Self-Reconfigurable Micro-Implants for Cross-Tissue Wireless and Batteryless Connectivity

Mohamed R. Abdelhamid, Ruicong Chen, Joonhyuk Cho, Anantha P. Chandrakasan, Fadel Adib
Massachusetts Institute of Technology
{mrhamid,raychen,joonhyuk,anantha,fadel}@mit.edu

ABSTRACT
We present the design, implementation, and evaluation of $\mu$medIC, a fully-integrated wireless and batteryless micro-implanted sensor. The sensor powers up by harvesting energy from RF signals and communicates at near-zero power via backscatter. In contrast to prior designs which cannot operate across various in-body environments, our sensor can self-reconfigure to adapt to different tissues and channel conditions. This adaptation is made possible by two key innovations: a reprogrammable antenna that can tune its energy harvesting resonance to surrounding tissues, and a backscatter rate adaptation protocol that closes the feedback loop by tracking circuit-level sensor hints.

We built our design on millimeter-sized integrated chips and flexible antenna substrates, and tested it in environments that span both in-vitro (fluids) and ex-vivo (tissues) conditions. Our evaluation demonstrates $\mu$medIC’s ability to tune its energy harvesting resonance by more than 200 MHz (i.e., adapt to different tissues) and to scale its bitrate by an order of magnitude up to 6Mbps, allowing it to support higher data rate applications (such as streaming low-res images) without sacrificing availability. This rate adaptation also allows $\mu$medIC to scale its energy consumption by an order of magnitude down to 350 nanoWatts. These capabilities pave way for a new generation of networked micro-implants that can adapt to complex and time-varying in-body environments.

CCS CONCEPTS
• Hardware → Full-custom circuits; • Computer systems organization → Sensor networks; • Applied computing → Life and medical sciences;

KEYWORDS
In-body IoT, Backscatter Communication, Wireless, Energy Harvesting, Batteryless, Reprogrammable

ACM Reference Format:

1 INTRODUCTION
The mobile networking community has recently witnessed mounting interest in wireless and batteryless sensors that can operate inside the human body [22, 33, 54, 63]. These sensors can power up by harvesting energy from RF (Radio Frequency) signals transmitted from outside the body, and they communicate at near-zero power via backscatter – i.e., by reflecting existing signals rather than transmitting their own carrier. The combination of energy harvesting and backscatter communication allows these sensors to be batteryless. Independence of batteries eliminates the need for surgical replacement, allows ultra-long term operation, and enables miniature, fully-integrated form factors [35]. As a result, such sensors could be used for continuous monitoring of biomarkers and tumors, ultra-long lasting drug delivery systems (e.g., for patients with Alzheimer’s or Osteoporosis), and closed-loop control systems with real-time feedback (e.g., artificial pancreas for Diabetes’ patients).

A key challenge that faces existing solutions for wireless and batteryless micro-implants lies in their rigid designs which cannot adapt to different tissues or to time-varying in-body conditions. This is particularly problematic for mobile sensors like ingestible capsules, which experience a variety of in-body environments as they travel through the digestive tract to deliver drugs or sense biomarkers. The ability to adapt to different in-body environments is also key to enabling these sensors to operate across different humans, whose bodies have different tissue compositions (fat, muscles, etc.). The
harvest energy across different tissues or maintain reliable backscatter communication inside tissues. The system introduces a network that allows it to scale to multiple sensors. Our evaluation in both in-vitro (fluids) and ex-vivo (tissues) conditions demonstrates the feasibility of our design.

We built a prototype of our design by fabricating it on an IC (shown in Fig. 1) and integrating it with a re-programmable antenna on a flexible substrate. The design also integrates a MAC protocol that allows it to scale to multiple sensors. Our evaluation in both in-vitro (fluids) and ex-vivo (tissues) conditions demonstrates the following results:

- μmedIC’s programmable resonance allows it to harvest energy across different types of tissues including fat, muscle, and multi-layer compositions with muscle, fat, and bone as well as different fluids. The resonance can be reconfigured by as much as 200 MHz inside tissues. In the absence of reconfigurability, the micro-implant’s ability to power up reduces to one or two tissues.
- μmedIC can support bitrates reliably up to 6 Mbps and as low as 625 kbps. Its rate adaptation can gracefully scale to different in-body conditions by incorporating feedback through sensor hints. In the absence of rate adaptation, the design becomes either limited to low availability or low throughput.

Contributions. We present the first batteryless micro-implanted system that is capable of self-reconfiguration for energy harvesting and backscatter communication inside tissues. The system introduces a reconfigurable architecture with programmable antennas, harvesting

---

1 In contrast, the MHz band around 400 MHz is used for larger battery-powered implants such as cardiac pacemakers.
Self-Reconfigurable Micro-Implants for Cross-Tissue
Wireless and Batteryless Connectivity

circuits, and backscatter throughput. The design also introduces a rate and resonance adaptation protocol for wireless micro-implants. We also present a prototype implementation on an integrated circuit on a flexible antenna substrate and evaluation in different tissues.

