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SYSTEM DEMONSTRATION AND CHARACTERIZATION OF A SELF-BIASED MAGNETOELECTRIC WIRELESS POWER TRANSFER SYSTEM FOR BIOMEDICAL IMPLANTS

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ABSTRACT
Magnetoelectric (ME) wireless power transfer systems (WPTS) offer a way to power small biomedical implants. We design a small (12 x 5 x 1.5 mm^3) self-biased ME WPTS to power a small, low-powered temperature sensor. At 80 μT, the ME WPTS can deliver 221.5 μW to a temperature sensor, allowing it to take a temperature reading and transmit its data via Bluetooth radio. We define a figure of merit, F_{ME}, to compare ME WPTS designed to power biomedical implants against one another. We find that our ME WPTS outperforms all but one of the ME WPTS designed for biomedical implants. However, we believe our self-biased ME WPTS is more viable for use inside the human body because it does not require permanent magnets to be implanted along with the ME receiver.

Keyword
Magnetoelectric, wireless power transfer, biomedical implants.

INTRODUCTION
The development of new biomedical implants is a growing field in which devices can be constructed to diagnose or treat a wide variety of alignments, from heart monitors to intrathecal pain pumps. However, the size of the biomedical implant is often constrained by the need to have a relatively large battery because it is impractical for a patient to undergo multiple surgeries just to have the physician change out the device's battery. Wirelessly recharging, or directly powering, the biomedical implant offers a solution that would allow for the miniaturization of the implant without reducing its functionality.

Miniaturizing implantable biomedical sensors enables the physician to insert a biomedical sensor via noninvasive surgery and increases the regions inside the body that are potentially compatible with the device. However, miniaturizing the device could potentially negatively impact the wireless power transfer (WPT) efficiency. A common WPT method for biomedical implants is using resonant inductive coupling between a transmit and receive coil. But, as the receive coil becomes smaller, the miniaturization affects the inductive wireless power transfer, whose efficiency dramatically decreases as the receiver becomes smaller than 1/10 of the electromagnetic wavelength [1].

An alternative method of WPT that may not be as affected by receiver miniaturization is ME WPT. ME WPTS are typically composed of a transmit coil and a ME transducer receiver composed of a magnetostrictive material laminated to a piezoelectric material. An AC current in the transmit coil induces a magnetic field (B-field) at the ME receiver. The ME WPTS first transforms the magnetic energy into mechanical vibrations through the interaction of the magnetostrictive material and the applied B-field. The vibrational kinetic energy is then converted to electrical form by the piezoelectric material. According to [1], for a given frequency, the characteristic wavelength of a ME antenna could be five orders of magnitude shorter than the electromagnetic wavelength, thus leading to more efficient energy transfer for small devices.

There are several characteristics for the design of an ideal ME WPTS-powered implantable biomedical sensor which we believe are critical for developing a useful prototype. First, the biomedical sensor's volume should be small, enabling it to be easily inserted into the body. Additionally, the biomedical implant should operate at low power since space for energy storage will be at a premium. And, the longer the implant can operate without a recharge, the more valuable it can be. Furthermore, the implant needs efficient power conditioning circuity since it will be recharged through the ME WPTS. Lastly, transmitting the sensor data wirelessly is essential for the implant to be useful.

This paper describes the overall system design of the ME WPTS and a low-powered biomedical temperature sensor. The ME WPTS system is characterized, and each step's power loss and efficiency are determined. The ME WPTS is then used to power a biomedical implant composed of an MCU, temperature sensor, and Bluetooth radio. This work is compared with other state-of-the-art ME WPTS designed explicitly for biomedical applications. We see that our ME WPTS, while having the smallest volume generally, outperforms all but one previously published ME WPTS designed for biomedical implants.

SYSTEM DESIGN AND EXPERIMENTAL VALIDATION

ME Reciever Design
ME transducer receivers for optimal wireless power transfer usually require a large external DC bias field. This DC bias field is generally accomplished through permanent magnets attached near, or on, the ME transducer. However, for implantable biomedical applications, installing large permanent magnets poses a health risk for the patient. Therefore, a self-biased ME transducer was built, shown in Figure 1, that mitigates the need for an external bias field [2]. Our self-biased ME transducer receiver uses two different magnetostrictive materials (nickel and metglas) with different permeabilities and coercivities fields, causing differences in saturation magnetization. This difference in saturation magnetizations causes an internal magnetostatic field transverse to the layers. The internal magnetostatic field

mitigates the need for an external DC magnetic bias field for the WPTS to operate at its peak efficiency [3]–[5].

The self-biased ME receiver is a 5-layer device composed of two layers of Metglas 2605SA1 (FeSiB), two layers of nickel, and a single layer of PZT 5A. The dimensions of each layer are given in Table *1*.

