Multi-access In-vivo Biotelemetry using Sonomicrometry and M-scan Ultrasound Imaging

Sri Harsha Kondapalli, Student Member, IEEE, Yarub Alazzawi, Student Member, IEEE, Marcin Malinowski, Tomasz Timek, and Shantanu Chakrabartty*, Senior Member, IEEE

Abstract—Objective: In this paper we investigate the use of commercial off-the-shelf (COTS) diagnostic ultrasound readers to achieve multi-access wireless in-vivo telemetry with millimeter-sized sonomicrometry crystal transducers. Methods: The sonomicrometry crystals generate ultrasonic pulses that supersede the echoes generated at the tissue interfaces in response to M-scan interrogation pulses. The traces of these synthetic pulses are captured on an M-scan image and the transmitted data is decoded using image deconvolution and de-blurring algorithms. Results: Using a chicken phantom and 1.3 MHz sonomicrometry crystals of diameter 1 mm, we first demonstrate that a standard ultrasound reader can achieve bio-telemetry data rates up to 1 Mbps for implantation depths greater than 10 cm. For this experiment the maximum power dissipation at the crystals was measured to be 20 µW and BER of the telemetry link was shown to be 10^-2. We also demonstrate the use of this method for multi-access bio-telemetry where several sonomicrometry crystals simultaneously transmit the data using different modulation and coding techniques. Using a live ovine model we demonstrate a sonomicrometry crystal implanted in the sheep’s tricuspid valve can maintain a continuous, reliable telemetry link at data rates up to 800 Kbps in the presence of respiratory and cardiac motion artifacts. Conclusion: Compared to existing radio-frequency and ultrasound based bio-telemetry devices, the reported data-rates are significantly higher considering the transducer’s form-factor and its implantation depth. Significance: The proposed technique thus validates the feasibility of establishing reliable communication link with multiple in-vivo implants using M-scan based ultrasound imaging.

Index Terms—Ultrasound, Wireless sensors, M-scan Imaging, Sonomicrometry Crystal, Bio-telemetry.

I. INTRODUCTION

ULTRASOUND imaging technology has undergone a revolution during the last decade due to the availability of transducers that can operate over a large range of frequencies and also due to the availability of high-speed, high-resolution analog-to-digital converters and signal processors. Existing clinical and FDA approved bench-top ultrasound systems are able to generate real-time high-resolution images at a rate of more than 7000 frames per second [1]. In literature ultrasound imaging systems with frame rate as high as 10000 frames per second [2] and with very high resolution [3] [4] have also been reported. On the other end of the spectrum, portable and hand-held ultrasound systems like GE VScan, Siemens P10, Philips Lumify and SignosRT (shown in Fig. 1) can also generate high-speed real-time scans and have been used for diagnostic imaging in non-clinical environments. The large data acquisition and computational bandwidth on these portable and bench-top ultrasound imaging systems could potentially be leveraged for designing in-vivo, high-speed, multi-access bio-telemetry links that can be used for communicating with multiple devices implanted in-vivo. Examples of implantable devices that require high-bandwidth telemetry links are shown in Fig. 1, and their applications range from swallowable imaging systems, neural implants, cochlear, retinal prosthesis etc. The typical data-rates for these implants and their respective implantation depths are summarized in Table I.

In literature, radio-frequency (RF) based telemetry systems have been proposed for high-speed telemetry [10]; however,
the implantation depths and form-factors are constrained by medical compliance limits (local Joule heating). As an example, an RF telemetry system reported in [11]- [12] was shown to achieve data-rates of 120-450 Kbps for implantation depths less than 2cm. Ultrasound based telemetry systems on the other hand can penetrate deeper into biological tissue while operating within the limits of medical compliance and without any side-effects due to long-term exposure [13]. In [14], an ultrasonic telemetry system (form factor of 10 mm) was reported to achieve data-rates of 70-700 Kbps, however, in an ex-vivo environment. Recently, [15] have demonstrated ultrasonic communication data rates of more than 20 Mbps using orthogonal frequency-division multiplexing (OFDM) technique. However the size of the 5MHz transducer used in [15] is relatively large for in-vivo implantation. In [16] an ultrasonic backscattering was used to achieve data-rates up to 500Kbps, however, the implantation depths that were reported were less than a cm. In our previous work [17], we have shown that a millimeter-scale sonomicrometry crystal could be used as transducer for ultrasonic communications in-vivo and hence can be easily implanted underneath the skin or deep inside the tissue. In [18] we also demonstrated that the crystal could be used to harvest energy from a COTS ultrasonic imager and the harvested energy was used for establishing a low-data-rate (less than 100bps) telemetry link. In this paper, we investigate the use of a COTS M-scan ultrasound imager to establish and maintain a communication link with multiple implanted crystals. We show that data rates up to 1 Mbps can be reliably achieved for implantation depths greater than 10 cm.

