Ultrasonic vs. Magnetic resonance communication for Mixed Wearable and Implanted Devices

Rajpreet K Gulati[1], Krishna Kant[1], Amitangshu Pal[2]

***Abstract*—Human body communication (HBC) has recently been explored extensively both for small wearable electronic gadgets and for implanted sensors to deliver relevant data to implanted therapeutic devices. In this paper we conduct an experimental comparison of two of the promising technologies but for on-body use, namely ultrasound coupling (USC) and magnetic resonance coupling (MRC) based communications. We find that both of these propagate much better through the body than in the air, thereby making them attractive for communications between in-body nodes, in-body to on-body nodes, and on-body nodes where the direct path includes substantial body area. USC also involves a surface acoustic wave (SAW) between on-body nodes which may be broken to varying extent by clothing. We find that with SAW component, USC works better than MRC, but otherwise has similar performance. MRC is very robust and can travel up to the entire body length with 25dB or less loss.**

***Index Terms*—Magnetic resonance coupling; magnetic communication; intra-body sensor network; wireless power transfer;**

I. Introduction

A. Motivation and background

Small electronic devices with smart sensing and communications continue to proliferate both for on-body and in-body use. The former, often described as *wearable computing devices* [1] are being used for an increasing array of assistive functions, the most basic ones being those that contact the body and measure some physiological parameters (e.g., temperature, blood pressure, etc.). The devices may also help augment/enhance human sensing capabilities (e.g., smart glasses or smart hearing aids). They may also be more intrusive and provide stimulus/medication in response to the sensed conditions. Implanted devices often perform all of these functions for dealing with chronic illnesses that continue to increase in an aging population in USA and elsewhere [2].

Although some of these devices can be self-contained, there is a compelling reason for networking these devices together, so that each can do its local function of sensing/actuation in a most energy efficient manner, and the complexities of combining multiple signals and decision making left to a more capable device. There are already several examples of such a need such as bladder control and others [3], [4] where signals from several parts around the bladder must be collected and analyzed to determine the electrical stimulation or drug release amounts. A wearable device such as a smart-watch (or a similar form-factor device attached at a suitable point on the body) can be used as a centralized hub for decision making based on signals from both on-body or in-body devices. The key advantages of an on-body device is high energy capacity (due to easily changed batteries), connection to external devices, and the possibility of supplying energy to other in-body nodes that cannot harvest enough energy on their own.

[1]Department of Computer and Information Sciences, Temple University, Philadelphia, PA, USA, [2]Department of Computer Science and Engineering, IIT, Kanpur, India

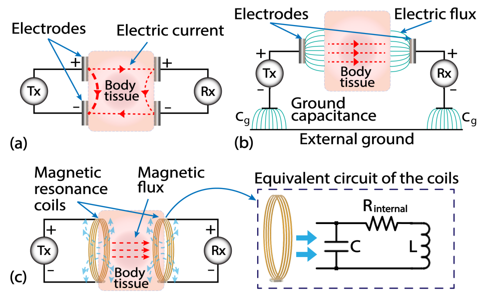

Fig. 1: Intra-body coupling methods: (a) galvanic coupling, (b) capacitive coupling, and (c) magnetic resonance coupling (MRC).

B. Human Body Communications and possible technologies

Thus the communications ability (whether for information or energy transfer) is crucial for all such nodes. Although RF can be used by on-body nodes, RF does have some security issues, e.g., possibility of eavesdropping, intrusion, or jamming by adversaries. Security vulnerabilities of IoT devices in general and in health-case specifically as well, as noted in the recent report (https://cps-vo.org/node/72664), which states that 82% of healthcare organizations' IoT devices have been targeted with a cyberattack within the last year (compared with 80% of organizations overall) and only 7% of attacks had no financial impact. Through the body communications, popularly known as *Human Body Communications (HBC)*, can lessen this concern since the attacker would need to be in close proximity of the person to conduct an attack. Unfortunately, RF itself does not travel well through the body [5], and other mechanisms are needed.

Several HBC mechanisms have been explored in the literature. They require various forms of coupling of electrodes into the body such as galvanic coupling (GC) [6], capacitive coupling (CC) [7], and magnetic resonance coupling (MR) [8], [9]. Fig. 1 briefly illustrates their working principle.