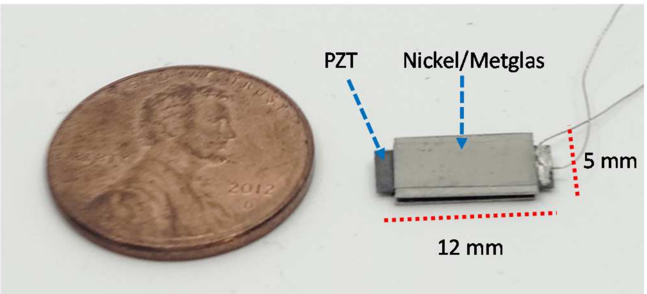

Figure 1 The self-biased ME transducer receiver is a 12 x 5 x 1.05 mm^3 composite beam composed of PZT, nickel, and Metglas.

Table 1 The self-biased ME transducer receiver

Materials	Length, mm	Width, mm	Thickness, mm
PZT-5A	12.36	3.81	0.5
Nickel	10.16	5	0.25
Metglas	10.16	3.81	0.023

Power Electronics and Low Powered MCU Design

Because of the potential for low coupling and hence low voltage produced in the ME WPTS, a voltage doubling (VD) rectifier was built. The VD rectifier consists of two Infineon Technologies BAT15-03W Schottky diodes with a typical forward voltage drop of 0.25 V and two 10 nF capacitors. These diodes were chosen because of their low voltage drop and small self-capacitance, 0.4 pF.

The rectified voltage is passed into an EPEAS AEM30940 energy harvesting chip. The EPEAS AEM30940 boosts or bucks the voltage to the appropriate level (3.3 V) for the MCU. The EPEAS chip was chosen because of its low minimum DC voltage input, 50 mV, and its low cold-start power requirement, 3 µW. In addition to boosting the voltage, the EPEAS chip also actively manages the power output to the MCU. If the MCU is consuming less power than is inputted via the ME receiver, the excess power is stored in a capacitor. If the MCU requires more power than is being inputted, then the EPEAS chip draws the necessary current from the storage capacitor. The MCU is an ATmega328P modified to operate in a low-power state. When the MCU is not sampling or transmitting data, all functionally of the chip is shut down except for the clock circuit. The temperature sensor is an Analog Device TMP36. The MCU transmits the sensor data via a low-powered NRF24L01 2.4 GHz radio. The system setup is shown below in Figure 2.

Transmit Coil System Description

The transmit coil is a 7.78 cm radius, nine turn, flat solenoid coil composed of 16 AWG (1.219 mm diameter) copper wire shown in Figure 3. An E&I 240L class A linear power amplifier was used to supply the transmitter coil with a high–frequency current. At 5 cm, the maximum B-field for this WPTS is 50 µT. At 2 cm, the maximum B-field for the WPTS is 80 µT. The maximum B-field is constrained by the maximum power output of our RF power amplifier.

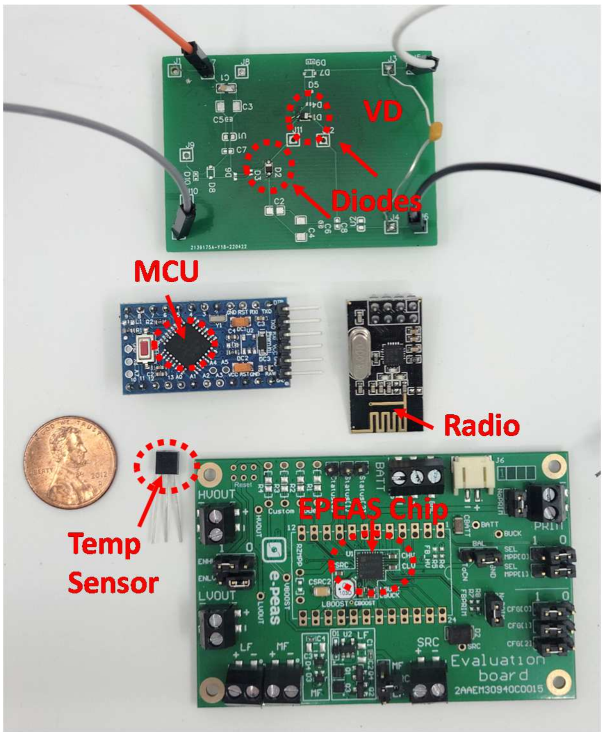

Figure 2 The power conditioning electronics and MCU, temperature sensor, and Bluetooth radio. The large printed circuit boards (PCBs), evaluation boards, and jumper cables are used for ease in the prototyping process.