This paper is organized as follows: Section II presents a brief overview of ultrasound imaging and its use for high-speed telemetry. We discuss different factors that will limit the data-rates that can be achieved using the proposed method. Section III presents experimental results using a chicken tissue as a phantom. We report communication strategies and results for both single crystal and multi-crystal experiments. Section IV describes experiments with a live ovine animal model where the crystal is implanted inside the tricuspid valve of a sheep and interrogated using a crystal probe sutured underneath the skin. Section V discusses about the limitations and factor effecting the proposed approach. Section VI concludes the paper.

II. PRINCIPLE AND LIMITS OF ULTRASOUND IMAGING BASED TELEMETRY

The principle of M-scan and B-scan ultrasound imaging is illustrated in Fig. 2. Using either a beam-forming or mechanical focusing technique, an ultrasound reader generates interrogation pulses directed along each of the imaging line, as shown in Fig. 2 a. In a conventional imaging mode, the pulses are reflected from the tissue boundaries (labeled as '1', '2' and '3') due to mismatch in their respective acoustic impedance. After generating an interrogation pulses (labeled as 'a' and 'b'), the ultrasound reader listens for reflected pulses (labeled as '1a-3a' and '1b-3b'). The time-of-arrival of the pulse indicates the depth of the interface and the magnitude of received pulses indicates the nature of the tissue boundary (for example muscle-blood or fat-muscle interface). A sample waveform of the received signal is shown in Fig. 2 (b) where the information has been mapped onto a grey-scale pixel value resulting in a single frame of an M-scan image. Information corresponding to multiple interrogation pulses are fused over time to generate a composite M-scan as shown in Fig. 2(c).

A. M-scan based Telemetry

In this paper we exploit the process of M-scan imaging to perform telemetry with implanted ultrasonic transmitters. The principle relies on mimicking the reflected echoes using implanted sonomicrometry crystals (Tx, Tx1 or Tx2), as shown in Fig. 3 (a),(c). As long as the magnitude of the pulses generated by the crystal is larger than the echoes generated due to the ultrasonic interrogation pulses, and as long as the frequency of the emitted pulses match the frequency of the interrogation pulses, the M-scan or a B-scan reader should be able to detect and decode the data encoded in the transmitted pulses. A synthetic M-scan image generated using the received pulses is illustrated in Fig. 3 (b). Similarly, when multiple crystals are emitting pulses based on their respective data streams (as shown in Fig. 3 (c)), the reader generates a composite M-scan image as shown in Fig. 3(d). The proposed M-scan imaging framework also supports telemetry with multiple crystals using multi-access techniques reported in standard communication literature [20] like frequency division multi access (FDMA) or code division multi access (CDMA). In this paper we have used on-off-keying (OOK) for data modulation and Walsh-Hadamard codes for multi-access communication. An example of multi-access communication protocol is illustrated in Fig. 3 (d) for two crystals, where each of the crystals uses its respective FDMA based orthogonal code (‘00001111’ for crystal 1, ‘00110011’ for crystal 2) to encode bit ‘1’. When transmitting bit ‘0’, both the crystals do not emit any echoes. Note that the size of the code determines the number of simultaneous telemetry links that can be established with the M-scan reader and determines the maximum data-rate for each individual crystal. Also, for the proposed telemetry method, the interrogation pulses generated by the ultrasound reader do not play a significant role. However, in our previous work [17] we demonstrated that the interrogation pulses could be used to remotely power the telemetry interface. While this attribute is important in realizing a fully integrated, remotely powered wireless device, our focus in this paper is to only investigate the limits of M-scan based ultrasound telemetry.