We note that μmedIC’s benefits extend beyond micro-implants that are entirely batteryless. For example, in higher data-rate applications (such as streaming images from endoscope capsules), today about half the energy is spent on RF transmissions [54, 61]. By enabling efficient and reconfigurable backscatter, μmedIC can significantly reduce the power consumption of such implants, allowing for battery-assisted implementations [44] that can function longer. As the technology evolves, it may also be integrated with recent proposals on battery-free cameras (which have been demonstrated outside the human body) [36]. Such designs are beyond the scope of this paper and are left for future work.

2 BACKGROUND

The past two decades have witnessed an increased interest in bringing wireless capabilities to implantable devices. Research in the early 2000’s focused on understanding the impact of RF signals on the human body [28, 43], and was propelled by the rise of body area networks [13]. The success of this research and technological agenda resulted in wide adoption of wireless communication in implantable medical devices such as implanted pacemakers, cardiac defibrillators, insulin pumps, and capsule endoscopes [20, 61]. These early systems were all battery-powered [13].

The success of this body of work has prompted researchers to extend the vision beyond wireless communication to in-body wireless power transfer [42, 59]. Power transfer can eliminate the need for batteries which would, in turn, allow implantable sensors to function longer (without surgical replacement) and can result in a significant reduction in their form factor (since batteries can occupy 50% or more of the sensor’s size [15]). These capabilities can significantly expand the potential use cases of in-body sensors to tumor monitoring, neural stimulation, and drug delivery [22, 33, 47, 54]. The promise of such sensors has prompted the US Office of Science and Technology to declare long-lasting wireless micro-implants as one of six grand challenges of the decade [46].

One of the major challenges that still faces in-body wireless applications is the low efficiency of implantable antennas [28, 49]. This low efficiency (around 1%) has been widely documented in literature on wireless communication with battery-powered medical implants [34, 37], and it becomes even more problematic for batteryless micro-implants that rely on harvested RF energy to power up [33, 54]. Recent advances in energy harvesting try to address this problem by resorting to resonant rectennas, where the antenna and the rectifier (energy harvester) are designed to resonate in order to maximize their harvesting efficiency [18, 24]. Such resonance, however, is significantly impacted by surrounding tissues; prior work has demonstrated that if tissue composition or depth changes, antennas can easily shift out of resonance, becoming inefficient [30, 32]. This is why the majority of existing in-body sensors still require batteries or remain limited to shallow depths where they can harvest enough energy to power up despite their low efficiency [53, 54]. Our work is motivated by this past literature on resonant rectennas and extends it to work across tissues by introducing reconfigurability to the design of wireless and batteryless micro-implants.

We also present a prototype implementation on an integrated circuit [53x676]As the technology evolves, it may also be integrated with recent proposals on battery-free cameras (which have been demonstrated outside the human body) [36]. Such designs are beyond the scope of this paper and are left for future work.

3 SYSTEM OVERVIEW

μmedIC is a fully-integrated wireless and batteryless sensor for micro-implants that operate in the UHF (Ultra-High Frequency) ISM Band (902-928 MHz). The sensor can be used to support a variety of in-body monitoring and sensing applications such as tracking biomarkers or long-term monitoring of internal vitals to allow for early intervention.

A μmedIC sensor powers up by harvesting energy from RF signals transmitted by a reader outside the body. The sensor decodes the reader’s downlink commands and transmits its own packets on the uplink to be decoded by the reader. The design extends to multiple sensors, each of which is uniquely addressable. In the presence of multiple sensors, the reader orchestrates medium access.

The overall architecture of a μmedIC sensor is shown in Fig. 3. The design consists of a system-on-chip (SoC) that supports energy harvesting, decoding, and backscatter communication. μmedIC’s SoC also incorporates a power management unit to support the various computing and communication tasks and an extensible interface that allows integrating the chip with external sensors. The 1 mm² chip is assembled on a flexible PCB with a custom printed antenna. μmedIC can self-reconfigure to adapt to different in-body environments. There are two key components of this self-reconfiguration: the first is a reprogrammable bi-loop antenna that can adapt to surrounding tissues (§4) and the second is a rate adaptation algorithm that tracks circuit-level sensor hints and adapts to channel conditions (§5). The antenna and rate reprogrammability can be orchestrated by the IC itself. The next sections describe these components in detail.

4 REPROGRAMMABLE IN-BODY RECTENNA

In this section, we describe the design of μmedIC’s reprogrammable rectenna and demonstrate how this design enables adapting to different tissues in order to ensure efficient energy harvesting across various in-body environments.

4.1 Resonant Rectenna Design

Before delving into μmedIC’s design, it is helpful to understand the challenges that face RF energy harvesting inside tissues and how prior designs address these challenges.

Harvesting RF energy inside human tissues is more challenging than harvesting in air for two main reasons. First, RF signals exponentially attenuate as they traverse human tissues [37, 41], while in air, their amplitude decays linearly with distance. This makes it difficult for an external reader to deliver sufficient energy to power up an...
energy harvesting micro-implant inside tissues.\(^2\) The second challenge facing in-body energy harvesting arises from the constrained form factor of micro-implants. Specifically, due to anatomical constraints, micro-implants have form factor requirements that vary between 2-3 cm [16] to sub-centimeter dimensions. The limited form factor makes it difficult to efficiently harvest energy since it constrains the dimensions of the micro-implant’s antenna with respect to the wavelength of the RF signal [37].\(^3\)

Because of the above challenges, state-of-the-art proposals for in-body energy harvesting fine-tune their designs to optimize the harvesting efficiency along two main dimensions:

- **Radiation efficiency in Bio-tissues:** This refers to the efficiency of antennas in transmitting and receiving RF signals within a specific frequency band of interest. Because of the limited antenna form factor and the conductive properties of human tissues, in-body antennas suffer from low radiation efficiencies (they are typically as low as 1\%) [37]. In order to minimize losses due to the surrounding tissue environments, antenna engineers typically simulate their designs in electromagnetic simulators which account for the impact of the dielectric properties of tissues. This allows them to fine-tune various design parameters (like shape, geometry, thickness of conductor) to achieve the highest possible efficiency given the limited form factor and simulated medium [12].

