat surprising. For example, the path-loss as a function of frequency shows a unimodal behavior, with minimum loss at a frequency that is determined by the coil size and number of turns. In the future we will explore such behavior further to understand the reasons and the impact of various factors that are difficult to control in practice.

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