B. Limits of Ultrasonic Imaging for Telemetry

From a theoretical point-of-view, there exists an upper and a lower limit on the data-rates that can be achieved using M-scan or B-scan based telemetry. One factor that will determine the theoretical upper limit on data-rate will be determined by maximum allowable heat dissipation in biological tissues. In literature this limit is given as $7.2 \text{ mW/mm}^2$ [21] and any telemetry interface has to ensure that its overall power-dissipation is below this limit. Another factor that will determine the upper-limit is the nature of the channel point spread.
function which will allow the reader to differentiate between two independent echoes. Note that a-priori knowledge of the channel response could be used to apply channel equalization techniques and recover overlapping echoes. While this will not be the main focus of this paper, we will illustrate the potential of channel equalization in section V.

The lower-limit on the data-rate is determined by the minimum transmitted power which in turn is determined by the following two factors: (a) ultrasonic attenuation characteristics of the in-vivo medium; and by (b) the noise in the medium. The attenuation characteristics of ultrasound in-vivo has been extensively studied [22], [24] and is typically modeled in terms of pressure level \( P_r \) as a function of depth \( d \) and frequency \( f \) according to

\[
P_r(f, d) = P_0 e^{-\alpha f^\beta d}
\]

where \( P_0 \) is pressure level generated at the surface of the ultrasonic transmitter and \( \alpha \) and \( \beta \) are constants which are determined by transmission media. Thus, based on the equation 1, the transmitter has to generate sufficient pressure to overcome channel attenuation such that a minimum detectable pressure level is received at the receiver. Table II shows typical values of the attenuation parameters obtained using a chicken tissue phantom. A first-order calculation using these parameters illustrates that at 1.3MHz frequency, ultrasound attenuates only by 10.83dB at a depth of 3cm. In comparison,

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Definition</th>
<th>Example (chicken tissue [19])</th>
</tr>
</thead>
<tbody>
<tr>
<td>( f )</td>
<td>Frequency</td>
<td>1.3MHz</td>
</tr>
<tr>
<td>( d )</td>
<td>Propagation Distance</td>
<td>3cm</td>
</tr>
<tr>
<td>( P_0 )</td>
<td>Initial Pressure</td>
<td>1 (N/m²)</td>
</tr>
<tr>
<td>( \alpha )</td>
<td>Attenuation Parameter</td>
<td>0.086 \times 10^{-\beta}</td>
</tr>
<tr>
<td>( \beta )</td>
<td>Attenuation Coefficient</td>
<td>1.5</td>
</tr>
</tbody>
</table>
a 1MHz radio-frequency wave attenuates by more than 50dB for the same depth [23].

The second factor which determines the minimum transmitted power is the level of channel noise. For the proposed M-scan based telemetry, the main source of noise is due to the echoes (due to specular reflection and scattering of incident interrogation pulse) generated at the tissue interface due to acoustic impedance discontinuities. Note that changes in pressure level due to thermal vibrations will be negligible compared to the magnitude of the echoes, and hence has not been considered. Table III shows the reflection coefficient ($\eta$) corresponding to different tissue interfaces and is determined by the ratio between the intensities of the reflected and incident acoustic wave [24].

Combining equation 1 with the respective reflection coefficient of the tissue interface located at depth $d$, the minimum transmitted power ($P_{min}$) can be estimated (relative to the interrogation power) as

$$P_{min}(dB) = 2 \log \eta - \alpha f^3 d$$  (2)

Table III summarizes the minimum transmission power estimated using equation 2 for a frequency of 1.3MHz and for a depth of 3cm. While these approximate values represent the lower limit of power in dB, it only takes into account the specular reflections (not considering the scattering and thus representing the lowest upper bound) for a single tissue interface. In reality, a tissue can be more complex and therefore may need much higher power to overcome the effects due to echoes.