- **Resonance:** Aside from optimizing the radiation efficiency of micro-implant antennas, state-of-the-art designs also exploit resonance [25, 45]. Resonance is a well-known electrical property that boosts harvesting energy efficiency by minimizing losses. It can be achieved by electrically matching the antenna impedance to the input impedance of the rectifier (energy harvester).

In order to maintain small form factor and optimize energy harvesting performance, state-of-the-art designs employ electrically small inductive loops, optimize their designs, and match them to the rectifying circuits [26]. The resultant rectennas are most efficient when they operate at their resonance frequency, defined by:

\[
f = \frac{1}{2\pi \sqrt{L_A C_{rect}}}
\]  

where \(L_A\) is the inductance of the loop antenna and \(C_{rect}\) is the input capacitance of the rectifier.

### 4.2 The Impact of Tissues on Resonance

In our above discussion, we have maintained that it is possible to simulate in-body environments that reflect practical real-world conditions. While that is true in principle, it is very difficult to truly optimize the energy harvesting to reflect practical environments. This is because the human body consists of multi-layer tissues, each layer with a different depth. Moreover, the tissue composition changes across different individuals as well as different body parts or organs. This makes it infeasible to design a one-size-fits-all resonant rectenna that has high efficiency across different body parts (e.g., for mobile micro-implants), let alone for different humans.

The difficulty in adapting to complex tissues arises from differences in their relative permittivity \(\epsilon_r\) (which reflect differences in the dielectric). Permittivity varies across different tissues, and directly impacts the antenna radiation pattern as mentioned earlier. For example, while the permittivity of muscle tissues is \(\epsilon_m = 55 - 17.4j\), the permittivity of fat is \(\epsilon_f = 11 - 2j\), both around the same frequency of 900 MHz which corresponds to the ISM band of interest [19].

To better understand the impact of such tissue variations on energy harvesting, recall Fig. 2 from the introduction which demonstrated how a design that is optimized to achieve high efficiency in muscle tissue within the ISM band (910 MHz) becomes inefficient at the same frequency when placed in a different tissue (e.g., fat). Indeed, we empirically verify this behavior in real-world measurements in §8. This behavior demonstrates the difficulty in scaling prior rigid designs to complex tissue environments. Because of this lack of scalability, existing designs are limited to shallow depths, where the received energy remains sufficient to power them up despite their low harvesting efficiency [53, 54].\(^4\)

### 4.3 Reconfigurable Coupled Rectenna

To deal with the above challenges and scale to different in-body environments, \(\text{medIC}\) adopts a self-reconfigurable architecture that enables it to adapt to surrounding tissues. This reconfigurability is made possible by synergistically combining two sub-components: (1) the first is a reprogrammable antenna that can adapt its radiation efficiency to the surrounding medium, and (2) the second sub-component is reprogrammable matching circuit that enables shifting the resonance to ensure that the energy harvesting circuit and antenna remain matched. By tuning both the radiation efficiency and the resonance, \(\text{medIC}\)’s design can adapt the two core dimensions that are typically pre-tuned to achieve high harvesting efficiency in tissues as set by §4.1. The rest of this section describes these two sub-components in detail.

#### 4.3.1 Antenna Reprogrammability

To enable antenna reconfigurability, \(\text{medIC}\) employs a coupled antenna design. Recall that coupling refers to the interaction between two antennas when they are in close physical proximity [11], and is typically considered to be harmful. In our design, however, we exploit a coupled design in order to adapt \(\text{medIC}\)’s radiation efficiency inside tissues. Technically, our goal is to change the current distribution along the radiating element in order to counteract the change in the surrounding medium’s permittivity.

Fig. 4 shows \(\text{medIC}\)’s coupled design, which consists of two loops: an outer rectangular loop and an inner circular loop (with a

---

\(^2\)Specifically, state-of-the-art RF rectifiers need around -34.5dBm of power to power up [25]. This minimum threshold is determined by transistor electronics.

\(^3\)In the absence of such form factor requirements, achieving good radiation efficiency would require antennas whose dimensions are of the same order of the wavelength [11].

\(^4\)At shallow depths, the RF signals experience less overall (exponential) attenuation due to a shorter path length inside tissues.
Antenna design is an art and is known to require extensive iterations, would exhibit the same behavior observed in the plots shown to the left as the capacitance is changed. This shows that the antenna is most efficient around that frequency. For example, right-most green curve has a peak around 900 MHz, indicating that the antenna is most efficient around that frequency. However, as we change the load on the outer loop, that peak shifts, indicating that the frequency of highest radiation efficiency also shifts. This shows that μmedIC’s design indeed enables programming the radiation efficiency. Such programmability is highly desirable because if the antenna becomes most efficient around 1.1 GHz due to surrounding tissues, this allows us to shift the highest efficiency region back to the 900 MHz ISM band.