Galvanic coupling (GC): The Galvanic HBC transmits the signal to the human body through a pair of electrodes that are placed in contact with the skin and act as transmitter (T_x) and receiver (R_x) respectively. The two electrode pairs across the body are shown in Fig. 1(a). Due to the low conductivity of the human body, the signal between the transmit and receive electrodes is rather small [10]. The short spacing between the positive and negative terminals on each end results in most of the current flowing locally. So, GC coupling is not an effective way to transfer energy or communicate across the body. To ensure most effective communication, the GC signal frequency ranges from 10 kHz to 100 MHz. A recent study reported a data rate of 1.23 Mbps when transmitting at 200 kHz with attenuation levels typically around 50 dB over a distance of 15 cm [11].

Capacitive coupling: In capacitive coupling (also known as electrostatic coupling), electrodes T_x and R_x are used as shown in Fig. 1(b). The ground electrodes are left floating while the signal electrodes are securely affixed to the body, creating a capacitance with the environment (ground or other objects around them). Capacitive coupling can be modeled using a distributed RC circuit [12]. Recent work shows an attenuation of 20-25 dB at 60 MHz and an on-body distance of 140 cm for capacitive coupling. Additionally, due to the weak nature of the received signal and high dependability on the surrounding environment, capacitive coupling in HBC usually works well only over a short-range [7] making it unusable for use with implantable/wearable devices at a longer distance.

Magnetic resonance coupling: Magnetic resonance coupling occurs when signals are coupled between the two coils $T_{[}x]$ and $R_{[}x]$ through magnetic flux as shown in Fig. 1(c). Both the transmitter and the receiver use an inductive coil in parallel with an identical capacitor to form a resonant LC circuit capable of transferring energy quite efficiently at resonance frequencies. Most MR coupling occurs over a spectral range between about 100KHz and 50 MHz, which produces a maximum attenuation of only 8.1 dB at a distance of 40 cm covered [11].

Ultrasonic coupling: Ultrasound coupling (USC) [13] is a very well-researched technology and has been widely used in various clinical applications [14], and specifically explored for both communications [15] and power transfer [16], [17]. USC is very popular for imaging in the human body, with typical frequencies in 3–6 MHz range. The USC velocity in human tissue varies in the range 1500–2000 m/s, which is quite slow but adequate for medical applications. However, this results in wavelengths of only 0.3–0.7 mm and has implications for penetration depth. Small USC have been used extensively in implants without any reported side effects, and provide a range of 5-10 cm communication range.

C. Our contributions

In this paper we conduct a detailed experimental comparison of two promising technologies for on-body communication devices, i.e. MRC and USC. Our prior work on MRC has shown that it works better than other forms of HBC [18], [19] and is very robust against variations that one would expect in on/in-body environment such as movement, posture, clothing, person to person variations (e.g., build, weight, etc.). It can also provide a range of almost the entire body-length with only about 25db loss. So, we believe that MRC is a good electronic communication mechanism for HBC use. On the other hand, ultrasound has also been explored extensively for intrabody use and therefore we chose this as a potential technology for HBC.

Our experiments indicate that USC can work quite well in on-body settings, particularly due to the phenomenon of Rayleigh surface acoustic waves (SAW) [20]. Surface acoustic waves 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 both scenarios, e.g., bare skin (typically quite smooth) and skin covered with clothing or other materials.

Our experiments also seem to suggest that USC transmission actually works better with increasing frequency. This is again a likely result of the extremely complex environment inside the body. Overall, we find that USC works similar or slightly better to MRC in 3-8 MHz.

D. Paper organization

The paper is organized as follows. In section II we discuss the magnetic and ultrasonic communication basics. Section III discusses our detailed experimental setup. Systematically comparison of magnetic and ultrasonic coupling through human body is discussed in section IV. The paper is concluded in section V.

II. Background on Relevant Communications Technologies

Because of their ubiquity, short-range RF based communication such as BlueTooth Low Energy (BLE) would be a natural choice for our application; unfortunately, RF is known to suffer high signal absorption in aqueous/tissue media [21]. We therefore discuss the brief characteristics of MRC and USC in this section, and study their performances on human body in subsequent sections.