### Table III

<table>
<thead>
<tr>
<th>Tissue Interface</th>
<th>$\eta$</th>
<th>$P_{min}(dB)$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Muscle-Blood</td>
<td>0.023</td>
<td>-7.485</td>
</tr>
<tr>
<td>Bone-Muscle</td>
<td>0.637</td>
<td>-0.903</td>
</tr>
<tr>
<td>Fat-Muscle</td>
<td>0.109</td>
<td>-4.433</td>
</tr>
<tr>
<td>Bone-Fat</td>
<td>0.697</td>
<td>-0.721</td>
</tr>
<tr>
<td>Skin-Blood</td>
<td>0.015</td>
<td>-8.347</td>
</tr>
<tr>
<td>Skin-Muscle</td>
<td>0.009</td>
<td>-0.489</td>
</tr>
</tbody>
</table>

A. Experimental setup

Omnidirectional sonomicrometry crystals were purchased from Sonomicrometry Inc. and were used to construct a single element M-scan ultrasound reader (’Rx’) and for implementing the in-vivo transmitters (’Tx 1-3’). The Sonomicrometry crystals shown in Fig. 4 (c) are made with PZT-5H ceramic material and are coated using teflon to ensure that the crystal is biocompatible and can be chronically implanted. The impedance characteristics and the frequency response of the crystals were first measured using an Omics Bode 100 vector network analyzer. Fig. 4 (d) shows the measured frequency response when the crystal is inserted in a water medium and Table IV summarizes the specifications of the sonomicrometry crystal. The ultrasound reader was implemented using a programmable GS200 echoscope (from gaMPT, Germany) and the ultrasonic transmitter was implemented using function generators (Tektronix DG4102, 100 MHz Arbitrary Waveform Generator), as shown in Fig. 4. M-scans from the echoscope were acquired using personal computer using a GS-Echoview software, which is an application provided by gaMPT mbH. Table V summarizes different measured parameters like transmitted power, peak-to-peak voltage applied to the piezo crystals and settings of the echoscope used for the experiments reported in this paper. In particular, ‘depth’ in Table V refers to the distance of implanted crystals from the surface of skin. Product of ‘Frame size’ (acquisition length) and ‘sampling frequency’ determines the number of rows in the scan reconstructed as illustrated in Fig. 2 (c). ‘Frame rate’ refers to pulse repetition frequency which determines the number of columns generated per second in the reconstructed image.

### Table IV

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Specification</th>
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</thead>
<tbody>
<tr>
<td>Resonant Frequency</td>
<td>1.3MHz</td>
</tr>
<tr>
<td>Material</td>
<td>PZT-5H Teflon coated</td>
</tr>
<tr>
<td>Diameter</td>
<td>1.0mm</td>
</tr>
<tr>
<td>Crystal Capacitance</td>
<td>129 – 290pF</td>
</tr>
<tr>
<td>Bandwidth</td>
<td>≈ 200KHz</td>
</tr>
<tr>
<td>Maximum Input Voltage</td>
<td>around 1kV</td>
</tr>
</tbody>
</table>

### Table V

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>Transmitted Power</td>
<td>0 – 18µW</td>
</tr>
<tr>
<td>Pulse amplitude ($V_{pp}$)</td>
<td>1-20 V</td>
</tr>
<tr>
<td>Pulse frequency</td>
<td>800 KHz</td>
</tr>
<tr>
<td>Shape</td>
<td>Square Pulse</td>
</tr>
<tr>
<td>depth</td>
<td>5cm - 12cm</td>
</tr>
<tr>
<td>H$_{ref}$ Gain</td>
<td>15 dB</td>
</tr>
<tr>
<td>Frame size</td>
<td>10-100 µsec</td>
</tr>
<tr>
<td>Speed of sound</td>
<td>1460m/sec</td>
</tr>
<tr>
<td>sampling frequency</td>
<td>10MHz – 100MHz</td>
</tr>
<tr>
<td>Frame rate</td>
<td>2.5 KHz</td>
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</table>

B. Methods

In this paper we have verified the proposed M-scan based telemetry using chicken tissue and live ovine model as two mediums for ultrasound propagation. The transmitted and received power at each of the crystals implanted in-vivo were measured using a source and a load resistor respectively, as shown in Fig. 5 (a)-(b). $P_T$ and $P_R$ shows the equations for time average measure of transmitted and received power at the crystal.