The right side of Fig. 5 shows a conceptual schematic of the resulting coupled design, which is a standard approach to reason about coupled engineering designs [50]. The schematic represents the inductive coupling between the two loops as an inductor, and represents the programmability via an on-chip variable capacitor. In practice, we implemented this programmability as a switched capacitor bank for simplicity and energy efficiency. This schematic would exhibit the same behavior observed in the plots shown to the left as the capacitance is changed.

Finally, we note that this programmable architecture was the result of many iterations in consultation with the relevant literature. One of the interesting designs we explored consisted of a single loop, where one side of the loop was connected to energy harvesting while the other was connected to a programmable load; such a design, however, had a much lower bandwidth than μmedIC’s coupled architecture which resulted in significantly lower data rates. Another interesting design element is the horizontal chord across the inner loop, which has also been recently demonstrated to achieve wider bandwidth and higher efficiency (albeit without reprogrammability) [25, 45]. Thus, because antenna design is a highly complex task, by introducing a simple design that enables reprogrammability, μmedIC can reduce the burden on pre-tuning all the parameters and allow for reconfigurability to adapt to different media.

4.3.2 Harvesting Reconfigurability

So far, we have focused only on improving the antenna’s efficiency by shifting its resonance frequency. However, if we just do that, then we cannot ensure it remains matched to the energy harvesting circuit. Thus, we won’t reap the benefit of resonance harvesting.

To optimize the energy harvesting efficiency, rectennas rely on the perfect resonance between an electrically small antenna with an inductive impedance profile with the conjugately matched capacitive input impedance profile of the rectifier circuit as shown in Fig. 6.

This can be expressed in the following equation:

$$Z_{A} = Z_{rect}$$  \hspace{1cm} (2)$$

where $Z_{A}$ is the antenna’s impedance while $Z_{rect}$ is the rectifier’s input impedance.

Now recall that in Fig. 5, the antenna programmability changes $Z_{A}$. So, we should change the rectifier accordingly to ensure resonance occurs at all programmable states.

To achieve this, we introduce programmable circuit matching. Since the programmable antenna is still an inductive one and the rectifier has a capacitive impedance, the required matching network can be implemented as a simple bank of capacitors. This capacitor bank can be used to compensate for the change in inductance from one state to the other, thus maintaining perfect conjugate matching. Effectively, by reprogramming both the antenna and the harvesting circuit, we have disposed of the need to pre-fine-tune the design and allowed ourselves to retune it inside the human body.

One might wonder whether modifying the antenna alone or the energy harvesting alone might be enough. Implementing any of these approaches alone will lead to less desirable performance. Changing the matching alone can allow a slight shift in the center frequency (e.g., a slight shift in the channel) [51], but if the desired shift is too large, the antenna becomes inefficient. On the other hand, reprogramming the antenna alone would not be enough because it would significantly reduce the efficiency without resonance. It is by

---

5Indeed, coupled designs are known to have larger bandwidth, and hence have been used in the design of wideband RFIDs and UWB front-ends [3, 17]. In contrast to prior coupled designs which are rigid, μmedIC’s design allows for reconfigurability.
combining both of these techniques together that µmedIC can enable cross-tissue wireless and batteryless connectivity for micro-implants.

5 RATE ADAPTATION FOR IN-BODY BACKSCATTER

So far, our discussion has focused on µmedIC’s energy harvesting reconfigurability. Next, we discuss how it can adapt its bitrate to deal with varying channel conditions. Specifically, recall that µmedIC communicates on the uplink via backscatter. Moreover, it needs to support a variety of applications, some of which may require bitrates up to few Mbps (e.g., capsule endoscopes) [15]. Thus, it is desirable to enable the sensor to backscatter at the highest possible bitrate needed by the application of interest whenever possible. Below, we discuss the need for in-body backscatter bitrate adaptation and how µmedIC’s reconfigurable design can address these needs.

5.1 The Need for Adaptation

Bitrate adaptation is today a core component of a variety of wireless network protocols, including WiFi and LTE. It refers to the ability of certain communication devices to adapt their throughput to the channel conditions. For example, if the channel is strong (i.e., has high SNR), the communication link can sustain higher throughputs and the transmitter should use a higher bitrate since the channel capacity is higher. On the other hand, if the channel is weak (i.e., low SNR), the transmitter should transmit at a lower bitrate.

Bringing such bitrate adaptation to in-body micro-implants is desirable for two main reasons. First, as the person moves (or as the micro-implant moves inside the body), the wireless channel changes and the bitrate must adapt to it. A second, and equally important, reason arises from the relationship between power consumption and backscatter bitrate. Specifically, higher bitrates consume more energy because the oscillator needs to be driven at a higher frequency. (The overall power consumption is directly proportional to bitrate according to $P = fCV^2$ where $C$ is the capacitance and $V$ is the voltage.) Thus, the bitrate also needs to be adapted to the harvested energy to ensure that the micro-implant does not consume all its energy and die off.