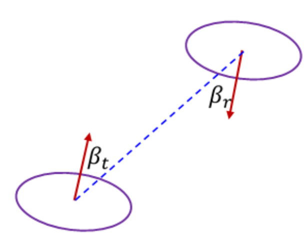

Fig. 2: The intersection angle between two unidirectional coils

MRC works on the principle of magnetic induction between two coils, and the matching of resonance frequency on transmit and receive sides enhances the energy transfer between the two. MRC uses a LC-circuit as antenna on both transmit and receive side. Since a coil with inductance L and a capacitor with capacitance C has resonance frequency of $1/(2\pi LC)$, the resonant energy transfer is easy to achieve, and the antenna can be quite small. We considered a flat coil of diameter 20 mm that contacts the skin directly. The energy transfer in this case can be described by Lenz's law and the detailed equations are given in [22].

Consider a transmit and receive coil pair separated by distance r with the plane of the coils tilted at angles β_t and β_r relative to the axis joining the coil centers, as shown in Fig. 2. Then the magnetic field induced in a receiver coil due to the current flowing through the transmit coil is given by Lenz's law. In particular, the mutual inductance in between the coils can described as [23].

$$M_{t\to r} = M_{r\to t} \approx \frac{\mu\pi\mathbb{N}_t\mathbb{N}_r\rho_t^2\rho_r^2}{2r^3}\left|\cos\beta_t\cos\beta_r - \frac{1}{2}\sin\beta_t\sin\beta_r\right|$$

Here ρ_x and $\mathbb{N}_x$ are the radius and the number of turns in the transmit ($x = t$) and receive ($x = r$) coils respectively, and μ is the magnetic permeability of the medium. In this paper, we assume that the transceivers are of identical dimensions, i.e. $\mathbb{N}_t = \mathbb{N}_r = \mathbb{N}$ and $\rho_t = \rho_r = \rho$.

However, our experiments indicate that these equations do not accurately describe the signal propagation through the human body. The reason is that these equations are intended for simple media like air, but the human body environment is extremely complex. Overall, our earlier experimental work indicated that MRC works substantially better in human body than in air [18], [19].

From equation(II) it can also be seen that the induced magnetic field (and hence the induced current) in the receive coil is maximum when the planes of the two coils are aligned (i.e., $\beta_t = \beta_r = 0$), and goes down rapidly as the mis-orientation increases. However, our extensive experiments did not show much sensitivity to this misalignment [18]. From a practical perspective, this is highly desirable since the relative alignment will often be quite different and may change with body movement and posture change.

As reported later in the paper, we find a similar issue even with USC for human body. The working principle of ultrasound communication can be described as follows: 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 (1,540 m/s in tissue) as follows: $I = P^2/(\rho c)$. As the US wave propagates, the pressure at distance d, denoted $P(d)$ decreases from the initial pressure P_0 as $P(d) = P_0 e^{-\alpha d}$ where α (in nano-Pascal/cm) is the attenuation coefficient. It turns out that α is a function of the carrier frequency f as $\alpha = af^b$ where a and b are attenuation parameters characterizing the media. With b close to 1 for body parts, α is approximately proportional to the frequency, which means that the attenuation in pressure at a given distance d should drop exponentially with f. However, experimentally we do not see such a drop; in fact, the communication range improves with frequency, at least for the frequency range that we were able to experiment with (745 KHZ to 8 MHz). Since US waves are mechanical, they should scatter at boundaries between two materials (e.g., soft tissue and bone), according to Snell's law, which makes the overall intrabody behavior very complex.

III. Our Experimental setup

A. Instruments used

Building out actual circuit boards for communication experiments is an extremely complex task; therefore, we have used available development platforms for this work. There are several such platforms [24], [25], and they typically utilize FPGAs or specialized CPU's for high-sample-rate digital signal processing. We chose USRP (Universal Software Radio Peripheral) [24] because of its widespread use and operating knowledge in the academia and industry. It consists of a motherboard and two daughterboards. The primary processing unit is the motherboard, which includes AD/DA converters (a dual 100 MSPS 14-bit ADC and a dual 400 MSPS 16-bit DAC) and an FPGA unit (Spartan 3A-DSP 3400). The daughterboards are radio frequency (RF) front ends that connect the device to a transmitter or receiving antennas. We utilize LFTX and LFRX daughterboards that run from DC to 30 MHz, which covers frequency range of interest that we are interested in.

We used USRP N210 and connected LFTX/LFRX to the antenna (USC or MRC) for our experiments. Such a setup allows us to transmit actual packets with suitable frame encoding. We used the simple BPSK (binary phase shift keying) in these experiments. It is certainly possible to use more sophisticated schemes (e.g., QPSK or higher) to increase the packet rate; however, high packet rate is usually not necessary in most healthcare/well-being applications; instead, the the more important aspect is energy consumption, for which the simplest scheme is the best.