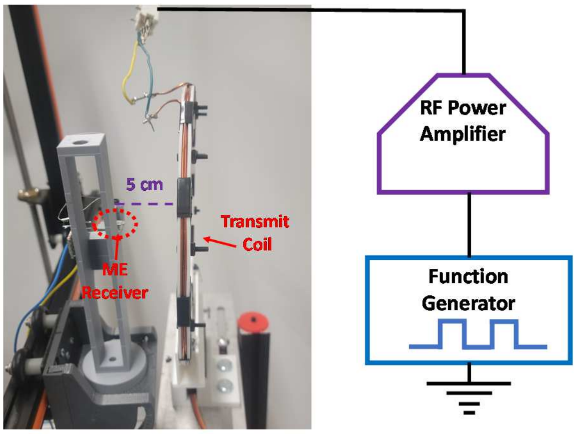

Figure 3 The setup of the ME WPTS and the 7.78 cm radius, nine turn, flat solenoid transmit coil. The function generator inputs a signal to the RF power amplifier, which excites high-frequency current in the transmit coil creating a B-field at the ME receiver.

Power Requirements and System Power Losses

Once turned on, the MCU requires 66 µW of power in its low power state. Reading the temperature via the TMP36 requires only 1.5 µW of power. The largest power requirement for the wireless biomedical sensor is transmitting the data via the low-power Bluetooth radio, which requires 99 µW.

The system power losses for the ME receiver when exposed to a 50 µT B-field are shown in Table 2.

At this B-field level, there is a significant loss of power in both the EPEAS chip and the VD rectifier. Most of the VD rectifier's power losses can be accounted for by the forward voltage drop of the two diodes. The EPEAS chip boosting the voltage to 3.3 V is the least efficient step, and the overall system efficiency is 38%. With the low input power of the ME receiver at 50 μT, the EPEAS chip is much more efficient, boosting to 1.8V (at this voltage level, the step efficiency to boost to 1.8V was 95%). However, the Bluetooth radio transmitter needs 3.3 V to operate. Note, the optimal load for the EPEAS was found to be the minimum resistance that the EPEAS chip could maintain a designated constant voltage output (e.g., below 310 kΩ, the voltage across the resistor would not be able to be held at 3.3 V, but would periodically drop below the set output level).

Table 2 Stepwise system power losses for ME WPTS at 50 uT

	Optimal Load (kΩ)	Voltage (V_{RMS})	Power (μW)	Step Efficiency	Overall Efficiency
No rectifier	4.4	0.637	92.13	X	X
VD rectifier	35	1.48	62.6	0.68	X
VD rectifier with EPEAS	310	3.3	35.1	0.56	0.38

Increasing the B-field to 80 μT improves the efficiency of the ME WPTS, as shown in Table 3.

Table 3 Stepwise system power losses for ME WPTS at 80 uT

	Optimal Load (kΩ)	Voltage (V - RMS)	Power (μW)	Step Efficiency	Overall Efficiency
No rectifier	4.4	1.24	349	X	X
VD rectifier	35	3.1	275	0.79	X
VD rectifier with EPEAS	52.3	3.3	208	0.76	0.60

With the higher B-field and subsequent higher voltage across the ME transducer, both the VD rectifier and the EPEAS chip efficiencies increase. Again, most of the power loss in the VD rectifier can be attributed to the forward voltage drop of the diodes. With the higher WPTS efficiency at this B-field, the ME transducer can directly power the biomedical sensor.

Full System Implementation and Experimental Validation

To validate the system power requirements, the full ME WPTS powered system was demonstrated and the power generation was characterized. The ME receiver was placed 2 cm from the cross-sectional plane of the transmit coil. A simplified circuit diagram of the receiver is shown in Figure 4. The transmit coil was supplied with 1.26 A of current at 179 kHz, which induced a B-field of 80 μT at the ME receiver. The B-field produced an AC voltage in the ME receiver which was rectified through the VD rectifier. The rectified voltage was boosted to 3.3 V through the EPEAS chip. The EPEAS chip produced a current of 20 μA to power the MCU, I_L. By measuring the change in the voltage, ΔV, over time, Δt, across the storage capacitor, C_s, the average current into the capacitor, I_s, was calculated according to

$$I_s = C_s \frac{\Delta V}{\Delta t}. \quad (1)$$

With a storage capacitor of 1 mF, $\Delta V/\Delta t$=0.32V/7.4s which gives I_s=43.2 μA. The total power going out of the EPEAS chip can be calculated by $V_{MCU} * I_L + V_{C_s} * I_s$, which is equal to 221.5 μW.