To gain more insight into the trade-off between bitrate and power consumption, we ran experimental trials with µmedIC (implementation detailed in §6). In each experimental trial, we pre-programmed the micro-implant to backscatter at a fixed bitrate. We repeated the experiment at three different bitrates (600kbps, 3Mbps, 6Mbps) and two different transmitted power levels (14dBm and 15dBm). We specifically selected two power levels that are close to demonstrate µmedIC’s reconfigurability in energy harvesting. To close the loop on bitrate adaptation, µmedIC monitors circuit-turn off threshold, and oscillator’s frequency to adapt to the available energy.

To gain more insight into the trade-off between bitrate and power consumption, we ran experimental trials with µmedIC (implementation detailed in §6). In each experimental trial, we pre-programmed the micro-implant to backscatter at a fixed bitrate. We repeated the experiment at three different bitrates (600kbps, 3Mbps, 6Mbps) and two different transmitted power levels (14dBm and 15dBm). We specifically selected two power levels that are close to demonstrate µmedIC’s reconfigurability in energy harvesting.

Next, we describe how this can be realized in practice. µmedIC’s backscatter switch is driven by a digitally controlled oscillator. The oscillator’s frequency can be in one of eight different states (i.e., 3 control bits). By employing a current-starved ring oscillator architecture, we can digitally tune its frequency by injecting more current to the core ring so that a lower frequency can be attained. The rectenna tuning logic senses the DC voltage and incrementally increases or decreases the oscillator frequency. This frequency is used to modulate and encode the sensor bits and backscatter to the reader. Alg. 5.1 summarizes this algorithm, where $V_{DC}$, $V_{TH}$, and $D_{fosc}$ represent the harvested voltage, turn-on threshold, and oscillator’s frequency configuration respectively.

Few additional points are worth noting:

• So far, we have described µmedIC’s bitrate adaptation algorithm. The same idea can be extended to resonance adaptation as shown in Fig. 8. Specifically, by sensing the harvested voltage using a sampler (ADC), the node can apply stochastic gradient descent to move to a more efficient harvesting state by updating the matching circuit and resonance capacitance. It is worth noting, however, that unlike rate adaptation, resonance and matching adaptation

\[ \text{Note that this bitrate frequency of the backscatter oscillator is different from the RF frequency of transmission.} \]
require the off-chip storage super-capacitor to have some residual energy in the first place. This “warm-start” approach is a standard assumption in the design of energy harvesters [45]. Extending the design to cold-start is beyond the scope of this paper.

• There are three types of reconfiguration parameters (matching capacitor bank, resonance capacitor bank, and oscillator frequency). Each of these parameters can be reconfigured independently. The two capacitor banks can be adapted on a packet-by-packet basis.

In the simplest implementation, we alternate between incrementing/decrementing each of the two capacitor banks. Looking ahead, it would be interesting to explore more optimal algorithms that can leverage μmedIC’s reconfigurable design to achieve higher throughput than our simple adaptation algorithms.

• Our discussion has focused on rate and resonance adaptation. However, μmedIC’s design extends to decoding packets on the downlink (via PIE encoding) and encoding backscattered packets on the uplink via FM0 encoding similar to RFID communication. The ability to decode downlink packet enables the reader to employ a master-slave Medium Access Control (MAC) protocol and extend to multiple micro-implanted sensors concurrently.

• In our evaluation, we noticed that the backscatter bitrate varied over time, even when the oscillator was expected to transmit at a fixed rate. This was caused by the susceptibility of the node’s ring oscillator to environmental variables. To deal with this issue and decode correctly, our reader continuously tracks the backscatter frequency and adapts to its variations. Practically, this was implemented via a bit-by-bit maximum likelihood decoder that corrects phase/frequency errors incrementally from one bit to the other. In sum, the rate adaptation operates on a packet-by-packet basis where the chip senses the voltage and increments or decrements the bitrate accordingly. On the receive side, the reader uses the preamble of the backscattered packet in order to determine the backscatter rate and correctly decode the sensor’s data.

6 IC DESIGN & ANTENNA FABRICATION

ASIC Implementation. μmedIC is implemented as an application-specific integrated circuit (ASIC), whose prototype is shown in Fig. 1. The integrated circuit was designed and fabricated in a 65nm low-power RF complementary metal-oxide-semiconductor (CMOS) technology. The ASIC incorporates the entire system-on-a-chip, and
which generates the clock signal for an FM0 encoder and a data cancellation technique to generate a constant voltage that while our final evaluation was performed with flexible substrate, allows for external configurability and testing each state independently. Moreover, an Agilent N9020A spectrum analyzer [1], hooked directly to an antenna, was used to initially characterize the backscatter modulation depth at different rates and center frequencies.

We evaluated µmedIC across five kinds of in-body environments including: minced meat, fatty tissues, mixed tissues with bone, saline water, and oil-based fluids. Since the majority of human tissues are oil-based or water-based, this allows us to cover a large variety of potential environments and multi-layer tissues. In each of these environments, we either submerged the sensor entirely in the fluid or covered the antenna with meat tissues as in Fig. 10. In addition to the different tissues, we evaluated our setup at different transmit powers, orientations, as well as distances ranging from 10cm (between the reader and the tissue with µmedIC) to low power levels to almost a meter at a transmit power of 20dBm. The pins of the IC are connected via a 1.27mm pitch bus to a Tektronix MSO3054 oscilloscope and to an FPGA to allow for testing and voltage measurements. Note that our design allows bypassing the self-reconfigurability to allow for external configurability and testing each state independently. Moreover, an Agilent N9020A spectrum analyzer [1], hooked directly to an antenna, was used to initially characterize the backscatter modulation depth at different rates and center frequencies.