We measured both the packet received and power received at the receiver at different distances and frequencies. The transmit power was maintained at 0.3 mW throughout the study. It is reported under the safety threshold limit according to the IEEE standard for safety levels with respect to human exposure [26]. An unregulated exposure to non-static electromagnetic fields may adversely affect the health of humans [27]. There was no sensation reported throughout the duration of the experiment due to the low transmission power, which did not cause any localized heating or absorption by the tissue [18] [19]. Note that all distances were through-the-body distances, and each frequency change required a change of transmit/receive antennas and a re-calibration to ensure accurate measurements.

B. Experimental protocol with human subjects

All experiments in this study were conducted on the human body and in the air, the former being done under a fully approved by IRB protocol #28089 at Temple. The experiments here mostly involve a single middle-aged volunteer. Our earlier work [18] had conducted experiments on 6 very different adults and found the variations in received power among them confined to a few dB. Therefore, we believe that our results are quite representative. Also note that since the many different points on the body were used for measurements, transmission path for both USC and MRC includes subcutaneous fat, muscles, bones, blood, etc. For practical reasons, all experiments had to be on-body; however, we have in the past done experiments by putting transmitter/receiver inside store-bought chickens. These results also indicate better transmission inside the body than in the air.

TABLE I: USC transceiver details used for experimentation

Parameter	USC-1MHz	USC-3MHz	USC-5MHz	USC-8MHz
1. Diameter(Dia)	20 mm	20 mm	20 mm	20 mm
2. Thickness(Th)	2.1mm	0.7mm	0.4 mm	0.2 mm
3. Frequency	1 MHz	3 MHz	5 MHz	8 MHz

TABLE II: MRC transceiver details used for experimentation

Parameter	MRC-3MHz	MRC-5MHz	MRC-8MHz
1. Diameter(Dia)	20 mm	20 mm	20 mm
2. Frequency	3 MHz	5 MHz	8 MHz
3. Capacitance(pF)	400pF	140pf	56 pF
4. Inductance (uH)	7 uH	7 uH	7uH
5, Number of turns	7	7	7

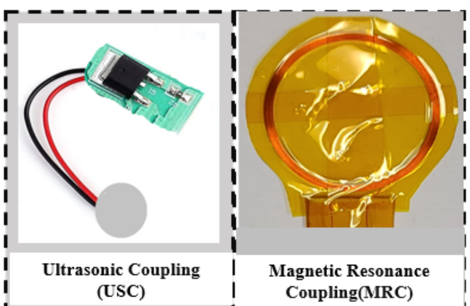

Fig. 3: Illustration of USC and MRC transceivers

Fig. 3 shows the pictures of USC and MRC transducers. USC transceivers were operated at 1, 3, 5, and 8 MHz frequencies. The USC transceivers were of disk form-factor with diameter of 20 mm, and thickness depending on the frequency; the thickness goes down with frequency as shown in Table I, and is already too small (0.2mm) at 8 MHz, thereby requiring extreme care in attachment and issues of fragility. We were unable to easily order USC transceivers operating at even higher frequencies, although they can be custom ordered. Note that the USC output on the receiver side needs to be rectified to have a DC output. We built a standard full-rectifier bridge for this purpose.

The MRC experiments were also conducted at 3, 5, and 8 MHz. For MRC, the antenna consists of a LC circuit (coil and a capacitor). The coil diameter was again 20 mm, and the physical size of MRC and USC were similar. The complete set of parameters for MRC LC circuits are shown in Table II. The inductance (made out of 7 turns of 34 AWG) of 7 μH is used with the planar coil. The transmitter and receiver coils were covered by a specialized magnetic shielding film (WMF200, Woremor) to minimize magnetic interference from nearby electronic equipment and over-the-air transmission [28]. The size is a crucial parameter for intra-body use, and a 20mm diameter is workable according to discussion with experts. The standard operating frequency for MRC is 13.56 MHz, but we tuned it down to lower frequencies by changing the capacitor and/or number of coil turns.