The acquisition of the M-scan data was performed using echoscope, connected to the ultrasound reader ‘Rx’, which was preset to low transmission gain and high receive gain. This ensures reduction of interference due to echoes generated during M-mode acquisition. Post-processing of the acquired ultrasound image (M-scan) was performed using MATLAB and was used to demodulate and decode the received data. The post-processing involved rasterizing the 2D intensity matrix.
Figure 4. (a) Experimental setup used for verifying and characterizing the M-scan telemetry link. (b) Setup showing the sonomicrometry crystals implanted inside chicken tissue. (c) Millimeter-scale sonomicrometry crystal used for transmission and reception. (d) Measured transmission response with respect to frequency for a sonomicrometry crystal immersed in water.

Figure 5. (a) Setup used for measuring the maximum transmitted power $P_T$ where the crystal was modeled using its equivalent load impedance $Z_T$. (b) Setup used for measuring the maximum received power $P_R$ where the crystal is modeled as a combination of voltage source driving an impedance $Z_T$ in series with a load resistor $R_L$.

Figure 6. Plot showing the variation in received power with the change in distance between the transmitter and the reader.

Figure 7. Plot showing the variation in the power received at Rx by varying the transmit power at Tx when they are at a distance of 3 cm and 10 cm. '1' or '0'. In this paper, each M-scan comprised of a million pixels and the window size varied from 10-20 pixels based on the pulse-frequency of the transmitter. The reconstructed bit sequence was then compared with transmitted bit sequence to
estimate the bit-error-rate (BER). In the case of multi-access communication, the reconstructed bits are compared with each of the encoded bit sequence and the information from each transmitter is recovered independently.

C. Single Crystal Experimental Results

The first set of experiments uses a phantom constructed using chicken breast tissue, as shown in Fig. 4, where the transmitter crystals are labeled as ‘Tx1-3’ and the reader crystal is labeled as ‘Rx’. While the transmitter crystal was implanted/sutured inside the tissue at different depths, the reader crystal was mechanically stabilized by suturing the probe underneath the top tissue layer. In this manner, any acoustic impedance artifacts due to the attachment of the crystal to the tissue surface was alleviated. The configuration of the function generators driving each of the transmission crystals are summarized in Table V, where the maximum transmission power has been measured using the sensing circuit shown in Fig. 5(a). In the first experiment, the distance between the transmitter crystal and the reader was varied. The attenuation in the tissue was measured (using the sensing circuit shown in Fig. 5(b)) and is shown in Fig. 6 and confirms to the log-linear attenuation model given by equation 1. Fig 7 shows the measured results where the received power (to a 50 KΩ resistive load) is plotted against different levels of transmitted power for two implantation depths (3 cm and 10 cm). The measured results show that for an upper-limit (7.2 mW/mm²) on the power dissipation on the transmitter, the receiver can receive power levels of 500 nW/mm² and 1 μW/mm² for respective implantation depths of 10 cm and 3 cm. These results are aligned with the experiments that were previously reported in [17] where we demonstrated that useful power can be harvested from interrogation pulses generated from a ultrasound reader.

The next set of experiments were designed to show the feasibility of ON-OFF keying using standard M-scan based telemetry. Fig. 8(a) and (b) show the M-scan images and the corresponding section of their A-scans for water and chicken tissue as the transmission medium respectively. The encoded data-stream shown in Fig. 8(c) where ‘1’ and ‘0’ are encoded using ON-OFF keying. Each of the M-scan images clearly show the modulation effect due to ON-OFF keying. However, the A-scans show that the spread around the peak is narrower in water medium as compared to the spreads when the medium is chicken tissue. The overlap between the consecutive ‘1’s which could arise due to channel effects inside the tissue and due to the pre-processing used for M-scan image generation by the echoscope. Note that for post-processing, each of the A-scan frames in the M-scan image needs to be arranged (reshaped as described in section III B in a form of 1D array and any error in aligning these frames by the echoscope would lead to an increase in bit error-rates. Also, the limited buffer size on the echoscope will affect the quality of the M-scan which in-turn will affect the quality of the communication link. Higher sampling rate at the reader leads to better quality of the M-scan and longer frame-size leads to accommodating more number of receiver packets which improves the quality of communication link. Hence the choice of frame-size and the sampling frequency, considering a constant buffer size at the reader, becomes crucial in M-scan based telemetry. For all the results presented in this paper the frame-size has been chosen to be 100 μsec and the sampling frequency has been set to 10 MHz.