Subsequently, our complete end-to-end evaluation was performed using a software radio setup which employed two USRP N210 [10] boards for collecting and post-processing a large number of packets and estimating the BER under different channel conditions. One USRP board (with an SBX daughterboard [9]) served as the transmitter for the micro-implant, powering the chip up and providing downlink commands. The other USRP (with LFRX daughterboard [7]) worked as the reader, capturing the backscatter signal at a sampling rate of 25MSps. Both USRP boards were synced together using the same reference clock of 10MHz of a CDA-2990 Octoclock 8-channel distribution module [4]. Moreover, their front-ends were
connected to log-periodic antennas [8] and their backends were connected via ethernet cables to allow for a central Linux machine to control both of them simultaneously. The data was then collected and post-processed using MATLAB to characterize the bit error rate, SNR, throughput and availability in the different tissues under test.

8 PERFORMANCE RESULTS

In this section, we report the evaluation results of µmedIC’s performance in real-world environments and various in-tissue conditions.

8.1 Energy Harvesting

First, we evaluated µmedIC’s ability to harvest energy across a variety of in-tissue environments. To do this, we used the Keysight N5183 MXG signal generator [6] as an RF source with fixed transmitted power. The signal generator was connected to a wideband log-periodic antenna covering the range of interest. We swept the transmitted frequency and recorded the signal received using an oscilloscope hooked up to the storage capacitor pin of µmedIC.

Fig. 11(a) shows the harvested voltage of a rigid design, i.e., when the tuning capacitor (of the antenna) and the matching capacitor are fixed. The figure plots the harvested voltage as a function of the frequency of the signal transmitted by the reader. We repeated the same measurement across the five tissue environments described in §7 with testing depths of around 10cm or less. The figure also plots a solid black line around 0.65V, which indicates the minimum voltage required to turn on the low-dropout voltage regulator (LDO).

The figure shows that for this fixed configuration, the micro-implant can be powered up in the ISM band only when it is placed in the oil-based tissue (red curve), where the harvested voltage is above the minimum threshold. For all other tissues, the peak is either shifted outside the ISM band (shaded grey region), or the rigid design is completely unable to power up due to low efficiency.

Next, we were interested in assessing whether µmedIC’s reconfigurability allows it to power up in the ISM band. So, for each tissue environment, we looped through all 4096 configurations (changing the tuning and matching capacitors). For each tissue, we chose the configuration that has the highest harvesting voltage in the UHF ISM band. Fig. 11(b) plots the resulting curves. The figure shows that across all tissues, µmedIC can harvest enough energy inside the UHF ISM band to power up (the harvested voltage is at or above the threshold). It also shows that the optimal configuration for energy harvesting is different across tissues. For example, optimal powering up in saline fluid can be achieved at with a matching capacitor $C_m = 0.5 \mu F$ and a tuning capacitor $C_t = 0.5 \mu F$; however, optimal powering in mixed tissue requires $C_m = 0.75 \mu F$ and $C_t = 0 \mu F$. This demonstrates that both the antenna programmability and matching programmability are necessary to adapt to different tissues.

Next, we were interested in understanding the extent of reconfigurability of µmedIC’s harvesting efficiency. This time, we fixed the tissue to an oil-based medium, and we looped again through all the configurations. For each frequency between 750 MHz and 1050 MHz, we chose the configuration that yielded the highest harvested voltage. Fig. 11(c) plots the envelope of these peaks across frequencies (in purple) as well as a sample subset of configurations. The figure shows that the peak indeed moves across different configurations by up to 200 MHz. Such range of frequency shift is desirable as it allows us to also exploit other bands. For example, the CCC band in China may also be used for UHF communication and is also shaded in the figure. Thus, µmedIC’s reconfigurability would also allow adapting to regulating bodies across countries.

Finally, we were interested in investigating whether deforming the antenna impacts is resonance (e.g., rolling it as a pill). Our experimental evaluation verified that such deformation may indeed shift the energy harvesting peak by up to 60 MHz. Moreover, similar to our earlier demonstration, µmedIC’s reconfigurability enables shifting the resonance back to within the ISM band.

8.2 Rate adaptation

Next, we wanted to assess the second dimension of reconfigurability in µmedIC’s design, namely its bitrate adaptation. Recall that bitrate adaptation is necessary to allow adapting to different channel conditions as well as harvested power levels, as highlighted in §5.