In a single run, 1000 packets were sent from the transmitter to the receiver for both USC and MRC. The distance between transmitter and receivers was incremented in steps of 3 cm, in 1-50 cm range. For both USC and MRC experiements, the transmitter was attached to the palm of the volunteer, and the receiver was placed on the ventral side of the arm and incrementally moved from the palm to the shoulder position. We ensured a good contact with the skin by using appropriate gel (i.e., ultrasonic jell for USC, and electrostatic jell for MRC). Both transmitter and receiver transducers were taped over to keep them on securely. In case of MRC, we used magnetic shielding of each to avoid through-the-air transmission but no such shielding is necessary for USC; however, we do need to mind the surface acoustic waves traveling along the skin surface between transmitter and receiver.

The packet frame structure is shown in Fig. 4 to send the data packets between transmitter and receiver. The minimum packet size used was 56B.

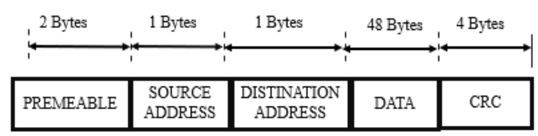

Fig. 4: Packet Frame Structure

By careful comparison between sent and received data, we found that in all cases, the CRC was able to detect the error; therefore, all packets received without the CRC error represent packets that do not suffer from any bit flips.

IV. Results and Discussion

A. Comparison of USC with different frequencies

We first demonstrate the performance of our USC transceivers operating at different frequencies. Fig. 5 shows the number of packets received correctly with varying transceiver distances. From this figure we can observe that the transmission range of USC-8MHz is $\sim$25 cm (with 100% packet delivery), which reduces with the decrease in operating frequencies. Therefore, we can conclude that the USC works better with increasing frequencies; we surmise that this is due to the complex propagation characteristics of the ultrasound signals inside human body.

B. Ultrasonic communication with different surfaces

We now demonstrate the effect of clothing and corrugated surfaces on USC. The results are shown in Fig. 6 and

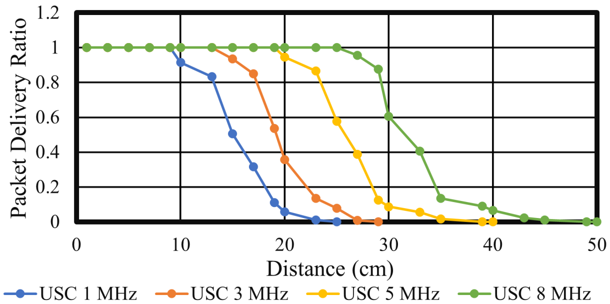

Fig. 5: Packet delivery ratio vs transceiver distances with different operating frequencies

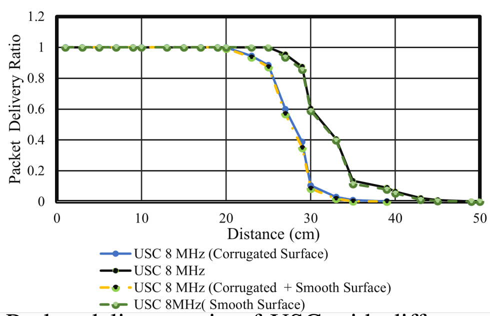

Fig. 6: Packet delivery ratio of USC with different surfaces

7. To show the effects on clothing, we have created two scenarios: (a) cover the transceivers with cloth, and (b) cover the medium (i.e. arm in our case) in between the transceivers with cloth. We use a cotton cloth with a thickness of $\sim$ 5mm to conduct this experiments. From Fig. 6, we can observe that the effect of clothing is almost negligible, which also makes it suitable for scenarios where the wearables are worn inside normal cloths.

We next use a corrugated cardboard to cover the arm instead of clothing and repeat the same experiment. We keep the thickness of the corrugated surface to be $\sim$5 mm, to make a fair comparison with the clothing effects. From Fig. 7, we can observe that the corrugated surface dampens the signal quite a bit, which results in lesser packet delivery as observed from Fig. 6. We surmise that this different effects of corrugated vs clothing is mainly due to the porosity of the clothing materials as compared to solid corrugated cardboard, however further studies are required to confirm the same. In short, the surface acoustic wave is greatly impacted by corrugated surface and is not affected by clothing materials.