The 221.5 μW delivered to the load was sufficient to power the MCU, even when transmitting its sensor data via the Bluetooth radio. The MCU successfully read and wirelessly transmitted the temperature sensor's reading to a laptop with a receiver station set up nearby. The power calculated here is slightly greater than the power previously calculated in the system power loss section. This slight difference in power is probably a result of the RF power amplifier dynamics, whose current output level varies slightly over time.

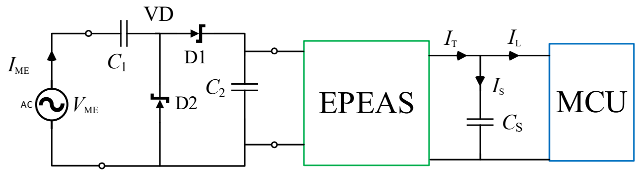

Figure 4 A simplified circuit diagram of the receiver side of the ME WPTS. The ME receiver is an AC voltage source that is rectified through the VD rectifier. The DC current is then passed into the EPEAS chip which outputs a total current, I_T, which powers the MCU with I_L. The excess current, I_s, charges a storage capacitor, C_s.

Comparison of ME WPTS for Biomedical Implants

To compare our ME WPTS to other WPTS designed explicitly for biomedical implants, we define a figure of merit, F_{ME}, [6] that normalizes the power by the device's volume, the square of the WPTS B-field, and frequency,

$$F_{ME} = \frac{P_L}{V_{rx} B^2 \omega} \quad (2)$$

where P_L is the power delivered to the load for the WPTS, V_{rx} is the volume of the receiver, B is the strength of the B-field at the receiver, and ω is the operating frequency of the ME WPTS. F_{ME} is defined in $WsT^{-2}m^{-3}$.

The comparative analysis in Table 4 shows that our device performs very well compared with other ME WPTS. The only ME WPTS that had a higher F_{ME} was a larger Galfenol-PZT device [7] that required a large DC bias field produced by large (relative to the receiver size) permanent magnets to achieve its high-power gain. Even then, our self-biased device has a F_{ME} only 47.7% lower with the bias field coming from the device's internal structure. If a biomedical sensor were to be built out of a non-self-biased ME receiver, then it would be essential to consider the volume of the permanent magnets that would be needed to define the overall volume of the complete ME transducer receiver.

Discussion of Safety Considerations for Maximum WPT

When designing WPTS for operation inside the human body, it is important to ensure the B-field does not exceed the recommended exposure limits [8]. Using the IEEE reference

Table 4 Comparison of ME WPTS for biomedical implants

Article	Method	Frequency (kHz)	Receiver Size	P_L	B-field (µT)	F_{ME} (MWs/T^{2}m^3)
[7] 2019	ME	150	10 x 20 x 1.76 mm^3	5 mW	101	9.283
[10] 2019	ME	202	10 mm ϕ x 2.4 mm	19.5 mW	1000	0.512
[12] 2019	ME	70	10 x 20 x 2.035 mm^3	600 µW	600	0.059
[11] 2008	ME	60	41 x 20 x 0.3 mm^3	160 mW	2000	2.694
This Work 2022	ME	179	12 x 5 x 1.05 mm^3	349 µW	80	4.854

levels [9], both [10], [11]operate at higher B-fields than the prescribed safety level. Our ME WPTS operated at fields significantly below the safety limits. Since ME WPTS power output scales with the magnetic field squared, our ME WPTS could increase its power by 2.03x if we operated the WPTS at 114 µT, which is the safety constraint at 179 kHz. Additionally, if our device's resonant frequency was lowered to 100 kHz and we operated our ME WPTS at the IEEE maximum allowable B-field limit of 615 µT, our receiver could generate 20.6 mW of power (a 59x increase), meaning that a 1-hour charge would enable 312 hours of operation of our sensor. For a fixed volume ME transducer, the device's resonant frequency (which is the operating frequency) could be lowered by increasing its length while decreasing its width.

CONCLUSION

In conclusion, a self-biased ME WPTS for a biomedical implant was successfully built and characterized. The self-biased ME receiver can deliver 389 µW of power to an optimal resistive load with a B-field of 80 µT. The designed power conditioning electronics achieved an end-to-end efficiency of 60%, resulting in 221.5 µW being used to power an implant that senses the temperature and successfully transmits its data wirelessly through a Bluetooth transmitter. This demonstration shows the potential usefulness of small ME transducer receivers to power biomedical implants. Future work will involve taking the benchtop demonstration of the ME WPTS and miniaturizing it into an mm-scale biomedical implant.

In addition to the device demonstration, a comparison was made between our ME WPTS to other ME WPTS designed for biomedical implants. Our ME WPTS performed better than all but one ME WPTS when the power delivered to the load was normalized by the device's volume and B-field. Additionally, since our device is a self-biased ME transducer, it would not need additional volume in the implant for large permanent magnets.

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