(a) Power Consumption vs Throughput. First, we were interested in measuring µmedIC’s power consumption as well as the impact of throughput on it. To perform this evaluation, we connected the IC to an external power supply of 0.5V rather than to its internal LDO and we measured the current drawn using the Keithly 2400...
We computed the BER for each bitrate and transmit power as the average BER following the expected theoretical trend. We ran 3 experimental trials, and we experimented with 10 bitrates in total (50kbps, 100kbps, 250kbps, 500kbps, 1Mbps, 2Mbps, 2.5Mbps, 3Mbps, 4Mbps, 5Mbps) and three transmit power levels to correspond to different in-tissue conditions (19dBm, 16dBm, 14dBm). The figure shows that as the bitrate increases, the BER decreases. This is because higher bitrate requires larger bandwidth, thus increasing the received noise. This figure also shows that if the transmit power increases, the BER decreases; this is also expected because a higher transmit power results in higher backscatter (reflected) SNR, resulting in better decoding ability. This is typically assessed by plotting the BER-SNR curve, which is shown as a heatmap where blue indicates the absence of the corresponding frequency. And finally, the bottom row shows the instantaneous bitrate inferred from the spectrogram. Figs. 15(a), (b), and (c) represent the low-bitrate, high-bitrate, and adaptive bitrate schemes respectively. We make the following remarks:

- For the low-bitrate scheme, throughput remains constant at around 625kbps without taking advantage of higher available power at better channels. This can be seen both from the spectrogram and the instantaneous bitrate. The oscillator rarely dies in this scheme due to low power consumption and conservative nature.
- When the node fixes its rate to the high mode regardless of the available power, it suffers from frequent power downs during transmission as illustrated in the spectrogram of Fig. 15(b).
- Finally, the adaptive scheme shown in Fig. 15(c) can smoothly adapt its bitrate from one packet to the other according to the received power variations.

To gain more insight into the tradeoff between availability and throughput, we collected measurements across three schemes, similar to the ones described above: a fixed low rate of 750kbps, a fixed high rate of 12Mbps, and an adaptive rate scheme capable of varying between the two. We ran around 30 experimental trials in total covering low-power, high-power, and varying-power environmental setups. In order to ensure the nodes experienced a time-varying channel in the latter, we programmed our downlink transmitter to quickly vary its transmit power by a factor of 2 each 20 ms. In each experimental trial, we estimated the node failure probability as our availability metric. The empirical probability was computed as the ratio of period of time the node fails (due to running out of power) to the full duration of the transmission.
Fig. 16 shows the results of this experiment by plotting the node failure probability across the different schemes and setups, where Fig. 16(a), (b), and (c) represent the low-power, high-power, and varying-power setups respectively. The bar graphs correspond to the median failure probability and the error bars show the standard deviation. We make the following remarks:

• $\mu$medIC’s low-rate scheme provides the highest availability. Specifically, even under the lowest-power mode, its failure probability remains below 0.06 (i.e., 6%), and it drops further to 0.0055 at higher power channels. This is expected since the low-rate scheme consumes the lowest power (as per Fig. 12).

• The high-rate scheme suffers from a node failure probability of 0.2 in the low-power regime. This demonstrates that high-rate is likely to be unsuitable for applications that require always-on availability for in-body sensing (and communication).

• In the high-power setup (Fig. 16(b)), almost all modes have high availability, with even the high-rate mode having a failure probability lower than 0.0003. This suggests that in high-power settings, there is little to be gained from using the low-rate scheme.

• Finally, $\mu$medIC’s adaptive scheme offers a middle ground between low-rate and high-rate. In principle, it is possible to make the adaptive rate more conservative or aggressive by adjusting the adaptation thresholds, which would result in performance that is closer to the low-rate or high-rate schemes respectively.

So far, we have demonstrated that $\mu$medIC’s ability to choose and/or adapt its bitrate enables it to adapt for availability. Next, we would like to understand how these different schemes compare in terms of effective throughput. To do this, we used the same experimental trials described above and performed trace-driven simulations to measure effective bitrate. We also compared different packet sizes to understand the impact of packet size on throughput. The effective bitrate is computed as the number of total packets successfully transmitted (before failing) multiplied by the number of payload bits per packet and divided by the entire transmit duration. Note that if a node fails in the middle of a packet transmission, we consider the entire packet to be lost, thus emulating practical flows.

Fig. 17 plots the effective bitrate for each chip configuration across the different schemes and setups, where Fig. 17(a), (b), and (c) represent the low-power, high-power, and varying-power setups respectively. The figure also shows the effective bitrate for different packet sizes, 100-bit packet (shown in blue), 1000-bit packet (shown in orange), and 5000-bit packet (shown in yellow). The bar plot represents the median rate while the error bar represents the standard deviation. We make the following remarks:

• The high-rate scheme provides the highest throughput (effective bitrate) across almost all setups, even the low-power setup. In particular, for the 100-bit packet size, it achieves bitrates between 8.5Mbps and 11.5Mbps across the different setups. While this may be counter-intuitive (given that the high-rate has the lowest availability as per Fig. 16), it is able to transmit more bits within the same duration of time due to its higher bitrate. This shows that if the goal is to achieve the highest throughput (rather than the highest availability), then it is more desirable to configure $\mu$medIC to its high-rate mode.

• The 100-bit packet size (blue) outperforms all other schemes, despite that it suffers from more overhead since it has the same header size as the other schemes but smaller payload. This is because the chip is prone to suffer from node failure; thus, it is more desirable to get a full packet across than it is to aim for transmitting a longer packet with less overhead.

• Interestingly, the adaptive rate mode performs the best for largest packet size (5000-bits) in time-varying channel, i.e., in Fig. 17(c). This is because it can transmit at higher bitrates than the low-rate and does not fail as frequently as the high-rate configuration.