C. Ultrasonic vs Magnetic Resonance coupling

From our experiments we can observe that both MRC (see [18]) and USC show promising propagation characteristics on human body; we therefore compare the effects of these two technologies in this section. The transceiver details used are reported in Table I-II. This comparative study is shown in Fig. 8, 9, 10 and 11, where we varied the operating frequencies from 3-8 MHz. From this comparative study we

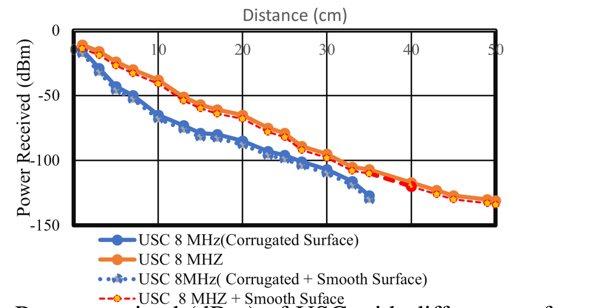

Fig. 7: Power received (dBm) of USC with different surfaces

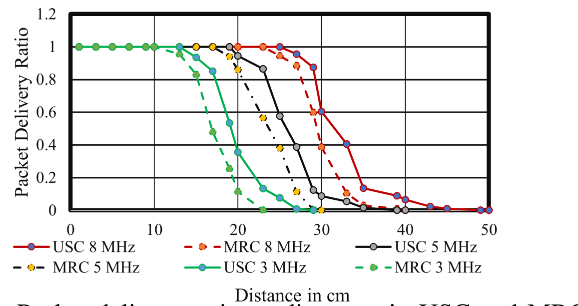

Fig. 8: Packet delivery ratio vs distances in USC and MRC

can observe that both USC and MRC work somewhat similar in 3-8 MHz, in fact USC performs marginally better that MRC in this band. For example at 8 MHz, the packet delivery drops to $\sim$80% at 30 cm for USC; with same delivery performance the range drops to $\sim$ 28 cm for MRC. Higher concentration of water in the body, might be helping ultrasonic waves achieve a longer range, however, further experimentation with other frequency bands need to be conducted to further strengthening this claim.

V. Conclusions and Future Work

In this paper, we have compared ultrasonic coupling versus magnetic resonance coupling-based communication for application in on-body and intra-body nodes (with the signal being communicated within the body). We showed that the ultrasonic coupling (USC) works much better than magnetic resonance coupling (MRC) for transmission through the body at 8MHz frequency. Specifically, it is seen that at 0.3 mW transmitted power, USC based communication shows a range of 50 cm without data loss whereas MRC shows comparable performance only up to 40 cm (25% increase in communication range for USC based communication). Future work will involve exploration into frequencies above 8 MHz. This was not done in the current study since higher frequency USC transducers need to be custom ordered and are not easy to procure. They would also be much thinner thus making their use challenging. We will also examine robustness of ultrasound communications under a variety of scenarios and with different volunteers similar to what we have already done with MRC.

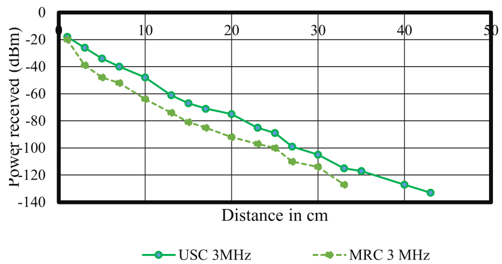

Fig. 9: Variation of power received(dBm) for ultrasonic and magnetic coupling in 3 MHz

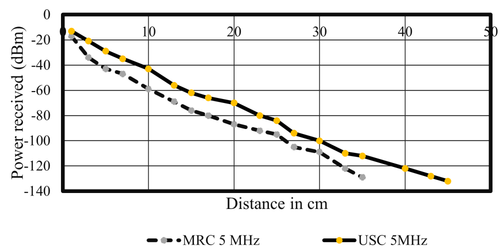

Fig. 10: Variation of power received(dBm) for ultrasonic and magnetic coupling in 5 MHz

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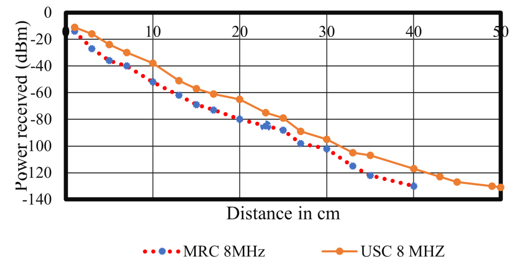

Fig. 11: Variation of power received(dBm) for ultrasonic and magnetic coupling in 8 MHz

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