D. Multi-access Telemetry Results

Multi-access telemetry is demonstrated using three transmitters communicating with the ultrasound reader at the surface of the tissue. In the first set of multi-access experiments, a 16 bit Walsh-Hadamard orthogonal code is used to encode the transmitted data, similar to that of a spread-spectrum based communication system. This approach is similar to the previously reported methods [14], [25] where a combination of pulse position modulation and pseudo random codes of variable length are used to provide multi-access using ultrasound. For this work, we used a fixed length, unique 16 bit codes to represent ‘0’ and ‘1’ at each of the transmitters. As shown in the Fig. 10, the received data can be demodulated
After post-processing the M-scan images for conditions where 20 KHz, 30 KHz and 50 KHz) to modulate the carrier frequency. 

The transmitters then used a periodic signal (at frequencies 20 KHz) to transmit a train of ultrasonic pulses at a frequency of 800 KHz. Each of the transmitters programmed to implement a multi-access scheme similar to frequency division multiple access (FDMA). The equivalent quantization noise (finite resolution) of the M-scan reader.

The system BER decreases with the increase in the transmitted bit streams corresponding to each of the transmitters. The bit streams have verified combined data-rates up to 2.4 Mbps. Note that FDMA encoding constitutes a special case of spread-spectrum communication and hence its BER performance will be similar to that of the results shown in Fig. 11.

### IV. Experiments using an Ovine Model

In the final set of experiments, the proposed M-scan telemetry was verified in a live ovine model. The objective was to verify that a reliable communication link can be established when the sonomicrometry crystal implanted in the tricuspid valve, as illustrated in Fig. 14(a), and is continuously subjected to motion artifacts (movement of the valve in this specific case). The surgical procedure for implanting the crystal was performed on an adult male sheep, in a fully equipped and accredited animal facility at West Michigan Regional Laboratory which is a part of the Spectrum Health Delivery System.

#### A. Adult male sheep preparation

The animal was studied in an acute, open-chest fashion to facilitate validation of the proposed telemetry technique. Briefly, the animal was sedated, intubated and under general anesthesia, a right thoracotomy was performed as surgical access to the heart. After full heparinization, cardiopulmonary bypass was established via the right carotid artery and the right internal jugular vein. While on cardiopulmonary bypass and with the heart beating, both cava were snared and the right atrium was opened to expose the tricuspid valve. A sonomicrometry crystal used for telemetry was sewn on the tricuspid valve, as illustrated in Fig. 14(a), and is continuously subjected to motion artifacts (movement of the valve in this specific case). The surgical procedure for implanting the crystal was performed on an adult male sheep, in a fully equipped and accredited animal facility at West Michigan Regional Laboratory which is a part of the Spectrum Health Delivery System.

![Example of multi-access waveforms based on Walsh-Hadamard coding](image1)

**Figure 10.** Example of multi-access waveforms based on Walsh-Hadamard coding, where Rx shows the received waveform as a result of simultaneous transmission from different transmitters.

![Bar plot showing the signal to noise ratios (SNR) and respective BER measured at the reader](image2)

**Figure 11.** Bar plot showing the signal to noise ratios (SNR) and respective BER measured at the reader, when the three transmitters are sending independent data streams of encoded data at a bitrate of 800 Kbps.
Figure 12. Frequency response of the signal received by Rx when each transmitter (a) Tx1, carrying a signal of 800 KHz which is modulated using 20 KHz (b) Tx2, carrying a signal of 800 KHz which is modulated using 30 KHz (c) Tx3, carrying a signal of 800 KHz which is modulated using 50 KHz, are trying to send their data stream independently.

Figure 13. Sample wave forms corresponding to each transmitter (Tx1, Tx2 and Tx3), in case of transmitting data (a) independently and (b) pairwise. (c),(d) Shows the spectrograms retrieved by the reader for the two FDMA transmissions shown in (a), (b) respectively.

located on the surface of the skin. The reader crystal was sutured underneath the surface of the skin and approximate distance to the implanted crystal is about 10 cm.

B. Experimental results

Fig. 15 shows the M-scan retrieved by the reader when transmit power is varied from 0 to 12 µW for data rates of 1Mbps and for a data sequence shown in Fig. 8 (c). The
Figure 14. Experimental biotelemetry setup using an adult sheep model: (a) illustration showing sonomicrometry crystals implanted in the tricuspid valve and sutured underneath the skin; (b) pictures taken during the surgery and crystal implantation (i)–(ii) and after surgery when the chest cavity has been closed (iii).