These results demonstrate that $\mu$medIC’s ability to switch between different configurations (whether fixed or adaptive bitrate) allows reconfiguring it to the application requirements. For instance, a streaming application may require continuous transmission, and thus favor availability over throughput (i.e., low-rate). Other applications may require higher throughput (e.g., taking one snapshot image and transmitting it quickly, with little buffering), and such applications may choose either high rate or aggressive rate adaptation. More generally, these results demonstrate the value of $\mu$medIC’s programmable throughput to serve different application requirements.
work has also explored mechanisms to improve signal-to-noise ratio (SNR) of in-body backscatter [60] but also ignored the impact of µ(b) Deep-Tissue RF. μmedIC’s design also builds on recent work on in-body RF backscatter. Recent designs have demonstrated RF backscatter in shallow tissues [22, 63] as well as the potential of operating deeper inside the body [33, 54]. Such designs, however need to isolate the antenna from nearby tissues by isolating the implant inside a test-tube or by forgoing energy harvesting [33, 63]. Prior work has also explored mechanisms to improve signal-to-noise ratio (SNR) of in-body backscatter [60] but also ignored the impact of surrounding tissues, which resulted in non-FCC compliant behavior. Our work directly advances this line of work and introduces resonance reconfigurability to enable embedding batteryless backscatter micro-implants directly in tissues.

(c) Backscatter Communication. Furthermore, μmedIC builds on a large and growing literature in backscattering different technologies, such as WiFi, TV signals, and Bluetooth [22, 23, 31, 62]. Our contributions are orthogonal and can be combined with these past proposals if one wishes to operate the micro-implants at their other frequency bands (albeit operating at higher frequencies would introduce more attenuation [40]). Prior work has also explored bitrate adaptation for backscatter, primarily in the context of RFID’s [39, 56], but also for ambient backscatter [55]. μmedIC’s rate adaptation is similarly motivated by the desire to reduce overhead, and it enables more efficient rate adaptation by tracking sensor hints in the IC itself.

(d) Non-RF In-Body Wireless. μmedIC is related to a large body of literature on in-body powering and communication using other approaches such as ultrasound [14], near-field [48], and midfield [21]. These approaches require either direct or near-direct contact with the body, and, as a result, have different application domains than RF-based systems like ours which can operate from larger distances outside the body [33, 63]. Thus, longer-range RF-based systems like μmedIC would result in more user-friendly implementations and pave way for easier remote and/or mobile healthcare solutions.

9 RELATED WORK

(a) Antenna Reconfigurability. Antenna reconfigurability can refer to a variety of techniques such as beamforming [38], polarization change [58], and frequency tuning [29]. μmedIC’s design is most related to past work on frequency reconfiguration that aims to reuse the same antenna front-end for different frequency bands (e.g., LTE vs 5G). Operating inside the body, however, introduces at least two new unique challenges: first, in contrast to air – which is a homogeneous medium from an RF perspective – the human body is neither homogeneous nor predictable as it consists of different layers of tissues which vary across individuals and across body parts. Second, μmedIC’s design not only needs to shift the antenna itself, but also the entire resonant structure (i.e., antenna+rectifier); which introduces additional complexity to the reconfiguration problem if one wishes to maintain high efficiency as we explained in §4. μmedIC’s architecture allows it to overcome these challenges while maintaining a near-zero power budget and operating inside tissues.

(b) Deep-Tissue RF. μmedIC’s design also builds on recent work on in-body RF backscatter. Recent designs have demonstrated RF backscatter in shallow tissues [22, 63] as well as the potential of operating deeper inside the body [33, 54]. Such designs, however need to isolate the antenna from nearby tissues by isolating the implant inside a test-tube or by forgoing energy harvesting [33, 54]. Prior work has also explored mechanisms to improve signal-to-noise ratio (SNR) of in-body backscatter [60] but also ignored the impact of surrounding tissues, which resulted in non-FCC compliant behavior. Our work directly advances this line of work and introduces resonance reconfigurability to enable embedding batteryless backscatter micro-implants directly in tissues.

(c) Backscatter Communication. Furthermore, μmedIC builds on a large and growing literature in backscattering different technologies, such as WiFi, TV signals, and Bluetooth [22, 23, 31, 62]. Our

5Specifically, in-air antennas can assume 50 Ohm matching, which would not be desirable for high harvesting efficiency [52].

10 CONCLUSION

This paper presents μmedIC, a self-reconfiguring fully-integrated wireless platform for batteryless in-body sensors. Our platform employs programmable structures in its antenna, harvesting circuits, and logic. The design is implemented on an IC and tested across tissues to demonstrate its ability to adapt its energy harvesting and backscatter throughput. Looking ahead, this technology paves way for a new generation for networked micro-implants capable of adapting to complex and time-varying in-body conditions.

Acknowledgments. We thank Mohamed Ibrahim for helpful discussions on antenna design and measurements. We also thank the Signal Kinetics group and the anonymous MobiCom reviewers and shepherd for their helpful feedback on the manuscript. This research is supported by an NSF CAREER Award, the MIT Media Lab, and the Endowed Fellowship of the Arab Republic of Egypt.


Mohamed R. Abdelhamid, Ruicong Chen, Joonhyuk Cho,
Anantha P. Chandrakasan, Fadel Adib