Figure 15. M-scan image corresponding to the ovine experiment where the crystal communicates (using a frequency modulation scheme) with the crystal located under-the-skin crystal and when the transmitted power is varied from 2 $\mu$W to 12 $\mu$W.

Figure 16. Measured bit error rate, for the data shown in Fig. 15, as the transmitted power is varied from 2 to 12 $\mu$W.

recorded M-scan images were used to reconstruct the data-stream and generate BER plots as shown in Fig. 16. As expected, the measured results show monotonic decrease in the error-rates with the increase in the transmission power. More importantly, BER rates of $10^{-2}$ can be achieved at power dissipation levels of only 11 $\mu$W which is well within the compliance limits.

V. DISCUSSION

Given that the maximum energy dissipated at data-rates of more than 1 Mbps was demonstrated to be well below the compliance limits, there is still room to push the limits of telemetry using M-scan imaging. The high-frequency sonomicrometry crystal could be used to support higher data-rates; however the attenuation in tissue will also increase. In this regard, an ultrasound reader with a sufficient receive gain and low noise-factor could be used to alleviate the effects due to attenuation. Another direction to increase the telemetry rate is by using channel equalization techniques [26] to compensate for overlap between neighboring transmission bits and also to compensate for effects due to multipath. The channel effect can be clearly seen in Fig. 17 where the bits '1' are aliased when transmitted in sequence as compared to the case when a sequence of '0's are interspersed in between. Like conventional channel equalization methods, a known preamble sequence would be used for estimating the instantaneous channel impulse response. This would then be used to demodulate the data bits. Note that in M-scan telemetry, the data (multi-access or single) is received as frames of images which can be post-processed using image equalization techniques (similar to channel equalization). Another possible solution to improve the data-rates is to reduce the echoes generated due to interrogation pulses. This can be achieved by lowering the transmit power at the Ultrasound reader and increasing the receiver gain to capture a higher resolution ultrasound image of the transmit signal. In terms of BER performance, like any
multi-access communication techniques the error-rates will increase compared to a point-to-point communication link [29]. However, as long as the modulation and coding basis functions are chosen to be orthogonal, the error-rates for different multi-access schemes are approximately the same. The high-channel capacity of the M-scan or B-scan telemetry link could be exploited in alternate ways. For instance, using forward error-correcting (FEC) techniques [27] one can establish reliable telemetry links with implants that are either located further away from the ultrasound imager or with devices implanted inside a material with low ultrasonic penetration - for example implants in tissue surrounded by bone [28]. In such cases the coding-gain is used to compensate for the signal attenuation and the entire M-scan image could be treated as a single block-code.

VI. CONCLUSION

The objective of this paper was to demonstrate that a standard ultrasound reader can be used to perform multi-access telemetry with devices implanted inside the body. Compared to an RF based telemetry system, ultrasound based telemetry can penetrate deeper into biological tissue and a millimeter scale sonomicrometry crystal was shown to be capable for high-speed data communications. Even at data rates of 1 Mbps and implantation depths greater than 10 cm with reasonable BER in the order of $10^{-2}$ which suits most of the in-vivo sensing applications [30], the measured power dissipation on sonomicrometry devices was shown to be well below the tissue heating limits. This implies that these devices could be powered using implanted energy sources or could be powered remotely. In [18] we showed that up to a few microwatts of energy can be harvested from the interrogation pulses generated by a standard ultrasonic reader. In the future, our goal will be to use the sonomicrometry crystal to harvest energy from the interrogation pulses to energize the sensing and telemetry functions. To summarize, this telemetry technique generates a M-scan containing the transmitted information and saves it as bit-map image and later can be demodulated using appropriate image processing algorithms. The use of COTS and medically compliant ultrasound readers for in-vivo telemetry will obviate the need to design dedicated ultrasound decoders and would simplify the adoption of the technology by practicing clinicians. Examples of such low-data-rate devices where the FEC technique could be applied include health-monitoring sensors for hip or knee-implants [28].

REFERENCES


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